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Perturbation-based balance training

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Perturbation-based balance training

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Editorial: Perturbation-based balance training

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KEYWORDS

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Editorial on the Research Topic
Perturbation-based balance training

Introduction

Perturbation-based balance training (PBT; or reactive balance training or perturbation training) is balance training that uses repeated, externally applied mechanical perturbations to trigger rapid reactions to regain postural stability in a safe and controlled environment (McCrum et al.). The goal of PBT is to specifically target and improve the ability to maintain and recover balance in situations that often lead to falls. There is evidence suggesting that perturbation-based balance training can reduce falls in everyday life by up to 40%–50% (1–3). This is particularly promising given the relatively short time needed to achieve these benefits, in comparison to traditional exercise programs. However, there were and are important knowledge gaps for this approach to fall prevention, especially regarding its efficacy, mechanisms, optimal dose, type and presentation of perturbation, transfer or generalisability to daily life tasks, application/feasibility in various clinical populations and retention of the improvements over time. Therefore, this Research Topic aimed to collect contributions on the latest developments related to perturbation-based balance training. Contributions could be broadly classified into three categories: (i) Balance measures using perturbations, (ii) effects and mechanisms and (iii) implementation of perturbation-based balance training.

Balance measures using perturbations

Five articles in this Research Topic contributed insight into assessing balance during perturbations. Two articles presented tests using instrumented treadmills: Lesch et al. proposed a perturbed postural balance test using an instrumented treadmill which has become increasingly common in biomechanics laboratories and clinical settings instead of purpose-built movable force plates; and Adams et al. proposed the Stepping Threshold Test using an instrumented treadmill and observation of stepping behaviours via video recording. Two studies examined specific outcomes derived from perturbation testing: Rieger et al. proposed a simple way to track balance recovery performance during gait perturbation training using the center of pressure data from an instrumented treadmill; and Gerards et al. demonstrated that adaptability to gait perturbation via treadmill belt accelerations was related to history of falls in older adults. Finally, in a comprehensive overview, Grabiner and Kaufman reviewed the literature and stated the need for developing and establishing biomechanical risk biomarkers for preventable falls such as those induced by trips. They proposed trunk kinematics as a biomarker for trip-specific falls.

Effects and mechanisms of PBT

Seven articles in this Research Topic addressed various mechanisms and effects of PBT. Two specifically addressed transfer between different tasks. A randomised cross-over trial by Song et al. directly compared acute motor adaptations to commonly used treadmill belt accelerations trips against obstacle trips on a walkway. They reported that older adults could learn to improve dynamic stability by repeated exposure to both perturbation modalities, but the adaptations to treadmill belt accelerations did not transfer to an actual trip. In slight contrast, Bhatt et al. compared groups of older adults who completed slip or trip training following novel perturbations of the untrained type. The training resulted in proactive adjustments that could worsen the reactive response to the opposite perturbation (interference) but older adults could generalise their improved reactive control to maintain dynamic stability (margin of stability), to preserve limb support control, and to reduce fall risk.

Four articles addressed aspects relating to muscular contributions and kinetics of balance recovery. Yoo et al. elucidated the kinetic and muscular mechanisms of balance recovery following a split-belt treadmill perturbation. Older people showed greater joint moments and muscle responses of the compensatory limb during the recovery period than in younger people. In contrast, older people showed greater co-contraction of biceps femoris/rectus femoris muscles during recovery, likely compensating for their muscle weakness. Debelle et al. studied kinematic and kinetic mechanisms of improved balance recovery to repeated backward slips simulated by treadmill belt accelerations in older and younger adults. Regardless of age, dynamic stability improved with repeated exposure, which was related to change in step length and ground reaction force angle. Staring et al. investigated kinematic and muscular mechanisms of improved stepping responses to backward and forward platform translations which were repeatedly applied to chronic stroke survivors. Although muscle onset became faster in gastrocnemius and likely in tibialis anterior, these were not related to increase in step length, duration and velocity which were related to a more upright position. Van Wouwe et al. compared the effects of a traditional 12-week resistance exercise program and a 3-week PBT program using support-surface perturbations of stance in older adults. The study found intervention-specific effects (improved strength due to resistance exercise and improved reactive balance during stance perturbations due to the PBT) and reported that muscle strength was not a limiting factor for reactive balance. However, neither intervention translated to improved performance of perturbation recovery during walking.

One final study was potentially a first for the field, in which Martelli et al. trained older adults using waist-pull perturbations on a treadmill. This study showed that a single session of perturbation-based balance training produces acute aftereffects in terms of increased cognitive performance and gait stability in healthy older adults.

Implementation

The final three articles of the Research Topic considered some implementation-related aspects of PBT. As a major barrier of PBT is the limited accessibility to perturbation equipment and a safety harness, Lee et al. proposed a novel manual technique for trip recovery training. Another gap in the current literature is how psychological factors play a role in the effectiveness of PBT and the article by Soh provides a reasoned overview of ways to measure and interpret falls efficacy, balance confidence, and balance recovery confidence in this context. Finally, in our review article McCrum et al. we provide a definition of PBT as "balance training that uses repeated, externally applied mechanical perturbations to trigger rapid reactions to regain postural stability in a safe and controlled environment." and discuss the current state of research on PBT from the perspectives of the basic principles, mechanisms and implementation in practice.

Conclusions

PBT is a promising approach to fall prevention. With each year, more studies provide insight into both the underlying mechanisms of this training and how to better implement it in practice. However, as we noted in McCrum et al. several fundamental and applied aspects of PBT still need to be investigated and understood in order for it to be widely and successfully applied in practice.

Author contributions

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ORIGINAL RESEARCH
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Development of a Balance Recovery Performance Measure for Gait Perturbation Training Based on the Center of Pressure

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Background: The availability of instrumented treadmills that can apply unexpected perturbations during walking has made gait perturbation training more popular in clinical practice. To quantify and monitor balance recovery while training, easy to use measures are needed and may be based on integrated force plate data. Therefore, we aimed to quantify and evaluate different implementations of the recovery performance measure based on center of pressure data.

Methods: Recovery performance was calculated based on differences in center of pressure trajectories between unperturbed walking and balance recovery after a perturbation. Five methodological choices leading to 36 different implementations were evaluated. Test-retest reliability, effect sizes, and concurrent validity were evaluated against trunk velocity measures.

Results: Differences in measures of (dis-)similarity, time normalization and reference data affected reliability, sensitivity and validity and none of the performance measure implementations based on center of pressure trajectories was superior on all criteria. Measures assessing perturbation effects on trunk velocities provided more reliable and sensitive recovery outcomes.

Discussion: Different implementations of the recovery performance measure can be chosen dependent on constraints imposed in the clinical setting.

Conclusion: Quantifying recovery performance based on center of pressure data is possible and may be suitable to monitor improvement in recovery performance after gait perturbations in specific clinical setups. Validity of performance measures in general requires further attention.

Keywords: postural balance [MeSH], walking, gait, accidental falls, physical functional performance, rehabilitation, aging

INTRODUCTION

Fall prevention training using gait perturbations during walking is becoming more popular (Gerards et al., 2017), but the application of standardized and sufficiently impactful perturbations is generally limited to setups that require a lot of space and are expensive and complex to control (e.g., over ground walkways or gait labs). Smaller devices, among which treadmills, are being developed to make gait perturbation training accessible in clinical settings. One advantage of advanced research setups is the ability to capture movements and record forces, to quantify the unperturbed gait kinematics and kinetics, the perturbation magnitude and impact, as well as the patient's balance recovery performance. Quantification of recovery performance is key for successful clinical application. It may also allow for identification of people at higher risk for falling and indicate necessity of fall-preventive interventions. Furthermore, it allows standardization and monitoring of the patient's progress over training sessions. This enables therapists to consistently adjust perturbation difficulty, to keep the patient challenged and motivated, and may support reporting outcomes to health care providers.

When loosing balance due to gait perturbations, the neuromotor system applies various strategies to regain balance (Hof et al., 2010; Reimann et al., 2018; van den Bogaart et al., 2020). According to Hof (2007), this can be achieved by adjusting the position of the center of pressure (CoP) relative to the vertical projection of the center of mass (CoM), by counter rotations of body segments around the CoM or by applying external forces (e.g., holding on to a handrail). Furthermore, older adults show compensatory stepping reactions (Jensen et al., 2001) and often take multiple steps in response to both anterior-posterior and medio-lateral perturbations (Mille et al., 2013).

In literature, recovery performance has been quantified in various manners. One commonly used parameter is called the margin of stability, which relates the movement of the center of mass (CoM) to foot placement (Hof et al., 2005). Although this measure is straightforward to use in unperturbed walking, its use is limited for large perturbations, due to the wide variability in balance recovery responses in terms of stepping direction, skipping instead of stepping, number of steps used, and use of other strategies than adjusting the position of the CoP, such as speeding up or slowing down (Bruijn et al., 2013; Hak et al., 2013). Consequently, it is not easy to infer how changes in the margin of stability contribute to recovery.

Maybe a more suitable approach is to quantify recovery performance after perturbations based on trunk kinematics (Owings et al., 2001; Grabiner et al., 2008; Bruijn et al., 2010; Sessoms et al., 2014; Roeles, 2018), as the trunk has a large impact on balance, given its large mass and cranial location. Trunk flexion angle at toe-off and trunk flexion velocity at recovery foot contact have been related to the successful balance recovery (van den Bogert et al., 2002; Sessoms et al., 2014). By combining linear and angular trunk velocities, the whole recovery movement of the trunk can be captured and compared to the trunk movements during normal walking. The deviation from normal walking can therefore be used to describe the perturbation impact and the rate of return to normal walking (Bruijn et al., 2010; Roeles, 2018; Rieger et al., 2020). The advantage of this approach is that it is less sensitive to variability in reactive stepping strategies.

While inertial measurements may allow low-cost motion capture compared to optical systems, at present, motion capture is often not available in clinical practice, so the measures mentioned above cannot be used. A simple solution, which would be more accessible (i.e., less time consuming), is the use of treadmills with embedded force plates. The cheapest option here is a one-directional force plate, which only records vertical ground reaction forces, and may provide sufficient information to quantify recovery after gait perturbations. To our knowledge, measures based on force plates to quantify balance recovery during gait have not been investigated previously. Finally, for successful clinical application, recovery performance should be quantified as single value, that is easy to interpret and available online during training, immediately after each perturbation.

Taking these constraints into account, the purpose of the study was to develop several implementations of a new potential measure of what we coined quantified recovery performance (QRP). These implementations all compare the CoP trajectory during balance recovery from a perturbation, but with small differences in data processing. We evaluated test-retest reliability, sensitivity and concurrent validity of these different measures against motion-captured based measures of trunk velocity. We hypothesized that the QRP has sufficient reliability, validity, and sensitivity to change to be used to monitor progress in fall-prevention training.

MATERIALS AND METHODS

Participants

Data of a previous perturbation intervention trial were used for this study (Rieger et al., 2020). The cohort consisted of 30 healthy older adults aged 65 years or older, who had no experience with perturbation training. Any neurological, cardiovascular or pulmonary comorbidity (i.e., stroke, heart attack, hypertension) that occurred in the past 12 months, as well as orthopedic complications (i.e., lower extremity fractures, joint replacements) within the past 6 months before the study, led to exclusion.

Experimental Setup

The setup and perturbation characteristics are explained in detail in the original study (Rieger et al., 2020). Briefly, participants walked on the GRAIL (Gait Real-time Analysis Interactive Lab, Motek Medical BV, Amsterdam, The Netherlands), a 3D instrumented dual-belt treadmill, with an integrated motion capture system (Vicon Motion Systems Ltd, Yarnton, UK). A model [Human Body Model (HBM), version 2.0, Motek Medical BV, Amsterdam, The Netherlands] based on 26 reflective markers placed on the feet, legs and trunk was used to capture the participants' movements.

A custom application (D-flow version 3.30.1, Motek Medical BV, Amsterdam, The Netherlands) triggered perturbations at heel strike (Zeni et al., 2008) of either the left or right leg while walking with a fixed treadmill speed at 1 m/s. ML perturbations consisted of a sideways movement of the treadmill platform to the side opposite of the foot contact, provoking cross-over stepping. AP perturbations consisted of belt decelerations at foot contact, provoking backwards balance loss. Participants were measured three times in total (Rieger et al., 2020). On the first day, their gait was perturbed eight times (four times in AP and four times in ML direction) before and after a short intervention consisting of 8 min of treadmill walking with 16 AP perturbations (experimental group) or without (control group). After a 1-week retention period, participants were measured again with the same eight perturbations. Results indicated, that short exposure to gait perturbations led to significant improvements in balance recovery (stabilization of the trunk during walking) that were retained over 1 week, which limits conclusion if training effects transfer between perturbation directions. Steady state gait parameters did not change compared to baseline, so we can conclude that improvements are based on improved reactive responses. Balance recovery was quantified based on trunk kinematics and we used this as a reference measure for testing concurrent validity in the current study. A detailed description can be found elsewhere (Rieger et al., 2020). In short, time series of trunk velocities of unperturbed and perturbed walking were normalized to 101 samples per stride. For unperturbed gait, averages over 100 strides and their variability for each percentage of the gait cycle were calculated. Deviations in perturbed gait relative to unperturbed gait were calculated for six degrees of freedom (Bruijn et al., 2010). Next, the deviations were divided by the standard deviation of the unperturbed gait cycle for each dimension and then combined as the Euclidian sum over degrees of freedom into a trunk velocity deviation measure (Bruijn et al., 2010). The integral of the deviation over the first three recovery strides following a perturbation, expressed as the area under the curve (AUC), was used to describe recovery performance for every perturbation. Initial exposure to perturbations caused an improvement in recovery performance in both groups. Hence, we used the pre- and post-intervention trials of the whole cohort to assess sensitivity. No change in recovery performance was found in the control group between the post-intervention and retention measurements. Hence, the post-intervention and retention trial of the control group were used to assess test-retest reliability. Finally, to assess concurrent validity we used the retention trial over the whole cohort, excluding the pre- and post-intervention trials that would add repeated (dependent) measurements.

Data Processing

Trunk marker data and force plate data, recorded at baseline, post-intervention and retention, were processed with Nexus software (version 2.7.0, Vicon Motion Systems Ltd, Yarnton, UK) and custom MATLAB scripts (version R2018a; MathWorks Inc, Natick, MA, USA). Three-dimensional marker data were smoothed using a second order 15 Hz low-pass Butterworth filter. Deviation in linear and angular trunk velocities from normal walking were used to quantify performance during balance recovery following a perturbation (Rieger et al., 2020). Vertical force and moment data of the instrumented dual-belt treadmill, recorded at 1,000 samples/s, were combined to simulate a single force plate and to estimate the CoP time series, which were then smoothed with a second order 6 Hz low-pass Butterworth filter and a second order 0.5 Hz high-pass Butterworth filter to correct for drift in the position on the treadmill. Finally, CoP data were resampled to 100 samples/s to match 3D marker data.

Quantified Recovery Performance

The QRP is based on the fact that humans have a relatively constant gait pattern during unperturbed walking with the movements in each gait cycle being approximately the same. This gait pattern also results in a relatively constant CoP trajectory. The CoP is the point of application of the ground reaction force vector. This single point on the supporting surface is an effect of the forces that the individual exerts on the surface during walking. The proposed QRP utilizes this property of gait, since any perturbation will result in a change from this pattern. The QRP was calculated in different ways based on five methodological choices in data processing, with two to three options each, leading to 36 different implementations (**Figure 1**). Here, we describe the different choices that were made in the algorithm.

(1) The change from the normal walking pattern was quantified using Pearson's correlation coefficient or using an area under the curve describing the difference in the time series of the CoP patterns between perturbed and unperturbed gait as a measure of deviation.

(2) The next choice considered which CoP dimension to use. A pilot study suggested that change from the normal walking pattern was larger in AP compared to ML direction, when perturbations were applied in AP direction and vice versa for perturbations in ML direction. To cover the whole recovery reaction, the two directions can be combined. For the correlation-based calculation AP and ML coefficients were averaged using the Fisher z-transformation before averaging to avoid interpretation bias if the sampling distribution of correlation values is skewed, followed by the inverse transformation. For deviation-based calculation, both dimensions are combined as the Euclidian sum over dimensions.

(3) For perturbed gait, time normalizing the gait cycles may result in an average gait cycle that may be stretched unnaturally due to missed gait events when using an automatic gait event detection algorithm. One the other hand, because absolute step times may be different after a perturbation, time normalization may improve the comparison between unperturbed and perturbed gait episodes.

(4) To obtain an unperturbed gait pattern as reference, either a short episode of pre-perturbed walking preceding a perturbation or a separate trial of unperturbed walking can be used. The latter would allow a more reliable estimate as many gait cycles can be measured, but this comes at the cost of a longer measurement time. In our study, we recorded a 2-min steady state walking trial at 1 m/s. A template of unperturbed walking is then created by repeating the average gait cycle of these different references.

(5) Finally, the short episode of pre-perturbed walking was determined either by a number of gait cycles or a number of seconds. A measure dependent on the number of recovery steps requires accurate gait event detection. If not robust enough, manual post-processing is needed to evaluate whether events are

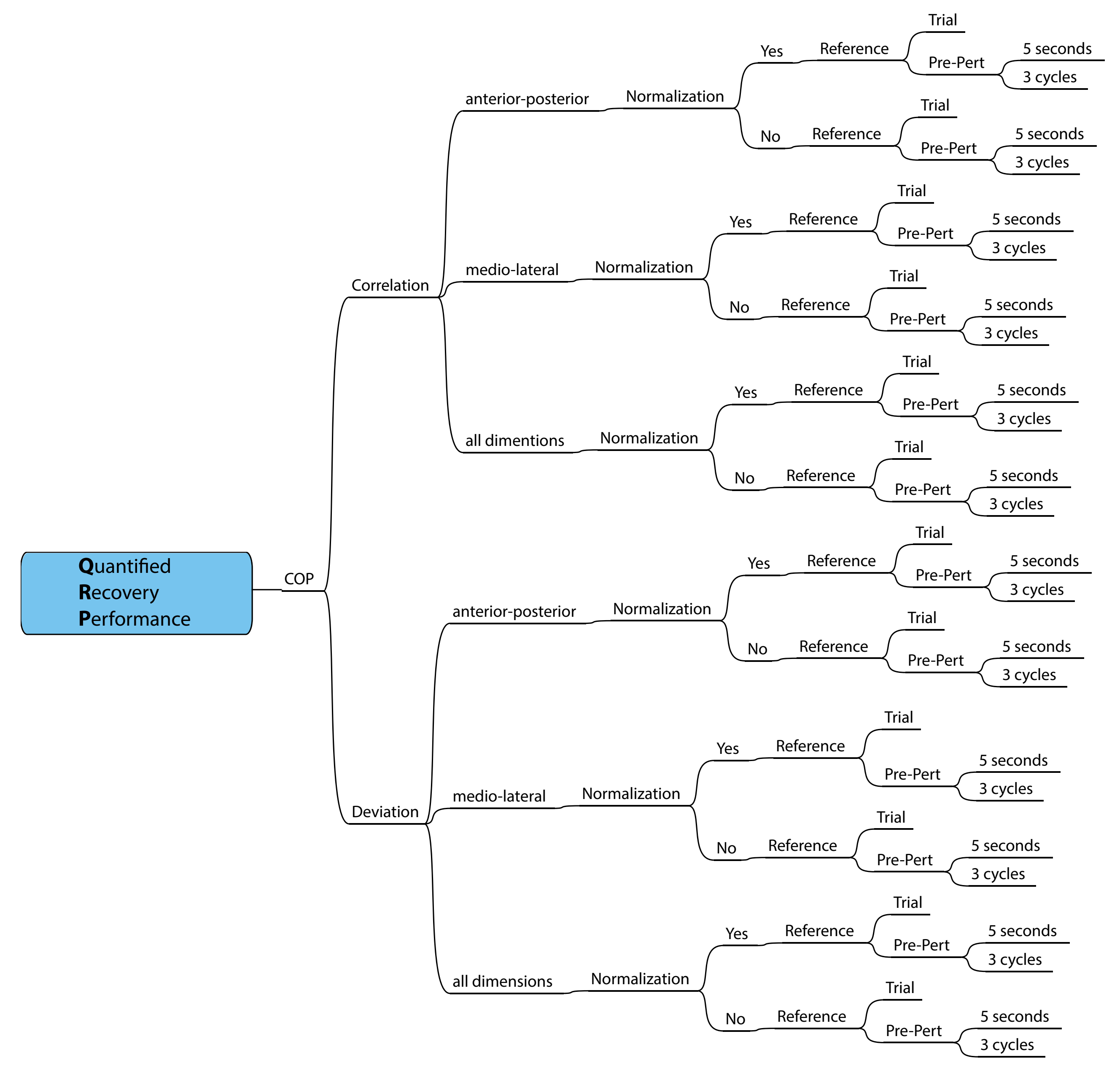

FIGURE 1 | Schematic overview of methodological choices in data processing, resulting in 36 QRP measures.

correct or missing. The variance of recovery strategies and time to recover normal walking is large between participants. In a pilot study with older adults, therapists were not able to visually observe any effect of a perturbation after 5 s and our previous study showed that three cycles are enough for recovery of normal walking based on changes in trunk velocities (Rieger et al., 2020). The length of the post-perturbed walking episode was equivalent to that of the pre-perturbation episode.

Example Calculation of a QRP Implementation

The section above introduced the different choices that can be made within the algorithm. Here we describe, as an example, the details of the QRP calculation using the CoP trajectory in the AP dimension with a non-normalized time window of 5 s before and after the perturbation trigger, which are then compared using Pearson's correlation:

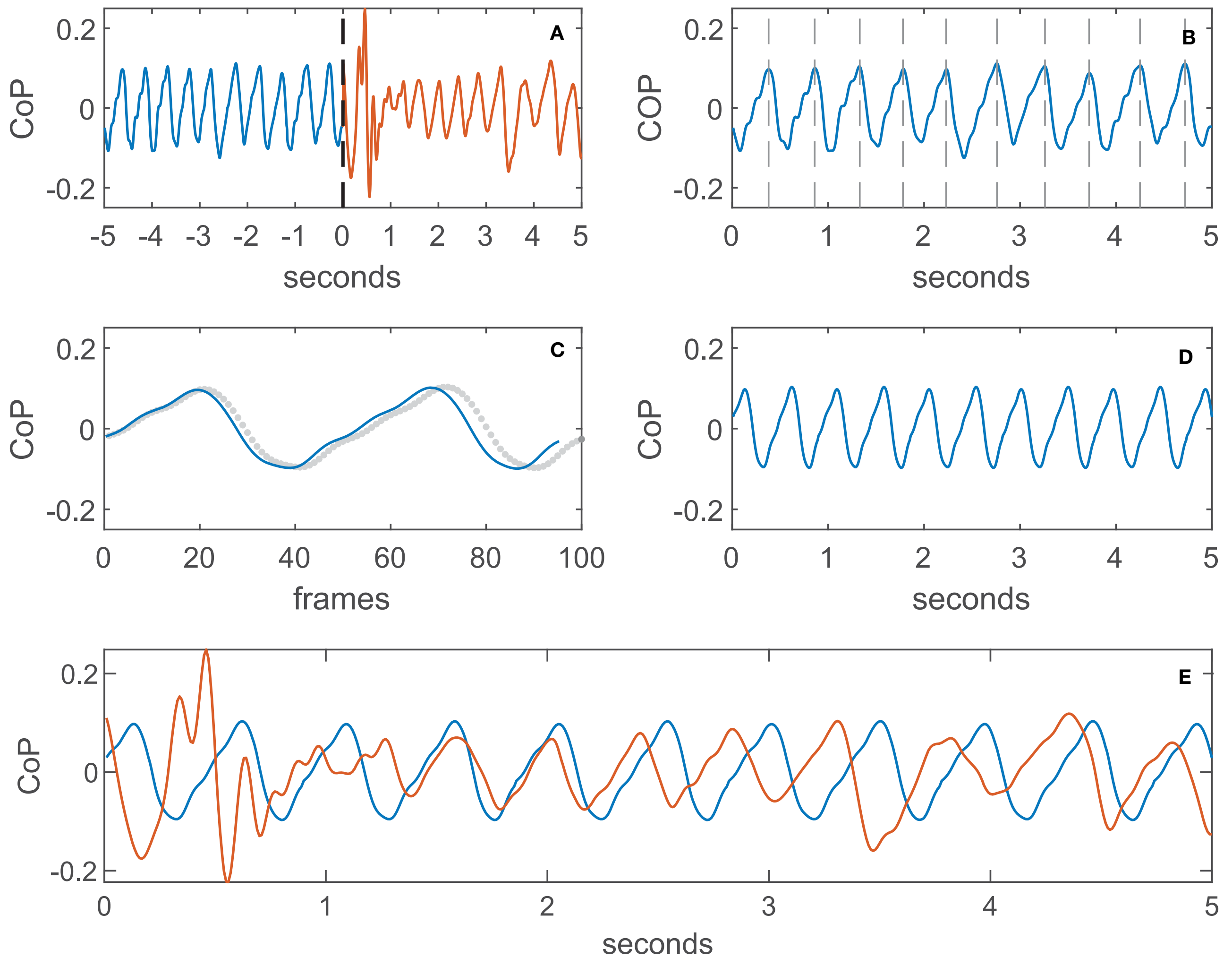

FIGURE 2 | Example calculation of the QRP, **(A)**: pre- and post-perturbation window (black dashed line is the trigger), **(B)**: divided into gait cycles, **(C)**: de-normalization of the gait cycle (dotted line is the normalized gait cycle), **(D)**: template of unperturbed walking (averaged), **(E)**: alignment of template (averaged) and perturbed (non-normalized) walking signal.

Step 1: The last 5 s before and the first 5 s after the perturbation of the CoP trajectory are selected and stored as a pre- and post-perturbation window (**Figure 2A**).

Step 2: In pre-perturbed CoP data, gait events are detected according to the method of Zeni and colleagues (Zeni et al., 2008). The gait cycles of the pre-perturbed episode are determined by the right or left heel strikes as the start and end of a gait cycle (**Figure 2B**).

Step 3: The average length of the gait cycles is calculated and the gait cycles of the pre-perturbed episode are time normalized to this average gait cycle length. In case of a non-normalized pre-perturbed episode, the gait cycle is de-normalized again after averaging (**Figure 2C**). For unperturbed walking, the average gait cycle is repeated multiple times to construct a template with reduced variance (**Figure 2D**).

Step 4: The cross-correlation between the post-perturbation CoP trajectory and the constructed unperturbed walking template is used to align the two signals (**Figure 2E**). The maximum correlation is used as a measure of recovery performance. In case of AUC, the template (unperturbed episode) is cut to the length of the perturbed episode.

Statistics

Statistical analyses were performed with SPSS version 25 (SPSS Inc, Chicago, IL, USA). First, we checked for normality of data using the Shapiro-Wilk test. Second, the interquartile range rule (IQR) was used to detect outliers, with no extreme outliers (exceeding three times IQR) being found and consequently no observations were excluded from the analysis. The level of significance was set at alpha = 0.05.

To evaluate between session test-retest reliability we calculated parametric intraclass correlation coefficients (ICC) with two-way mixed single measure analyses for consistency between the post-intervention and the retention trials of the control group. ICCs were interpreted according to Shrout (1998) as indicating insufficient reliability (<0.40), fair reliability (0.40–0.60), moderate reliability (0.60–0.80), and substantial reliability (>0.80). Next, sensitivity to change was assessed by the effect size (ES) calculated with the mean difference over time divided by the standard deviation of the difference for the pre- and post-intervention trials of the whole cohort. An ES of 0.2 and lower reflects a mean difference of two measurements of <0.2 standard deviations, which can be interpreted as a trivial effect, even if results are significant. An ES between 0.5 and 0.8 is considered as a medium effect and above 0.8 as a large effect (Cohen, 1992). Finally, concurrent validity of the retention trial was tested with Pearson's correlation coefficient between the QRP and recovery performance based on trunk velocity deviation (Rieger et al., 2020). Pearson's r were interpreted as weak (below $r = 0.3$), moderate ($r = 0.31$–0.69), and strong (above $r = 0.7$).

RESULTS

Measures of trunk velocity deviation resulted in substantial between session reliability of ICC = 0.897 for ML perturbations and ICC = 0.855 for AP perturbations (**Figure 3**). The sensitivity to change was ES = 0.977 for ML perturbations and ES = 1.028 for AP perturbations which is considered to be a large effect.

For the QRP measures, between session reliability ranged from fair to substantial for ML perturbations with ICC = 0.486–0.935 and from insufficient to substantial for AP perturbations with ICC = 0.290–0.882 (**Figure 3**). The sensitivity to change ranged between small ES = 0.005–0.434 for ML perturbations and between small to medium ES = 0.028–0.520 for AP perturbations (**Figure 3**). Concurrent validity with the recovery performance based on trunk velocities ranged from weak to very strong $r = 0.09$–0.938 for ML perturbations and from weak to strong $r = 0.009$–0.775 for AP perturbations (**Figure 3**).

No single QRP method was superior to the other calculations. See **Supplementary Table 1** for full details.

DISCUSSION

We developed and evaluated a new method of quantifying recovery performance after treadmill-based gait perturbations using center of pressure data (CoP) obtained from a treadmill-embedded force plate. We compared various implementations of the QRP, with respect to test-retest reliability, sensitivity to change and concurrent validity. Results showed a wide range across these implementations for reliability, sensitivity, and concurrent validity, suggesting that no option is superior and that a choice between these implementations must be made dependent on the constraints and demands of the setting in which the QRP will be used. Theoretically, when evaluating a perturbation protocol only using decelerations of the belt, a QRP based on a non-normalized pre-perturbation episode of 5 s combining AP and ML dimension as input would provide substantial reliability (ICC = 0.855), medium sensitivity to change (ES = −0.496) and moderate validity (r =-0.476). For a ML perturbation protocol, a QRP based on a normalized unperturbed walking trial with only the AP dimension as input would provide high reliability (ICC = 0.935) with small sensitivity to change (ES = −0.434), which is still the largest effect size across all options for ML perturbations, and strong validity ($r = -0.854$).

The change from the normal walking pattern can be quantified using Pearson's correlation coefficient or using an area under the curve describing the difference in the time series of the CoP patterns between perturbed and unperturbed gait as a measure of deviation. Both options provide similar results for test-retest reliability and sensitivity to change across all options. In general, the correlation-based option yielded less variable results compared to the deviation-based options, especially for concurrent validity, suggesting correlation-based calculations of the QRP to be more consistent across different perturbations.

Our pilot study suggested that the change from the normal walking pattern was larger in the AP compared to the ML dimension, when perturbations were applied in AP direction and vice versa for perturbations in ML direction. However, similar performance of the QRP was found when using CoP data from either the AP or ML dimension or when AP and ML dimensions were combined. When using the QRP in a setting where perturbations in AP and ML direction are applied, then a correlation-based option using combined AP and ML CoP data provides more reliable and sensitive results for both AP and ML perturbations compared to a deviation-based option combining AP and ML CoP data. Further, when combining CoP data from the AP and ML dimension, the recovery reaction can be captured more completely yielding a more comprehensive analysis of the recovery performance.

For clinical practice, the QRP should preferably not rely on detection of gait events, as automatic detection of gait events from a COP trajectory may not always produce reliable results, due to the large variability in reactive stepping response. Multiple recovery strategies have been observed for trips (Eng et al., 1994) and slips (Yang et al., 2008) and some subjects perform cross-over steps (Vlutters et al., 2016) and backward steps (Yang et al., 2014). In the present study, we manually checked for false or missing gait events. In a clinical setting, this may not be possible and the gait event detection algorithm needs to detect gait events accurately, which may be limited due to the manifold stepping responses after a perturbation. Test-retest reliability and sensitivity to change of QRP based on a gait episode consisting of 5 s were comparable to those of QRP based on a gait episode consisting of three gait cycles, although concurrent validity was lower, when using time-based episodes. Moreover, as detection of gait events is required to calculate spatial-temporal gait parameters such as step length and step time, such measures that use the time or steps required to recover to baseline values of spatial-temporal parameters (Krasovsky et al., 2012) are less suitable for clinical practice than the QRP measures we proposed.

Given the natural variation in walking behavior, segmenting data into gait cycles almost always results in gait cycles of different duration. Therefore, time-normalization is commonly used for comparison of gait patterns. Similarly, during recovery performance, time normalization of the gait cycles may improve the comparison of perturbed and unperturbed gait cycles. However, it may unnaturally stretch gait cycles in case of false or missing gait events. In the current study we corrected false or

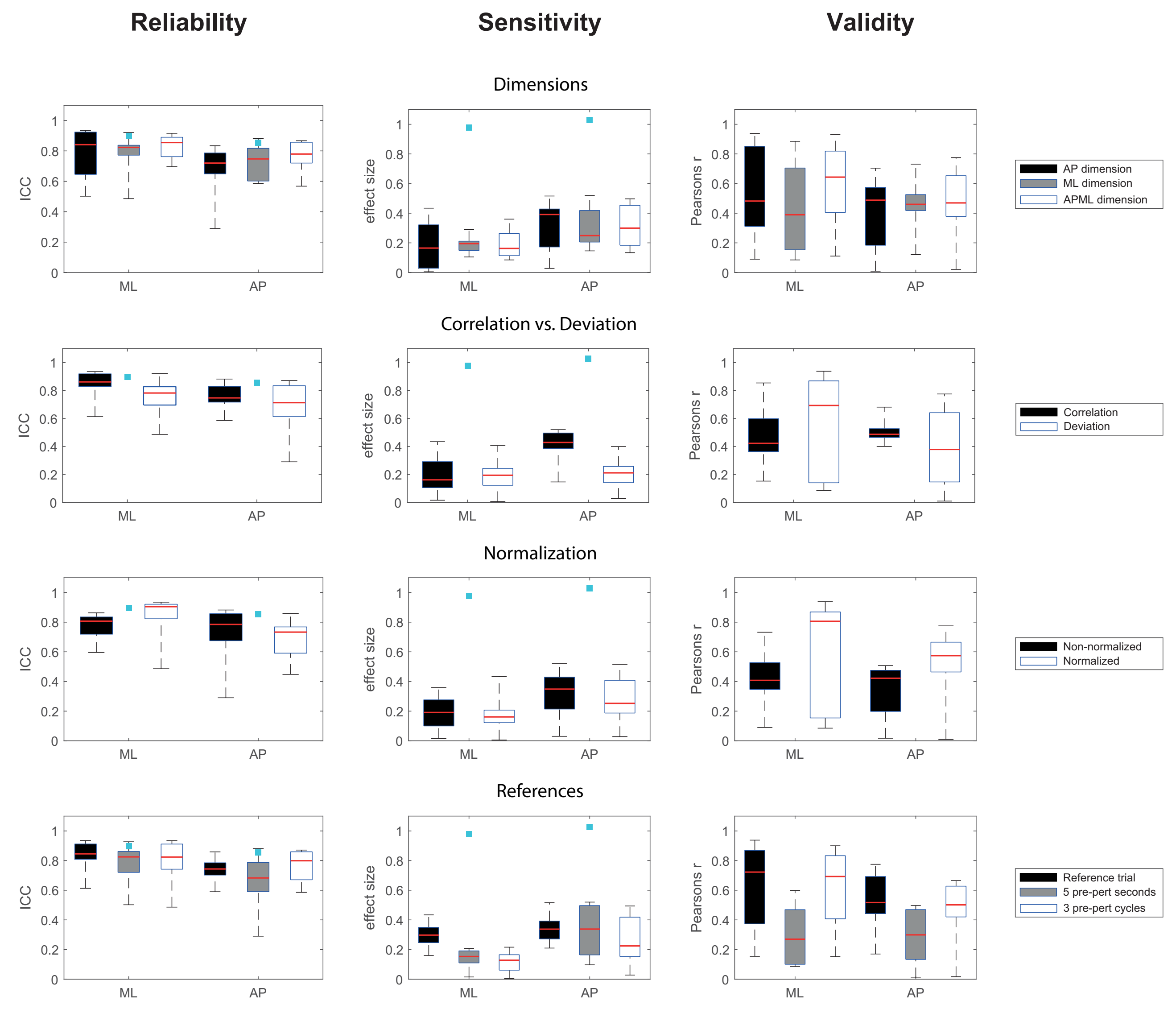

FIGURE 3 | Boxplots of outcomes for different methodological choices of QRP implementations, x-axis: ML, medio-lateral perturbation; AP, anterior-posterior perturbation, light blue squares indicate the results of the trunk velocity deviation measure.

missing gait events and our data suggests that normalizing the data provides comparable performance for test-retest reliability and sensitivity to change, but increases variability for concurrent validity results compared to non-normalized data.

If no separate reference trial is available, correlation-based QRP measures provide more reliable outcomes than deviation-based measures. However, using a normalized separate reference trial provides the highest concurrent validity across all options. This may be because the same choice was made for the reference measure based on trunk velocities (Rieger et al., 2020). A reference containing three pre-perturbation gait cycles provide similar performance on reliability, sensitivity and validity than a separate reference trial and both options have higher concurrent validity compared to a 5 s pre-perturbation time window as reference. With respect to the length of the time window analyzed, we have previously shown that recovery of the trunk kinematics is achieved within three gait cycles after perturbations of a magnitude such as applied here. This suggests that a pre- and post-perturbation episode of three cycles would be sufficient (Rieger et al., 2020). However, this depends on reliable automatic gait event detection and as mentioned before, this may limit this option.

In clinical practice, online feedback of recovery performance to the therapist is key for monitoring and adjusting perturbation difficulty within a training session. It provides the therapist with objective recovery performance for each perturbation as is preferable over subjective visual judgement. This is possible when the measurement uses pre-perturbed walking as reference. As an alternative, the use of data from a separate reference trial is possible and the advantage may be that anticipation to perturbations does not affect the reference COP pattern. In addition, more gait cycles can be recorded, resulting in a more precise average gait cycle. However, such a trial needs to be taken in the beginning of a training session and may be affected by a lack of familiarization. Therefore, recording a separate reference trial may be time consuming and may need to be recorded in every training session, as a participant's walking speed may change over sessions. Finally, participants may adapt their gait pattern between two perturbations, which is likely to be different compared to the gait pattern of a separate, unperturbed walking reference trial. This would favor for using three pre-perturbed gait cycles, as this provides similar results as a separate reference trial of unperturbed walking. Moreover, we have previously shown that improvement in recovery performance is independent of adaptive changes in the gait pattern (Rieger et al., 2020).

In the current study we evaluated the CoP based QRP measures against a recovery performance based on trunk velocity deviations obtained with motion capturing. It could be argued that the recovery performance measure based on trunk velocity deviations is not yet accepted as a golden standard for quantifying balance recovery after gait perturbations. Alternatively, the concept of Margins of Stability (MoS) has been used to quantify stability of walking. However, stepping responses after a perturbation are manifold, including jumping, skipping, repositioning of the perturbed foot, various side or cross-over steps (Mccrum et al., 2018) and these are difficult to analyze within this framework. Moreover, the MoS concept is based on the assumption that the body can be modeled as an inverted pendulum. In responses after gait perturbations, this assumption is likely to be violated (Bruijn et al., 2013; Hak et al., 2013) Consequently, gait adaptations after experiencing a perturbation, e.g., walking with flexed knees to lower the CoM, may limit the applicability of the MoS (Hof et al., 2005). Therefore, measures based on the deviation in trunk velocities from unperturbed walking provide high test-retest reliability and sensitivity to change and has potential to become the golden standard to quantify recovery after gait perturbations.

Limitations

In our study, only treadmill belt decelerations were used in the AP direction to provoke backward balance loss and contra-lateral sway perturbations in the ML direction. These perturbations were selected as they are considered the most challenging for each direction eliciting the strongest recovery responses (Roeles et al., 2018; Rieger et al., 2020). However, this implies that information is lacking for perturbations using belt accelerations and ipsilateral sway perturbations. In addition, only one intensity level as perturbation difficulty was used and treadmill speed was fixed for all participants. Further investigation of validity for a variety of perturbation types after more or less challenging perturbations, at various gait speeds and across different target groups are recommended. Finally, we recommend that future recovery performance measures could be based on inertial measurement units (Faber et al., 2009; Miller and Kaufman, 2019) or a simple camera system, such as the Kinect, to capture trunk kinematics (Shani et al., 2017), as this provides more reliable and sensitive outcomes compared to CoP based measures.

For this study we used selected conditions from a previous study (Rieger et al., 2020). For test-retest reliability we selected data exclusively from the control group in that study as some systematic changes we found between time points in the training group. We additionally evaluated reliability over the whole cohort, which confirmed our conclusion that the QRP measure was less reliable than the reference measure based on trunk kinematics. To assess concurrent validity, we used the retention trial over the whole cohort, excluding the pre- and post-intervention trials. Additional evaluation of the effect sizes per group did not lead to a different conclusion. Furthermore, additional analysis of the concurrent validity in the baseline and post-intervention trials did not yield substantially different results.

Conclusion

Quantifying recovery performance using center of pressure data from a force-plate embedded treadmill device can achieve sufficient reliability and concurrent validity, although less reliable and sensitive to change than trunk velocity measures.

DATA AVAILABILITY STATEMENT

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Vaste Commissie Wetenschap en Ethiek (VCWE), Vrije Universiteit Amsterdam. The patients/participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

MR collected the data and wrote the first draft of the manuscript. MR and JvD analyzed the data. SP, FS, MP, and JvD edited the draft version. All authors contributed to conception and design of the study, manuscript revision, read, and approved the submitted version.

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SUPPLEMENTARY MATERIAL

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fspor.2021.617430/full#supplementary-material

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Conflict of Interest: MR is an early stage researcher, employed by Motek Medical BV and performed the analysis of the data, in collaboration with Vrije Universiteit Amsterdam. Further, SP and FS are employed by Motek Medical BV and a device of this company was used for this experiment.

The remaining authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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BRIEF RESEARCH REPORT
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Adaptability to Balance Perturbations During Walking as a Potential Marker of Falls History in Older Adults

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Given that falls most commonly occur during walking due to unexpected balance perturbations like trips and slips, walking-based balance assessment including walking stability and adaptability to such perturbations could be beneficial for fall risk assessment in older adults. This cross-sectional study reanalyzed data from two larger studies conducted with the same walking protocol. Participants completed unperturbed walking trials at speeds of 0.4 m/s up to 1.8 m/s in 0.2 m/s steps. Ten unannounced treadmill belt acceleration perturbations were then applied while participants walked at equivalent stability, assessed using the margins of stability. Retrospective (12 months) falls incidence was collected to divide participants into people with and without a history of falls. Twenty older adults (mean age 70.2 ± 2.9 years) were included in this analysis; eight people with one or more recent falls and 12 people without, closely matched by sex, age and height. No significant differences were found in unperturbed walking parameters or their variability. Overall perturbation-recovery step behavior differed slightly (not statistically significant) between the groups after the first perturbation and differences became more pronounced and significant after repetition of perturbations. The No-Falls group significantly reduced the number of recovery steps needed across the trials, whereas the Falls group did not show these improvements. People with a previous fall tended to have slightly delayed and more variable recovery responses after perturbation compared to non-fallers. Non-fallers demonstrate more signs of adaptability to repeated perturbations. Adaptability may give a broader indication of the ability of the locomotor system to respond and improve responses to sudden walking perturbations than unperturbed walking variability or recovery to a single novel perturbation. Adaptability may thus be a more useful marker of falls history in older adults and should be considered in further research.

Keywords: accidental falls, risk assessment, perturbation, aging, adaptation, stability recovery, falls prevention

BACKGROUND

Falls are a principal cause of injury, leading to disability and hospitalization in older adults (Berry and Miller, 2008). Therefore, adequate identification and treatment of older fallers are critical. Approximately 60% of outdoor falls in older adults occur when unexpected balance perturbations during walking (e.g., slips or trips) cause a sudden change in the relationship between the center of mass (CoM) and base of support (BoS) of the body (Berg et al., 1997). Thus, balance assessment during walking, focusing on walking stability and adaptability may be beneficial for fall risk assessment in older adults (Woollacott and Tang, 1997; Pai et al., 2010a; McCrum, 2020a).

In response to balance perturbations such as slips and trips, older adults show less effective initial recovery responses than younger adults (Pijnappels et al., 2005; Karamanidis and Arampatzis, 2007; Pai et al., 2010b). Still, the literature reports that older adults seem fully capable of improving their responses when exposed to repeated perturbations (Pai et al., 2014; Bohm et al., 2015; McCrum et al., 2017). As a result, walking stability in response to single and repeated perturbations may capture different underlying mechanisms. However, how adaptability to repeated perturbations relates to real life falls has not been the topic of many studies. Pai et al. (2010a) associated adaptability to repeated slip perturbations during a sit-to-stand task with a lower likelihood of future falls in daily life in older adults. Adaptability was indicated by less balance loss and falls during the task and improved recovery performance during the final slip. This association has not yet been thoroughly investigated for mechanical perturbations during walking, which are more task-specific to the most common causes of falls in older adults.

In this study, we aim to address the extent to which stability following a single perturbation and adaptability following repeated perturbations relate to falls history in older adults. Stability of the body configuration during walking will be measured using the margin of stability (MoS) (Hof et al., 2005). Due to previous indications of differences between older adults with and without a history of falls (Hausdorff et al., 2001; Mortaza et al., 2014) we also analyze step variability during unperturbed walking, to examine how these potential differences relate to those seen in the perturbation tasks. These analyses may give indications of the usefulness of such tasks and properties for falls risk assessments and falls prevention. We hypothesize that there will be not only higher step variability during walking, but also a reduced ability to cope with and adapt to unexpected balance perturbations during walking in older adults who fell in the past 12 months compared to older adults who did not fall.

Abbreviations: CoM, Center of Mass; BoS, Base of Support; METC, Medical Ethics Committee; MUMC+, Maastricht University Medical Center; Base, Baseline of the eleventh to second last step before each perturbation; Pre, The final step before each perturbation; Post1–8, The recovery steps following each perturbation (1–8).

METHODS

Setting and Subjects

This cross-sectional study reanalyzed data from two larger studies that included the same walking protocol (McCrum et al., 2020; Grevendonk et al. submitted). Older adults were recruited from the city of Maastricht, the Netherlands, and the surrounding area. Inclusion criteria were; community-dwelling, 65–80 years old, no known musculoskeletal or neurological deficits and no history of dizziness, balance or walking complaints. All subjects provided written informed consent. Both studies were approved by the medical ethics committee (METC) at Maastricht University Medical Centre (MUMC+) (NL58205.068.16 & NL59895.069.17) and were conducted in accordance with the declaration of Helsinki. Prior to the walking measurements, participants were given a short falls history questionnaire based on the recommendations of Lamb et al. (2005) and Lord et al. (2011), that led with the question: "In the past year, have you had any fall including a slip or trip in which you lost your balance and landed on the floor or ground or lower level?" This was followed by other questions about the number, location and cause of the fall(s) and about any injuries sustained. The questionnaire is available from https://osf.io/hmjef/ (McCrum, 2020b). Participants were divided into two groups based on their answers to this questionnaire. The Falls group including those participants who reported one or more falls in the past year, and the No-Falls group including those who did not fall.

For the current secondary analysis, a sample size calculation was conducted to determine the required sample size for $\alpha = 0.05$, $\beta = 0.8$ and estimated effect size of $f = 0.5$ for the group effect (falls history vs. no falls history) on MoS in a two-way ANOVA, with step as the other (repeated measures) factor (Baseline, pre-perturbation and the first eight recovery steps). This effect size for the MoS across the steps corresponds to a Cohen's d of 1 and to an approximately three-step difference in recovery to baseline MoS based on previous analyses (McCrum et al., 2020), which we interpret to be clinically meaningful. This revealed a required total sample of 20 participants. All available fallers from the existing datasets were included in the reanalysis, and a group of non-fallers was formed from participants who most closely matched the fallers in sex, age, and height.

Setup

Measurements were conducted with the Computer Assisted Rehabilitation Environment Extended (CAREN; Motekforce Link, Amsterdam). This comprises of a dual-belt force plate-instrumented treadmill (1,000 Hz), a 12 camera Vicon Nexus motion capture system (100 Hz; Vicon Motion Systems, Oxford, UK) and a 180° virtual environment providing optic flow. A safety harness connected to an overhead frame was worn by the participants. Six retroflective markers were attached to anatomical landmarks (C7, sacrum, left and right trochanter and left and right hallux) to calculate MoS.

Procedures

Participants completed familiarization trials followed by measurement trials from speeds of 0.4 m/s up to 1.8 m/s in

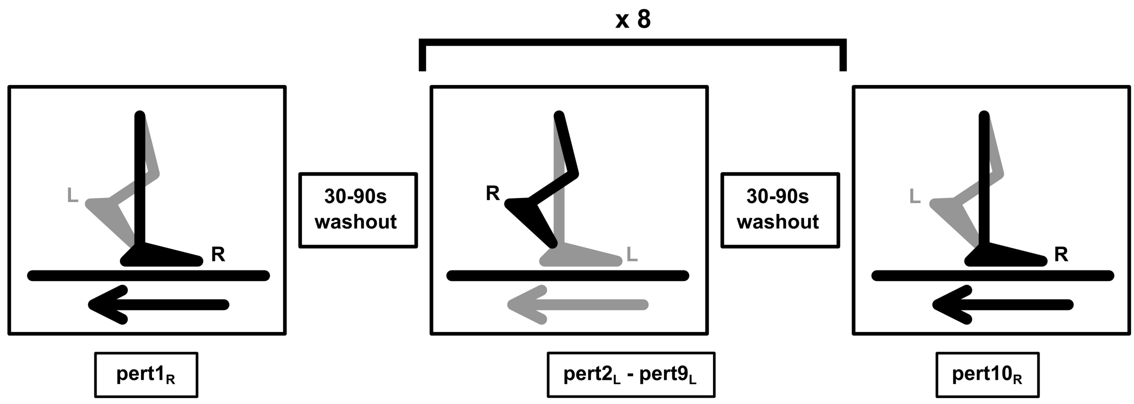

FIGURE 1 | Gait perturbation protocol [image previously shown in McCrum et al. (2019b)]. The right leg (R) was perturbed by the treadmill belt acceleration first ($Pert1_R$), followed by eight perturbations ($Pert2_L$ – $Pert9_L$) to the left leg (L), and the final perturbation ($Pert10_R$) was again applied to the right leg (R). In all, 30–90 s of unperturbed walking occurred between each perturbation. The perturbation was designed to cause a forward rotation and acceleration of the upper body, relative to the lower body, leading to a forward loss of dynamic stability.

0.2 m/s steps. To ensure equivalent stability across participants and groups during the perturbation trials, the stability-normalized walking speed was then calculated using the mean anteroposterior MoS of the final 10 steps of each walking trial [(0.4–1.8 m/s) (McCrum et al., 2019b)]. The method and effectiveness of this approach are described in detail elsewhere (McCrum et al., 2019b). For each participant, the walking speed that would result in MoS of 0.05 m was calculated. The walking perturbation protocol then began with participants walking at the stability-normalized speed for 3-4 min, followed by 10 unilateral treadmill belt acceleration perturbations, which occurred unannounced every 30–90 s. The perturbation was a 3 m/s^2 acceleration of the treadmill belt to a maximum speed equal to 180% of the stability-normalized walking speed. The acceleration began when the hallux marker of the to-be-perturbed limb passed the hallux marker of the opposite foot in the sagittal plane. The belt decelerated at toe-off of the perturbed limb. Participants were naïve to the specifics of the perturbation protocol (i.e., limb, type, number, timing, magnitude). The first and tenth accelerations perturbed the right leg, while the second to ninth accelerations perturbed the left leg. This way, not only balance recovery after a novel perturbation, but also adaptation to repeated perturbations can be studied within the same protocol. A schematic overview of the perturbation protocol is shown in **Figure 1**. Further technical details of the perturbations can be found elsewhere (McCrum et al., 2018).

Data Processing

Data processing was conducted in MATLAB (2016a, The MathWorks, Inc., Natick). The three-dimensional coordinates of the markers were filtered using a low pass second order Butterworth filter (zero-phase) with a 12 Hz cut-off frequency. Foot touchdown and toe-off were detected using marker and force plate data, as described previously (McCrum et al., 2019a). The anteroposterior MoS at foot touchdown were calculated as the anteroposterior distance between the anterior boundary of the base of support (BoS) and the extrapolated center of mass, adapted for our validated reduced kinematic model (Hof et al., 2005; Süptitz et al., 2013). The MoS was calculated for the following steps: baseline for each perturbation was the mean MoS of the eleventh to second last step before each perturbation (Base); the final step before each perturbation (Pre); and the first eight recovery steps following each perturbation (Post1-8). The number of steps to return to baseline stability following the perturbation was determined by calculating the number of steps that were within 0.05 m of the MoS value of Base for each individual, counting back from the eighth recovery step, using custom written R code (R version 3.6.0; R Core Team, 2019). Additionally, the means and coefficients of variation of step length, width and time, as well as double support time, were calculated using the foot marker data for 0.4, 0.8, 1.2, and 1.6 m/s unperturbed walking trials.

Analysis

The effects of falls history on MoS recovery after the first perturbation to each leg ($Pert1_R$ and $Pert2_L$; representing the un-adapted response) and the final perturbation to the left leg ($Pert9_L$; representing the adapted response), were analyzed using repeated-measures two-way ANOVA with group (Falls/No-Falls) and step (repeated measures: Base, Pre, Post1-8) as factors for each of the perturbations separately. Additionally, Mann-Whitney tests were applied to compare the groups on number of recovery steps needed for each perturbation and Friedman tests were used to assess the change in steps across perturbations within each group. Finally, the spatial (step length and width means and variability) and temporal (step and double support time means and variability) parameters of gait at a range of walking speeds (0.4, 0.8, 1.2, and 1.6 m/s) were compared between the Falls and No-Falls groups using a two-way ANOVA with group (Falls/No-Falls) and walking speed (repeated measure) as factors.

RESULTS

Twenty older adults (8 with, and 12 without falls in the previous year) were included in this study. Characteristics of participants described by group (Falls/No-Falls) can be found in **Table 1**: participant characteristics. Six of the eight participants in the Falls group fell only once in the previous year, one reported two falls, and one fell three or more times.

Step Parameters

Spatial and temporal parameters of gait, as well as their variability, were compared between groups using two-way repeated-measures ANOVAs. From these analyses, no significant effects of group (Falls vs. No-Falls), and no interaction effects (Group x Speed) were found for any parameter (the complete effect and interaction results can be found in **Supplementary Table 1**).

TABLE 1 | Participant characteristics (mean ± SD).

	Falls group	No-falls group
Men/women (*n*)	4/4	6/6
Age (years)	70.6 ± 3.6	70 ± 2.4
Height (cm)	168.2 ± 15.4	169.4 ± 7.2
Weight (kg)	75 ± 16.3	75.6 ± 10.3
Body mass index	26.3 ± 3.3	26.3 ± 2.9
Stability-normalized walking speed (m/s)	1.29 ± 0.13	1.31 ± 0.14
Falls in the previous year n (frequency)	1 (6), 2 (1), ≥3 (1)	0 (12)

Stability and Adaptability

All participants were able to recover from the walking perturbations without harness assistance. However, due to a technical failure during the first perturbation, one participant was excluded from the analyses involving Pert1$_R$. Two-way repeated-measures ANOVAs for Pert1$_R$, Pert2$_L$ and Pert9$_L$ did not reveal significant effects of falls history on MoS [Pert1$_R$: $F_{(1,\,17)} = 0.89$, $P = 0.36$; Pert2$_L$: $F_{(1,\,18)} = 3.07$, $P = 0.097$; Pert9$_L$: $F_{(1,\,18)} = 3.3$, $P = 0.085$). Significant step by falls history interaction effects on MoS were found for Pert2$_L$ and Pert9$_L$ (Pert1$_R$: $F_{(9,\,153)} = 0.31$, $P = 0.97$; Pert2$_L$: $F_{(9,\,162)} = 5.25$, $P < 0.0001$; Pert9$_L$: $F_{(9,\,162)} = 3.63$, $P = 0.0004$). Dunnett's tests for multiple comparisons were used to compare the MoS for each step to the Base value (results indicated in **Figure 1**). Sidak's tests for multiple comparisons were used to compare the MoS between groups and revealed that only Post2 in Pert2$_L$ was significantly different (**Figure 2**; note that the study was not powered for these pairwise comparisons). Complete Dunnett and Sidak results can be found in the **Supplementary Material**.

The Falls group required averages of 6.3, 5.6, and 5.4 recovery steps and the No Falls group required averages of 6.4, 6.6, and 4.4 recovery steps for Pert1$_R$, Pert2$_L$, and Pert9$_L$, respectively (see **Figure 3**). Mann-Whitney tests did not find significant group differences in number of recovery steps ($U = 37$, $P = 0.7$; $U = 37.5$, $P = 0.44$; $U = 31$, $P = 0.19$). A Friedman test revealed a significant effect of perturbation number on the number of recovery steps in the No Falls group (Friedman statistic = 12.41, $P = 0.002$), with Dunnett's multiple comparisons tests revealing significant differences between Pert9$_L$ and both Pert1$_R$ ($P = 0.018$) and Pert2$_L$ ($P = 0.007$). Due to the missing participant in the Falls group at Pert1$_R$, Wilcoxon signed rank tests were

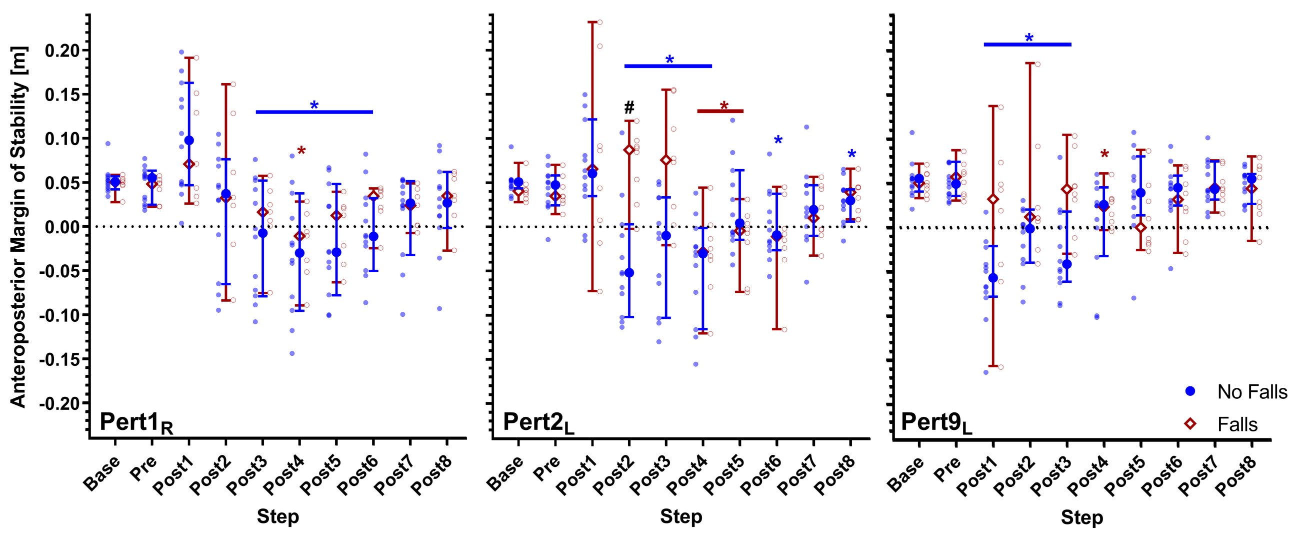

FIGURE 2 | Median and 95% confidence intervals (with individual data points) of the anteroposterior margins of stability during the first, second and ninth perturbations (Pert1$_R$, Pert2$_L$, and Pert9$_L$, respectively) including unperturbed walking prior to each perturbation (Base), the final step prior to each perturbation (Pre) and the first eight recovery steps following the perturbations (Post1–8) for Falls and No-Falls groups. Blue * and Red *: significant difference to Base for the No Falls and Falls groups, respectively ($P < 0.05$; adjusted using Dunnett's multiple comparisons test). #Significant difference between the No Falls and Falls groups ($P < 0.05$; adjusted using Sidak's multiple comparisons test).

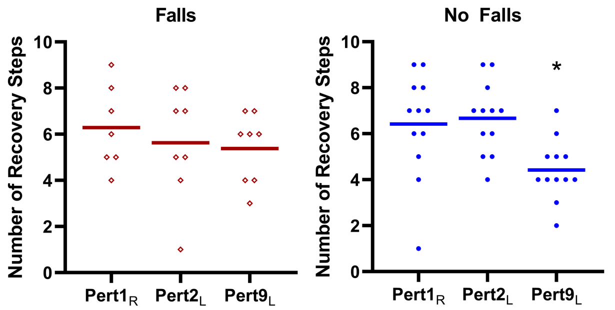

FIGURE 3 | The number of recovery steps (means and individual values) required by the Falls group (left panel) and No Falls group (right panel) for the first, second and ninth perturbations ($Pert1_R$, $Pert2_L$, and $Pert9_L$, respectively). *Significant difference to $Pert1_R$ and $Pert2_L$.

used for this group and did not reveal significant differences in the number of recovery steps needed between $Pert1_R$ and $Pert2_L$ ($P = 0.25$), $Pert1_R$ and $Pert9_L$ ($P = 0.53$) and $Pert2_L$ and $Pert9_L$ ($P > 0.99$).

DISCUSSION

The aim of this study was to address the extent to which walking stability following a single perturbation and walking adaptability following repeated perturbations relate to falls history in older adults. We hypothesized that older adults with a history of falls would demonstrate decreased stability and adaptability compared to older adults without a history of falls. Additionally, we analyzed step variability during unperturbed walking, due to previous indications of increased variability in older adults with a history of falls (Mortaza et al., 2014).

Previous studies indicate differences in variability during unperturbed walking between older adults with and without a history of falls [for a review see Mortaza et al. (2014)]. However, in this study, no significant between-group differences in variability during unperturbed walking were found. Our study used set walking speeds instead of self-selected walking speeds, which may have resulted in differences compared to previous studies [7 out of the 13 studies reviewed in Mortaza et al. (2014) that found significant differences between fallers and non-fallers do not mention accounting for walking speed]. The results of this study (**Supplementary Table 1**) show significant walking speed effects on nearly all parameters, but no significant group effects.

Our results showed no significant effects of falls history on MoS during the first left or right leg perturbations ($Pert1_R$ and $Pert2_L$). However, significant step by falls history interaction effects on MoS were found for $Pert2_L$, with a significant between-group difference in the second recovery step. The middle panel in **Figure 2** shows that the No-Falls group had negative MoS on the second recovery step, while the Falls group still had positive MoS. This may be due to a difference in the recovery response directly after the perturbation, in which the Falls group shows a slightly delayed recovery compared to the No-Falls group. These differences are less pronounced but consistent with findings from another study (McCrum et al., 2020), which compared reactive stability between healthy young and older adults using the same walking perturbation protocol. In that study, older adults had a more posterior extrapolated center of mass in response to the perturbation, resulting in initially more positive MoS but a delayed stability recovery. Additionally, notably greater inconsistency in perturbation recovery responses across the Falls group compared to the No-Falls group can be observed in **Figure 2**, indicating there may be inconsistent recovery strategies in older adults with a history of falls. Despite more inconsistency however, the highest MoS value in the first recovery steps consistently belongs to participants in the Falls group. Combined, these results might hint at a decreased ability to coordinate the dual tasks of maintaining stability and continuing walking on the treadmill with age, and a further decrease in older adults with a history of falls compared with older non-fallers. This is consistent with findings from a study by dos Santos et al. which suggested a tendency for older fallers to favor a "stability-first" strategy, when facing other motor dual-tasks (Dos Santos et al., 2018). In their study, older fallers showed similar walking stability but decreased accuracy when placing a dowel over a target compared to non-fallers. The differences between the Falls and No-Falls groups after the first perturbation found in this study, are insufficiently pronounced to be a useful indicator of falls risk. However, corroborated with the presented literature, they suggest that the ability to coordinate a physical dual-task (combined stability recovery after a walking perturbation and continued treadmill walking) may be related to fall risk in older adults. To clarify this relationship and how it relates to daily-life situations of older adults, future studies may focus on the ability to coordinate various dual-tasks with stability recovery from perturbations during overground walking.

While the results showed no significant group effect, a significant step by falls history interaction on MoS was found for the last left leg perturbation ($Pert9_L$). This indicates a difference between the groups for specific steps after this perturbation. Additionally, high variation in MoS after $Pert9_L$ in the Falls group is observed (indicated by the wider confidence intervals and individual data points), as there was during the early perturbations, and the presence of some high MoS values in the first recovery steps remains. In contrast, the variability in MoS in the No-Falls group has visibly decreased by $Pert9_L$, and there are no longer any high MoS values in this group in the first few recovery steps. Together, this indicates better adaptation in the No-Falls group, who by $Pert9_L$, seem to respond with more consistent and effective recovery responses. Statistically this is substantiated by the significant differences in the number of recovery steps needed to reach close to normal stability values between perturbation 9 and the first two perturbations in the No-Falls group, with no significant differences in the Falls group. These findings are in alignment with results from a study by Pai et al. who demonstrated that adaptability to repeated perturbations during a sit-to-stand task may give an indication of falls risk (Pai et al., 2010a). These findings suggest that with further research, adaptation to repeated walking perturbations may be a useful measure to distinguish between older adults with and without a history of falls.

We hypothesize that recovery to a single novel treadmill acceleration perturbation is too specific a task to assess overall fall risk. The task-specificity of balance is now well established (Patla et al., 1990; Kiss et al., 2018; Ringhof and Stein, 2018) and given that falls can occur in a multitude of ways, this one specific perturbation might not represent or generalize to all possible causes of falls. Reduced adaptability, however, may give a broader indication of the ability of the locomotor system to respond and improve reactive responses to sudden perturbations, which may better generalize to the many situations that could lead to falls. It may also serve as a marker for the health of the locomotor control system (which may, in turn, be linked with falls risk), as reduced adaptability to such perturbations has often been shown in sensory and neurological pathology (Karamanidis et al., 2020). How the proposed relation between adaptability to repeated perturbations, locomotor system health and falls risk presents in daily-life remains unclear, and should be studied further. Additionally, there are many ways that walking adaptability can be assessed, and it is currently unclear if the method of assessment is critical (Geerse et al., 2019). Further research on walking adaptability in various tasks, including repeated external perturbations such as slips or trips, in older fallers and non-fallers, could help address this gap in knowledge.

We included a relatively healthy sample of older adults, resulting in mostly older adults who had experienced a single fall in the Falls group (with no known musculoskeletal or neurological deficits and no history of dizziness, balance or walking complaints), which may decrease the generalizability of the results to more frail populations. However, it is in this relatively healthy part of the older population where other clinical tests are known to have ceiling effects, which makes it important to determine other methods of indicating increased risk of falls for this population (Petterson et al., 2020). Having experienced one or more previous falls is one of the strongest predictors for future falls in community-dwelling older adults (OR 2.8 for all fallers; OR 3.5 for recurrent fallers) (Deandrea et al., 2010).

In conclusion, this study found some small but significant differences in reactive stability and adaptability between older adults with and without a history of falls, but no differences in variability of unperturbed walking. The results indicate that older adults with a history of falls may have decreased ability to coordinate the dual tasks of regaining stability and continuing to walk on the treadmill. The differences between the groups were more pronounced after repeated perturbations, with evidence of better adaptation in the No-Falls group, while increased variability of recovery responses and signs of a different recovery strategy remained in the Falls group. The results from the present study indicate that further research on adaptability to repeated walking perturbations as an indicator of falls history, and how this presents in the daily life of older adults, is warranted.

DATA AVAILABILITY STATEMENT

The original contributions presented in the study are included in the article/**Supplementary Material**, further inquiries can be directed to the Corresponding author.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Medisch Ethische Toetsingscommissie azM/UM, Maastricht University Medical Center/Maastricht University. The patients/participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

MG: conceptualization, investigation, formal analysis, resources, writing – original draft, and writing – review & editing. AL: conceptualization, resources, writing – review & editing, and supervision. KK: methodology, writing – review & editing, and supervision. LG: investigation, resources, data curation, and writing – review & editing. JH: resources, writing – review & editing, supervision, project administration, and funding acquisition. KM: conceptualization, methodology, resources, writing – review & editing, and supervision. CM: conceptualization, methodology, investigation, formal analysis, resources, data curation, writing – original draft, writing – review & editing, visualization, supervision, project administration, and funding acquisition. All authors contributed to the article and approved the submitted version.

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SUPPLEMENTARY MATERIAL

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ORIGINAL RESEARCH
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Aging Affects Lower Limb Joint Moments and Muscle Responses to a Split-Belt Treadmill Perturbation

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Age-related changes cause more fall-related injuries and impede the recoveries by older adults compared to younger adults. This study assessed the lower limb joint moments and muscle responses to split-belt treadmill perturbations in two groups (14 healthy young group [23.36 ± 2.90 years] and 14 healthy older group [70.93 ± 4.36 years]) who performed two trials of unexpected split-belt treadmill perturbations while walking on a programmable split-belt treadmill. A motion capture system quantified the lower limb joint moments, and a wireless electromyography system recorded the lower limb muscle responses. The compensatory limb's (i.e., the tripped limb's contralateral side) joint moments and muscle responses were computed during the pre-perturbation period (the five gait cycles before the onset of a split-belt treadmill perturbation) and the recovery period (from the split-belt treadmill perturbation to the baseline gait relying on the ground reaction forces' profile). Joint moments were assessed by maximum joint moments, and muscle responses were quantified by the normalization (%) and co-contraction index (CCI). Joint moments and muscle responses of the compensatory limb during the recovery period were significantly higher for the YG than the OG, and joint moments (e.g., knee flexion and extension and hip flexion moments) and muscle responses during the recovery period were higher compared to the pre-perturbation period for both groups. For CCI, the older group showed significantly higher co-contraction for biceps femoris/rectus femoris muscles than the young group during the recovery period. For both groups, co-contraction for biceps femoris/rectus femoris muscles was higher during the pre-perturbation period than the recovery period. The study confirmed that older adults compensated for muscle weakness by using lower joint moments and muscle activations and increasing muscle co-contractions to recover balance after split-belt treadmill perturbations. A better understanding of the recovery mechanisms of older adults who train on fall-inducing systems could improve therapeutic regimens.

Keywords: falls, split-belt treadmill perturbation, aging, joint moments, muscle responses

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INTRODUCTION

Falls are the leading cause of injury and death in young and older adults (Berg et al., 1997; Timsina et al., 2017). Falls affect quality of life due to fall-related physical injuries (fractures) and loss of confidence and fear of falling while engaging in daily activities (Tinetti et al., 1988; Salkeld et al., 2000; Gallagher et al., 2001). Falls have occurred during a variety of activities (e.g., walking, running, playing sports, going up/down the stairs, moving between sitting and standing positions, etc.) in young and older adults (Timsina et al., 2017), and walking and sports/exercise activities are the first and second fall-related activities throughout all ages, respectively (Talbot et al., 2005). Considering that walking is an important activity in daily activities (Lawton and Brody, 1969; Gill et al., 1997), unexpected gait perturbations, known as trips and slips, are the major causes of falls in young adults and the community-dwelling elderly (Berg et al., 1997; Heijnen and Rietdyk, 2016; Timsina et al., 2017). For sports activities, a concussion is one of the major sports-related injuries (Hutchison et al., 2015; Kendall et al., 2020). For example, in ice hockey and football sports, a fall to the ice or the ground and a trip are the causes of concussions (Hutchison et al., 2015; Kendall et al., 2020). Concussion results in impaired emotional, neurocognitive, and physical functioning (Conder and Conder, 2015).

Age-related cognitive and visual impairments, changes in neuromuscular mechanisms, and declining muscle strength increase the probability of tripping and falls in older adults (Pijnappels et al., 2008; Bento et al., 2010). Given that lower limbs contribute to balance recovery by creating a new base of support and generating joint moments to control balance stability after a gait perturbation (Wang et al., 2019, 2020; Yoo et al., 2019), reduced lower limb muscle strength caused by aging can be an important fall predictor (Horlings et al., 2008; Pijnappels et al., 2008). Decreased lower limb muscle strength is a limiting factor in preventing falls in older adults (Pijnappels et al., 2008). A previous study indicated that older adults with relatively weaker lower limb muscle strength showed a slower recovery process and higher fall rates after slip perturbations induced by a vinyl tile surface coated with oil compared to young adults (Lockhart et al., 2005). One study found that young adults used higher ankle and knee joint moments to recover balance after slip perturbations induced by a slippery mixture compared to older adults (Liu and Lockhart, 2009), while another found that young adults used higher knee and hip joint moments to avoid an obstacle after trip perturbations induced by the external obstacle compared to older adults (McFadyen and Prince, 2002).

Changes in the neuromuscular system such as decreases in motor unit firing rate and fewer motor units, affect older adults' muscle responses after losing balance (e.g., reduced muscle activation and increased co-contraction of agonist and antagonist muscles) (Tang and Woollacott, 1998; Okada et al., 2001; Lockhart and Kim, 2006; Chambers and Cham, 2007; Pijnappels et al., 2008; Bento et al., 2010; Watanabe et al., 2018). As muscle responses correlate to muscle strengths and joint moments (Buchanan et al., 2005; Watanabe et al., 2018), electromyography (EMG) has been analyzed to understand different muscle responses to gait perturbations for young and older adults (Tang and Woollacott, 1998; Lockhart and Kim, 2006). Older adults showed lower activation rates, smaller amplitudes, and longer onset latencies of lower limb muscles compared to young adults after slip perturbations induced by a moveable platform or a vinyl tile surface coated with a soap and water mixture (Tang and Woollacott, 1998; Lockhart and Kim, 2006).

When older adults confront balance challenges, they increase joint stiffness through muscle co-contraction to compensate for muscle weakness, which is a compensatory strategy to recover balance stability by decreasing the degree of freedom of the body segment's movement (Nagai et al., 2011; Nelson-Wong et al., 2012; Schinkel-Ivy and Duncan, 2018). Increased co-contraction of the gastrocnemius (GAS) and tibialis anterior (TA) muscles was found in older adults after balance perturbations induced by a moving platform (Okada et al., 2001). Older adults also showed higher co-contraction and a longer co-contraction duration at knee and hip muscles after slip perturbations induced by a moveable platform or a contaminated floor (Tang and Woollacott, 1998; Chambers and Cham, 2007).

Gait perturbation systems with external mechanisms have practical limitations because they require enough physical space for overground walking, have fixed locations, etc. (Schillings et al., 1996; Pavol et al., 1999; Cham and Redfern, 2002; Okubo et al., 2018). One alternative is a programmable treadmill with a single belt or dual belts (Sessoms et al., 2014; Mueller et al., 2016). Recently, we developed a fall-inducing system incorporating a split-belt treadmill (Lee et al., 2017a,b; Lee et al., 2019, 2020; Yoo et al., 2019). In our previous studies (Lee et al., 2017a,b, 2019, 2020; Yoo et al., 2019), we confirmed kinematic changes (i.e., increased maximum trunk flexion angle, maximum trunk flexion velocity, and maximum center of mass (COM) velocity) following a split-belt treadmill perturbation induced by instantaneously stopping one belt of our fall-inducing system. These results were similar to the results of previous studies demonstrating increased maximum trunk flexion angle, maximum trunk angular velocity, and COM position in the anterior direction after trip perturbations induced by mechanical obstacles (Pavol et al., 2001; Bieryla et al., 2007).

Although trip perturbations contribute to more fall-related injuries than slip perturbations in older adults (Timsina et al., 2017), and trip and slip perturbations result in different body responses (e.g., the forward and backward loss of balance for trip and slip perturbations, respectively) (Grabiner et al., 1993; Pavol et al., 2001; Pai et al., 2010), most studies have investigated the differences in lower limb joint moments and muscle responses to slip perturbations, rather than trip perturbations, by young and older adults. Therefore, this study assessed lower limb joint moments and muscle responses to split-belt treadmill perturbations in young and older adults. We hypothesized that 1) young adults would show higher lower limb joint moments and muscle activations after split-belt treadmill perturbations and 2) older adults would show higher muscle co-contractions after split-belt treadmill perturbations.

TABLE 1 | Statistical analysis results of the demographic characteristics, walking speeds, and MOCA of the participants ($n = 28$) for young group (YG) and older group (OG).

	YG ($n = 14$)	OG ($n = 14$)	*p* value
Gender (male, %)	Male, 50%	Male, 29%	–
Age (years)	23.36 ± 2.90	70.93 ± 4.36	< 0.0001***
Weight (kg)	70.35 ± 14.20	72.43 ± 12.87	0.688
Height (cm)	172.11 ± 10.59	167.56 ± 7.08	0.192
BMI (kg/m^2)	23.67 ± 3.75	25.85 ± 4.59	0.178
Walking speed (m/s)	0.86 ± 0.06	0.63 ± 0.14	< 0.0001***
MOCA	28.86 ± 1.10	28.07 ± 1.07	0.067

*BMI, body mass index; MOCA, montreal cognitive assessment (*** $p < 0.0001$).*

METHODS

Participants

Based on the results of our previous studies (Lee et al., 2017a,b, 2019, 2020; Yoo et al., 2019) and our pilot study, the power analysis indicated a minimum of 20 participants, with an effect size (f) = 0.67 (large effect size) (Cohen, 1992), power ($1\text{-}\beta$) = 0.80, and $\alpha = 0.05$. Fourteen healthy young adults [7 females and 7 males; Young Group (YG)] and 14 healthy older adults [10 females and 4 males; Older Group (OG)] participated as shown in **Table 1**. No participants had a major operation in the previous 6 months, musculoskeletal dysfunctions, or neurological and peripheral sensory diseases. All participants scored 26 or more in the Montreal Cognitive Assessment (MOCA), representing normal cognitive ability. All participants read and signed a consent form prior to the study, which was approved by the Institutional Review Boards of the University of Houston.

Experimental Procedures

The fall-inducing system consisted of one load cell (LC101-250, Omega Engineering Inc., CT, USA), a programmable split-belt treadmill embedded with two force plates underneath (Bertec Corporation, Columbus, OH, USA), and a VICON motion capture system consisting of 35 reflective passive markers and 12 near-infrared cameras (Vicon Motion Systems Ltd., Oxford, UK), and custom software as shown in **Figure 1**. A wireless EMG system (Trigno ™IM, Delsys Inc., Natick, MA, USA) was used to measure muscle responses.

The Nexus 1.8 software synchronized with the EMG system and the custom software sampled the EMG signals at 2000 Hz and the marker positions and the ground reaction forces (GRFs) at 100 Hz, respectively [see the detailed algorithm information in (Lee et al., 2017a); see the custom software information in (Lee et al., 2017a, 2019; Yoo et al., 2019)]. The custom software generated split-belt treadmill perturbations (at the foot level) by stopping the treadmill's left belt within 100 ms at a rate of 10 m/s^2 at 10% of the gait cycle (approximately the loading phase) determined by GRFs (Lee et al., 2017a, 2020; Yoo et al., 2019). After the non-tripped foot's first heel strike (i.e., the first heel strike of the right foot), the stationary treadmill belt returned to the pre-perturbation speed within 100 ms at a rate of 10 m/s^2 (Lee et al., 2017a, 2020; Yoo et al., 2019). If the peak loading force measured by the load cell exceeded 30% of a participant's body weight, a trial was considered a fall incident (Bhatt et al., 2013; Okubo et al., 2018).

The 35 reflective passive markers were attached bilaterally on the body and the 10 wireless surface EMG sensors were attached to biceps femoris (BF), rectus femoris (RF), TA, and GAS (lateralis and medialis) muscles as shown in **Figure 1B**. All participants wore a safety harness and selected their own comfortable walking speed (0.86 ± 0.06 m/s for YG and 0.63 ± 0.14 m/s for OG) by increasing or decreasing the treadmill's speed until they felt comfortable as shown in **Table 1**. No participants received instructions (e.g., how to respond, recover, etc.) and there were no practice trials.

Since a previous study indicated that motor adaptation to trip perturbations occurred from the third trial compared to the first trial, all participants performed 2 consecutive split-belt treadmill perturbation trials (Wang et al., 2012) consisting of standing (15 s standing), pre-perturbation (steady walking at the self-selected walking speed with 31 to 40 gait cycles), and recovery (steady walking at the self-selected walking speed after the split-belt treadmill perturbation) periods. There was a 20 s rest period between the trials. There were no marker or system malfunctions.

The fall-inducing treadmill randomly induced a split-belt treadmill perturbation between the 31st and 40th steps at the left foot based on a study indicating that the number and percentages of recovery and fall were not affected by the side of the tripped foot (Pavol et al., 1999). Each trial ended after 15 steps from the split-belt treadmill perturbation to provide adequate recovery for the OG, since healthy young adults needed nearly 5 s for recovery (Lee et al., 2017a).

Data Processing

Using the Nexus 1.8 software, the full body Plug-in-Gait model filtered the 35 reflective passive marker positions by a sixth-order Butterworth filter with a cut-off frequency of 6 Hz (Yoo et al., 2019) and computed the ankle, knee, and hip moments in the sagittal plane based on the filtered marker positions. The Plug-in-Gait model is a standard and reliable tool in biomechanics research for analyzing joint moments, especially for the sagittal plane (Kadaba et al., 1989). Previous studies found that the whole-body movement predominated in the sagittal plane after trip perturbations (Lee et al., 2017a, 2019). The low-pass filter was applied to the GRFs by a second-order Butterworth filter with a cut-off frequency of 10 Hz using MATLAB (The MathWorks, Natick, MA, USA) (Lee et al., 2019). A band pass filter was applied to the EMG signals by a fifth-order Butterworth filter with a low cut-off frequency of 20 Hz and a high cut-off frequency of 300 Hz (Lee et al., 2019).

Since the compensatory limb's stepping response (the contralateral side of tripped limb) is the general response to gait perturbations (Jensen et al., 2001; Maki and McIlroy, 2006; Yoo et al., 2019), joint moments (i.e., maximum right ankle dorsiflexion and plantarflexion, knee flexion and extension, and hip flexion and extension moments) and EMG signals (right BF, RF, TA, and GAS muscles) of the compensatory limb

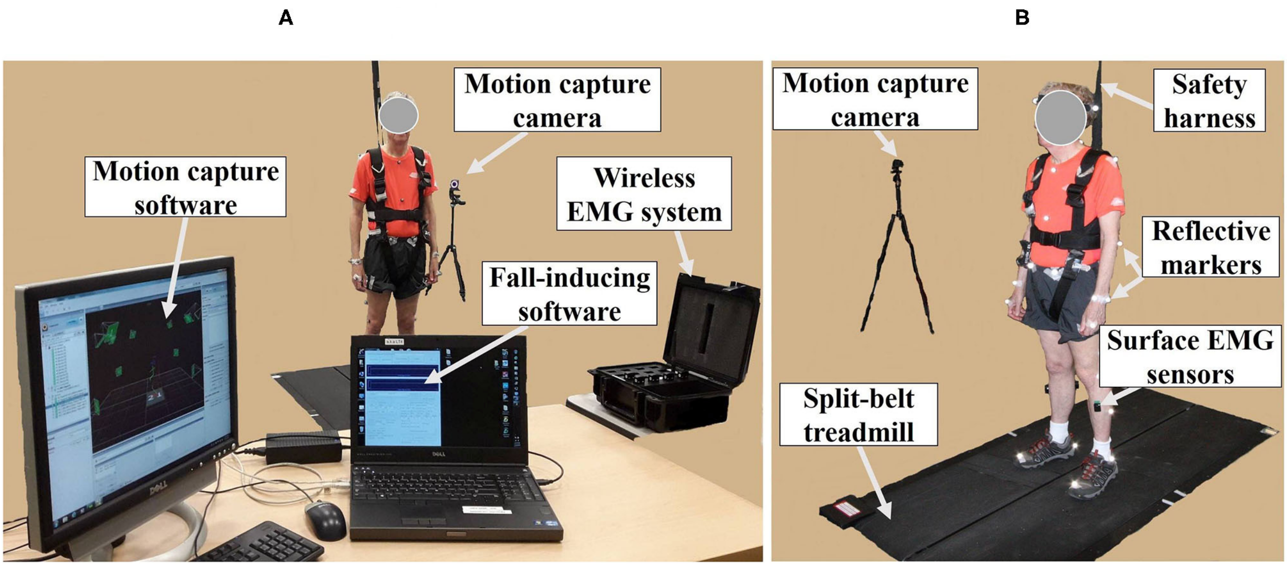

FIGURE 1 | Experimental apparatus. **(A)** Motion capture system; wireless electromyography (EMG) system; fall-inducing system. **(B)** Reflective markers and surface EMG sensors placed on body landmarks; split-belt treadmill; safety harness; motion capture camera.

were analyzed. Since our previous study found no significant difference between EMG signals from the medial and lateral GAS before and after split-belt treadmill perturbations (Lee et al., 2019), the EMG signals from the medial and lateral GAS were averaged and analyzed. The joint moments and EMG signals were computed for the standing, pre-perturbation, and recovery periods. The recovery period was defined as the period from the instant of split-belt treadmill perturbation to return to baseline gait depending on the GRF's profile (i.e., when a correlation coefficient attained 95%). To compute the recovery period, each of the 5 gait cycle's GRFs before the split-belt treadmill perturbation was normalized (i.e., the GRFs ranged from 0 to 100% corresponding to the gait cycle) and then averaged. Each gait cycle's GRFs after the split-belt treadmill perturbation was also normalized and compared to the averaged GRFs before the split-belt treadmill perturbation with Pearson's correlation coefficient.

Consistent with our previous kinematic analysis (Lee et al., 2019), joint moments of pre-perturbation and recovery periods were normalized to the averaged joint moments of the standing period to remove baseline joint moments (i.e., averaged joint moments while standing) from joint moments during pre-perturbation and recovery periods. For EMG analysis, the filtered EMG signals were rectified and enveloped by the root-mean-square (RMS) using a 20 ms moving window (Begalle et al., 2012). EMG signals were normalized based on the maximum value during normal gait (Heiden et al., 2006; Qu et al., 2012). The EMG signals of 5 normal gait cycles during the pre-perturbation period were averaged. Next, the maximum value and the pre- and post-values from the maximum on the averaged signals were averaged for normalization and defined as a normalization value. For normalization, the mean of averaged EMG signals during the pre-perturbation period and the mean of EMG signals during the recovery period were divided by the normalization value and multiplied by 100, respectively.

$$\frac{\text{Mean of averaged EMG signals during pre} - \text{perturbation period}}{\text{normalization value}} * 100\%$$

$$\frac{\text{Mean of EMG signals during recovery period}}{\text{normalization value}} * 100\%$$

CCI (i.e., between TA and GAS (the average of the medial and lateral GAS) and BF and RF, respectively) was computed for the pre-perturbation and recovery periods, respectively, as shown in **Figure 2**, based on Falconer and Winter (Falconer and Winter, 1985) as:

$$CCI = \frac{2I_{ant}}{I_{total}} * 100\%,$$

where I_{ant} is the area of the lower signals at any point between two muscle activity signals and I_{total} is the sum of the area of the lower signals and the area of higher signals at any point between two muscle activity signals.

Statistical Analysis

Statistical analysis was performed by SPSS (IBM Corp., Armonk, NY, USA) for join moments, normalized muscle activations, and CCI. Levene's test and the Shapiro-Wilk test, respectively, confirmed the homogeneity of variances and normal distributions of the outcome measures. An independent *t*-test was used to compare the YG and OG demographic characteristics, self-selected walking speeds, and MOCA. Two-way analysis of variance (ANOVA) was performed for six join moments (maximum right ankle dorsiflexion and plantarflexion, knee flexion and extension, and hip flexion and extension moments), four normalized muscle activations (EMG signals for right BF, RF, TA, and GAS muscles), and two CCI (CCI between

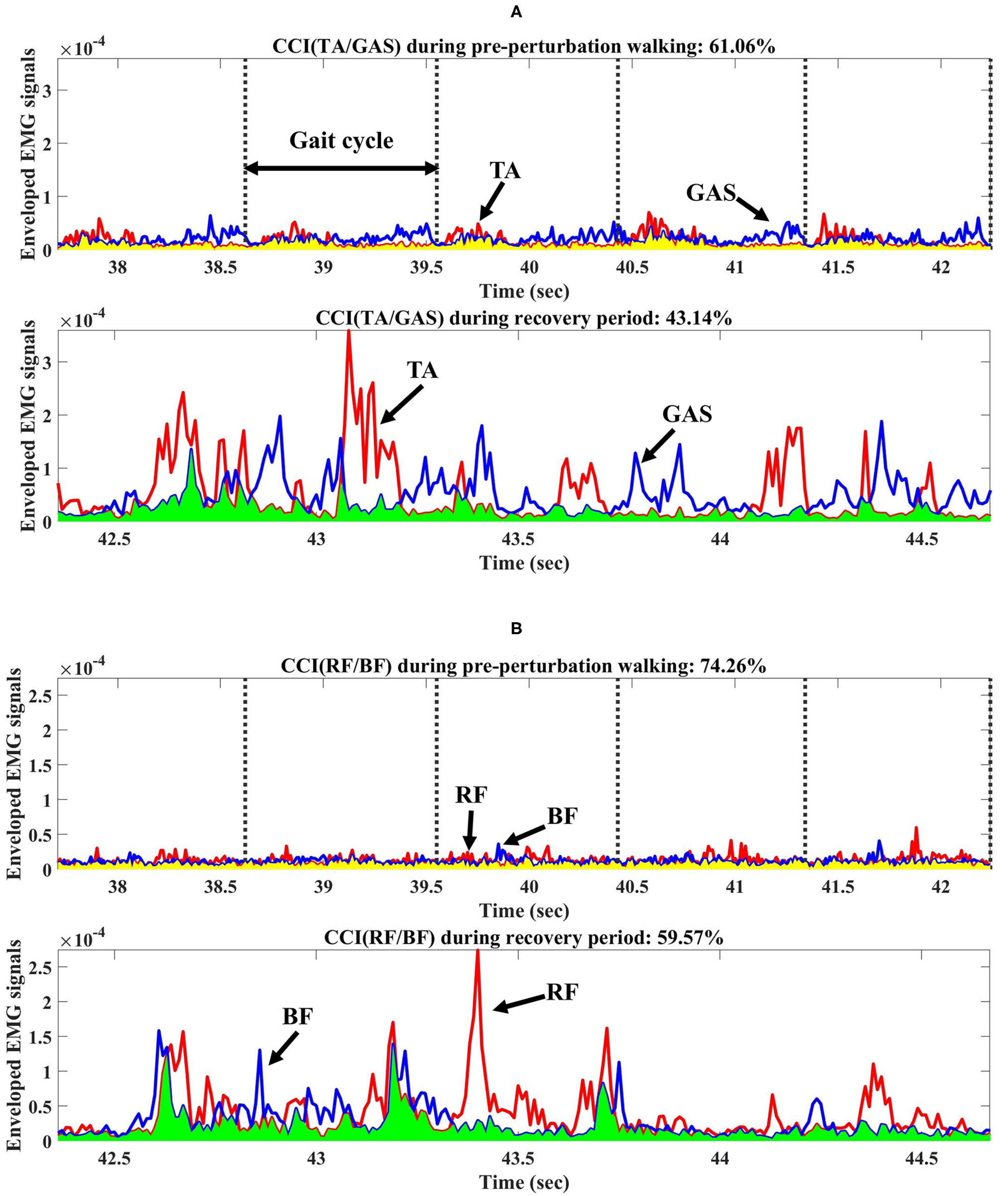

FIGURE 2 | Co-contraction index (CCI) from one participant during pre-perturbation and recovery periods. **(A)** CCI of TA/GAS during pre-perturbation and recovery periods. **(B)** CCI of RF/BF during pre-perturbation and recovery periods. TA, GAS, RF, and BF indicate tibialis anterior, gastrocnemius, rectus femoris, and biceps femoris, respectively. Yellow shaded area indicates the area under the lower signals at any point between two muscle activity signals during pre-perturbation period used for CCI calculations. Light green shaded area indicates the area under the lower signals at any point between two muscle activity signals during recovery period.

TABLE 2 | Statistical analysis results of joint moments for group (G), period (P), and interaction (G × P) (*$p < 0.05$, **$p < 0.01$, and ***$p < 0.0001$).

Joint moments	Effects	DF	*F* value	*p* value
Dorsiflexion moment	G	1, 108	9.448	0.003**
	P	1, 108	0.011	0.916
	G × P	1, 108	2.296	0.133
Plantarflexion moment	G	1, 108	1.860	0.175
	P	1, 108	2.074	0.153
	G × P	1, 108	0.003	0.953
Knee flexion moment	G	1, 108	3.886	0.051
	P	1, 108	49.796	< 0.0001***
	G × P	1, 108	6.411	0.013*
Knee extension moment	G	1, 108	5.813	0.018*
	P	1, 108	60.595	< 0.0001***
	G × P	1, 108	4.241	0.042*
Hip flexion moment	G	1, 108	27.684	< 0.0001***
	P	1, 108	107.505	< 0.0001***
	G × P	1, 108	11.955	0.001**
Hip extension moment	G	1, 108	1.274	0.262
	P	1, 108	0.118	0.732
	G × P	1, 108	0.188	0.665

The period indicates the pre-perturbation period and recovery period.

TA and GAS and CCI between BF and RF) to evaluate the main effects of the period (pre-perturbation and recovery periods), groups, and their interactions. An F test was used to identify the main effects and the interaction effects, and *post-hoc* analysis (Šídák method) was conducted to confirm the influence of any factors on the main and interaction effects. The significance levels for statistical analyses were set at $p < 0.05$.

RESULTS

Demographic Characteristics and Recovery Steps

The results of the independent t-test showed that age and walking speed differed significantly, whereas there were no significant differences for weight, height, BMI, and MOCA between the two groups as reported in **Table 1**. Following the split-belt treadmill perturbations, the YG and OG returned to their normal walking in 1.93 ± 1.02 steps and 2.43 ± 1.23 steps, respectively.

Joint Moments

The results of the two-way ANOVA showed a significant main effect of the group for ankle dorsiflexion moments and no significant main and interaction effects for ankle plantarflexion moment as reported in **Table 2**. *Post-hoc* analysis indicated that the YG showed significantly higher ankle dorsiflexion moments during the recovery period than the OG ($p = 0.002$) as shown in **Figure 3A**. However, ankle plantarflexion moments were not significantly different within and between groups as shown in **Figure 3B**.

For knee joint moments, the results of the two-way ANOVA showed a significant main effect of the period and the interaction effect for knee flexion moments, and significant main effects of the group and period, and the interaction effect for knee extension moments, respectively as shown in **Table 2**. *Post-hoc* analysis indicated that the YG showed significantly higher knee flexion moments ($p = 0.002$) and extension moments ($p = 0.002$) during the recovery period than the OG as shown in **Figures 3C,D**. Both groups showed significantly increased knee flexion moments (YG: $p < 0.0001$ and OG: $p = 0.002$) and knee extension moments (YG: $p < 0.0001$ and OG: $p < 0.0001$) during the recovery period compared to the pre-perturbation period.

For hip joint moments, the two-way ANOVA indicated significant main effects of the group and period, and the interaction effect for hip flexion moment, and no significant main and interaction effects for hip extension moment as shown in **Table 2**. *Post-hoc* analysis indicated that the YG showed significantly higher hip flexion moments during the recovery period compared to the OG ($p < 0.0001$) as shown in **Figure 3E**. Both groups showed significantly increased hip flexion moments during the recovery period compared to the pre-perturbation period (YG: $p < 0.0001$ and OG: $p < 0.0001$). However, hip flexion moments were not significantly different within and between groups as shown in **Figure 3F**.

Muscle Responses

The results of the two-way ANOVA indicated a significant main effect of the period for the TA and GAS muscles as shown in **Table 3**. *Post-hoc* analysis indicated that the YG showed significantly higher TA muscle activation during the recovery period than the OG ($p = 0.034$). Both groups showed significantly higher muscle activations during the recovery period compared to the pre-perturbation period (YG: $p < 0.0001$ and OG: $p < 0.0001$) as shown in **Figure 4A**. Both groups showed significantly higher muscle activations for the GAS muscle during the recovery period compared to the pre-perturbation period (YG: $p < 0.0001$ and OG: $p < 0.0001$) as shown in **Figure 4B**.

The results of the two-way ANOVA indicated significant main effects of the group and period and the interaction effect for the RF and BF muscles as reported in **Table 3**. *Post-hoc* analysis indicated that the YG showed significantly higher muscle activations during the recovery period than the OG (RF ($p < 0.0001$) and BF ($p < 0.0001$)) as shown in **Figures 4C,D**. Both groups showed significantly higher RF (YG: $p < 0.0001$ and OG: $p = 0.001$) and BF (YG: $p < 0.0001$ and OG: $p < 0.0001$) muscle activations during the recovery period compared to the pre-perturbation period.

The results of the two-way ANOVA indicated a significant main effect of the period [$F_{(1, 108)} = 44.170$, $p < 0.0001$] and the interaction effect [$F_{(1, 108)} = 7.725$, $p = 0.006$] for the CCI of BF/RF (**Figure 4F**), and an insignificant main effect and interaction effect for the CCI of TA/GAS (**Figure 4E**). *Post hoc* analysis indicated that the OG showed significantly higher CCI of BF/RF during the recovery period ($p = 0.001$) than the YG. Both groups showed higher CCI of BF/RF during the pre-perturbation period compared to the recovery period (YG: $p < 0.0001$ and OG: $p = 0.007$).

DISCUSSION

This study investigated young and older adults' joint moments and muscle responses of the compensatory limb to split-belt

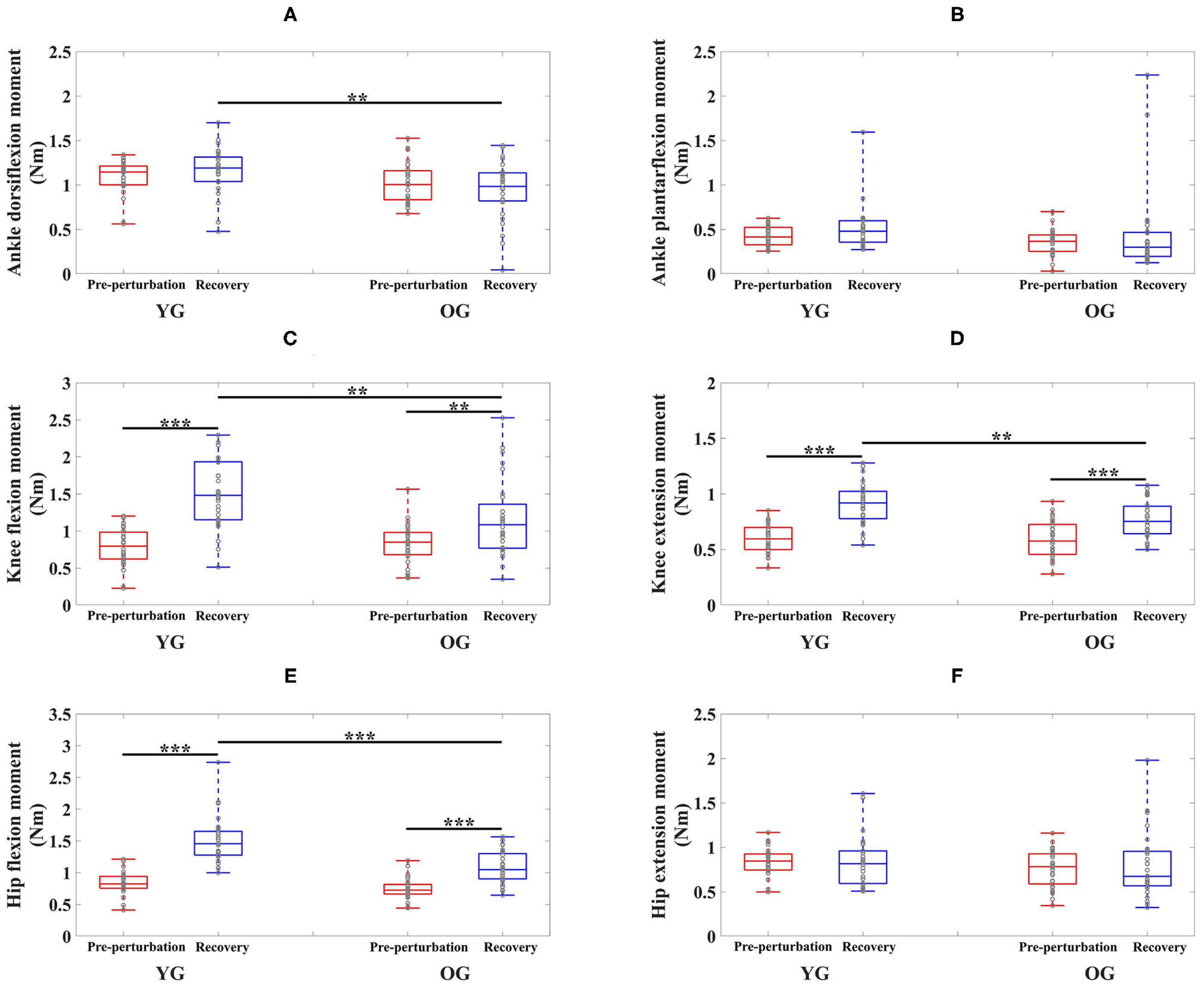

FIGURE 3 | Box plots of joint moments as a function of group (YG and OG) and period across all participants (**$p < 0.01$ and ***$p < 0.0001$). **(A)** Ankle dorsiflexion moment. **(B)** Ankle plantarflexion moment. **(C)** Knee flexion moment. **(D)** Knee extension moment. **(E)** Hip flexion moment. **(F)** Hip extension moment. YG and OG indicate young group and older group, respectively. Gray circles superimposed on box plots indicate individual data points from all participants.

treadmill perturbations. Both groups showed overall increased joint moments and muscle activations during the recovery period compared to the pre-perturbation period. The YG showed higher joint moments and muscle activations during the recovery period than the OG, whereas the OG showed higher muscle co-contractions of BF/RF during the recovery period.

This study confirmed previous findings of increased joint moments of the compensatory limb (e.g., increased peak ankle, knee, and hip joint moments) by older adults to control the body's forward rotation and to gain time and clearance for repositioning the tripped foot after trip perturbations induced by an external obstacle (Pijnappels et al., 2004; 2005). This study, however, indicated insignificant increases in plantarflexion and hip extension moments after split-belt treadmill perturbations induced by stopping one belt of a split-belt treadmill, which contradicted a previous study indicating increased plantarflexion and hip extension moments after trip perturbations induced by an external obstacle (Pijnappels et al., 2005). Since a small sample size affect statistical power (Suresh and Chandrashekara, 2012), this result could not be generalized. However, we speculate that the plantarflexion and hip extension moments contributing to push-off for foot clearance of the tripped limb were unnecessary after split-belt treadmill perturbations (Pijnappels et al., 2005).

Two studies found increased TA, GAS, RF, and BF muscle activations of the compensatory limb to control the dynamic body's forward angular momentum after trip perturbations induced by an external obstacle (Pijnappels et al., 2005) and by the split-belt treadmill compared to normal walking (Lee et al., 2019). This study's similar results for the YG and OG indicated that increased TA (125.8% for the YG and 87.99% for the OG), GAS (98.70% for the YG and 78.44% for the OG), RF (225.94% for the YG and 108.49% for the OG), and BF (201.37% for the YG

TABLE 3 | Statistical analysis results of normalized muscle activation for group (G), period (P), and interaction (G×P) ($^{**}p < 0.01$ and $^{***}p < 0.0001$).

Normalization	Effects	DF	*F* value	*p* value
Tibialis anterior	G	1, 108	1.709	0.194
	P	1, 108	120.132	< 0.0001***
	G × P	1, 108	2.969	0.088
Gastrocnemius	G	1, 108	3.224	0.075
	P	1, 108	75.514	< 0.0001***
	G × P	1, 108	0.021	0.885
Rectus femoris	G	1, 108	13.456	< 0.0001***
	P	1, 108	65.802	< 0.0001***
	G × P	1, 108	10.211	0.002**
Biceps femoris	G	1, 108	12.730	0.001**
	P	1, 108	81.904	< 0.0001***
	G × P	1, 108	9.659	0.002**

The period indicates the pre-perturbation period and recovery period.

and 106.54% for the OG) muscle activations during the recovery period compared to the pre-perturbation period contributed to restraining the body's forward rotation. Given that joint moments are associated with muscle activations (Zajac and Gordon, 1989; Pijnappels et al., 2005), both the YG and OG showed significantly increased muscle activations and joint moments after split-belt treadmill perturbations with the exception of the dorsiflexion, plantarflexion, and hip extension moments. For insignificantly changes in the dorsiflexion, plantarflexion, and hip extension moments, two possible explanations could be speculated. First, a relatively small sample size may contribute to these results. Second, considering joint moments in the frontal and transverse planes and the sagittal plane contributed to balance recovery after gait perturbations induced by a moveable platform (Liu and Lockhart, 2009), increased TA, GAS, BF muscle activations may in fact contribute to generating ankle and hip joint moments in the frontal and transverse planes.

Previous studies indicated significantly higher lower limb joint moments and muscle responses (e.g., amplitude and activation rate) in young adults after gait perturbations compared to older adults (Tang and Woollacott, 1998; McFadyen and Prince, 2002; Lockhart and Kim, 2006; Liu and Lockhart, 2009). This study indicated that the YG showed higher joint moments after split-belt treadmill perturbations compared to the OG (23.94% higher dorsiflexion moment, 28.84% higher knee flexion moment, 18.46% higher knee extension moment, and 38.86% higher hip flexion moment, respectively). The YG also showed higher muscle activations after split-belt treadmill perturbations than the OG (15.35% higher TA, 72.65% higher RF, and 57.95% higher BF, respectively). Muscle strength in older adults progressively decreased due to sarcopenia (Roubenoff, 2000; Akima et al., 2001), and muscle mass reduced nearly 30–50% caused by the reduction volume and number of muscle fibers with aging (Lexell et al., 1988; Granacher et al., 2008). Considering that aging decreased older adults' muscle strength related to joint moments and muscle activations (Horlings et al., 2008; Pijnappels et al., 2008; Watanabe et al., 2018), decreasing muscle strength may contribute to relatively lower joint moments and muscle activations after split-belt treadmill perturbations. Since resistance training contributed to reversing the age-related loss of muscle strength (Pijnappels et al., 2008), resistance training for lower limb muscles may help to prevent falls.

This study indicated significantly higher CCI of BF/RF after the OG's split-belt treadmill perturbations (18.51% higher than the YG). Previous studies demonstrated higher the CCI of hip muscles (e.g., RF and BF) after slip perturbations induced by a moveable platform or a contaminated floor in older adults compared to young adults (Tang and Woollacott, 1998; Chambers and Cham, 2007). Older adults increase co-contraction of agonist and antagonist muscles to compensate for muscle weakness after a loss of balance (Okada et al., 2001; Chambers and Cham, 2007; Pijnappels et al., 2008; Nagai et al., 2011). Co-contraction increases joint stiffness, which contributes to balance stability by reducing the degree of freedom of the movements of the body segments (Nagai et al., 2011; Nelson-Wong et al., 2012; Schinkel-Ivy and Duncan, 2018). Based on the results of this study, older adults increase the CCI of BF/RF as a compensatory strategy to recover balance after split-belt treadmill perturbations.

This study indicated that the CCI of BF/RF for both groups was significantly lower during the recovery period compared to the pre-perturbation period, unlike a previous study which found a higher CCI of vastus lateralis/hamstring after slip perturbations compared to normal walking. The difference may be due to the increased trunk and COM range of motions after slip perturbations compared to trip perturbations (Lee et al., 2019), and the smaller hip range of motion and limits of stability of the feet in a backward direction than in a forward direction (Humphrey and Hemami, 2010; Lee et al., 2014). This study assumed that slip perturbations may require higher co-contraction by activating both hamstring and vastus lateralis muscles for joint stiffness as a compensation strategy to control for difficult stability challenges. This study indicated that RF muscles related to knee extensions and hip flexions activated more than BF muscles to control the forward rotation of the body after split-belt treadmill perturbations, and may have resulted in a relatively lower CCI during recovery compared to pre-perturbation periods as shown in **Figure 2B**.

Self-selected comfortable walking speeds were slower than previous studies (∼1.15 and 1.05 m/s preferred walking speeds on a treadmill for young adults and older adults, respectively) (Plotnik et al., 2015; Lazzarini and Kataras, 2016). Since no participants received instructions (e.g., how to respond to a perturbation) and there were no practice trials, this study assumed they walked carefully during trials. Previous studies indicated that awareness of upcoming perturbations affected walking performance (e.g., slower walking speeds and shorter step lengths) (Bohm et al., 2015; Okubo et al., 2018).

Our fall-inducing system could be used for train balance-constrained individuals and athletes to improve their responses (e.g., joint moments and muscle responses) after multiple split-belt treadmill perturbations. Compared to fall-inducing systems requiring external mechanisms (Schillings et al., 1996; Pavol et al., 1999; Cham and Redfern, 2002; Okubo et al., 2018), our fall-inducing system using a split-belt treadmill requires less space,

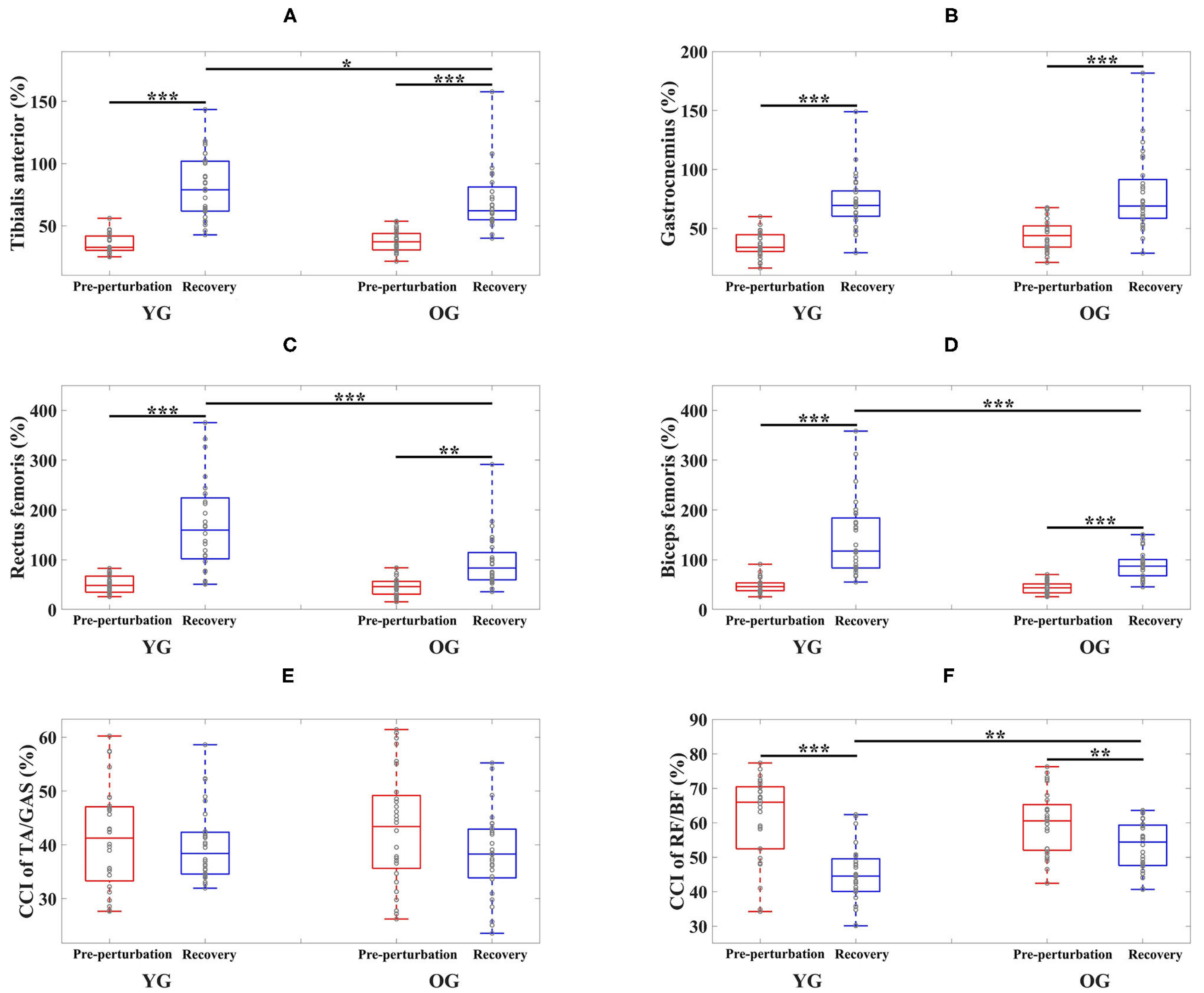

FIGURE 4 | Box plots of normalized muscle activations and co-contraction index (CCI) as a function of group (YG and OG) and period across all participants (*$p < 0.05$, **$p < 0.01$, and ***$p < 0.0001$). **(A)** Tibialis anterior. **(B)** Gastrocnemius. **(C)** Rectus femoris. **(D)** Biceps femoris. **(E)** CCI of TA/GAS. **(F)** CCI of RF/BF. YG and OG indicate young group and older group, respectively. TA, GAS, RF, and BF indicate tibialis anterior, gastrocnemius, rectus femoris, and biceps femoris, respectively. Gray circles superimposed on box plots indicate individual data points from all participants.

offers more precise control of perturbation intensity or increases perturbation intensity gradually, and provides less predictability of gait perturbations with any number of steps before the onset of perturbations. Since perturbation-based gait training reduced and prevented falls in different populations (McCrum et al., 2017), a fall-inducing system using a split-belt treadmill is expected to perturbation-based gait training in clinical or athletic settings.

This study was limited by relatively small sample size. Given that sample size is positively correlated with statistical power (Suresh and Chandrashekara, 2012), a relatively small sample size could limit to generalization of our findings. This study was also limited by gender imbalance in the OG. Joint moments in the frontal and transverse planes, joint stiffness, and the onset detection of muscle activity were not investigated. Future research will increase the sample size, balance gender, and examine joint moments in the frontal and transverse planes, joint stiffness, and the detection of muscle onset. Future research will also investigate the level of physical activity impacts on falls.

CONCLUSION

This study characterized the joint moments and muscle responses of the compensatory limb after split-belt treadmill perturbations. Older and younger adults' compensatory limb's joint moments and muscle responses to split-belt treadmill perturbations were compared. Overall, young adults showed higher joint moments and muscle activations during recovery periods after split-belt treadmill perturbations. Older adults showed a higher CCI of

BF/RF during recovery periods. This study characterized the joint moments and muscle responses of the compensatory limb after split-belt treadmill perturbations by older adults. Given that gait perturbations encountered during normal walking are a major cause of falls in older adults, the results could teach older adults who train on fall-inducing systems how to compensate for unexpected gait perturbations.

DATA AVAILABILITY STATEMENT

The original contributions presented in the study are included in the article, further inquiries can be directed to the corresponding author.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Institutional Review Boards of the University of Houston. The patients/participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

DY collected data, performed data and statistical analysis, interpreted results, and drafted the manuscript. JA assisted with data collection, data analysis, and manuscript preparation. K-HS reviewed the manuscript. B-CL conceived the study, designed the experimental protocols, supervised the research, and edited and reviewed the manuscript. All authors contributed to the article and approved the submitted version.

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Acute Effects of a Perturbation-Based Balance Training on Cognitive Performance in Healthy Older Adults: A Pilot Study

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Aging is accompanied by an alteration in the capacity to ambulate, react to external balance perturbations, and resolve cognitive tasks. Perturbation-based balance training has been used to induce adaptations of gait stability and reduce fall risk. The compensatory reactions generated in response to external perturbations depend on the activation of specific neural structures. This suggests that training balance recovery reactions should show acute cognitive training effects. This study aims to investigate whether exposure to repeated balance perturbations while walking can produce acute aftereffects that improve proactive and reactive strategies to control gait stability and cognitive performance in healthy older adults. It is expected that an adaptation of the recovery reactions would be associated with increased selective attention and information processing speed. Twenty-eight healthy older adults were assigned to either an Experimental (EG) or a Control Group (CG). The protocol was divided in 2 days. During the first visit, all participants completed the Symbol Digit Modalities Test (SDMT) and the Trail Making Test (TMT). During the second visit, a cable-driven robot was used to apply waist-pull perturbations while walking on a treadmill. The EG was trained with multidirectional perturbations of increasing intensity. The CG walked for a comparable amount of time with cables on, but without experiencing perturbations. Before and after the training, all participants were exposed to diagonal waist-pull perturbations. Changes in gait stability were evaluated by comparing the distance between the heel of the leading leg and the extrapolated Center of Mass (Heel-XCoM Distance—HXD) at perturbation onset (PON) and first compensatory heel strike (CHS). Finally, the cables were removed, and participants completed the SDMT and the TMT again. Results showed that only the EG adapted the gait stability ($p < 0.001$) in reaction to diagonal perturbations and showed improved performance in the SDMT ($p < 0.001$). This study provides the first evidence that a single session of perturbation-based balance training produce acute aftereffects in terms of increased cognitive performance and gait stability in healthy older adults. Future studies will include measures of functional activation of the cerebral cortex and examine whether a multi-session training will demonstrate chronic effects.

Keywords: gait, perturbations, balance, cognition, adaptation, aging

INTRODUCTION

Gait, balance, and cognitive disorders are serious problems in late life (Holtzer et al., 2007; Snijders et al., 2007). Aging is generally accompanied by a declining capacity to resolve cognitive tasks (Holtzer et al., 2007; Staudinger, 2015), ambulate (Snijders et al., 2007; James et al., 2016), and react to external balance perturbations (Maki and Mcilroy, 2006; Martelli et al., 2017a). These are regarded as apparent signs of many pathologies leading to falls, a major public health concern for our society (Berg et al., 1997).

The ability to walk and the efficacy of compensatory responses to maintain balance rely not only on the sensorimotor system, but also critically depend on cognitive functioning (Horak, 2006; Snijders et al., 2007; Sturnieks et al., 2012; Morris et al., 2016). Cognitive performance is strongly associated with characteristics of gait and balance (Morris et al., 2016), compensatory responses (Sturnieks et al., 2012), and locomotor adaptability (Caetano et al., 2017). When older adults walk or are exposed to balance perturbations while simultaneously engaged in a cognitively demanding task, performance is impaired in one or both tasks (Woollacott and Shumway-Cook, 2002). The cerebral cortex is directly involved in controlling rapid balance reactions but also keeping the central nervous system prepared to optimize balance recovery reactions even when a future threat to stability is unexpected (Bolton, 2015).

The aptitude to adapt to the environment is essential for walking and compensating for instabilities (Snijders et al., 2007; Caetano et al., 2017). Cognitive (Staudinger, 2015) and locomotor (Bohm et al., 2015; Krishnan et al., 2018) adaptability is still preserved in older age. The control of the compensatory responses required after a balance perturbation can be strengthened (Pai and Bhatt, 2007; Pai et al., 2014; Bohm et al., 2015; Liu et al., 2017). The exposure to repeated external disturbances induces motor adaptations that lead participants to better correct their balance during the recovery phase (i.e., adaptation of the reactive strategy) and modify the volitional control of stability in the face of a possible perturbation (i.e., adaptation of the proactive strategy) (Bohm et al., 2015). As a result, after repeated perturbation-based balance training (PBT) sessions, participants may also show longer-term effects of improved recovery after unexpected loss of balance encountered in daily life, thus reducing their risk for falling (Grabiner et al., 2014; Pai et al., 2014; Mansfield et al., 2015; Gerards et al., 2017; Mccrum et al., 2017; Okubo et al., 2017). Similarly, single bouts of moderate exercise are able to induce acute physiological responses that have a positive impact on the brain and on cognitive performance, as assessed by behavioral measures (Lambourne and Tomporowski, 2010; Chang et al., 2012; Netz, 2019). Cumulative effects of exercise have been associated with increases in brain volume (Colcombe et al., 2006; Hillman et al., 2008; Herold et al., 2019) and cognitive performance (Angevaren et al., 2008; Smith et al., 2010), more efficient brain functioning (Voelcker-Rehage et al., 2011), and attenuated cognitive decline (Lautenschlager et al., 2012).

To our knowledge, the impact of exposure to repeated balance perturbations on cognitive performance has not yet been studied. The compensatory reactions generated to control dynamic balance in response to external perturbations are not merely segmental reflexes organized at the level of the spinal cord, but rather depend on the integration of proprioceptive, visual, and vestibular information implicating many levels of the central nervous system (Horak, 2006; Maki and Mcilroy, 2006; Jacobs and Horak, 2007; Bolton, 2015; Varghese et al., 2017). Cognitive resources are needed to recognize a disturbance of balance and then rapidly initiate a recovery step, maintain balance on a single limb, and navigate the contralateral limb to regain stability. Besides the sense of balance, it requires specific cognitive abilities such as selective attention and speed of information processing (Snijders et al., 2007). This suggests that training dynamic balance recovery reactions also should show acute cognitive training effects.

The present study aims to investigate to what extent the exposure to repeated balance perturbations while walking can produce acute improvements in gait stability and cognitive performance in community-dwelling, healthy older adults. We hypothesized that a single-session of gait training encompassing unpredictable waist-pulls would be more effective than unperturbed walking in improving reactive and proactive control of gait stability and cognitive performance in terms of information processing speed and selective attention.

MATERIALS AND METHODS

Participants

Twenty-eight healthy community-living older adults were randomly assigned to either the Experimental Group (EG: 14 subjects, 3 males) or the Control Group (CG: 14 subjects, 3 males). Inclusion criteria included: (i) living independently in the community, (ii) at least 65 years old; (iii) absence of acute, severe, or unstable medical illness; (iv) not reporting any significant neural, muscular, or skeletal disease, and (v) able to safely walk on a treadmill without mobility aids. Participants were informed about the research procedure and signed a written consent form approved by the Institutional Review Board of Columbia University, before participating.

Procedure

Participants were asked to come to the lab on two occasions (**Figure 1**). During the first session, they completed

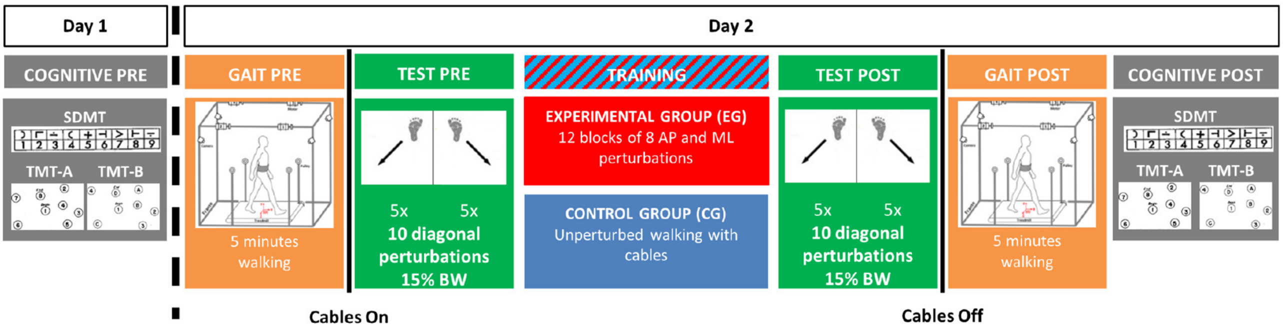

FIGURE 1 | Experimental Protocol. The experiment was broken in two visits. During Day 1, (COGNITIVE PRE), participants completed the Symbol Digit Modalities Test (SDMT) and the Trail Making Test Part A (TMT-A) and Part B (TMT-B). During Day 2, participants first walked without cables at their preferred speed on the treadmill for 5 minutes (GAIT PRE). Then, cables were attached, and they were exposed to 10 diagonal perturbations (TEST PRE). In the TRAINING session, the Experimental Group (EG) was trained with 12 blocks of 8 antero-posterior (AP) and medio-lateral (ML) perturbations of increasing intensity. The Control Group (CG) walked on the treadmill with the cables attached for the same amount of time as the EG (i.e., 24 minutes). Both groups were then exposed to the same set of perturbations delivered before the training session (TEST POST). Then, the cables were removed, and all participants walked again for 5 minutes (GAIT POST). Finally, after a 10-minute resting period, all participants completed the same cognitive tests of Day 1 (COGNITIVE POST).

questionnaires describing their study cohort, a battery of functional tests and they were tested for baseline cognitive performance (COGNITIVE PRE) using the Symbol Digit Modalities Test (SDMT) (Sheridan et al., 2006) and the Trail Making Test (TMT) (Tombaugh, 2004). During the second session, usually occurring within 1 week of the first, the experimental intervention took place using the Active Tethered Pelvic Assist Device (A-TPAD), an innovative cable-driven robot conceived for gait rehabilitation able to apply controlled force-moments at the human pelvis in any direction and precise instants of the gait cycle (Vashista et al., 2015) (**Figure 2**). In this configuration, the A-TPAD is used to apply multidirectional waist-pull perturbations while walking on a treadmill (Martelli et al., 2016, 2017b,c, 2018).

Participants were equipped with the pelvic brace necessary to apply the perturbations, a harness to protect them from falling, and reflective markers that allow to collect kinematic data (**Figure 2**). Preferred treadmill walking speed was determined for each participant and then maintained during the experiment. Speed was determined by gradually increasing the speed by 0.1 m/s until the subject reported that was too fast and then reducing it by 0.1 m/s. All participants first walked on the treadmill for 5 min. Subsequently, cables were attached to the brace. All subjects were exposed to 10 diagonal perturbations while walking (TEST PRE). Perturbations consisted of 5 pulls with Motor 2 (back-right perturbation) triggered at right heel strike and 5 pulls with Motor 4 (back-left perturbation) triggered at left heel strike (**Figure 2**, left panel). The first perturbation was delivered at right heel strike and then the order of perturbations was alternated. Peak force was fixed at 15% of the Body Weight (BW). Then, the EG was exposed to 12 blocks of 8 Antero-Posterior (AP) and Medio-Lateral (ML) perturbations of increasing intensities (TRAINING). In each block, 4 directions (forward, backward, leftward and rightward), and 2 events (right and left heel strikes) were used. At the beginning, the peak force was 15% and 5% BW for AP and ML perturbations, respectively. Every four blocks, the peak force was increased by 5% BW. The order of the perturbations in each block was chosen randomly. The range of intensity of the perturbations was determined based on previous experiments with healthy young subjects (Martelli et al., 2016, 2017b, 2018). The CG did not receive any perturbation during the training session, but they walked on the treadmill with the cables attached for the same amount of time as the EG. In order to reduce the risk of fatigue, the treadmill was stopped every four blocks (for the EG) or 8 min (for the CG) and subjects were told they could rest at any time if they felt tired. All participants were then exposed to the same set of perturbations delivered before the training session (TEST POST). All perturbations were delivered while walking at constant speed and consisted of a trapezoidal force profile (rise, hold and fall times of 150 ms duration each). The time between perturbations was chosen randomly (5–15 s). Participants were aware that they could be perturbed at the waist when the cables were attached, but were not informed about the magnitude, the direction or the timing of the perturbations. Before the intervention started, they were instructed to maintain balance and keep walking. Then, the cables were removed, and all participants walked for another 5 min (GAIT POST). For all the duration of the experiment, subjects wore a safety harness to prevent them from falling but without restricting their movements. Finally, after a 10-min resting period, all participants completed the SDMT and the TMT as in the first session (COGNITIVE POST). All participants completed the experiment without difficulty. Inspection of video recording images confirmed that all participants were able to recover their balance without being assisted by the safety harness. Technical problems resulted in missing the TEST POST data for one participant in the CG group.

Measures

Cognitive Performance

The SDMT requires individuals to identify nine different symbols corresponding to the numbers 1 through 9, and manually fill the

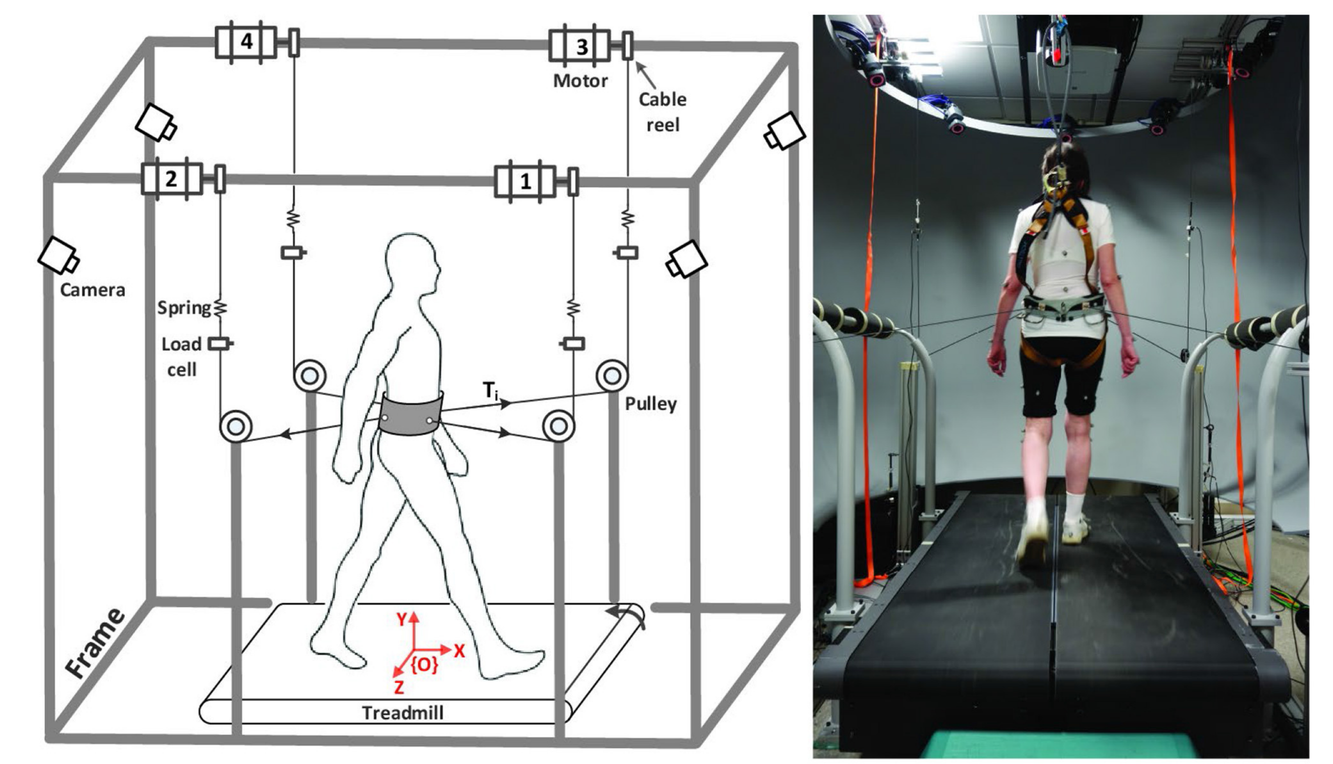

FIGURE 2 | Experimental Setup. Schematic of the Active Tethered Assistive Pelvic Device (A-TPAD) and a picture of a participant while walking with it. Four AC servo motors are mounted on a rigid frame and connected through cables to a fabric hip belt worn by the subject. A load cell and a spring are installed in series with each cable. A closed-loop controller ensures delivery of the correct tensions in the motors. Cables are routed using pulleys to be diagonally directed. The heights of the pulleys were changed for each subject such that during standing each cable was almost parallel to the floor (range: 5°, −20°). Participants walk at constant speed on a split-belt treadmill (Bertec Instrumented Treadmill). A ten-camera motion capture system (Vicon Bonita-10 series), the load cells, and the force plates embedded in the treadmill are used as a part of the controller. The motion capture system is used to track the cable orientation. The force plates are used to detect heel strikes in real time (vertical ground reaction force threshold at 50 N) and time the application of perturbations. When cables were attached to the subject, a constant force of 25 N is applied by each motor to prevent cable slackening. Waist-pull perturbations with peak force of a desired amplitude proportional to the subject's body weight (BW) are provided by applying a transient pulse on one or two of the four cables. The controller is implemented on a LabVIEW, (National Instrument, PXI real time system).

blank space under each symbol with the corresponding number as fast as possible. Two scores were calculated: total number of correct answers given in 90 s (SDMT-C), and time to complete all 110 blank spaces (SDMT-T). The TMT consists of two parts (TMT-A and TMT-B). TMT-A requires an individual to draw lines sequentially connecting 25 encircled numbers distributed on a sheet of paper. Task requirements are similar for TMT-B except the person must alternate between numbers and letters (e.g., 1, A, 2, B, 3, C, etc.). The score on each part represents the amount of time required to complete the task without considering the number of mistakes. In case an error was made, the participant was instructed to return to the "circle" where the error originated and continue. Participants were instructed to complete each test as quickly and accurately as possible.

Biomechanical Measures

The trajectories of 55 reflective markers were collected at 200 Hz using a 10-camera motion capture system (Vicon Bonita-10 series). Missing kinematic data were estimated by means of cubic spline interpolations. High-frequency related noise was removed from digitized coordinates by low-pass filtering data (zero-lag, fourth-order Butterworth low-pass filter) with a cut-off at 10 Hz. Heel strikes were determined as the first point in the descending phase of the lateral malleolus' marker in which the vertical position did not decrease more than 2 mm for two consecutive time frames (Alton et al., 1998). Missing or false gait events were manually checked. A 13-segment biomechanical model was used to calculate the trajectory of the body Center of Mass (CoM) (Martelli et al., 2017c).

In order to maintain balance, it is necessary to control the relative position and velocity between the moving body's CoM and the moving base of support (BoS) (Patla, 2003; Hof, 2008). The extrapolated center of mass (XCoM) (Hof et al., 2005; Hof, 2008) represents the state of the CoM when taking into account both its position and velocity and was calculated as:

$$\mathrm{XCoM}_{x,y} = \mathrm{CoM}_{x,y} + \mathrm{VCoM}_{x,y}/\sqrt{g/l}$$

where $\mathrm{CoM}_{x,y}$ and $\mathrm{VCoM}_{x,y}$ are the AP and ML components of the CoM position and velocity vectors, l is the estimated pendulum length based on the instantaneous distance between the body CoM and the ankle joint of the leading leg and g is the gravitational acceleration. The $VCoM_{x,y}$ was calculated as the first derivative of $CoM_{x,y}$ by using the three-point central differences method. The treadmill speed was added to the $VCoM_x$.

Gait stability was quantified during the TEST PRE and TEST POST using the 2D Euclidean distance between the heel marker of the leading leg and the XCoM (i.e., Heel-XCoM Distance—HXD). Two events were identified: Perturbation onset (PON) and the compensatory heel strike (CHS—first heel strike after PON). HXD-PON was used to identify stability before the perturbation started and possible proactive adaptations in the gait pattern (note that at PON the perturbation force is still at zero). HXD-CHS was used to identify stability after the perturbation and possible reactive adaptations of the early compensatory reaction.

Anthropometric, Socio-Demographic, and Functional Measures

Further assessments were performed during the first visit to verify that participants' age, body height, body weight, socio-demographic, and levels of balance, mobility, and fear of falling were comparable. Socio-demographic data were assessed with a self-report questionnaire that provided information regarding education, family status, housing, occupation, life satisfaction and general health status. The life satisfaction and general health subparts score ranged from 1 (low satisfaction/health) to 5 (high satisfaction/health). Functional measures included the Berg Balance Scale (BBS) (Berg et al., 1992), the Short Physical Performance Battery (SPPB) (Guralnik et al., 1994), and the Falls Efficacy Scale International (FES-I) (Yardley et al., 2005).

Statistical Analysis

Anthropometric characteristics (i.e., age, body height, body mass), preferred treadmill speed, levels of health status and life satisfaction, scores obtained in the BBS, SPPB and FES-I in the two groups were compared by means of independent samples t-tests. Level of education was compared by the Kruskal-Wallis test.

Mixed design analyses of variance (ANOVAs) were performed using the HXD and cognitive test scores (SDMT-C, SDMT-T, TMT-A, and TMT-B) as dependent variables. For the cognitive test scores, a 2-way ANOVA was used. Group (EG and CG) and session (PRE and POST) were used as between- and within-subject factors, respectively. For the HXD, a preliminary ANOVA was performed to confirm that values were similar for perturbations delivered at right or left heel strikes. Since that no significant effect of side was detected, the average value obtained during the 5 perturbations delivered at right heel strike was used in the analysis. A 3-way ANOVA was used with an additional within-subject factor: the time of the gait cycle (PON and CHS) at which the HXD was evaluated. At TEST PRE, it is expected that all participants would show a significant difference between HXD-PON and HXD-CHS due to the effect of the waist-pulls. At TEST POST, it is expected that the CG would still show differences while the EG—that has been exposed to repeated multidirectional perturbations during the TRAINING—would be able to adapt gait stability and "cancel" the effect of the perturbation in a single step, such that the HXD at CHS would be similar to the HXD at PON. Given the exploratory nature of the study, the alpha level of the ANOVAs tests was not adjusted. Significant interaction effects were followed up by Tukey's Honest significance tests. The Lilliefors test, Levene's test for equality of error variances, and the Mauchly's tests were performed to check the normality, homoscedasticity, and sphericity assumptions, respectively. Statistical significance was set at $p < 0.050$.

RESULTS

Table 1 describes characteristics EG and CG. The two groups did not differ on any of the sample characteristics ($p > 0.539$, **Table 1**). All participants reported high levels of subjective well-being and subjective health and were positively biased toward higher levels of education.

At PRE, participants showed an average HXD-PON of 123.5 $\pm$ 38.7 mm. Waist-pull perturbations caused a disruption of normal walking, such that, at the following heel strike, participants showed an HXD-CHS of 182 $\pm$ 7 mm. The results of the ANOVA revealed that the HXD showed a significant effects of session ($p < 0.001$), time of the gait cycle ($p < 0.001$), group$\times$session interaction term ($p = 0.006$), time$\times$session interaction term ($p = 0.019$), and group$\times$session$\times$time interaction term ($p = 0.030$, **Figures 3A,B**). Further analysis revealed that: (i) the HXD changed from PRE to POST for participants in the EG (PRE: 160.4 $\pm$ 61.9 mm; POST: 122.5 $\pm$ 39.8 mm; $p < 0.001$) but not for the participants in the CG (PRE: 145.8 $\pm$ 67.0 mm; POST: 136.7 $\pm$ 56.2 mm; $p = 0.431$); (ii) both HXD-PON (PRE: 132.1 $\pm$ 32.9 mm; POST: 115.6 $\pm$ 31.5 mm; $p = 0.005$) and HXD-CHS (PRE: 188.8 $\pm$ 71.8 mm; POST: 129.4 $\pm$ 46.9 mm; $p < 0.001$) showed significant changes from PRE to POST for the EG; (iii) due to the steep decrements of HXD-CHS at POST, significant differences between HXD-PON and HXD-CHS for the EG were observable only at PRE ($p = 0.010$) but not at POST training ($p = 0.686$). On the contrary, neither HXD-PON (PRE: 115.0 $\pm$ 43.2 mm; POST: 106.2 $\pm$ 35.8 mm; $p = 0.680$) and HXD-CHS (PRE: 176.5 $\pm$ 73.7 mm; POST: 167.3 $\pm$ 57.3 mm; $p = 0.951$) showed significant changes from PRE to POST for the CG. As a result, the HXD-CHS and the HXD-PON were significantly different at both PRE and POST sessions for the CG ($p < 0.006$).

In regards to the cognitive tests, the ANOVA revealed significant effects of the session (SDMT-C: $p = 0.001$; SDMT-T: $p < 0.001$) and group$\times$session interaction term (SDMT-C: $p = 0.022$; SDMT-T: $p = 0.040$) for both SDMT-C and SDMT-T. Further analysis showed that: (i) only the EG increased the number of correct answers given in 90 s from PRE to POST (PRE: 41.3 $\pm$ 8.4; POST: 49.5 $\pm$ 11.3; higher SDMT-C, $p < 0.001$, **Figure 3C**); (ii) only the EG completed the 110 items more quickly from PRE to POST (PRE: 244.5 $\pm$ 41.1 sec; POST: 206.5 $\pm$ 42.1 sec; lower SDMT-T, $p < 0.001$, **Figure 3D**); and (iii) the EG showed a higher SDMT-C ($p = 0.030$, **Figure 3C**) and lower SDMT-T ($p = 0.036$, **Figure 3D**) compared to the CG at POST. On the contrary, the CG did not modify neither SDMT-C (PRE: 39.4 $\pm$ 6.3; POST: 40.9 $\pm$ 8.2, $p = 0.421$) or SDMT-T from PRE to POST (PRE: 253.4 $\pm$ 33.3 sec; POST: 243.1 $\pm$ 45.3 sec, $p = 0.264$, **Figure 3D**). No significant main or interaction effects of

TABLE 1 | Descriptive statistics[a] of participants in Experimental Group (EG) and Control Group (CG).

	Experimental Group (EG) (n = 14)	Control Group (CG) (n = 14)	p-values
Age [years]	70.7 ± 3.7	71.6 ± 4.7	0.598
Height [m]	1.66 ± 0.73	1.65 ± 0.80	0.652
Body mass [kg]	70.5 ± 14.4	71.6 ± 12.6	0.824
Treadmill Speed [m/s]	0.83 ± 0.18	0.83 ± 0.20	1.000
BBS [0–56]	55.0 ± 1.8	54.9 ± 1.1	0.903
SPPB [1–12]	10.7 ± 1.2	10.4 ± 1.2	0.539
FES-I [16–64]	21.1 ± 5.3	21.4 ± 4.7	0.881
Life Satisfaction [1–5]	4.3 ± 0.7	4.4 ± 0.6	0.633
Subjective Health [1–5]	4.5 ± 0.5	4.2 ± 0.6	0.194
Education [≤Highschool, >Highschool]	7.1, 92.9%	7.1, 92.9%	1.000

[a]*Values are reported as Means ± Standard Deviations. BBS, Berg Balance Scale; SPPB, Short Physical Performance Battery; FES-I, Falls Efficacy Scale International.*

group and session were found for either the TMT-A ($p > 0.203$, **Figure 3E**) or the TMT-B ($p > 0.086$, **Figure 3F**).

DISCUSSION

This study aimed to investigate if the exposure to repeated balance perturbations delivered while walking would induce acute adaptations of gait stability and cognitive performance in community-dwelling, healthy older adults. Research on cognitive and locomotor adaptability during balance-demanding tasks is highly important, as it may contribute to the design of effective methods to early detect and remediate gait and cognitive deficits. As hypothesized, results showed that the exposure to multidirectional waist-pull perturbations induced acute modifications of the (i) recovery reaction in terms of stability both before and after the perturbation onset; and (ii) cognitive task performance, as measured by the SDMT. This study provides the first evidence that systematic perturbations of gait induce acute changes in cognitive functioning.

Replicating our previous results (Martelli et al., 2017b,c, 2018), we showed that participants in the EG were able to adapt their capacity to counteract diagonal waist-pull perturbations. Both reactive (HXD-CHS) and proactive (HXD-PON) adaptations in the EG were primarily accounted for by a reduced distance between the XCoM of the body and the heel of the leading leg (i.e., lower HXD-PON and HXD-CHS at POST, **Figures 3A,B**). Such changes allowed the EG to compensate for the instability created by the waist-pull in a single step (i.e., at POST, the HXD at CHS was similar to the HXD at PON). The ability to better control the relationship between the XCoM and the BoS while walking and in reaction to different kinds of perturbations has been shown in young adults as compared to older adults (Bierbaum et al., 2010) and has been associated with a reduced risk of falling (Lugade et al., 2011). It can be argued that proactive adjustments were made predominantly with feedforward control implemented by the central nervous system to increase stability before the perturbation actually started (Bhatt et al., 2006). This modification was beneficial to start the reaction to the perturbation from a more stable position and ideally reduce the reliance on the reactive corrections after the onset of the perturbation. Reactive adjustments to external unanticipated perturbations are largely influenced by the central nervous system as well (Horak, 2006; Jacobs and Horak, 2007; Bolton, 2015; Varghese et al., 2017). The recovery reactions against repeated balance perturbations can bypass some stages of information processing due to a change in the central set developed from prior experience (Horak, 2006). These fast responses recalibrate a previously constructed motor memory without the need of developing a new motor pattern. After the initiation of the compensatory reaction, the cerebral cortex can also modulate late-phase or change-in-support responses characteristics through direct control (Bolton, 2015).

As hypothesized, participants in the EG also showed acute changes in cognitive functioning. The compensatory reactions to perturbations delivered while walking are characterized by fast changes in the base of support that requires challenges of spatial navigation, coordination and affordances in the surrounding environment (Maki and Mcilroy, 2007). The amount of sensorimotor and cognitive processing required to maintain balance and the specific domains involved depend on the type and complexity of the task. We can assume that exposure to repeated perturbations may have been linked with increased activation of cognitive control processes, especially the ones dedicated to processing speed, integration of motion, and navigation to a higher degree than unperturbed walking (Snijders et al., 2007; Sturnieks et al., 2012; Senden et al., 2014; Patel and Bhatt, 2015; Wittenberg et al., 2017). This cognitive activation may have continued to facilitate the speed of mental processes in the EG once the cables were removed and may have contributed to the improvements in the SDMT.

For the first time, we were able to show that a single session of perturbation-based balance training (PBT) can affect cognitive performance in older adults. In relation to the exercise-cognitive relationship, activities can be classified into physical (i.e., aerobic and strength) and motor training (i.e., balance, coordination and flexibility) (Netz, 2019). Studies that compared

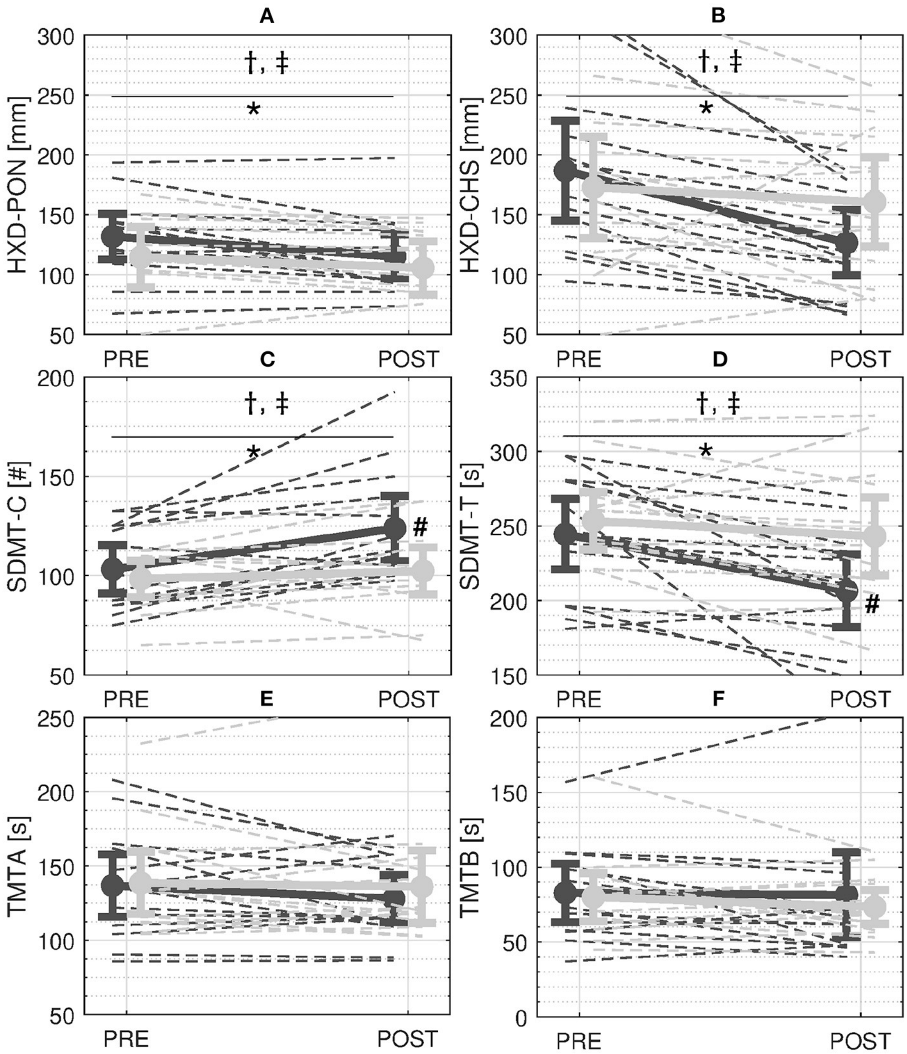

FIGURE 3 | Results. **A** and **B**: Stability measures during perturbed walking with cables—Average Heel-XCoM Distance (HXD) at perturbation onset (PON) **(A)** and compensatory heel strike (CHS) **(B)**. **C** and **D**: Scores of the Symbol Digit Modalities Test (SDMT)—Average number of correct answers in 90 s (SMDT-C) **(C)** and time to complete 110 answers (SMDT-T) **(D)**. **E** and **F**: Scores of the Trail Making Test (TMT)—Time to complete part A (TMT-A) **(E)** and time to complete part B (TMT-B) **(F)**. Solid and dashed dark and light gray lines represent the mean and single-subject changes in the Experimental Group (EG) and the Control Group (CG) from PRE to POST, respectively. Bars refer to 95% confidence interval of the mean. †, ‡ symbols indicate a significant main effect of session and group × session interaction term respectively. *and # symbols indicate a significant effect of the Tukey's Honest significance test for the within-subject and between-subject factors, respectively.

the cognitive improvements of motor and physical training concluded that both are beneficial, but motor training may better stimulate changes in information processing, especially the ability to handle visual and spatial information (Paffenbarger et al., 2001; Netz, 2019). Another important difference between these two training modes is the driving mechanism that affects the cognitive function (Netz, 2019). During physical training, it is the intensity of the exercise that influences neuroplasticity. In contrast, during motor training, it is the complexity of the task that has an effect on cognitive improvements (Carey et al., 2005; Pesce, 2012; Netz, 2019). PBT falls inside the second category of exercise mode and it can be considered as a motor training task with high complexity and neuromuscular demands. Our preliminary results confirm that balance training is effective in improving speed of information processing and the introduction of unanticipated balance perturbations in the task could be particularly beneficial for this domain of cognitive functioning. However, more research is needed to assess the dose-response relationship between level of complexity and cognition for motor activities (Netz, 2019). While intensity is measurable, complexity is hard to measure, and thus the dose-response effect of motor activities on cognition is difficult to determine. PBT is usually implemented with platforms able to impose controlled, standardized and repeatable perturbations (Mccrum et al., 2017). Therefore, they could be ideal for analyzing the dose-response relationship by comparing the effects on cognitive performance of PBT sessions in which complexity is controlled by adjusting the amplitude of balance perturbations. Future studies will further investigate this critical aspect.

Even though in this study no measures of cortical neurophysiological functioning were collected, we can speculate that the repeated exposure to perturbations may have caused changes in brain activity, the so-called perturbation-evoked response (Bolton, 2015; Mierau et al., 2015; Varghese et al., 2017), that were functional to improve the performance in the SDMT. A number of cognitive structures become activated in response to both expected and unexpected perturbations including brain areas generally considered to be involved in executive control such as the pre-frontal cortex and the fronto-central cortical region (Bolton, 2015; Patel et al., 2018). Similarly, the SDMT requires recruiting cerebral networks interconnecting fronto-parietal areas related to selective attention processes, occipital areas related to visual attention, the cerebellum (Forn et al., 2013), and the anterior and posterior corpus callosum known to connect to pre-frontal, parietal and motor cortical areas involved in sensory integration, decision making and motor response (Gawryluk et al., 2014).

Despite an increment in the SDMT, both TMT-A and TMT-B did not show any significant modifications in both groups (**Figures 3E,F**). This may be because the cognitive mechanisms that underlie the TMT are not the ones that are mainly involved while reacting to balance perturbations. Even if the TMT is one of the most widely used instruments in neuropsychological assessment as an indicator of speed of cognitive processing and executive function, an in depth analysis reveals that the TMT-A requires mainly visuoperceptual abilities, and the TMT-B primarily reflects working memory and secondarily task-switching ability (Sanchez-Cubillo et al., 2009). On the contrary, the SDMT is a neuropsychological test with high reliability and ideal to measure information processing speed and selective attention. This is because it is an easy task involving a short time-frame, and carrying out it does not allow use of alternative strategies as is often the case for other tasks indexing executive functioning (Forn et al., 2013).

Even though the results of this study are promising, several limitations need to be considered when interpreting results. Our sample of 28 older adults was rather small which limited the power of the statistical tests. Yet, the fact that nevertheless significant group differences were found is encouraging. Preferred treadmill speed was slow for both groups and equal to 0.83 m/s. This may be related to the procedure used to determine it. Participants were reminded that the walking speed would stay the same throughout the experiment once determined. This may have led the participants to choose a more conservative speed as 'too fast' to avoid getting tired. In other words, we think that this is more of a psychological than a functional effect. This assumption is supported by the fact that for both groups the BBS and the SPPB showed performance higher than 54 (out of 56) and 10 (out of 12) points. These scores would rank both groups as highly functioning older adults with low risk of falls (Bogle Thorbahn and Newton, 1996; Veronese et al., 2014). Determining the preferred walking speed on the treadmill was a measure to tailor the experimental procedure to the respective participant. We do not think that walking at slower speed influenced the results concerning the impact of the gait perturbation intervention. Similar experiments, without testing the cognitive function, were conducted with healthy young subjects (Martelli et al., 2016, 2017b). In these experiments, young participants walked at a speed of about 1.1 m/s. Results were similar and showed an adaptation of the post-training gait pattern as well as the recovery reactions. Accordingly, we do not expect that a faster walking speed for older participants would have yielded a different outcome. Moreover, by keeping walking speed slow, we further reduced the risk of fatigue, a possible confounding effect on cognitive and motor performance. Even if the effects of acute physical fatigue on cognitive performance post-exercise have been unclear (Brisswalter et al., 2002; Lambourne and Tomporowski, 2010), several factors associated with peripheral fatigue could lead to the appearance of central fatigue and a decrease in cognitive performance. Further studies should control for metabolic expenditure and analyze the propensity and functional implications of fatiguability.

We only included two cognitive tests, consequently we are not able to delineate more precisely which dimensions of cognitive aging may profit from the gait perturbation intervention and which may not. In addition, we did not include any specific cognitive screening to determine cognitive impairments. However, performance levels in the SDMT and TMT were similar to the ones obtained by non-clinical adults of similar age in the literature (Tombaugh, 2004; Sheridan et al., 2006) (**Figures 3C–F**). Accordingly, we can infer that our sample was not characterized by any severe cognitive deficits. The same cognitive test was presented at baseline and post-training, thus creating possible practice effects (Goldberg et al., 2015). The introduction of a control group partially obviated this problem, yet we cannot rule out that the perturbation-based balance training facilitated the learning of solving the task rather than processing speed itself. Further studies involving larger samples, a wider range of older adults, multiple baseline assessments and additional cognitive tests are necessary to confirm our conclusions. Moreover, we cannot reject the hypothesis that the perturbations experienced by the EG may have also increased participants' vigilance and arousal level as compared to the CG. Possibly such arousal could have been maintained through the resting period after the intervention and facilitated speeded mental processes and improved cognitive task performance (Lambourne and Tomporowski, 2010). Future studies should therefore include an attention control group to better tease apart arousal from sensorimotor effects on cognitive improvements. Moreover, both the participants and the investigators should be blinded to group assignment to ensure objectivity. Future studies should also include measures of cortical function to investigate directly alterations of the brain function. Finally, the present study focused on acute and not lasting effects. Future studies have to examine whether a multi-session and more extensive training will demonstrate chronic effects on walking balance and cognitive performance.

DATA AVAILABILITY STATEMENT

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by the Institutional Review Board of Columbia University. The participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

DM, US, and SA conceptualized and designed the study. DM, JK, and FA collected the data and worked on the hardware and software of the system. DM cleaned and analyzed the data, performed the statistical analysis, prepared a first draft of the manuscript, and created the figures and tables. US and SA supervised the project. All authors contributed extensively to the work presented in this paper, commented on the manuscript throughout the editorial process, and approved the final submitted version.

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Perturbation Training for Fall-Risk Reduction in Healthy Older Adults: Interference and Generalization to Opposing Novel Perturbations Post Intervention

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This study examined the effects of perturbation training on the contextual interference and generalization of encountering a novel opposing perturbation. One hundred and sixty-nine community-dwelling healthy older adults (69.6 ± 6.4 years) were randomly assigned to one of the three groups: slip-perturbation training (St, $n = 67$) group received 24 slips, trip-perturbation training (Tt, $n = 67$) group received 24 trips, and control (Ctrl: $n = 31$) group received only non-perturbed walking trials (ClinicalTrials.gov NCT03199729; https://clinicaltrials.gov/ct2/show/NCT03199729). After training, all groups had 30 min of rest and three post-training non-perturbed walking trials, followed by a reslip and a novel trip trial for St, a retrip and a novel slip trial for Tt, and randomized novel slip and trip trials for Ctrl. The margin of stability (MOS), step length, and toe clearance of post-training walking trials were compared among three groups to examine interferences in proactive adjustment. Falls, MOS at the instant of recovery foot touchdown, and hip height of post-training perturbation trials were investigated to detect interferences and generalization in reactive responses. Results indicated that prior adaptation to slip perturbation training, resulting in walking with a greater MOS (more anterior) and a shorter step length ($p < 0.01$) than that of the Ctrl group, would be associated with a greater likelihood to forward balance loss if encountered with a trip. The trip adaptation training mainly induced a higher toe clearance during walking ($p < 0.01$) than the Ctrl group, which could lead to reduced effectiveness of the reactive response when encountered with a novel slip. However, there was no difference in the reactive MOS, limb support, and falls between the control group and the slip and trip training groups on their respective opposing novel perturbation post-training (MOS, limb support, and falls for novel slip: Tt = Ctrl; for the novel trip: St = Ctrl, both $p > 0.05$). Current findings suggested that, although perturbation training results in proactive adjustments that could worsen the reactive response (interference) when exposed to an unexpected opposing perturbation, older adults demonstrated the ability to immediately generalize the training-induced adaptive reactive control to maintain MOS, to preserve limb support control, and to reduce fall risk.

Keywords: SLIP, TRIP, perturbation, fall, contextual interference

INTRODUCTION

Falls are the leading cause of injury-related deaths among older adults regardless of their physical function and activity level (Rubenstein et al., 1994; Morley, 2002; Spaniolas et al., 2010). Falls often occur without any signs or warnings, even among the healthiest older adults. Large environmental postural disturbances most often lead to slip- or trip-related falls, which comprise 28–53% of outdoor falls (Luukinen et al., 2000; Talbot et al., 2005; Antes et al., 2013). Both types of falls are highly dangerous and can result in fatal injuries such as hip fractures from slips and traumatic brain injuries from trips (Parkkari et al., 1999; Smeesters et al., 2001). The subsequent cost is high after fatal or non-fatal falls (Milat et al., 2011; Towne et al., 2014), and the induced fear of falling leads to activity reduction (Tinetti et al., 1986), diminishing the quality of life of older adults. Due to such vast consequences of falls (social and economic), strengthening the defenses of older adults against falls is imperative.

Efforts toward designing and implementing fall-prevention programs have relied on multifactorial/multicomponent (Hopewell et al., 2018) and single-component interventions (e.g., exercise) (Sherrington et al., 2019). Overall, there is a reduction of 20–30% in the rate of falls by multifactorial/multicomponent interventions and exercises such as Tai Chi (Wu et al., 2010), balance exercises, and functional exercises (Clemson et al., 2012; Arantes et al., 2015). However, it was suggested that the lack of specificity of applying gains from training under a prepared voluntary environment to an unexpected postural disturbance, a scenario that causes slip and trip falls in daily life, might limit the effectiveness of the abovementioned approaches in fall reduction (Grabiner et al., 2014).

Emerging task-specific perturbation-based training paradigms that involve inducing repeated disturbances to the alignment of the center of mass (COM) relative to the base of support (BOS) are known to enhance fall-resisting skills in older adults (Bhatt et al., 2006a,b; Mansfield et al., 2010, 2015; Bhatt et al., 2012; Wang et al., 2012; Pai et al., 2014; Patel and Bhatt, 2015). Perturbations given in a block with pure repetitive slips or trips have been shown to lead to prominent adaptive changes in the performance (Bhatt et al., 2006b; Wang et al., 2012). Critical body disequilibrium in the first perturbation that quickly reduces over a course of repetitive perturbations and is associated with both improved feedforward and feedback control through adaptation (Pai and Bhatt, 2007). Feedback control makes the ongoing reactive adjustments to compensate for motion errors after a perturbation occurs (Wolpert and Ghahramani, 2000), while feedforward control occurs before or in anticipation of a perturbation (Scheidt et al., 2001). Feedforward control makes proactive adjustments to alter the postural control relying on previous experience, and it can also influence the feedback control-related reactive adjustments. Adaptive proactive and reactive stability are achieved by improved control of the relative COM state (i.e., either its position and/or its velocity relative to the BOS; Pai et al., 2003; Pai and Bhatt, 2007). Other than stability, repeated perturbation training is known to significantly improve the control of vertical limb support required to maintain an upright position and minimize hip descent upon a large-scale perturbation. Previous research indicates that such an increase in the post-perturbation reactive limb support is achieved by increased production of the net vertical lower limb joint torque (Pai et al., 2003; Pai and Bhatt, 2007), which, in turn, is influenced by the rate and magnitude of muscle force production. Although adequate studies have reported significant improvements in the reactive balance control and fall reduction following a block of repetitive perturbations generated in the same manner, such a predictive gait alteration induced by predictable block perturbations might obscure the reactive improvements. For example, if participants adopted a high toe clearance before anticipating a trip, it is very likely that they would avoid contacting the tripping obstacle and would make it highly challenging to examine the response of the feedback control (Wang et al., 2019).

It is known that fall mechanisms and corresponding preventive adaptive responses for recovery from slips vs. trips are opposite in nature. For example, a slip or a slip-like perturbation moving the feet/BOS anterior to the COM induces a backward balance loss and associated falls (Bhatt et al., 2006b) and a trip-like perturbation moving the feet/BOS posterior to the COM induces a backward balance loss and associated falls (Wang et al., 2012). Recovery from both thus involves specific directional responses for the control of COM stability. For example, while controlling trunk momentum is crucial for preventing forward falls upon novel trips (Pavol et al., 2001; Wang et al., 2019), a backward compensatory stepping contributes more to slip-induced recoveries (Pai and Bhatt, 2007). Therefore, it is questionable whether such adaptive changes acquired from a highly predictable fixed condition can be transferred to more unexpected conditions with perturbations occurring at random.

To address the above issues, mixed exposure of opposing perturbations (slip and trip) can minimize the anticipation and evaluate the reactive balance response during gait perturbation. A vital form of functional plasticity of the central nervous system (CNS) is its ability to take motor adaptations obtained from one situation and apply them appropriately to different "contexts." Previous findings have shown the ability of CNS to generalize the adaptive gains in stability and limb support across different environmental contexts (treadmill-slips to over-ground-slips; Lam and Dietz, 2004; Morton and Bastian, 2004a,b; Seidler et al., 2004) or across different tasks (gait-slip to a sit-to-stand slip; Pai et al., 2003; Bhatt and Pai, 2009; Yang et al., 2009, 2013). However, when the contextual difference is large (slip vs. trip), sensorimotor adaptation to a perturbation that requires opposing motor adjustments could, in fact, interfere (negative transfer) with each other, at least in the proactive control of stability. For example, the CNS learns to anteriorly shift the COM position and/or to increase its velocity with feedforward and feedback mechanisms after repeated slip exposure (Pai et al., 2010). Yet, when facing a trip, the CNS must learn to posteriorly shift the COM position and/or to reduce its velocity (Wang et al., 2012). Contextual interference of exposure to slip and trip was proved in young adults. Bhatt et al. (2013) found that proactive adjustments, shown as the anterior shifting of the COM position relative to BOS adapted from prior slip-perturbation training, persisted at the pre-trip instance in a novel trip following the prior slip training. Such proactive adjustments immediately resulted in a greater anterior instability compared with a control group not receiving prior slip training.

It is postulated that the training-induced vulnerability to the opposite perturbation, if existing, could be quickly amended based on the capability of CNS to trigger an adapted reactive control that rapidly enhances post-perturbation stability (improved trunk control and protective stepping) and limb support (improved net vertical joint torque), thus, minimizing the need for an entirely new motor program or immediate improvements in the physical conditions (strength, balance, etc.) of an individual (Morton et al., 2001). The CNS gradually recalibrates and optimizes the stability and limb support gains and its representation of fall risk limits against both forward and backward balance losses. Such a postulation was partially validated in a study conducted in young adults (Bhatt et al., 2013), where such interference seen was, however, mitigated at the post-trip instance of recovery touchdown—a possible generalization of the reactive response resulting in no difference in the vertical limb support and stability values between the training and control groups. Similarly, Okubo et al. (2018) reported that young adults had an improved margin of stability (MOS) when recovering from a trip after exposure to random slip and trip perturbations. There is limited evidence to determine to what extent the interference of the opposing perturbation could affect the proactive and reactive stability control in older adults.

The aim of this study was thus to determine the effects of perturbation-specific training (slip-only or trip-only) in inducing interference or generalization within proactive (feedforward) and reactive (feedback) mechanisms for the control of stability and limb support, the two likely essential defense elements against falls in older adults. Our prior preliminary results from young adults showed that post-perturbation training, adaptation within proactive control (feedforward), which is involved with the upcoming context prediction, will be prone to a greater interference when exposed to an opposing perturbation (Bhatt et al., 2013). Because we expected that the impact of training-induced improvement in the reactive control of stability and limb support will be higher than that in the proactive control, a proper and effectively trained reactive response can be commonly applicable against falls even under diametrically different precursors. Specifically, we hypothesized that, though perturbation-specific training will induce a negative interference in the proactive control of stability when exposed to the opposing perturbation, it could induce a significant amount of (positive) generalization in the reactive control of stability and limb support, thus, leading to greater gains in these variables and lowering the laboratory-induced falls when exposed to the opposing perturbation compared to that of their controls (**Figure 1**). Findings from this study can contribute to optimizing the design of an effective perturbation training in older adults.

MATERIALS AND METHODS

Participants

Three hundred and five older adults (>60 years) were initially screened to pass a descriptive questionnaire without the self-reported recent (<6 months) neurological, musculoskeletal, or systematic disorders. Two hundred and forty-one qualified older adults were then screened onsite to pass a cognitive test [>25 on the Folstein Mini-Mental Status Exam (MMSE)] (Mf et al., 1975), a calcaneal ultrasound screening (T score > −2.0) (Thompson et al., 1998), a mobility test [Timed-Up-Go (TUG) score < 13.5 s] (Podsiadlo and Richardson, 1991), and a monofilament foot sensation test (able to detect the Weinstein 5.07 monofilament at all nine locations of both feet; Kumar et al., 1991). One hundred and sixty-five qualified community-dwelling healthy older adults (69.6 ± 6.4 years) were finally included in the study. Participants also received other commonly used clinical measurements and questionnaires, including the Berg Balance Scale (BBS), Activities-specific Balance Confidence (ABC) Scale, a fall history questionnaire, and a 6-min walking test. All participants provided written informed consent, and this study was approved by the Institutional Review Board in the University of Illinois at Chicago.

Study Design

Qualified participants were randomly assigned following simple randomization procedures to one of the three groups: slip-perturbation training group (St, $n = 67$), trip-perturbation training group (Tt, $n = 67$), and control group (Ctrl: $n = 31$) with a 2:2:1 allocation. This study is the first part of a larger study (ClinicalTrials.gov NCT03199729; https://clinicaltrials.gov/ct2/show/NCT03199729) specifically examining generalization and/or interference effects in older adults when exposed to a directionally opposing perturbation after a slip-only or trip-only training. We had conducted an *a priori* power analysis based on the preliminary data, and because we expected the total rate of slip and trip falls (∼30%) in the slip-only and trip-only group to be half of that in the control group (∼60%), we needed a larger sample size for slip- and trip-only groups to detect a large effect size between these two groups. The current sample size provided a >80% statistical power to detect a large effect size (=0.5) between the training groups and the control group (slip-only vs. control and trip-only vs. control) and between the two training groups. The randomization option was adopted to maintain sufficient power yet reduce the recruitment burden. A randomization sequence was created using Excel. Group St received 24 repetitive slip perturbations, Group Tt received 24 repetitive trip perturbations, and Group Ctrl received no training but only walking trials. Post-training walking trials were studied to show proactive (feedforward) control. Post-training perturbation trials were studied to indicate reactive (feedback) control (**Figure 1**).

Experimental Setup

Slip perturbations were induced by the sudden release of a pair of low-friction, movable platforms on sliding tracks mounted to supporting frames. The two platforms were embedded in the middle of the left and right sides of the 7-m walkway. During slip trials, the movable platform was released when the vertical ground reaction force (GRF) under the perturbed (right) limb exceeded 10% of body weight after the touchdown of the

Aim: Examining interference and generalization effects to novel opposing slips/trips post-perturbation training			
Group R = *(2:2:1)*	**Training**	**Post-training Test**	**Hypothesis (Planned comparisons)**
Group St Slip-only (n=67)	24 Slips	Walking trials Trial1 = Retest slip Trial2 = Novel trip	**Walking trials** **Proactive Stability:** *St = Tt > Ctrl* **Step length**: *Tt = Ctrl > St* **Toe clearance:** *Tt > Ctrl = St*
R → Group Tt Trip-only (n=67)	24 Trips	Walking trials Trial1 = Retest trip Trial2 = Novel slip	**Reactive Stability & Limb Support** ***Slip trial:*** *St* > Tt ≥ Ctrl ***Trip trial:*** Tt > St ≥ Ctrl
Group Ctrl Control (n=31)	NONE	Walking trials r [Novel trip Novel slip	**Fall** ***Slip trial falls:*** *St* < Tt ≤ Ctrl ***Trip trial falls:*** *Tt* < St ≤ Ctrl

R=Randomization

FIGURE 1 | The research design for the hypothesis and the planned comparisons were performed. Post-training walking trials (PW) were compared to examine interferences of training adaptations on the responses to opposing perturbations in the proactive control. Post-training perturbation trials were studied to investigate interferences of training adaptations on the responses to opposing perturbations in the reactive control. R = randomized assignment of subjects among groups.

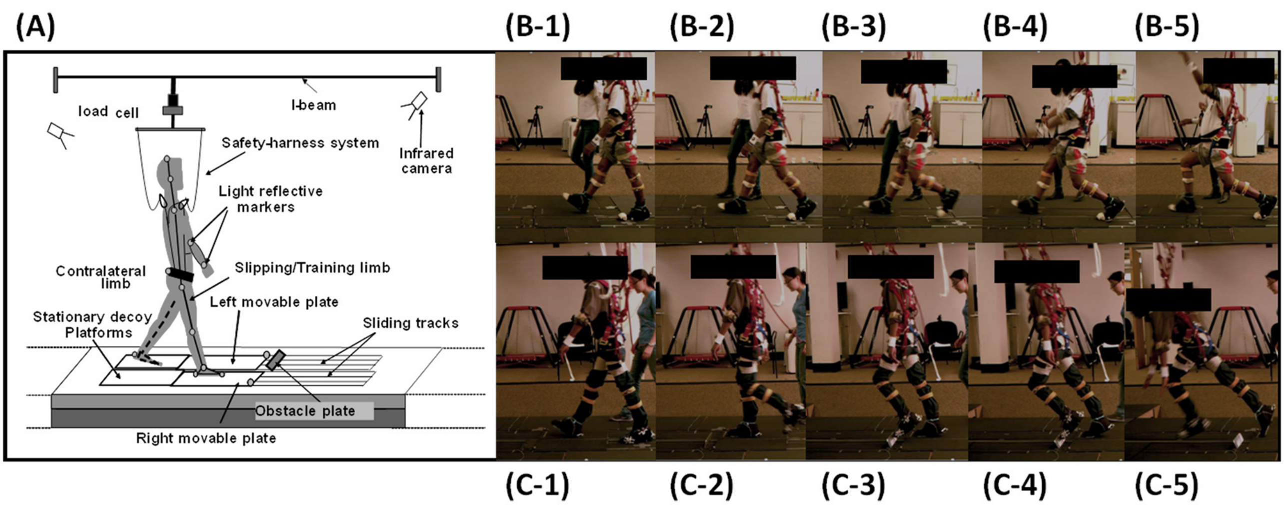

FIGURE 2 | (A). The experimental setup of the over-ground walkway, the overhead harness, and the motion system. **(B-1)** to **(B-5)** Still images indicate the instance of right foot touchdown (RTD) before the slip onset to recovery left foot touchdown (LTD). **(C-1)** to **(C-5)**. Still images indicate the instance of RTD before left foot hitting the obstacle to recovery foot touchdown.

right foot. The left platform was automatically released after the recovery (left) foot landed on it.

This would guarantee that all slips occurred at the beginning of the double-stance phase (**Figures 2A,B**). Trip perturbations were induced by an obstacle device (height: 8 cm; width: 27 cm; thickness: 0.5 cm), which was embedded on the left side of the walkway (**Figures 2A,C**). During trip trials, the trip plate was unlocked after 50 ms of the instant when the vertical GRF under the unperturbed (right) limb exceeded 90% of body weight after its touchdown. Once the trip plate was triggered, it stayed unlocked. This would guarantee that all trips occurred in the late-swing phase. The GRF was detected by the force plates (AMTI, Newton, MA) installed beneath the right platform. During regular walking, both the movable platform and the trip plate were locked by a pair of electromagnets. Participants were protected by a full-body safety harness connected by shock-absorbing ropes to a load cell (Transcell Technology Inc., Buffalo Grove, IL). The load cell was mounted to an overhead trolley on a track over the walkway. The harness enabled participants to walk freely while providing them protection against body impact with the floor. Kinematics from a full-body marker set (30 retro-reflective markers) were recorded

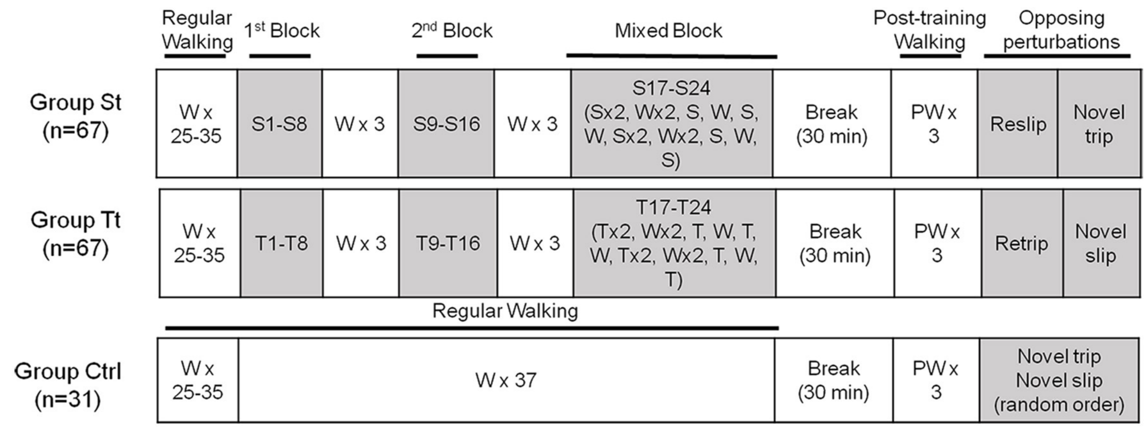

FIGURE 3 | The training protocol used for the study. The slip-training (St) group received 24 repetitive slip perturbations, the trip-training (Tt) group received 24 repetitive trip perturbations, and the control (Ctrl) group received no training but only walking trials (W). Specifically, after 25–35 unperturbed normal walking trials (W) received by all groups, Group St received a block of eight repeated slip trials (S1–S8), followed by three unperturbed trials, another block of eight slip trials (S9–S16), an additional three unperturbed trials, and a final block of 15 mixed trials (including eight slip and seven unperturbed trials) (S17–S24). Group Tt experienced trials in the same design as Group St but trips as perturbation (T). Group Ctrl experienced an additional 37 unperturbed walking trials. After a 30-min break, all groups received three unperturbed post-perturbation walking trials (PW). Then, Group St received a reslip followed by a novel trip, Group Tt received a retrip followed by a novel slip, and Group Ctrl experienced these two perturbations in random order.

by an eight-camera motion capture system (Motion Analysis Corporation, Santa Rosa, CA). Kinetic data were sampled at 120 Hz and synchronized with the force plate and load-cell data, which was collected at 600 Hz.

Study Protocol

All participants experienced 25–35 unperturbed walking trials on a 7-m walkway to become familiar with the laboratory walking environment. Their starting position was adjusted during walking trials to ensure that the upcoming perturbations were consistently induced in the same gait phase for all participants. Specifically, after normal walking trials in the training session, Group St received a block of eight repeated slip trials, followed by three unperturbed trials, another block of eight slip trials, an additional three unperturbed trials, and a final block of fifteen mixed trials (including eight slip and seven unperturbed trials) (**Figure 3**). Group Tt experienced trials in the same design of Group St but trips as perturbation. Group Ctrl experienced an additional 37 unperturbed walking trials following the familiarization walking session to match the total trials received by the other two groups. After a 30-min break, all groups received three unperturbed post-walking trials. Group St received a reslip followed by a novel trip, Group Tt received a retrip followed by a novel slip, while Group ctrl experienced these two perturbations in a random order. For all three groups, participants were informed that "a slip or trip may or may not occur during your walking" at the beginning of each trial and that, if the perturbation occurred, they should "try to recover and continue walking."

Outcome Variables

Perturbation outcome from a slip or a trip was defined as a fall (**Figures 2B-5,C-5**) if the load cell detected more than 30% of body weight of the participant after perturbation onset and was further verified using motion videos (Yang and Pai, 2011). If the perturbation outcome did not meet this criterion, it was defined as a recovery. Because both slip and trip were triggered by detecting the right foot touchdown (RTD), the instances of RTD right before a slip (**Figures 2B-1**) or a trip (**Figures 2C-1**) onset were chosen to reflect a proactive performance anticipating a perturbation. Following a slip or a trip, the training foot (left foot) quickly touched down for a recovery step; therefore, the instance of the left foot touchdown (LTD) was selected to reflect the reactive response to a perturbation (**Figures 2B-4** for the slip and **Figures 2C-5** for the trip). All instances were identified from the synchronized vertical GRF and motion analysis data.

Margin of stability was selected to qualify the balance status of an individual, which was calculated as follows (Hof et al., 2005):

$$\mathrm{MOS} = (\mathrm{xCOM} + \frac{\mathrm{vCOM}}{\sqrt{\frac{g}{l}}} - BOS_{pos})/BOS_{len}$$

Here, the xCOM indicates the COM position in the anterior–posterior (AP) direction, and vCOM indicates the COM velocity in the AP direction. Body COM kinematics were calculated using a 13-segment rigid-body model with gender-dependent segmental inertial parameters. g is the gravitational acceleration and l represents the leg length calculated using the markers attached on the greater trochanter of femur. BOS represents the area beneath an individual encircled by the points of contact that the foot or feet of an individual make(s) with the supporting surface, and BOS_{pos} is the posterior edge of BOS, which was calculated using the position of a heel marker. In this study, we normalized the MOS by the length of BOS (BOS_{len}), which is the length of BOS in the anteroposterior direction. In this case, the MOS, whose value is >1, indicates that the extrapolated COM exceeds the anterior boundary of BOS, while a negative MOS indicates that the extrapolated COM exceeds the posterior boundary of BOS. A larger MOS indicates better stability against slip perturbation but a greater forward instability against trip perturbation; conversely, a smaller MOS indicates better stability against trip perturbation but a greater backward instability against slip perturbation.

Previous studies indicated that the step length was related to the stability and could affect the risk of slip-induced falls (Espy et al., 2010; Wang et al., 2020). The step length was calculated by subtracting the heel position of the stepping foot from the heel position of the stance foot in the AP direction at RTD. The toe clearance was shown to be highly related to the risk of tripping in older adults (Hamacher et al., 2014). The toe clearance was measured as the maximum vertical distance from the ground to the toe marker in the gait cycle before LTD (from the liftoff of the left foot to its touchdown). Both the step length and the toe clearance were calculated for the post-training walking trial in three groups to indicate the proactive adjustments. The hip height was calculated as the midpoint of the two hip markers. In one gait cycle after perturbation onset (from RTD to LTD), the minimum value of the hip height, calculated when the midpoint of the two hip markers reached the lowest position in the end, was examined during post-training slip and trip trials to indicate the reactive responses. All of these variables were normalized by body height.

Because proactive control quickly improves in the first block through adaptation and remains stable in the subsequent perturbation trials (Bhatt et al., 2006b; Wang et al., 2012), MOS at RTD for S1, S8, and S24 in Group St and for T1, T8, and T24 in Group Tt were analyzed to detect the slip and trip adaptation. MOS at RTD for the reslip and retrip trials were also compared to detect the retention of adaptation. MOS at RTD, step length, and toe clearance in the first post-training walking trials for three groups were compared to examine the proactive adjustment and interference. The first post-training walking trials were chosen to represent the adaptive proactive adjustments because this trial reflects the immediate gait changes used to anticipate a perturbation. MOS at LTD and minimum hip height of reslip and novel trip trials in St, of retrip and novel trip trials in Tt, and of novel slip and trip trials in Ctrl were calculated to examine the reactive interference and generalization.

Statistical Analysis

One-way ANOVAs were performed to examine any differences in the baseline demographics (age, height, weight, BBS, MMSE, TUG, and ABC) of the participants among the three groups. One-way repeated measures ANOVAs were first performed to examine the adaptive changes and the retention of these changes in MOS (S1, S8, S24, and reslip for St and T1, T8, T24, and retrip for Tt) at RTD and LTD, respectively. Follow-up comparisons were resolved using the paired t-tests between two trials. The Benjamini–Yekutieli procedure is a multiple testing method that controls the false discovery rate under the arbitrary dependence of the *p*-values (Benjamini and Yekutieli, 2001). This procedure was applied to reduce the type I error for multiple comparisons across different groups (corrected $\alpha = 0.02$). A chi-squared test was performed to compare the fall outcomes of a reslip in St, a novel slip in Tt, and a novel slip in Ctrl. A chi-squared test was also conducted to compare the fall outcomes of a retrip in Tt, a novel trip in St, and a novel trip in Ctrl. Furthermore, a chi-squared test was performed between two groups out of the three groups as the *post-hoc* analysis. A fall was coded as 1 and recovery was coded as 0 in the analysis.

TABLE 1 | Baseline demographics and clinical measurements of the participants in the slip-training (St) group, the trip-training (Tt) group, and the control (Ctrl) group.

	St ($N = 67$)	Tt ($N = 67$)	Ctrl ($N = 31$)	*p*-value
Age (yrs)	69.6 ± 6.8	69.9 ± 6.2	68.8 ± 6.4	0.52
Weight (kg)	79.1 ± 18.2	75.7 ± 15.2	78.2 ± 17.3	0.72
Height (m)	1.69 ± 0.1	1.68 ± 0.1	1.65 ± 0.1	0.1
TUG (s)	8.5 ± 1.8	8.2 ± 1.6	8.5 ± 1.5	0.57
BBS	53.5 ± 2.4	53.2 ± 2.9	52.2 ± 2.97	0.11
MMSE	28.6 ± 1.7	28.2 ± 2.3	28 ± 1.8	0.27
ABC	84.4 ± 13.3	85.6 ± 12.3	83.1 ± 15.3	0.7

The mean and SD of individual variable in each group are provided. P-values of one-way ANOVAs for individual variables among four groups are provided.
TUG, Timed-Up-and-Go test; BBS, Berg Balance Scale; MMSE, Mini-Mental Status Exam; ABC, Activities-specific Balance Confidence Scale.

A one-way ANOVA was conducted to analyze the training effect (level=3 for Group St, Tt, and Ctrl) on the MOS, step length, and toe clearance at RTD in the post-training walking trials to indicate the proactive adjustments. Independent *t*-tests were used as a *post hoc* test for a two-group comparison (corrected $\alpha = 0.02$). Two-way ANOVA was conducted to analyze the training effect (level=3 for Group St, Tt, and Ctrl), the perturbation effect (level=2 for slip and trip), and the interaction on MOS at LTD, as well as to analyze the minimum hip height in the reslip and novel trip trials of the St group, the retrip and novel slip trials of Tt group, and the novel slip and trip trials of the Ctrl group. Independent *t*-tests were used as a *post-hoc* test for a two-group comparison (corrected $\alpha = 0.02$). Linear regressions were used to examine the relationship between the proactive MOS and the reactive MOS in slip and trip trials. Proactive MOS was input as the independent variable to predict the reactive MOS, which was input as the dependent variable for the slip trials (including the reslip trial in Group St and the novel slip trials in Groups Tt and Ctrl) and the trip trials (including the novel trip trials in Groups St and Ctrl and the retrip trial in Group Tt). All statistical analyses were performed using SPSS 22 (IBM Corp, Armonk, NY).

RESULTS

Adaptation and Retention

There was no significant difference in the baseline demographics of the participants (**Table 1**).

There were significant differences in the proactive MOS over time in Group St ($F = 4.85$, $p = 0.003$; **Figure 4A**). MOS rapidly improved in the first eight trials (S8 > S1, $p = 0.007$), and by the end of the slip training, MOS was significantly greater in S24 than in S1 ($p = 0.005$). Training effects remained for 30 min

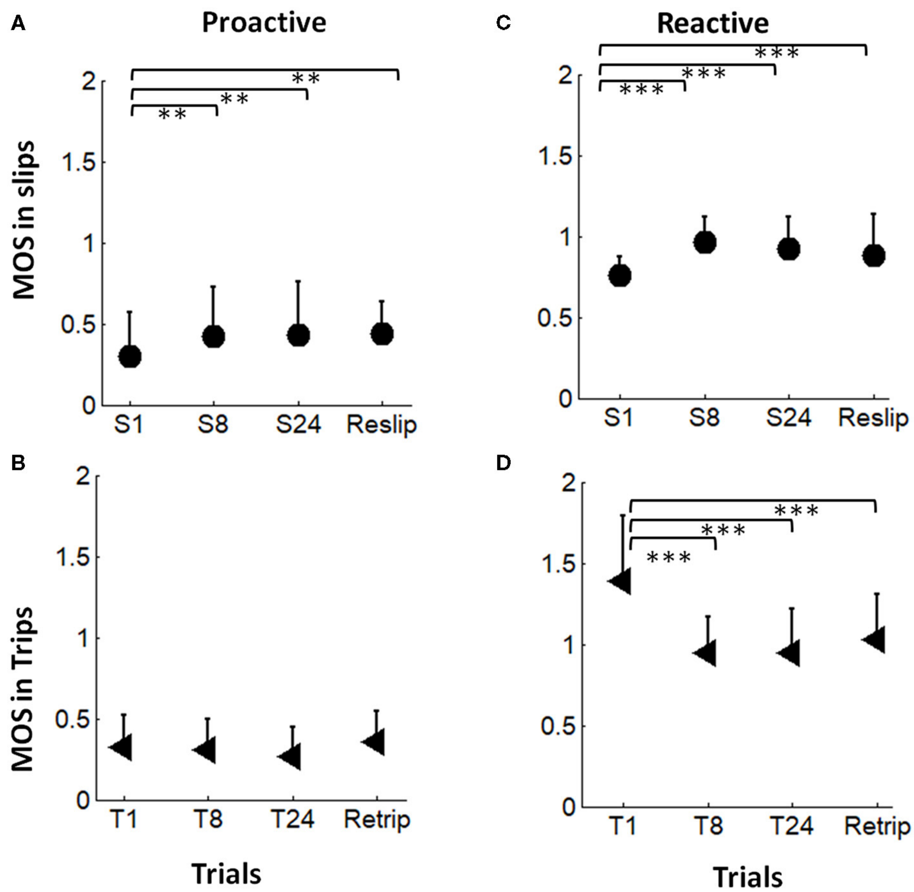

FIGURE 4 | (A,B) The proactive adaptation and the short-term (30-min) retention in the margin of stability (MOS) during slip perturbation (indicated by filled circles) and trip perturbation (indicated by filled triangle) trainings. **(C,D)** The reactive adaptation and the short-term (30-min) retention in MOS during slip perturbation (indicated by filled circles) and trip perturbation (indicated by filled triangle) trainings. S1, S8, and S24 indicated the 1st, 8th, and the 24th slips, respectively, during the slip-training session. Reslip indicated the retest slip after a 30-min break. T1, T8, and T24 indicated the 1st, 8th, and 24th trips, respectively, during the trip-training session. Retrip indicated the retest trip after a 30 min break. The mean value of MOS and the positive value of standard deviation for each trial are displayed. $^{**}p < 0.01$; $^{***}p < 0.001$.

such that reslip had comparable proactive MOS to S24 ($p > 0.05$). There was a trend of reduced proactive MOS from T1 to T24 during trip training ($p = 0.07$) (**Figure 4B**). There was no significant difference in the proactive MOS between retrip and T24 ($p > 0.05$).

Adaptation of the reactive MOS (at LTD) was observed in both slip training and trip training groups (**Figures 4C,D**). There were significant differences in the reactive MOS over time for both slip ($F = 18.5$, $p < 0.001$) and trip training ($F = 28.2$, $p < 0.001$) groups. The reactive MOS in S8 and S24 was significantly improved in comparison with that in S1 ($p < 0.001$ for both). Training effects remained for 30 min during which reslip had a comparable reactive MOS to S24 ($p > 0.05$). The reactive MOS in T8 and T24 was significantly lower than that in T1 ($p < 0.001$ for both). Training effects remained for 30 min during which retrip had a comparable reactive MOS to T24 ($p < 0.001$).

Fall Outcomes

Results of the chi-squared test indicated that fall incidences were significantly different among reslip in the St group, novel slip in the Tt group, and novel slip in the Ctrl group [χ^2 (2) $= 63.0$, $p < 0.001$], and the results were significantly different across retrip in the Tt group, novel trip in the St group, and novel trip in the Ctrl group [χ^2 (2) $= 30.1$, $p < 0.001$] (**Figure 5**). For slip-induced falls, the participants in Group St had fewer falls (0%) in reslip than in novel slip in Group Tt (64%) and in novel slip in Group Ctrl (58%) ($p < 0.001$ for both; **Figure 5**, indicated by filled columns), while no difference was found between Groups Tt and Ctrl ($p = 0.57$). For trip-induced falls, the participants in Group Tt had fewer falls (3%) in the retrip trial than in the novel trip trials in Group St (42%) and in Group Ctrl (39%) ($p < 0.001$ for both; **Figure 5**, indicated by unfilled columns), while no difference was found between Groups St and Ctrl ($p = 0.78$).

Interferences in Proactive Adjustments

There was a main effect of training on the proactive MOS among post-training walking trials in three groups ($F = 8.37$, $p < 0.001$; **Figure 6A**). The *post-hoc t*-test showed that the participants in Group St had a significantly larger MOS compared with those in Group Tt ($p < 0.001$) and Group Ctrl ($p = 0.003$), while the MOS was comparable between Groups Tt and Ctrl ($p = 0.45$). There was also a main effect of training on the step length ($F = 11.2$, $p < 0.001$) (**Figure 6B**) and toe clearance ($F = 15.6$, $p < 0.001$) (**Figure 6C**). The participants in Group St took a significantly shorter step than those in the other two groups ($p < 0.01$ for both), and no difference in the step length was found between Groups Tt and Ctrl ($p > 0.05$). However, Group Tt had a higher toe clearance compared with other groups ($p \leq 0.001$ for both), and no difference in the toe clearance was found between Groups St and Ctrl ($p > 0.05$).

Interferences and Generalization in Reactive Adjustments

There was a main effect of the training ($F = 40.3$, $p < 0.001$) and perturbation types ($F = 230.1$, $p < 0.001$), as well as a significant interaction between the training and perturbation types ($F = 4.8$, $p = 0.009$) on the reactive MOS (**Figure 7A**). Overall, there was a larger reactive MOS in Group St and a larger reactive MOS for the slip perturbations. The *post-hoc t*-test revealed that reslip of Group St had a significantly larger reactive MOS than that in novel slip of Groups Tt and of Ctrl ($p < 0.001$ for both) (**Figure 7A**, indicated by filled circles), and there were no significant differences in the reactive MOS between novel slips of Groups Tt and Ctrl ($p > 0.05$). The *post-hoc t*-test also indicated that retrip of Group Tt had a significantly smaller reactive MOS than that in novel trip of Group St and of Group Ctrl ($p < 0.001$ for both) (**Figure 7A**, indicated by triangles), and there were no significant differences in the reactive MOS between the novel trips of Groups Tt and Ctrl ($p > 0.05$). Similarly, there was also a main effect of the training ($F = 5.08$, $p = 0.007$) and perturbation types ($F = 5.32$, $p = 0.02$), as well as a significant interaction between the training and perturbation types on the reactive limb support (hip height) ($F = 13.74$, $p < 0.001$) (**Figure 7B**). Overall, the hip height was larger in Group St and in trip perturbations. The *post-hoc t*-test indicated that the reslip of Group St had a significantly higher hip height compared with that in novel slips of Groups Tt and Ctrl ($p < 0.001$ for both) (**Figure 7B**, indicated by filled circles), and there were no significant differences between novel slips of Group Tt and that of Group Ctrl ($p > 0.05$). However, there was no difference in the hip height among retrip in Group Tt, novel trip in Group St, and novel trip in Ctrl ($p > 0.05$ for all). Linear regressions indicated that, for both slip and trip trials, the proactive MOS was a significant predictor (both $p < 0.001$) of the reactive MOS. Specifically, 11.9% ($r^2 = 0.119$) and 13.9% ($r^2 = 0.139$) of variances in the reactive MOS were accounted for by the proactive MOS for slip and trip trials, respectively.

DISCUSSION

Our central hypothesis was that the CNS can still recalibrate its motor strategies based on the commonalities in the reactive control of stability to generalize (positively transfer) the previously learned strategies and to mitigate or overcome any negative interference in the proactive control of stability induced by the opposing perturbation. Specifically, this study supported that post perturbation training, adaptation within proactive control (feedforward), which is involved with the upcoming context prediction, will be prone to a greater interference when exposed to an opposing perturbation. In addition, the current study partially supported that, even with the given negative interference in the proactive control induced by the opposing perturbation training, the training-induced improvement in the reactive control of stability and limb support will be more generalizable than that in the proactive control, as shown by the results that subjects had equal but not inferior reactive stability and post-perturbation limb support gains in comparison to those in the control group who did not receive any opposing perturbations.

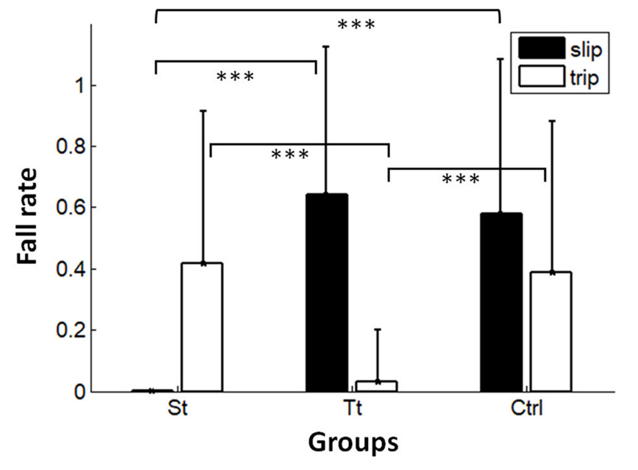

FIGURE 5 | Fall outcomes among the reslip trial in the St group, the novel slip trial in the Tt group, and the novel slip trial in the Ctrl group. Significant differences were shown as the top two lines. Fall outcomes among the retrip trial inTt and the novel trip trial in St and in Ctrl. Significant differences were shown in the middle two lines. ***$p < 0.001$.

This study adopted a design with the first training session of repetitive perturbations of the same type (i.e., all slips or all trips) and a latter part of the mixed exposure of opposing perturbations—one reslip followed by a novel trip for the slip-training group, one retrip followed by a novel slip for the trip-training group, and a randomized novel slip and a novel trip for the control group. Consistent with the previous findings, at the completion of the first training session, subjects demonstrated a trial-to-trial improvement in the reactive control of stability (**Figures 4C,D**) shown by an increased MOS (more stable against a backward loss of balance) from the 1st slip to the 8th and 24th slips, as well as a decreased MOS (more stable against a forward instability) from the 1st trip to the 8th and 24th trips. Furthermore, the non-significant difference between the retention slip (before the opposing trip) and the 24th slip and the non-significant difference between the retention trip (before the opposing slip) and the 24th trip indicated that such adaptive improvements were able to retain for at least 30 min. Hence, it is reasonable to postulate

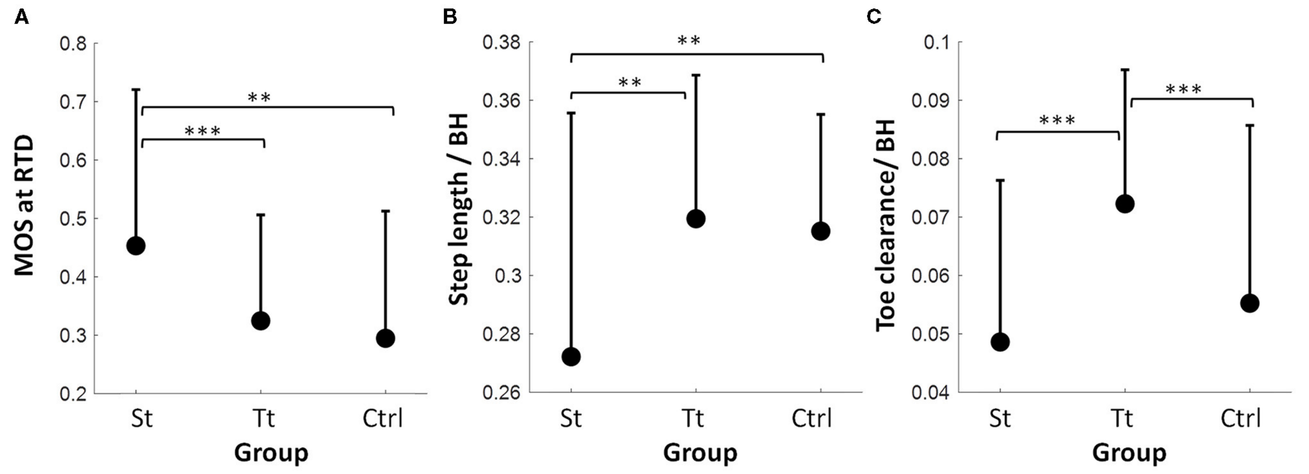

FIGURE 6 | (A) Proactive adjustments among PW in the St group, the Tt group, and the Ctrl group in a) the MOS at the RTD, **(B)** the step length normalized by body height (BH), and **(C)** the toe clearance normalized by BH. The mean value of MOS, step length, and toe clearance and the positive value of SD for each variable in each trial are displayed. $^{**}p < 0.01$; $^{***}p < 0.001$.

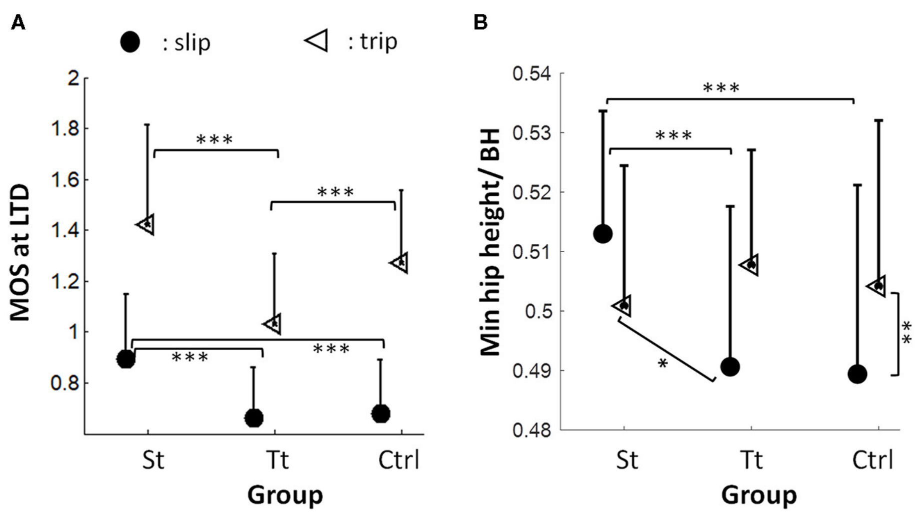

FIGURE 7 | (A) Reactive MOS at the recovery LTD among the reslip trial in the St group and the novel slip trials in the Tt group and in the Ctrl group (indicated by filled circles). Significant differences are shown as the bottom two lines. The reactive MOS at the recovery LTD among the retrip trial in the Tt group and the novel trip trials in St and in Ctrl groups (indicated by triangles). Significant differences are shown in the top two lines. **(B)** Limb support (represented by the minimum hip height normalized by BH) among the reslip trial in the St group () and novel slip trials in the Tt group and in the Ctrl group (indicated by filled circles). Significant differences are shown in the top two lines. Limb support among the retrip trial in the Tt group and the novel trip trial in St and in Ctrl groups (indicated by triangles). The vertical line indicates the significant differences between the novel slip and the novel trip of the Ctrl group. The oblique line indicates the significant differences between the novel trip in the St group and the novel slip in the Tt group. The mean value of MOS and limb support and the positive value of the SD for each variable in each trial are displayed. $^{*}p < 0.05$; $^{**}p < 0.01$; $^{***}p < 0.001$.

that, based on the recent perturbation history, the proactive and reactive control of stability would improve or at least remain unchanged given the upcoming perturbation in the same context (Bhatt et al., 2006a; Wang et al., 2019). To meet the demand of sufficient reactive stability against a fall, the CNS has to proactively regulate gait while anticipating an upcoming perturbation in the same context. As shown in our results (**Figure 6**), proactive stability against a predicted slip was achieved by shortening the step length in the regular gait (step length: St < Tt = Ctrl), and proactive adjustment against a predicted trip was achieved by the increased toe clearance before hitting an obstacle in the regular gait (toe clearance: Tt > St = Ctrl).

From the mechanistic perspective, it is postulated that such an adaptation occurs *via* updating of the internal representation of stability limits based on the immediate and past experiences

(Lam and Dietz, 2004; Morton and Bastian, 2004a,b; Seidler et al., 2004). Such an update results in the modification of motor responses (predominantly proactive changes *via* feedforward mechanisms) when the CNS is simultaneously expecting a similar perturbation. When the expected and experienced perturbations match up, it results in an enhanced performance (adapted/learnt response). However, when the CNS experiences a different (in our case, an opposing) perturbation, the proactive adaptive changes could lead to an interference. For example, positive slip adaptations are shown as the shortened step length and a forward-shifted COM position (Bhatt et al., 2006b). Although a smaller step length in gait could bring COM closer to the BOS at the trailing limb lift-off, which, in turn, initiates a better (forward) stability at the slip onset, such a strategy would increase the forward instability for the upcoming trip and could worsen the reactive recovery response (Bhatt et al., 2013). Improved trip adaptations are indicated as a higher toe clearance and a posterior shift of COM (Wang et al., 2019). While sufficient toe clearance can reduce the impact of the trip or completely prevent contact with an obstacle, and a less anterior COM position can establish a stable initial status against forward instability (Wang et al., 2019), such a strategy might increase the predisposition to a backward balance loss and reduce the overall effectiveness of the reactive response upon a slip.

Based on an expected interference resulting from a proactive adaptation, it could be postulated that both St and Tt groups would have more falls, worse reactive stability, and lower vertical limb support than the Ctrl group when they experienced an opposing novel perturbation (novel trip for Group St and novel slip for Group Tt). However, if the generalization of the adaptive improvements through training was demonstrated when experiencing an opposing perturbation, the performance on the novel opposing perturbation would be better than, or at least equal to, that of the control group receiving no prior training.

Our results partially supported such an interference based on the prior expectation for both the St and Tt groups. On the post-training walking trial for the St group, just prior to the 30-min reslip test, we saw that the participants maintained their proactive changes in the slip-training group with a higher pre-slip MOS and a shorter step length than those in the control group and the trip group (who did not get any slip training) (**Figures 6A,B**). Such proactive changes could have interfered with the reactive response to trips as indicated by a slightly greater forward post-slip/reactive MOS on that trial than that on the novel trip of the Ctrl group (although not significant) (**Figure 7A**). However, such an interference was probably mitigated by the reactive response demonstrated in the vertical support limb at touchdown of the compensatory step as the proactive MOS only accounted for ∼10% variance in the reactive MOS. There was no difference in limb support (**Figure 7B**) and fall outcomes (**Figure 5**) on the novel trip trial between St group and the Ctrl group. On the post-training walking trial for the Tt group, immediately before the 30-min retrip test, the toe clearance was higher than that of the Ctrl group and the St group (**Figure 6C**). However, for trip training, the proactive changes in MOS may not have been retained as robustly as the slip group after 30 min. Thus, there was possibly a lesser proactive interference seen in the trip group, as indicated by a similar proactive MOS between Tt and Ctrl groups (**Figure 6A**). Subsequently, there was no significant difference in the reactive MOS and limb support on the novel slip between the Tt group and the Ctrl group (**Figures 7A,B**). However, it must be noted that slips might be more challenging perturbations to recover from than trips, which may help to explain that the limb support on the novel slips for Tt and Ctrl groups (**Figure 7**, filled circles for Tt and Ctrl groups) was lower than that on the novel trips experienced in St and Ctrl groups (**Figure 7**, triangles for St and Ctrl groups).

Despite interferences in the proactive control as shown in the Results section, the findings of the non-significant differences in falls, MOS, and hip height in the novel opposing perturbation of the training groups (either St and Tt) in comparison with those in the novel perturbation of the Ctrl group supported our second hypothesis that the reactive control of stability and limb support will be more generalizable than the proactive control, which was consistent with the previous findings. Bhatt et al. (2013) reported that young adults exhibited a lack of difference in the reactive stability after being exposed to opposing perturbations instead of a worsening outcome than their controls without prior interference. This could be explained by a more flexible responding strategy in the feedback control than in the feedforward control. For feedforward adjustment, the CNS relies on prior experience, such as repeated perturbations to recalibrate its internal representation of the fall threshold, and further alters postural response synergies to meet the demand of that specific type of perturbation (Vetter and Wolpert, 2000; Scheidt et al., 2001; Wang et al., 2001; Witney et al., 2001; Davidson and Wolpert, 2003). Hence, the postural responses to an anticipated perturbation are consistent such as a reduced step length when anticipating to a slip or an elevated toe clearance when anticipating to a trip (Bhatt et al., 2006b; Wang et al., 2019). However, the feedback control of gait recovery has more flexibility based on the ongoing COM status. Multiple joint segments together contribute to the global COM state changes and limb support after perturbation onset, and such a multilink mechanism allowed versatile recovery strategies to be applied during gait perturbation

TABLE 2 | Demographics and clinical measurements of the participants grouped by leg dominance.

Dominant leg	Right ($N = 159$)	Left ($N = 6$)	*p*-value
Age (yrs)	69.4 ± 6.4	73.7 ± 7.7	0.24
Weight (kg)	77 ± 16	76.2 ± 14.4	0.9
Height (m)	1.67 ± 0.1	1.68 ± 0.1	0.82
TUG (s)	8.4 ± 1.7	8.2 ± 0.9	0.60
BBS	53.1 ± 2.8	53.5 ± 2.7	0.76
MMSE	28.1 ± 2.9	28.5 ± 2	0.67
ABC	84.5 ± 14.7	74 ± 13	0.1

The mean and SD of individual variables in each group are provided. P-values of the independent t-test analysis for individual variables among two groups are provided.

(Pijnappels et al., 2004, 2005; Yang and Pai, 2010). For example, after the onset of slip, alteration of stance and swing limbs of the ankle, the knee, and the hip joint led to a change in COM stability (Yang and Pai, 2010), and sufficient knee and hip extensions before training limb liftoff together were major factors preventing a limb collapse (Pai et al., 2006). While after a trip onset, large ankle plantar flexion, knee flexion, and hip extension moments were key to generating the necessary push-off reaction and to restraining the forward angular moment (Pijnappels et al., 2004). Other than lower extremities, a larger peak shoulder flexion post-slip perturbation contributed to a lower fall rate by reducing the trunk extension velocity (Troy et al., 2009). In addition to multiple degrees of freedom adopted in the recovery of gait perturbation and despite proactive interference, muscle responses were rapid enough (usually under 100 ms after a perturbation onset before a recovery step) to allow the online adjustment of reactive control to some extent in both young and older adults (Pijnappels et al., 2005; Pai et al., 2006; Troy et al., 2009).

The findings of this study must be interpreted in light of its limitations. The slips were always introduced during RFT, while the trips were always triggered during left foot swing due to physical constraints in designing the floor for conducting such an experiment; however, in daily life, the slip or trip could occur on either leg. Hence, it was unclear whether such a design would increase or reduce the contextual interference. Moreover, 4% of subjects reported their left leg as the dominant legs, and differences in the dominant leg might contribute to the altered performances in response to perturbations. However, subjects who were left-footed had comparable age, height, and weight, as well as performance in the BBS, TUG, MMSE, and ABC (all $p > 0.05$), in comparison with those who were right-footed (**Table 2**). Moreover, most of the studies showed no differences between dominant and non-dominant legs in performing dynamic balance tasks in non-athletic adults (Paillard and Noé, 2020). In addition, only healthy older adults were included in the current study, which does not represent more vulnerable older populations who are more likely to fall.

In summary, similar to young adults, older adults who received repetitive perturbation training showed the ability to quickly generalize training-induced improvement in the reactive control to overcome negative interference in the proactive control to some extent during novel opposing perturbations. The findings suggest that a future design of perturbation training with mixed opposing conditions may reduce the reliance on feedforward adjustments but enhance the feedback control, which would better prepare older adults to prevent falls in a more complex, highly unpredictable situation that includes realistic environmental fall-risk factors.

DATA AVAILABILITY STATEMENT

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Institutional Review Board in the University of Illinois at Chicago. The patients/participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

TB designed and directed the project. YW, SW, and LK performed the experiments. SW and LK assisted with data analyses. TB, SW, YW, and LK helped with interpretation of results. TB, YW, and SW wrote the manuscript. All authors provided critical feedback and helped shape the research and manuscript.

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Perturbation-Based Balance Training Using Repeated Trips on a Walkway vs. Belt Accelerations on a Treadmill: A Cross-Over Randomised Controlled Trial in Community-Dwelling Older Adults

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Background: Walkway and treadmill induced trips have contrasting advantages, for instance walkway trips have high-ecological validity whereas belt accelerations on a treadmill have high-clinical feasibility for perturbation-based balance training (PBT). This study aimed to (i) compare adaptations to repeated overground trips with repeated treadmill belt accelerations in older adults and (ii) determine if adaptations to repeated treadmill belt accelerations can transfer to an actual trip on the walkway.

Method: Thirty-eight healthy community-dwelling older adults underwent one session each of walkway and treadmill PBT in a randomised crossover design on a single day. For both conditions, 11 trips were induced to either leg in pseudo-random locations interspersed with 20 normal walking trials. Dynamic balance (e.g., margin of stability) and gait (e.g., step length) parameters from 3D motion capture were used to examine adaptations in the walkway and treadmill PBT and transfer of adaptation from treadmill PBT to a walkway trip.

Results: No changes were observed in normal (no-trip) gait parameters in both training conditions, except for a small (0.9 cm) increase in minimum toe elevation during walkway walks ($P < 0.01$). An increase in the margin of stability and recovery step length was observed during walkway PBT ($P < 0.05$). During treadmill PBT, an increased MoS, step length and decreased trunk sway range were observed ($P < 0.05$). These adaptations to treadmill PBT did not transfer to a walkway trip.

Conclusions: This study demonstrated that older adults could learn to improve dynamic stability by repeated exposure to walkway trips as well as treadmill belt accelerations. However, the adaptations to treadmill belt accelerations did not transfer to an actual trip. To enhance the utility of treadmill PBT for overground trip recovery performance, further development of treadmill PBT protocols is recommended to improve ecological authenticity.

Keywords: perturbation, balance training, older adults, gait, exercise, accidental fall

INTRODUCTION

Falls in older people are a major health issue associated with significant morbidity, mortality (James et al., 2020), and economic burden (Davis et al., 2010). One-third of community-dwelling older adults fall annually (Lord et al., 1993), of which, 10–20% will require hospitalisation for complications such as hip fracture (Rubenstein, 2006). Evidence for fall prevention interventions consistently shows combinations of balance and functional exercises reduce the rate of falls, with an average effect of 34% (Sherrington et al., 2019). However, it has been suggested that the effects of conventional balance exercise are limited due to a lack of "task-specificity" to the balance recovery responses required to prevent falls (Grabiner et al., 2014). This has led to the development of perturbation-based balance training (PBT) which is a task-specific intervention exposing participants to repeated unexpected perturbations to improve reactive balance control (Mansfield et al., 2015; Gerards et al., 2017). A recent clinical trial found that PBT using an instrumented treadmill incorporated into conventional physiotherapy significantly reduced injurious falls in daily life, compared to physiotherapy alone (Lurie et al., 2020). Furthermore, systematic reviews and meta-analyses of randomised controlled trials (RCTs) have shown that PBT reduces the rate of falls by ∼50% in older adults and individuals with neurological conditions (Mansfield et al., 2015; Okubo et al., 2017).

Whilst the reported efficacy of PBT is promising, several important questions are yet to be answered. Many heterogeneous perturbation methods have been used to simulate and train reactive balance and the most effective method is unknown (Gerards et al., 2017; Okubo et al., 2017). Therapist-applied perturbations such as push, pull, and lean-and-release during stance have been used in clinical settings as they require minimal space and can be administered easily (Gerards et al., 2017; Mansfield et al., 2018). In contrast, overground perturbation systems with hidden tripping obstacles and low-friction surfaces have been generally used only in laboratory studies. These overground systems can more closely resemble "real-life" perturbations including trips and slips during gait (Pai et al., 2014; Okubo et al., 2019b; Wang et al., 2020), thus having the advantage of "task-specificity." However, since many of these systems require a long walkway and overhead harness track, their clinical feasibility is limited.

In contrast, an instrumented treadmill can deliver sudden perturbations during gait through belt acceleration and therefore offer a viable method for administering clinically feasible trip- and slip-like PBT. A study in 166 community-dwelling older adults reported significant transfer of training effects from treadmill-based slip training to improvement in balance recovery responses following an overground slip (Wang et al., 2019). Since trips are the most common cause of falls in community-dwelling older adults (Berg et al., 1997), previous studies used several treadmill methods to evoke trip-like balance responses such as belt accelerations (McCrum et al., 2018), ankle cable pulls (break-and-release) (Epro et al., 2018), and dropping an obstacle onto the belt (King et al., 2019). Although treadmill belt accelerations do not involve obstruction of the swinging foot, they simulate the overall forward trunk rotation and stepping during a trip to a certain degree (Sessoms et al., 2014). Thus, belt accelerations have been used as part of PBT in recent studies (McCrum et al., 2018, 2020; Lurie et al., 2020; Gerards et al., 2021). Because treadmill accelerations do not require additional perturbation devices other than an instrumented treadmill, the clinical feasibility of this approach may be high. However, it is important to clarify whether PBT using treadmill belt accelerations can provide meaningful adaptation to balance recovery from an actual trip. To our knowledge, no previous studies have examined whether adaptations to PBT with treadmill belt accelerations can transfer to actual overground trips.

The aims of this study were to (i) compare the training adaptations to repeated overground trips and treadmill belt accelerations in community-dwelling older adults and (ii) determine if any adaptations gained during treadmill PBT transferred to improved responses to a naïve overground trip. Based on previous studies (Bhatt and Pai, 2009; Wang et al., 2012; Okubo et al., 2019b), we hypothesised that both PBT regimes would induce significant and similar adaptations in dynamic stability during a trip and that participants with prior treadmill PBT would have significantly better responses in the overground trip, compared to those without prior PBT.

METHODOLOGY

Study Design

This study was a randomised crossover trial comparing treadmill and overground PBT, conducted at Neuroscience Research Australia (19 July 2019–3 March 2020). The study protocol was approved by the University of New South Wales Human Research Ethics Committee (HC16227).

Participants

Prospective participants were recruited via a research volunteer database. Eligibility criteria were aged 65+ years, living independently, ability to walk 20 min unassisted, and no neurological impairments or osteoporosis. Written informed consent was obtained from all the participants.

Randomisation

Thirty-eight participants were randomly allocated into either Group W-T ($n = 19$) or Group T-W ($n = 19$) based on the flip of a coin. Group W-T completed the walkway PBT first followed by a 15-min break and then the treadmill PBT. In contrast, Group T-W completed the treadmill PBT before the break, followed by walkway PBT (**Figure 1**).

Baseline Measurements

Participants were assessed regarding their concern about falling [Falls Efficacy Scale – International (Yardley et al., 2005)], mental health [Hospital Anxiety and Depression Scale (Snaith, 2003)], and falls in the past year.

Experimental Protocol

Participants initially walked at the usual pace for three repeated trials over an 8-m course with a 5.7-m long electronic mat

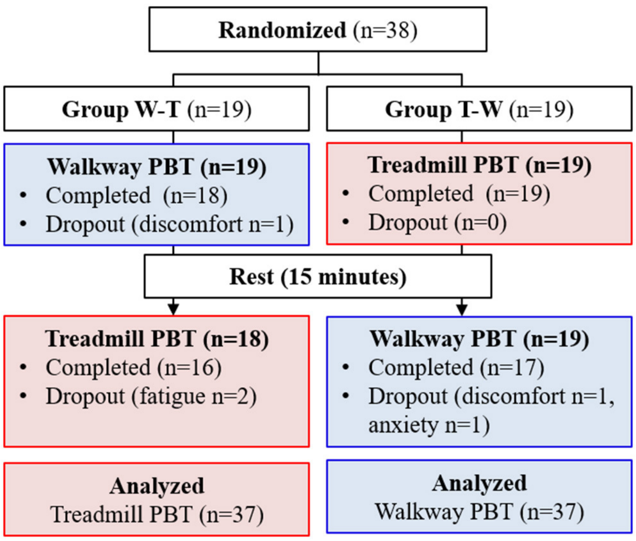

FIGURE 1 | Flow chart of the cross-over randomised controlled trial.

(GAITRite, CIR Systems, New Jersey, USA) to determine their step length, cadence, and gait speed to be used in the PBT conditions. In preparation for both walkway and treadmill PBT, participants were fitted with a ceiling-mounted full-body harness adjusted such that when hanging in the harness, their knees were 10 cm above the floor to prevent contact with the ground in the event of a fall.

Walkway PBT Setup

Walkway PBT involved 11 trips and 18 normal walks on a custom-built 10 m wooden walkway (**Supplementary Figure 1**) (Okubo et al., 2018, 2019b). Target stepping tiles were placed along the walkway at 95% of individual usual step length, whilst a metronome was set to 95% of their usual cadence. During a 3-min practice and throughout training, participants were instructed to walk while stepping on the target tiles in time with the metronome beat, yielding a walking speed of 90% of their usual speed. If the gait of the participant did not match the metronome timing and stepping tile locations (by visual inspection), then additional familiarisation trials were undertaken.

Trips were induced by a 14 cm height spring-loaded tripping board which flipped up when activated by the participant moving over an optical foot detection sensor hidden in the walkway. The tripping board was positioned at the late-swing phase (at ∼60–70% of the gait cycle from foot contact) to increase the likelihood of a lowering strategy (Eng et al., 1994) to induce a similar response to the treadmill condition (an elevating strategy never occurs on a treadmill). To minimise prediction of a trip, 18 normal walk (no-trip) trials were interspersed with 11 trip trials, presented in various locations (left or right side and near, middle, or far position) in a pre-determined, pseudo-random order (**Table 1**; **Supplementary Figure 1**). Participants were instructed that they may experience a hazard anywhere and at any time whilst walking on the walkway but to try to continue walking normally. To evaluate potential training effects in a consistent manner, the first (T1), fourth (T4), seventh (T7), and eleventh (T11) trips were delivered to the left leg in the middle of the walkway.

Treadmill PBT Setup

The treadmill PBT was conducted on a dual-belt, instrumented treadmill (M-Gait, Motekforce Link, Amsterdam, The Netherlands) controlled by custom-written software within D-Flow 3.30.2 (Motek Medical B.V., Amsterdam, The Netherlands) interfaced with an 8-camera Vicon motion capture system (Bonita, Vicon Motion Systems Ltd., Oxford, UK). During a 3-min practice period and throughout training, participants walked on the treadmill with the belt speed set to 90% of their individual walking speed. A perturbation was induced by a sudden acceleration of one side of the treadmill belt at 8 m/s^2 to up to 200% of the walking belt speed. The belt acceleration began at approximately mid-swing of the gait cycle (triggered by a hallux marker of the to-be-perturbed limb passed the hallux marker of the stance limb in the sagittal plane) so that perturbation was delivered at the subsequent foot strike (McCrum et al., 2018). Each treadmill belt perturbation was delivered for 30% of stride taken from the average time of the previous three strides. Participants were instructed that they may experience a hazard at any time whilst walking but to try to continue walking normally. Similar to the walkway, treadmill PBT involved 11 belt accelerations interspersed with 18 (30–90 s) long bouts of normal (no-trip) walking. Each walk was however in a continuous sequence (**Table 1**). To minimise prediction, belt accelerations were induced to both left and right legs in a pre-determined pseudo-random order. The first (T1), fourth (T4), seventh (T7), and eleventh (T11) belt accelerations were induced on the left leg to be used for analysis.

Outcome Measures

Falls Incidence and Recovery Strategy

A fall was defined by a post-trip harness supported load of >30% of the body weight of the participant (Yang and Pai, 2011) as measured by a load cell in series with the harness line. Walkway trip recovery strategies were classified as either a lowering strategy (i.e., when the obstructed foot immediately stepped down in front of the obstacle) or an elevating strategy (i.e., when the obstructed foot was elevated to clear the obstacle).

Kinematics

Eight-camera motion capture systems (Vicon Motion Systems Ltd., Oxford, UK) were used to collect 3D kinematic data during the treadmill (Bonita cameras) and walkway (Vantage cameras) PBT sessions. Thirty-nine 14-mm diameter retroreflective markers were attached to anatomical landmarks according to the Plug-in-Gait full-body model marker set (Vicon Motion Systems,

TABLE 1 | The training protocol used for both treadmill and overground training.

Trial type	Trip location	Tripped foot	Anxiety and perceived difficulty
N1			Check
N2			
N3			
N4			
N5			Check
T1	Middle	Left	
N6			
T2	Near	Left	
N7			
T3	Middle	Right	
N8			
N9			
T4	Middle	Left	
N10			
T5	Far	Left	Check
N11			
T6	Middle	Right	
N12			
N13			
T7	Middle	Left	
N14			
T8	Far	Left	Check
N15			
T9	Middle	Right	
N16			
T10	Near	Left	
N17			
N18			
T11	Middle	Left	Check

"T" denotes a trip trial. "N" denotes a normal (no-trip) walk trial. Trials on the treadmill were conducted continuously, thus a series of 30 steps was considered a trial, equivalent to one return walk over the 10 m overground walkway. The trip location denotes the position of the tripping board on the walkway, which was not relevant on the treadmill because participants walk in place over the moving treadmill belt. Shaded trials were used for statistical analysis.

2017). Kinematic variables were calculated from sagittal-plane marker trajectories using custom software in MATLAB R2019b (The MathWorks, Inc., MA, USA) (see **Supplementary Table 1** for detail).

To assess predictive and reactive gait adaptations during trip trials, the following kinematic parameters were calculated one step before (Pre1) and the first (Rec1), second (Rec2), and third (Rec3) steps after trip-onset (i.e., one previous and three recovery steps). On the walkway, the step that cleared the tripping board was treated as Rec1, that is, the tripped (left) footstep in an elevating strategy and the contralateral (right) footstep in the lowering strategy.

As a measure of dynamic stability, the margin of stability (MoS) in the anterior–posterior direction was calculated at step touchdown. The MoS is the distance (cm) between the closest edge (usually the toe) of the base of support and extrapolated centre of mass (*XCoM*), which accounts for the velocity of the *CoM* (Hof et al., 2005; Süptitz et al., 2013):

$$XCoM = P_{CoM} + \frac{V_{CoM} + \overline{V}_{BoS}}{\sqrt{\frac{g}{L}}}$$

where P_{CoM} is the position of the *CoM* estimated by the Dynamic Plug-in-Gait model (relative to the ankle marker of the trailing limb), *VCoM* is the velocity of the *CoM*, $\overline{V}_{BoS}$ is the velocity of the heel marker on the belt (averaged during stance phase), g is gravitational acceleration (9.81 m/s^2), and L is the sagittal distance between the *CoM* and the ankle joint centre. V_{BoS} was assumed zero for walkway trials. A positive MoS indicates the *XCoM* is within the base of support and therefore a stable body configuration. A negative MoS indicates an unstable body configuration and a requirement to take additional steps to avoid a fall.

To quantify the magnitude of the balance perturbation, anteroposterior distance (cm) between *XCoM* and the rear ankle joint centre (marker) of the trailing limb was calculated at step touchdown. A positive value indicates the forward location of the *XCoM* relative to the rear foot ankle joint centre. Maximum toe elevation was also measured at previous and recovery steps. Trunk sway range was defined as maximal angular displacement of the trunk over one previous step or three recovery steps.

During normal walking trials, spatiotemporal gait parameters including step length, cadence, gait speed, and minimum toe elevation were calculated for subsequent analysis.

Self-Reported Anxiety and Difficulty Levels

Participants were asked to report their level of anxiety and perceived difficulty prior to N1, following N5 (prior to T1), T5, T8, and T11. Participants reported anxiety using a 5-point scale with one representing "not at all" and five representing "extremely anxious." Participants reported their perceived difficulty in the last trial using a five-point scale with one representing "easy" and five representing "too hard." The anxiety of the participants and perceived difficulty scores at the five time points were averaged for the analyses.

Statistical Analysis

Approximate normality of variable distributions was confirmed with the Shapiro-Wilk test and visual inspection of Q–Q plots, and logarithmic (base 10) transformation for skewed data was conducted if required to allow parametric analysis. Changes in anxiety and perceived difficulty (N1 vs. N5/T5/T8/T11) during treadmill and walkway PBT were tested using Wilcoxon signed rank test with Bonferroni adjustments. Average anxiety and perceived difficulty scores for the trip trials (N1, N5, T5, T8, and T11) were also compared between the walkway vs. treadmill PBT using a paired t-test. Potential predictive gait adaptations during normal walks (N5 vs. N6/N9/N13/N18) were examined using the spatiotemporal gait parameters with a generalised linear mixed model with robust estimation (robust against violations of model assumptions) and sequential Bonferroni adjustments. Potential training effects (predictive and reactive gait adaptation) on pre- and post-trip kinematics were examined using a generalised linear mixed model with time (T1, T4, T7, and T11), step (Pre, Rec1, Rec2, and Rec3) and condition (treadmill, walkway) entered as factors and interaction effects adjusted for the group (i.e., training order). Changes from T1 to T4/T7/T11 within each step were examined by *post-hoc* pairwise comparisons with sequential Bonferroni corrections. Transfer of any training effects from treadmill PBT to a walkway trip was examined by comparing the first walkway trip (T1) parameters between Group W-T (prior to training) and Group T-W (after treadmill PBT) using independent-samples t-tests. All statistical tests were conducted using IBM SPSS Statistics 25 (IBM Corp., New York, USA). $P < 0.05$ was considered statistically significant.

TABLE 2 | Participant characteristics, fall history, and usual gait parameters.

Variables	Total sample ($n = 38$)	Group O-T ($n = 19$)	Group T-O ($n = 19$)	P
Age (years)	73.6 (4.7)	74.0 (5.1)	73.2 (4.3)	0.632
Sex, N (% female)	21 (55.3%)	11 (57.9%)	10 (52.6%)	0.744
Height (m)	1.69 (0.10)	1.69 (0.02)	1.69 (0.09)	0.992
Weight (kg)	74.3 (13.1)	75.2 (12.5)	73.4 (14.0)	0.674
BMI (kg/m^2)	26.0 (3.5)	26.4 (3.4)	25.6 (3.6)	0.502
Dominant leg, N (% right)	38 (100%)	19 (100%)	19 (100%)	1.000
FES-I (score)	18.8 (3.6)	18.7 (4.4)	18.9 (2.8)	0.861
HAD (score)	3.95 (3.42)	3.68 (3.59)	4.21 (3.33)	0.642
Fallers, N (%)*	19 (50%)	11 (57.9%)	8 (42.1%)	0.330
Multiple fallers, N (%)**	10 (26.3%)	5 (26.3%)	5 (26.3%)	1.000
Step length (m)	0.65 (0.10)	0.65 (0.08)	0.64 (0.12)	0.877
Cadence (steps/min)	107.5 (9.0)	106.3 (7.4)	108.7 (10.4)	0.402
Gait speed (m/s)	1.16 (0.21)	1.15 (0.15)	1.17 (0.26)	0.709

Data are mean (SD) or N (%). BMI, body mass index; FES-I, Falls Efficacy Scale-International; HAD, Hospital Anxiety and Depression scale.

**Number of people reporting at least 1 fall for the previous 12 months.*

***Number of people reporting 2 or more falls for the previous 12 months.*

RESULTS

Participant Characteristics

Thirty-eight participants were recruited and randomised into Group W-T ($n = 19$) or Group T-W ($n = 19$). Five participants (13%) could not complete all the protocols due to fatigue ($n = 2$), discomfort ($n = 1$), and anxiety ($n = 2$). Three out of 38 participants (7.9%) dropped out during the walkway PBT and 2 out of 37 participants (5.4%) dropped out during the treadmill PBT (**Figure 1**).

The characteristics, falls, and usual gait parameters of the participants are summarised in **Table 2**. There were no differences between the W-T and T-W groups in the proportion of women or participant age, height, weight, body mass index, leg dominance, past falls, fear of falling, depressive symptoms, or gait parameters ($P > 0.05$).

Anxiety and Perceived Difficulty

On average, participants reported significantly higher anxiety during treadmill PBT compared to walkway PBT (1.82 ± 0.83 vs. 1.58 ± 0.59, $P = 0.030$). Average perceived difficulty scores during treadmill PBT were also significantly higher compared to walkway PBT (2.02 ± 0.74 vs. 1.65 ± 0.54, $P = 0.001$). There were no significant changes of anxiety or perceived difficulty over time during both training conditions, except for a significant decrease of perceived difficulty from N1 to N5 on the walkway ($P = 0.04$).

Walkway PBT

No significant differences in gait speed and step length were detected among the walkway normal walks (N5 vs. N6/N9/N13/N18, $P > 0.05$, **Figure 2**). Minimum toe elevation during normal walks in both groups was significantly increased in N6 (2.6 ± 1.3 cm), N9 (2.5 ± 1.1 cm), N13 (2.9 ± 1.6 cm), and N18 (2.9 ± 1.6 cm) compared to N5 (2.0 ± 0.9 cm) prior to the first trip ($P < 0.01$). A significant increase in walkway cadence was seen at N6 (110 ± 11.0 steps/min) and N9 (110 ± 10.3 steps/min) compared to N5 (107 ± 9.8 steps/min) ($P < 0.05$).

There were no changes in any of the kinematic parameters during the previous step in walkway trip trials (T1 vs. T4/T7/T11, $P > 0.05$) (**Figure 3**). A significant increase in step length was observed in Rec1 (T1: 62.9 ± 12.7 cm, T11: 70.5 ± 12.1 cm) and Rec2 (T1: 51.4 ± 18.6 cm, T11: 60.7 ± 15.9 cm) ($P < 0.05$). The MoS also significantly improved in Rec1 (T1: −19.2 ± 13.8 cm, T11: −7.8 ± 13.6 cm), Rec2 (T1: −12.5 ± 14.8 cm, T11: −1.2 ± 10.4 cm), and Rec3 (T1: −6.4 ± 14.4 cm, T11: 3.0 ± 8.9 cm) ($P < 0.01$). No significant changes over time were found in recovery step *XCoM*, trunk sway range, and maximum toe elevation on the walkway ($P > 0.05$).

Treadmill PBT

During the treadmill normal walks, no significant changes were observed in any of the spatiotemporal gait parameters over time (N5 vs. N6/N9/N13/N18, $P > 0.05$) (**Figure 2**). Similarly, during the belt acceleration trials on the treadmill, there were no significant changes in the previous step kinematic parameters ($P > 0.05$) (**Figure 3**). A significant increase from T1 to T11 was found in Rec3 MoS (T1: 2.3 ± 8.7 cm, T11: 6.1 ± 5.6 cm), Rec2 step length (T1: 30.1 ± 18.3 cm, T11: 43.9 ± 18.1 cm), Rec2 *XCoM* (T1: 46.6 ± 18.7 cm, T11: 60.3 ± 17.4 cm), and Rec3 maximum toe elevation (T1: 3.8 ± 2.9 cm, T11: 5.6 ± 2.9 cm). A significant reduction in recovery trunk sway range on the treadmill was observed from T1 (17.3 ± 7.2 deg) and T11 (13.5 ± 5.4 deg) ($P < 0.05$).

Interactions Between Time and Condition

A significant time- and condition-interaction was detected in MoS ($P < 0.001$) indicating greater improvements during the walkway PBT, compared to during treadmill PBT. Another

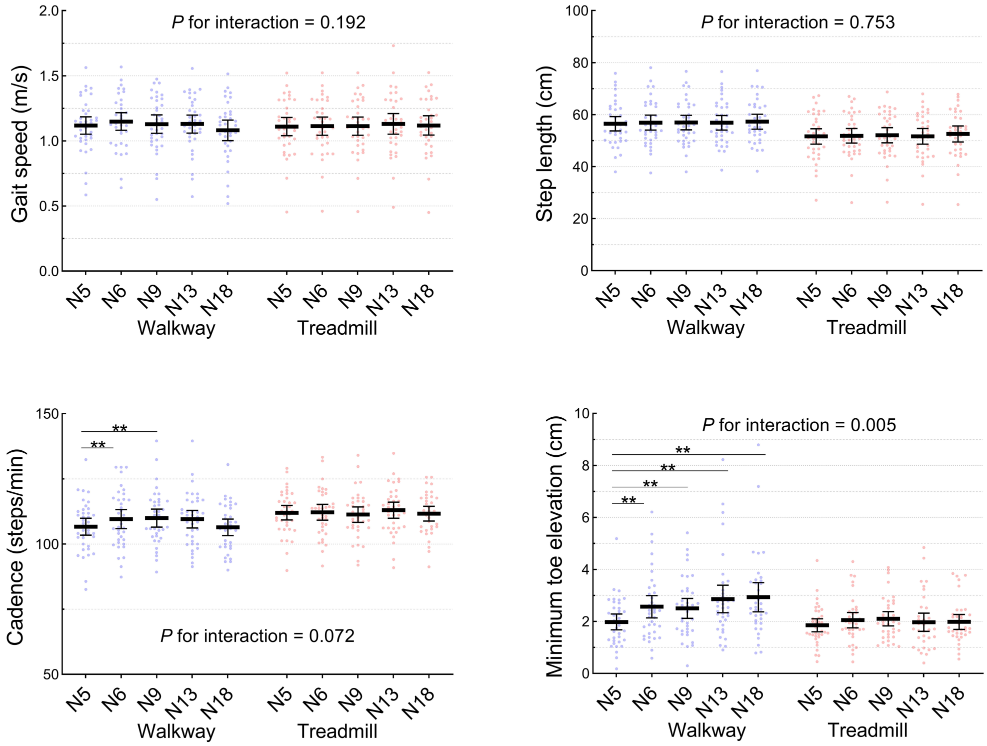

FIGURE 2 | Spatiotemporal parameters during normal walks on the treadmill and overground walkway ($n = 38$). N5 and N6 were prior to and following the first trip, respectively. The middle and error bars represent mean and 95% confidence interval. $^{**}P < 0.01$.

significant interaction in *XCoM* ($P = 0.024$) indicated a greater increase in treadmill PBT than walkway PBT. No significant interactions were observed in step length, maximum toe elevation, and trunk sway range ($P > 0.05$).

Transfer of Training Adaptations From the Treadmill PBT to a Walkway Trip

During the first walkway trip (T1), there were no significant differences in any kinematic parameters in any steps between Group T-W (who had previously completed treadmill PBT) and Group W-T (who had no prior training) ($P > 0.05$) (**Figure 4**).

DISCUSSION

This cross-over trial is the first to directly compare PBT involving walkway trips against PBT involving belt accelerations on a treadmill. The walkway PBT resulted in improved dynamic stability and greater step length following a trip, which supports our first hypothesis. The treadmill PBT also resulted in improved MoS, *XCoM*, step length, and less trunk sway following a belt acceleration, but contrary to our second hypothesis, treadmill PBT did not transfer to better recovery to a first trip on the walkway.

Adaptations to PBT Using Trips on a Walkway

This trial demonstrated that older adults could improve their balance recovery following walkway trips. A similar increase in dynamic stability has been reported by previous studies that trained young and older adults with 8–24 walkway trips (Wang et al., 2012, 2020; Bhatt et al., 2013). However, previous studies administered all trips to the left foot in a fixed location resulting in a significant predictive gait adaptation seen as increased toe elevation (8–10 cm) (Wang et al., 2012, 2020; Bhatt et al., 2013) and the majority (12–60%) of participants avoided the obstacle on the last trip. In contrast, our walkway method maintained a high level of unpredictability in repeated trials by randomly inducing trips to both feet in various hidden locations. Therefore, we detected no predictive gait adaptations except for a small increase in minimum toe elevation (0.9 ± 1.1 cm) during normal walk trials. Maximum toe elevation in the previous step was 8.6 ± 2.1 cm and the tripping board was sufficiently high (14 cm) to induce legitimate trips to examine reactive adaptation during balance recovery. The unchanged gait speed, recovery step *XCoM*, and trunk sway suggest that the magnitude of balance perturbation induced by the trips was constant throughout the repeated trials. Thus, the increased MoS during the recovery step likely reflects the improved balance recovery response to trips. The walkway trips involved obstruction of the swing foot, substantial forward shift of *XCoM*, and trunk sway. Thus, it was necessary to take

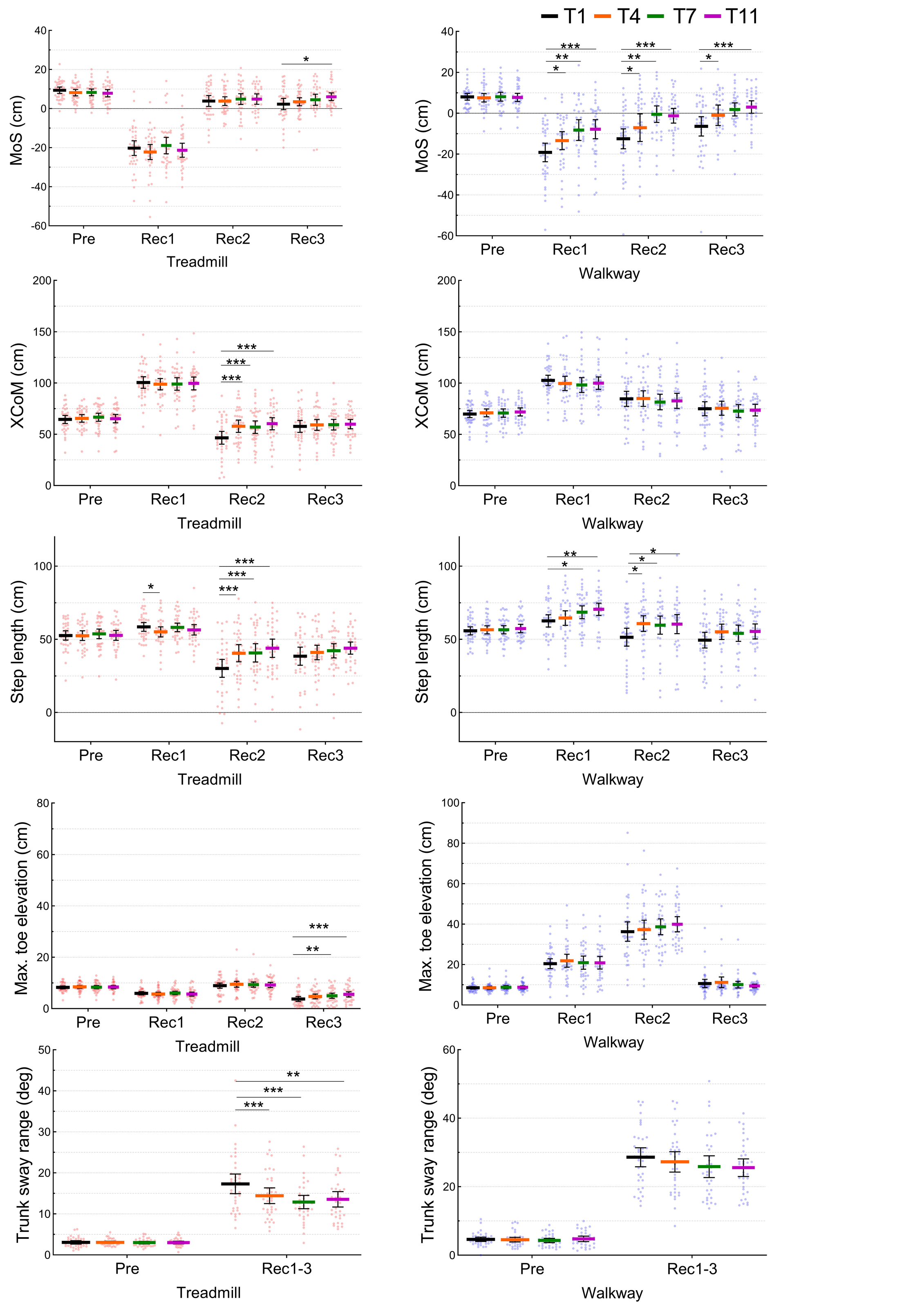

FIGURE 3 | Kinematic parameters at one previous (Pre) and three recovery (Rec1, Rec2, and Rec3) steps during four trip trials (T1, T4, T7, and T11) on the treadmill and overground walkway ($n = 38$). The mid lines and error bars represent means and 95% confidence intervals. $^{*}P < 0.05$, $^{**}P < 0.01$, $^{***}P < 0.001$.

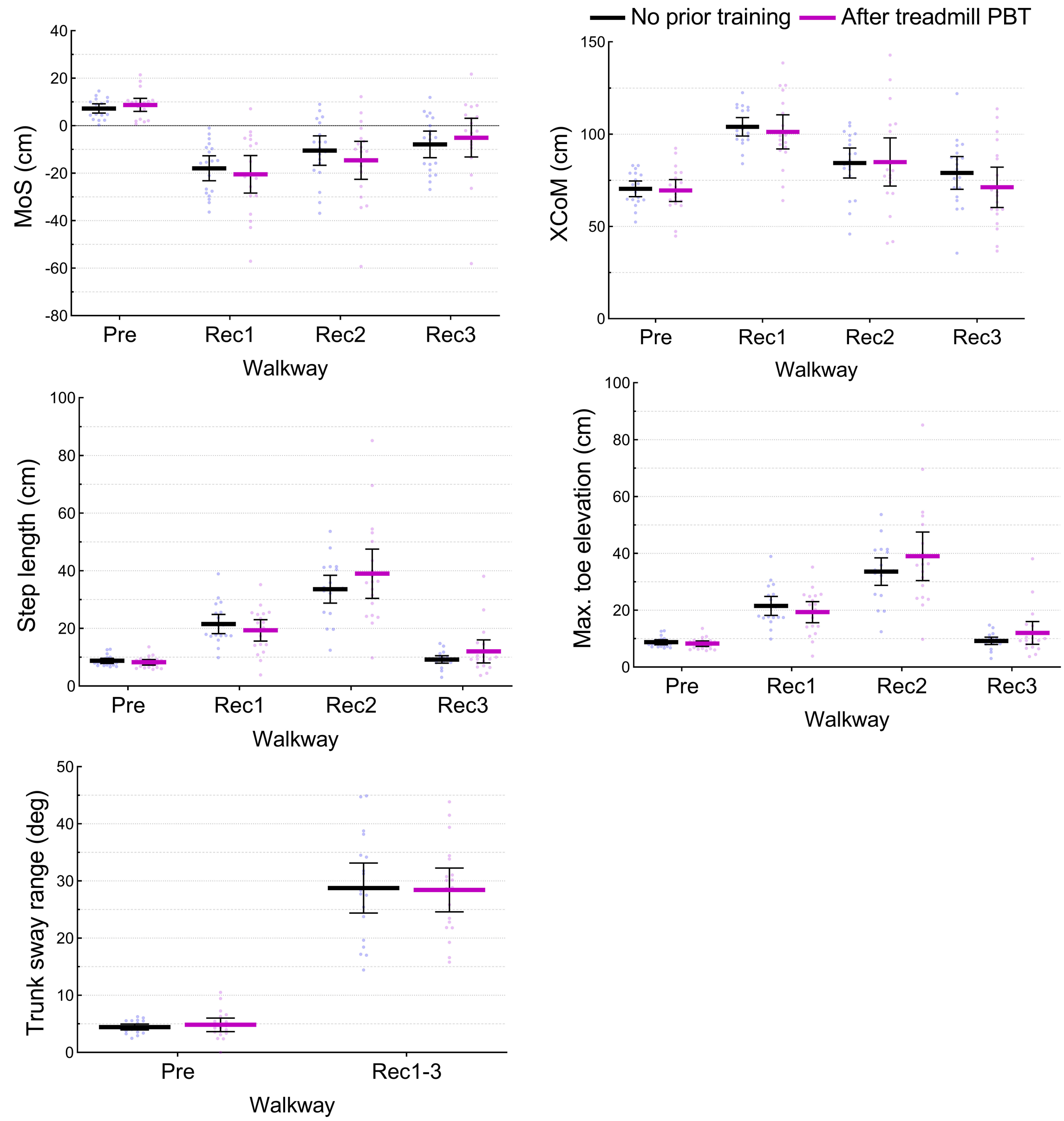

FIGURE 4 | Test of transfer from treadmill PBT to an actual trip. Kinematic parameters at one previous (Pre) and three recovery (Rec1, Rec2, and Rec3) steps during the first walkway trip (T1) were compared between Group T-W (who had previously completed treadmill PBT) and Group W-T (who had no prior training). The mid lines and error bars represent means and 95% confidence intervals. No significant differences were detected ($P > 0.05$).

a longer recovery step (i.e., base of support) (Okubo et al., 2018) to provide a counter torque to catch the falling upper body. The ability to rapidly generate an extensor moment and position in the recovery limb has been identified as one of the key intrinsic limitations to balance recovery in older adults (van Dieen et al., 2005). This study shows that unexpected walkway trips can successfully train older adults to take longer recovery steps to increase the likelihood of balance recovery.

Adaptations to PBT Using Treadmill Belt Accelerations

The improvement in MoS was found in the third recovery step over the 11 treadmill belt accelerations (8 m/s^2 to 200% of walking speed). This is consistent with a previous study that exposed young and older adults to 10 belt accelerations (3 m/s^2 to 180% of walking speed) and found improved MoS during the third to fifth recovery steps (McCrum et al., 2020). We also found increased *XCoM* and step length in the second recovery step which also replicated the results reported by McCrum et al. who also reported full retention over 1 month (McCrum et al., 2018). A reduction in trunk sway during recovery steps was also found as reactive balance adaptation to repeated exposure to treadmill belt accelerations. This agrees with a study in 16 stroke patients who underwent a single session of 15 treadmill perturbations from standing (22 cm displacement, acceleration/deceleration ± 13.89 m/s^2, velocity 0.56 m/s) and reported a reduction in trunk flexion but no improvement in MoS (Nevisipour et al., 2019). An RCT of 30 older adults who walked on a treadmill also reported an improvement in trunk control (i.e., reduction in trunk velocity) following both anterior–posterior (deceleration -9 m/s^2 for 0.12 s) and medio-lateral (displacement 5 cm in 0.31 s) perturbations, which was retained after 1 week (Rieger et al., 2020). Interestingly, they found no difference between the intervention (16 perturbations) and control group indicating exposure to eight perturbations during the baseline assessment was sufficient to improve trunk control. This rapid adaptation coincides with our finding showing a reduction in trunk sway by T4 and T7. It is possible that the body has rapidly adapted to relax and reduce stiffness and thus less momentum is transferred from the foot on the suddenly accelerated treadmill belt to trunk flexion. Thus, these significant improvements in MoS, *XCoM*, recovery step length, and trunk sway reaffirm there is some capacity for reactive adaptation during treadmill PBT but the benefit of such adaptation needs to be examined.

Transfer From Treadmill PBT to an Actual Trip

Following completion of treadmill PBT (Group T-W), the response to the first walkway trip was not significantly different from those with no prior training (Group W-T). Treadmill PBT has high clinical feasibility requiring less space, time, and human resources compared to walkway PBT. However, our findings indicate that the adaptation to treadmill belt accelerations may not improve recovery from real-life overground trips; likely because treadmill PBT did not provide the motor skills to deal with obstacles. A small increase in maximal toe elevation (on average 3.8 cm in T1 to 5.6 cm in T11 in Rec3) during the treadmill PBT was clearly not sufficient during the actual trip that required much higher foot elevation (on average 20.4 cm in Rec1, 26.3 cm in Rec2, 10.6 cm in Rec3 in T1). Our findings contrast to a previous study conducted on 34 young adults reporting significant beneficial effects of treadmill slip training on overground slip recovery (Yang et al., 2013). Such differences in transferability of slip and trip recovery training effects between treadmill and walkway PBT likely reflect the degree of shared biomechanical properties. Simulated slips induced by the deceleration or reverse rotation of the treadmill belt can replicate forward slipping of the leading foot. However, belt accelerations on the treadmill involve rapidly shifting the stance foot backward to induce forward rotation of the upper body, which requires rapid reactive stepping. Although the overall body response may be similar, a treadmill belt acceleration differs from an overground trip where the swing foot is physically obstructed requiring immediate elevation or lowering of the foot (Eng et al., 1994). It is likely that adaptations induced by PBT are highly task-specific and the greater the difference in biomechanical properties of the training, the more limited the transferability of training effects across different conditions. Indeed, König et al. found no transfer of training effects from treadmill trip training using ankle cable pulls to performance on a lean-and-release task that did not involve obstruction of the foot (Konig et al., 2019).

Two studies have reported training obstacle-clearing from an initial stance position on a treadmill can improve balance responses to actual trips. Grabiner et al. conducted an RCT involving 52 healthy middle-aged women in which intervention participants were trained to recover from sudden treadmill accelerations from rest (stance) by stepping over a 5 cm high foam obstacle (Grabiner et al., 2012). They found that following 120–150 treadmill-induced perturbations over 4 weeks, intervention participants had significantly fewer falls on an overground trip test compared to controls. Similar findings were reported by Bieryla et al. in that a training program involving 20 treadmill accelerations from rest requiring a step over a 7.6 cm high obstacle produced improved trunk control during an overground test trip in a small trial of 12 older adults (Bieryla et al., 2007). It is also possible to administer obstacle-induced trips on a treadmill by dropping an obstacle onto the belt (King et al., 2019) but the increased complexity limits its feasibility in clinical settings. An instrumented treadmill that provides belt accelerations may be a useful way to train balance responses to backward slips at heel strikes (Yang et al., 2013) and forward slips at the late stance phase (Debelle et al., 2020) but may not be sufficient in preparing older adults for an actual trip. Further refinement of treadmill PBT protocols including belt kinematics and/or methods of delivering foot obstruction, as well as determination of optimal training doses and longer-term follow-ups are required to better clarify the clinical role of treadmill PBT training.

Anxiety and Perceived Difficulty

Anxiety can negatively affect reactive balance control (e.g., delayed and more rigid responses) (Carpenter et al., 2004; Okubo et al., 2021), and thus should be minimised for better training outcomes. However, only a few studies have quantified anxiety during PBT (Okubo et al., 2019a,b). Anxiety and perceived difficulty were higher during PBT on a treadmill compared to PBT on the walkway. Since the magnitude of perturbations induced by the treadmill was not greater than that on the walkway, this higher anxiety and perceived difficulty were likely due to unfamiliarity to treadmill walking and the elevated surface of the large, instrumented treadmill. The provision of a surrounding platform at the level of the treadmill belt surface may assist in reducing anxiety during treadmill PBT.

Limitations

This study has some limitations that warrant attention. First, study participants were healthy older adults who may not be representative of the older population. Older adults in poorer health or with increased fear of falling may show lower acceptability to PBT. Second, whilst we used the walkway trip as a surrogate for real-world trips, our study findings should be verified with a sufficient sample size and follow-up for evaluating the effect of PBT on falls in daily life.

CONCLUSIONS

This study demonstrated that older adults can learn to improve dynamic stability and stepping by repeated exposure to walkway trips. Exposure to belt accelerations on the treadmill may also improve dynamic stability, stepping, and trunk control in older adults. However, these adaptations obtained on a treadmill are likely not generalisable to an overground trip. Further refinement of treadmill trip training protocols to improve ecological authenticity while maintaining clinical feasibility is required.

DATA AVAILABILITY STATEMENT

The datasets analysed for this study can be provided to upon reasonable request approved by the University of New South Wales Human Research Ethics Committee. Requests to access the datasets should be directed to yoshiro_okubo@yahoo.co.jp.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by University of New South Wales Human Research Ethics Committee. The participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

YO, DS, PS, and SL contributed to the conception of the study. YO, PS, and DS contributed to the experimental setup. PS, YO, and MD conducted the participant recruitment, data acquisition, and processing. PS and YO conducted statistical analyses and drafted the manuscript. DS, MD, and SL revised the article for important intellectual content. All authors approved the final version.

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SUPPLEMENTARY MATERIAL

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fspor.2021.702320/full#supplementary-material

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ORIGINAL RESEARCH
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A Perturbed Postural Balance Test Using an Instrumented Treadmill – Precision and Accuracy of Belt Movement and Test-Retest Reliability of Balance Measures

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A perturbed postural balance test can be used to investigate balance control under mechanical disturbances. The test is typically performed using purpose-built movable force plates. As instrumented treadmills become increasingly common in biomechanics laboratories and in clinical settings, these devices could be potentially used to assess and train balance control. The purpose of the study was to investigate how an instrumented treadmill applies to perturbed postural balance test. This was investigated by assessing the precision and reliability of the treadmill belt movement and the test-retest reliability of perturbed postural balance test over 5 days. Postural balance variables were calculated from the center of pressure trajectory and included peak displacement, time to peak displacement, and recovery displacement. Additionally, the study investigated short-term learning effects over the 5 days. Eight healthy participants (aged 24–43 years) were assessed for 5 consecutive days with four different perturbation protocols. Center of pressure (COP) data were collected using the force plates of the treadmill while participant and belt movements were measured with an optical motion capture system. The results show that the treadmill can reliably deliver the intended perturbations with <1% deviation in total displacement and with minimal variability between days and participants (typical errors 0.06–2.71%). However, the treadmill was not able to reach the programmed 4 m/s^2 acceleration, reaching only about 75% of it. Test–retest reliability of the selected postural balance variables ranged from poor to good (ICC 0.156–0.752) with typical errors between 4.3 and 28.2%. Learning effects were detected based on linear or quadratic trends ($p < 0.05$) in peak displacement of the slow forward and fast backward protocols and in time to peak displacement in slow and fast backward protocols. The participants altered the initial location of the COP relative to the foot depending on the direction of the perturbation. In conclusion, the precision and accuracy of belt movement were found to be excellent. Test-retest reliability of the balance test utilizing an instrumented treadmill ranged from poor to good which is, in line with previous investigations using purpose-built devices for perturbed postural balance assessment.

Keywords: postural balance, instrumented treadmill, perturbation, reliability, accuracy, precision

INTRODUCTION

Human postural balance has been defined as the ability to sustain an upright posture (Papengaaij et al., 2014). Low et al. (2017) defined postural control as maintaining, achieving, or restoring postural balance despite executable tasks. Sufficient postural control is crucial for executing activities of daily living. Thus, postural control has an important role in everyday life (Jancova, 2008; Anson et al., 2017). Postural control requires the integration and smooth coordination of multiple sensorimotor systems, namely, visual, vestibular, somatosensory, and higher-level premotor, and motor systems (Mancini and Horak, 2010). Impaired postural control may result in falls because of loss of balance. Around one-third of people aged over 60 years fall yearly, and fall risks increase substantially with advancing age (Gerards et al., 2017). Neurological and musculoskeletal disorders deteriorate postural control, thus having a negative effect on safe mobility (Mancini and Horak, 2010). Therefore, maintaining and improving postural control and balance are an essential goal of clinical interventions (de Jong et al., 2020), and research is needed to support the development of effective interventions.

Based on a traditional definition, balance control can be divided into static balance control in which the center of mass movements maintains the balance, while the base of support remains stationary; and dynamic balance control in which both the center of mass and base of support are moving (Shumway-Cook and Woollacott, 2016). This traditional definition does not capture all the important aspects of balance control; thus, Shumway-Cook and Woollacott (2016) suggest postural balance control to be divided into four types: static steady-state balance: maintaining a steady position while sitting or standing; dynamic steady-state balance: maintaining a steady position during movements such as walking; proactive balance: anticipation of a predicted postural disturbance; reactive balance: response to an unpredicted postural disturbance. Numerous postural balance tests exist in clinical use such as the Berg Balance Scale (BBS) and Timed Up and Go (TUG). These tests are easy and quick to perform and thus, are often used in clinical practice. However, they may be subjective, lack responsiveness to small changes, and are not always sensitive enough to detect early deterioration in postural balance or changes due to interventions (de Jong et al., 2020). Moreover, these tests simultaneously assess many of the above-mentioned four types of balance control but provide little information for research on mechanisms of balance improvements or targets for practical interventions.

Recently, mainly because of technological development, computerized dynamic posturography with purpose-built devices has been increasingly used for measuring postural balance. These devices typically consist of a force plate on top of a movable platform which allows perturbation applied through the base of support. These computerized posturography devices measure the adaptive mechanisms of the whole postural control system including sensory, motor, and central mechanisms (Yuntao et al., 2017). The benefit of these devices is that they can assess multiple aspects of balance, namely, static steady-state, proactive, and reactive balance. A drawback of typical computerized dynamics posturography is that the test is performed in a standing posture, but most falls occur during walking or sit-to-stance transfers. Still, balance control under perturbed standing conditions predicts future falls (Sturnieks et al., 2013), and training on perturbed standing conditions reduces fall incidences (Rosenblatt et al., 2013). Thus, the controlled environment that a standing condition provides has a value in both balance testing and training contexts despite not being the particular task in which falls typically occur.

Purpose-built computerized dynamic posturography devices can only be used for a single purpose, which makes them costly investments for research institutes but, unlike clinical tests, allow the measurement of a specific aspect of postural balance performance,. There has been an increase in the use of treadmills with integrated force sensors (i.e., instrumented treadmills) for the investigation of human locomotion, thus, this type of treadmill has become accessible for an increasing number of researchers and clinical practitioners. Instrumented treadmills can measure the required parameter for postural balance assessment, namely, the center of pressure (COP). Additionally, they can be used to perturb balance. Therefore, they provide instrumentation for performing dynamic posturography measurements to assess static and dynamic steady-state, proactive, and reactive balance with devices already existing in many laboratories. However, the reliability and validity of instrumented treadmills for postural balance measurements have been questioned. Instrumented treadmills are susceptible to errors especially in ground reaction force and COP measurements (Sloot et al., 2015) because of mechanical noise or vibrations induced by the treadmill structure to the sensors (Willems and Gosseye, 2013). On the other hand, Fortune et al. (2017) showed that the COP measurement accuracy of an instrumented treadmill can be on par with a traditional ground-mounted force plate, and Collins et al. (2009) showed that COP error can be reduced to a similar level compared with a ground-mounted force plate using a calibration procedure. There can be also differences in the accuracy between devices from different manufacturers to deliver perturbations, which Crenshaw et al. (2019) speculated to be due to unique control strategies and computations. Nevertheless, encouraging results were provided by a preliminary feasibility study conducted by Yuntao et al. (2017), who evaluated the use of an instrumented treadmill (FTM-1200WA; Tec Gihan, Kyoto, Japan) as a standing postural balance measurement device. The study indicated that the reliability of the treadmill-based measurement is comparable with that of computerized dynamic posturography measurement using a purpose-built device (MPS-3102; Balance Master, NeuroCom, Clackamas, USA; ICC $r = 0.67–0.7$). In contrast, results obtained using the instrumented treadmill and purpose-built device differed substantially.

The purpose of this study was to examine if an instrumented treadmill in combination with an optical motion capture system can be used to assess perturbed postural balance. This study concentrated on reactive postural balance with a proactive component included in the assessment as the direction of perturbation was known and the perturbation could be anticipated although exact timing was unknown. COP trajectory in the antero-posterior direction was used as the outcome measure. Following a previous study utilizing a purpose-built perturbed postural balance assessment device (Piirainen et al., 2013), we tested the balance with four perturbation protocols (slow and fast, forward and backward directions). From a theoretical point of view, it is of interest to include perturbations in both directions as the balance maintenance requires the use of different muscle groups when recovering from the perturbation in different directions and it may involve different balance strategies such as ankle strategy and hip strategy.

To this end, we performed a between-days test–retest study that allowed us to evaluate the reliability of the balance assessment as well as short-term learning effects over 5 days. We defined changes that occurred between days as short-term learning, whereas acute changes that occurred within a day were considered as habituation. We hypothesized that: (1) the instrumented treadmill can be used to induce perturbations of the base of support precisely, accurately and reliably, (2) the parameters calculated from the COP trajectory to quantify balance performance show similar reliability as previously reported for purpose-built devices, and (3) learning is observed in balance performance over 5 days. If the study supports the hypotheses, instrumented treadmills in combination with an optical motion capture system could provide a tool to analyze perturbed postural balance in research settings. Furthermore, this setup can be potentially used for postural balance training with continuous monitoring of the progression of balance performance.

METHODS

Participants and Protocol

Eight people without current musculoskeletal pain or physical limitations volunteered for the study (two females, six males, aged between 24 and 43 years, body mass 64.2–105.6 kg). They were informed about the study, testing protocols, and the use of data according to the institutional guidelines.

Testing was conducted for 5 consecutive days with an identical test setup each day. Four test protocols were commenced with a single protocol and included 10 perturbations with a given direction (backward or forward) and speed (slow or fast) at random intervals. Each day, the protocols were performed in the same order: slow backward, fast backward, slow forward, and fast forward. Participants were made aware of the perturbation direction and speed before commencing the test. Each protocol was performed twice. The first performance was considered as habituation, and the results were calculated from the second performance. Habituation was performed to accustom the participants to the protocol and to mitigate the potential order effect. Stepping response was not allowed, and the habituation trial successfully removed the need for taking a step to maintain balance, which was occasionally observed in the habituation trial but there was none in subsequent trials. Habituation was included in the test setup each day to keep the test setup the same for the examination of the reliability between days. The perturbation intervals were different for habituation and the measurement protocol, but the same across participants and days. The slow protocols lasted, in total, 48–49 s depending on the direction, and the fast protocols lasted 52–53 s. The delay between perturbations was 4.5 ± 0.9 s (mean ± SD). The delay was confirmed to be sufficient for recovering a stable balance between the perturbations. During the measurements, we confirmed that COP recovered close to the initial location and that the COP trajectory was stable before a new perturbation was delivered.

Initially, the participants stood barefoot on a split-belt instrumented treadmill (M-gait, Motek Medical, Houten, The Netherlands) with feet pointing forward with a standardized width of 30 cm (center to center distance) both feet on different force plates/belts, hands on the sides of the body, and gaze fixed to a point at the level of the eyes on the opposing wall. A previously published (base of support) movement pattern (Piirainen et al., 2013) was implemented using the D-Flow software (Motek Medical, Houten, The Netherlands) controlling the treadmill. The test setup comprised four protocols. A single protocol included only slow or fast perturbations in one direction. The software allows setting the target velocity for the belt, the maximal acceleration that the motor can utilize, and the duration that the belt is driven with the target velocity. In slow perturbations, the belt was set to move with a maximal acceleration (and deceleration) of 0.3 m/s^2 targeting 0.15 m/s belt velocity for 0.5 s resulting in a ramp-like velocity profile without plateau (**Figure 1**). The resulting calculated ideal belt movement was 75 mm. For fast perturbations, the target speed was set to 0.25 m/s for 0.5 s, while the maximal acceleration was limited to 4 m/s^2 and then decelerated to a full stop with the same acceleration. The resulting calculated ideal belt movement was 125 mm. Unlike in the study of Piirainen et al. (2013) in which electromechanical cylinders could move the force plate forward and backward, the opposite directions of perturbations in this study were enabled by changing the direction the subjects were facing, i.e., the belt only moved in one direction. This is a limitation of the system that can be overcome by updating the software and may not apply to all corresponding systems.

Data Analysis

To account for the relative movement of the belt (base of support) and the treadmill structure (force plate), the movement of the belt was recorded by tracking three reflective markers placed on the treadmill using an optical motion capture system (100 Hz, Vicon Vero, Vicon Motion Systems Ltd., Oxford, United Kingdom), while the COP was measured with the instrumented treadmill (1,000 Hz). Measurement of the belt movement allowed us to express the COP trajectory relative to the base of support similarly as in the case where the force plate would be moving. The optical motion capture system was also used to measure the location of four markers on each foot (big toe, heel, and medial and lateral malleolus). The malleolus markers were used to express the COP location relative to the ankle joint center. This information can be used to evaluate potential anticipation of the coming perturbations by shifting the COP location toward the toes in case of forward perturbation or toward the heel in case of backward perturbation. Heel and toe markers can be used to express the COP trajectory relative to foot length, but here we

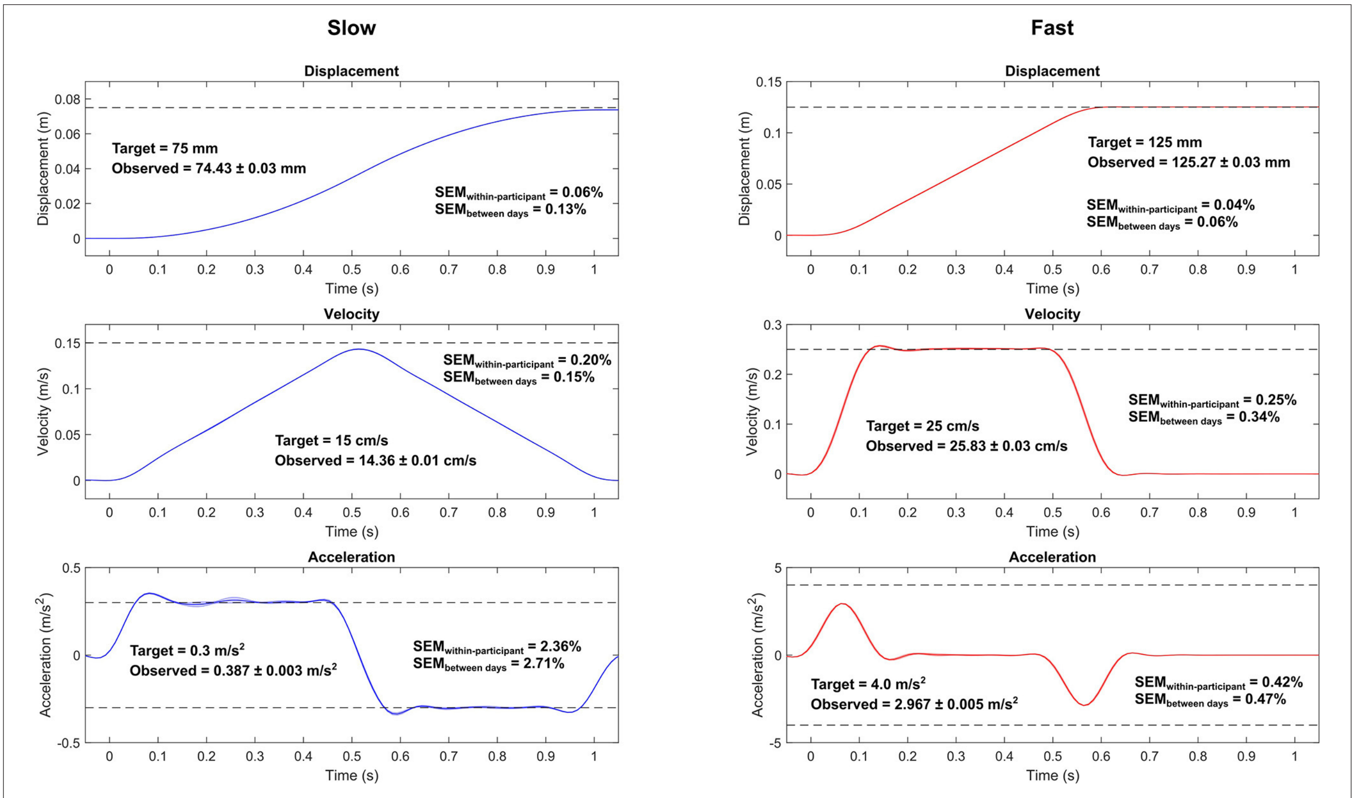

FIGURE 1 | Measured belt movement and its reliability metrics for the slow (left) and fast (right) protocols. The curves represent the mean and between-participant standard deviation (shaded area; note that the standard deviations are extremely small). Only the backward perturbations are shown, since the patterns are identical in both directions. Dashed vertical lines show the target values of the peak displacement, velocity, and acceleration based on programmed control signals of the belt. Time zero indicates the identified start of the perturbation. Note that the time scales are the same for both protocols, but the vertical scales differ. The observed values are presented as the mean and within-participant standard deviation of the peak values. $SEM_{within-participant}$, within-participant standard error of measurement; $SEM_{between\ days}$, between-day standard error of measurement.

chose to report the results in absolute units consistent with a previous study (Piirainen et al., 2013).

COP and marker data were filtered using a fourth-order zero-lag 5 Hz low-pass Butterworth filter, and COP data were interpolated to 100 Hz to match the sampling frequency of motion capture data. In the analysis, we only considered the anteroposterior direction of the COP trajectory. The displacement of the COP relative to the base of support was calculated by subtracting the COP displacement from the belt displacement. The onset and the end of the perturbations were detected from the marker-based belt velocity profile using 3 and 2 cm/s thresholds for onset and end detections, respectively, followed by constant time shifts for locating the actual onset and end that depended on the protocol (slow or fast). Three outcome measures reported by Piirainen et al. (2013) were quantified from the COP trajectories: peak displacement, time to peak displacement, and recovery displacement, which allowed the results to be compared with those measured using a purpose-built movable force plate system. Peak displacement and time to peak displacement were defined as the peak of the COP trajectory relative to position at the instance of perturbation onset, and the time to the peak, respectively. Recovery displacement was defined as the peak-to-peak displacement of the COP trajectory during a 500-ms time window after the end of belt movement. The 500-ms recovery period has been used previously by Piirainen et al. (2013) and Chien and Hsu (2018), with the authors of the latter study justifying the selection by averaged active response time observed in previous studies. Additionally, we calculated the COP location relative to the ankle joint center (the midpoint between medial and lateral malleolus) at the instance of perturbation onset (initial COP location) to evaluate potential anticipation behavior. The extraction of the outcomes from the COP data was done with two different approaches: from ensemble average trajectory and individual trajectories. For the extraction of the outcomes from the ensemble average COP trajectory, the COP trajectory was cut into sections defined by the above-mentioned onset and end detention, and ensemble average COP trajectory was calculated for the left and right legs to improve the signal-to-noise ratio of the data. The trajectories were set to zero at the instance of perturbation onset and, finally, the mean trajectory of the left and right leg COP trajectories was calculated. Then, the three outcome measures were determined from this average trajectory (**Figure 2**). Additionally, extraction of

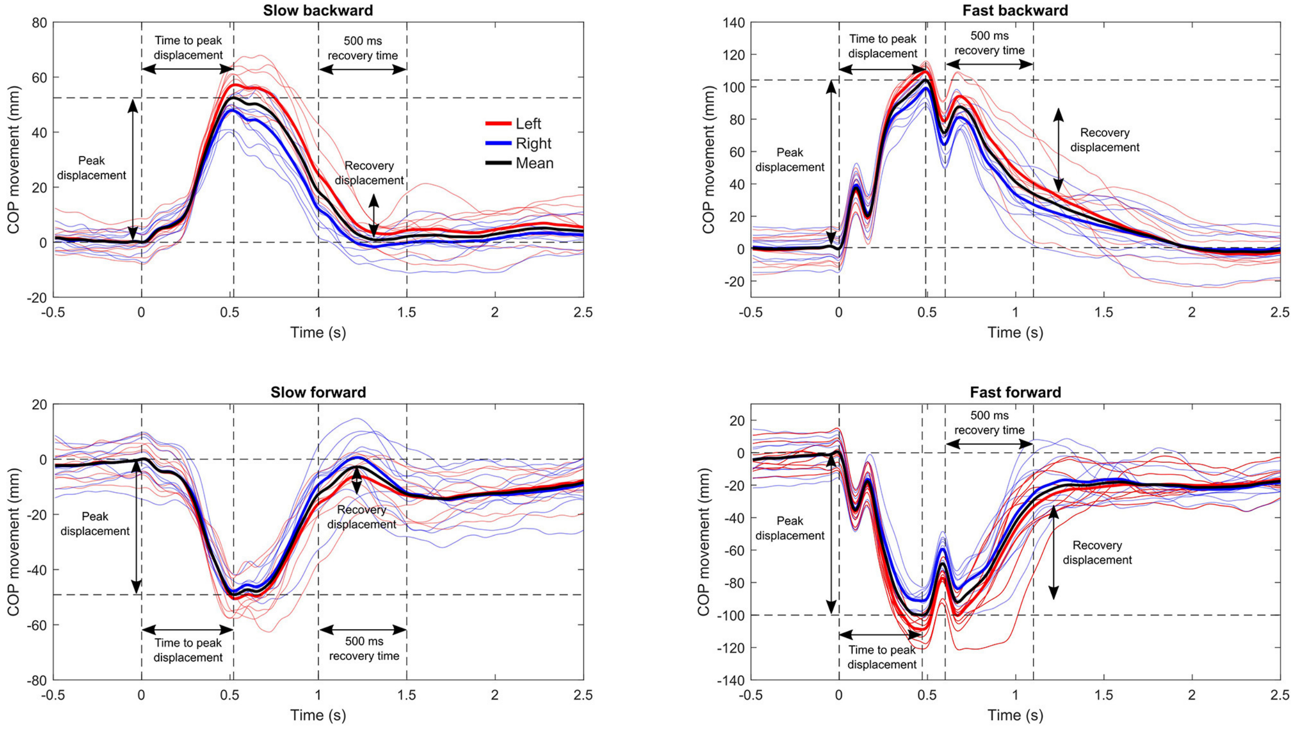

FIGURE 2 | Example of the center of pressure (COP) trajectories relative to the belt during the four perturbation protocols from a single participant on a single session. The thin lines represent the response to each perturbation, and the thick lines are the ensemble average trajectory for each leg. The outcome measures were calculated from the ensemble average trajectory of both legs (black) and alternatively from each response to perturbation separately (average of right and left legs) with subsequent averaging of the results to obtain the final outcome. The horizontal dashed lines denote the peak displacement and the zero displacement level, and the vertical lines mark the instance of perturbation onset, time at peak displacement, and 500-ms post peak displacement for the calculation of recovery displacement.

the outcome measures was performed from each response to the perturbations separately (average of right and left legs), and the final outcome was calculated as the average of the outcomes from individual trajectories. Body mass was calculated by dividing the mean vertical force recorded during the trial by 9.81 m/s.

Statistical Analysis

The reliability of the postural balance outcomes calculated from the COP trajectory was evaluated using intraclass correlation (ICC), a measure of relative reliability and standard error of measurement (SEM), a measure of absolute reliability (Weir, 2005). For ICC calculation, we used a single rater two-way random-effect model for absolute agreement (ICC 2.1). ICC values were interpreted according to Koo and Li (2016) with the following cutoff points: < 0.5 poor, 0.5–0.75 moderate, 0.75–0.9 good, and >0.9 excellent reliability. SEM was calculated by repeated measured analysis of variance (ANOVA) that partitions the observed variability to the variability arising from between-days and within-day effects. The within-day variability is further partitioned into between participants and error variability. The error variability is an estimate of the variability within-day that is not accounted for by between-participant differences and, therefore, estimates the random variability within-participant. By taking the square root of the mean square within-day error, we estimated the typical measurement error (Weir, 2005) and later referred to this as $SEM_{within-participant}$. For reliability analysis of the belt movement, in addition to the $SEM_{within-participant}$, we reported the estimate of typical between days difference (standard deviation between days), which was calculated by taking the square root of the between days mean squares. We referred to this later as $SEM_{betweendays}$. The SEM values are presented as the percentage of the mean and, additionally, in original units in the supplement material for the reliability of postural balance outcomes. To investigate if the body mass of the participant affected the movement of the belt, we calculated Pearson correlation coefficients of belt peak displacement, velocity, and acceleration with body mass. In this analysis, we utilized the data from all days and both perturbation directions within a speed resulting in 80 observations (forward and backward protocols, 5 days, and eight participants) for each analysis. Pooling the data was justified by the assumption that the influence of body mass is much larger than any potential effect of measurement day or standing direction of the participant; thus, each trial could be considered as an independent observation. In case of a significant correlation, linear regression analysis was performed to determine the magnitude of the effect that the body mass had on the belt movement. Learning effects were assessed using repeated-measures ANOVA comparing the results obtained on different days followed by an additional analysis of linear and quadratic trends to assess systematic patterns in the values observed on different days. A linear trend was considered to model a situation in which learning is occurring throughout the 5 days, whereas a quadratic trend was considered to model learning with a ceiling effect during the 5 days. Reliability and learning effect analyses were performed in the IBM SPSS Statistics software (version 27, SPSS Inc., IBM, Armonk, NY, United States), and the correlation and regression analysis between belt peak displacement, velocity, and acceleration and participant body mass (derived from the force data) was performed in MATLAB (R2019b, The MathWorks, Inc., Natick, MA, United States). The statistical significance was set at $p < 0.05$.

RESULTS

During the data analysis, we noticed that the first perturbation in a set was systematically different from the rest of the perturbations in the set, showing higher acceleration, especially in the slow protocol (**Supplementary Figure 1**). Hence, we removed the first perturbation from the analyses and calculated all outcome measures based on the remaining nine perturbations in the set. The movement of the belt was highly repeatable and accurately followed the control signal. In both the slow and fast protocols, the displacement of the belt was always less than half a millimeter from the target value (**Figure 1**). Belt peak velocities showed < 1 cm/s error. The largest deviation from the target values was observed in peak acceleration in the fast protocol in which the belt reached about 75% of the target value. In the slow protocol, peak accelerations overshot the target by an average of 29%. The largest within-participant and between-day standard errors in the belt movement were observed in the peak accelerations of the slow protocol in which these errors were <3% of the mean. All the other SEMs were <1% of the mean.

We did not observe statistically significant correlations between participant body mass and the measured peak displacements or velocities, but a weak correlation was observed between body mass and peak acceleration (slow protocol $r = 0.262$, $p = 0.019$; fast protocol $r = 0.298$, $p = 0.007$, **Supplementary Figure 2**). Regression analysis indicated that with each 1 kg increase in body mass, the peak acceleration would increase by 0.1% in both the slow and fast protocols.

Visual observations indicated that COP trajectories show repeatable patterns between days (**Figures 3**, **4**). One participant (participant 5) showed clearly different COP movement patterns in both slow protocols for day 5 compared with other days. This probably reflects an altered balance maintenance strategy for day 5. We excluded the participant from the reliability analyses of the slow protocols, as these vastly different results would have inflated the reliability metrics (**Table 1**). This result probably reflects an altered balance maintenance strategy. The reliability results using the whole dataset are provided in **Supplementary Table 1**. Overall, the reliability results were not markedly influenced by the analysis methods, i.e. if the outcomes were calculated from the ensemble average COP trajectory or individual trajectories and subsequently averaged. The absolute reliability (SEM) of time to peak displacement and recovery displacement was better when the perturbation direction was backward compared to forward. Based on ICC values the reliability in different variables and perturbation directions and speeds ranged from poor to good.

Over the five consecutive testing days, time to peak displacement showed a linearly increasing trend in the slow backward ($p = 0.033$) and a linearly decreasing trend in the fast backward ($p = 0.011$) protocols (**Figure 5**). Peak displacement showed a linearly decreasing trend in the slow forward protocol ($p = 0.003$) and a quadratic trend in the fast backward protocol ($p = 0.027$) with an initial decrease as a function of time. Additionally, peak displacement from day 1 significantly differed from day 5 in the slow forward protocol ($p = 0.043$). The COP was located approximately 4–5 cm anterior from the ankle joint center. The COP location was systematically approximately 1 cm more from the anterior in the forward perturbation protocols, which is related to the fact that the participants were aware of the perturbation direction and anticipated it by moving the COP location anteriorly in case of forward or posterior perturbation in case of backward perturbation to provide a possibility for a larger movement amplitude of the COP. COP location relative to the ankle joint at the instance of perturbation onset showed a quadratically decreasing trend (i.e., COP was closer to the ankle joint on later days) in the slow forward protocol ($p = 0.021$). Most of the decrease occurred between days 1 and 2.

DISCUSSION

The purpose of this study was to examine the capability of an instrumented treadmill for testing perturbed standing postural balance. We hypothesized that the instrumented treadmill is precise and accurate in delivering intended perturbations and that the measured outcomes show comparable values with previously reported ones using purpose-built devices and with comparable test–retest reliability. Additionally, we examined potential short-term learning effects that are important to acknowledge when designing longitudinal studies and provided indications if the system could also be useful as a balanced training method. The results indicate that the treadmill can repeatedly deliver perturbations with low between-session and between-participant variations in displacement, speed, and acceleration. Postural balance evaluated with the treadmill in combination with a motion capture system (**Figure 5**) showed comparable results with Piirainen et al. (2013) using a purpose-built device (numeric values not given, data provided as a bar chart). The only marked difference between the studies was the recovery displacement of the fast perturbation protocol in which the results of this study were about half of those observed in the study of Piirainen et al. (2013). The observed test–retest reliability was also on par with the previous report using purpose-built devices (Yuntao et al., 2017). Finally, the analysis provided evidence for short-term learning effects on multiple outcome measures. In some variables, the results seemed to plateau within 5 days; whereas in others, continued learning effects were observed throughout 5 days. Overall, the

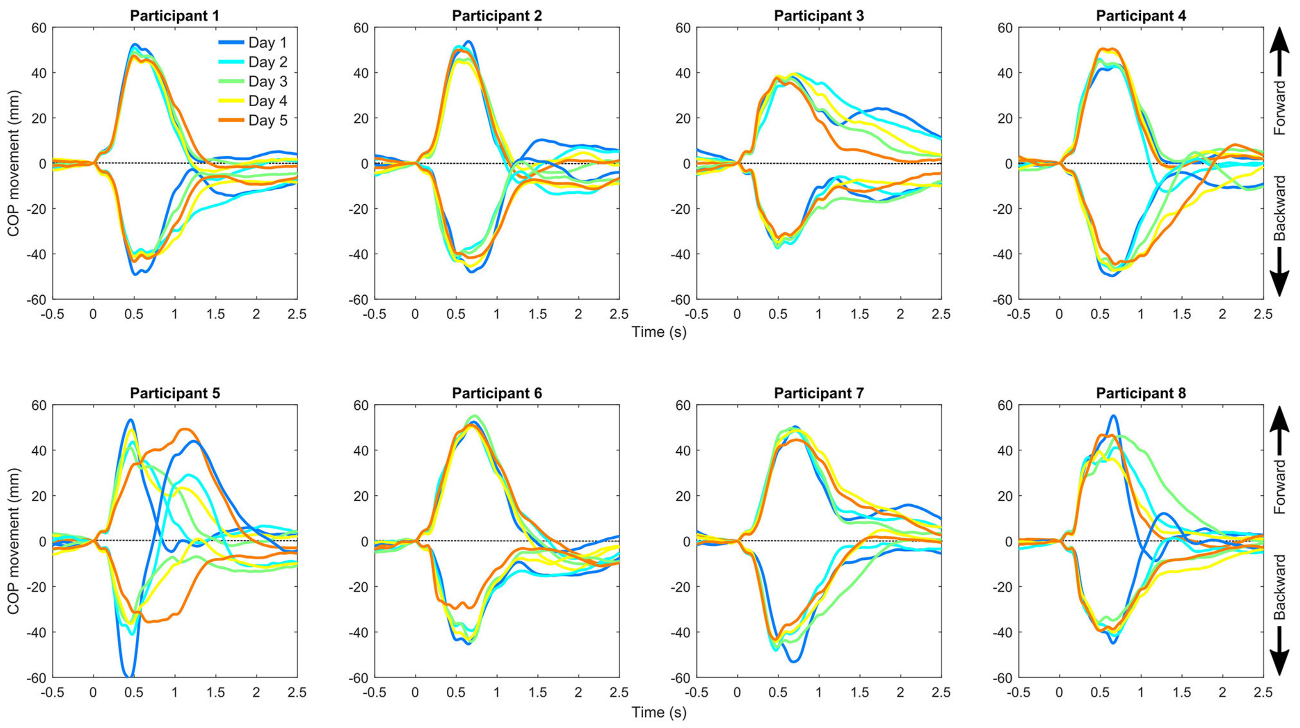

FIGURE 3 | Center of pressure (COP) movement relative to the belt in the slow backward and forward perturbation protocols separately for each participant and measurement day. The positive direction of the center of pressure movement is forward (direction of gaze) and occurs in response to backward perturbation of the belt.

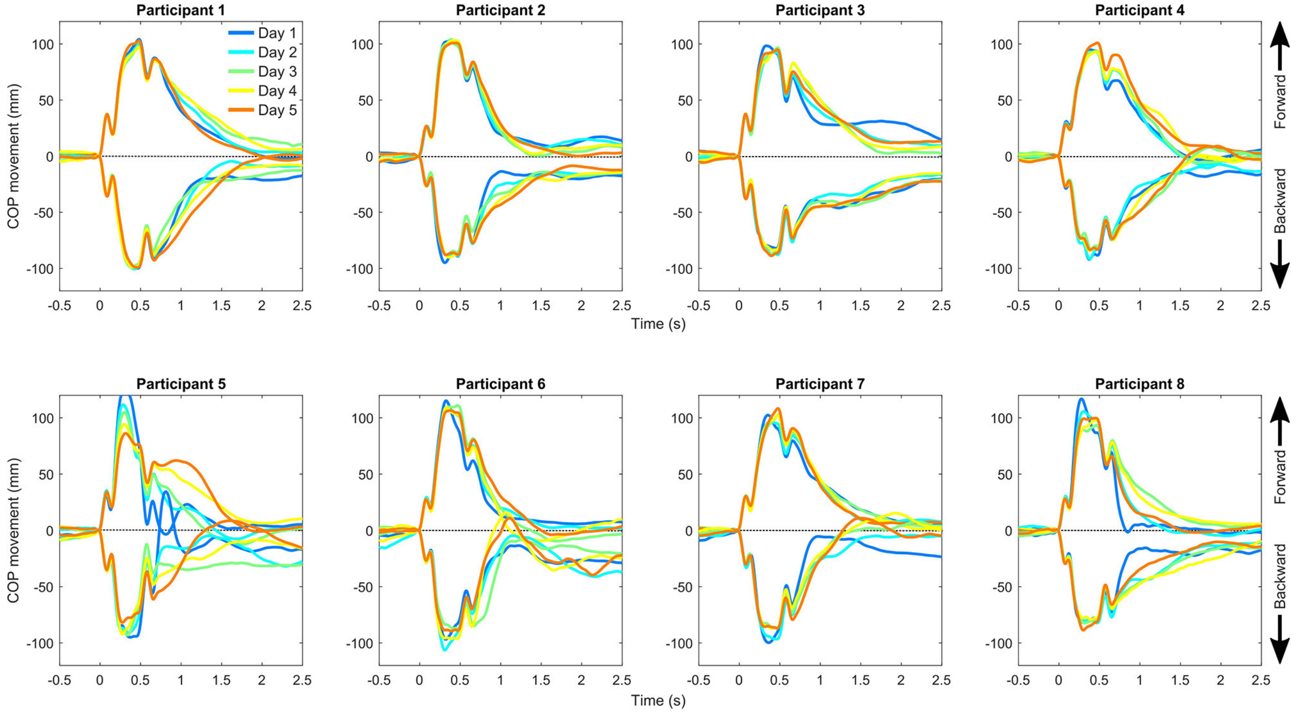

FIGURE 4 | Center of pressure (COP) movement relative to the belt in the fast backward and forward perturbation protocols separately for each participant and measurement day. The positive direction of the center of pressure movement is forward (direction of gaze) and occurs in response to backward perturbation of the belt.

TABLE 1 | Test–retest reliability of the selected outcome measures describing perturbed postural balance performance.

		Slow backward	Slow forward	Fast backward	Fast forward
Based on ensemble average trajectory					
Peak displacement (mm)	ICC	0.598 (0.277–0.893)	0.571 (0.238–0.882)	0.252 (0.015–0.671)	0.547 (0.247–0.854)
	SEM	3.30	2.69	6.27	4.13
	SEM%	7.00	6.32	6.16	4.56
Time to peak displacement (s)	ICC	0.503 (0.187–0.854)	0.506 (0.183–0.857)	0.549 (0.228–0.856)	0.549 (0.228–0.856)
	SEM	0.06	0.07	0.04	0.05
	SEM%	9.93	11.78	8.90	12.66
Recovery displacement (mm)	ICC	0.719 (0.430–0.932)	0.669 (0.346–0.918)	0.738 (0.468–0.929)	0.445 (0.153–0.803)
	SEM	3.70	4.64	7.65	13.31
	SEM%	18.14	27.65	15.83	28.22
Based on individual trajectories					
Peak displacement (mm)	ICC	0.638 (0.326–0.907)	0.574 (0.235–0.884)	0.349 (0.083–0.743)	0.513 (0.215–0.838)
	SEM	2.73	2.41	5.53	3.97
	SEM%	5.56	5.41	5.32	4.28
Time to peak displacement (s)	ICC	0.388 (0.069–0.804)	0.156 (−0.098 to 0.653)	0.562 (0.255–0.861)	0.550 (0.237–0.858)
	SEM	0.06	0.08	0.03	0.04
	SEM%	9.82	12.65	7.55	8.37
Recovery displacement (mm)	ICC	0.752 (0.478–0.942)	0.658 (0.336–0.915)	0.680 (0.389–0.909)	0.438 (0.147–0.800)
	SEM	3.29	4.47	7.26	13.23
	SEM%	14.96	23.65	14.34	25.91

Outcome variables are calculated from ensemble average COP trajectory and alternatively from each individual trajectories.
ICC, intraclass correlation and 95% confidence interval; SEM, standard error of measurement expressed as percentage of the mean. SEM reported here refers to the $SEM_{within-participant}$ explained in the statistical analysis chapter. Data from one participant on one day was excluded from the slow protocols.

study supports the usability of an instrumented treadmill in combination with a motion capture system for testing perturbed postural balance.

Accuracy and Reliability of Treadmill Belt Movement

One of the aims of the study was to quantify the precision, accuracy and reliability of the belt movement for delivering perturbations. The belt movement was highly repeatable and accurately replicated the target velocity. In the fast protocol, the belt reached only about 75% of the maximal allowed acceleration. The reason for this was the limit in the rate of rising of the acceleration. However, regardless of not reaching the maximal allowed acceleration set for this protocol, the peak accelerations were still highly repeatable both within (between days) and between-participants with a typical error of $< 0.5\%$ of mean in the fast protocol. Thus, the deviation from the intended acceleration does not invalidate comparisons between sessions and participants. In both slow and fast perturbations, the mean belt displacement during perturbation was within < 1 mm of the calculated target. This is noteworthy since the controls for the belt movement only included target velocity maximal acceleration/deceleration to be used.

We also examined the impact of body mass on treadmill belt movement using correlation analysis and found that body mass did not significantly correlate with the displacement amplitude or peak velocity of the belt, but a significant correlation was observed between body mass and peak acceleration. The observed correlation may be related to the control system of the treadmill and the increased demand for the adjustments of motor torque due to added body mass. Regression analysis showed that a 1 kg increment in body mass had a 0.1% effect on belt peak acceleration. Thus, for example, a 50 kg between-subject variation on body mass is expected to have a 5% effect on belt peak acceleration. The effect is not negligible but is comparable with the within-participant typical error in peak acceleration in the slow protocol. Thus, we consider that the effect that body mass has on belt movement does not invalidate between-participant comparisons. The results regarding belt movement accuracy ($<$ 1% deviation in total displacement from target value) are generally in line with those of a previous report (2–5%) (Crenshaw et al., 2019), although we reported a slightly better accuracy except for peak acceleration in both slow and fast protocols (between 26 and 29% in this study versus $\leq$5% in that of Crenshaw). In addition, Crenshaw et al. (2019) found an effect of body mass on belt displacement and velocity accuracy but not on acceleration. This difference may be due to the use of different treadmills.

Interestingly, we noticed that the belt movement in the first perturbation systematically differed from the rest of the perturbations in the set (**Supplementary Figure 1**). The reason for this behavior was that a brake is released simultaneously with the start of the first movement of the belt resulting in a slower start of the perturbation followed by abrupt acceleration. This can be avoided by adding a period of zero velocity at the beginning of the control script that releases the brake.

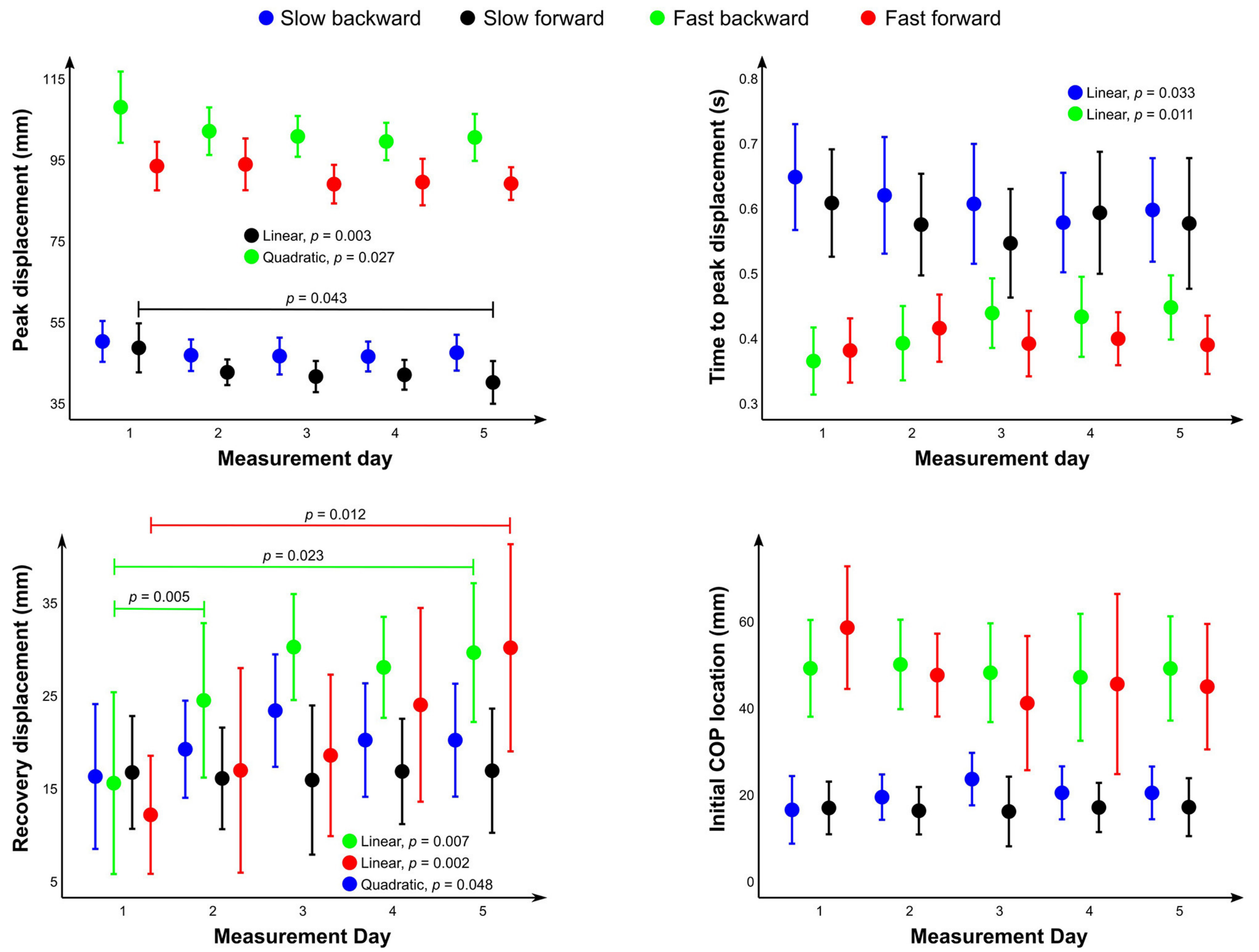

FIGURE 5 | Mean values of the different outcome measures and center of pressure location at the instance of perturbation onset relative to the ankle joint. Whiskers represent the 95% confidence interval of the mean. Statistically significant differences between days and statistically significant linear or quadratic trends across the days are shown.

Reliability of the Perturbed Postural Balance Outcome Measures

Test–retest reliability of the selected outcome measures of balance performance was mostly moderate and not markedly affected by the calculation of the outcomes from the ensemble average COP trajectory or individual trajectories (**Table 1**). In time to peak displacement, the relative reliability (ICC) was better when calculated from the ensemble average COP trajectory compared with the calculation from individual trajectories, but the calculation type did not affect absolute reliability (SEM). Poor reliability based on ICC was observed in peak displacement of the fast backward protocol and recovery displacement of the fast forward protocol when outcomes were calculated from the ensemble average COP trajectory. Poor reliability was observed in time to peak displacement of the slow backward and forward protocols, peak displacement of the fast backward protocol, and recovery displacement of the fast forward protocol when outcomes were calculated per trajectory. The poor reliability is partly due to observed learning effects, as we used the absolute agreement definition of the ICC calculation as opposed to consistency definition. The low end of the ICC values reported in this study (ICC 0.16) is worse than that which has been reported in previous investigations of perturbed postural balance assessments, which have reported ICCs ranging from 0.61 to 0.96 (Yuntao et al., 2017; Crenshaw et al., 2019). However, it should be noted that the test protocols, devices and outcome measures differ between the studies.

We want to point that visually inspecting the shape of the COP trajectories showed repeatable patterns between repeated perturbation within a session and between days (**Figures 3**, **4**). It seems that there is a COP trajectory "fingerprint" that is somewhat unique to the participant, although the reliability of the selected outcome measures was only modest. The finding also indicates that postural balance correction strategies are relatively stable within a participant. Hence, the modest reliability is probably not related to the instrumentation but has issues with the used outcome metrics. It could be of interest to further investigate individual COP trajectory shapes in future studies and identify outcome metrics that better capture the individual features of COP responses.

Between-Day Differences and Learning Effect

We quantified potential learning effects by investigating between-day differences and between-day linear and quadratic trends. Statistically significant differences in the postural balance outcome measured were detected only for peak displacement in the slow forward protocol in which the observed peak displacement was larger on day 1 compared with day 5. Statistically significant trends were observed for peak displacement (slow forward and fast backward) and time to peak displacement (slow and fast backward). Both peak displacements and times to peak displacement decreased with time in the slow backward protocol. However, in the fast backward protocol, time to peak displacement increased with time. These are probably a result of short-term learning or habituation. The increase in time to peak displacement in the fast protocol was coupled with a decrease in peak displacement. The result may be due to the participants learning to start the balance-correcting muscle activity earlier, which slows down the anterior COP movement velocity and results in the observed later occurrence of peak displacement.

Interestingly, the perturbation velocity, which was also known by the participants, did not affect the COP location in the backward perturbation condition. In the forward perturbation condition, the COP location was more anterior with the fast perturbation speed. No significant between-day differences were observed in the initial COP location, but there was a significant linearly decreasing trend in the slow forward protocol, which probably indicates habituation to the perturbation protocol allowing the participant to stand with COP closer to the ankle joint center while maintaining balance.

The trends observed in peak displacements and times to peak displacement may indicate short-term learning effects and, therefore, support the use of instrumented treadmills as a potential postural balance training apparatus. However, in this study, perturbation intervals were randomized within the protocol, but the protocol was the same between the days. Hence, it is not clear if the improvements reflect memorizing the protocol or learning in balance control. Earlier studies have shown the importance of task-specific training. Training methods that influence postural balance might be more effective than basic and general exercises (Hrysomallis, 2011; Gerards et al., 2017). Perturbations of the base of support can provide task-specific training and have been named perturbation-based balance training (PBT). The goal of PBT is to improve reactive balance control after destabilizing perturbations (Gerards et al., 2017). PBT performed during walking has been shown to improve perturbed postural balance (Chien and Hsu, 2018). In addition, based on a meta-analysis, PBT seems to be effective for reducing fall risk among older adults and individuals with Parkinson's disease (Mansfield et al., 2015). Future studies could investigate if PBT performed during locomotion is more effective in reducing fall risk than PBT during standing as performed here.

Comparison With Previous Studies Utilizing Purpose-Built Devices

The perturbation protocols used in this study were based on a previous study by Piirainen et al. (2013). COP peak displacements, times to peak displacement, and recovery displacement showed comparable results with the group of young adults in that study. Moreover, the peak displacements observed in the current study were in line with the study by Walker et al. (2020) which utilized a protocol closely resembling the one used here in the fast condition. The finding suggests treadmill-based perturbed postural balance assessment has good concurrent validity compared with the test performed using a purpose-built device consisting of a commercial force plate driven by electromechanical cylinders. This finding suggests that instrumented treadmills can be utilized for perturbed balance assessments despite the lower accuracy of the COP measured due to mechanical vibrations transmitted to the force sensors and concern regarding the accuracy of movement due to, e.g., belt slackness.

LIMITATIONS

The following limitations related to this study should be acknowledged. First, the small sample size limited the ability of the authors to detect short-term learning effects. With a larger sample size, we could have most probably detected learning effects from more of the parameters. A larger sample size could have also resulted in higher confidence for reliability estimates apparent in reduced confidence intervals. Second, we examined only young and healthy individuals. Thus, reliability estimates for balance outcome measures are not generalizable to other populations, but the technical suitability of an instrumented treadmill for perturbed postural balance measurement is not expected to depend on the population of interest. Third, in this study, the treadmill belt only moved in one direction, which allowed the participants to anticipate perturbation even when the time-lag between perturbations is randomized. However, even when performed with a uni-directional treadmill, the results of the study were in accordance with previous investigations using a multi-directional movable force plate (Piirainen et al., 2013). Also, on average, the difference in COP location relative to the ankle joint center at the instance of perturbation onset was only 1.2 cm. The difference was systematic, so we can conclude that knowing perturbation direction causes anticipation, but the magnitude of anticipation was only around 5% of the total foot length, which is about 10–20% of COP trajectory length in response to the perturbation.

Suggestions for Future Studies Utilizing Instrumented Treadmills for Perturbed Postural Balance Assessment

The test described in this study can be easily supplemented with measurement of joint kinematics and kinetics (inverse dynamics-based), as the necessary equipment for those measurements are force plates and a motion capture system. Also, adding electromyography measurements, in addition to joint kinematics and kinetics, would allow for a comprehensive assessment of balance maintenance mechanisms. Muscle activity could give more information about the motor control of postural balance by quantification of factors such as anticipatory muscle activity, latency or reaction time, reflective activity, and muscle co-activation. Furthermore, previous studies have coupled measurement with percutaneous electrical stimulation of peripheral nerves to assess H-reflexes during perturbations (Piirainen et al., 2013; Miranda et al., 2019). This measurement can be used to assess spinal sensitivity during postural balance maintenance. When investigating participants with unilateral musculoskeletal conditions or neurological conditions affecting the body asymmetrically, it may be of interest to consider COP trajectories separately for both legs. It may be also of interest to investigate medio-lateral COP movement in response to the perturbations. In future studies, it is advisable to mix directions and speeds of perturbations within a trial when the hardware allows this. This would allow one to include more than 10 perturbations in a trial. In this study, we were able to detect short-term learning effects during the 5 days in some parameters and also observed indications of instantaneous habituation between the training and measurement trials, and within the habituation trial (see details on the **Supplementary Material**). Future studies should utilize sufficient habituation period to prevent habituation effect biasing the results. At minimum, it needs to be ensured that all experimental groups compared have received equal habituation to the testing procedures. Based on the results of this study, we are not able to give a recommendation on the required amount of habituation, and this aspect should be more thoroughly investigated in the future. We noticed that COP peak displacements occurred around the instance of belt deceleration. This may be because of corrective angular impulse relative to the body center of mass that the base of support deceleration creates. Future studies should investigate protocols in which the velocity plateau is longer and, thus, the base of support deceleration would not help in balance maintenance. Finally, we suggest that COP accuracy during belt movement should be investigated if the information is not available for the particular device. A previous study showed that COP accuracy with an instrumented treadmill depended on belt speed and mass applied on the belt (Fortune et al., 2017).

We noticed that the first perturbation in the trial provides an acceleration profile different from the rest of the perturbations in a trial. The reason for this behavior was that a brake is released simultaneously with the first input to the treadmill. The issue can be resolved in future studies by implementing a short section with zero velocity at the beginning of the protocol.

CONCLUSIONS

The results indicate that an instrumented treadmill combined with an optical motion capture system can be utilized for testing perturbed postural balance similarly as has been previously done using purpose-built motorized force plates. This opens up possibilities for research laboratories and rehabilitation centers with access to such equipment for perturbed balance assessments. However, it should be noted that the results may not be generalized to equipment from different manufacturers. The observed learning effects suggest that the system and protocols can be potentially used for training to improve postural balance, but further research is needed to confirm this. The data presented can be used to inform future studies that will utilize instrumented treadmills for perturbed postural balance assessments regarding required sample sizes and selection of protocols.

DATA AVAILABILITY STATEMENT

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

ETHICS STATEMENT

Ethical review and approval was not required for the study on human participants in accordance with the local legislation and institutional requirements. Written informed consent for participation was not required for this study in accordance with the national legislation and the institutional requirements.

AUTHOR CONTRIBUTIONS

KL, VH, MV, HT, PV, and LS designed the study. KL, JL, PV, and LS conducted the experiments. JL and LS wrote the analysis code. KL, JL, and LS analyzed and interpreted the data. KL and LS wrote the initial manuscript draft. JL, VH, PV, MV, and PK revised the manuscript. HT and PK contributed to funding acquisition and project administration. MV, PV, LS, PK, and HT contributed to supervision. KL, PV, LS, HT, and PK contributed to the development of the infrastructure enabling this study. All the authors read and approved the final manuscript.

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SUPPLEMENTARY MATERIAL

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Biomechanical Mechanisms of Improved Balance Recovery to Repeated Backward Slips Simulated by Treadmill Belt Accelerations in Young and Older Adults

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Aim: Exposure to repeated gait perturbations improves the balance of older adults (OAs) and decreases their risks of falling, but little is known about the underpinning mechanical adjustments. We aimed to quantify the changing temporo-spatial and kinetic characteristics of balance recovery following repeated backward slips to better understand the mechanical adjustments responsible for improved balance.

Methods: We exposed 17 young adults (YAs) (25.2 ± 3.7 years) and 17 OAs (62.4 ± 6.6 years) to 10 backward slips simulated on an instrumented treadmill by unilateral backward belt accelerations. We measured the balance of the participants (margin of stability: MoS), balance recovery (n_{steps}: number of steps necessary to return to a steady gait for at least three consecutive steps), temporo-spatial (step length), and kinetics [ground reaction force (GRF) angle, lower limb joint moments] for 15 steps following each slip. The results were compared with baseline.

Results: Participants in both groups improved their MoS and n_{steps} with repeated exposure to the slips, but no significant effect of age was detected. During the perturbed step, the GRF vector was directed more posteriorly during mid-stance and more anteriorly during push-off than baseline, which resulted in a longer step. These adjustments were maintained from the first (Slip01) to the last (Slip10) slip, and by Slip10 were correlated with better balance (MoS) on the second recovery step. During the first recovery step following Slip01, participants developed lower plantarflexor and larger knee extensor moments whilst taking a shorter step, these adjustments were correlated with poorer balance and were not maintained with repeated slips. Joint moments and step length of the first recovery step returned to normal levels by Slip10.

Conclusion: Young adults and OAs improved their balance with repeated slips. The adjustments that were positively correlated with balance (changes in step length, GRF angle) were maintained whilst those that were not (changes in joint moments) were discarded. All the responses observed in Slip10 were observed in Slip01. The observed balance improvements were achieved by refining the initial strategy rather than by developing a new one. The underlying mechanics were correlated with step length of the first recovery steps, which was associated with balance and should be monitored in fall prevention interventions.

Keywords: gait perturbation, balance, recovery mechanisms, age, kinetics, temporo-spatial variables, slip, step length

INTRODUCTION

Older adults are at greater risk of falling than young adults (YAs), and these falls can result in life-threatening injuries (Spaniolas et al., 2010). Especially for community-dwelling older adults (OAs), most of the falls are triggered by trips or slips (Berg et al., 1997). Although some inconsistencies in the definitions of trips and slips exist, trips can be described as gait perturbations resulting from the sudden arrest of the swing foot that triggers a forward loss of balance, and slips as perturbations to balance resulting from sliding of the stance foot over the ground. Typically, slips arise either when the stance foot slides forward mainly shortly after heel strike (in this article: forward slips), or when the stance foot slides backward typically from mid to end stance (in this article: backward slips). Historically, backward slips have been considered less dangerous, as an individual has the opportunity to quickly regain balance with the contralateral foot. However, when participants walked on a contaminated oily surface, backward slips ($n = 20$) were observed up to 2.5 times more often than forward slips ($n = 8$) (Nagano et al., 2013). Additionally, when investigating the dangerousness of slips, Myung (2003) reported that half of the observed backward slips (5 out of 10 slips) were classified as dangerous (were arrested by a fall arresting system) when only 4 out of 14 forward slips triggered a dangerous fall. Therefore, backward slips and their recovery strategies require further attention.

Recent studies on recovery from gait perturbations show that large internal joint moments are required in response to backward slips (Debelle et al., 2020), forward slips (Yoo et al., 2019), and trips (King et al., 2019) to arrest the abnormally large angular momentum and regain control of the centre of mass (COM) position and velocity. Accordingly, the age-related deterioration in plantarflexor and knee extensor muscles' strength and tendons' reduced stiffness has been correlated with impaired balance in static (Onambele et al., 2006) and dynamic (Karamanidis et al., 2008) conditions, and linked to poorer control of the body angular momentum following trips (Pijnappels et al., 2005b). However, even though resistance training interventions successfully improve balance in static and dynamic situations (for review, see Chang et al., 2004; Papa et al., 2017), they do not necessarily directly transfer to better balance recovery when OAs are exposed to gait perturbations (Pijnappels et al., 2008). This has led to hypothesise that task-specific training (i.e., exposure to simulated slip- or trip-like perturbations) may be more beneficial than resistance exercise (for review, see Grabiner et al., 2014). The rationale for developing such interventions is that they better mimic the sensory feedback experienced during real, outside lab environment, perturbations than resistance training. They could potentially be used to adapt well-known motor schemes (here, gait pattern) to closely match the requirements induced by the change to compensate (in the present context: the perturbation), and this new behaviour could be retained and automatised for future exposure to similar conditions (Doyon and Benali, 2005). This may also be efficacious for OAs, as the ability to learn new motor skills is maintained with ageing (Durkina et al., 1995; Boyke et al., 2008; Pai et al., 2010). Further supporting the advantages of task-specific over resistance training interventions, OAs exposed to both interventions did not display better improvements than those exposed to only task-specific training (Epro et al., 2018b). Fall recovery training protocols have confirmed the ability of young and older adults to improve their balance following exposure to multiple perturbations, within one session (Konig et al., 2019b; McCrum et al., 2020) and in the long term (Bhatt et al., 2012; Epro et al., 2018b), although when compared with YAs, long-term retention appears less efficacious in OAs (Konig et al., 2019b).

To implement fall recovery training interventions, it is necessary to use protocols that apply perturbations that are as realistic as possible. Diverse protocols have been developed to achieve this, including, among others, movable low-friction platforms (Bhatt et al., 2012; Okubo et al., 2018), or split-belt instrumented (SBI) treadmills to study trips (King et al., 2019), forward slips (Yoo et al., 2019) and backward slips (McCrum et al., 2018; Debelle et al., 2020). To optimise the delivery of these protocols in fall prevention interventions, it is necessary to understand the underlying biomechanics of successful fall recovery strategies and the evolution of these strategies that result in improved balance recovery following repeated exposures. By understanding the mechanisms underlying an optimal recovery strategy, we might be able to design interventions that will specifically target these mechanisms and might be coachable outside lab environments to a wider public. To date, the mechanisms underlying balance recovery strategies with repeated perturbations have not been fully investigated, partly because of the relative novelty of this field, and also because of difficulties in recording complete kinematic, kinetic, and temporo-spatial data sets from multiple consecutive steps. In this regard, protocols utilising SBI treadmills are advantageous, because they can produce sudden unanticipated perturbation of the foot during stance, and record rich data sets during recovery.

In our previous study documenting the biomechanics of recovery from backward slips simulated by belt accelerations (Debelle et al., 2020), we detailed a protocol developed in our lab to trigger single backward slips in YAs using an SBI treadmill. We reported that in response to an induced backward slip, YAs needed four recovery steps to return their balance to normal levels, increased the length of their base of support during the perturbation by about 8% and decreased it on the following step by about 21%, and developed larger hip (+125% at peak hip extensor moment) and knee (+200% at peak knee extensor moment) moments and lower plantarflexor moments (−25% at peak plantarflexor moment) on the first recovery step, than in typical gait. As balance recovery has been shown to improve with repeated backward slip-like perturbations in YAs and OAs (McCrum et al., 2020), it is possible that the mechanical responses to a single backward slip that we previously measured might change with repeated exposures as the recovery strategy improves.

Therefore, the goal of this study was to establish whether and how young and older adults modified their gait pattern to improve their balance recovery following repeated exposure to backward slip-like perturbations. We used the aforementioned protocol to expose young and older participants to 10 repeated backward slips, and for each slip, we measured their balance on 15 recovery steps (margin of stability) and recovery of balance (n_{steps}), and their kinetic and temporo-spatial variables before slip onset, during the perturbed stance and on the following recovery steps.

First, we hypothesised that both YAs and OAs would improve their balance with repeated exposures to the backward slips. Second, we hypothesised that with repeated slips the recovery strategy will be optimised to better accommodate the effects of the perturbation, through an adjustment of the recovery steps' length and a redistribution of the joint moments to rely more on the hip and knee joints. Finally, we hypothesised that OAs would develop a similar recovery strategy and recovery strategy adjustments as YA, whilst possibly needing more steps to recover their balance.

METHODS

Participants and Protocol

Seventeen young (eight males, nine females, age 25.2 ± 3.7 years, height 176.1 ± 8.1 cm, body mass 71.8 ± 10.1 kg) and 17 older (3 males, 14 females, age 62.4 ± 6.6 years, height 161.8 ± 7.2 cm, body mass 66.5 ± 11.3 kg) adults volunteered to take part in this study. All participants were able to walk unassisted for at least 15 min, and were free from any lower limb injury in the last 6 months, surgery in the last 2 years, and balance, neurological or musculoskeletal disorders.

Participants were exposed to 10 backward slip-like perturbations while walking on an SBI-treadmill (300 Hz, M-Gait, Motek; Motekforce Link, Amsterdam, The Netherlands), and the protocol of the perturbation has been described in detail previously (Debelle et al., 2020). Briefly, after 5 min of familiarisation with participants walking at 1.2 $m \cdot s^{-1}$, we first recorded baseline data (Normal), and then triggered the perturbation at random and unexpected times by an acceleration (5 $m \cdot s^{-2}$) of the belt, followed by a return to normal speed. Perturbations were randomly assigned either to the right or the left side using MinimPy (http://sourceforge.net/projects/minimpy) and applied consistently to that limb. Belt accelerations were designed to start at 20% stance phase with the belt speed returning to 1.2 $m \cdot s^{-1}$ at 70% stance. Mechanical latency and quicker stance phase during the perturbation than in normal gait meant that the perturbation actually occurred slightly later than these timings (**Figure 1**). Previous results on the experimental validity of the protocol indicated a very good consistency between the timings of acceleration beginning [25 (SD 1.2) % of stance, CV = 5%] and return to 1.2 $m \cdot s^{-1}$ [86.5 (SD 3.4) % of stance, CV = 4%, respectively]. These accelerations produced a forward loss of balance during the second half of stance, from which the participants had to adjust to avoid falling.

For safety, participants wore a full-body safety harness attached to a frame above the treadmill. They were instructed beforehand that should they experience a trip or a slip, they should try to recover their balance and continue walking as if they had experienced one outside of the lab. The participants were also asked to avoid using the handrails, and although vigorous arm movements were occasionally observed, none grabbed the handrails. To ensure that participants' balance had returned to normal levels, participants continued walking on the treadmill for 1 to 2 min before the next perturbation trigger. This was repeated until 10 perturbations had been triggered.

Data were recorded for both the ipsilateral (Pre2) and contralateral (Pre1) steps prior to the perturbed step (Pert) and up to the 15th recovery step (Rec15). Zero-dimensional data (margin of stability and step length) were measured at heel strike and one-dimensional data (kinetics and temporo-spatial) over 100% of stance phase (**Figure 1**).

Since the primary aim of this study was to determine the mechanisms by which the participants achieved better recovery and a very large data set was developed during the experiment, it was necessary to include in the main manuscript only the results that helped achieving this aim. Therefore, we only report kinetics and temporo-spatial results (1) if they were significantly different from normal and (2) if they were correlated to balance recovery. Results not meeting these criteria are reported in the **Supplementary Material**.

This study was carried out with the approval of the Liverpool John Moores University and National Health Service (NHS) ethics committees (18/NW/0700). Written consent was obtained in accordance with the declaration of Helsinki.

Evaluation of Balance

Participants' balance was quantified from the margin of stability (MoS) (Hof et al., 2005), measured as the distance between the anterior boundary of the base of support (BoS) (anterior-posterior position of the second toe marker of the leading foot) and the extrapolated centre of mass (XCoM) at heel strike. A positive MoS indicated that the XCoM was located behind the anterior boundary of the BoS and that the participant was stable. Balance was assessed in the two steps prior to each slip (Pre2 and Pre1) to test for changes in walking pattern with repeated slips resulting from the anticipation of a potential upcoming perturbation due to any sensory cue (visual, auditory, or vibration). Balance was also assessed for 15 recovery steps following the slip (Rec1-15) to establish the time course of balance recovery. The MoS data are reported as mean ± SD. For Normal condition, MoS SD was computed as the average of each participant's SD on that trial, for the other trials, MoS SD was computed as the group's SD.

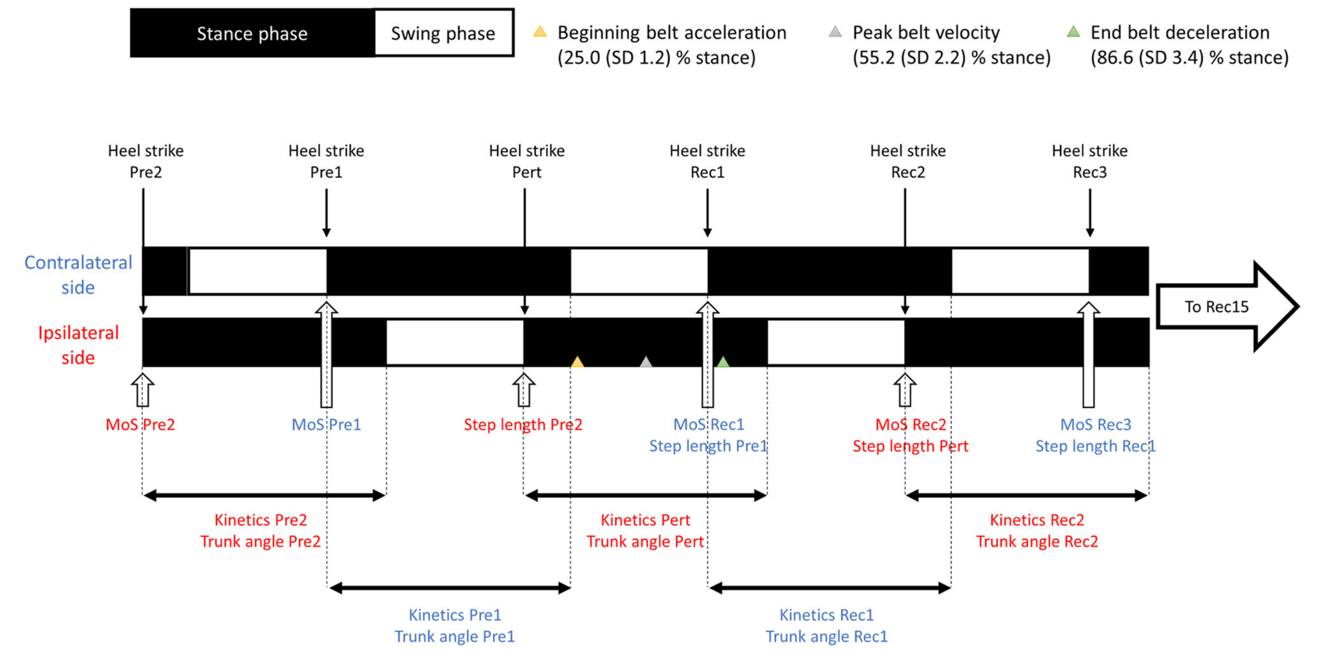

FIGURE 1 | Diagram of steps and outcome measures. In red: outcome measures from the ipsilateral side to the perturbation; in blue, outcome measures from the contralateral side.

The position and velocity of the feet's markers and participants' COM were computed and filtered using a low-pass fourth-order Butterworth filter with a cut-off frequency of 8 Hz in Visual3D (C-Motion; Germantown, MD, United States) before being exported to Matlab (R2020; MathWorks, Natick, MA, United States) for calculation of the MoS.

To quantify how long it took the participants to recover their balance following each slip, we quantified n_{steps} as the first step of at least three consecutive steps within one standard deviation of normal MoS, which was determined as the average of five gait cycles recorded during steady gait on the treadmill after the familiarisation period. When participants did not reach stable gait by the last recorded step (Rec15), we set n_{steps} to 16 (n_{steps} was set at 16 for 14 participants (7 YAs and 7 OAs) during Slip01, and for 4 participants (2 YAs and 2 OAs) during Slip10).

Mechanics of Recovery

We used a 6DoF marker set with 68 retroreflective markers tracked by 12 motion capture cameras (120 Hz; Vicon Motion Systems, Oxford, United Kingdom) to measure three-dimensional whole-body kinetic and kinematic data while the participants were walking on the treadmill. Force data were recorded using Vicon at a sample rate of 1,200 Hz.

We evaluated changes in trunk angle (sagittal plane) and step length (anterior-posterior distance between the centres of mass of each foot at heel strike of the leading foot) for each step of each slip trial. These parameters were chosen, as they could be easily targeted in a fall prevention intervention. To allow comparisons between the participants, step length was computed in percentage of body height (% BH). To understand how participants adapted their gait pattern between the first and last slips, we measured the internal joint moments at the hips, knees, and the ground reaction force angle to the vertical (GRF_θ, $+ =$ anterior) as the inverse tangent of the ratio between the anterior-posterior and vertical GRF vectors.

Kinetics (joint moments) and kinematics (trunk angle) data were computed in Visual3D, using inverse dynamics for the joint moments, and filtered using a low-pass fourth-order Butterworth filter with a cut-off frequency of 8 Hz. The same filter was used on the anterior-posterior and vertical GRF vectors and temporo-spatial (location of the feet's COM) data that were then exported to Matlab where the GRF_θ and step length were computed.

Statistical Analysis

All the variables were tested for normality by Shapiro–Wilk's test.

To test whether the participants changed their gait pattern in anticipation of the slip, we compared the MoS, kinetic and temporo-spatial variables during normal with Pre2 and Pre1 of Slip01 and Slip10. When main effects of Age (YAs, OAs) or Conditions (Normal, Slip01_Pre2, Slip01_Pre1, Slip10_Pre2, Slip10_Pre1) were detected, Bonferroni *post hoc* tests were performed and alpha adjusted to the number of tests (α = 0.01 or α = 0.0063, respectively). For the MoS, step length, and joint moments during Pre, we performed non-parametric tests [Mann2Whitney (Age: YAs, OAs), Friedman (Conditions: Slip01 to Slip10), and Wilcoxon signed rank tests for *post-hoc* comparisons], and parametric mixed-design ANOVAs for the trunk angle.

To test whether participants' balance was different from normal following each slip, we compared the MoS of each recovery step (Rec1–Rec15) with Normal using mixed-design ANOVAs: Age (YA, OA), Conditions (for each slip trial: Normal, Rec1 to Rec15), Age*Conditions. Because we repeated the analysis 10 times, α was adjusted to 0.005.

To test for differences in the number of recovery steps required to return to normal balance with repeated slips, we compared n_{steps} between each slip trial using Mann-Whitney (Age: YAs, OAs), Friedman (Conditions: Slip01 to Slip10), and Wilcoxon signed rank tests for *post-hoc* comparisons with Bonferroni adjustments ($\alpha = 0.0011$).

To test for differences in the biomechanics of recovery following the first (Slip01) and last (Slip10) slips, we evaluated the changes in the reactive kinetic and temporo-spatial variables measured during the perturbation and the first (Rec1) and second (Rec2) recovery steps of Slip01 and Slip10. The following conditions were included in the analysis: Normal, Slip01_Pert (perturbed step of Slip01), Slip01_Rec1 (first recovery step of Slip01), Slip01_Rec2 (second recovery step of Slip01), Slip10_Pert, Slip10_Rec1, and Slip10_Rec2. Although the perturbed step cannot be considered as a recovery step per se, we included it in the present analysis to then evaluate whether and how changes in the biomechanics during the perturbation affected the balance and balance recovery in the following steps. When a main effect of Age was detected, Bonferroni *post-hoc* tests were performed and alpha adjusted to the number of tests ($\alpha = 0.007$). When a main effect of Condition was detected, Bonferroni *post-hoc* tests were performed (n =9: Normal vs. Pert, Rec1, and Rec2 for both Slip01 and Slip10, Slip01 vs. Slip10 in Pert, Rec1, and Rec2) and alpha adjusted to the number of tests performed ($\alpha = 0.0056$).

Finally, when we found a significant effect of condition on the MoS in the Pre steps, kinetics, kinematics and temporo-spatial variables, we used bivariate parametric and non-parametric correlations to understand whether and how these variables affected the balance (MoS) on the following recovery steps, and the balance recovery (n_{steps}) on that trial. Specifically, we ran a correlation analysis to understand (1) whether and how participants' balance (MoS) prior to the slip was related to participants' balance following the slip, (2) whether and how participants' balance (MoS or n_{steps}) in the first slip trial was related to the balance in the last slip trial, and (3) whether and how the kinetic, kinematic, and temporo-spatial adjustments made when recovering from Slip01 and Slip10 affected the MoS of the next recovery steps and n_{steps} on that slip. For (1), we ran a correlation analysis between the MoS during Slip01_Pre2 and Slip01_Rec1, Slip01_Pre1 and Slip01_Rec1, Slip10_Pre2 and Slip10_Rec1, and between Slip10_Pre1 and Slip10_Rec1. For (2), we ran a correlation analysis to understand whether the MoS of Slip01_Rec1 and Slip01_Rec2 was correlated with the MoS of Slip10_Rec1 and Slip10_Rec2, respectively, and whether n_{steps} of Slip01 was correlated with n_{steps} of Slip10. For (3), when a kinetic, kinematic, or spatio-temporal variable was significantly different from Normal levels, we ran a correlation analysis to evaluate whether this variable was related to the MoS of the next steps or to n_{steps} of that slip trial. To use one-dimensional variables (kinetic and temporo-spatial variables for which a significant effect of condition was found between a step and Normal) in the correlation analysis, we used the average from the region of interest (region of significant difference from Normal as determined by statistical parametric mapping, SPM). Because n_{steps} was not normally distributed, the correlations between kinetics or temporo-spatial parameters and n_{steps} should be treated with caution. Participants whose n_{steps} was set to 16 were excluded from the correlation analysis including n_{steps}.

Statistical analysis was performed using SPSS 26 (IBM, NY) for zero-dimensional data (i.e., MoS, n_{steps}, and step length), and we performed statistical parametric mapping in Matlab for one-dimensional data (i.e., GRF_θ, joint moments and trunk angle).

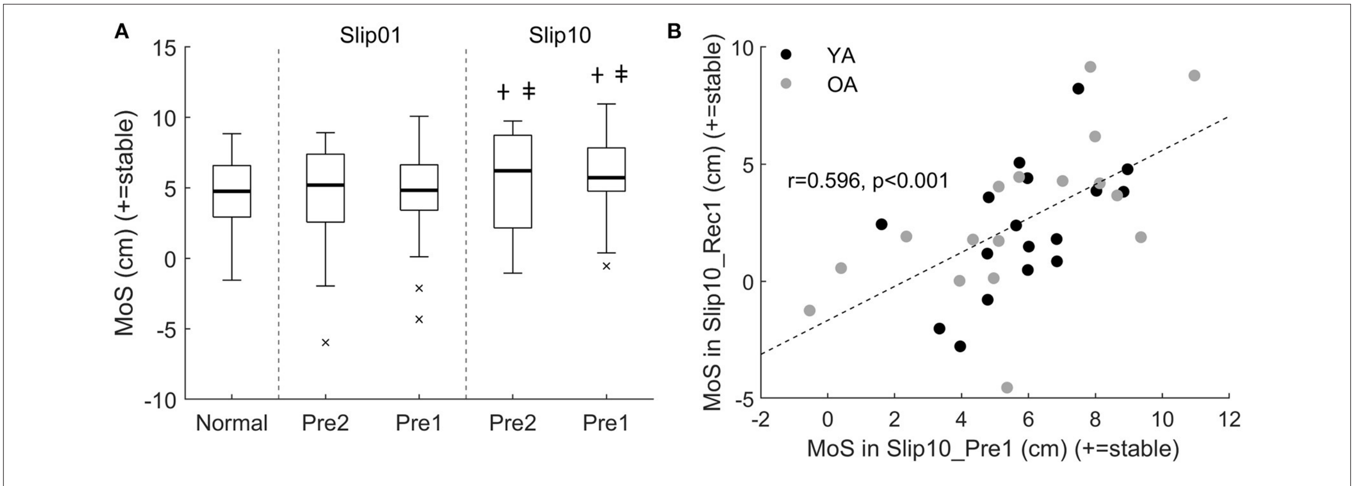

FIGURE 2 | (A) Boxplots of the margin of stability (MoS) for all participants in the two steps prior to Slip01 and Slip10 (Pre2 and Pre1). Thick horizontal black lines: median; thin horizontal black line: first and third quartiles; ×: outliers. †: Slip10 significantly higher than normal ($p \leq 0.005$); ‡: Slip10 significantly higher than Slip01 ($p \leq 0.023$). See **Supplementary Figure 1** for participants' data points. **(B)** Correlation between MoS in Slip10_Pre1 and MoS in Slip10_Rec1, $r = 0.596$, $p < 0.001$. Black circles: young adults (YAs); grey circles: older adults (OAs).

RESULTS

Anticipatory Adjustments

The margin of stability at heel strike of the two steps prior to the first and last slips (Pre2 and Pre1) was not significantly different between the age groups ($p = 0.309$). Irrespective of age, there was no difference in the MoS between the first slip (Pre2 or Pre1) and Normal, but the MoS was larger in the Pre2 and Pre1 of the last slip than both Normal and the equivalent steps of the first slip ($p \leq 0.023$, **Figure 2A**). These anticipatory adjustments of the MoS before the last slip were correlated with better balance following that last slip (Spearman's rho = 0.565, $p = 0.001$ between the MoS in Slip10_Pre2 and Slip10_Rec1, and $r = 0.596$, $p < 0.001$ between the MoS in Slip10_Pre1 and Slip10_Rec1, **Figure 2B**). No significant difference existed between Slip01_Pre2 and Slip01_Pre1 ($p = 0.614$), or between Slip10_Pre2 and Slip10_Pre1 ($p = 0.567$), suggesting that although the participants might have adjusted their balance in anticipation of a potentially upcoming perturbation, they did not anticipate its timing.

We found no main effect of either Age or Conditions on the knee and ankle moments or trunk angle during the steps preceding the perturbations ($p > 0.05$). Significant main effects of Age and Conditions existed for both step length and hip moments during the pre-slip steps, but these changes were not correlated with balance (neither the MoS of the first and second recovery steps of Slip10 nor n_{steps} in Slip10, $p > 0.05$).

Recovery Adjustments

We found no significant effect of Age on the MoS or n_{steps} ($p > 0.005$ for MoS and $p = 0.052$ for n_{steps}). We found a significant effect of Conditions on the MoS from Slip01 to Slip06 ($p < 0.005$) and on n_{steps} ($p < 0.001$). A *post hoc* analysis showed that following Slip01 and Slip02, the MoS was significantly lower than Normal (MoS Normal = 4.6 ± 1.3 cm) until the sixth and fifth recovery steps, respectively (MoS Slip01_Rec6 = 2.8 ± 3.6 cm, $p = 0.005$, and MoS Slip02_Rec5 = 3.3 ± 3.6 cm, $p < 0.001$). From Slip03 to Slip06, only Rec1 had a significantly lower MoS compared with Normal (MoS Slip03_Rec1 = 2.0 ± 3.7 cm, $p = 0.007$; MoS Slip06_Rec1 = 1.9 ± 3.1, $p = 0.001$) (**Figure 3A**). Significant positive correlations between the MoS of Slip01_Rec1 and the MoS of Slip10_Rec1 ($r = 0.698$, $p < 0.001$), and between the MoS of Slip01_Rec2 and the MoS of Slip10_Rec2 (Spearman's rho = 0.661, $p < 0.001$) indicate that the participants who recovered well during the first slip trial tended to also recover well during the last trial.

Accordingly, n_{steps} decreased with the number of slips (n_{steps} Slip01 = 10.3 ± 5.6 steps, n_{steps} Slip10 = 4.9 ± 5 steps, $p < 0.001$) until Slip03, from which n_{steps} was not significantly larger than in Slip10 (n_{steps} Slip02 = 9.1 ± 5.5 steps, $p < 0.001$; n_{steps} Slip03 =7.2 ± 6.2 steps, $p = 0.016$; n_{steps} Slip04 = 6.1 ± 5.1 steps, $p = 0.066$). From Slip03 to Slip10, n_{steps} was constantly lower than n_{steps} in Slip01 ($p \leq 0.001$) (**Figure 3B**).

Trunk angle data have not met the criteria for being reported in the main manuscript and are reported in the **Supplementary Material**.

During the perturbation, the participants' sagittal GRF_θ was directed more posteriorly during mid stance than in Normal condition for Slip01 ($p < 0.001$ from 18 to 67% of stance, **Figure 4A**), and then directed more anteriorly than in Normal at the end of stance ($p < 0.001$ from 71 to 90% of stance, **Figure 4A**). These modifications were maintained during Slip10 ($p < 0.001$ from 15 to 60% and from 68 to 90% of stance, **Figure 4A**), and we found that in Slip10, the participants whose GRF_θ was directed more posteriorly during mid stance (averaged from 15 to 60% stance) were those who better recovered their

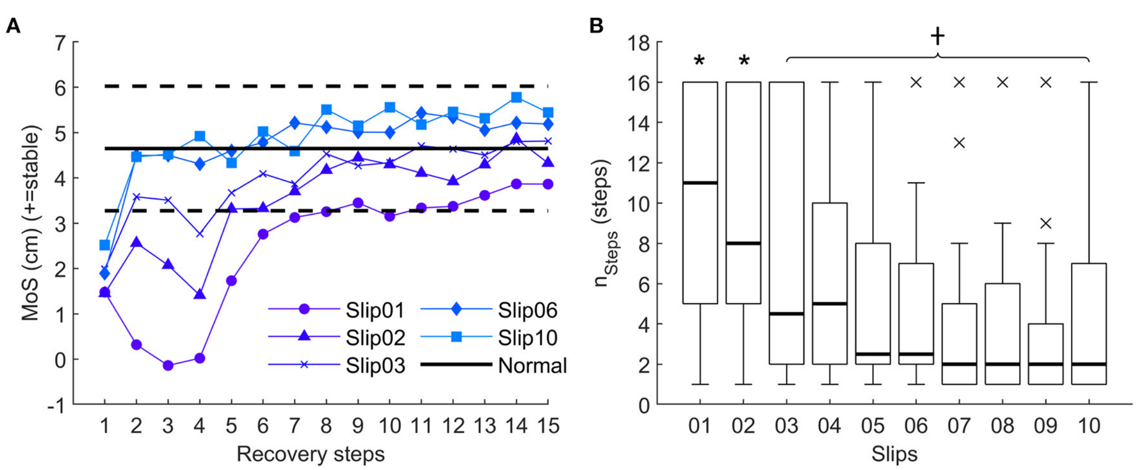

FIGURE 3 | (A) Average margin of stability for all participants for the 15 recovery steps recorded for Slip01, Slip02, Slip03, Slip06, and Slip10. Solid and dotted horizontal black lines represent Normal ± 1 standard deviation (SD), respectively. See **Supplementary Figure 2** for YA and OA curves. **(B)** Boxplots of n_{steps} (i.e. first step of at least three consecutive steps back to ± 1 SD of Normal MoS) from Slip01 and Slip10. Thick horizontal black lines: median; thin horizontal black line: first and third quartiles; ×: outliers. *: significantly larger than Slip10, $p \leq 0.0011$; †: significantly lower than Slip01, $p \leq 0.0011$. See **Supplementary Figure 3** for participants' data points.

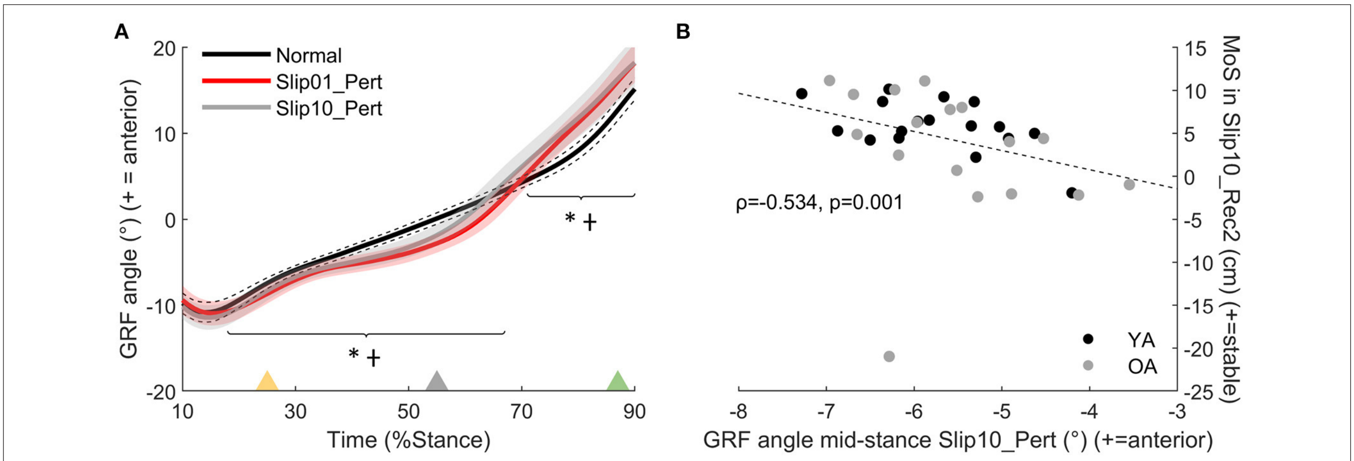

FIGURE 4 | (A) Ground reaction force angle (GRF$_\theta$) (°), effect of condition between Normal (black line), Slip01_Pert (red line), and Slip10_Pert (grey line); *: Slip01 significantly different from Normal $p < 0.001$ from 18 to 67% and from 71 to 90% of stance; †: Slip10 significantly different from Normal $p < 0.001$ from 15 to 60% and from 68 to 90% of stance. Yellow triangle: beginning of belt acceleration; grey triangle: peak belt speed; green triangle: belt speed returns to 1.2 m·s^{-1}. See **Supplementary Figure 4** for YA and OA curves. **(B)** Correlation between GRF$_\theta$ during mid stance of Slip10_Pert and MoS in Slip10_Rec2 (Spearman's rho = −0.534, $p = 0.001$). Black circles: young adults; grey circles: older adults.

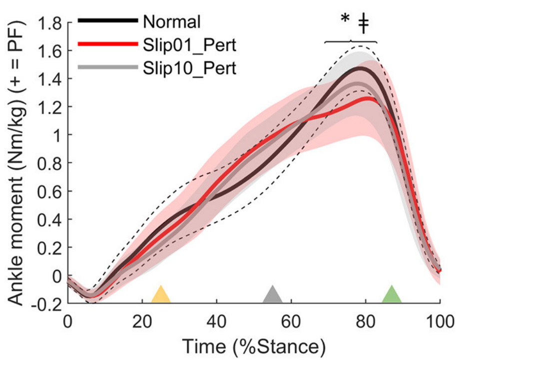

FIGURE 5 | Ankle moment, effect of condition between Normal (average: solid black line; standard deviation: dotted black lines), Slip01_Pert (average: solid red line; standard deviation: red shaded area), and Slip10_Pert (average: solid grey line; standard deviation: grey shaded area). *: significant difference between Normal and Slip01 (69–83% stance, $p = 0.002$); ‡: significant difference between Slip01 and Slip10 (70–79% stance, $p < 0.001$). Yellow triangle: beginning of belt acceleration; grey triangle: peak belt speed; green triangle: belt speed returns to 1.2 m·s^{-1}. See **Supplementary Figure 5** for YA and OA curves.

balance during Rec2 of Slip10 (Spearman's rho = −0.534, $p = 0.001$, **Figure 4B**).

We found that participants developed a lower than Normal plantarflexor moment during push-off of the first slip (Slip01_Pert: $p = 0.002$ from 69 to 83% stance, **Figure 5**), which returned to Normal levels by the last slip ($p > 0.0056$ between Slip10 and Normal; $p < 0.001$ from 70 to 79% stance between Slip01 and Slip10, **Figure 5**).

The OAs developed a lower plantarflexor moment than YAs at push-off of the perturbed step during the last slip than YA ($p = 0.004$ from 69 to 79% stance for Slip10_Pert, **Figure 6A**, not significant for Slip01). During that same step, participants who developed larger plantarflexor moments were the ones with the higher MoS at heel strike of the second recovery step (Slip10_Rec2, Spearman's rho = 0.368, $p = 0.032$, **Figure 6B**). These suggest that the OAs, who were grouped lower on the Ankle Moment – MoS correlation graphs (**Figure 6B**), might be at higher risk of falling than the YAs, at least partly due to an inability to produce enough propulsive force at push-off of the leg that slipped.

On average, participants took a longer step in the first and last slips than in Normal (Slip01_Pert and Slip10_Pert, $p < 0.001$, **Figure 7A**). The participants who took a longer step in Slip10_Pert were those who had the GRF$_\theta$ directed more anteriorly in Slip10_Pert (Spearman's rho = 0.644, $p < 0.001$, **Figure 7B**), and those who better recovered their balance by the third recovery step of the last slip (Slip10_Rec3, Spearman's rho = 0.437, $p = 0.01$, **Figure 7C**).

During the first recovery step, participants developed a larger knee extensor moment ($p = 0.002$ from 23 to 79% stance, **Figure 8A**) and a lower ankle plantarflexor moment ($p = 0.002$ from 37 to 90% stance, **Figure 8B**) in the first slip (Slip01_Rec1) compared with Normal. These joint moments had returned to Normal levels by the last slip ($p > 0.0056$ between Slip10_Rec1 and Normal, $p = 0.002$ for knee moment from 32 to 80% of stance between Slip01_Rec1 and Slip10_Rec1 (**Figure 8A**), and $p < 0.001$ for ankle moment from 34 to 89% of stance between Slip01_Rec1 and Slip10_Rec1, **Figure 8B**).

Knee and ankle extensor moments in Slip01_Rec1 had moderate to good correlations with MoS of the second recovery step (Slip01_Rec2, r = −0.434, $p = 0.01$, **Figure 9A**; $r = 0.496$, $p = 0.003$, **Figure 9B**, respectively). Therefore, participants who developed a larger knee extensor moment

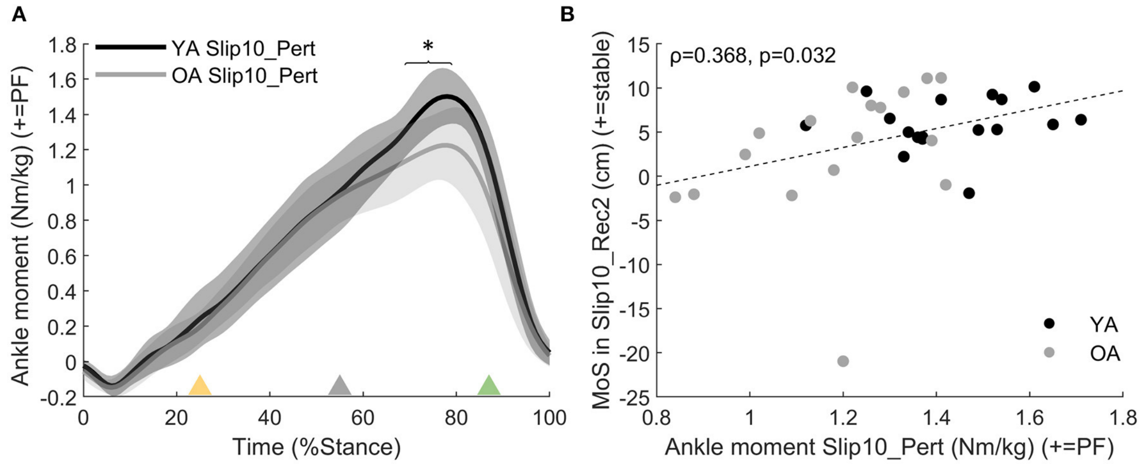

FIGURE 6 | (A) Ankle moment age difference between YAs and OAs for Slip10_Pert. Solid black line: YA in Slip10_Pert; solid grey line: OA in Slip10_Pert. *: significant effect of Age ($p = 0.004$ from 69 to 79% stance). Yellow triangle: beginning of belt acceleration; grey triangle: peak belt speed; green triangle: belt speed returns to 1.2 $m \cdot s^{-1}$. **(B)** Correlation between ankle moment in Slip10_Pert (69–79% stance) and the MoS of Slip10_Rec2 (Spearman's rho = 0.368, $p = 0.032$). Black circles: young adults; grey circles: older adults.

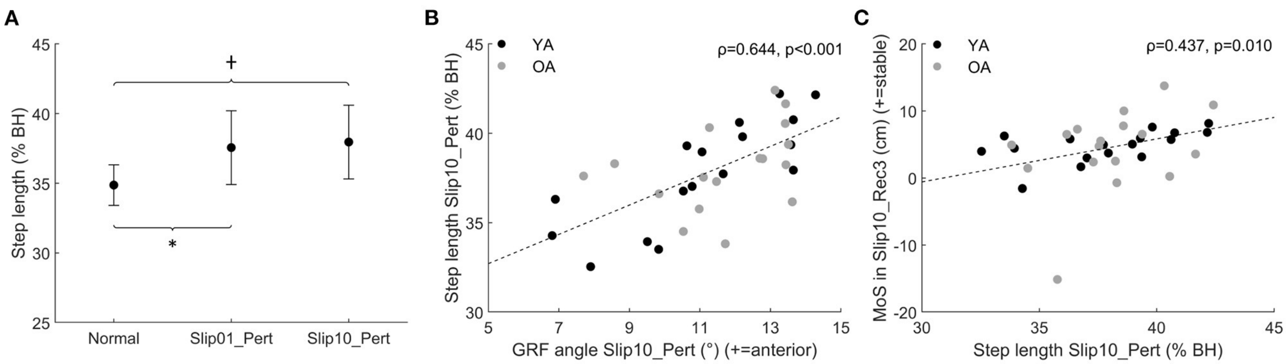

FIGURE 7 | (A) Step length of Slip01_Pert and Slip10_Pert compared with Normal (average ± SD as error bars); *: significant difference between Slip01 and Normal, $p < 0.001$. †: significant difference between Slip10 and Normal, $p < 0.001$. See **Supplementary Figure 6** for participants' data points. **(B)** Correlation between GRF_θ during push-off of Slip10_Pert (68–90% of stance) and step length in Slip10_Pert (Spearman's rho = 0.644, $p < 0.001$). **(C)** Correlation between step length in Slip10_Pert and MoS in Slip10_Rec3 (Spearman's rho = 0.437, $p = 0.01$). Black circles: young adults; grey circles: older adults. % BH: % body height.

and a lower ankle plantarflexor moment in mid stance of Slip01_Rec1 seemed to have a poor balance during the following steps.

On average, participants took a smaller step in Slip01_Rec1 that returned to Normal length by the last slip ($p < 0.001$ between Slip10_Rec1 and Slip01_Rec1, **Figure 10A**). There were moderate to good correlations between the length of the first recovery step of the first slip (Slip01_Rec1) and the MoS of the next step (Slip01_Rec2, $r = 0.648$, $p < 0.001$, **Figure 10B**) and between step length in Slip01_Rec1 and n_{steps} in Slip01 (Spearman's rho = −0.48, $p = 0.004$, **Figure 10C**), suggesting that during the first slip participants who took a longer step in Rec1 seemed to be those who had a better balance on the following step and required fewer steps to return to a stable balance during that trial.

We found that the length of the first recovery step during the first slip trial (Slip01_Rec1) was correlated with ankle and knee moments during that step, with the participants who developed the larger ankle plantarflexor moment during push-off being those who took the longer step ($r = 0.75$, $p < 0.001$, **Figure 11A**), and those who developed the larger knee extensor moment in mid stance being the ones who took the shorter step ($r = -0.477$, $p = 0.004$, **Figure 11B**).

None of the variables that differed from Normal *during the second recovery step* were correlated with balance. Therefore, results related to the mechanisms of recovery in the second recovery step are reported in the **Supplementary Material**.

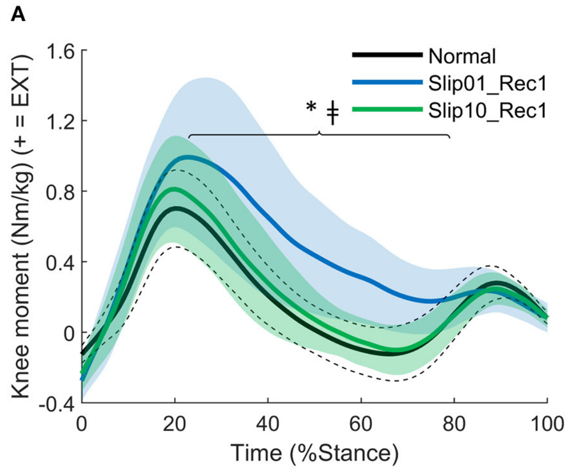

FIGURE 8 | (A) Knee moment: effect of condition between Normal, Slip01_Rec1, and Slip10_Rec1. *: significant difference between Normal and Slip01 (23–79% stance $p = 0.002$); ‡: significant difference between Slip01 and Slip10 (32–80% stance, $p = 0.002$). **(B)** Ankle moment: effect of condition between Normal, Slip01_Rec1 and Slip10_Rec1. *, significant difference between Normal and Slip01 (37–90% stance, $p = 0.002$); ‡: significant difference between Slip01 and Slip10 (34–89% stance, $p < 0.001$). Normal: average: solid black line; SD: dotted black lines; Slip01_Rec1: average: solid blue line; SD: blue shaded area; Slip10_Rec1: average: solid green line; SD: green shaded area. See **Supplementary Figure 7** for YA and OA curves.

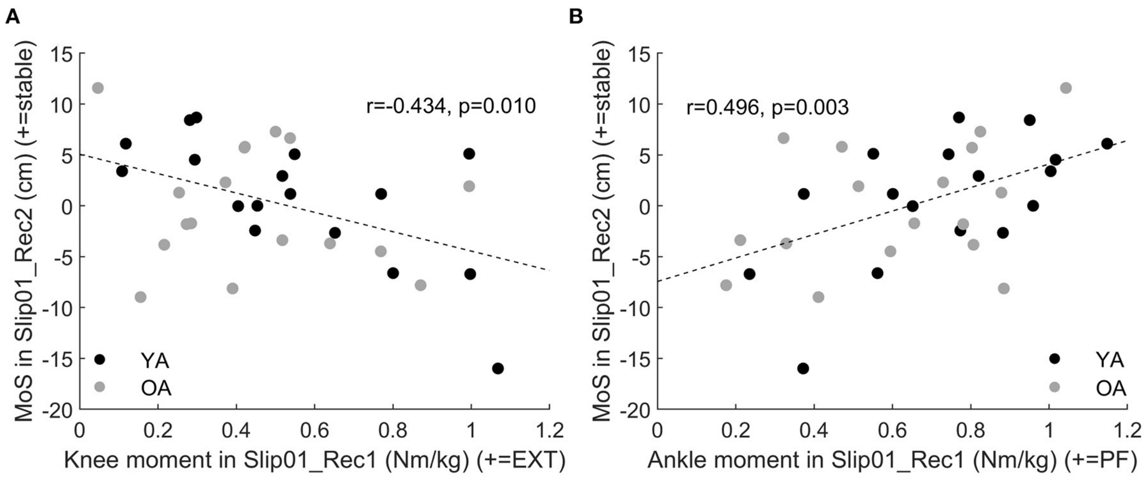

FIGURE 9 | (A) Correlation between knee moment in Slip01_Rec1 from 23 to 79% stance and the MoS of Slip01_Rec2 ($r = -0.434$, $p = 0.01$). **(B)** Correlation between ankle moment in Slip01_Rec1 from 37 to 90% stance and the MoS of Slip01_Rec2 ($r = 0.496$, $p = 0.003$). Black circles: young adults; grey circles: older adults.

DISCUSSION

With this study, we showed that: (a) balance recovery following repeated slip-like perturbations simulated by treadmill belt accelerations can be improved with repeated exposure in young and older adults, which supports our first hypothesis, and more importantly, (b) the older adults demonstrated improvements that were not different to those of younger adults. Following the first slip, participants utilised biomechanical responses that were associated with both better and worse recovery. However, the recovery strategy was optimised with repeated exposures to preferentially retain only the responses associated with better recovery or which resulted in a rapid return to normal balance following the slip. Generally, this optimal recovery strategy requires changes in the orientation of the GRF vector (**Figures 4A,B**), length of the perturbed and recovery steps (**Figures 7A**, **10A**), and internal moments around the knee and ankle joints (**Figures 5**, **8A,B**). This improvement in balance recovery after repeated exposure was, in part, achieved by adopting a length for the first recovery step closer to normal,

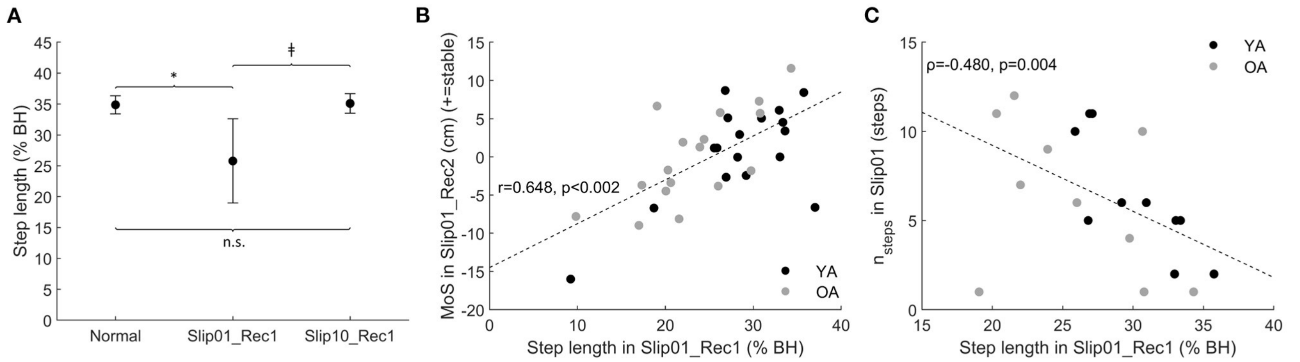

FIGURE 10 | (A) Step length of Slip01_Rec1 and Slip10_Rec1 compared with Normal (average ± SD as error bars); *: significantly shorter than Normal, $p < 0.001$; ‡: Slip10 significantly longer than Slip01, $p < 0.001$; n.s.: no significant difference between conditions. See **Supplementary Figure 8** for participants' data points. **(B)** Correlation between step length in Slip01_Rec1 and MoS of Slip01_Rec2 ($r = 0.648$, $p < 0.001$). Black circles: young adults; grey circles: older adults. **(C)** Correlation between step length in Slip01_Rec1 and n_{steps} (first of at least three consecutive steps within one standard deviation of normal MoS) in Slip01 (Spearman's rho = −0.480, $p = 0.004$), participants whose n_{steps} had not returned to Normal by Rec15 were removed from the correlation analysis. Black circles: young adults; grey circles: older adults. % BH: % of body height.

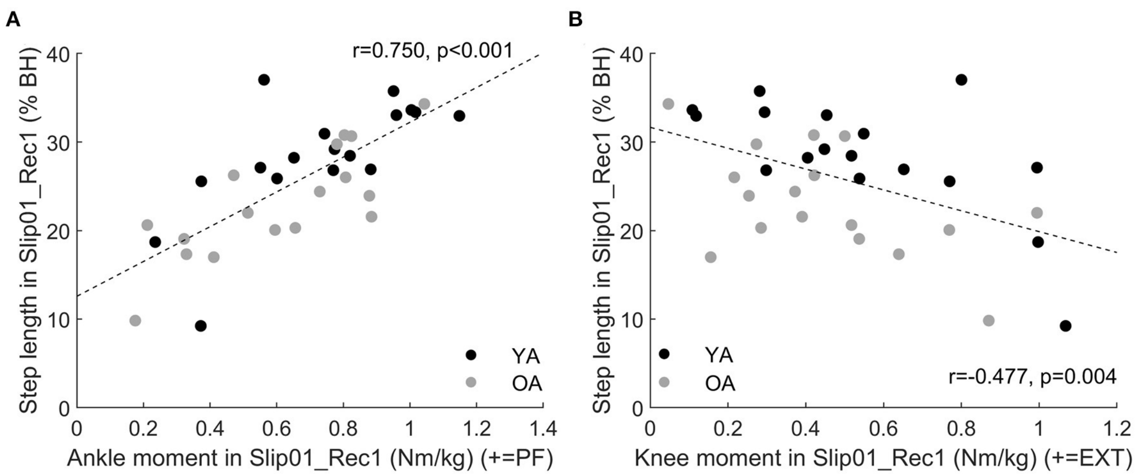

FIGURE 11 | (A) Correlation between ankle moment in Slip01_Rec1 from 37 to 90% stance and step length in Slip01_Rec1 ($r = 0.75$, $p < 0.001$). **(B)** Correlation between knee moment in Slip01_Rec1 from 23 to 79% stance and step length in Slip01_Rec1 ($r = −0.477$, $p = 0.004$). Black circles: young adults; grey circles: older adults. % BH: % body height.

which offers an easily explained and monitored strategy to teach in fall prevention interventions. Together, these findings give further evidence that fall prevention interventions that use repeated backward slip-like perturbations on an instrumented treadmill as a form of training have the potential to be effective for this mechanism of falling.

During Slip01, the direction of the GRF vector was adjusted and the step length increased during the perturbed step; these characteristics were associated with a more optimal strategy. However, participants also developed low ankle plantarflexor moments during the slip and first recovery step, high knee extensor moments during Rec1, and took a small step during Rec1, which were all associated with poor balance recovery. The recovery strategy did not differ between the age groups; therefore, we accept the hypothesis that OAs developed a similar recovery strategy as YAs on the first slip. However, independent of age, this gross, generalised reaction to the first slip was then fine-tuned to retain only the beneficial characteristics by the 10th slip, in which the participants demonstrated a more optimal recovery strategy. Specifically, by Slip10, participants' GRF vector was still directed more posteriorly during mid stance and anteriorly during push-off, they took a longer step during the slip, generated larger plantarflexor moments compared with Slip01 during both Pert and Rec1, and had returned their knee moments and step length of Rec1 back to normal levels. We, therefore, accept the hypothesis that step length is altered with repeated backward slip-like perturbations, but reject the hypothesis that the joint moments would be redistributed to rely more on the hip and knee joints and less on the ankle joint.

This improved balance and shift towards an optimal strategy by Slip10 for both YAs and OAs validate our hypothesis on the recovery strategy developed by young and older adults, and show that task-specific perturbation training by exposure to multiple mechanical perturbations can be used as an intervention to improve balance recovery from backward slip-like perturbations, as already demonstrated elsewhere (McCrum et al., 2020). With this study, we have established biomechanical strategies by which the improvement in recovery is achieved. However, whether this can be used to reduce the risk of falling in outside-lab, real-world conditions remains to be examined for this particular type of perturbation. Particularly, we showed that keeping the step length close to normal levels was an important component of balance recovery. More studies are needed to understand whether interventions training older adults to maintain a normal step length in response to external perturbations can prevent falls in real-world conditions. However, as the step length is (1) easy to monitor outside lab settings and (2) easily understandable by participants, fall prevention interventions targeting the step length and not requiring specialised treadmills should be developed, and if successful in decreasing fall risks, could be used to reach larger cohorts.

Increased step length, as it can compensate for larger COM displacement, independently of the direction and type of perturbation, has often been associated with better balance for postural perturbations (Owings et al., 2001), trips (Okubo et al., 2018), forward slips (Patel and Bhatt, 2015), and now backward slips. The contribution of the joint moments to balance recovery, however, is task specific, as the mechanical requirements vary widely, but the general consensus tends towards the development of large internal moments at the lower limb joints as a reaction to the perturbation (Pijnappels et al., 2005a; Liu and Lockhart, 2009; King et al., 2019; Yoo et al., 2019). Surprisingly, in this study, we found that optimal recovery strategy did not require the development of larger than normal joint moments, and that large knee extensor moments were actually correlated with poor balance. In the aforementioned studies, the adjustments of the joint moments in response to repeated perturbations were not accounted for, neither were the joint moments directly correlated with balance performance. Therefore, the effect of these kinetic changes on fall avoidance and balance recovery was assumed but not actually demonstrated. We cannot rule out that large joint moments may be important for recovery from perturbations more mechanically demanding than the one we applied, but as reported in the **Supplementary Material**, hip moments during the perturbed stance and the first recovery step, as well as knee moments during the perturbed step, were larger than in normal condition and did not return to normal levels by Slip10. As we could not link these changes to improved balance recovery, we reject the hypothesis that the kinetics of improved recovery strategy mainly relies on higher knee and hip joints moments. However, as these changes are likely energy demanding, they would have been dampened by Slip10 if they were not necessary or did not provide some margin of safety in recovery. Therefore, although these increased internal joint moments were not correlated with improved balance as measured in this study (MoS or n_{steps}), they might be correlated with other markers of balance. These adjustments may have been maintained because they had a positive impact on the vertical state of the COM rather than its horizontal (anterior posterior) state, as evaluated in this study (MoS), or on the regulation of the whole-body angular momentum. Therefore, the optimal recovery strategy described here only reflects the optimal strategy used to recover balance as measured by the anterior-posterior MoS, and other factors might affect dynamic stability.

Contrary to previous research quantifying balance recovery in static (Onambele et al., 2006) and dynamic conditions (Bierbaum et al., 2010; Pai et al., 2010; Konig et al., 2019b; McCrum et al., 2020) in YAs and OAs, we did not detect an effect of age on the balance ability of the participants (neither on MoS nor on n_{steps}). One possibility for this lack of difference between the age groups could be that the perturbation triggered in this study did not present a mechanical demand high enough to discriminate the two groups. Indeed, although the MoS was lower than in normal condition it remained, on average, positive. Despite this lack of significant effect of age, trends were apparent on both the balance and the recovery strategy developed by the participants. We have previously shown that the MoS of YAs was lower than normal up to the fourth recovery step on the first exposure to a backward slip-like perturbation (Debelle et al., 2020); whereas here, the results show that when both the age groups are analysed together, participants need on average six recovery steps to return to normal MoS, indicating a tendency from OAs to be less stable than the YAs. Despite this tendency, the lack of significant age effect refutes the hypothesis that the number of recovery steps required to recover balance would be greater in OAs than in YAs. The OAs also tended to be grouped towards the low end of the correlation figures between kinetic or temporo-spatial variables and MoS (**Figures 6B**, **10B**). Another explanation for the lack of differences between the OAs and the YAs in this study might simply be that the OAs we recruited were healthy, able to walk unassisted, and of relatively young age (62.4 ± 6.6 years), which may have shifted the results towards an undetectable effect of age. Therefore, caution should be exercised before extrapolating these results to frailer populations.

Regardless of age, participants' balance (MoS and n_{steps}) improved with repeated exposures to backward slip-like perturbations. This is consistent with findings that ageing does not affect the capacity to learn new motor tasks per se (Bock and Schneider, 2002), and functionally that both YAs and OAs can improve their balance when exposed to repeated perturbations (Bierbaum et al., 2010; Pai et al., 2010; Konig et al., 2019b; McCrum et al., 2020). We observed high inter-individual variability in our results (**Figure 3B**), which could suggest a need for further training in participants performing poorly to achieve the same performance levels than the most proficient ones. Also, the MoS was measured as the distance between the anterior boundary of the BoS and the extrapolated centre of mass. This is based on the false assumption that the centre of pressure can travel infinitely fast (Hof and Curtze, 2016) and therefore overestimates the location of the anterior boundary of the BoS, which leads to an underestimation of the instability. An alternative way to measure the MoS would have been to downsize the BoS. To the best knowledge of the authors, there is no agreement yet on the proportion of the BoS that should be used to measure the MoS during perturbed walking conditions. However, results from the functional BoS measured during standing tasks show that the size of the functional BoS decreases with age (King et al., 1994; Tomita et al., 2021), which, if accounted for, could have affected the between-groups results. A further study is needed to fully understand which factors, if not age, can explain these limits in the improvement of balance with multiple exposures.

The time course by which participants reached the optimal recovery strategy, i.e., whether they gradually adjusted their response after each perturbation until reaching it or whether they selected and applied the optimal recovery strategy from pre-existing motor programs following repeated perturbations, was not investigated in this study. However, the results on balance (MoS and n_{steps}) suggest an improvement with repeated slips within one session, which is consistent with previous studies on different kinds of gait perturbations (Pai et al., 2010; Konig et al., 2019b). Contradictory results show that the adaptation from repeated forward slips might only happen after an initial observational stage of three perturbations, in which the activity of the prefrontal cortex and the kinematics response to the perturbations were not modified (Lee et al., 2020). Therefore, further study is necessary to understand the time course of strategy adjustment, which is important to optimise the delivery of interventions utilising this approach.

Our results indicate a stabilisation of the effect no later than Slip06, but the number of repetitions required to provide a lasting effect is not known. In this study, we observed a plateau in the MoS improvement past the sixth slip, and from the third slip the number of steps required to return to normal balance did not differ from Slip10. These results suggest that three to six perturbations might be enough to trigger an online learning effect for backward slip-like perturbations, which is consistent with findings on forward slips (Pai et al., 2010) and trips (Epro et al., 2018a). However, the large standard deviation found in this study on both the balance and the mechanisms of balance recovery suggests that this threshold might be individual dependent.

Similar to the online learning effect, the long-term retention of the balance improvements, which is outside the scope of this study, is also conditioned by the number of perturbations triggered. Indeed, a single perturbation was not enough to trigger a long-term retention (Konig et al., 2019a), but a small number of trials ($n = 8$) successfully induced a lasting improvement in balance (Epro et al., 2018b; Konig et al., 2019b). Additionally, balance ability improvements may be retained at least over 1 month for backward slips in YAs (McCrum et al., 2018), and 1 year for forward slips in OAs (Pai et al., 2014). Although OAs are able to retain the balance improvements from the perturbation training, Konig et al. (2019b) have found that they lose the benefits of the first session quicker than YAs: exposed again to a lab-induced trip 14 weeks following the initial training, OAs' MoS was significantly lower than during the last perturbation of the first session when the YAs did not display this drop. Further work is needed to understand what the optimal perturbation dose (Karamanidis et al., 2020) is, i.e., the threshold above which additional perturbations would not improve the balance further and would trigger long-term retention of balance improvements.

Other considerations that were outside the scope of this study, such as transferability and generalisability of task-specific interventions, should also be investigated. Evidence exists for OAs that an inter-limb transfer of backward slip-like perturbations is possible (McCrum et al., 2020); however, transfer to other mechanical tasks is yet to be investigated. To the knowledge of the authors, generalisation of the benefits from treadmill induced backward slips to overground backward slips has not been investigated yet, but encouraging results on forward slips show that within session and long-term generalisation of the balance improvement following treadmill-induced slips is possible, although not as efficacious as overground-slip training (Liu et al., 2020).

Some limitations exist in this study that should be taken into account. First, we used a fixed walking speed in this study (1.2 $m \cdot s^{-1}$); therefore, because step length and stability (Bhatt et al., 2005) depend on walking speed, caution should be used when extrapolating the results to other walking speeds. However, as OAs have been shown to improve their balance recovery with repeated perturbations in self-selected (Bhatt et al., 2006), fixed (Epro et al., 2018a), and stability-normalised (McCrum et al., 2020) walking speeds, we are confident that the conclusions on balance improvements with repeated backward slip-like perturbations are not limited to this specific speed. Second, we did not find a significant correlation between the changes in kinetic (GRF angle and ankle moment) or temporo-spatial (step length) variables observed during the perturbed step (Pert) and the balance of the first recovery step (Rec1), which is probably even more important than Rec2 for fall avoidance. Therefore, other factors not investigated in the present study, such as participants' ankle plantarflexor and knee extensor muscles' strength and associated tendons' stiffness might be of significant importance in fall avoidance during the first recovery step. This lack of correlation between the mechanics of recovery during Pert and the balance of Rec1 might also be explained by the concomitance of the changes in these kinetic and temporo-spatial variables and the belt acceleration. Whether the observed changes (compared with Normal) are linked to an actual attempt to maintain a stable balance or to the belt acceleration (and therefore centre of pressure displacement) remains unknown. Third, as visual inspection of the moment-time curves did not identify notable changes in the timing of peak moments during the perturbed step or the first recovery step, we did not study the changes in the sequential organisation of the joint moments and how they may have affected the balance on the following steps. However, as the onset of knee moment generation seems to discriminate older fallers from young adults following trips (Pijnappels et al., 2005b), the timing of moment generation should be accounted for in future studies. Fourth, we took great care in recording baseline data during completely normal walking (no lateral or anterior-posterior displacements of the participants were visually observed) and used an average of five steps after a period of familiarisation, which might not be representative of the actual variability of the MoS during Normal condition. Using only five steps to measure the MoS in Normal conditions, we could have overestimated the MoS variability and therefore underestimated the number of participants who did not reach 1SD from Normal balance following the slips (n_{steps}). However, participants' MoS variability ranged from 0.4 to 3.1 cm, which is not larger than the variability reported by McCrum et al. (2020), which ranged from ∼1 to ∼3.5 cm and was measured over 10 consecutive unperturbed steps. This, and recent observations by Fallahtafti et al. (2021) showing that treadmill walking leads to lower MoS variability compared with overground walking, suggest that the within-subjects MoS variability might not have been overestimated but rather underestimated, which may explain the large number of participants who did not return to stable gait by the 15th step in this study ($n = 14$ following the first slip). Careful considerations should be made concerning the number of steps used to determine the MoS variability in future studies, particularly when transferring from treadmill to overground tasks. Lastly, we found that participants did not make anticipatory adjustments in their MoS prior to Slip01, which is consistent with results reported on predictive changes in balance in unexpected perturbations (Okubo et al., 2018), did not anticipate the exact timing of the perturbation (no difference between Pre2 and Pre1 neither for Slip01 nor Slip10), but that following repeated exposures to backward slip-like perturbations, participants developed a more conservative gait pattern (increased MoS in Pre2 and Pre1 of Slip10), which is consistent with previous reports for trips (Wang et al., 2020) and forward slips (Pavol et al., 2004; Heiden et al., 2006; Lawrence et al., 2015). We found significant positive correlations between the MoS in Slip10_Pre2 and Slip10_Pre1 and the MoS of the first recovery step, which suggest, as already demonstrated by Bhatt et al. (2006), that the anticipatory adjustments in balance modulate the reactive ones, and possibly the outcome of the perturbation (fall or recovery). Therefore, the generalisability of our findings to recovery from real-world backward slips, for which there is likely no balance adjustment prior to the actual perturbation, might be dampened. This is a problem for any fall prevention intervention that utilises repeated perturbation exposures. However, the results reported by Pai et al. (2014) are encouraging, as OAs exposed to repeated ($n = 24$) lab-induced forward slips were found to be 2.3 times less likely to fall within a year than those exposed to a single slip.

To summarise, we showed that independent of age, participants improved their balance with repeated exposure to backward slip-like perturbations. We found that the length of the first recovery step following the slip is an important variable for the improvement of balance recovery and was optimised with repeated slips by returning it close to normal levels. As this variable can easily be measured and controlled, instructing OAs to increase their step length when their gait is perturbed may help them recover their balance and potentially avoid falling more effectively.

DATA AVAILABILITY STATEMENT

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Liverpool John Moores University ethics committee National Health Service (NHS) ethics committee (18/NW/0700). The patients/participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

HD, TO'B, and CM: contributed to the conception and design of the research, interpretation of the results, edited and revised the manuscript, and agreed to manuscript submission for publication. HD: data acquisition and data analysis and drafted the manuscript. All authors contributed to the article and approved the submitted version.

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SUPPLEMENTARY MATERIAL

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fspor.2021.708929/full#supplementary-material

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REVIEW
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Developing and Establishing Biomechanical Variables as Risk Biomarkers for Preventable Gait-Related Falls and Assessment of Intervention Effectiveness

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The purpose of this review is to position the emerging clinical promise of validating and implementing biomechanical biomarkers of falls in fall prevention interventions. The review is framed in the desirability of blunting the effects of the rapidly growing population of older adults with regard to the number of falls, their related injuries, and health care costs. We propose that biomechanical risk biomarkers may be derived from systematic study of the responses to treadmill-delivered perturbations to both identify individuals with a risk of specific types of falls, such as trips and slips as well as quantifying the effectiveness of interventions designed to reduce that risk. The review follows the evidence derived using a specific public health approach and the published biomedical literature that supports trunk kinematics as a biomarker as having met many of the criteria for a biomarker for trip-specific falls. Whereas, the efficacy of perturbation training to reduce slip-related falls by older adults appears to have been confirmed, its effectiveness presently remains an open and important question. There is a dearth of data related to the efficacy and effectiveness of perturbation training to reduce falls to the side falls by older adults. At present, efforts to characterize the extent to which perturbation training can reduce falls and translate the approaches to the clinic represents an important research opportunity.

Keywords: injury, intervention, perturbation training, prevention, risk factors

INTRODUCTION

Over 60 years ago, although falls and fall-related injuries among older adults were concerning issues, the published biomedical literature addressing the issues was "very meager" (Sheldon, 1960). This is easily confirmed. A PubMed search (25 May 2021) using the search terms "(falls OR falling) AND (older adults OR elderly) AND (prevention OR intervention)" returned 29,705 titles since 1956. Since 1956 the number of papers published annually has doubled approximately every 6.5 ± 2.7 years. So, the biomedical literature addressing issues related to falls and fall-related injuries among older adults is no longer meager.

Recently, the question was asked "Are we moving the needle on fall prevention?" (Hicks, 2019). The answer, according to the author was a qualified yes; the motion of the needle was said to be in a positive direction. It seems, however, that there is a growing gap between (1) the rate at which the number of older adults, who, by virtue of their age and the relentless effects of biological aging, are at increased risk of falls and injuries, is increasing and (2) the rate at which the needle may be moving and the extent to which it is moving with regard to fall prevention.

We agree that the overall motion of the needle has been in the positive direction. However, the magnitude of its motion is challenging to assess. The full range of possible motion for the needle and its current position within the range have not been, perhaps cannot be, estimated. Nevertheless, referencing Tinetti et al. (1988), Sherrington et al. (2019) wrote that "at least one-third of community dwelling people over age 65 years of age fall each year." In their paper, Tinetti et al. (1988) referenced Prudham and Evans (1981) and Campbell et al. (1981). Thus, based on the persistence of the "one-third" statistic, as general as it may be, one may legitimately wonder about the actual extent to which the needle has been moved in over four decades.

The increase in the population of community dwelling people 65 years of age and older since the early 1980s translates to a large increase in the estimated absolute number of falls. Furthermore, as will be addressed in another section of this paper, the absolute number of falls by older adults, particularly injurious falls, are reasonably expected to increase with the growth of the population of older adults who at risk of falling as well as the number of them who do fall.

It seems reasonable to ask if the effectiveness of fall prevention intervention has reached or is approaching its maximum. The set of 355 US national health and well-being objectives for the coming decade has explicit objectives to both decrease emergency room visits due to falls by older adults and to decrease fall-related deaths by older adults (Healthy People, 2030). This may reflect that there are those who believe there remains room for improvement. Therefore, the question related to whether the effectiveness of fall prevention intervention has reached or is approaching its maximum is quite practical, the long-term impact of which ranges from the personal to the societal levels. The rate at which the older adult population is presently growing will, expectedly, increase the number of falls by older adults that occur annually. This, in turn, will increase the expected number of injurious falls. Again, in turn, this will increase the health- and economic-/financial-related impact of these falls.

The purpose of this review is to address these issues from the point of view that it is both desirable and achievable to slow, to the greatest extent possible, the effects of an already ongoing exponential growth of the population of older adults with regard to falls, particularly injurious falls, and their associated health-care costs. We propose that an effective means by which this may be approached is by greater specificity in fall-prevention interventions. That is, considering the stepping responses that may contribute to avoiding a fall following slips, trips, and falls to the side, as being separate motor skills that may be acquired, or learned, through practice. We will summarize the approach that we have implemented to identify biomechanical variables that are causally related to trip-related falls, clinically-modifiable, and that, following modification, convincingly appear to possess efficacy and effectiveness with regard to decreasing prospectively measured trip related falls by older adults. Collectively, a set of biomechanical variables has emerged that may qualify as risk biomarkers for trip-related falls. This approach is generalizable to other types of falls such as slips and falls to the side, which along with trip-related falls, share a common feature. The common feature is that in many cases, a fall may be prevented by a temporally and spatially appropriate stepping response.

When we set out to answer questions about the biomechanical causes of gait-related falls by older adults it was with the long-term goal of reducing the incidence of these falls, specifically trip-related falls. A key motivation was the belief that once a trip had occurred, the trip-related fall, or at least some of them, could be prevented by performing a spatially and temporally appropriate compensatory strategy, specifically a stepping response. We reasoned that identifying biomechanical variables that could be shown to be causally associated with failed stepping responses and, importantly, were clinically modifiable, could be followed by determining the extent to which clinically modifying these variables would decrease the prospectively measured incidence of a trip-related falls by older adults. Decreasing the incidence of trip-related falls by older adults would consequently lead to fewer trip-related injuries.

We first described the biomechanical differences between successful and failed recoveries by older adults following a laboratory-induced trip (Pavol et al., 2001). In a separate paper, we described an experimental protocol in which a treadmill was used to deliver a perturbation to a standing subject and that required a forward-directed step to avoid falling (Owings et al., 2001). Failed recovery efforts by older adults following these perturbations shared general biomechanical mechanisms of failed recoveries following a laboratory-induced trip during overground walking (Pavol et al., 2001). Although this perturbation did not induce a trip as seen during overground walking by obstructing the forward progression of the swing leg, it seemed to be a reasonable, and practical model for a protocol that is not clinically-friendly. A seminal experimental result was that exposure to a single, treadmill-delivered, trip-specific perturbation was associated with changes to key biomechanical variables that led to a successful recovery for 75% of women following subsequent within-session trip-specific perturbations (Owings et al., 2001). These trip-specific perturbations were delivered as participants stood on the treadmill belt and, thus, do not include an actual obstruction of the forward motion of the swing limb as would be associated with a tripping event. Rather, it is the biomechanics of the initial recovery step, the trunk and of the weight-acceptance phase of the recovery that are "trip-specific." This finding was consistent with predictions in accordance with the specificity of training principle (Henry, 1968). This principle holds, that "motor skill performance during a 'test' condition is enhanced when both the sensorimotor and environmental contexts of the 'practice' and test conditions are similar" (Grabiner et al., 2014). Collectively, our early hypotheses and results contributed to a conceptual

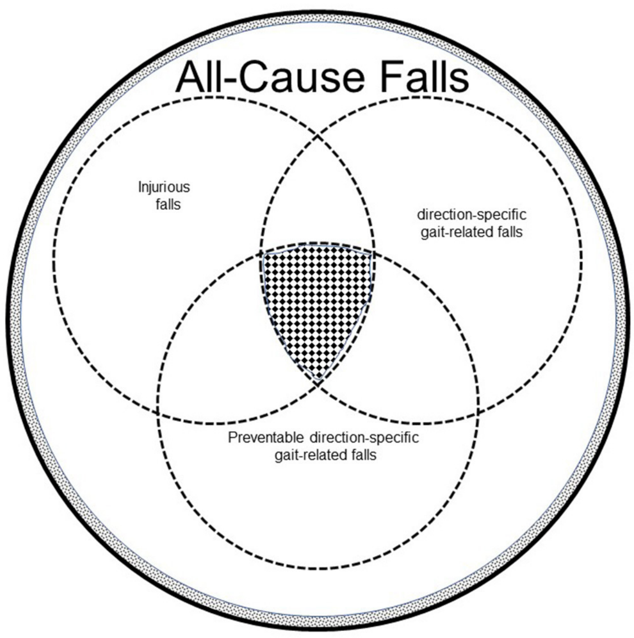

FIGURE 1 | The total encircled area represents all-cause falls by all older adults. The white area represents all-cause falls by independent community dwelling older adults, who represent 95% of older adults in the United States (Administration on Aging, 2012, http://www.aoa.gov/Aging_Statistics/Profile/2012/docs/2012profile.pdf). The dotted area represents all-cause falls by older adults living in assisted living facilities. Direction-specific gait-related falls are a subset of all-cause falls. Slips and trips, direction-specific falls, that occur both in and outside the home, may generally account for about 50% gait-related falls by independent community-dwelling older adults (Berg et al., 1997; Crenshaw et al., 2017). A substantial number of falls by independent community-dwelling older adults are laterally-directed (Crenshaw et al., 2017). Because laterally-directed falls may have recovery solutions that involve stepping responses, the percentage of preventable direction-specific gait-related falls is increased. Of these gait-related falls, some proportion leads to injury, both minor and serious, the latter of which can be life-threatening. Of these direction-specific gait-related falls by independent community-dwelling older adults, some are demonstrably preventable through various clinical interventions. The hatched area at the intersection of injurious falls, direction-specific gait-related falls. and preventable, direction-specific gait-related falls represents out target for perturbation-based intervention.

representation of our long-term goal of reducing the incidence of trip-related falls (**Figure 1**).

Overall, the experimental approach that informed our subsequent approach to fall prevention intervention was that described in a paper titled "Public Health Model of a Scientific Approach to Prevention" (Mercy et al., 1993). Notably, although this approach was presented in a context related to violence prevention, the very same model was presented 17 years later in a paper titled "An older adults research agenda from a public health perspective" (Stevens et al., 2010). The basis of the model was a four element pathway, the first of which is comprehensive description of the problem(s) and its/their implication(s), including both of the problem being successfully or not successfully addressed. The second element involves the identification of risk factors and causes, particularly causes that are modifiable. The third element is the implementation of the results from the second element to design interventions, target interventions, and evaluate the efficacy and effectiveness of the interventions. Here, efficacy reflects how well an intervention works under "ideal and controlled circumstances" such as those found under laboratory conditions. In contrast, effectiveness reflects how well the intervention works in the more messy "real world" (Singal et al., 2014). Lastly, the demonstrably effective intervention and implemented. These are topics to which we shall return.

In particular, we advocate here a transition toward the clinical inclusion of valid biomechanical biomarkers for falls with validated risk factors for falls. Generally, a risk factor is a characteristic or an exposure that increases the probability of a negative outcome. On the other hand, a biomarker is an objective and quantitative measure of biological structure, function, and/or process, the purpose of which is to evaluate health status and/or response(s) to intervention(s) (Micheel and Ball, 2010). There is a distinction, and potentially meaningful difference between risk factors and biomarkers. By definition, although biomarkers may be shown to be risk factors, risk factors are not necessarily biomarkers.

Fall-risk assessment has historically consisted of assessing the presence of risk factors. Thus, in the present context, a risk factor may be considered an attribute, characteristic, or exposure that increases the probability of a fall. Common fall risk factors include muscle weakness, gait deficits, and balance deficits (Rubenstein and Josephson, 2002). Each of these risk factors can be, and are, measured in numerous ways. In addition, each may be, clinically modifiable via exercise and, in some cases have been shown to be associated with falls, fall rate, and fall-related injury. Other risk factors that are associated with of falls but that are not modifiable include one's sex, age, and history of previous falls.

BIOMARKERS FOR FALLS

There has been an increase in the frequency with which the terms "fall" and "biomarker" have been conjoined in the published literature. Gait-related biomechanical variables have been suggested as potential biomarkers for Parkinson's disease (Horak and Mancini, 2013), dementia (Montero-Odasso et al., 2017), cognitive impairment (Beauchet et al., 2013), Alzheimer's Disease (Varma et al., 2021) osteoarthritis (Mezghani et al., 2017), and ALS (Inam et al., 2010). Further, the term biomarker has been used in the context of Parkinson's disease-related falls; (Brodie et al., 2014). Interestingly, serum levels of malondealdehyde, a biomarker for oxidative stress, was reported to be prospectively associated with falls (Verghese and Ayers, 2017). Based on this result the authors recommended malandealdehyde for further study, "as a fall *risk biomarker*" (italics added for emphasis) and as "a potential target to prevent falls." The term risk biomarker has a very specific connotation.

There are eight recognized categories of biomarkers, the uses of which include, for example, diagnosis, prognosis, exposure, surveillance, and risk/susceptibility (FDA-NIH Biomarker Working Group, 2016). Risk biomarkers, aka susceptibility biomarkers, indicate the potential for developing a disease or medical condition in an individual for whom clinically apparent disease or medical condition is absent (FDA-NIH Biomarker Working Group, 2016). Examples given are women who inherit the BRCA1 and BRCA2 genes, and who are at significantly higher lifetime risk of breast and ovarian cancer, and the APOEe4 gene that increases the risk of Alzheimer's disease. In these examples of risk biomarkers, a specific gene is attributed as the cause for the clinical outcome and the associated disease risk present from the time of fertilization, decades before clinical manifestations of the diseases.

General requirements for clinical biomarkers include their capacity to be accurately measured (analytical validity), to accurately estimate structural/functional status (clinical validity), having an acceptable risk-benefit ratio (clinical utility), being clinically feasible (i.e., being time-effective and cost-effective), and being non-invasive (Ferlini et al., 2013). With respect to biomechanical biomarkers for falls, we propose three additional "ideal" criteria. First, a biomechanical biomarker for falls should ideally be established as causally related to falls. This criterion is based on the notion that the clinical benefit of monitoring a biomarker that bears no, or little influence on the outcome, in this case falls, may have a high cost:benefit ratio. Secondly, a biomechanical biomarker for falls should ideally be clinically modifiable. This criterion is based on the notion that the clinical benefit of monitoring a biomarker that is not sensitive to some type of intervention, will not influence the outcome, i.e., falls, and consequently have a high cost:benefit ratio. Thus, following a targeted intervention, a biomarker established as causally related to falls should be altered and this change in the biomarker should be accompanied by an appropriate directional change in fall risk. Thirdly, a biomechanical biomarker for falls should ideally demonstrate direction specificity. That is, the biomarker should reflect the risk for a specific type of fall, for example a slip as opposed to a trip, to which a person is susceptible in contrast to an any-cause fall. We consider specificity to be instrumental to the prescription of targeted, fall-specific prevention interventions.

Biomarker specificity addresses a key limitation of both currently acknowledged risk factors for falls and variables that have previously been suggested as potential biomarkers for falls. The limitation is that the risk factors and proposed biomarkers are associated with *all-cause* fall risk. Aside from slips, trips, and falls to the side, all cause falls also include those that occur while ascending or descending stairs, falls from bed, chairs, and ladders, falls due to carrying external loads, collisions, and using mobility assistance devices. In addition, there is also the ambiguous, if not necessary, fall type that include "other", "non-classifiable," and "unknown" that can often account for meaningful percentages of the total number retrospectively and prospectively reported falls (Talbot et al., 2005; Pai et al., 2014b).

Presently, our focus is on biomarkers of "direction-specific, preventable gait-related falls" by independent community-dwelling adults. By direction specific, we distinguish between trips that cause anteriorly-directed falls, slips that cause posteriorly-directed falls, and falls to the side. We operationally define a preventable gait-related fall as one that occurs by a community-dwelling adult while in the home or in the community during conditions in which the external and/or internal (prevailing physiological conditions) environments do not limit the initiation and completion of a compensatory-stepping response. To this last point, following a loss of balance that, unless corrected, would inevitably lead to a fall, the physical environment in which the loss of balance occurs can exert a substantial effect on the probability of performing a temporally and spatially sufficient stepping response to avoid a fall. If a particular environment shortens either the time available to perform a stepping response or decreases the physical space that would accommodate the stepping response, then the probability of successfully performing the stepping response is diminished. Similarly, a sudden change in blood pressure, onset of dizziness, or a change in vision could render a loss of balance unrecoverable. We offer five premises in support of our argument that discovery of biomarkers of "direction-specific, preventable gait-related falls" can increase the effectiveness of fall-prevention interventions.

Premise 1

From a practical perspective, some gait-related falls by older adults are preventable and some are not preventable. An important characteristic of preventable gait-related fall is that neither the internal nor external environments preclude or limit the execution of a successful stepping response. Rather, it is the functional capacity of an individual to initiate and complete successful stepping response being exceeded by the requirements imposed by external and/or internal environments. Improved functional capacity is the focus of a fall-prevention intervention.

Premise 2

Different types of preventable gait-related falls by older adults may not be equally preventable. This premise remains to be experimentally confirmed or refuted. However, for example, once initiated, avoiding a slip-related fall appears to be more challenging for older adults than avoiding trip-related fall. Following a laboratory-induced trip, 100% young adults were able to avoid a fall (Grabiner et al., 1993, 1996). Under similar conditions, following a laboratory-induced slip using a slippery surface mimicking ice, 74% of older adults were able to avoid a fall (Pavol et al., 2001; Grabiner et al., 2012). However, following a laboratory-induced slip on a slippery surface, 86% of young adults were able to avoid a fall, but only 14% of older adults were able to do so (Troy et al., 2008). Neural and biomechanical explanations for the apparent greater difficulty by older adults to avoid a slip-related fall compared to a trip-related fall have been proposed (Grabiner et al., 2014).

Premise 3

The potential for interventions to decrease the incidence of preventable gait-related falls by older adults may not be similar for different fall types. This premise remains to be experimentally confirmed or refuted. The rationale for this premise follows logic similar to that for slip-specific and trip-specific falls noted in Premise 2. Aging-related degradation of the neuromuscular and musculoskeletal systems that includes, but is not limited to muscle strength, muscle power, flexibility, coordination, and response time, are normal and expected although there is substantial between-person variability. Collectively, these aging-related changes render the recovery solution for slip-related falls an increasingly challenging motor skill for older adults, in general, to perform, acquire, retain, and generalize/transfer the skill to different tasks having similar performance attributes (e.g., König et al., 2019). Therefore, the deleterious aging-related changes to the motor fitness may require an intervention approach requiring longer intervention time and/or larger exposure to slow or reverse.

Premise 4

Certain types of preventable gait-related falls may not be as prevalent as others. Most of the data regarding the prevalence of fall types has been acquired retrospectively and has the limitations that come with that type of reporting. Nevertheless, slips and trips combined generally represent the largest reported proportion of gait-related falls by older adults. Prospectively-measured trip-related falls appear to occur more frequently than slip-related falls by community-dwelling adults (Berg et al., 1997; Hill et al., 1999; Decullier et al., 2010; Pai et al., 2014b; Crenshaw et al., 2017), by older adults in long-term care (Robinovitch et al., 2013), and older adults with intellectual disabilities (Smulders et al., 2013). Limited evidence suggests that gait-related falls that occur to the side may occur at rates that approach those of trips and slips (Stevens et al., 2014; Crenshaw et al., 2017). Similar to trip-related and slip-related falls, many falls that occur to the side may potentially be preventable by executing a temporally and spatially appropriate compensatory stepping response. Indeed, laterally-directed stepping responses have been shown to be improved by task-specific training, at least for young adults (Hurt et al., 2011).

Premise 5

Preventable gait-related falls have stepping response solutions, which as motor skills, may be practiced, acquired, and retained. Our definition of a preventable gait-related fall specifies that the external and/or internal environments do not limit the initiation and completion of a stepping response.

In addition to the above premises, our focus on biomarkers of "direction-specific preventable gait-related falls" by independent community-dwelling adults reflects the growing magnitude of fall-related issues, the range of which spans from the individual older adults and their families to societal. In 1960, there were approximately 16.6 million adults 65 years of age and older in the United States (U.S. Bureau of the Census, 1996). In 2012 this number had more than doubled to about 43 million, is expected to double once again by 2050, and further increase to 98 million by 2060 (Colby and Ortman, 2014). In 2012 there were 3.2 million non-fatal falls by older adults reported in the US. Given that about 25% of older adults are estimated to fall annually (Bergen et al., 2016; Centers for Disease Control Prevention, 2018) the number of falls is under-reported. Of the reported non-fatal falls in 2012, 18% required hospitalization, 55% involved an emergency department visit, and 27% led to office-based/outpatient visits (Burns and Stevens, 2016). The estimated total medical-related cost of these non-fatal, reported falls in 2012 was $30.3 billion.

Based on the 2012 statistics above, and assuming that the percentage of serious fall-related injuries and/or costs of medical care for these injuries will not change appreciably, the costs for treating 6.5 million reported non-fatal falls by older adults will logically increase proportionately to the population by 49% to $45 billion. Overall, at the societal level, the costs to the government of healthcare are the focus of a great deal of political energy. That said, at the level of individual older adults and their families, the specter of fall-related morbidity including the loss of mobility and independence, mortality, and financial burden represent a source of everyday concern for many, if not most older adults.

Some fall-related injuries, such as hip fracture, head injury, and spine injury impose significant morbidity and post-injury mortality. For example, 20% of hip fracture survivors require long-term medical care. Notably, in the US, Medicare will not pay for long-term medical care and <15% of older adults in the US have long-term care insurance (Tajeu et al., 2014). Ignoring, for the moment, the expectation that the Medicare Hospital Insurance Trust Fund has been projected to become depleted by about 2030 (Centers for Medicare Medicaid Services, 2019), for many, if not most older adults, personal retirement savings is insufficient to bridge the gap between the total medical costs of a fall-related injury and that provided by Medicare coverage. Therefore, a reasonable potential solution, albeit partial, to this is to prevent the injurious fall from occurring in the first place.

The above arithmetic is compelling. If the expected incidence of falls by older adults cannot be decreased beyond that which is currently possible, then the growth of the population of older adults assures that the number of falls and the corresponding personal and societal problems will increase proportionately, at least through the end of this century. This, then, represents a growing gap that is important to narrow if possible. We propose that the gap may be narrowed to some extent by three means. The first is to increase the percentage of older adults who engage is exercise. The second is to increase the effectiveness of clinical assessment of fall risk. The third is to increase the effectiveness of fall-prevention interventions. All three can be addressed by focusing attention on falls that are demonstrably preventable.

CURRENTLY USED FALL-RISK ASSESSMENTS HAVE LIMITED UTILITY

The effectiveness of clinical tests that are commonly used to assess fall risk is dependent upon the population for which the assessment is desired. When applied to independent community-dwelling older adults, who represent more than 95% of the entire population of older adults (Institute on Aging; Ortman et al., 2014), systematic reviews and meta-analyses have consistently reported that included studies were of low to moderate quality, the sensitivity of many presently used clinical fall risk assessment methods is low, sometimes unacceptably low, and, consequently did not warrant the authors recommendations (Scott et al., 2007; Gates et al., 2008; Muir et al., 2010; Beauchet et al., 2011; Rydwik et al., 2011; Schoene et al., 2013; Omaña et al., 2021).

EXERCISE-BASED INTERVENTIONS TO REDUCE FALL RISK ARE EFFECTIVE BUT PARTICIPATION RATES BY OLDER ADULTS ARE LOW

Fall prevention interventions are effective at reducing falls and exercise may be the most effective fall prevention intervention. Exercise decreases fall incidence by about 23% and decreases fall rate between 17 and 29% (Sherrington et al., 2019). Of the falls that occur and are reported, exercise decreases injurious falls by about 50% (Tricco et al., 2017). However, the participation rate by older adults in exercise is low. Based on the US guidelines for leisure time activity, fewer than 30% of adults older than 75 years of age meet the aerobic activity guidelines (Clarke et al., 2017). Fewer than 9% of adults older than 75 years of age meet the guidelines for aerobic plus muscle strengthening activities (Clarke et al., 2017). Thus, we suggest that exercise, the most robust fall-prevention intervention, likely falls quite short of its potential. *However, increased participation rate in exercise programs by older adults, in addition to increased adherence to the programs, in conjunction with increased effectiveness, that is the specificity of exercise to target fall prevention, represent achievable targets that could effectively decrease the incidence of falls.*

THE CASE FOR POTENTIAL BIOMECHANICAL BIOMARKERS FOR FALLS AND A PATH FORWARD

Establishing biomechanical biomarkers of preventable, gait-related, and cause-specific falls is achievable. The extent to which a biomarker for falls qualifies as causal can be evaluated using the Bradford Hill criteria (Hill, 1965; Aronson, 2005). These criteria are possibly the most frequently cited framework for inferring causality (Fedak et al., 2015). The nine criteria are strength of association, consistency, specificity, temporality, biological gradient (dose-response), plausibility, coherence, experiment, and analogy. Accumulating evidence convincingly demonstrates that sagittal plane trunk kinematics of older adults following trip-specific treadmill-delivered disturbances and following laboratory-induced trips meet many of the Hill criteria.

The first criterion, strength of association, is met by sagittal plane trunk kinematics following a laboratory-induced trip. Sagittal plane trunk kinematics are positively, strongly, statistically significantly, and repeatably related to the outcome (fall or recovery) of a laboratory-induced trip. The positive association between trunk flexion angle, measured at the recovery step completion (the instant at which the foot of the recovery step limb contacts the ground) and trip-related falls was demonstrated when young adults and older adults were subjected to laboratory-induced trips (Grabiner et al., 1993, 1996, 2012, 2014; Pavol et al., 2001).

Following a laboratory-induced trip the difference between the trunk flexion angle at recovery step completion of older adults who did not fall and that of those who did fall was significant (Pavol et al., 2001; Grabiner et al., 2012). The 95% confidence intervals for trunk flexion angle at recovery step completion for those who fell and for those who did not fall were (37.3–41.3°) and (20.4–26.4°), respectively. Further, the 95% confidence interval for trunk flexion angle at recovery step completion of the young adults, who did not fall, overlapped with that of the older adults who did not fall (23.3–31.5°). Collectively, these results are also consistent with the fifth Hill criterion of biological gradient, or dose-response. This is similarly so for the trunk velocity at recovery step completion. The 95% confidence intervals for the older adults who did not fall and that of those who did fall were −35 to −14°/s (the negative sign indicates trunk extension velocity) and 36–72°/s (the positive values indicate trunk flexion velocity), respectively (Pavol et al., 2001; Grabiner et al., 2014). The extension velocity of the trunk at recovery step completion means that these older adults were able to both arrest and reverse the trunk velocity prior to contact of the recovery foot with the ground.

The second Hill criterion, consistency, refers to the extent that the association has been reported by different laboratories and/or for different populations. In addition to healthy, community-dwelling older adults, similar relationships have been reported between falls following trip-specific treadmill-delivered perturbations and trunk kinematics for people with unilateral, bilateral, below-knee, and above-knee lower extremity amputation (Kaufman et al., 2014), for women with knee joint osteoarthritis (Pater et al., 2016; Foucher et al., 2020), and post-stroke patients (Honeycutt et al., 2016). However, it warrants noting that recently, a protocol consisting of both trip-specific and-slip specific training was reported to not have had an effect on trunk flexion angle by older adults at recovery step completion following a laboratory-induced trip (Allin et al., 2020).

The third criterion, specificity, restricts the association to an explicit outcome. Trip-specific training appears to decrease prospectively measured falls by middle age and older women following laboratory-induced trips during overground walking (Grabiner et al., 2012). Subsequent to that work, trip-specific training appears to have decreased prospectively measured trip-specific fall rate over a 12 month period by middle age and older women by about 50% (Rosenblatt et al., 2013). Notably, the trip-specific training was not associated with a change in the rate of falls due to causes other than trips. To date, the results of Grabiner et al. (2012) and Rosenblatt et al. (2013) have not been subjected to replication studies. However, in another study, young and older adults were subjected to both forward and backward platform-delivered perturbations. The stepping responses of older adults were improved in both directions. However, the stepping responses of young adults improved only in the forward direction. Importantly, in the present context, neither the older nor younger adults improved following laterally-directed perturbations (Dijkstra et al., 2015). Recently, trip-specific training was reported as superior to Tai Chi training with regard to reducing trunk flexion following treadmill-delivered trip-specific disturbances (Avils et al., 2019). This may be the first published, direct comparison of the efficacy of trip-specific training to Tai Chi, which has become a standard fall-prevention intervention (Stevens and Burns, 2015). Collectively, the published literature suggests that the effects of trip-specific perturbation training are, indeed, specific to trip-related falls.

The fourth criterion, temporality, specifies that the effect cannot occur before the cause. Trunk flexion velocity at recovery step completion, discussed in a previous section on the first criterion, fulfills this criterion. In the present context, trunk flexion velocity, in particular, is viewed as a biomechanical cause and the fall is the effect. Similar to trunk angle, the trunk angular velocity is measured at recovery step completion. This means that those older adults who did not fall were able to *arrest* and *reverse* the trip-induced trunk flexion velocity prior to recovery step completion. In contrast, the trunk velocity of those who fell was in the direction of flexion. A similar finding was reported from an experiment in which laboratory-induced trips were delivered to middle age and older women who had participated in a trip-specific perturbation protocol or who had served in the control group (Grabiner et al., 2012). This also addresses the criterion of consistency.

The fifth criterion is biological gradient, or dose-response. This criterion appears to be met by the extent to which increased trunk flexion angle and direction of trunk velocity are related to the probability of a fall. Following a laboratory-induced trip, at recovery step completion, the trunk flexion angle of the women who fell was $37 \pm 5°$ and the trunk velocity was $40 \pm 41°/s$ (flexion). In contrast, the trunk flexion angle of the women who did not fall was $22 \pm 13°$ and the trunk velocity was $-13 \pm 44°/s$ (extension). The between-group differences for trunk flexion angle and trunk velocity were significant. Interestingly, the recovery step length of women who fell was significantly shorter than that of women who did not fall (28.4 ± 19.3% body height vs. 58.2 ± 16.9% body height, respectively. This result was consistent with a suggestion that, in addition to decreasing dynamic stability, increased trunk flexion following a trip would interfere with performance of the recovery step through the action of antagonistic biarticular muscles (Grabiner et al., 1993).

The remaining four Hill criteria include plausibility, coherence, experiment, and analogy. The biomechanical underpinnings of the causal biomechanical roles of trunk flexion angle and trunk angular velocity are plausible, given that that the head, arms, and trunk represent about 50% of the total body mass. The demonstrable and expected age-related decrease in trunk muscle strength and muscle power would reasonably be predicted to require more time and/or distance, i.e., range of motion, to arrest and reverse the motion of this mass caused by the trip. Coherence refers to the extent to which the association is consistent with generally available knowledge. Hill's criterion of experiment relates to the evidentiary strength for claims of causality of experimentally derived results when supported by the established association. Lastly, using the criterion of analogy, Hill suggested that, in some cases, having established a variable as causally related to an outcome may lower the standard of evidence for subsequent variables if they are, in some way, related to those previously established.

In the aggregate, the established relationship between trunk kinematics following trip-specific perturbations and the subsequent success or failure of the recovery efforts appear to meet many of the requisite cause-effect characteristics defined by Hill (1965). In addition, trunk kinematics following a trip-specific perturbation appear to meet many of the general requirements for biomarkers. The kinematic variables may be accurately measured, they are clinically feasible and are non-invasive. Furthermore, they are clinically modifiable and direction specific. However, there are criticisms of perturbation-based fall prevention. A frequently raised criticism is the scalability and cost-effectiveness that reflect the cost and availability of the required instrumentation. This is a legitimate concern although one could argue that the cost-benefit is reasonable, especially if the issue of "moving the needle" with respect to fall prevention is of sufficient importance to reconsider the resources to be invested in the necessary infrastructure. Further, the previously described limitations of existing clinical tests of fall risk and the very low rates of exercise participation by older adults must be recognized as elements in the equation. Another frequently raised criticism related to perturbation-based fall prevention interventions has been directed at one of its strengths, that is, its specificity. Trip-specific perturbation training appears to decrease prospectively measured falls following laboratory-induced trips (Grabiner et al., 2014) and prospectively measured trip-related fall rate in the community (Rosenblatt et al., 2013). It warrants mention that the study design of Rosenblatt et al. (2013) was designed to be powered, a priori, to detect a 50% reduction in all-cause falls and not trip-specific fall rate. Nevertheless, trip-specific perturbation training did not appear to have an effect on other types of falls. To date, the experiment of Rosenblatt et al. (2013) has not had the benefit of being repeated. Nevertheless, the implication is that trip-specific protocols may have to be conducted separately from slip-specific protocols and protocols directed at reducing falls to the side. This, of course increases the time demands of the protocols. However, prior to crossing that bridge it seems reasonable to first demonstrate if perturbation-based interventions are actually effective in reducing the incidence of these type of falls in the community.

It is abundantly clear from published data that trunk kinematics improve as a result of trip-specific perturbation training and the published literature points to this improvement being causally associated with decreased probability of a trip-specific fall. What remains to be done, from our point of view, is to systematically determine if the trunk biomarkers measured on an individual who has not participated in an intervention, general, or trip-specific, can prospectively predict trip-specific falls by that individual. This will prove challenging. First, exposure to even a single trip-specific perturbation is sufficient to induce a significant improvement in the performance of the recovery of a second perturbation (Owings et al., 2001). Secondly, prior knowledge of an imminent perturbation, such as that disclosed during the informed consent, is sufficient to induce a significant improvement in the performance of the recovery, at least for young adults (Oludare et al., 2018). Challenging as it may be, the clinical value of having a subject-specific baseline record, in the form of biomarkers for trip-specific falls, would allow quantitatively informed opinions regarding the effectiveness of an intervention, general or trip-specific, to reduce trip-specific fall risk. In addition, such a baseline and follow-up records could contribute to a clinical decision regarding when, or if, the maximum effect of an intervention had been achieved for an individual.

We previously described how our work related to trip-specific falls followed the model originally proposed by Mercy et al. (1993). In light of the reasonably positive outcomes of having done so, we propose that, moving forward, this same approach may be useful to consider and apply to other types of potentially preventable gait related falls, particularly slip-related falls, and falls that occur to the side. Both of these types of falls share a common biomechanical characteristic critical to the recovery solution(s). The common characteristic is that successfully avoiding a fall is reliant, to a large extent, on the ability to execute an appropriate stepping response within the available time and space.

Slip-Related Falls

There is ample evidence to support the claim of efficacy of slip-specific perturbation training. That is, slip-specific perturbation training has repeatably been reported to decrease falls following platform-induced slips in the laboratory by older adults (e.g., Pai et al., 2010, 2014a; Lee et al., 2018; Okubo et al., 2019; Wang et al., 2019; Allin et al., 2020).

Thus far, though, the effectiveness of slip-specific perturbation training to reduce slip-related falls in the community is not known. To our knowledge there may be only one published study reporting the effects of slip-specific training on prospectively measured slip-specific falls (Pai et al., 2014b). In this study, 67 community-dwelling older adults were subjected to a series of 24 slip-specific perturbations using a sliding platform. A control group of 75 community-dwelling older adults were subjected to a single slip-specific perturbation. Subsequently, all-cause falls, of which there were five types, were prospectively assessed for 12 months. The fall types were slips, trips, those that were caused by ADL and transfers, those that were caused by external hazards, and those for which the causes were categorized as "others/unknown." The between-group difference for all-cause falls, i.e., inclusive of all fall types, achieved significance. However, the study was not designed to directly compare between-group difference for slip-related falls (20 vs. 16.7% of the total number of falls for the trained and control groups, respectively, for the on-treatment analysis; 28.6 and 15.4% of the total number of falls for the trained and control groups, respectively, for the intention-to-treat analysis). It is possible that the significant main effect of the perturbation training may have been influenced by the number of falls that were categorized as "other" or "unknown." In this category of falls, the percentages of the total number of falls for the control group were more than two and six times that of the training group for the on-treatment analysis and intention-to-treat analysis, respectively. Nevertheless, based only on the results of this single experiment which, similar to the study of Rosenblatt et al. (2013), has not had the benefit of being repeated, a conclusion related to the effectiveness of slip-specific perturbation training on the incidence of slip-related falls in the community by older adults, is premature. That said, there is a trial currently being conducted, but for which the results do not yet appear to have been disseminated, that is focused on the effects of slip- and trip-specific perturbation training on the prospectively measured slip- and trip-specific falls (Rieger et al., 2020).

Falls to the Side

At present and to our knowledge, there is no published research related to the use of perturbation training to decrease the incidence *laterally-directed* falls either in the laboratory or in the community. Falls to the side, which may account for up to 33% of falls by older adults (Crenshaw et al., 2017), are a particular concern because of their association with hip fracture. Hip fracture is the cause of a high level of morbidity, mortality, debility, and destitution in the population of older adults (Tajeu et al., 2014). However, in the light of the framework for evaluating causality, it is notable that the ability to perform laterally-directed stepping responses following waist-pull perturbations has been associated with prospectively measured all-cause falls, by older adults (Hilliard et al., 2008; Mille et al., 2013; Rogers et al., 2021). For young adults, laterally-directed stepping following treadmill-delivered disturbances is improved by practice (e.g., Hurt et al., 2011). In another experiment, young adults walking on a treadmill were subjected to direction-specific waist-pull perturbations. The quality of the stepping responses was measured as the margin of stability (Hof et al., 2005) and improved in the anterior-posterior direction, but not the medial-lateral direction (Martelli et al., 2017). In the aggregate, although the ability to perform stepping responses following laterally-directed perturbations appear to be clinically modifiable, at present there does not appear to be established risk factors or biomechanical biomarkers for laterally-directed falls.

The rate of growth of the population of older adults, for which a slowing has not been predicted, has outpaced the ability of currently used clinical interventions and general exercise participation by older adults to reduce the incidence of falls by older adults. The ensuant health and health care issues, which cannot be easily separated from political and economic issues, will likely continue to grow in parallel. We have tried to make the case for systematically establishing biomarkers for falls by older adults. We have proposed that a family of biomechanical biomarkers for direction-specific and gait-related falls can increase the effectiveness of fall prevention interventions and, potentially, serve as indices for the extent to which an intervention has achieved or is nearing its maximum effect for a specific individual. In this case, the maximum effect is with regard to preventable, direction-specific, gait-related falls by older adults. Presently, perturbation training directed toward decreasing the incidence of trip-related falls by older adults has the largest body of supportive evidence supporting both its efficacy and effectiveness. The efficacy of perturbation training to reduce slip-related falls by older adults appears to have been confirmed although its effectiveness of to reduce slip-related falls by older adults is an open and important question. Both the efficacy and effectiveness of perturbation training to reduce falls to the side falls by older adults, understudied and reported, represents an important research opportunity. In light of the exponential nature of the growth of both the population of older adults and the incidence of falls, particularly injurious falls, the present seems to be a perfect time to at least consider the possibility that we can do better at refining and implementing the state-of-the art with regard to fall prevention.

A key challenge to the design of prospective experiments that will be necessary to convincingly confirm or refute efficacy and effectiveness of perturbation training to reduce direction-specific falls by older adults to is the estimated sample size. Sample sizes will necessarily be much larger to ensure *a priori* statistical power to detect between group-differences in the incidence of falls due to direction-specific types of falls such as trips, slips and falls to the side compared to all-cause falls (Karamanidis et al., 2020). This translates to higher costs and more time to conducting such studies. We support the recommendation for a solution involving large, collaborative efforts (Karamanidis et al., 2020). Such an collaborative team approach has the potential to link international research and clinical facilities, scientists and clinicians, thereby serving as a force multiplier for shared and increasingly important interests.

AUTHOR CONTRIBUTIONS

Both authors contributed equally to the drafting and revising of this manuscript.

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The Stepping Threshold Test for Reactive Balance: Validation of Two Observer-Based Evaluation Strategies to Assess Stepping Behavior in Fall-Prone Older Adults

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Introduction: Measurement of reactive balance is critical for fall prevention but is severely underrepresented in the clinical setting due to the lack of valid assessments. The Stepping Threshold Test (STT) is a newly developed instrumented test for reactive balance on a movable platform, however, it has not yet been validated for fall-prone older adults. Furthermore, different schemes of observer-based evaluation seem possible. The aim of this study was to investigate validity with respect to fall risk, interpretability, and feasibility of the STT using two different evaluation strategies.

Methods: This study involved 71 fall-prone older adults (aged ≥ 65) who underwent progressively increasing perturbations in four directions for the STT. Single and multiple-step thresholds for each perturbation direction were determined *via* two observer-based evaluation schemes, which are the 1) consideration of all steps (all-step-count evaluation, ACE) and 2) consideration of those steps that extend the base of support in the direction of perturbation (direction-sensitive evaluation, DSE). Established balance measures including global (Brief Balance Evaluations Systems Test, BriefBEST), proactive (Timed Up and Go, TUG), and static balance (8-level balance scale, 8LBS), as well as fear of falling (Short Falls Efficacy Scale—International, FES-I) and fall occurrence in the past year, served as reference measurements.

Results: The sum scores of STT correlated moderately with the BriefBEST (ACE: $r = 0.413$; DSE: $r = 0.388$) and TUG (ACE: $r = -0.379$; DSE: $r = -0.435$) and low with the 8LBS (ACE: $r = 0.173$; DSE: $r = 0.246$) and Short FES-I (ACE: $r = -0.108$; DSE: $r = -0.104$). The sum scores did not distinguish between fallers and non-fallers. No floor/ceiling effects occurred for the STT sum score, but these effects occurred for specific STT thresholds for both ACE (mean floor effect = 13.04%, SD = 19.35%; mean ceiling effect = 4.29%, SD = 7.75%) and DSE (mean floor effect = 7.86%, SD = 15.23%; mean ceiling effect = 21.07%, SD = 26.08). No severe adverse events occurred.

Discussion: Correlations between the STT and other balance tests were in the expected magnitude, indicating convergent validity. However, the STT could not distinguish between fallers and non-fallers, referring to a need for further studies and prospective surveys of falls to validate the STT. Current results did not allow a definitive judgment on the advantage of using ACE or DSE. Study results represented a step toward a reactive balance assessment application in a clinical setting.

Keywords: reactive balance, assessment, step threshold, perturbation, validity, fall prevention, fallers

INTRODUCTION

Approximately every third person aged 65 and above experiences at least one fall annually (World Health Organization, 2007). Early detection of individuals at high risk for falls could help prevent falls and reduce health care costs. The most commonly used measurements to detect impairments in postural control are measures of static and dynamic balance (Sibley et al., 2011), such as the single-leg stance, the Berg Balance Scale (Berg, 1989), and the Timed Up and Go (TUG) Test (Podsiadlo and Richardson, 1991). However, neither these nor other fall risk assessments demonstrate sufficient ability to distinguish between older adults at high and low risk for falls (Balasubramanian et al., 2015; Lusardi et al., 2017; Park, 2018).

Reactive balance control, which is the ability to recover from an unexpected loss of balance, is a critical component of postural control for fall prevention (McIlroy and Maki, 1996). At the same time, reactive balance is the least assessed component of postural control in the clinical setting (Sibley et al., 2011). In a cross-sectional survey by Sibley et al. (2013), nearly 80% of the clinicians, who reported to assess reactive balance, used only non-standardized observation-based methods of assessing reactive control. Even clinicians who rely on standardized tools must cope with severe limitations. Validated tools, such as the Balance Evaluation Systems Test (BEST) (Horak et al., 2009) and the Tinetti Balance and Gait Test (Tinetti, 1986), that include reactive balance items, have limited accuracy due to few items and a coarse scale. These tests do not reproduce the unpredictability of unexpected loss of balance, which is an important requirement for testing reactive control (Maki and McIlroy, 2006) but is difficult to ensure in a standardized test. Accordingly, there is a concerning lack of clinical approaches for measuring reactive balance ability.

In the scientific setting, several approaches have been developed, e.g., perturbations by cable pull (Hilliard et al., 2008), sudden cable release of tethered lean (Carty et al., 2015), and platform motions (Maki and McIlroy, 2006; Madigan et al., 2018; Aviles et al., 2019). Emerging technologies enable the computerized application of perturbations in various directions, intensities, time intervals, and under controlled, safe conditions (Shapiro and Melzer, 2010). This provides the opportunity to simulate the unpredictability of events that lead to loss of balance in daily life.

In previous studies, reactive single-step and multi-step responses (respectively lower step thresholds) have been shown to be independent predictors of future falls in community-dwelling older adults (Hilliard et al., 2008; Batcir et al., 2020; Crenshaw et al., 2020). A recent meta-analysis of 12 studies came to the results that reactive stepping tests can distinguish moderately between fallers and non-fallers (Okubo et al., 2021), but the studies differ greatly in their applied methods and results. A unified and standardized measurement procedure of reactive control in healthy older adults that is both valid and feasible for clinical uptake is still missing.

In this context, the study of Handelzalts et al. (2019a,b) presented a promising test approach. They applied perturbations by platform translations in four directions and at six progressive intensity levels to assess reactive balance ability in healthy adults and individuals after stroke. The single-step and multiple stepping thresholds were determined. The assessment tool developed, hereafter referred to as the Stepping Threshold Test (STT), proved to be inter-observer reliable in both populations and convergent validity for individuals after stroke (Handelzalts et al., 2019b). However, data on the validity of the STT in healthy older adults are not yet available.

In previous studies, each step after a perturbation was counted to determine the number of steps required to regain balance (Mille et al., 2013; Crenshaw et al., 2020) or the step and stepping thresholds (Batcir et al., 2018, 2020; Handelzalts et al., 2019a,b). The study of Handelzalts et al. (2019a) defined steps on the basis of an extension of the base of support (BoS). The study of Arampatzis et al. (2008) also considered the direction of perturbation in their definition. They used a cable release system and defined a multiple stepping as any second step taken by the recovery limb or an anterior exceeding of the first step by the contralateral limb (Arampatzis et al., 2008).

From a biomechanical view, a consideration of the extension of the BoS and the direction of perturbation could lead to a further refinement of the step evaluation strategy of the STT. Perturbations lead to a movement of the center of mass (CoM) (Maki and McIlroy, 1997). Step and stepping strategies aim to modify the BoS in order to maintain the CoM within the stability limits of the BoS (Maki and McIlroy, 1997). Thus, if a step extends the BoS in a different direction than the CoM movement, it cannot directly support rebalancing and cannot be considered as part of an efficient reactive strategy. Accordingly, an efficient step and stepping strategy at the step threshold extends the BoS toward CoM motion and therefore opposite to the direction of surface translation. Other strategies could reflect an inadequate reaction or might merely serve to increase standing comfort. When considering multiple steps, it should be taken into account that the BoS has already changed after the first step. Consequently, every single step that follows the first step should be evaluated based on the actual (newly formed) BoS.

For this reason, we developed a new strategy to evaluate the step and stepping behavior of the STT, which we called the ‘direction-sensitive evaluation’ (DSE). As opposed to counting every step is taken (Handelzalts et al., 2019b; Batcir et al., 2020; Crenshaw et al., 2020), which we called the ‘all-step-count evaluation’ (ACE), our approach considers two important characteristics in the step and stepping behavior. First, our approach leads to a direction-specific consideration since steps counted only in the opposite direction to the surface translation. Second, single steps and multiple steps are counted only if they extended the actual BoS.

This investigation had three aims. Our first aim was to test the convergent validity of the STT in fall-prone older adults with respect to fall risk. For this purpose, we used an established method and investigated associations between widely used clinical measures of balance and fall risk (Handelzalts et al., 2019b) and the STT sum score (convergent validity). We expected to find moderate correlations with the Brief Balance Evaluations Systems Test (BriefBEST, global balance), moderate correlation with the TUG (proactive balance), low to moderate correlations with the 8-level balance scale (8LBS, static balance), and low to moderate correlations with the Short Falls Efficacy Scale—International (Short FES-I, fear of falling). This expectation is based on the results of Handelzalts et al. (2019b), Crenshaw et al. (2018), and a meta-analysis by Kiss et al. (2018) who found associations between reactive balance and other balance domains. Our second aim was to explore the association between the STT and the experience of at least one fall in the past 12 months. We hypothesized to find significant differences in the STT sum score between fallers and non-fallers in the past year (discriminative validity). Past falls are among the strongest risk factors for future falls (Ek et al., 2019) and fallers use significantly more recovery steps after perturbations than non-fallers (Okubo et al., 2021). Our third aim was to evaluate the feasibility and interpretability of the STT. We hypothesized the test to be safe and feasible in fall-prone older adults. The study of Handelzalts et al. (2019b) successfully applied the STT in the vulnerable group of individuals with stroke. Our fourth aim was to compare the validity of the ACE and DSE in order to explore the advantages of a differentiated step evaluation and to advance the standardization of the measurement process. We hypothesized to find stronger evidence for convergent and discriminative validity in the DSE compared with the ACE since the DSE leads to a more differentiated consideration of stepping behavior.

METHODS

Study Participants

This methodological study used baseline data of an intervention study on perturbation-based balance training registered at clinicaltrials.gov (trial register number: NCT04087512). A sample of 71 community-dwelling adults aged 65 and older was recruited. We contacted 3,350 people *via* a random selection by the local resident registration office. Eligibility criteria were assessed in a two-step procedure consisting of a standardized telephone screening and a face-to-face screening (**Figure 1**). Eligible subjects were invited to the baseline assessment. Subjects had to be able to walk for at least 20 min without a walking aid and had to be fall-prone. The latter could be met in two ways. It was identified either the subject has experienced a fall in the last 12 months or a subjective feeling of a decrease in balance ability in the past year and a deficit in balance ability, defined as a loss of balance ability on the 8LBS (Clemson et al., 2012; Weber et al., 2018) to level 4 (tandem standing with eyes closed). Exclusion criteria included severe metabolic, cardiovascular, pulmonary, neurological, or orthopedic diseases. Moreover, subjects were excluded if cognitive impairment was suspected due to a score below eight on DemTect (Kessler et al., 2000). Other reasons for exclusion were strong dizziness, a body mass index above 30, significant visual or sensory impairments, and participation in balance training in the last 3 months. This study was carried out in accordance with the Declaration of Helsinki and approved by the ethics committee of Heidelberg University (reference: AZ Schwe 2019 /1-2).

Measurements

Demographic characteristics and falls within the last 12 months (retrospective) were assessed and recorded during standardized interviews. For this purpose, a fall was operationally defined as an unexpected event in which a person walking, standing, sitting, or lying down involuntarily, suddenly, and uncontrollably comes to rest on the ground or another lower level (Hauer et al., 2006). Participants were classified as non-fallers and fallers (at least one fall in the past 12 months) (Crenshaw et al., 2020).

For the testing procedure of the STT, we used a commercial perturbation treadmill (Balance Tutor, MediTouch, Israel) (**Figure 2**). The study of Shapiro and Melzer (2010) described the system configuration. Starting from approaches of previous studies (Batcir et al., 2018, 2020; Handelzalts et al., 2019a,b) that use step and stepping thresholds to estimate reactive balance, we defined the STT as follows: Participants were instructed to stand on the Balance Tutor in their shoes with their both feet together and to respond to unannounced surface translation perturbations (backward, forward, left, and right) with as few compensatory steps as possible. The test was composed of six levels with increasing intensity (**Table 1**). Each level contained four unannounced surface translations, one in each direction. An additional perturbation that was not included in the analysis was added to the sequence (in level 4 of 6) to ensure the unpredictability of the perturbation direction. The order of directions varied randomly between the levels (**Supplementary Material 1**). The order of perturbation intensity was not randomized but gradually increasing because we aimed to determine participants' single-step and multiple stepping thresholds. Participants were exposed to each perturbation only once. The perturbations lasted 0.5 s and the intervals in between were 10 and 19.5 s (**Supplementary Material 1**). Familiarization with the perturbation treadmill consisted of 10–20 min of normal walking on the treadmill at the face-to-face screening (12.3 ± 4.7

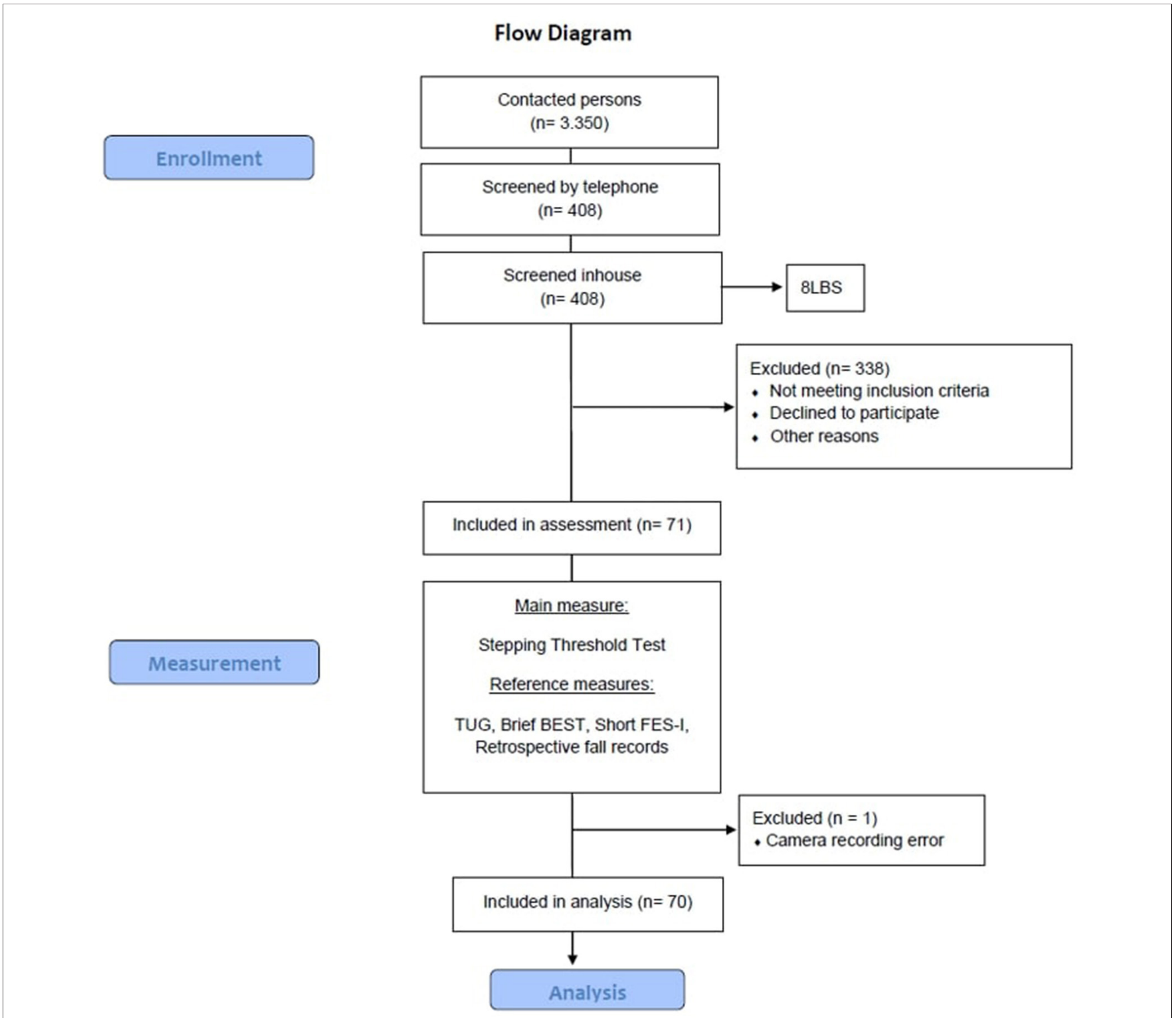

FIGURE 1 | The course of study. BriefBEST, Brief Balance Evaluations Systems Test; FES-I, Short Falls Efficacy Scale—International; TUG, Timed Up and Go Test; 8LBS, Eight level balance scale.

days before the actual assessment), a full body weight relief into the harness system, and two perturbations at the lowest intensity level prior to the test. Subsequently, the STT was performed.

The stepping behavior of the participants was evaluated for all 24 surface translations. In order to avoid injuries, the participants wore a safety harness that protected them from falling. The rope length was adjusted so that in the event of a fall, the knees of the participants would come to rest ∼10 cm above the treadmill surface. In case of a fall or excessive fear by the participant, the test was terminated prematurely. The testing process was recorded on video from the thoracic spine of the participant downwards. The camera system (Logitech C920HD Pro Webcam, Logitech, Apples, Switzerland) was placed at a distance of 2.1 m and an angle of 35° dorsolateral to the participant (**Figure 3**) and recorded at frame rates of 30 Hz.

The evaluation of the stepping behavior of the participants was assessed by video analysis. Stepping behavior was scored as no step, single step, or multiple stepping. For the ACE, we counted each step up to the point where the subject regained balance, based on the observational judgment of a static steady-state balance, i.e., maintaining a steady position while standing with a stable trunk. For this purpose, we defined a step as an observable change in the bipedal BoS. In the DSE, we specified a step as reaction behavior that leads to a sensible extension of the BoS in the opposite direction of the surface translation. To be counted as a single step, the BoS in the basic test position

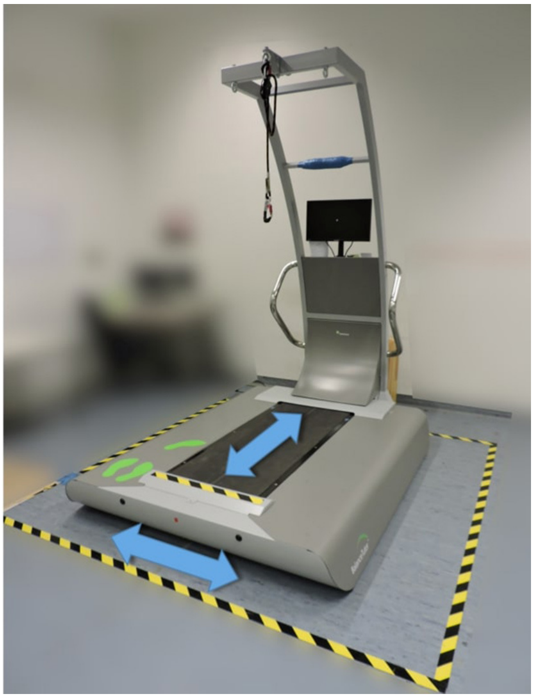

FIGURE 2 | Balance tutor and directions of surface translation.

TABLE 1 | Characteristics of surface translations.

Level of the STT	Displacement anteroposterior (cm)[a]	Displacement mediolateral (cm)[b]
1	7.4	3.3
2	12.9	6.3
3	18.5	9.2
4	23.9	12.1
5	29.5	15.1
6	35.0	18.0

[a] *Displacement of treadmill surface in the forward and backward direction.*
[b] *Displacement of treadmill surface in left and right direction.*

had to be extended by one step in the opposite direction to the surface translation. To be counted as a multiple-step, the actual BoS had to be additionally extended by one or more steps in the opposite direction to the surface translation. Subsequently, single-step and multiple stepping thresholds were determined for each direction of surface translation (forward, backward, left, and right). In the increasing perturbation protocol, the single-step threshold was defined as the first perturbation displacement from which the subject needed to take a step to recover. The multiple stepping threshold was defined as the first perturbation displacement from which the subject needed to take multiple steps, i.e., at least two steps, to recover. To ensure that the threshold was reached, two successive perturbations in the same direction each had to result in a single step or multiple stepping for the threshold to be scored (Batcir et al., 2018). The first of these two consecutive perturbations was set as the threshold.

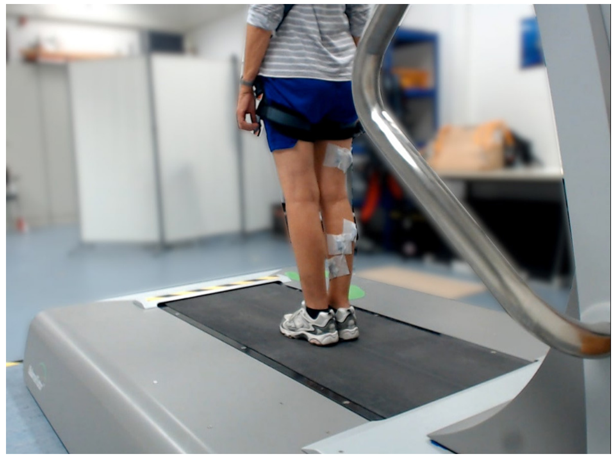

FIGURE 3 | The perspective of the camera.

Some participants did not reach all of the eight step and stepping thresholds (original thresholds). In this case, the threshold value was set at one level above the highest executed level as conducted before (Handelzalts et al., 2019a). The thresholds were termed according to the direction of the surface translation (e.g., single-step threshold forward).

We used several established and widely used clinical assessments for balance and fall risk as reference measures. The Brief Balance Evaluations Systems Test was obtained by an assessor as described elsewhere (Marques et al., 2016). It is a shortened version of Horak's BESTest (Padgett et al., 2012) and consists of six items, measuring aspects of static, dynamic, proactive, and reactive postural control in standing and walking. The Timed Up and Go Test is a widely used performance-based assessment of dynamic balance and fall risk (Podsiadlo and Richardson, 1991). It was assessed per protocol, by measuring the time needed by the participant to stand up from a chair, walk three meters at a brisk but safe pace, turn 180 degrees, and walk back to the chair to sit down. The 8-level balance scale is a further development of the Short Physical Performance Battery (Guralnik et al., 1994). It comprises eight static balance tasks with increasing difficulty. Every task needed to be performed for 15 s without external support, the use of a reactive step, or compensatory arm movements (Clemson et al., 2012; Gordt et al., 2020). Fear of falling was assessed by the interviewer using the Short FES-I (Kempen et al., 2008). Participants rated their level of confidence during seven activities of daily life on a 4-point Likert scale, with a lower value representing more confidence.

Statistical Analysis

Statistical analyses were performed using IBM SPSS Statistics Version 26 (IBM, New York, NY, USA) and MS Excel 2010 (Microsoft, Redmond, Washington, USA). Hypotheses were two-sided evaluated at the alpha level at $p < 0.05$. The primary outcome was the STT sum score, calculated as the sum of all eight original single-step and multiple stepping thresholds. Secondary outcomes included the STT subscores, i.e., sums of single-step thresholds, multiple stepping thresholds, mediolateral (left and right) step and stepping thresholds, anteroposterior (forward and backward) step and stepping thresholds, and the original thresholds for each (forward, backward, left, and right) single-step and multiple stepping threshold.

Descriptive statistics were used to characterize the study population. Differences between non-fallers and fallers with regards to demographics, postural balance capacity, and fear of falling were analyzed by means of Chi²-Test for categorical variables, and by either the Mann-Whitney-U test or the independent t-test, as indicated, for continuous variables. Normal distribution was tested by means of the Shapiro-Wilk W test. For the estimation of convergent validity, we investigated the association of the STT with the TUG, BriefBEST, 8LBS, and the Short-FES-I applying the Spearman's rank correlation coefficient. Correlation coefficients of $r = 0.1–0.29$ indicate a small, $r = 0.3–0.49$ moderate and $r \geq 0.50$ strong correlations (Cohen, 1988). Discriminative validity was calculated by the Mann-Whitney U test and non-parametric receiver operating characteristic (ROC) curve analysis. The Mann-Whitney U statistics were applied to determine differences between the groups of fallers with respect to the STT. The receiver operating characteristic curve analysis was used to determine the prognostic value in order to evaluate a difference between fallers and non-fallers, by means of the area under the curve (AUC). As the non-parametric ROC analysis is based on the Mann-Whitney U statistic, we reported only the ROC curves of the STT variables that were significantly different between different groups of fallers. The area under the curve values of the ROC were classified into non-informative (AUC $= 0.5$), less accurate ($0.5 < \text{AUC} \leq 0.7$), moderately accurate ($0.7 < \text{AUC} \leq 0.9$), very accurate ($0.9 < \text{AUC} < 1$), and perfect (AUC $= 1$) (Greiner et al., 2000). Information about feasibility was examined based on the rate of early test terminations and the occurrence of adverse events during the STT. Adverse events were defined as any unfavorable or unintended event that occurs in the course of this study (Ory et al., 2005). Floor and ceiling effects occur when a distinct percentage of subjects achieve the worst or best possible score and reflect an incomplete distribution of sample within a test and insufficiency to distinguish subjects at the lower and upper ends of the measurement system (McHorney and Tarlov, 1995). They were defined to be present if more than 15% of subjects reached the highest and lowest level, respectively (McHorney and Tarlov, 1995). A sensitivity analysis using G∗Power 3.1.9.7 (Faul et al., 2009) showed that with a sample size of $n = 70$ a correlation of 0.327 can be shown with power 0.8 using a significance level of 0.05.

RESULTS

Demographics

A consecutively recruited sample of 70 fall-prone older adults with a mean age of 74.8 years ($SD = 6$) was included in the analysis (**Table 2**). From the 71 recruited participants, one had to be excluded from analysis due to technical problems and incomplete data. Among the included participants, 32 (46.5%) had experienced at least one fall in the past 12 months and were therefore classified as fallers. There were significantly more women categorized as fallers than as non-fallers ($p = 0.007$). No further significant differences were found in regards to age, gender, balance capacity, and fear of falling between non-fallers and fallers (**Table 2**).

TABLE 2 | Study population characteristics.

	All participants ($n = 70$)	Non-fallers ($n = 38$)	Fallers ($n = 32$)	Sign.
N (%) Women	45 (64.3)	19 (50.0)	26 (81.3)	0.007
Mean Age ± SD	74.8 ± 6.0	75.4 ± 6.4	74.0 ± 5.3	0.385
Median BriefBEST (IQR)	18 (4.25)	17.5 (5)	18 (3.75)	0.374
Mean TUG ± SD	7.8 ± 1.3	8.1 ± 1.4	7.5 ± 1.0	0.600
Median 8LBS (IQR)	5 (1)	5 (2)	5 (1)	0.104
Median Short FES-I (IQR)	8 (2.25)	8 (3)	8.5 (2)	0.370

Sign., One-tailed significance level was set to $p < 0.05$. BriefBEST, Brief Balance Evaluations Systems Test; FES-I, Short Falls Efficacy Scale—International. IQR, Interquartile range; TUG, Timed Up and Go Test; 8LBS, Eight level balance scale; P-value calculated by means of the Mann-Whitney-U test.

Convergent Validity

The Stepping Threshold Test sum score (ACE) correlated moderately with the BriefBEST ($r = 0.413$) and the TUG ($r = -0.379$). In addition, the STT sum score (ACE) correlated low with the 8LBS ($r = 0.173$) and the Short FES-I ($r = 0.108$) (**Table 3**). The Stepping Threshold Test subscores (ACE) correlated low ($r = 0.102$ to $|-0.297|$) in 8 of 16 values, moderately in 6 values ($r = 0.312$ to $|-0.433|$), and did not correlate with the reference measures in 2 values (**Table 3**). The single-step thresholds (ACE) correlated in low 8 of 16 values ($r = 0.107$ to $|-0.293|$), moderately in 5 values ($r = 0.300$ to $|-0.390|$), and did not correlate in 3 values with the reference measures (**Supplementary Material 2.1**). The multiple stepping thresholds (ACE) correlated low in 12 of 16 values ($r = 0.104$ to $|-0.292|$), moderately in 1 value ($r = 0.309$), and did not show correlations with the reference measures in 3 values (**Supplementary Material 2.1**). Correlation plots for visual inspection are presented in the appendices (**Supplementary Material 3.1**).

The Stepping Threshold Test sum score (DSE) correlated moderately with the BriefBEST ($r = 0.388$) and the TUG ($r = -0.435$). In addition, the STT sum score (DSE) correlated low with the 8LBS ($r = 0.246$) and Short FES-I ($r = -0.104$) and (**Table 4**). The Stepping Threshold Test subscores (DSE) correlated low ($r = |-0.104|$ to $|-0.279|$) in 7 of 16 values, moderately in 6 values ($r = 0.305$ to $|-0.447|$), and did not correlate with the reference measures in 3 values (**Table 4**). The single-step thresholds of the DSE correlated low in 9 of 16 values ($r = |-0.105|$ to $|-0.238|$), moderately in 2 values ($r = |-0.342|$ to 0.415), and did not correlate with the reference measures in 5 values (**Supplementary Material 4.1**). The multiple stepping thresholds (DSE) correlated low in 13 of 16 values ($r = 0.106$ to $|-0.267|$), moderately in 1 value ($r = |-0.318|$), and did not show correlations with the reference

TABLE 3 | Correlation between STT sum score and subscores (ACE) and reference measures.

	STT-Thresholds		BriefBEST	TUG	8LBS	Short FES-I
Primary outcome	STT sum score	r	0.413	−0.379	0.173	−0.108
		CI95	0.19–0.6	−0.57–0.15	−0.07–0.39	−0.34–0.13
Secondary outcome	SS subscore	r	0.425	−0.433	0.144	−0.021
		CI95	0.2–0.61	−0.61–0.21	−0.1–0.37	−0.25–0.22
	MS subscore	r	0.318	−0.280	0.135	−0.120
		CI95	0.08–0.52	−0.49–0.04	−0.1–0.36	−0.35–0.12
	AP subscore	r	0.372	−0.297	0.161	−0.195
		CI95	0.14–0.56	−0.5–0.06	−0.08–0.38	−0.41–0.04
	ML subscore	r	0.312	−0.314	0.102	−0.008
		CI95	0.08–0.51	−0.52–0.08	−0.14–0.33	−0.24–0.23

Sign., Two-tailed significance; SS, Single-step; MS, Multiple Stepping; AP, Anteroposterior; ML, Mediolateral; CI95, confidence interval of 95%; r, correlation coefficient rho, calculated by means of the Spearman-rank-correlation; TUG, Timed Up and Go Test; BriefBEST, Brief Balance Evaluations Systems Test; 8LBS, Eight level balance scale; FES-I, Short Falls Efficacy Scale—International.

TABLE 4 | Correlation between STT sum score and subscores (DSE) and reference measures.

	STT-Thresholds		BriefBEST	TUG	8LBS	Short FES-I
Primary outcome	STT sum score	r	0.388	−0.435	0.246	−0.104
		CI95	0.16–0.58	−0.61–0.21	0.01–0.46	−0.33–0.13
Secondary outcome	SS subscore	r	0.276	−0.354	0.055	−0.120
		CI95	0.04–0.48	−0.55–0.12	−0.18–0.29	−0.35–0.12
	MS subscore	r	0.377	−0.378	0.305	−0.068
		CI95	0.15–0.57	−0.57–0.15	0.07–0.51	−0.3–0.17
	AP subscore	r	0.430	−0.447	0.272	−0.249
		CI95	0.21–0.61	−0.62–0.23	0.04–0.48	−0.46–0.01
	ML subscore	r	0.227	−0.279	0.113	0.085
		CI95	−0.01–0.44	−0.49–0.04	−0.13–0.34	−0.15–0.31

Sign., Two-tailed significance; SS, Single-step; MS, Multiple Stepping; AP, Anteroposterior; ML, Mediolateral; CI95, confidence interval of 95%; r, correlation coefficient rho, calculated by means of the Spearman-rank-correlation; TUG, Timed Up and Go Test; BriefBEST, Brief Balance Evaluations Systems Test; 8LBS, Eight level balance scale; FES-I, Short Falls Efficacy Scale—International.

measures in 2 values (**Supplementary Material 4.1**). Correlation plots for visual inspection are presented in the appendices (**Supplementary Material 3.2**).

Discriminative Validity

The Stepping Threshold Test sum score and subscores (ACE) showed no significant differences between fallers and non-fallers (**Table 5**). Significant differences were found in the single-step threshold backward, with advantages for the fallers compared with the non-fallers ($p = 0.034$) (**Supplementary Material 2.2**).

The Stepping Threshold Test sum score and subscores (DSE) showed no significant differences (**Table 6**) between fallers and non-fallers. Significant differences were found in the single-step threshold right ($p = 0.015$) with higher thresholds for the non-fallers compared with the fallers (**Supplementary Material 4.2**). The subsequent ROC-analysis indicated an AUC of 0.634 (95CI = 0.511–0.775).

Interpretability of the STT

The Stepping Threshold Test sum score of both ACE and DSE showed no floor (0%) or ceiling effect (0%). The subscores of both the ACE and DSE also revealed no ceiling or floor effects (0–4.29%) (**Supplementary Materials 5.1, 5.2**). In both the ACE and DSE, floor effects occurred in the single-step thresholds forward (41.43–57.14%) and backward (20–21.43%) (**Tables 7**, **8**). In the ACE ceiling effect occurred only in the multiple stepping threshold left (21.43%) (**Table 7**). In the DSE ceiling effects were observed for the multiple stepping threshold backward (54.3%), left (48.6%), and right (54.3%) (**Table 8**).

Feasibility

In total, 1,593 of 1,680 (94.8 %) perturbations were applied. The test was terminated prematurely in 18 subjects (25%) with an average of 19.7 out of 25 applied perturbations ($SD = 2.7$). In the ACE, 17 of these 18 (94.44%; in total, 69 of 70, 98.57%) subjects had already reached all single-step thresholds and 10 of these 18 subjects (55.55%) had already reached all multiple stepping thresholds. In the DSE, 9 of these 18 (50%; in total, 61 of 70, 87.14%) subjects had already reached all single-step thresholds and none of these 18 subjects (0%) had already reached all multiple stepping thresholds. Accordingly, 62 of 70 (88.57%) participants reached all thresholds in the ACE and 52 of 70 (75%) participants reached all stepping thresholds in the DSE during the testing procedure. For five participants (7.14 %), fall thresholds were documented (mean perturbation = 21.4, $SD = 2.3$), whereas the earliest fall appeared in perturbation number 18 and the latest in perturbation 23. We were able to include every participant but one due to technical problems (98.59%) in the analysis of both ACE and DSE using the calculated thresholds. There were no adverse events, but some participants reported high stress levels and anxiety during the higher intensities of the STT.

DISCUSSION

This study is the first empiric investigation of the psychometric properties of the STT in fall-prone older adults. We provided evidence of the convergent validity of this reactive balance test with respect to fall risk and introduce a newly developed DSE to evaluate stepping behavior. Discriminative validity could not be demonstrated. Floor and ceiling effects were found in the original thresholds for ACE and DSE, but not in the sum scores and subscores. Completion rates of the STT indicated sufficient feasibility for the ACE, but not for the DSE.

Convergent Validity

Previous studies reported correlations between reactive balance and measures of other balance domains between 0.03 and 0.691 (Crenshaw et al., 2018; Kiss et al., 2018; Handelzalts et al., 2019b). The Brief Balance Evaluation Systems Test is a testing battery that contains measures of all four balance domains (Marques

TABLE 5 | Differences between non-fallers and fallers in the STT sum scores (ACE).

		Non-fallers (n = 38)					Fallers (n = 32)					
		Mean	**Median**	**IQR**	**Min**	**Max**	**Mean**	**Median**	**IQR**	**Min**	**Max**	***P*-value**
Primary outcome	STT sum score	25.61	27	7	12	36	26.22	26	8.5	15	37	0.897
Secondary outcome	SS subscore	9.05	9	3.25	4	14	9.56	9	3	6	14	0.571
	MS subscore	16.55	16.5	5.25	8	25	16.66	17	4.75	9	26	1.000
	AP subscore	10.71	10	4	4	18	11.25	10	5	5	21	0.647
	ML subscore	14.89	15	3.5	7	21	14.97	15	5.75	10	21	0.817

SS, Single-step; MS, Multiple Stepping AP, Anteroposterior; ML, Mediolateral; Two-tailed p-value calculated by means of the Mann-Whitney-U test. The significance level was set to $p < 0.05$.

TABLE 6 | Differences between non-fallers and fallers in the STT sum scores (DSE).

		Non-fallers (n = 38)					Fallers (n = 32)					
		Mean	**Median**	**IQR**	**Min**	**Max**	**Mean**	**Median**	**IQR**	**Min**	**Max**	***P*-value**
Primary outcome	STT sum score	35.5	35	7.25	26	44	35.13	35	7.75	27	43	0.799
Secondary outcome	SS subscore	12.68	12	3	6	18	12.32	12	3	8	16	0.501
	MS subscore	22.82	23	4.25	16	27	22.81	23.5	5	17	28	0.972
	AP subscore	14.29	14	3	7	20	14.31	14	4.75	9	19	0.976
	ML subscore	21.21	21.5	2	16	27	20.81	21.5	5	16	26	0.807

SS, Single-step; MS, Multiple Stepping AP, Anteroposterior; ML, Mediolateral; Two-tailed p-value calculated by means of the Mann-Whitney-U test. The significance level was set to $p < 0.05$.

TABLE 7 | Floor and ceiling effects of the STT (ACE).

	Floor effect[a]		Ceiling effect[b]	
	Single step	**Multiple step**	**Single step**	**Multiple step**
Forward	57.14%	14.29%	0.00%	0.00%
Backward	21.43%	1.43%	0.00%	21.43%
Left	5.71%	0.00%	0.00%	2.86%
Right	4.29%	0.00%	0.00%	10.00%

Floor or ceiling effect exists if the value is above 15 %. [a]Percentage of participants who reached the lowest level in the single or multiple stepping thresholds. [b]Percentage of participants who reached the highest single or multiple stepping thresholds.

TABLE 8 | Floor and ceiling effects of the STT (DSE).

	Floor effect[a]		Ceiling effect[b]	
	Single step	**Multiple step**	**Single step**	**Multiple step**
Forward	41.43%	0.00%	0.00%	7.14%
Backward	20.00%	0.00%	0.00%	54.29%
Left	0.00%	0.00%	1.43%	48.57%
Right	1.43%	0.00%	2.86%	54.29%

Floor or ceiling effect exists if the value is above 15 %. [a]Percentage of participants who reached the lowest level in the single or multiple stepping thresholds. [b]Percentage of participants who reached the highest single or multiple stepping thresholds.

et al., 2016) defined by Shumway-Cook and Woollacott (2017), i.e., static, dynamic, proactive, and reactive balance. Accordingly, we expected to find moderate correlations between the STT and global balance as measured by the BriefBEST. Thus, moderate correlations between the STT sum score and the BriefBEST of 0.413 (ACE) and 0.388 (DSE) confirmed our hypothesis related to convergent validity.

The meta-analysis by Kiss et al. (2018) found a low correlation between reactive balance and proactive balance ($r = 0.14$), but this result was based on only a single study (Owings et al., 2000). The study of Handelzalts et al. (2019b) performed the STT in 15 persons with stroke and correlated balance measures with the fall thresholds, i.e., the perturbation intensity that could not be compensated and led to unambiguous support by the harness. They found correlations of $r = 0.691$ between the STT and the Berg Balance Scale, a test battery that primarily consists of proactive balance items. Accordingly, we also hypothesized to find moderate correlations between measures of reactive and proactive balance. Correlations between the STT sum score and the TUG of $r = -0.379$ (ACE) and $r = -0.435$ (DSE) confirmed our hypothesis. Lower correlations in our study compared with the study of Handelzalts et al. (2019b) may be attributed to the different sample characteristics, i.e., stroke patients vs. older adults.

The study of Crenshaw et al. (2018) explored correlations between standing postural control and anteroposterior step and stepping thresholds and revealed low to moderate correlations ($r = 0.21$–0.38). The study of Kiss et al. (2018) included five studies in their analysis with respect to the relationship of static balance and reactive balance and found a correlation coefficient of $r = 0.19$. Accordingly, we expected low to moderate correlations between reactive balance and static balance in our study. Correlation coefficients between the SST sum scores and the 8LBS of $r = 0.173$ (ACE) and $r = 0.246$ (DSE) confirmed our hypothesis of a low to moderate the relationship between reactive and static balance measures.

In a recent study, Batcir et al. (2020) applied mediolateral perturbations in a comparable sample and found moderate correlations (single-step threshold: $r = -0.398$ and multiple stepping threshold: -0.302) between the single-step and multiple stepping thresholds and the fear of falling. The study of Crenshaw et al. (2018) applied anteroposterior perturbations and found low correlations ($r = 0.19$–0.20) of single-step thresholds and moderate correlations ($r = 0.39$–0.40) of multiple stepping thresholds with activity-specific balance confidence, a construct which is similar to fear of falling. Accordingly, we hypothesized to find low to moderate correlations between the STT sum score and the Short FES-I. We determined lower correlation coefficients of $r = -0.108$ (ACE) and $r = -0.104$ (DSE), which are, however, within the expected range. Interestingly, anteroposterior subscores were higher (ACE: $r = -0.195$ and DSE: $r = -0.249$) and mediolateral subscore did not indicate any correlation (ACE: $r = -0.008$ and DSE: $r = 0.085$). These findings are in line with experiences gained during the testing procedure that AP perturbations seemed to be the most uncomfortable especially for anxious participants. Anteroposterior step and stepping thresholds might be closer related to fear of falling since backward perturbations require a particular fast step reaction (Sturnieks et al., 2013). In addition, forward step and stepping motion is a very common lower extremity motion in daily life and is also addressed in the Short FES-I (Kempen et al., 2008). The absence of more and higher correlations can be explained by the fact, that the median Short FES-I score was very low in our study population. A reason may be that mainly individuals with a lower fear of falling were willing to participate in our study (recruitment bias).

These results are supplemented by numerous correlations between the reference measures, the STT subscores (ACE: 14 of 16 values, $r = 0.102$ to $|-0.433|$; DSE: 13 of 16 values, $r = |-0.104|$ to $|-0.447|$) and the original single-step and multiple stepping thresholds (ACE: 26 of 32 values, $r = 0.104$ to $|-0.390|$; DSE: 25 of 32 values, $r = |-0.105|$ to 0.415). Due to the high numbers of variables in our secondary outcomes the possibility of type-I error must be considered here. However, only the primary outcome, i.e., the STT sum score, was considered in hypothesis testing and secondary outcomes do not affect the conclusion of this study. In summary, our hypothesis regarding the convergent validity of the STT with other assessments of balance and fall risk was confirmed.

Discriminative Validity

Our initial hypothesis regarding the discriminative validity of the STT could not be confirmed. None of the sum scores or subscores showed significant differences in the comparison of fallers and non-fallers. In the DSE, we found one original threshold, i.e., single-step threshold right, at which non-fallers performed significantly better than fallers. However, since we conducted several analyses for the same hypothesis, single results should be interpreted with caution and could be due to chance (Streiner and Norman, 2011). In addition, we also found a threshold in the ACE at which fallers performed significantly better. Several previous studies showed reactive step and stepping thresholds to be capable to distinguish between non-fallers and fallers (Hilliard et al., 2008; Batcir et al., 2020; Crenshaw et al., 2020). However, our results are aligned with other studies that could not show significant differences between non-fallers and fallers by means of reactive balance tests (Mille et al., 2013; Sturnieks et al., 2013; Fujimoto et al., 2015).

On one hand, the lack of significant results might be due to our inclusion criterion of fall proneness resulting in low heterogeneity between fallers and non-fallers. Although normal age-related physiological changes, balance deficits, and fear of falling are relevant to falls (Ambrose et al., 2013), we did not find any significant difference between fallers and non-fallers. On the other hand, retrospective fall assessment is accompanied by a risk of inaccurate data because of recall bias (Ganz et al., 2005), and prospective fall assessment is preferable. Previous studies compared non-fallers with recurrent fallers (at least two falls) (Balasubramanian et al., 2015; Lima et al., 2018; Batcir et al., 2020) to increase discriminatory power between the groups and to ensure that subjects are not classified as fall-prone because of an unavoidable event that leads to a fall, but because of endogenous factors that significantly increase fall risk. However, the number of recurrent fallers in our study sample was too small to allow this, and further studies with a higher number of recurrent fallers are needed. In addition, strong floor and ceiling effects had occurred that may have limited the validity of the test procedure. Determining a fall threshold, i.e., the level of perturbation at which participants fall into a harness system, as done in the study by Handelzalts et al. (2019b), could lead to benefits in terms of discriminative validity. However, in our study population, only five participants had experienced a fall into the harness system during test use, so statistical evaluation of this threshold was not possible.

Interpretability

Neither for the sum score nor subscores floor or ceilings effects were found. However, strong floor and ceiling effects were observed in consideration of the individual step and stepping thresholds in both, the ACE and DSE. Since the criteria for whether a step is counted as such are more demanding in the DSE, it is plausible that stronger ceiling effects occurred here, whereas floor effects were more pronounced in the ACE. The greatest floor effects appeared in the single-step threshold forward. This threshold represented a forward displacement of the surface and thus a backward displacement of the CoM of the participants. Center of mass translations in the backward direction require a particular fast step reaction, as the location of the CoM is relatively close to the base-of-support border (Sturnieks et al., 2013) and muscular stabilization in this direction is more demanding (Hall and Jensen, 2002). In addition to the higher demand for this perturbation, the fact that the first perturbation was applied in this direction might have increased the floor effect even due to insecurity.

Even though we used the highest perturbation intensities that the utilized perturbation treadmill (Balance Tutor) is capable of, ceiling effects in the multiple stepping thresholds appeared in the DSE in all directions except in the forward translations. This is surprising, since mediolateral reactive stepping strategies, such as the cross-over step are also very demanding for older adults (Mille et al., 2013). Thus, depending on the target population higher intensities in mediolateral and backward surface translations might be necessary to evaluate multiple stepping thresholds with the DSE. Due to the limited system, one might consider other ways of increasing demand, e.g., limiting arm movements and reducing BoS in standing, but taking into account ecological validity (Reis and Judd, 2000) and the construct of reactive balance. Another potential improvement could be the inclusion of more levels of perturbation to expand the ability to stratify participants. However, this would increase the duration of the test and thus further increase the psychological and physical stress.

Feasibility

Previous studies regarded feasibility as sufficient if at least 85% of the measurements were successful (Malmberg et al., 2002; Waninge et al., 2011). For a clinical setting, even a higher rate of completion than 85% would be desirable. In this study, 75% of the participants performed all perturbations, but more than half (55.55%) had already reached all thresholds in the ACE leading to sufficient completion rates for this evaluation strategy. In the DSE, none had already reached all thresholds, resulting in an insufficient completion rate of 75%. Accordingly, we presented preliminary evidence that the STT is feasible using ACE in the scientific setting. For the feasibility of the DSE, higher completion rates should be achieved for the multiple stepping thresholds.

Sufficient test completion rates are already present for the single-step thresholds (ACE: 98.57% and DSE: 87.14%). In this study, results for the Single-step subscore were similar or only slightly different to the STT sum score and the Multiple stepping subscore in both ACE and DSE. The study of Crenshaw et al. (2018) found correlations of 0.29–0.68 between anteroposterior single-step and multiple stepping thresholds. Future studies should examine whether there is a substantial benefit by multiple stepping thresholds compared with single-step thresholds that justify the significantly higher burden placed on participants during the assessment.

Since the perturbation treadmill and the camera system we used are commercially available, the test application is also transferable to other settings. To increase feasibility, especially in the DSE, stress and anxiety levels should be reduced for example by more extensive familiarization with the test prior to the actual test administration. Further studies are needed to investigate the different areas of feasibility such as acceptability, practicality, and implementation (Bowen et al., 2009) of the STT in the scientific and clinical setting. Since no adverse events occurred, the STT can be considered safe.

ACE vs. DSE

When comparing ACE and DSE, we observed differences in the evaluation of 660 out of 1,593 (41.43%) applied perturbations. This high frequency of differences in the evaluations confirmed the need for a differentiated view of these two observer-based evaluation methods. For both evaluation strategies, namely the ACE and DSE, we presented evidence for convergent validity. Based on discriminative validity, neither DSE nor ACE shows advantages over the other evaluation strategy. The total number of correlating thresholds was slightly higher in the ACE compared with the DSE, the subscores, and the original thresholds. There are tendencies that the ACE might be more valid in mediolateral single-step thresholds and the DSE be more valid in anterior multiple-stepping thresholds (**Supplementary Materials 2.1, 4.1**). Furthermore, ceiling effects in the DSE suggested that that the full potential of this approach has not yet been exploited. Higher perturbation intensities could lead to an even more precise and differentiated assessment of reactive balance capacity, especially in the DSE, and to even clearer results regarding validity. In conclusion, we cannot make a clear recommendation on which evaluation strategy should be used in future assessments of reactive balance in community-dwelling, fall-prone older adults. Nonetheless, our results showed that a differentiated consideration of these two approaches is an important step on the way to a valid and feasible reactive balance test for this population. This will require further studies comparing the results of both approaches with other measurements of reactive balance and, if available, with a gold standard. To compare the utility of both approaches in assessing fall risk, prospective studies with higher numbers of participants and a less homogeneous population should be conducted.

Limitations

A limitation of this study is the retrospective characterization of participants as fallers or non-fallers. The number of recurrent fallers in our sample was too small to conduct an analysis of such a subsample. While our STT protocol was unpredictable with respect to perturbation direction, the gradual increase of the perturbation intensity might have been predictable. While our results indicated convergent validity, future validation studies could use a specific reactive balance test as a reference measure.

Recommendations for Future Research

Finally, we would like to provide recommendations based on our findings and experiences during the study process:

1) To avoid floor and ceiling effects, future studies should determine the optimal intensity in terms of magnitude, velocity, acceleration, duration of and the number of surface translations for each direction for different populations. In community-dwelling, fall-prone older adults this includes both higher and lower magnitudes than applied in this study. On the same note, care must be taken to avoid excessive demands and to ensure safety. Particular attention should be paid to the proper balance of mediolateral, anterior, and posterior perturbations intensities.

2) In consideration of the floor effects that occurred only in anteroposterior perturbations as well as the described associations between anteroposterior step and stepping thresholds and fear of falling, we recommend either starting the STT with mediolateral perturbations or with a lower intensity.
3) When using unexpected perturbations, participant anxiety and stress levels should be considered when planning studies. Future investigations should refrain from extending the duration of the test, e.g., by a higher number of applied perturbations, to avoid a further increase in the stress level and a resulting physical and psychological overload of the participants. In this context, we would like to point out psychological consequences of fall experiences such as post-fall anxiety syndrome (Rubenstein, 2006) and advice against pushing fall thresholds in older adults at risk for falls, especially those with previous fall experiences.
4) Perform perturbation treadmill familiarization consisting of treadmill walking, being caught by the harness, and small perturbations to keep stress and anxiety levels as low as possible. At the same time, the learning effect must be considered and kept as low as possible when performing a reactive balance test.
5) The calculation of sum scores, as presented in this study, contributes to higher validity and should be considered as a further variant with regard to the analysis of step and stepping thresholds.
6) Further validation studies are needed that compare results of the STT with other measures of reactive balance, e.g., the lean and release test (Inness et al., 2015).

CONCLUSION

The Stepping Threshold Test is a promising assessment tool of reactive balance applicable on commercially available computerized treadmill systems. We demonstrated evidence for convergent validity in fall-prone older adults. Furthermore, we presented a new approach with respect to the evaluation of reactive step and stepping behavior and gave concrete recommendations for further application of the test. Although current evidence is not sufficient to use the STT as fall risk assessment, we recommend further research in order to optimize the test protocol with respect to different target populations. If this succeeds, the STT has the potential to be applied as a regular, valid assessment for reactive balance in the clinical setting.

DATA AVAILABILITY STATEMENT

The datasets presented in this study can be found in online repositories. The names of the repository/repositories and accession number(s) can be found below: https://heibox.uni-heidelberg.de/d/cfd40c46c8be43b1b519.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Ethics Committee of Heidelberg University (reference: AZ Schwe 2019 /1-2). The patients/participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

MA was involved in data analysis and interpretation and drafting of the manuscript. LB was involved in the conception, experimental design, data acquisition, data interpretation, and drafting of the manuscript. ML was involved in data interpretation and critical revision of the manuscript. MS was involved in the conception, data interpretation, and drafting of the manuscript. All authors contributed to the article and approved the submitted version.

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SUPPLEMENTARY MATERIAL

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fspor.2021.715392/full#supplementary-material

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Adaptations in Reactive Balance Strategies in Healthy Older Adults After a 3-Week Perturbation Training Program and After a 12-Week Resistance Training Program

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Both resistance training (RT) and perturbation-based training (PBT) have been proposed and applied as interventions to improve reactive balance performance in older adults. PBT is a promising approach but the adaptations in underlying balance-correcting mechanisms through which PBT improves reactive balance performance are not well-understood. Besides it is unclear whether PBT induces adaptations that generalize to movement tasks that were not part of the training and whether those potential improvements would be larger than improvements induced by RT. We performed two training interventions with two groups of healthy older adults: a traditional 12-week RT program and a 3-week PBT program consisting of support-surface perturbations of standing balance. Reactive balance performance during standing and walking as well as a set of neuro-muscular properties to quantify muscle strength, sensory and motor acuity, were assessed pre- and post-intervention. We found that both PBT and RT induced training specific improvements, i.e., standing PBT improved reactive balance during perturbed standing and RT increased strength, but neither intervention affected reactive balance performance during perturbed treadmill walking. Analysis of the reliance on different balance-correcting strategies indicated that specific improvements in the PBT group during reactive standing balance were due to adaptations in the stepping threshold. Our findings indicate that the strong specificity of PBT can present a challenge to transfer improvements to fall prevention and should be considered in the design of an intervention. Next, we found that lack of improvement in muscle strength did not limit improving reactive balance in healthy older adults. For improving our understanding of generalizability of specific PBT in future research, we suggest performing an analysis of the reliance on the different balance-correcting strategies during both the training and assessment tasks.

Keywords: reactive balance, perturbation-based balance training, resistance training, ankle strategy, hip strategy, older adults, stepping threshold, fall prevention

INTRODUCTION

About one third of adults over the age of 65 fall each year (Tinetti et al., 1988). Approximately 30% of falls result in injuries requiring medical attention, with fractures occurring in ∼10% of these falls (Berry and Miller, 2008). Therefore, targeted interventions based on a good understanding of the circumstances in which falls occur and the underlying neuro-muscular mechanisms of increased fall risk are of great importance. Both resistance training (RT) (Liu-Ambrose et al., 2004) and perturbation-based training (PBT) (Mccrum et al., 2017) have been proposed and applied as interventions to reduce fall incidence in older adults. RT aims to improve or restore muscle strength by performing repetitive contractions against an external resistance. In previous studies, RT interventions have at most resulted in limited improvements in balance performance (Orr et al., 2008) and fall incidence (Faber et al., 2006; Fairhall et al., 2014). PBT is a more recent training paradigm and aims at improving reactive balance control by applying repeated unpredictable mechanical perturbations, typically during a small number of training sessions (Dijkstra et al., 2015; Mccrum et al., 2017). PBT is considered a promising intervention as it has been shown to improve balance during the trained task (Pai and Bhatt, 2007; Dijkstra et al., 2015) and seems to reduce fall risk (Pai and Bhatt, 2007; Mansfield et al., 2015; Gerards et al., 2017; Mccrum et al., 2017). From the principle of training specificity, improvements in reactive balance control in the task trained during PBT are expected. However, the adaptations in balance-correcting mechanisms that underlie improvements in reactive balance control after PBT and RT are not well-understood. Moreover, it is unclear whether PBT and RT induce adaptations in balance control that generalize to movement tasks that were not part of the training.

The most common circumstances in which falls occur in older adults are incorrect body-weight shifts, slips and trips (Berg et al., 1997; Robinovitch et al., 2013), indicating that maintaining reactive balance performance with age is key to prevent falls. Reactive balance is the skill to perform a balance-correcting response following a perturbation in order to avoid a loss of balance (Woollacott and Shumway-Cook, 2005). From a mechanical point of view, three strategies constitute a reactive balance-correcting response each requiring different muscular coordination (Hof, 2007; Halvorsen, 2010) (**Figure 2**): (1) a center-of-pressure (COP) or ankle strategy relies on a change in the ankle torque, shifting the COP location under the foot, in order to return the center-of-mass to the equilibrium position while the body sways around the ankle joint as a simple inverted pendulum; (2) a hip or inertial strategy relies on counter-rotation of body segments with respect to the center-of-mass (COM) to dissipate the change in angular momentum induced by the perturbation; and (3) a step strategy increases the base-of-support by taking a step. The timely and appropriate combination of these balance-correcting strategies determines reactive balance performance.

Older adults use these balance-correcting strategies differently than younger adults in response to perturbations during standing and walking. As compared to young adults, older adults rely less on COP (Gruben and Boehm, 2014) and inertial strategies (Afschrift et al., 2017) to attenuate the effect of perturbations during both standing (Runge et al., 1999; Jensen et al., 2001) and walking (Afschrift et al., 2019). Hence, older adults initiate stepping strategies at lower balance disturbances (Pai et al., 1998) resulting in more frequent use of stepping strategies and adapt step length more in response to perturbations during walking (Afschrift et al., 2019) than young adults. The lower stepping threshold during standing (Pai et al., 1998) and increased reliance on stepping strategies during walking (Afschrift et al., 2019) might suggest an age-related change in a common mechanism underlying both standing and walking balance. If that were the case, a training intervention that induces adaptation in that common mechanism would thus reduce the use of stepping strategies during both standing and walking.

Age-related changes in balance control might have multiple origins as the timely and appropriate implementation of balance-correcting strategies depends on the accuracy of integrated sensory information, the transformation of this information into motor commands and finally the functional capacity of the motor system that executes these commands (Pasma et al., 2014). Several age-related changes in sensorimotor function have been associated with decreased reactive balance performance and/or increased fall risk: reduced muscle function (e.g., maximal strength, rate of force development) (Morley, 2008), decline of visual (Lord and Dayhew, 2001), vestibular (Herdman et al., 2000), and proprioceptive acuity, increased neuromuscular noise (Singh et al., 2012), and decreased peripheral nerve conduction velocity (Pasma et al., 2014). In addition, several studies suggest that altered sensorimotor transformations, i.e., the transformation of sensory information into motor commands, can explain age-related changes in postural control (Bugnariu and Fung, 2006; Yeh et al., 2014) and other tasks such as reaching (Goodman et al., 2020). Hence, we expect that interventions might improve reactive balance control by (a) increasing acuity of sensory information, (b) inducing adaptations in sensorimotor transformations (Safavynia and Ting, 2013; Welch and Ting, 2014; Afschrift et al., 2021), or (c) improving capacities of the motor system (e.g., increasing muscle strength).

RT interventions have been successful in increasing muscle strength but have induced limited improvements in reactive balance performance (Hess et al., 2006) and reductions of fall risk (Faber et al., 2006; Cadore et al., 2014; Fairhall et al., 2014; De Labra et al., 2015). Age-related decreases in muscle strength and rate of force development potentially limit the COP strategy (Robinovitch et al., 2002; Hess et al., 2006) and impair the potential to quickly increase the base of support (BOS) when taking a step (Karamanidis et al., 2008). Higher maximal strength has been associated with better reactive balance performance during non-stepping responses in response to perturbations of standing (Mackey and Robinovitch, 2006) and lower fall risk (Melzer et al., 2004; Lin and Woollacott, 2005; Pijnappels et al., 2008; LaRoche et al., 2010; Cattagni et al., 2014; Gadelha et al., 2018). Some studies found similar associations between the rate of force development and balance performance (Pijnappels et al., 2008), whereas others did not (LaRoche et al., 2010; Kamo et al., 2019). Yet RT has not consistently led to improved reactive balance and it is unclear whether RT induced improvements in muscle strength lead to more efficient application of the COP strategy in healthy older adults.

PBT is a promising approach to improve reactive balance, but the mechanisms underlying improvements in reactive balance are not yet understood. PBT has induced improvements in reactive balance for the task being trained within a session (Sakai et al., 2008; Bierbaum et al., 2010; Tanvi et al., 2012) or after a couple of sessions (Dijkstra et al., 2015; Alizadehsaravi et al., 2021). Improvements in the trained task have been shown to be retained (Pai and Bhatt, 2007), and there is even some evidence of decreased fall risk following PBT (Mansfield et al., 2015; Gerards et al., 2017; Mccrum et al., 2017). Although PBT interventions yield some general exercise, they are not expected to introduce peripheral adaptations in skeletal muscle that lead to higher muscle strength. Such changes in muscle strength are especially unlikely because PBT is typically limited to a couple of days or weeks (Mansfield et al., 2007; Mccrum et al., 2017). Similarly, physiological changes at the cell level improving the acuity of sensory and motor systems are unlikely at this time scale (Aman et al., 2015). More likely, PBT affects the sensorimotor transformations that govern how the different balance-correcting strategies are combined. However, to the best of our knowledge, no studies describe how PBT alters the application of balance-correcting strategies.

In addition, it is unclear whether alterations in balance control during the trained task generalize to other tasks. A limited number of studies indicate improvements in reactive balance during tasks that were not trained. Parijat and Lockhart (2012) showed that practicing a slip-perturbation during walking, applied by means of translating a movable part of the walkway upon heel strike, improved balance when walking on a slippery surface. In this study however, the training exercises were very similar to the actual task performed during the pre- and post-training assessment. Kurz et al. (2016) demonstrated that a training intervention based on unexpected perturbations during walking improves the ability to voluntarily step rapidly in older adults. This is a generalization of reactive balance to a different task, but the non-trained task is not challenging reactive balance directly. Next, Gimmon et al. (2018) showed that a training intervention based on unexpected perturbations during walking induced adaptations in the nominal gait kinematics, but also here it is unclear whether reactive balance performance improved. Studies by Arampatzis et al. (2011) and Bierbaum et al. (2013) showed that older adults who trained on the hip and stepping strategy mechanisms in a functional way improved in an untrained lean-and-release task and untrained perturbed walking task, respectively. The lean-and-release task was again similar to the training exercises and so limited insight on the generalizability of the improved mechanisms to other locomotion tasks is provided. In both studies it is unclear which adapations in the application of the balance-correcting strategies occurred and how these were implemented in order to improve performance in the untrained task.

In this study, we evaluated whether PBT using support-surface perturbations during standing as the training task improved reactive balance performance during perturbed walking and walking on a narrow beam more than RT in healthy older adults. In addition, we explored the effect of both training paradigms on the application of balance-correcting strategies and on sensorimotor acuity. We performed two training interventions with two groups of healthy older adults: a 12-week RT and a 3-week PBT consisting of support-surface perturbations of standing balance. The dosages of both training interventions were in line with common dosages for RT and PBT (Latham et al., 2004; Pai and Bhatt, 2007; Mccrum et al., 2017). For both interventions, the chosen dosage has been shown to induce specific improvements in previous studies (Latham et al., 2004; Dijkstra et al., 2015; Mccrum et al., 2017). Reactive balance performance during standing and walking as well as a set of neuro-muscular properties were assessed pre- and post-intervention. We hypothesized that the training programs would induce specific improvements, where PBT during standing would improve reactive balance performance during standing and RT would improve muscle strength. Therefore, we tested whether:

(1) step incidence in response to perturbations of standing balance at specific perturbation magnitudes decreased more after PBT than after RT;
(2) maximal isometric strength improved after RT, but not after PBT.

We hypothesized that standing PBT outperformed RT in improving reactive balance control during tasks that were not included in the training. More specifically we hypothesized that:

(3) step length corrections during perturbed walking decreased more after PBT than after RT;
(4) the distance covered in a narrow-beam walking task increased more after PBT than after RT.

We performed an explorative analysis to describe adaptations in the use of the different balance-correcting strategies to perturbed standing. In addition, we explored whether RT and PBT induced changes in sensory and motor acuity.

METHODS

Two groups of healthy older adults (>65 y) participated in a longitudinal study that consisted of a training intervention with pre- and post-intervention assessment sessions (**Figure 1**). Participants had not been enrolled in similar training programs previously. Assessors were for practical reasons not blinded, which is a limitation of the presented study. The perturbation-based training (PBT) group consisted of 16 individuals that followed a perturbation-based balance training program during 3 weeks. The resistance training (RT) group consisted of 14 individuals that were part of a larger group participating in a study that was focused on observing effects of omega-3 supplementation during a 12-week RT training program on muscle strength, and muscle anabolic insensitivity (Dalle et al., 2021). Those that indicated their interest, combined the RT training program with the assessment sessions of the present study. The PBT group was not omega-3 supplemented, but it

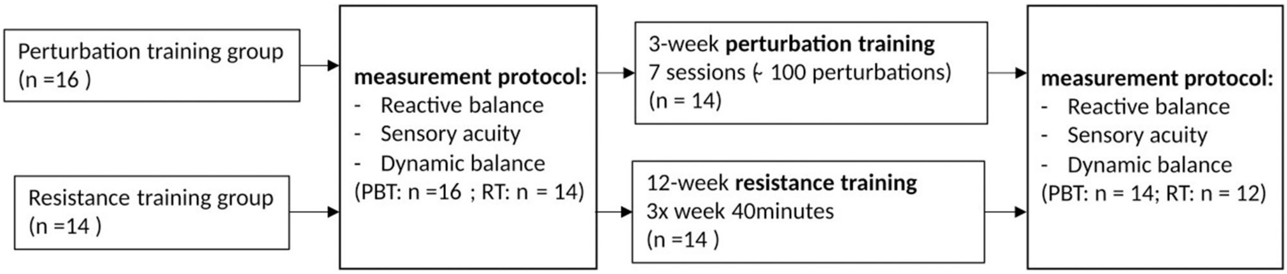

FIGURE 1 | Flow chart of intervention study with pre- and post-intervention measurements.

was assumed that this difference could not directly account for any of the differences in reactive balance performance between groups. Sample sizes were based on previous studies that demonstrated specific improvements for both resistance training (Latham et al., 2004) and perturbation-based training (Dijkstra et al., 2015). A limitation was that no sample size estimation based on expected effect sizes for non-specific improvements was performed as we did not dispose of estimated effect sizes for these potential improvements.

To qualify the group of older adults as healthy, they performed a 5x sit-to-stand test (Cesari et al., 2009) and a test measuring handgrip strength (Dodds et al., 2014) and filled out the Fall Efficacy Scale-International (FES-I) questionnaire (Yardley et al., 2005; Kurz et al., 2016). None of the included older adults scored under the cut-off points for both the 5x sit-to-stand test or hand grip strength test that would indicate frailty (Cruz-Jentoft et al., 2019). None of the participants had a "high concern" score for the FES-I. Participants that suffered from musculoskeletal injury or pathologies that could impair balance were excluded.

From the PBT group, two participants did not complete the full intervention (one participant due to COVID-19 impact and another participant because they were diagnosed with health issues affecting balance control during the intervention and therefore no longer met the inclusion criteria). From the RT group, one participant did not participate in the post-intervention assessment and one participant was excluded due to lower back pain problems during the post-intervention assessment.

The assessment performed pre- and post-intervention served to quantify reactive balance performance and balance-correcting strategies during standing and walking, muscle strength and sensory acuity. We quantified standing reactive balance performance and strategies (CAREN platform, Motek); walking reactive balance performance and strategies (instrumented treadmill); maximal isometric knee-extension torque (Biodex); motor acuity (Biodex); sensory acuity (NeuroCom Balance Master); and dynamic balance [beam walking test (Mansfield et al., 2015)].

Reactive Balance

Standing Reactive Balance Protocol

To assess reactive balance performance during standing we quantified the step incidence for specific perturbation directions and magnitudes, where perturbations were applied as support-surface translations using a CAREN platform (Motek Medical, Netherlands) (Van Wouwe et al., 2020). Participants stood barefoot on the movable platform with their feet at shoulder width looking forward and wore a safety harness to catch them in case of a loss of balance. Participants were instructed to maintain balance without taking a step when perturbed and were allowed to move their arms freely. If the perturbation elicited a stepping response, participants were instructed to return themselves to their original position before the next perturbation. To standardize foot placement, the heel position was marked on the platform. Participants received support-surface perturbations in six directions: anterior and posterior translations, lateral left and right translations, and pitch rotations in two directions inducing either ankle plantar- or dorsiflexion. The protocol consisted of a familiarization part and a randomized part. Subjects were first familiarized with the motion of the platform while being informed on the direction of the upcoming perturbation. During this familiarization, perturbations were applied with progressively larger magnitudes until subjects needed to take a step, which ended familiarization with the specific perturbation direction. The first perturbation magnitude that induced a step response was the highest magnitude included in the second, randomized part of the protocol. When no step response was evoked at the highest magnitude, all perturbations for that direction were included. Up to six different perturbation magnitudes were presented for posterior translations, whereas up to four different perturbation magnitudes were presented in the other directions (**Figure 2**). Next, during the randomized part of the protocol, each perturbation condition was applied five times in random order. Perturbations were provided in random order to minimize anticipatory postural adjustments. We quantified step incidence for the backward and forward platform translations at each presented perturbation intensity.

Walking Reactive Balance Protocol

To assess reactive balance performance during treadmill walking we quantified changes in step length in response to treadmill belt accelerations and decelerations with four different magnitudes (Afschrift et al., 2019). The protocol is the same as the part of the protocol with sagittal perturbations in the study by Afschrift et al. (2019). Participants walked with shoes on an instrumented treadmill and wore a safety harness to prevent falling in case of a loss of balance. The protocol consisted

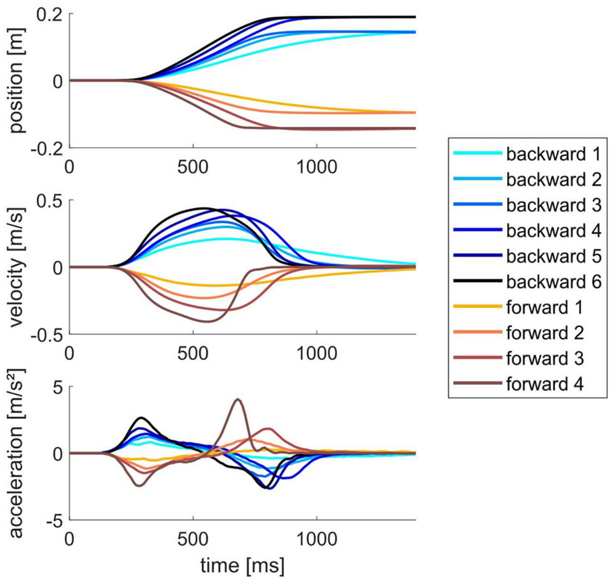

FIGURE 2 | Position, velocity and acceleration profiles of the platform for the six backward and four forward support-surface translations.

of a familiarization part and unperturbed walking part and a randomized perturbation part (perturbed walking). During the familiarization part subjects walked on the treadmill until they were comfortable at the baseline speed of 1.0 m/s, a speed at which healthy older adults are comfortable to execute the whole protocol (Afschrift et al., 2019). During the unperturbed walking part, subjects got used to treadmill walking at a speed of 1.0 m/s for 2 min and their baseline walking pattern was collected. During the randomized perturbation part, subjects were exposed to 16 different perturbations, belt accelerations or decelerations with four different magnitudes applied at two different instants in the gait cycle. Perturbations were applied immediately after left heel strike (7.5% gait cycle, first double support), and during late stance (37.5% gait cycle). These timings and perturbation magnitudes were motivated based on the work of Afschrift et al. (2019) that showed largest differences between young and older adults for these perturbations within a larger set of timings at which perturbations were applied. The order of the perturbations was randomized. A next perturbation was applied when the operator indicated that it was safe to apply a perturbation to the participant and the participant had reached steady state walking, for which the standard deviation of the stride time of the last five strides had to be below 0.05 s.

Motion Capture Data During Standing and Walking

Motion capture data was collected during both reactive standing and walking balance assessments. Subjects were instrumented with 33 reflective markers on anatomical landmarks (full body plug-in-gait) and cluster markers on the left and right shanks and thighs. Platform motion during perturbations of standing was measured using three markers. The marker trajectories were captured using seven and fifteen Vicon cameras during standing and walking, respectively, at a frequency of 100 Hz. Both the treadmill and CAREN platform were instrumented with two force plates, measuring contact forces and moments between the subjects and the support-surface at 1,000 Hz. A static trial in anatomical position was acquired before starting the experiments.

Data was preprocessed to get joint, COM, and COP kinematics and joint kinetics. All marker trajectories were labeled in Vicon Nexus 2.4. Generic musculoskeletal models (gait2392 - OpenSim 3.3) were scaled based on the subject mass and anatomic marker positions acquired during the static trial (Delp et al., 2007; Seth et al., 2018). Joint angles were computed using OpenSim's Inverse Kinematics tool (OpenSim3.3). Finally, OpenSim's Body Kinematics tool was used to compute segment and whole body kinematics. Center of pressure (COP) locations were derived from the forces and moments recorded by the force plates. Joint kinetics were computed using OpenSim's Inverse Dynamics tool (OpenSim3.3), with the scaled musculoskeletal models, force plate data and joint kinematics as input. A correction of the force plate data was performed to remove forces and moments registered due to the inertia of the force plate (Roberts et al., 2019). We corrected for these forces and moments by subtracting the forces and moments registered while the platform was moving without any load on it from the data acquired with the subject on the platform (Van Wouwe et al., 2020).

Outcome Variables for Reactive Balance Performance During Standing and Walking

During perturbed standing, step incidence within each perturbation type was computed by detecting trials in which the vertical ground reaction force (GRF) was below 10 N during more than 50 ms. The step incidence at the largest perturbation magnitude applied pre-intervention was used as outcome variable for both pre -and post-intervention assessments.

During perturbed walking, step length was computed as the sagittal plane distance between the left and right ankle joint center at heel-strike, where heel-strike was defined as the first instant at which the vertical GRF was higher than 10 N after swing phase. Rather than using the step length during unperturbed walking as reference we computed the reference step length as the average length over all last steps before a perturbation during perturbed walking. We did this because some subjects adapted their step length during perturbed walking compared to the familiarization part of the protocol.

Outcome Variables to Quantify Reliance on Balance-Correcting Strategies

To explore adaptations in sensorimotor transformation that lead to changes in the application of the different balance-correcting strategies, we quantified the strategies as described in the following paragraphs.

The reliance on the COP strategy was quantified by the feedback gain (K_{COP}) that linearly relates the deviation of the delayed (100 ms) anteroposterior extrapolated COM position

($\Delta xCOM$) with the corrective ankle joint torque (ΔT_A) (Afschrift et al., 2021):

$$\Delta T_A\ (t) = K_{COP}\Delta xCOM\ (t - 100ms)\ \ with\ \ t = 50\ldots200ms$$

We reduced the feedback model used in (Afschrift et al., 2021), to the $xCOM$ rather than both COM position and velocity in order to only have a single feedback gain quantifying the sensorimotor transformation. During standing, deviations of T_A and $xCOM$ were computed with respect to T_A and $xCOM$ at perturbation onset. All quantities and results were non-dimensionalized using COM height during quiet standing ($l_{\max}$), the gravitational acceleration (g) and body mass (m). COM positions were normalized by $l_{\max}$ and torques by $mgl_{\max}$ (Gruben and Boehm, 2014). Subject-specific feedback gains were estimated from the measured kinematics and joint moments by solving a robust least squares regression (MATLAB R2020a; "*lmfit*" with robust fitting option), pooling the data of all anterior-posterior perturbation trials (Afschrift et al., 2021). An increase in K_{COP} indicated that the subjects increased their reliance on COP strategies.

The reliance on a hip strategy was quantified by the relation between the $xCOM$ position 300 ms after perturbation onset ($xCOM_{300ms}$) and the maximal trunk lean angle $\theta_{trunk,\max}$ during non-stepping responses K_{hip} (Van Wouwe et al., 2020). The maximal trunk lean angle $\theta_{trunk,\max}$ is a measure of the reliance on a hip strategy for a specific trial. However, $\theta_{trunk,\max}$ depends on the perturbation magnitude and, as we recently demonstrated, the initial posture of the subject (Van Wouwe et al., 2020). $xCOM$ 300 ms after perturbation onset ($xCOM_{300ms}$) captures the effect of both the perturbation magnitude and initial posture (Van Wouwe et al., 2020). Therefore, the relation between $\theta_{trunk,\max}$ and $xCOM_{300ms}$ established based on several perturbation trials better quantifies an individual's reliance on a hip strategy than $\theta_{trunk,\max}$. To allow for better comparison between subjects $xCOM_{300ms}$ was normalized by the subject-specific BOS ($xCOM_{300ms}/BOS$). The BOS at perturbation onset was computed as the horizontal distance from the toes to the ankle joint, for the anterior direction, and from the heel to the ankle joint for the posterior direction. Subject specific robust linear regression models were generated for each individual with $\theta_{trunk,\max}$ as outcome variable and $xCOM_{300ms}/BOS$ as predictor variable. The model had a fixed intercept and the slope coefficients were variable with respect to the categorical variable time (pre vs. post). An increase in slope coefficient (K_{hip}) indicated that the subjects increased their reliance on hip strategies.

The reliance on a stepping strategy was quantified by the stepping threshold, which was defined as the maximal extrapolated COM excursion in non-stepping trials. We normalized again by the base-of-support $xCOM_{\max,non-stepping}/BOS$. This outcome variable captures how strongly a subject's balance was disturbed before they initiated a stepping response. For each non-stepping trial, we computed the largest within trial $xCOM$ values. For each subject, pre -and post-intervention, we computed the mean over the three largest values $xCOM_{\max,non-stepping}/BOS$. An increase of $xCOM_{\max,non-stepping}/BOS$ post-intervention indicated that subjects increased their stepping threshold and thus relied less on a stepping strategy.

Muscle Strength

Muscle strength was assessed for the knee-extensor muscles on a Biodex System 4 PRO dynamometer (Shirley, NY). Subjects were seated on the Biodex equipment with their knee at a 90° flexion angle with the back support in fully upright position. Subjects performed three maximal voluntary isometric knee extension contractions of 3 s with 1-min rest between trials. The torque was measured at 1,000 Hz. From these three trials the maximal voluntary isometric knee extension torque [MVIKT (Nm)], normalized by the body weight, was computed.

Motor Acuity

Motor acuity was tested by measuring force fluctuations during submaximal isometric knee extension (Singh et al., 2010, 2012, 2013). Force fluctuations were tested at 15 and 20% of the measured maximal isometric knee extension torque. To assess force fluctuations three different torque tracking tasks where executed three times in random order. Task 1 and 2 consisted of generating a constant torque for 15 s at, respectively, the 15 and 20% level, task 3 consisted of tracking a ramp-up torque from the 15 to 20% level during 15 s. The target torque profile was displayed on a monitor and participants were instructed to match the torque level as well as they could for the duration of each test by generating knee-extension torque at 90° flexion angle. The torque generated by the subjects (GT) was overlaid in real-time on the target torque (TT).

The recorded torque profiles were first low-pass filtered using a fourth-order low-pass Butterworth filter with a cut-off frequency of 25 Hz (Singh et al., 2013). Force fluctuation for each of the trials was computed using the normalized standard deviation (SD) of the absolute error (NSAE) between the target and generated torque during the middle 10 s of each trial (Christou and Carlton, 2001; Singh et al., 2010):

$$NSAE = SD(TT\ (t) - GT(t))/mean(GT\ (t))$$

A composite score to quantify motor acuity for each subject was then computed by taking the average over the nine trials.

Sensory Acuity

Sensory acuity in the context of balance control was tested by performing a sensory organization test (SOT) (Nashner and Peters, 1990) on a NeuroCom Balance Master, yielding a composite score, subscores to quantify visual, vestibular and proprioceptive acuity and a preference score that quantifies the subjects' ability to organize and select the appropriate sensory information to maintain balance (Ford-Smith et al., 1995). The composite score was the main outcome measure, but subscores were analyzed as well.

Dynamic Balance

To quantify non-reactive dynamic balance control, subjects performed a narrow beam walking task (Speers et al., 1998; Sawers and Hafner, 2018a). Subjects were asked to walk on a narrow 3.66 m long beam (width: 2 cm; height: 2 cm) six times

(Sawers and Hafner, 2018a). The only instructions were to start with one foot on the beam with the heel lining up with the start of the beam and to maximize the distance covered on the beam without touching the ground. The distance was measured between the start of the beam and the position where one of the feet touched the ground or the end of the beam if subjects touched the ground beyond the end of the beam. An overall score was computed by summing the distances of all six trials and dividing this sum by the length of the beam (21.96 m), yielding a percentage score of the maximum distance that could have been covered. Before their first measured trial, subjects performed four familiarization trials. Subjects were free to use their arms.

12-Week Resistance Training

The 12-week supervised RT program for the lower limbs consisted of three 40-min sessions per week (Dalle et al., 2021). Each training session consisted of a 10-min warm-up on a bicycle ergometer at low intensity followed by the RT program. The RT program consisted of a leg press exercise, a leg extension exercise and calf raises. During the first 6 weeks, participants performed two sets of 12–15 repetitions at about 70% of the one repetition maximum with 1 min rest between sets and 2 min rest between exercises. During the last 6 weeks, participants completed three sets of 10–12 repetitions at ∼80% of the one repetition maximum. Participants performed these exercises at moderate velocity, i.e., 3 s for the concentric and eccentric phase.

3-Week Standing Perturbation Training

The 3-week supervised PBT program consisted of 7 sessions in total excluding the assessment sessions. Participants stood on a movable platform and received unpredictable support-surface translations. The participants were instructed to maintain balance without taking a step. About 100 perturbations were performed each session. Similar to during assessment, support-surface perturbation were applied randomly in six directions. For each perturbation direction, different magnitudes were applied depending on the participant's level. The maximal perturbation magnitude for each perturbation direction during the first training session was determined during the familiarization part of the pre-intervention assessment. The stepping frequency for each perturbation type and magnitude was determined during each session. Based on this information, the perturbation magnitudes and number of repetitions for each perturbation magnitude for the next session were determined to increase the difficulty each session. When subjects exhibited step incidence below 25% for the largest included perturbation magnitude in a specific direction, the next session contained a perturbation with larger magnitude for the same direction. The number of perturbations for each magnitude within a perturbation direction was such that higher magnitudes were applied more than lower magnitudes.

Statistical Analysis

An overview of all outcome variables is provided in **Table 1**. For the statistical analysis we reported data as means ± SD. Normality of data was tested by applying a Shapiro-Wilkes test for all outcome variables for within intervention group pre- to post-intervention differences in outcome variables.

TABLE 1 | Description of outcome variables.

Outcome variable	Measured for/by	Quantifies
Step incidence	Anterior and posterior perturbations of standing with different perturbation magnitudes	Reactive standing balance performance
Step length correction	Anterior and posterior perturbations of walking with different magnitudes at different instances of the stance phase	Reactive walking balance performance
K_{COP}	Anterior and posterior perturbations of standing with different perturbation magnitudes	Reliance on COP strategy: sensorimotor transformation from COM kinematics to ankle torque during standing
K_{hip}	Anterior and posterior perturbations of standing with different perturbation magnitudes	Reliance on hip strategy: relation between $xCOM_{300ms}$ and $\theta_{trunk,\ max}$
$xCOM_{max,non-stepping}/BOS$	Anterior and posterior perturbations of standing with different perturbation magnitudes	Reliance on stepping strategy: maximum *xCOM* excursion withstanded without initiating a step response
Maximal voluntary isometric knee-extension torque (MVIKT)	Maximum over three trials	Knee-extensor strength
NSAE of force fluctuations	Averaged over nine torque tracking tasks on the Biodex dynamometer	Motor acuity
SOT composite and subscores	Sensory organization test	Sensory acuity and organization during balance control
Beam walking score	Averaged over six trials of narrow-beam walking	Dynamic balance

Repeated-measures ANOVA (Matlab R2020a "ranova") with time (pre- vs. post-intervention) as within-subject factor and intervention group (RT vs. PBT) as between-subject factor was performed to test for time (pre- vs. post-intervention) and time*intervention interaction effects. *p*-value corrections for violations of compound symmetry within the repeated-measures ANOVA model were performed. When a significant interaction effect was observed, paired *t*-tests within intervention groups were executed to detect changes pre- to post-intervention within intervention groups.

RESULTS

At baseline the resistance training group had a MVIKT that was 20% lower than the perturbation training group (**Table 2**). For all other outcome parameters and for age, body mass, length and BMI the two groups did not differ. Results for all outcome variables are reported in **Tables 3**, **4**.

TABLE 2 | Baseline comparison for the included groups of healthy older adults.

	Resistance training	Perturbation-based training	*p*-value
Age (y)	72.2 ± 3.56	70.8 ± 4.6	0.41
Gender	7 male; 5 female	7 male; 7 female	
Body mass (kg)	73.1 ± 8.1	70.7 ± 14.0	0.60
Length (cm)	167 ± 7.3	169 ± 9.4	0.61
BMI (kg/m²)	26.2 ± 2.4	24.6 ± 3.0	0.15
Step incidence (backward)	0.79 ± 0.35	0.84 ± 0.36	0.66
Step incidence (forward)	0.87 ± 0.28	0.93 ± 0.19	0.40
MVIKT *(Nm/kg)*	1.85 ± 0.38	2.32 ± 0.54	0.03*
NSAE (%)	1.79 ± 0.46	1.72 ± 0.39	0.10
SOT	68.2 ± 9.63	71.4 ± 6.79	0.32
Beam walking (%)	30 ± 15	36 ± 13	0.32
Δl_{step} *(m) early stance (backward)*	0.062 ± 0.032	0.065 ± 0.017	0.70
Δl_{step} *(m) early stance (forward)*	−0.12 ± 0.040	−0.13 ± 0.049	0.77
Δl_{step}*(m) late stance (backward)*	0.078 ± 0.030	0.068 ± 0.029	0.48
Δl_{step}*(m) late stance (forward)*	−0.003 ± 0.057	−0.032 ± 0.045	0.22
K_{COP}	0.66 ± 0.20	0.65 ± 0.10	0.39
K_{hip}	32.6 ± 35.5	35.2 ± 21.5	0.34
$xCOM_{max,non-stepping}/BOS$	1.13 ± 0.19	1.14 ± 0.18	0.70

**significance at 0.05 alpha level.*

Summary of the Main Observations

Both PBT and RT induced training specific improvements. PBT reduced step incidence during backward perturbations of standing more than RT (hypothesis 1; **Figure 3**), while RT increased maximal strength of the knee extensors whereas PBT did not (hypothesis 2).

Reduced step incidence after PBT could not be explained by changes in the sensorimotor transformation from xCOM to T_A that determine the reliance on the COP strategy (**Figure 3**). Exploring other potential mechanisms reveals that an increased stepping threshold might have reduced step incidence after PBT, whereas an increased reliance on hip strategies might have contributed to the reduced step incidence in part of the subjects (**Figure 3**).

Improvements in balance during standing after PBT do not seem to generalize to perturbed walking or narrow beam walking. Neither PBT nor RT induced changes in step length corrections during perturbations of walking (hypothesis 3). Finally, beam walking performance improved significantly after the interventions but the improvements after PBT and RT were not different (hypothesis 4).

Reactive Balance

Standing Reactive Balance

Step incidence in response to backward platform translations (backward perturbations) of standing presented both a time main effect [$F_{(1, 24)} = 22.8$, $p < 0.001$] and time*intervention interaction effect [$F_{(1, 24)} = 6.5$, $p = 0.0175$]. Both RT [$t_{(11)} = 2.55$, $p = 0.0254$] and PBT [$t_{(13)} = 4.20$, $p = 0.0011$] induced a significant reduction in step incidence but PBT induced a significantly larger pre- to post-intervention reduction in step incidence, confirming hypothesis 1 for backward perturbations.

Step incidence in response to forward platform translations (forward perturbations) presented a time main-effect [$F_{(1, 24)} = 16.1$, $p < 0.001$], but not a time*intervention interaction effect [$F_{(1, 24)} = 2.5$, $p = 0.13$]. Step incidence decreased significantly after the intervention, but the decrease was not different between PBT and RT.

K_{COP} during perturbed standing did not present a time main effect [$F_{(1, 24)} = 0.29$, $p = 0.59$] nor a time*intervention interaction effect [$F_{(1, 24)} = 0.06$, $p = 0.81$] indicating that neither intervention induced a change in the sensorimotor transformation between $xCOM$ *and* T_A. The subject-specific linear regression models fitted the data well with mean R-squared values over all subjects being 0.90, 0.88, 0.90, and 0.89 for the RT pre-intervention, RT post-intervention, PBT pre-intervention and PBT post-intervention data, respectively. The minimal R-squared value was 0.73. These findings indicate that PBT does not increase the reliance on COP strategies.

Reliance on hip strategies was quantified by the slope coefficient K_{hip} that linearly relates the $xCOM_{300ms}$ and $\theta_{trunk,\,\max}$. We did not detect significant changes in K_{hip}, i.e., no time main-effect [$F_{(1, 24)} = 3.3$, $p = 0.081$] nor a time*intervention interaction-effect [$F_{(1, 25)} = 3.0$, $p = 0.096$].

PBT induced significantly larger increases in stepping thresholds than RT. $xCOM_{\max,non-stepping}/BOS$ showed both a time main effect [$F_{(1, 24)} = 12.8$, $p = 0.0015$] and a time*intervention [$F_{(1, 24)} = 4.6$, $p = 0.043$]. $xCOM_{\max,non-stepping}/BOS$ increased significantly in the PBT group [$t_{(13)} = -3.65$, $p = 0.0029$], but not in the RT group [$t_{(11)} = -1.25$, $p = 0.23$]. This indicates that older adults increase their stepping threshold after PBT but not after RT.

Walking Reactive Balance

Step length changes in response to perturbations did not change after PBT or RT. Step length corrections during perturbed walking did not present a time main effect nor time*intervention interaction effect during treadmill belt accelerations in early [$F_{\text{time}(1, 24)} = 1.13$, $p_{\text{time}} = 0.31$; $F_{\text{time*intervention}(1, 24)} = 0.03$, $p_{\text{time*intervention}} = 0.86$] and late stance [$F_{\text{time}(1, 24)} = 1.14$, $p_{\text{time}} = 0.30$; $F_{\text{time*intervention}(1, 24)} = 1.17$, $p_{\text{time*intervention}} = 0.16$] and during treadmill belt decelerations during early [$F_{\text{time}(1, 24)} = 0.003$, $p_{\text{time}} = 0.96$; $F_{\text{time*intervention}(1, 24)} = 0.002$, $p_{\text{time*intervention}} = 0.97$] and late stance [$F_{\text{time}(1, 24)} = 0.30$, $p_{\text{time}} = 0.59$; $F_{\text{time*intervention}(1, 24)} = 1.97$, $p_{\text{time*intervention}} = 0.18$]. We thus reject the hypothesis that the correction in step length decreased more after PBT than after RT (hypothesis 3).

TABLE 3 | Pre -to post-training comparisons for different outcome variables that were defined a-priori.

	Resistance training		Perturbation-based training		*P*-value
	Pre	**Post**	**Pre**	**Post**	
Step incidence (backward)	0.79 ± 0.35	0.63 ± 0.41* −0.25 ± 0.38⁺	0.84 ± 0.36	0.34 ± 0.40** −0.50 ± 0.46	$p < 0.001^a$ $p = 0.0175^b$
Step incidence (forward)	0.87 ± 0.28	0.71 ± 0.37 −0.15 ± 0.33	0.93 ± 0.19	0.58 ± 0.35** −0.35 ± 0.33	$p < 0.001^a$ NS ($p = 0.13$)[b]
MVIKT *(Nm/kg)*	1.85 ± 0.38	2.15 ± 0.38*** 0.30 ± 0.17⁺⁺⁺	2.32 ± 0.54	2.27 ± 0.43 −0.043 ± 0.24	$p = 0.0044^a$ $p < 0.001^b$
NSAE (%)	1.79 ± 0.46	1.66 ± 0.33 −0.14 ± 0.33	1.72 ± 0.39	1.51 ± 0.40 −0.21 ± 0.38	$p = 0.0198^a$ NS ($p = 0.61$)[b]
SOT	68.2 ± 9.63	73.2 ± 6.58* 5.00 ± 6.61	71.4 ± 6.79	76.1 ± 6.73** 4.64 ± 5.00	$p < 0.001^a$ NS ($p = 0.87$)[b]
Beam walking (%)	30 ± 15	33 ± 20 2.4 ± 11	36 ± 13	42 ± 17** 6.0 ± 6.4	$p = 0.021^a$ NS ($p = 0.31$)[b]
Δl_{step} *(m) early stance (backward)*	0.062 ± 0.032	0.056 ± 0.039	0.065 ± 0.017	0.027 ± 0.037	NS ($p = 0.30^a$) NS ($p = 0.86^b$)
Δl_{step}*(m) early stance (forward)*	−0.12 ± 0.040	−0.12 ± 0.024	−0.13 ± 0.049	−0.13 ± 0.035	NS ($p = 0.96^a$) NS ($p = 0.97^b$)
Δl_{step}*(m) late stance (backward)*	0.078 ± 0.030	0.061 ± 0.024	0.068 ± 0.029	0.070 ± 0.023	NS ($p = 0.30^a$) NS ($p = 0.16^b$)
Δl_{step}*(m) late stance (forward)*	−0.003 ± 0.057	−0.015 ± 0.06	−0.032 ± 0.045	−0.005 ± 0.046	NS ($p = 0.59^a$) NS ($p = 0.18^b$)

All statistical model residuals were normally distributed, and so repeated-measures ANOVA and parametric tests were performed for the different outcome variables. Values are means ± SD. Where relevant the changes from pre- to post-assessment values are reported on the second line in the post-assessment column. p-values obtained by repeated-measures ANOVA with time as within-subject effect and training as between-subject effect.

[a]*Main time-effect (pre -to post-training).*

[b]*Time*intervention interaction effect.*

****p < 0.001, **p < 0.005, *p < 0.05 significant difference within group from pre -to post-training. ⁺⁺⁺p < 0.001, ⁺p < 0.05 significant difference in pre- to post-training changes between PBT and RT groups.*

TABLE 4 | Exploratory outcome variables.

K_{COP}	0.66 ± 0.20	0.63 ± 0.14 −0.036 ± 0.14	0.65 ± 0.10	0.63 ± 0.17 −0.022 ± 0.15	NS ($p = 0.59$)[a] NS ($p = 0.81$)[b]
K_{hip}	32.6 ± 35.5	32.8 ± 30.8	35.2 ± 21.5	43.2 ± 29.6	NS ($p = 0.0813^a$) NS ($p = 0.096^b$)
$xCOM_{max,non-stepping}/BOS$	1.13 ± 0.19	1.18 ± 0.13 6.2 ± 20.1%⁺	1.14 ± 0.18	1.27 ± 0.17** 13.6 ± 16.3%	$p = 0.0015^a$ $p = 0.043^b$

[a]*Main time-effect (pre -to post-training).*

[b]*Time*intervention interaction effect.*

***p < 0.005 significant difference within group from pre -to post-training. ⁺p < 0.05 significant difference in pre- to post-training changes between PBT and RT groups.*

Maximal Strength

MVIKT presented both a time main effect [$F_{(1,\,24)} = 9.91$, $p = 0.0044$] and time*intervention interaction effect [$F_{(1,\,24)} = 17.6$, $p < 0.001$], indicating that RT induced larger improvements in normalized MVIKT than PBT, confirming hypothesis 2. RT induced significant increases of normalized MVIKT comparing pre- to post-measurements [$t_{(11)} = 6.3$, $p < 0.001$]. PBT did not result in significant increases of MVIKT from pre- to post-intervention [$t_{(13)} = 0.67$, $p = 0.51$].

Narrow Beam Walking

The beam walking score presented a time main effect [$F_{(1,\,24)} = 6.0$, $p = 0.021$] but not a time*intervention interaction effect [$F_{(1,\,24)} = 1.1$, $p = 0.31$]. Subjects walked further on the beam post- than pre-interventions but these improvements were not significantly different between the two interventions. We thus reject our hypothesis that the distance covered in a narrow-beam walking task increased more after PBT than after RT (hypothesis 4).

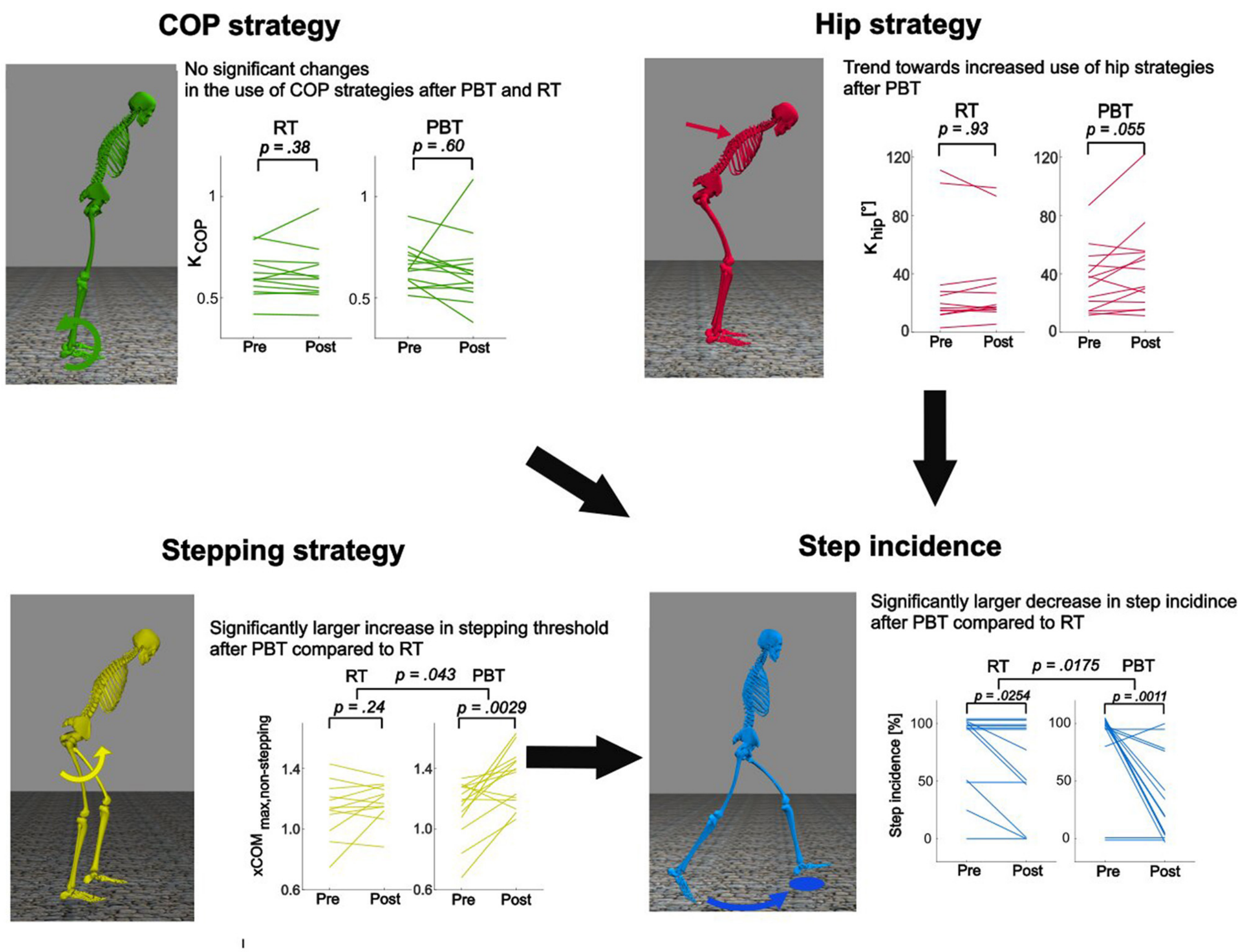

FIGURE 3 | Effects of resistance training (RT) and perturbation-based training (PBT) on reactive standing balance performance and strategies. Step incidence during backward perturbations of standing decreased more after PBT than after RT. Improvements in step incidence were not explained by changes in the COP strategy defined by the sensorimotor transformation from xCOM to T_A. A trend toward increased use of hip strategies might have contributed to the decreased step incidence. Increased reliance on hip strategies was observed in some but not all individuals. Most likely, decreased step incidence might be due to changes in reliance on step strategies quantified by a significant change in stepping threshold after PBT that was not observed after RT.

Motor Acuity

NSAE presented a time main effect [$F_{(1, 24)} = 6.2, p = 0.0198$] but no time*intervention interaction effect [$F_{(1, 24)} = 0.27, p = 0.61$].

Sensory Acuity

Composite SOT scores presented a time main effect [$F_{(1, 24)} = 313.4, p < 0.001$] but not a time*intervention interaction effect [$F_{(1, 24)} = 0.43, p = 0.88$]. Further analysis of the SOT sub scores revealed no significant changes for the visual and somatosensory scores. The vestibular sub score presented a time main effect [$F_{(1, 24)} = 5.4, p = 0.0285$], but not a time*intervention interaction effect [$F_{(1, 24)} = 3.4, p = 0.076$]. The improvements in the sensory acuity task might thus be the result of a learning effect. The observed changes are not unexpected based on literature reporting on learning effects of this test (Wrisley et al., 2007; DiFrancisco-Donoghue et al., 2015).

DISCUSSION

PBT of standing balance did not improve balance control during non-trained walking tasks in healthy older adults and neither did RT (hypotheses 3 and 4 not confirmed). Both PBT and RT induced training specific improvements, i.e., standing perturbation training improved reactive balance during perturbed standing (hypothesis 1) and resistance training increased strength (hypothesis 2). Improvements in reactive standing balance after PBT, measured in terms of step incidence, were the result of an increased stepping threshold, possibly in combination with increased reliance on hip strategies, but not of an increased reliance on the COP strategy. The strong specificity of PBT should be considered in the design of an intervention as it might lead to limited effects on fall prevention, the ultimate purpose of such interventions. Increasing strength was not effective in improving reactive balance in healthy older adults. This suggests, in line with previous studies, that as long as muscle strength remains above a threshold, it is not the primary limiting factor for reactive balance.

Training-Induced Alterations in Reactive Balance Strategies in Older Adults

Subjects increased their stepping threshold and tended to rely more on hip strategies after PBT, suggesting that the lower stepping threshold in older adults compared to young adults (Pai et al., 1998) can thus be increased through PBT. Similar to previous intervention studies that targeted perturbed standing (Dijkstra et al., 2015), and perturbed walking (Pai and Bhatt, 2007; Sakai et al., 2008), we found that PBT improved reactive balance performance for the trained task. Our observations suggest that the decreased step incidence was due to changes in control strategy, i.e., the translation of sensory information to motor commands, rather than changes in muscle strength or sensorimotor acuity. We indeed found that subjects increased their stepping threshold, expressed as the maximum *xCOM* excursion resisted without initiating a step response, by 13.6% on average after PBT. This indicates a change in the sensorimotor transformation that coordinates the muscles to initiate a step response at a specific threshold of balance disturbance.

The increase in stepping threshold after PBT, might be interpreted as relying on a higher risk strategy by allowing a larger disturbance of balance from equilibrium before initiating a step. The repeated PBT might decrease the fear that healthy older adults experience during the platform perturbations, which might allow them to better achieve the task goal of suppressing a step response. However, subjects' fear of falling questioned using a FES-I questionnaire showed low fear of falling both pre- and post-intervention in both groups.

Alternatively, step incidence can be reduced without changing the step initiation threshold by relying more on COP and hip strategies. On the one hand, we found limited changes in the application of COP strategies after RT and PBT. This might indicate that the COP strategy is limited by a factor that is not affected by the specific PBT or RT training. For example, intrinsic foot muscle capacities might be a factor limiting the COP strategy in older adults (Koyama and Yamauchi, 2017; Zhang et al., 2017). On the other hand, we found that some, but not all, subjects of the PBT group increased their reliance on hip strategies. When examining the subject-specific linear regression models we could observe whether specific subjects increased their reliance on hip strategies significantly. In the PBT group this was the case for eight out of fourteen subjects, in the RT this was the case for four out of twelve subjects. These findings indicate that PBT might induce an increased reliance on hip strategies in some subjects. Our previous simulation study showed that inter-subject differences in reliance on the hip strategy can be explained by differences in the trade-off between effort and stability in a group of young adults. Similar differences might be present in older adults (Van Wouwe et al., 2020) that might shift their strategy to maximize stability in exchange for increased effort. It remains to be investigated whether explicitly coaching subjects to rely more on a hip strategy, would have reduced step incidence further.

It is unlikely that improvements in reactive balance performance during standing were the result of changes in sensory, although we cannot exclude this based on our observations. We found improvements in sensory acuity as measured by SOT in both PBT and RT groups. It is unlikely that these improvements in SOT reflect changes in the sensory system given the short duration of the PBT, the focus of RT on strength, and the absence of differences between PBT and RT. Changes in SOT scores post- vs. pre-intervention might reflect a learning effect, in line with previous findings (Wrisley et al., 2007; DiFrancisco-Donoghue et al., 2015).

Although motor acuity measures improved after both PBT and RT, it is unlikely that these explain the changes in reactive balance performance. Our assessment of motor acuity was indirect as it involved a torque-tracking task and it is therefore possible that there was a learning effect. Considering the short duration of PBT since it is unlikely that physiological changes took place. In contrast, RT induced physiological changes in the muscle and might therefore have had an effect on motor acuity, yet changes after PBT and RT did not differ.

Increasing muscle strength should not be the main target when aiming to improve reactive balance in healthy older adults. RT decreased step incidence during a reactive standing balance task but to a smaller extent than PBT, which did not alter muscle strength. These findings are in line with previous intervention studies that found little or no improvements in fall risk (Faber et al., 2006; Cadore et al., 2014; Fairhall et al., 2014; De Labra et al., 2015) or reactive balance (Hess et al., 2006) following RT. It is likely that learning effects, rather than increases in strength after RT, explain the observed reduction in step incidence after RT. Indeed, the pre- and post-assessments of reactive standing balance can be considered a perturbation-based training session and PBT had a large effect on step incidence. Although healthy older adults use stepping strategies at lower perturbation magnitudes than young adults that are stronger (Pai et al., 1998), these alterations in reactive balance might not be due to reductions in strength. Simulation studies that have elicited causal relations between muscle strength and balance-correcting responses following perturbations of standing (Robinovitch et al., 2002; Mackey and Robinovitch, 2006; Afschrift et al., 2015), indicate that muscle strength influences the efficacy of the COP strategy. Increases in the rate of force development allow the COP to shift faster, and increases in maximal ankle plantar flexion torque allow the COP to shift further toward the edge of the BOS. Our exploratory analysis did not reveal any changes in the application of the COP strategy after RT. This is in line with our prior simulation study, which suggests that only severe muscle weakness, not present in our group of community -dwelling healthy older adults, limits the COP strategy (Afschrift et al., 2015). Hence, muscle strength might not have limited the use of

COP strategies in our group of healthy older adults explaining why increases in strength after RT did not result in improved reactive balance performance. Note that ankle plantarflexion strength might be more relevant for reactive standing balance than knee-extensor strength, which was assessed in this study. We do, however, expect that RT induced similar improvements in maximal ankle plantarflexion and knee-extension torque given that the RT protocol imposed similar training demands on both muscle groups.

Specific Adaptations Induced by PBT and RT Do Not Improve Balance in Non-trained Tasks

Improvements in reactive standing balance after PBT did not generalize to walking balance. PBT studies rarely evaluated reactive balance in tasks different from the trained task and thus little is known on generalizability of PBT. van Duijnhoven et al. (2018) used a training exercise paradigm that was similar to the one used in this study, i.e., platform translations, and found improvements in reactive balance performance in a lean-and-release task after training. Since they had no control group, it is unclear whether improvements in the lean-and-release task were due to the perturbation training or to a learning effect. Bierbaum et al. (2013) showed that reactive balance to perturbations during walking improved in a group that performed a 14-week training consisting of functional exercises of the balance-correcting mechanisms but not in a group that combined these same exercises with strength training. The evidence of this study is thus not straightforward but, but similar to our study, these results indicate again that strength should not be the main target of fall prevention but also that generalizability of such reactive balance exercises is complicated to understand. We propose that the analysis of how the different balance-correcting mechanisms are applied throughout the tested reactive balance task, is useful to provide more insight on the generalizability of training interventions.

The reduced reliance on stepping during reactive standing balance did not generalize to walking, which might suggest that different mechanisms drive the selection of a strategy during both tasks. Older adults step more in response to perturbations of standing (Pai et al., 1998; Afschrift et al., 2017) and use larger corrections in step length in response to perturbations of walking (Afschrift et al., 2019) than young adults, which suggests an age-related change in a common mechanism that determines stepping strategies during both standing and walking. Yet, the effect of PBT on stepping during perturbed standing and not during perturbed walking, questions the existence of such a common mechanism. After PBT, the step initiation threshold increased reducing step incidence (Pai et al., 1998). However, healthy older adults did not reduce step length corrections during perturbed walking after PBT, and thus relied similarly on stepping strategies walking pre- and post-intervention. This lack of generalizability might have been related to different instructions, older adults were explicitly instructed to avoid step responses during perturbed standing, whereas no instructions related to step length corrections were given during perturbed walking. Alternatively, the perturbations during perturbed walking might not have been challenging enough to detect a reduced reliance on step length corrections. Yet our protocol was based on our previous work (Afschrift et al., 2019) in which we found differences in step length corrections between young and healthy older adults for the applied forward and backward perturbations. Hence, age-related changes in sensorimotor function or motor control that were not affected by PBT or RT must be at the basis of these age-related increases in step length corrections.

Narrow beam walking performance did not increase more after PBT than after RT. The observed main effect might thus originate from learning effects or both trainings have a similar effect. The difference in training paradigms between PBT and RT would suggest a learning effect, rather than training specific improvements. The task of beam walking challenges reactive balance, but in a very different way than the unexpected perturbations applied during standing and walking. First, the perturbations during beam walking are self-induced. Second, balance is mainly challenged in the frontal plane (Sawers and Hafner, 2018b). The PBT training focused strongly on balance in the sagittal plane. We did provide medio-lateral perturbations during training, but these did not present a strong balance perturbation as standing with the feet at shoulder width is stable. The main purpose of the medio-lateral perturbations was a better randomization, making the perturbation direction more unpredictable.

The two interventions were not standardized in terms of training dosage, which is a limitation when to interpret the efficiency of the two interventions for improving reactive balance. However, the main goal was to analyze whether training specific improvements would generalize to a task that was not trained. Further, we doubt that increasing the dosage of the perturbation training intervention to a dosage similar to the resistance training might lead to better generalization to other tasks, since the achieved improvements in the trained task were already strong. The purpose of this study was not to provide equal dosages of each intervention, but rather to compare two interventions—including their typical dosages—that have previously been shown to induce specific improvements.

CONCLUSION

By comparing the effects of a RT and PBT interventions we aimed to get more insight into the mechanisms through which PBT affects reactive balance. Our findings indicate that the effects of PBT are specific to the trained task, which was standing balance, and do not necessarily generalize to other modes of locomotion. An exploratory analysis suggests that reductions in step incidence following perturbations of standing were the result of changes in the sensorimotor transformation that determine the initiation of a stepping response, rather than adaptations in the application of hip and COP strategies. Muscle strength might play a role in the age-related decrease in reactive balance performance but muscle strength was not the primary factor limiting reactive balance in healthy older adults as reactive standing balance did not improve after increasing strength (RT). In contrast, reactive balance improved after PBT, which did not alter strength. For better insight in the generalizability of training different balance-correcting strategies we suggest quantifying the reliance on these separate strategies for both the trained and assessment tasks in future research.

DATA AVAILABILITY STATEMENT

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

ETHICS STATEMENT

The studies involving human participants were reviewed and approved by Ethical Committee Research UZ/KU Leuven. The patients/participants provided their written informed consent to participate in this study.

AUTHOR CONTRIBUTIONS

TV, MA, SD, EV, KK, and FD conceived and designed research. TV, MA, and SD performed experiments. TV and MA analyzed data. TV, MA, and FD interpreted results. TV prepared figures and drafted manuscript. TV, MA, SD, EV, KK, and FD edited and revised manuscript. All authors contributed to the article and approved the submitted version.

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Perturbation-based balance training: Principles, mechanisms and implementation in clinical practice

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Since the mid-2000s, perturbation-based balance training has been gaining interest as an efficient and effective way to prevent falls in older adults. It has been suggested that this task-specific training approach may present a paradigm shift in fall prevention. In this review, we discuss key concepts and common issues and questions regarding perturbation-based balance training. In doing so, we aim to provide a comprehensive synthesis of the current evidence on the mechanisms, feasibility and efficacy of perturbation-based balance training for researchers and practitioners. We address this in two sections: “Principles and Mechanisms” and “Implementation in Practice.” In the first section, definitions, task-specificity, adaptation and retention mechanisms and the dose-response relationship are discussed. In the second section, issues related to safety, anxiety, evidence in clinical populations (e.g., Parkinson’s disease, stroke), technology and training devices are discussed. Perturbation-based balance training is a promising approach to fall prevention. However, several fundamental and applied aspects of the approach need to be further investigated before it can be widely implemented in clinical practice.

KEYWORDS

aged, slips, trips, gait adaptation, balance disorders, rehabilitation, accidental falls

Introduction

Large mechanical destabilizing disturbances during walking (such as slips and trips) lead to most falls among community-dwelling older adults (1–8). Interventions to reduce falls among older adults and clinical populations with balance impairment have received much attention in the literature, with multiple Cochrane reviews on the topic (9–14). Physical exercise is the most evidence-based approach for preventing falls, with challenging balance exercise among the most successful approaches (13, 15, 16). This aligns with the notion of task-specificity in exercise-based fall prevention (17–24), and the development of perturbation-based balance training (PBT).

Interest in the use of large mechanical perturbations as a method of preventing falls has steadily increased since the mid-2000s. In this period: Pai and Bhatt (18) presented a framework for using repeated slip perturbations to reduce slip-related falls; Grabiner et al. (19) presented evidence and theory on how the task-specific training of limiting trunk motion during slips and trips might reduce fall risk; Oddsson et al. (17) presented a balance training programme with a focus on training specificity, incorporating perturbations; and Mansfield et al. (25) published the first protocol for an RCT of PBT in older adults. Two subsequent large trials showed promising effects of PBT interventions on daily life fall incidence in older adults (26, 27) and another highlighted the clinical feasibility of this approach (28). Subsequent reviews and meta-analyses have further supported these encouraging results (29–33). More recently, a large, pragmatic RCT conducted in a clinical setting (34) and a smaller experimental trial (35) also reported positive fall-related outcomes. In contrast, a recent trial conducted in individuals with chronic stroke reported inconclusive results (36). Further RCTs of PBT are currently underway (37–41).

Despite the accumulating research on PBT, there is much still to be learned. Even so, practitioners are open to implementing PBT (42, 43) and desire more knowledge on the topic (43). In this review, we discuss some of the key concepts and common issues and questions around PBT. In doing so, we aim to provide a comprehensive synthesis of the current evidence on the mechanisms, feasibility, and efficacy of perturbation-based balance training for researchers and practitioners. We address this in two sections: "Principles and Mechanisms" and "Implementation in Practice."

Principles and mechanisms

What is PBT?

Various names for the same, or similar, training concepts to PBT can be found in the literature. These include reactive balance training, perturbation training, reactive step training and fall-resisting skills training, and there is not yet clear consensus on the best terminology. Here, we define PBT as *balance training that uses repeated, externally applied mechanical perturbations to trigger rapid reactions to regain postural stability in a safe and controlled environment*. The goal of PBT is to specifically target and improve the ability to recover stability in destabilizing situations like those that lead to falls in daily life. To meet this definition of PBT, the training should meet two key criteria (Figure 1): (1) the training should use external perturbations that induce a sudden motor response and, (2) these perturbations should be of sufficient magnitude to induce a loss of stability that would lead to a fall without a sufficient motor response (or use of the safety harness). Biomechanically, a loss of stability occurs when the position and motion characteristics of the center of mass exceed certain spatial and temporal limits relative to the base of support, whereby a fall becomes imminent without further action (44, 45). For this article and from a functional standpoint, we use the term balance as an umbrella term for all mechanisms and skills contributing to the maintenance of stability, with the term stability referring to the outcome or state (e.g., mechanically stable/unstable, fall/no fall).

What is task-specificity in the context of PBT?

Our criteria for defining training as PBT, described in What is PBT?, specify that the training should use external perturbations that induce a sudden response and that are of sufficient magnitude to induce a loss of stability. In other words, if the perturbations used do not, (a) require a sudden response to compensate for the disturbance or, (b) lead to a loss of stability, we contend they are not sufficiently similar to the common causes of falls in daily life and are therefore, not task specific. For example, "internal perturbations" or instability induced by narrowing one's base of support or standing on an unstable wobbly surface are not considered PBT. A second consideration is that the method of perturbation delivery should be similar to common perturbations experienced in daily life. In this regard, pop-up obstacles on a walkway [like those used by Pavol et al. (46), Pavol et al. (47), Pijnappels et al. (48), Pijnappels et al. (49), Pijnappels et al. (50), Okubo et al. (51), Okubo et al. (52)] more closely simulate real life trips than a treadmill belt acceleration or deceleration [like those used by Sessoms et al. (53), Owings et al. (54), Grabiner et al. (55), McCrum et al. (56), for example], with cable-trip systems [e.g., as in Senden et al. (57), McCrum et al. (58) or Epro et al. (59)] lying in-between. While some studies suggest that the kinematics of the recovery actions triggered by treadmill-delivered perturbations are similar to more ecologically valid perturbations (53, 54), another study that directly compared treadmill belt accelerations with obstacle-induced trips while walking reported significant differences in trunk and stepping

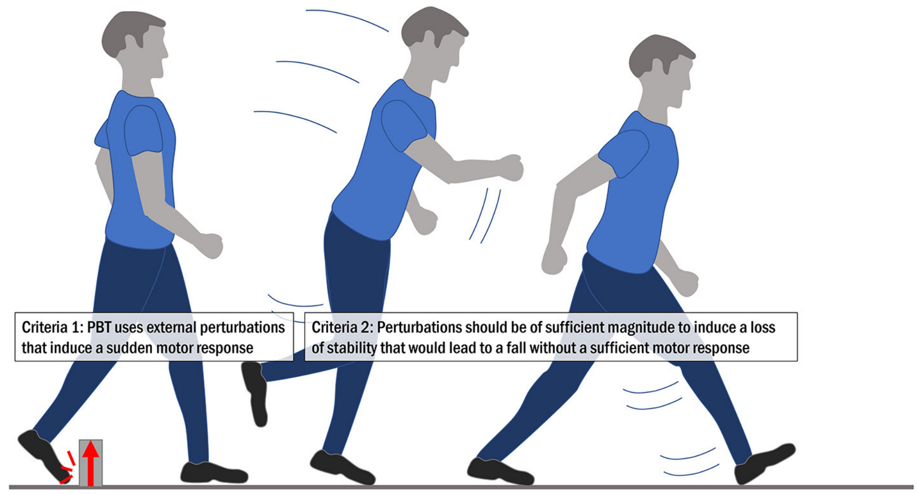

FIGURE 1
The two key criteria for perturbation-based balance training.

kinematics and their adaptations (60). As discussed in sections How does PBT lead to the retention and generalization of fall-resisting skills? and What technology is required for PBT?, the degree of similarity between the training and real-life trips and slips may have implications for the generalizability of PBT training approaches.

A third aspect of task specificity relates to whether perturbations are applied during standing, walking or other common movements (i.e., sit-to-stand transitions). As most falls in community-dwelling older adults occur during walking (1–6, 8, 61), this may be the most relevant task for PBT training for this group. However, frail older people, such as those living in residential care facilities, often experience falls during transitions (62–65), thus may benefit from standing and sit-to-stand perturbation training. Finally, due to the task-specific nature of PBT, training benefits may be restricted to improvements in dynamic and perturbed balance tasks with little or no transfer to less dynamic / static balance tasks (66–68). Some examples of various task-specific elements to consider are shown in Figure 2.

How does PBT differ from other task-specific approaches to fall prevention?

Task-specific walking or balance training, even in the narrowed context of fall prevention, may take many forms. In addition to PBT these include: volitional step training with responses to various stepping targets, cues and constraints [for a review see: Okubo et al. (32)]; gait adaptability training using virtual or real obstacles [for example tasks see Geerse et al. (69) or Timmermans et al. (70)]; adapted forms of agility training [e.g., Donath et al. (71) and Lichtenstein et al. (72)]; and training with ongoing disturbances, simulating situations like uneven ground (73–75). Our criteria for PBT described above, however, distinguish PBT from these complementary approaches, in that regardless of how the perturbation is delivered (trip, slip, push, pull, to the trunk, to the foot, etc.), the participant must quickly identify and respond to a 'sudden' perturbation. In contrast, in the other approaches, changes in the environment can be perceived prior to contact or the response is to a cue separate from a loss of stability. During PBT, the "cue" *is* destabilization, detected by the sensory systems, which triggers rapid stability-recovery responses. PBT facilitates sensorimotor adaptations in these stability-recovery responses through trial-and-error practice. Coupled with the criteria that PBT triggers a sudden response is the requirement that the perturbation causes a loss of stability. This element is conceptually similar to the definition of "challenging balance training" in previous reviews [e.g., balance training including two or three of the following criteria: movement of the center of mass; narrowing of the base of support; minimizing upper limb support (76)]. However, during PBT, usually conducted with a safety harness, participants' stability can and should be further challenged so that destabilization *always* occurs. This is also distinct from volitional step training and gait adaptability training, where appropriate stepping behavior to *avoid* stability loss is trained. Practically, in PBT, this sudden response following a loss of stability during standing or walking usually manifests as reactive stepping or reaching (when an appropriate support is available), as described by Maki and McIlroy (77) as a change-in-support strategy. If participants

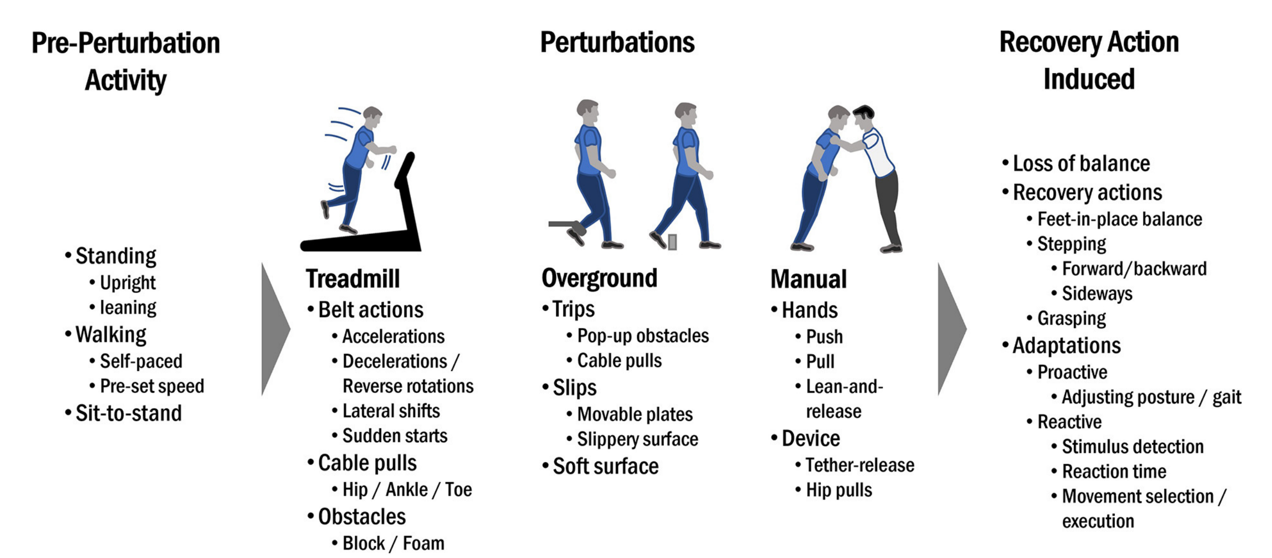

FIGURE 2
Examples of various task-specific elements to consider in PBT.

cannot retain their stability following a perturbation, they are caught by a safety harness or therapist. Such events form part of the sensorimotor skill learning, although whether complete recovery failure is necessary for successful intervention is currently unclear. There are also practical considerations that may affect how perturbation intensities leading to these failures are administered. These issues are addressed in sections What is the dose-response relationship for PBT?, How can anxiety be alleviated during PBT? and What is the evidence for PBT in clinical populations. Finally, while muscle strength training can't be considered a task-specific fall prevention intervention, strength training targeting functionally relevant muscle groups and actions could potentially be used in conjunction with PBT. One RCT (35) reported that PBT combined with hip muscle strengthening may further improve stepping performance and reduce daily life falls compared to PBT alone or strength training alone. This may suggest possible synergistic benefits of using PBT and targeted strengthening approaches. However, another study found no synergistic effects of PBT and training of plantar flexor muscles stability following trip perturbations (59), thus further investigation into such combined approaches is required.

What are the mechanisms for PBT improving fall-resisting skills?

Early research demonstrated that a single session of repeated-perturbations (such as slip- or trip-like perturbations) results in acquisition of fall-resisting skills through implicit learning (without instruction) (78–81) across age-groups (young and old) (78) and tasks (standing, sit-to-stand transitions and walking) (82). In such single training sessions, the reduction of "in-task" falls can occur rapidly, i.e., in three–five trials (83). These improvements in recovery are associated with rapid improvements in both the feedforward/proactive control of stability (anterior shift of the center of mas) (83, 84) and the provision of proper limb support against collapse (78, 85–87), reflected in the form of improved recovery stepping responses, both during stance and walking perturbations. Depending on the perturbation type, the control of stability and limb support is achieved *via* changes in kinematic parameters such as recovery step length, trunk angle and velocity resulting from changes in neuromuscular output (88–90).

Motor adaptations, like those induced by PBT, may be predictive or reactive in nature (91–94). Predictive adaptation to a perturbation utilizes prior experience and knowledge of the upcoming perturbation in a feedforward manner to proactively adjust locomotor control and output (e.g., modifications of the base of support and/or center of mass position). Predictive adaptation can reduce the impact of a perturbation, reducing the magnitude of the required balance recovery response (95). Reactive adaptation, conversely, is a change in the motor responses to an unexpected perturbation. Reactive adaptation can manifest as: earlier detection of the perturbation or stability loss and faster stability recovery initiation (96, 97); optimization of motor programmes for stability recovery including facilitation and suppression of functionally relevant and irrelevant reflexes and reactions, respectively (98–100); and altered coordination in skeletal (especially weight bearing) muscles for rapid motor actions (18, 88, 94, 101–103). As discussed in section How does PBT differ from other task-specific approaches to fall prevention?, only PBT aims to improve the reactive stability recovery responses to destabilizing perturbations, as opposed to other task-specific approaches (e.g., gait adaptability training) that target predictive adaptations only. However, predictive adaptation, to some extent, is likely inherent in most PBT programmes (92). As such, PBT programmes should consider, and possibly monitor and account for, the role and influence of predictive adaptation, since it might reduce the impact of the administered perturbations and reduce generalization effects (31, 104, 105). Indeed, it has been demonstrated that perturbation impact might be significantly reduced if participants are aware they might encounter an unspecified hazard that may perturb their balance (105–107). Okubo et al. (108) also found that predictive adaptations are less readily observed when perturbation type, location and timing are unpredictable. However, other studies have shown that awareness of upcoming perturbations (109) or even observation training (watching videos of the perturbation task) (110) can lead to predictive adaptations but their effects were not comparable with those from actual physical experience of the perturbations.

How does PBT lead to the retention and generalization of fall-resisting skills?

Promising results from early studies using overground slip perturbation training revealed that the skills acquired during a *single* repeated-slip training session can be retained for up to a year by developing protocols incorporating random practice [contextual interference (111–113)] and overlearning [continued task practice after reaching a success criterion (114–116)] *via* high intensity training (24 repeated slips) among healthy (young and older) adults (86, 117). Several other studies have subsequently shown good retention after exposure to repeated perturbations in healthy young adults (84, 86, 118, 119), older adults (59, 119), people with stroke (120–123) and Parkinson's disease (124–126). In terms of training dose, studies have included single sessions (117–119, 124, 127) and multiple sessions (59, 122, 123, 128) and retention intervals from as short as 30 min to up to 1.5 years post intervention.

A vital function of the central nervous system is its ability to apply motor adaptations obtained in one situation to a different situation. The central nervous system can generalize response adaptations to similar perturbations to an untrained limb (56, 129–131); untrained tasks [e.g., gait-slip to sit-to-stand slip (82)]; untrained contexts [e.g., moveable platform to vinyl floor (132–135)]; and to different perturbation types [slips to trips (136) and waist pull perturbations to treadmill slips (137), though minor interference has also been reported (81, 136)]. Generalization between contexts (treadmill to overground slips) may also be retained over longer periods (138). Based on evidence from locomotor training studies it is postulated that when an acquired internal representation is more general (i.e., not specific to certain effectors, environments or tasks) more motor transfer will ensue (139–143). This postulation seems applicable for PBT as well for fall prevention.

In summary, most reports indicate a positive transfer of adaptations between different conditions of the same perturbation, i.e., from treadmill gait-slips to a 'novel' overground slip, or from training gait-slips on a moveable platform to an untrained slip on an oily surface (97, 133, 135, 144–146). However, several recent investigations have shown that improved balance skills resulting from repeated exposure to trip-like perturbations does not transfer to the recovery response to a similar large mechanical perturbation in the anterior direction (60, 119, 147). Critical components in neuromotor control (e.g., module composition and time-coordinated recruitment of motor modules) due to different neuromechanical task constraints (e.g., muscle activity patterns and body dynamics) may discriminate between perturbation types, possibly explaining the discrepancy between findings for generalization of adaptations from repeated gait perturbation exposure. Thus, although generalization is possible within the human stability control system, it may require a certain degree of similarity, if not consistency, between tasks which may be determined by factors other than shared limb mechanics. A recent study investigated potential factors limiting inter-task generalization within the stability control system (147). Differences were detected in the synergistic spatiotemporal organization of muscle activations indicating a diverging modular response to different perturbations, seemingly covered by the same main balance skill (i.e., rapid stepping). Hence, it may be argued that the transfer of adaptations in stability control between different balance tasks may be influenced by differences in muscle synergies in the perturbation recovery responses. Thus, while generalization of adaptation is in principle possible within the human stability control system, it seems limited if neuromotor factors discriminate perturbation responses in different motor tasks e.g., discrepancies in the spatiotemporal organization of the motor system between balance tasks (147).

What is the dose-response relationship for PBT?

For a training protocol to be clinically accepted and implemented, the training dose-response relationship in addition to the training effect needs to be established (21, 148). A training dose can be varied by altering the intensity of the perturbation (making it more challenging), the amount of practice per session (increasing the number of perturbations) or the number of training sessions provided (148, 149).

For overground slip perturbations, earlier studies showed that a high practice dose (in terms of intensity) provided in a single session led to significant retention over the longer-term (4-6 months) (117, 128). Increasing the session frequency in terms of providing a booster dose did not lead to greater retention in younger adults (86, 150) but did so in older adults (128). However, increasing intensity, frequency and duration of such protocols could also have disadvantages such as activity-induced fatigue and reduced participation, particularly in certain clinical populations with significant health issues and balance impairments (151, 152). Another alternative for those unable to tolerate a high dose within a single session is to provide more sessions with fewer training trials or min per session (152). For example, studies have shown that a single slip exposure administered in separate, frequent sessions can induce lasting effects within the same environment (i.e., laboratory) (80, 150). No studies have examined dose effects for overground trip perturbations (148).

Several studies involving young adults, older adults and people with stroke have used different practice durations per session and number of sessions in their studies using treadmill belt perturbations. The number of trials per session have ranged from 11 to 80 and number of sessions have ranged from 1 to 24 (144, 145, 149, 153, 154). Retention periods have ranged from 30 min (144, 153, 155) to 6 months (138, 154).

There is a clear need for further dose-response studies (in particular, for the more clinically applicable treadmill-based protocols) to examine retention and generalization of the adaptations made, as the optimal dose for within-session or within-training programme adaptation may not necessarily be the same as the optimal dose for long term retention and generalization. Further, most studies have used only a single type of perturbation direction which may result in the limited real-life generalization observed. More studies are needed to examine the effect of bidirectional or multidirectional perturbation training on longer-term retention and generalization. Lastly, the type of perturbation training that yields maximum efficacy also remains unknown. These gaps are important to fill to provide recommendations to clinicians and develop clinical practice guidelines.

Implementation in practice

What are the primary safety issues in PBT?

PBT requires additional safety measures compared to conventional balance training. In this regard, safety harnesses are often used when administering large external perturbations. The benefit of a safety harness is that the participant can move in an almost unrestricted manner, and the therapist can focus on training delivery, with the assurance that any unsuccessful balance recovery will be safely arrested by the harness. Many different options are available, ranging from a fixed harness which can be attached to the ceiling in the middle of the exercise room or above a treadmill, or ceiling rail system harnesses which enable the wearer to move freely through a room. A portable/movable support frame is another option if appropriately certified for supporting a participant's body weight and does not interfere with reactive stepping responses. Harnesses also need to be well-fitted and comfortable to prevent harness-induced bruising and soreness after training.

Few adverse events from PBT training have been reported in the literature. Most studies report no or relatively minor adverse events such as soreness at the contact points between the body and the harness or muscle soreness (156–160). In 12 RCTs (20, 36, 52, 97, 122, 159, 161–165) summarized by Mansfield et al. (166), pain and delayed onset muscle soreness were the most commonly reported adverse events (16.4% of participants), with no severe adverse events reported in these trials. One other study reported 6 mild to moderate adverse events related to lateral waist-pull perturbations, including knee pain and groin injury, although the authors stated that this perturbation approach was generally well tolerated by the participants (35). Muscle soreness during or after training cannot be entirely prevented but may be decreased by adjusting training intensity for each individual. If an individual experiences a fall into the safety harness, follow-up assistance is often required to help them regain their stability and composure.

When working with less intensive external perturbations, such as therapist-applied perturbations, training is possible without additional safety equipment. However, it is crucial that both the therapist and patient know their limits and having a second therapist present to provide stability support is advised. Transfer belts can also assist the therapist apply perturbations as well as support their patients as required.

How can anxiety be alleviated during PBT?

Anxiety and fear about upcoming perturbations and/or falling is a practical challenge in PBT (167). In their overview of 12 RCTs of PBT, Mansfield et al. (166) noted that about 5% of the included participants reported PBT-related fear or anxiety (some of which withdrew for this reason) and a more recent meta-analysis confirms that anxiety and fear occur more frequently in PBT than in control interventions (33). Anxiety during training is higher in older adults compared to younger adults and increases with greater uncertainty about the upcoming perturbations (51). In one study, older adults reported higher anxiety during PBT on a treadmill compared to PBT on an overground walkway (60). The authors suggest that this higher anxiety may have been due to unfamiliarity with treadmill walking and the elevated surface of the treadmill. Anxiety is higher in those with poor reactive balance, but heightened anxiety can also impair reactive balance control *via* delayed, more rigid and/or (poorly adapted) startle responses

(168–170), and thus should be minimized for a better training outcome. Monitoring of anxiety levels using a custom 5-point scale and adjustments of training intensity (e.g., 5–10% reduction in gait speed) have been effective in easing anxiety during reactive balance training using overground trips and slips (52). Interviews with participants who underwent PBT using an instrumented treadmill system (171) revealed that while some participants experienced anxiety during training, most described feeling a "good kind of nervousness" during training, rather than anxiety. Participants that reported being initially anxious often found that their anxiety diminished or resolved after the first training session when they had experienced PBT and were confident they could recover from the perturbations, a finding also reported by Jagroop et al. (167). The presence of safety equipment (especially a safety harness), and ensuring participants are heard and informed during the training sessions have been identified as important factors that mitigate anxiety (171). In cases where sufficiently large destabilizing perturbations increase anxiety and possibly prompt withdrawal, it may be prudent to administer training intensities that are less threatening until anxiety is reduced. This may reduce the effectiveness of the initial training period and may not qualify as PBT as per our definition but may retain patients in training and allow them to become more comfortable with the training regime and take part in higher intensity PBT in subsequent trials. Uncertainty about the timing, location, type or direction of perturbations (in situations in which these are modifiable options) can also be gradually increased congruent with the comfort and performance levels of participants.

What is the evidence for PBT in clinical populations?

To date, PBT has been studied primarily in healthy community-dwelling older adults. However, there is also emerging evidence for the effectiveness of PBT in 'high risk' older adults (for example assisted living residents, or older adults with a history of falls or balance problems), and people with Parkinson's disease, stroke and multiple sclerosis (121–123, 156, 158, 159, 172–176). PBT trials have also been conducted in people with chronic obstructive pulmonary disorder (152) and incomplete spinal cord injury (165), but due to limited findings will not be discussed in detail in this article. Previous reviews (29, 30) showed significant fall reductions in community-dwelling older adults, frail/high-risk older adults and people with Parkinson's disease and stroke following PBT. PBT has also been found to improve perturbation recovery measures (156, 159, 160, 177) and some studies have reported improvements in clinical balance tests such as the Berg Balance Scale in people with Parkinson's disease (173, 174, 178). However, while there appears to be interest in the potential for PBT to improve a broad range of gait and balance measures in clinical populations [see reviews of Hulzinga et al. (179), Coelho et al. (180)], as outlined in section What is task-specificity in the context of PBT? and How does PBT differ from other task-specific approaches to fall prevention?, the effects do not necessarily generalize to less-reactive balance and gait measures. To our knowledge, no current studies in clinical populations have reported non-responders in terms of adaptation of the stability recovery response to PBT. However, on an individual level, those who cannot tolerate being exposed to perturbations (due to, for example, anxiety or pain) may not be able to benefit from PBT immediately, and perhaps initially require more basic balance training.

There are some important factors to consider before applying PBT in less able populations. First, decreasing training intensity to an acceptable level for the participant may mean that the total training volume is increased to compensate. Second, frailer people may require a walking aid in daily life. To our knowledge, no studies have focused on the feasibility of using walking aids during PBT, but we hypothesize that the use of a full-body harness with partial bodyweight support may enable PBT for these people. Future studies may focus on this gap in knowledge.

What technology is required for PBT?

Several mechanical perturbation systems can evoke the balance disturbances required for PBT and trigger error-driven motor learning in the control of postural balance. As there is a growing body of evidence suggesting both the efficacy and efficiency of PBT for improving fall resisting skills, there is also a need to further develop devices which are capable of mimicking disturbances experienced during daily-life mobility in clinical settings.

An ideal system for training reactive balance recovery should be capable of applying unpredictable mechanical perturbations of different magnitudes and directions and/or types at pre-specified timepoints that elicit a loss of balance and thus mimic near-fall situations in a safe, controllable environment (31, 181, 182). This system should also be able to measure the participant's stability and stability recovery to facilitate assessment and personalized training.

Several perturbation systems have been used to disturb stability during walking, including floor obstacles in both overground (46, 51, 113, 183–186) and treadmill setups (181, 187), unexpected surface compliance changes [overground; (188)], overground slips or surface translations (133, 189, 190), cable or rope trips both in overground (191, 192) and treadmill setups (57, 88, 193–196), as well treadmill-based belt speed changes (53, 55, 61, 118, 197–199), platform translations or tilts (200) and waist/torso pushes and pulls (137, 201–204). Several commercially available systems are also available (e.g., BalanceTutor, ActiveStep, C-Mill React). It is important to highlight that no system is without its limitations. For example, overground setups suffer from the limitation of limited walkway length and that the location of the perturbations may not be entirely unpredictable (31), though this limitation can be, at least partly, addressed by including multiple possible perturbation locations [see, for example, (108)]. The obvious advantage of the treadmill in comparison to such overground setups is that predicting when a perturbation will be applied is more difficult, as there is no location-based reference point (31), which ensures that predictive adjustments in in anticipation of perturbations are reduced [though not necessarily completely absent (123, 155)]. However, walking on a treadmill can provide additional challenges in some populations at increased fall risk, due to lack of familiarity and the requirement to maintain a specific speed [walking speed can be instantaneously adjusted in an overground setup but maintaining it provides an additional challenge during perturbed treadmill walking (205)]. Another inherent limitation in some setups is that the perturbations themselves may not strictly mimic common causes of falls like slips and trips (60) despite the subsequent recovery mechanics being suggested to be similar (53, 54) (see also Figure 2 in section What is task-specificity in the context of PBT?). A recent study reported that adaptations observed with repeated treadmill belt accelerations did not transfer to obstacle-induced trips while walking (60). However, it is not currently known if and how this affects transfer to daily life situations. Another factor that should be considered is the ease with which PBT dose can be altered. The number of perturbations and training sessions can be easily manipulated but not all systems can provide a wide range of perturbation magnitudes which is critical to ensure that participants are safely and sufficiently destabilized, even late in their training. This is of particular relevance for the conceptualization of fall prevention interventions in clinical settings because the hypothesis of a non-linear dose-response relationship (148) implies that adaptation may not be directly related to the applied practice dose and that a dose threshold exists beyond which any additional stimuli may not induce further changes.

In summary, based on current evidence, we believe that the primary factors for a successful PBT system are that it can; (a) administer perturbations that are difficult for participants to predict (in time of onset but perhaps also in body location, mode or magnitude of application); (b) suddenly destabilize participants with these perturbations; and (c) easily adjust the magnitude of perturbations.

Despite the potential advantage of using such systems to destabilize participants and create near fall situations, the costs associated with the equipment, as well as the expertise required to operate PBT systems may hinder their application in clinical settings. Thus, there is a need to develop alternative, feasible PBT programmes that do not require these devices. Therapist-applied perturbations, as described above, are the natural alternative and can be easily applied if appropriate safety measures are followed. However, managing the training and perturbation dosage may be problematic due to the perturbations being more predictable and the intensity of therapist-applied perturbations being less precise. Such limitations, however, do not discount the potential effectiveness of this approach when they constitute the only feasible option in at least the short term. For a useful resource on the therapist-applied perturbation approach, we refer readers to Mansfield et al. (166).

Is PBT appropriate in at-home, group or semi-supervised settings?

The application of PBT in home or group settings has been little investigated to date. Clearly, it is not safe to apply large external perturbations, with the possibility of an unsuccessful balance recovery in the absence of a safety harness. Smaller perturbations however, such as therapist-applied perturbations, may be applied in home and group settings. For example, Oddsson et al. (17) successfully applied perturbations in a group setting through training in couples with partner or therapist-applied perturbations.

As discussed above, it is crucial the participant feels safe during training, and everyone involved know their limits. Portable safety equipment, such as a transfer belts, can assist the therapist apply the perturbations as well as support patients during training. However, if an appropriate training stimulus cannot be reached this way, transferring the training to a one-on-one basis, or using more specific equipment should be considered. Future studies are necessary to elucidate the feasibility of PBT in a group or semi-supervised setting.

Recommendations

Taking the previous sections into account, several recommendations for both research and clinical application of perturbation-based balance training can be made.

Research

Studies are required to:

- Determine optimal training doses and the potential effects of repeated training or booster sessions.
- Identify the relative contribution of different aspects of training dose (e.g., perturbation impact, perturbation training intensity (displacement, velocity, acceleration settings), perturbation number, training session number) to the training effects.

- Compare the effects of different laboratory-based PBT methods with respect to stability outcomes and daily life fall prevention.
- Further elucidate and compare the criteria by which adaptations gained by training one type of perturbation transfer to other similar perturbations (e.g., between legs, a movable plate to a slippery floor; see How does PBT lead to the retention and generalization of fall-resisting skills? above).

Clinical application

There is a need to:

- Develop effective, affordable and clinically feasible methods for applying perturbations.
- Conduct feasibility studies to explore opportunities and barriers for implementation.
- Determine strategies to alleviate anxiety in participants undertaking PBT to ensure clinical feasibility.
- Identify which clinical populations with balance impairment benefit from PBT
- Elucidate PBT dose-response relationships in these populations.

Finally, it is worth highlighting that there have been only a few randomized controlled trials with sample sizes large enough to have statistical power to evaluate the role of PBT in reducing daily life falls. Lurie et al. (34) with their multicenter pragmatic (non-standardized protocol based on therapist judgement) trial is the largest. This 12-month trial included 187 participants (of 253 allocated) who received PBT and 190 (of 253 allocated) participants who received standard balance training. Once some of the issues relating to training and practice mentioned above have been further elucidated, we recommend large, definitive trials following CONSORT guidelines are conducted. In the meantime, we recommend that studies on PBT collect and report prospective falls data as secondary outcomes to assist future meta-analyses. Template forms for collecting falls information following recommendations by Lamb et al. (206) and Lord et al. (207) can be downloaded at http://doi.org/10.17605/OSF.IO/HMJEF (208).

Conclusions

Perturbation-based balance training is a promising approach to fall prevention. This task-specific training of balance using repeated exposure to sudden perturbations may present a paradigm shifting approach that may improve effectiveness and efficiency of a fall prevention exercise intervention. However, several fundamental and applied aspects of the approach need to be further investigated before this approach can be widely implemented in clinical practice.

Author contributions

CM and YO: conceptualization, project administration, writing-original draft, and writing-review and editing. TB, MG, and KK: writing-original draft and writing-review and editing. MR: writing-review and editing. SL: conceptualization and writing-review and editing. All authors contributed to the article and approved the submitted version.

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A preprint version of this work is available at OSF Preprints (209).

Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Measures of falls efficacy, balance confidence, or balance recovery confidence for perturbation-based balance training

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KEYWORDS

falls efficacy, balance confidence, balance recovery confidence, self-efficacy, perturbation-based training, falls prevention, falls management, fear of falling

Introduction

There is a growing interest in using perturbation-based balance training (PBT) to reduce falls (1). PBT is a skill training intervention that aims to improve reactive balance control in response to destabilizing perturbations in a safe and controlled environment (2). Studies have often posited that the training mechanisms of PBT improve physical abilities, such as generating more effective recovery step response and trunk movement to arrest falls in the face of a slip, trip or a loss of balance caused by volitional movement (3). This explanation has also been offered for studies employing a single PBT session (4, 5). PBT is likely to influence psychological factors. However, the impact on this aspect remains unclear. Psychological factors are well-established predictors of falls and play a role in determining performance, such as balance and gait (6). Yet, several studies have reported a limited influence of PBT on falls efficacy or balance confidence (7, 8). PBT could affect other self-efficacies, such as balance recovery confidence, safe landing confidence, or fall recovery confidence, but there are scarce studies on them. Since falls are a complex phenomenon, the concepts of the different falls-related self-efficacy (falls efficacy) constructs must be clarified. Having better clarity allows appropriate measures to be selected to elucidate the impact of PBT on the perceived ability to deal with falls.

Deciphering falls efficacy has not been easy because several falls-related psychological factors have been used interchangeably in the literature. Falls efficacy is closely related to fear of falling or balance confidence, but it is necessary to recognize that these constructs are distinct (9, 10). While some research papers have presented falls efficacy and balance confidence as isomorphic (11), this paper will consider balance confidence to be a subdomain of falls efficacy. A recent methodological quality review of the content development of falls efficacy-related measurement instruments reported that falls efficacy has been viewed as a self-efficacy construct that covers different perceived abilities needed to prevent and manage falls (12). Rooted in Bandura's self-efficacy theory (13, 14), falls efficacy refers to the general belief in capabilities required to overcome various falls-related situations. This belief incorporates different self-efficacies presented across four stages surrounding falls (Figure 1) (15). In the pre-fall stage, balance confidence refers to the perceived ability to perform activities without losing balance. In the near-fall stage, balance recovery confidence focuses on the perceived ability to arrest a fall in response to destabilizing perturbations. These two stages surround the perceived capability to prevent falls (16). In the fall-landing stage, safe landing confidence relates to the perceived ability to fall safely on the ground when the balance is irrecoverable. In the post-fall stage, fall recovery confidence refers to the perceived ability to get up from the floor independently. The latter two stages surround the perceived capability to manage falls (16).

In contrast, fear of falling refers to the concerns about falling and that the individual would avoid the activity despite being able to perform (9). Fear of falling is likely to incorporate efficacy and outcome expectancies (17). Outcome expectancy is a judgement about performance outcomes, whereas efficacy expectancy is a judgement of the capability to perform in a given situation. Fear of falling measures, such as the Falls Efficacy Scale-International (18), Fear of Falling Questionnaire (19), and Fear of Falling Avoidance Behavior Questionnaire (20), do not solely assess falls efficacy expectations. Applying appropriate measurement instruments is imperative to understand PBT's role and helps reduce the risk of misinterpreting the results (21). The commentary aims to highlight some falls efficacy measures for PBT research so that researchers can make an informed decision when selecting the most suitable measures to determine perceived capabilities to deal with falls.

Measures of falls efficacy for PBT research

Bandura's self-efficacy theory (13) states that the efficacy belief system is a differentiated set of self-efficacy beliefs linked to distinct realms of functioning. Researchers need to be clear about the intended self-efficacy beliefs or the confidence of the ability to accomplish a task or succeed in a particular situation (22). When PBT research plans for certain types of perturbations to be delivered, some balance control mechanisms and self-efficacies are predominantly targeted (23). The following examples are presented:

Example 1: Destabilizing perturbations to be delivered at an insufficient intensity to cause a fall, yet having the equilibrium perturbed adequately would likely train balance control *in situ* (23). The targeted mechanisms could be proactive, anticipatory, or reactive fixed support systems for the person to perform the task or activity more steadily, as shown in Figure 1: Pre-fall stage. Some real-world situations are standing (not holding a handrail) on a moving train or walking on a wet sidewalk. These PBT may benefit from using measures of balance confidence. The most commonly used measures are the "Falls Efficacy Scale (FES)" (24) and the "Activities-specific Balance Confidence (ABC) Scale" (25). Both scales aim to measure the confidence level to perform activities of daily living steadily. Both scales have excellent psychometric properties, such as the FES has good test-retest reliability (0.71) (24), internal consistency (0.90) (25), and scalability (0.44) (25) and the ABC scale has excellent test-retest reliability (0.92) (25), internal consistency (0.96) (25), and scalability (0.59) (25). The 10-item FES is suitable for low-functioning older adults, whereas the 16-item ABC scale is designed for higher-functioning seniors (25). Both scales are not difficult to administer, and each takes about five to ten minutes to complete.

Example 2: Large mechanical destabilizing perturbations to be delivered in such a way that insufficient or inadequate recovery reactions (i.e., reach-to-grasp or compensatory stepping) would result in a fall (1). This training aims to improve reactive change-in-support balance control, as shown in Figure 1: Near-fall stage. Real-world applications refer to individuals arresting falls in situations such as experiencing a slip when walking on a puddle of water or a trip when a foot gets caught by a curb. Such training may benefit using the measures of balance recovery confidence. One candidate measure is the Balance Recovery Confidence (BRC) Scale (26). The BRC scale measures the perceived reactive balance recovery ability in response to perturbations such as a slip, a trip or a loss of balance from volitional movement (26). The BRC scale has good psychometric properties, such as test-retest reliability (0.94) (26) and internal consistency (0.97) (26). The 19-item BRC scale has a list of pictures accompanying each item's descriptor to provide a consistent interpretation of the scenarios (26). The scale is designed for community-dwelling older adults and takes about seven to ten minutes to complete.

Example 3: PBT supplemented with other interventions, such as cognitive behavioral therapy and strength and balance exercise training, could consider multi-domain measures of falls efficacy. Multi-domain measures reveal a general sense of personal efficacy to produce certain attainment (14) and, in this context, overcome falls. This approach transcends the separate subdomains, as noted in Figure 1, where a more meta-efficacy measure could demonstrate an overall change in falls efficacy. One candidate measure is the Perceived Ability to Prevent and Manage Falls Risks (PAPMFR) scale (27). The six-item PAPMFR scale aims to measure confidence in the ability to prevent and manage falls. Items included: "Steadiness on their feet", "Balance while walking", "Ability to walk in their homes", "Ability to walk outdoors", "Ability to prevent falls", and "Ability to find a way to get up if they fall". The PAPMFR scale was conceptually designed to measure the perceived ability to deal with falls. The scale has good psychometric properties, such as excellent internal consistency (0.94), good structural validity and construct validity (27). The PAPMFR scale is developed for community-dwelling older adults and takes about 5–7 min to complete.

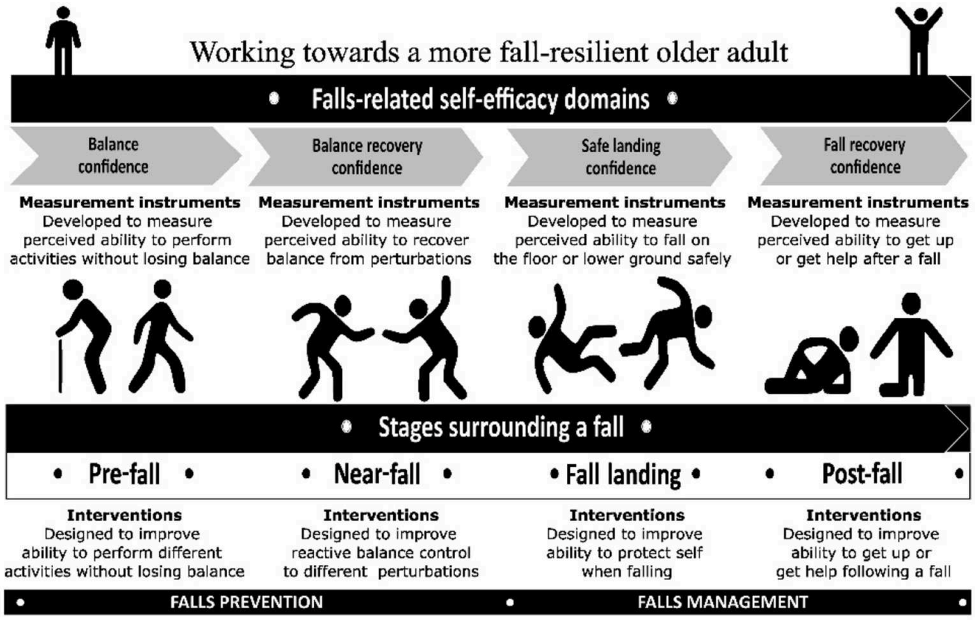

FIGURE 1
The various domains of falls efficacy and the different stages surrounding a fall. The figure has been adapted and reproduced with permission (15).

There are lacking measures for other constructs, such as safe-landing confidence (Figure 1: Fall landing stage) and fall recovery confidence (Figure 1: Post-fall stage). Selecting items from multi-domain measures may be considered but should be done circumspectly. One item is the "Protect yourself if you fall" from the Perceived Ability to Manage Risk of Falls or Actual Falls scale (28) for safe-landing confidence (Figure 1: Fall-landing stage). Another is the "Ability to find a way to get up if they fall" from the PAPMFR scale (27) for fall recovery confidence (Figure 1: Post-fall stage). However, these measures have not been rigorously validated, unlike the FES or the ABC scale. Researchers must be cautious when using these measures or selecting certain items to evaluate specific constructs or falls efficacy. There is an urgent need for validation studies to critically evaluate these measures using the COSMIN methodology (29) to present their psychometric properties (i.e., content development and validity, structural validity, construct validity, reliability, responsiveness, measurement error).

Discussion

Bandura's self-efficacy theory has been an enduring concept for understanding behavior outcomes and would be applicable for PBT in falls prevention and management. The self-efficacy theory explains how efficacy expectations can determine whether coping behaviors will be initiated, how much effort will be expended, and how long the self-efficacy will be sustained in the face of obstacles and adverse experiences (30). PBT research must clarify the self-judged efficacy of interest when designing different perturbation strategies to help older people overcome falls. In other words, which of the constructs, such as the overall confidence to prevent and manage falls (falls efficacy), or the specific constructs, such as the balance confidence, balance recovery confidence, safe landing confidence, and fall recovery confidence, are being targeted? The most suitable measure should then be applied. Potentially, PBT could address the fear of falling by having graded perturbations prescribed with the starting perturbations set at lower strengths of self-judged efficacy. Appropriate identification of the targeted self-efficacy allows PBT to be planned appropriately for individuals to achieve performance mastery and build their self-efficacy (31). Previous studies have shown that falls efficacy plays a mediator between fear and functional abilities (32, 33). PBT could be purposefully designed to alleviate fear by enhancing falls efficacy and achieving improved performance such as balance and gait.

Given that there are varying capabilities to deal with falls, researchers need to discern the objectives of the PBT. Measures of falls efficacy could be employed in various ways. Some researchers may be keen to use PBT to address falls efficacy and thus apply the measures as outcome tools to evaluate the effectiveness of the intervention. Others may wish to use PBT to address the fear of falling and activity-related avoidance behaviors using the self-efficacy theory. Falls efficacy measures can then act as a conduit to inform the design of the PBT's perturbations. For example, the BRC scale contains 19 different "potential near-fall" scenarios depicting a range of perturbations-types (e.g., a slip or a trip), direction-specific (e.g., forward or backward), environmental constraints (e.g., indoor or outdoor), and set-ups for balance recovery strategies (e.g., availability of handrail or uneven ground level). The BRC scale can help researchers plan suitable perturbations by identifying challenging scenarios reported by certain groups of individuals.

Falls efficacy measures should be used alongside other assessments in PBT research to understand perceived and actual abilities. Unlike observable parameters such as kinematic changes, reactive skill performances or reduction in falls, latent psychological factors require researchers to be explicit about the construct of interest. Selecting the most appropriate measures is imperative to elucidate the psychological impact of PBT to help older people overcome falls (9). Moreover, a greater use of appropriate fall efficacy measures in PBT research allows "patient-centered" data captured to demonstrate measurable and meaningful improvements (34). Presenting the perceived capabilities of the individual in real-world falls-related scenarios will provide empirical evidence that the effects of PBT are translatable from a simulated environment to real-life generalization.

Author contributions

The author confirms being the sole contributor of this work and has approved it for publication.

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Conflict of interest

The author declares that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Improvements in spatiotemporal outcomes, but not in recruitment of automatic postural responses, are correlated with improved step quality following perturbation-based balance training in chronic stroke

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Introduction: People with stroke often exhibit balance impairments, even in the chronic phase. Perturbation-based balance training (PBT) is a therapy that has yielded promising results in healthy elderly and several patient populations. Here, we present a threefold approach showing changes in people with chronic stroke after PBT on the level of recruitment of automatic postural responses (APR), step parameters and step quality. In addition, we provide insight into possible correlations across these outcomes and their changes after PBT.

Methods: We performed a complementary analysis of a recent PBT study. Participants received a 5-week PBT on the Radboud Fall simulator. During pre- and post-intervention assessments participants were exposed to platform translations in forward and backward directions. We performed electromyography of lower leg muscles to identify changes in APR recruitment. In addition, 3D kinematic data of stepping behavior was collected. We determined pre-post changes in muscle onset, magnitude and modulation of recruitment, step characteristics, and step quality. Subsequently, we determined whether improvements in step or muscle characteristics were correlated with improved step quality.

Results: We observed a faster gastrocnemius muscle onset in the stance and stepping leg during backward stepping. During forward stepping we found a trend toward a faster tibialis anterior muscle onset in the stepping leg. We observed no changes in modulation or magnitude of muscle recruitment. Leg angles improved by 2.3° in forward stepping and 2.5° in backward stepping. The improvement in leg angle during forward stepping was accompanied by a −4.1°change in trunk angle, indicating a more upright position. Step length, duration and velocity improved in both directions. Changes in spatiotemporal characteristics were strongly correlated with improvements in leg angle, but no significant correlations were observed of muscle onset or recruitment with leg or trunk angle.

Conclusion: PBT leads to a multi-factorial improvement in onset of APR, spatiotemporal characteristics of stepping, and reactive step quality in people with chronic stroke. However, current changes in APR onset were not correlated with improvement in step quality. Therefore, we suggest that, in addition to spatiotemporal outcomes, other characteristics of muscle recruitment or behavioral substitution may induce step quality improvement after PBT.

KEYWORDS

perturbation-based balance training, balance recovery, stroke, reactive stepping, spatiotemporal outcomes, step quality

Introduction

People with stroke (PwS) often exhibit balance impairments, even in the chronic phase (1). These balance impairments have a detrimental effect on their mobility and daily life independence (2) and contribute to the high fall risk post stroke (3). Specifically, a stroke results in an at least two-fold increased risk of falls (4), which may have severe consequences such as hip fractures on the paretic side (5). While exercise interventions with a balance component may be effective in reducing fall risk after stroke (6), it remains unclear which type of balance training is most effective. For the development of more effective falls prevention programs, it is crucial to understand how stroke-related balance deficits respond to training.

One promising training target is the capacity to recover from balance perturbations. The main rationale for focusing on balance-recovery responses to help prevent an actual fall lies in their utility in a wide variety of situations that may induce a loss of balance (e.g., misstep, trip or collision with another pedestrian). In PwS, this so-called reactive balance capacity is often impaired (7), with stepping responses showing more profound deficits than feet-in-place balance recovery strategies. Stroke-related deficits have been observed in different aspects of reactive stepping. First, PwS exhibit deficient automatic postural responses (APRs). APRs are fast automatic muscle responses evoked by balance perturbations and act as a first line of defense to counteract loss of balance (8). These deficiencies are evident by delayed muscle onsets (9) and reduced amplitudes of APRs (10), as well as poorer muscle coordination patterns during the APR time window (11). Second, a stroke affects the spatiotemporal characteristics of reactive stepping. Specifically, PwS may demonstrate a delayed step onset (12, 13) and a smaller step length (14) compared to healthy individuals. Lastly, balance recovery steps in PwS are less effective in 'catching' the falling center of mass (CoM), as shown by outcome measures that capture the relationship between CoM and the base of support at the instance of foot contact (9, 15).

Recent evidence suggests that training involving repeated exposure to balance perturbations (perturbation-based balance training; PBT) can improve stroke-related deficits in reactive stepping (16, 17) and may reduce fall risk (18). At the level of APRs, faster muscle responses were found after PBT in healthy individuals (19), but the effect of PBT on the size and direction-dependent amplitude modulation of the APRs have not yet been studied. At the level of spatiotemporal step characteristics, PBT has shown to improve step length (20), but its effect on other variables such as step velocity or step time remains unclear. Importantly, studies also show that reactive stepping performance improved following PBT (21, 22). Specifically, after PBT, PwS required fewer steps and exhibited a more favorable body configuration at stepping foot contact, as evidenced by a larger leg angle (stepping foot placed further ahead of the pelvis in the direction of perturbation) (16).

Still, limited evidence is available on the effects of PBT at the level of APRs, spatiotemporal variables and stepping performance in PwS. Moreover, it is unclear whether improvements in one respect would translate to the other. As such, specific insights into the mechanisms underlying improvement of reactive stepping performance through PBT in PwS is limited. To obtain more insight in these mechanisms, we performed additional analyses on the data collected by van Duijnhoven et al. (16), who previously reported the beneficial effects of PBT on stepping leg angles following perturbations induced by a moveable platform. The aim of our study was threefold. Our first aim was to provide a comprehensive characterization of PBT-induced changes in APRs and spatiotemporal characteristics of reactive stepping in PwS. At the level of APRs we were interested in onset latencies, response amplitudes and direction-dependent modulation of activity of the prime movers (in the forward and backward direction). Spatiotemporal variables involved step onset, step duration, step velocity and step length. Our second aim was to determine whether step quality in response to forward and backward perturbations would improve in terms of trunk angle at foot contact, in addition to the previously reported improvements in leg angle. Our third aim was to determine which improvements in APR and spatiotemporal variables would underlie improvements in step quality, as quantified by the body configuration at foot contact (i.e., leg and trunk angles).

TABLE 1 Participant characteristics ($n = 18$).

Sex (m/f)	10/8
Age (years)	59 (8.3)
Months since stroke	54 (40)
Stroke type (Ischemic / hemmorhagic)	12/6
Affected body side (left /right)	11/7
MMSE	27,8 (2)
MI-LE (range 0–100%)	64 (19)
ABC-6 Scale (range 0–100%)	42 (25)
Fall history (number of falls in previous year)	1.4 (1.8)
FMA-LE (range 0–28)	20 (5)
FAC (4/5)	4/14

MMSE, Mini Mental State Examination; MI-LE, Motricity Index lower extremity; ABC-6 Scale, Activities-specific Balance Confidence 6 Scale, FMA-LE, Fugl-Meyer Assessment lower extremity score without coordination domain; FAC, Functional Ambulation Categories. Values are presented in means (SD).

Methods

Participants

As described in van Duijnhoven et al., 20 participants in the chronic phase (> 6 months) of stroke were recruited from Nijmegen and the surrounding area. Detailed patient characteristics can be found in Table 1. Participants had to be able to stand and walk independently (Functional Ambulation Categories >3). They were excluded if (1) they had other neurological or musculoskeletal conditions affecting balance, (2) used drugs affecting balance, (3) had severe cognitive problems (Mini Mental State Examination <24) or persistent unilateral spatial neglect (Star Cancellation Test <44), or (4) had behavioral problems interfering with compliance to the protocol. For the current study we only selected those participants of whom data was collected at pre- and post-training assessments and who stepped with the same leg during both assessments ($N = 18$). The study protocol was approved by the Medical Ethical Board of the region Arnhem-Nijmegen and all participants gave written informed consent in accordance with the Declaration of Helsinki.

Study design and intervention

Participants received a 5-week PBT on the Radboud Falls Simulator, with a total of 10 sessions. The Radboud Falls Simulator is a movable platform that can induce balance perturbations by horizontal translations in multiple directions (see Supplementary Videos). Participants received 45 min of training, twice a week. Training difficulty was increased during each training session in a participant-specific manner, more specifically, through increasing the intensity of the perturbation, the unpredictability of perturbation direction, or by adding secondary (cognitive) tasks (dual-tasking). Detailed information on the study design can be found in van Duijnhoven et al. (16) and the training protocol can be accessed directly *via* this link, https://www.frontiersin.org/files/Articles/410755/fneur-09-00980-HTML/image_m/fneur-09-00980-t002.jpg.

Experimental procedure

During pre- and post-training assessments participants were exposed to unpredictable translational platform perturbations that consisted of an acceleration phase (300 ms), a constant velocity phase (500 ms), and a deceleration phase (300 ms) (23) initiated after a random delay. Participants stood on the platform with their preferred shoes and wore a safety harness. They were instructed to recover their balance with one single step. To standardize the perturbation difficulty across individuals with different balance recovery capacities, a participant-specific perturbation intensity was determined during the pre-training assessment. To this end, the multiple stepping threshold (MST) was determined by gradually increasing the perturbation intensity (acceleration) in each direction. The perturbation direction refers to the direction of stepping, such that a forward perturbation would induce a forward step (i.e., platform moving backwards). The direction-specific MST was defined as the maximum intensity at which a participant was able to recover his/her balance with one step. To allow comparison of stepping characteristics between the pre- and post-training, trials at the same intensity (MST and MST+0.125 m/s^2) were collected during both assessments. Within this respective study the average MST value for forward perturbations was 3.03 m/s^2 (SD = 1.4 m/s^2) and for backward perturbations 2.2 m/s^2 (SD = 0.88 m/s^2).

Data collection

Reactive stepping responses were recorded at 100 Hz using an 8-camera 3D motion capture system (Vicon Motion Systems, Oxford, UK). Reflective markers were placed on anatomical landmarks according to the Plug-in-Gait full body model. In addition, we recorded bilateral electromyography (EMG) from Rectus Femoris (RF), Tibialis Anterior (TA), and Gastrocnemius Medialis (GM). EMG electrodes were placed according to SENIAM guidelines (24) and recorded at 1,000 Hz (ZeroWire by Aurion, Italy).

Data processing

Marker data were low-pass filtered at 5 Hz (2nd order Butterworth). Subsequently marker data and EMG data were processed by a custom-written MATLAB (MATLAB2018a) script. Raw EMG was bandpass filtered between 10 and 450 Hz, rectified and low-pass filtered at 40 Hz. Rectified EMG was averaged across trials with similar platform direction. Subsequent analysis was performed on the averaged rectified EMG.

Data analysis

The onset of muscle activity was determined for the prime movers during the APR for each perturbation direction (GM for forward perturbations and TA and RF for backward perturbations). Onset latencies were determined by means of a semi-automatic algorithm that selected the instant at which the averaged rectified EMG exceeded a threshold of 2 SDs above the mean of pre-perturbation activity (500 ms) (11). Perturbation onset was determined through a synchronization trigger sent from the platform. Visual inspection of the trials was performed to verify the accuracy of the onset.

To assess muscle recruitment and modulation during the APR we calculated the integrated rectified EMG (iEMG) and the Modulation Index during the first 75 ms after the first muscle onset in a given direction. Thus, during forward stepping, iEMG was calculated post GM onset and, vice versa, during backward stepping post TA onset.

As described by the studies of Lang et al. (25) and Kelly et al. (26) the Modulation Index can be used to quantify relative changes in activity when a muscle would serve as an agonist compared to when it would be an antagonist. In addition, it can serve as a measure to express a persons' ability to scale task-specific muscular recruitment. As for the APR, the GM acts as an agonist during forward perturbations and as an antagonist during backward perturbations, whereas the opposite is true for the TA. The Modulation Index was defined as described by EQ1.

$$Modulation\ Index = 100^{*}\frac{EMG^{ag}_{75ms} - EMG^{ant}_{75ms}}{EMG^{ag}_{75ms}}. \quad (1)$$

EMG^{ag}_{75ms} is the mean EMG within the 75 ms window of muscle in the agonistic direction and EMG^{ant}_{75ms} is the mean EMG within the 75 ms window in the antagonistic direction. A higher MI indicates a greater recruitment of the muscle in its expected behavior.

Spatiotemporal stepping behavior was assessed by determining step onset, step length, step duration and step velocity. Step onset was defined by marker data as the moment at which the vertical velocity component of the heel or toe marker surpassed a threshold of 0.2 m/s. To calculate the subsequent spatiotemporal stepping behavior parameters, foot contact was defined as the moment when the vertical velocity component of the heel or toe went below 0.2 m/s after step onset. Step length was then calculated as the distance covered by the toe marker from step onset to foot contact. Step duration was defined as the duration of step onset until foot contact, and step velocity was calculated as step length divided by step duration.

To characterize step quality, body configuration at first stepping foot contact was determined. Body configuration outcomes included the vertical inclination angle of the leg segment (defined as the angle between the vertical and a line connecting the mid-pelvis and the second metatarsal of the stepping leg; see Figure 1), i.e., the leg angle, and the trunk segment (defined as the angle between the vertical and a line connecting the mid-pelvis and mid-shoulder position), i.e., the trunk angle.

Statistical analysis

Since outcome measures did not significantly differ between the two perturbation intensities (MST and MST+0.125 m/s^2), subsequent comparisons were performed on the two intensities collectively. To evaluate between session differences ($\Delta_{\text{pre}-\text{post}}$) we performed two-tailed paired *t*-tests on the averaged outcomes. Additionally, we identified potential determinants of improvements in body configuration (leg and trunk angle) through a similar procedure described by de Kam et al. (9, 15). First, we selected those spatiotemporal and muscular outcome measures that improved significantly after training. Subsequently, from those outcome measures Pearson's correlations coefficient were used to identify which outcome measures were significantly correlated ($p < 0.05$) with changes in body configuration. Variables that were significantly correlated in these univariate analyses into multivariate forward step-wise linear regression models with changes in leg and trunk angle as dependent variables for each platform direction. Statistical

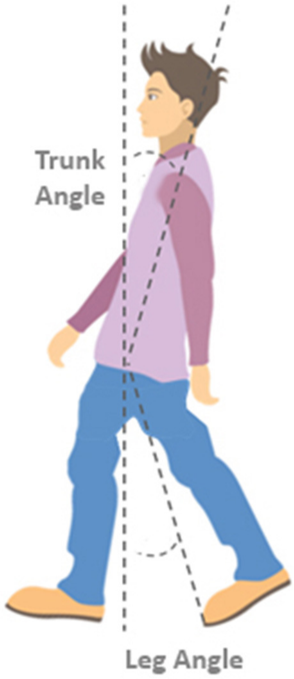

FIGURE 1
Definitions of leg and trunk angles. The trunk angle is defined as the angle between the vertical and a line connecting the mid-shoulder to the mid-pelvis position at the instant of stepping foot contact. The leg angle is defined as the angle between the vertical and a line connecting the mid-pelvis and the 2nd metatarsal of the stepping foot.

analyses were performed with SPSS (version 25.0). P < 0.05 was considered statistically significant.

Results

Amplitude and modulation of automatic postural responses

In the post-training assessment, we observed faster automatic postural responses compared to pre-training, while APR amplitude and activity modulation did not change. Specifically, during forward stepping we observed a small but significant shortening in the gastrocnemius onset latency in the stance and stepping leg (*stepping leg:* $\Delta_{pre-post}$ *5 ms* $\pm$ *7,* $p = 0.01$*; stance leg:* $\Delta_{pre-post}$ *5 ms* ± 7, $p = 0.02$*)*. For backward stepping, we observed a trend toward a faster tibialis anterior onset in the stepping leg ($\Delta_{pre-post}$ *4.4ms* $\pm$ 9.2, $p = 0.06$*)* but not in the stance leg ($p = 0.16$). Onset latencies of the rectus femoris did not change in response to training (*stepping leg* $p = 0.20$, *stance leg* $p = 0.68$).

TABLE 2 Mean (SD) pre and post-training values of step quality and step characteristics.

	Pre intervention	Post intervention	*P*-value
Forward stepping			
Leg angle (°)	21.1 (3.4)	23.4 (2.9)	**0.007**
Trunk angle (°)	22.3 (7.8)	18.1 (7.3)	**0.002**
Step onset (s)	0.29 (0.04)	0.28 (0.04)	0.117
Step length (m)	0.48 (0.2)	0.55 (0.1)	**<0.001**
Step duration (s)	0.3 (0.04)	0.33 (0.04)	**0.002**
Step velocity (m/s)	1.56 (0.4)	1.66 (0.4)	**0.011**
Backward stepping			
Leg angle (°)	−1.6 (6.6)	0.9 (5.6)	**0.008**
Trunk angle (°)	2.1 (7.2)	2.3 (7)	0.55
Step onset (s)	0.32 (0.05)	0.32 (0.04)	0.19
Step length (m)	0.37 (0.2)	0.44 (0.2)	**0.001**
Step duration (s)	0.25 (0.05)	0.29 (0.04)	**<0.001**
Step velocity (m/s)	1.4 (0.5)	1.47 (0.4)	*0.06*

Statistically significant different outcome measures are highlighted in bold $P < 0.05$.

Step quality and spatiotemporal characteristics

As previously reported by van Duijnhoven et al. (16), participants demonstrated improved reactive step quality following the training intervention. The leg angles increased for both the forward and backward perturbations [forward (fwd): $\Delta_{pre-post} = 2.3 \pm 3.0°$, $p < 0.01$; backward (bwd): $\Delta_{pre-post} = 2.5° \pm 3.5°$, $p < 0.01$; Figure 2], indicating that the stepping foot was placed further ahead of the pelvis. Complementary to our previous study, we found and improvement in trunk angle for forward stepping following training. Specifically, the trunk angle decreased, which indicates a more vertically oriented trunk ($\Delta_{pre-post} = -4.1 \pm 1.1°$, $p < 0.01$; Figure 2). On the other hand, no changes were observed in trunk angles after backward perturbations (Table 2).

The observed improvements in step quality were accompanied by changes in some, but not all, spatiotemporal step characteristics. Specifically, greater step lengths were observed in the post vs. pre-training assessments for both forward and backward perturbations (fwd: $\Delta_{pre-post} = 0.06 \pm 0.06$ m, $p < 0.01$; bwd: $\Delta_{pre-post} = 0.06 \pm 0.06$ m, $p < 0.01$);. Similarly, step durations increased for both perturbation directions (fwd: $\Delta_{pre-post} = 0.02 \pm 0.02$ s, $p < 0.01$; bwd: $\Delta_{pre-post} = 0.03 \pm 0.03$ s, $p < 0.01$). An increased step velocity was observed for forward perturbations ($\Delta_{pre-post} = 0.09 \pm 0.14$ m/s, $p = 0.01$), with a similar trend for backward steps ($\Delta_{pre-post} = 0.07 \pm 0.16$ m/s, $p = 0.06$). Step onsets were not different between the pre- and post-training assessments.

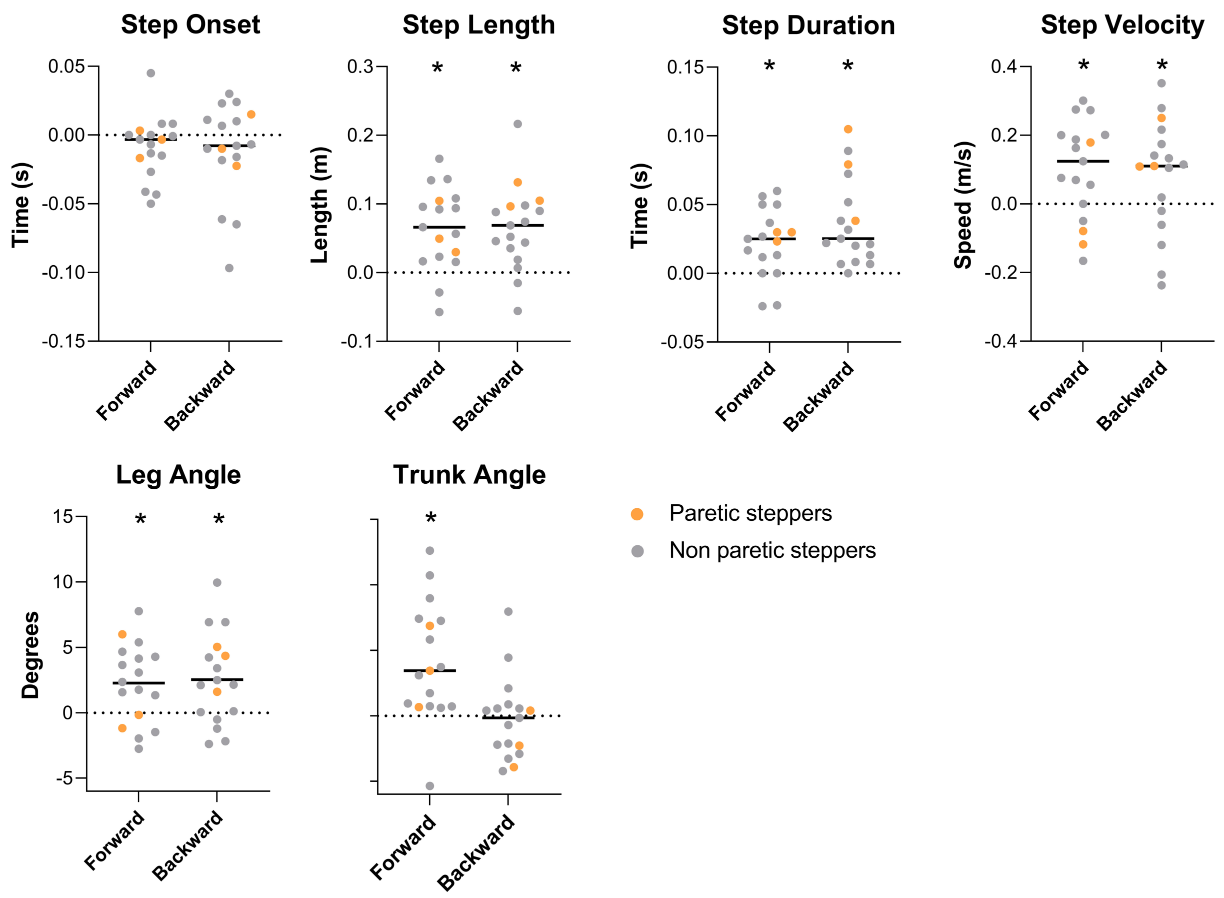

FIGURE 2
Spatiotemporal and body configurational outcome effect of training. Outcome measures are expressed as pre-post training differences. Bars represent group average, dots represent individual data points; participants who stepped with the paretic leg are indicated as orange dots. * Represents a significant pre-post training difference ($P < 0.05$).

Strong correlations between leg angle and spatiotemporal outcomes

The univariate analyses yielded moderate to strong correlations of leg angle with step length ($r = 0.87$, $p < 0.01$), step duration ($r = 0.52$, $p < 0.01$), and step velocity ($r = 0.91$, $p < 0.01$) for forward stepping. For backward stepping moderate to strong correlations of leg angle were observed with step length ($r = 0.87$, $p < 0.01$), step duration ($r = 0.64$, $p < 0.01$) and step velocity ($r = 0.67$, $p < 0.01$). None of the outcomes in muscle recruitment were correlated with changes in leg angle. The subsequent multivariate step-wise regression analyses for leg angle revealed that during forward perturbations, an increase in step velocity and step length together explained 90% of the variance in the pre-post leg angle differences ($F = 62.7$ $R^2 = 0.90$, $p < 0.001$, $\beta^{velocity} = 12.01$, $p < 0.01$, $\beta^{length} = 22.00$, $p < 0.01$). For backward stepping, an increase in step length explained 76% of the variance in pre-post leg angle differences ($F = 47.57$, $R^2 = 0.76$, $p < 0.001$ $\beta = 48.77$). None of the outcome measures used in the current study were significantly correlated to changes in trunk angle for either backward or forward perturbations.

Discussion

We aimed to evaluate whether PBT would improve APRs and spatiotemporal characteristics of reactive stepping in people in the chronic phase of stroke. In addition, we determined whether improvements in leg and trunk angles could be induced by underlying spatiotemporal and neuromuscular components. At the level of APRs, our findings showed a slight shortening in onset latencies during forward perturbations, with a similar tendency in the stepping leg during backward perturbations, whereas no differences were observed in magnitude or modulation. Analysis of spatiotemporal characteristics revealed improvements in step length, step duration, and step velocity. Leg angles improved after PBT in both perturbation directions, whereas trunk angles only significantly improved in forward steps. Improvements in leg angle were largely explained by larger step lengths and faster step velocities after training, whereas improvements in APR latencies did not contribute significantly to the changes in leg angle. None of the spatiotemporal or EMG parameters correlated with improvements in trunk angle.

The presently reported changes in trunk angle at step contact following forward perturbations complement the previously reported improvements in leg angle (16). Specifically, we observed a more upright trunk position at the post-training assessment, which is in accordance with the effects of trip-specific training as reported by Pigman et al. (20) and Nevisipour et al. (27). Smaller forward trunk rotation angles or angular velocities are correlated with better postural stability (28, 29). The more upright position after training indicates that participants were able to generate a more effective response to counteract the induced forward angular momentum. This finding, in combination with the larger leg angles, demonstrates that our participants had improved their reactive stepping performance following PBT, which is in agreement with observations from other PBT studies in people with stroke (22, 30). Yet, despite the growing evidence of beneficial effects of PBT on performance-related outcomes, there is a lack of insight into the mechanisms underlying these improvements. Therefore, we performed a comprehensive analysis of possible determinants that may have contributed to the observed improvements in reactive step quality.

We investigated training-induced changes in APR recruitment, because PwS often have persistent deficits in APR onset latency, magnitude and coordination patterns, mainly on the paretic side (9–11). The vast majority of our participants used their paretic leg as the stance leg ($n = 14$), and Table 3 shows that APR recruitment commenced ∼10–20 ms later with lower magnitudes and poorer modulation in the stance compared to the stepping leg. These observations suggest that participants who presented with APR deficits could potentially improve by training. Indeed, after PBT we observed faster APR onsets in lower-leg muscles, which finding complements the very sparse studies that have reported effects of training on APRs in PwS (31, 32), with only a single study thus far showing significant improvements in the paretic leg (33). The average shortening of onset latencies, however, was rather modest (5–6 ms), which limits the potential clinical relevance.

In accordance with studies indicating direction-specific deficits in coordination of APRs in the paretic leg of people with chronic stroke (9, 34), the modulation index was generally lower (i.e., poorer) in the (predominantly paretic) stance leg of our participants. Indeed, the mean pre-intervention modulation indices in the stance leg of 20–70% are all below the average of 71% as reported for a combined group of healthy older individuals and people with Parkinson's disease (26). This result shows that there was room for improvement in stance-leg APR recruitment in our participants, yet we did not observe PBT-related gains in the magnitude of agonistic muscle recruitment

TABLE 3 Mean (SD) pre and post-training values of muscle onset latencies, amplitudes and modulation indices.

	Pre intervention	Post intervention	*P*-value
Stepping direction			
Forward			
Stepping leg			
Gm Onset (ms)	157 (16)	152 (15)	**0.01***
Gm iEmg (μV)	11.2 (10.1)	11.0 (5.8)	0.9
Stance leg			
Gm Onset (ms)	174 (23)	168 (23)	**0.02***
Gm iEmg (μV)	5.9 (3.8)	6.4 (3.9)	0.55
Backward			
Stepping leg			
Ta Onset (ms)	153 (17)	149 (17)	***0.06***
Rf Onset (ms)	169 (22)	165 (22)	0.16
Ta iEmg (μV)	17.0 (7.4)	15.6 (10.6)	0.33
Rf iEMg (μV)	2.7 (2.6)	2.9 (2.4)	0.66
Stance leg			
Ta Onset (ms)	163 (17)	162 (19)	0.68
Rf Onset (ms)	187 (19)	184 (18)	0.21
Ta iEmg (μV)	11.5 (5.7)	10 (6.7)	0.52
Rf iEMg (μV)	3.2 (4.2)	2.6 (2.4)	0.49
Modulation index (%)			
Stepping leg			
Ta	82 (15)	80 (31)	0.83
Rf	57 (26)	58 (32)	0.73
Gm	81 (10)	84 (11)	0.37
Stance leg			
Ta	70 (36)	78 (23)	0.43
Rf	20 (104)	40 (35)	0.10
Gm	60 (45)	65 (47)	0.24

Statistically significant different outcome measures are highlighted in bold $P < 0.05$.

(iEMG), nor in direction-dependent modulation (Modulation Index). The present lack of chance in APR recruitment is reminiscent of previous work on recovery of gait in the subacute phase after stroke (35, 36). These studies showed that walking ability substantially improved in the absence of significant changes in aberrant muscle coordination, presumably through behavioral substitution rather than restoration of function. Likewise, we surmise that the beneficial effects of our PBT intervention on reactive step quality, in the absence of improvements in APR recruitment, may also point at behavioral substitution rather than restoration of function (37, 38).

Following the PBT intervention, we found significant improvements in spatiotemporal outcomes of reactive stepping, with greater step lengths, step duration and step velocity being observed in both perturbation directions. These findings complement previous studies that reported variable effects of PBT on one or more spatiotemporal step characteristics, in addition to consistently positive effects of PBT on reactive stepping performance (17, 20, 34). Our regression analyses provided novel insight into the relationships between changes in spatiotemporal step characteristics on the one hand, and gains in reactive step quality on the other hand. Increased step lengths were significantly associated with larger leg angles (i.e., better step quality) in both backward and forward perturbation directions, whereas increased step velocity was identified as an additional significant contributor in the forward direction only. Under the assumption of unchanged perturbation-induced CoM dynamics, a larger step length provides a greater lever arm for the ground reaction force to produce torques that counteract the CoM movement. A faster stepping velocity, in turn, reduces the time for the CoM to accelerate and displace relative to the base of support, thus making it easier to be "caught" at foot strike. Hence, these improvements in spatiotemporal step characteristics provide a biomechanically plausible explanation for the observed gains in reactive step quality.

While in our regression models, faster APR onset latencies did not additionally contribute to the explained variance in step quality improvements, it cannot be excluded that a faster response in the support leg (paretic leg for the majority of participants) may have been beneficial. The faster response could reduce the angular momentum generated by the perturbation, thereby providing extra time for the stepping leg and allowing better clearance for proper positioning of the stepping limb (39). Yet as stated before, the modest 5–6 ms shortening that we observed in the present study may not represent a substantial benefit.

A limitation of our study is that our analysis did not permit comparing the effects of training on the paretic and non-paretic legs separately. We allowed our participants to self-select their stepping limb as this would resemble their most instinctive stepping response, but this instruction resulted in few participants selecting their paretic leg. Therefore, we suggest for future studies to also examine imposed stepping with the paretic and non-paretic leg separately. In addition, to improve the generalizability of the current results toward a greater population of PwS, it would be preferable to increase the sample size.

In summary, the current improvements in step quality after PBT in our group of participants people with chronic stroke were largely explained by improved spatiotemporal characteristics and not by changes in APR recruitment. For gaining further insight into the observed effects of training on spatiotemporal characteristics of reactive stepping, it may be of interest to study changes in muscle recruitment during execution of the recovery step itself, in addition to the present focus on the APR that precedes stepping.

Data availability statement

The original contributions presented in the study are included in the article/Supplementary material, further inquiries can be directed to the corresponding author.

Ethics statement

The studies involving human participants were reviewed and approved by the Medical Ethical Board of the region Arnhem-Nijmegen. The patients/participants provided their written informed consent to participate in this study.

Author contributions

WS: analyzed data and wrote paper. HD and JR: designed, conducted experiment, and critical revision. SZ: supervision throughout analysis and draft of the manuscript. JB and FL: eligibility assessment of participants and critical revision. AG: eligibility assessment of participants, supervised WS throughout the study, and provided feedback. VW: designed experiment, supervised WS throughout the study, and provided feedback. All authors contributed to the article and approved the submitted version.

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Conflict of interest

Authors FL, AG, and VW were employed by Sint Maartenskliniek.

The remaining authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Supplementary material

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fspor.2022.1008236/full#supplementary-material

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A proposed methodology for trip recovery training without a specialized treadmill

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Falls are the leading cause of accidental injuries among adults aged 65 years and older. Perturbation-based balance training is a novel exercise-based fall prevention intervention that has shown promise in reducing falls. Trip recovery training is a form of perturbation-based balance training that targets trip-induced falls. Trip recovery training typically requires the use of a specialized treadmill, the cost of which may present a barrier for use in some settings. The goal of this paper is to present a methodology for trip recovery training that does not require a specialized treadmill. A trial is planned in the near future to evaluate its effectiveness. If effective, non-treadmill trip recovery training could provide a lower cost method of perturbation-based balance training, and facilitate greater implementation outside of the research environment.

KEYWORDS

trip recovery, training, falls, perturbation, balance, treadmill

Introduction

Falls are the leading cause of both non-fatal and fatal injuries among adults aged 65 years and older in the United States (1, 2). Falls are also costly in that the 2015 direct medical costs associated with falls among older adults in the United States totaled $50 billion (3). Falls and fall-related injuries are prevalent among older adults largely because of the declines in physical (4) and/or cognitive (5) capabilities with aging.

Trips account for 29%—53% of falls among community-dwelling older adults (6–8). These trip-induced falls frequently result from an ineffective balance recovery response to the trip-induced loss of balance (LOB) (9). Perturbation-based balance training (PBT) has received growing interest as an exercise-based fall prevention intervention (10–13), and accumulating evidence supports its ability to improve balance recovery responses as well as reduce fall rates (10). While some PBT studies have targeted disease-specific populations (14, 15), most aim to reduce falls among older adults (10, 12). The goal of PBT is to train and thus improve this recovery response. Many PBT efforts specifically target trip-induced falls (16–19). This so-called trip recovery training can improve balance recovery responses to lab-induced trips (16, 18, 20), and decrease fall rates after both lab-induced trips (16, 18) and real world trips (21). Trip recovery training has been employed using varied means to induce trips or trip-like perturbations. For example, (13, 22) used an electromechanical tripping obstacle embedded within a laboratory walkway that abruptly raised during early/mid-swing to induce a trip. Other studies have employed a specialized treadmill to elicit trip-like perturbations. For example, (16, 17) had participants stand on a stationary treadmill belt and suddenly accelerated the belt posteriorly to induce a forward LOB, while (19) had participants walk on a treadmill and applied sudden belt accelerations. Treadmill-assisted trip recovery training has been conducted using commercially-available specialized treadmills marketed for PBT (19, 23–25) as well as a lower cost option using a modified off-the-shelf treadmill (17, 20). The cost and/or space requirements associated with an electromechanical tripping obstacle within a walkway or a specialized treadmill can present a barrier to wider application trip recovery training (26). A trip recovery training regimen that does not require either may facilitate its use outside the research setting.

This paper reports a proposed methodology for trip recovery training that does not require an electromechanical trip obstacle or specialized treadmill. Successful balance recovery after a trip-induced LOB has three primary requirements: (1) quickly step anteriorly to extend the base of support and enable the ground reaction force line of action to be anterior to the whole-body center of mass; (2) quickly decelerate the forward angular velocity of the trunk segment; and (3) maintain sufficient stance limb hip height to enable stepping over the obstacle (9, 27, 28). Similar to other treadmill-based trip recovery training programs (17, 25, 29–31), the so-called non-treadmill training (NT) regimen proposed here targets these requirements through volitional step training and reactive step training, both of which can improve fall rates and risk factors for falls (32). Moreover, the step training within NT closely mimics the postures and movements required during trip recovery to leverage the specificity of training principle and thus enhance transfer to trip recovery. NT was developed by the authors based upon their expertise and experience studying trips and administering PBT among older adults. Approximately 20 pilot participants were used to refine the NT procedure described below, although no formal evaluation of its effect on trip-induced LOB responses has been completed. A trial is planned in the near future to evaluate its effectiveness on laboratory-induced trips in comparison to treadmill-based trip training and a control among community-dwelling older adults. If effective, NT could provide a lower cost method for trip recovery training and facilitate greater implementation outside of the research environment (12, 26).

Methods

Non-treadmill training is performed over an area ∼1.2 m wide by 4 m long, and uses an 8-cm-tall wooden tripping obstacle fastened to a sheet of plywood on the floor with padding affixed to vertical face where foot contact is anticipated during a trip. Each NT session includes four phases of training with increasing difficulty and similarity to actual trip recovery. Time lapse photographs of each phase are illustrated in Figure 1. Each NT session involves a single participant, is designed to be ∼40 min in duration, and begins with a 3-min warm-up of walking and light stretching. The number of trials recommended within each phase is not specified because no empirical evidence is available at this time to support any such recommendation. In general, trainers should endeavor to complete a large number of trials because learning will increase with added practice, but also maintain a comfortable and enjoyable pace for the participant with time for trainer encouragement, feedback, and possible rest breaks. We anticipate multiple NT sessions being completed by participants to achieve meaningful and lasting improvements in trip recovery.

Phase 1 – Rapid Stepping targets the need to quickly step anteriorly to extend the base of support (27, 33, 34). It involves volitional stepping exercises from bilateral standing during which the participant starts to tip and fall forward by rotating about their ankles, and then takes quick steps to recover balance. This is repeated numerous times while the participant steps initially with the left and right feet with approximately equal frequency since trip recovery may require both. In this and all phases, participants are encouraged to complete multiple steps and achieve a stable gait even though instructions are only focused on the initial step. When the participant appears to execute these steps with little difficulty, the difficulty can be increased by encouraging the participant to fall as far forward as possible before starting to step, and also to take a long initial recovery step. Moreover, stepping is first performed without the tripping obstacle installed, and then with the tripping obstacle to elicit a step over an obstacle such as during trip recovery. The distance from the participant's initial standing position and the obstacle should initially be at a comfortable distance for stepping over (∼7–20 cm), with this distance being increases as a part of making this phase more challenging.

Phase 2 – Trunk Control targets the need to quickly decelerate the forward angular velocity of the trunk segment (33, 35–37). It involves similar volitional stepping exercises as Phase 1, but with explicit instructions and emphasis on arresting trunk motion. To accomplish this, the participant is instructed to control their trunk segment angular orientation to be vertical at touchdown of the first recovery step. While achieving a vertical trunk segment orientation at touchdown is not a requisite for successful trip recovery, we found this to be a useful mnemonic to encourage participants to focus on controlling trunk motion. As in Phase 1, this is repeated numerous times while stepping initially with the left and right feet, initially without the tripping obstacle, and later with it.

Phase 3 – Lean Release targets the need to accomplish the same requirements as in Phases 1 and 2, but in response to an unexpected LOB rather than in a volitional sense. The participant leans forward while being supported bilaterally at

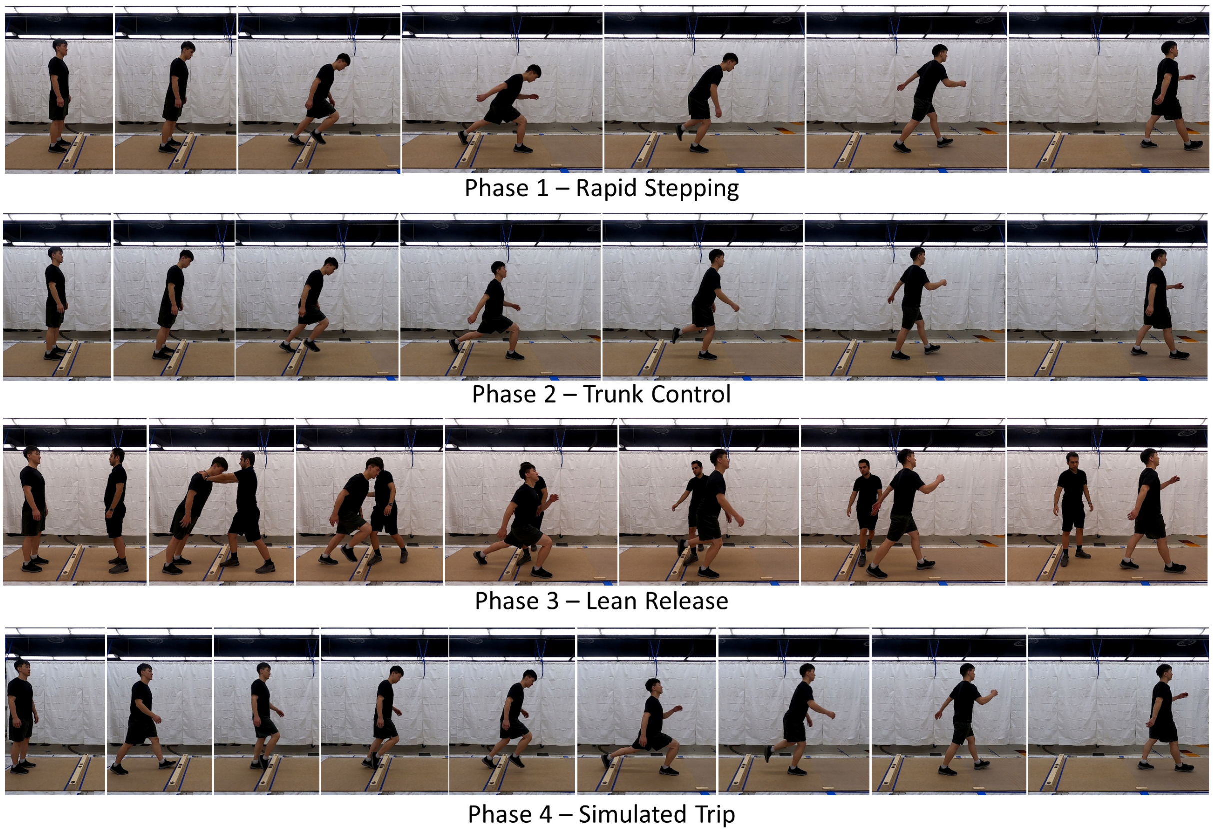

FIGURE 1
Time lapse photos of the four phases of non-treadmill training. In Phase 1, participants practiced a long and quick initial step after purposefully rotating forward about the ankles to induce a forward fall. To increase the difficulty as performance improved, participants were encouraged to delay the start of their stepping as long as possible. In Phase 2, participants also practiced a long and quick initial step after purposefully rotating forward about the ankle for as long as possible. However, emphasis was placed on controlling the sagittal plane trunk angle during the initial recovery step by aiming to achieve a vertical trunk orientation at the time of touchdown of the initial recovery step. In Phase 3, we added a reactive component by releasing participants from a static forward lean without warning. Participants focused on a long, quick initial step and trunk control as emphasized in Phases 1 and 2. In Phase 4, participants were asked to self-induced a trip while walking and practice a long, quick initial step and trunk control as emphasized in earlier Phases.

the shoulders by a trainer standing and facing the participant with their arms fully extended. Without warning, the trainer releases the participant and steps to the side. The participant quickly takes recovery steps to recover balance. The participant is reminded to emphasize trunk segment control as in Phase 2. As in Phases 1 and 2, this is repeated numerous times while stepping initially with the left and right feet, initially without the tripping obstacle, and later with it. A verbal cue of release can be provided to the participant, if needed for confidence or frequent success. The cue can be eliminated later in training as performance improves.

Phase 4 – Simulated Trip attempts to integrate the requirements targeted in Phases 1 and 2 into a realistic trip. The participant starts by standing one step away from the tripping obstacle. The participant then steps with their first foot, and during the subsequent swing phase purposefully trips on the obstacle. The participant then executes an elevating strategy by using the obstructed foot to step over the obstacle, and then continues walking. As in earlier phases, this is repeated numerous times while stepping initially with the left and right feet. The participant will be instructed to emphasize taking a long initial recovery step, and controlling trunk segment by achieving a vertical angular orientation at touchdown of the first recovery step. To increase difficulty later in training, the participant can start more than one step away from the tripping obstacle.

The goal for the NT trainer should be to include all four phases during each session. However, NT can and should be individualized to each participant's capability, and completing all four phases should not come at the expense of participant comfort. Spending additional time in early phases early in the training to ensure the participant does not overexert themselves and to build the comfort and confidence with the training is likely important. Also, depending upon the physical capability of the participant and the speed at which they are able to learn the movements involved, some participants may need to spend additional time in early phases and not complete all four phases during initial NT sessions.

Anticipated results and discussion

We anticipate NT to have a measure of acceptability among older adult participants. This expectation is based upon qualitative similarities between NT and treadmill-based trip recovery training programs and the acceptability that has been provided to the latter (26). We also anticipate NT to elicit improvements in trip recovery after laboratory-induced trips when compared to a control involving general balance and strength exercises not specific to tripping. More specifically, we anticipate improved stepping responses and trunk control following NT. This expectation is based upon a systematic review and meta-analysis indicating volitional step training and reactive step training among older adults improve fall rates and fall risk factors such as reaction time, gait, balance, and balance recovery (32). It is unclear at this time how the efficacy of NT will compare to trip recovery training using a treadmill, as well as comparing how both are received by the targeted population of community-dwelling older adults. Subsequent studies will be needed to determine how well NT transfers to fall reduction in the real-world environment.

Participants with significant lower limb joint pain gait impairment, or who are dependent upon a walking aid may not be a good fit for the proposed NT. No explicit age range is provided either given that eligibility should be based upon gait and balance ability. NT does have safety risks. As with other PBT regimens, NT risks include exacerbation of preexisting medical conditions, overexertion, tissue strains, and fall-induced injury. To minimize these risks, participants should be screened by a qualified health professional prior to NT, warmup exercises and stretching are recommended, and rest breaks can be includes as needed. The need for a safety harness is dependent upon participant physical capability and confidence level. Our upcoming trial will involve community-dwelling older adults, whom we will attempt to train with a spotter and no harness to avoid added infrastructure. NT participants should also be encouraged to wear suitable clothing and footwear.

The trainer administering NT should have requisite traits to enhance training efficacy and safety. NT as proposed here has no formal or objective quantification of participant performance during training. Because of this, modulating perturbation magnitude and difficulty so that training can be individualized and progress as the participant improves requires trainer experience and intuition. If no safety harness is used, then they should also have sufficient size and physical capacity to provide fall-arresting assistance when needed. Regardless as to whether a safety harness is used, we anticipate the trainer standing near the participant during all phases of NT to demonstrate the movements, facilitate feedback, provide physical support when needed, and also for encouragement.

We anticipate participants most likely needing to complete multiple NT sessions to achieve meaningful improvements in trip recovery. However, the number of sessions of NT needed to elicit meaningful improvements in trip recovery, as well as the optimal training schedule, have not been evaluated. We also acknowledge that only one of the four phases of the trip recovery training proposed here involves reactive stepping to perturbations that occur without warning (Phase 3). Many other trip recovery training methods reported elsewhere fully involve reactive stepping responses to sudden perturbations. While such reactive stepping appears to be more specific to balance recovery responses after perturbations, volitional stepping exercises such as those used in Phases 1, 2, and 4 can also improve fall risk factors and reduce fall rates (32), and support the potential benefits of the training proposed here.

In conclusion, a methodology for trip recovery training that does not require a specialized treadmill is presented. If acceptability by participants and effective, this training could provide a lower cost implementation of trip recovery training, and facilitate greater implementation outside of the research environment.

Data availability statement

The original contributions presented in the study are included in the article/supplementary material, further inquiries can be directed to the corresponding author.

Ethics statement

Written informed consent was obtained from the individual(s) for the publication of any potentially identifiable images or data included in this article.

Author contributions

YL: conceptualization, methodology, and writing—original draft. NA: supervision and writing—review and editing. MM: conceptualization, methodology, writing—original draft, and project administration. All authors contributed to the article and approved the submitted version.

Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

Publisher's note

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