Human resilience to forward falls: adaptation and transfer of stability control

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Statutory declaration

I hereby affirm in lieu of an oath that the research presented within this thesis is my own original work unless stated. All citations, either direct quotations or passages which were reproduced verbatim or nearby verbatim from publications are indicated and the respective references are named. The same is true for tables and figures. No contents have been submitted either in whole or in part for a degree at London South Bank University or any other institution.

London, 18th of October 2022

J Werth

Mr Julian Werth

Acknowledgments

“I can’t see a way through”, said the boy*. “Can you see your next step?”*
“Yes”. “Just take that”, said the horse.

Charlie Mackesy

I have gone through countless moments when everything seemed overwhelming, when I couldn't see any light at the end of the tunnel. Surely, I would not have been able to achieve where I am without being surrounded by an amazing team, without constant support of friends and colleagues, and without my caring mother.

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Summary

Scoping fall resilience requires knowledge of factors enabling the neuromotor system to transfer stability control between different postural perturbations. This thesis addressed this objective in comprising three different studies on adults across the lifespan. The first study examined the intra- and inter-session reliability of recovery performance across 97 participants at several research centres using two different protocols of a clinical assessment method (lean-and-release task) simulating sudden anterior stability loss, i.e. gradual increase to maximal forward-lean angle vs. predefined lean angle. Independent of the protocol used and participants' age, reliable assessment of common stability recovery performance parameters using the lean-and-release task could be confirmed. The second study used single exposures to both lean-and-release and a treadmill-based gait trip to investigate the association of recovery performance between unpractised perturbations. We revealed that recovery performance in one task could not significantly explain performance in the other task, indicating limited transfer of fall-resisting skills for anterior perturbations. The third study examined factors (particularly practising stability recovery responses with different perturbation magnitudes) that could elicit or limit transfer to unpractised perturbations. Participants walking on a treadmill were exposed to eight trip perturbations of either low or high magnitude or walked unperturbed (control group). To investigate transfer to unpractised anterior perturbations following walking tasks, all participants underwent a lean-and-release task and an overground trip. Adaptation in stability to repeated gait-perturbations did not lead to enhanced stability recovery in the lean-and-release task but did improve overground trip performance, independent of the practised perturbation magnitude. Lower limb joint angle differences between treadmill- and lean-and-release perturbations for the swing phase of recovery steps were more prolonged and greater as opposed to the comparison of the two gait perturbation tasks. In conclusion, the current work indicates that practising stability control enhances human resilience to unpractised perturbations which is not necessarily dependent on the perturbation magnitude but may partly be subject to similarity in motor response patterns between tasks.

List of publications related the thesis' topic

1. **Werth, J.**, Epro, G., König, M., Santuz, A., Seeley, J., Arampatzis, A. & Karamanidis, K. (2022). Differences in motor responses to stability perturbations limit fall-resisting skill transfer. *Scientific Reports*, *12*(21901).
2. Weber, A., Hartmann, U., **Werth, J.**, Epro, G., Seeley, J., Nickel, P., & Karamanidis, K. (2022). Limited transfer and retention of locomotor adaptations from virtual reality obstacle avoidance to the physical world. *Scientific Reports*, *12*(1), 1-9.
3. König, M., Santuz, A., Epro, G., **Werth, J.**, Arampatzis, A., & Karamanidis, K. (2022). Differences in muscle synergies among recovery responses limit inter-task generalisation of stability performance. *Human Movement Science*, *82*, 102937.
4. Weber, A., **Werth, J**., Epro, G., Friemert, D., Hartmann, U., Lambrianides, Y., Seeley, J., Nickel, P. & Karamanidis, K. (2022). Head-mounted and hand-held displays diminish the effectiveness of fall-resisting skills. *Sensors*, 22(1), 344.
5. Weber, A., Friemert, D., Hartmann, U., Epro, G., Seeley, J., **Werth, J**., Nickel, P. & Karamanidis, K. (2021). Obstacle avoidance training in virtual environments leads to limb-specific locomotor adaptations but not to interlimb transfer in healthy young adults. *Journal of Biomechanics*, 120, 110357.
6. Bosquée, J., **Werth, J.**, Epro, G., Hülsdünker, T., Potthast, W., Meijer, K., Ellegast, R. & Karamanidis, K. (2021). The ability to increase the base of support and recover stability is limited in its generalisation for different balance perturbation tasks. *European Review of Aging and Physical Activity*, *18*(1), 1-10.
7. **Werth, J.**, Bohm, S., Klenk, J., König, M., Sczuka, K. S., Schroll, A., Epro, G., Mandla-Liebsch, M., Rapp, K., Potthast, W., Arampatzis, A. & Karamanidis, K. (2021). Stability recovery performance in adults over a wide age range: A multicentre reliability analysis using different lean-and-release test protocols. *Journal of Biomechanics*, 125, 110584.
8. **Werth, J**., König, M., Epro, G., Seeley, J., Potthast, W., & Karamanidis, K. (2021). Volitional step execution is an ineffective predictor of recovery performance after sudden balance loss across the age range. *Human Movement*

Science, K. (2022). Differences in muscle synergies among recovery responses limit inter-task generalisation of stability performance. *Human Movement Science*, 82, 102937.

List of congress proceedings

1. **Werth, J.**, Epro, G., Santuz, A., König, M., Seeley, J., Arampatzis, A., Karamanidis, K. Impact of short-term perturbation exercise dose on transfer of fall-resisting skill adaptations (European College of Sport Science Virtual Congress 2021; oral; 26-0746); *5th place within Young Investigator Award*.
2. **Werth, J.**, Santuz, A., Epro, G., König, M., Arampatzis, A., Karamanidis, K. Modular control of motor output as a limiting factor for fall-resisting skill learning (European College of Sport Science Virtual Congress 2020; oral; 25-1446).
3. **Werth, J.**, König, M., Epro, G., Seeley, J., Potthast, W., Karamanidis, K. Voluntary step execution is a limited predictor for recovery performance after sudden balance loss (European College of Sport Science Congress 2019 in Prague/Czech Republic; poster; 24-2076).
4. König, M., Epro, G., **Werth, J.**, Seeley, J., Potthast, W., Karamanidis, K. Task-specific adaptability but diminished retention of fall-resisting skills in old age (European College of Sport Science Congress 2019 in Prague/Czech Republic; oral; 24-1233); *5th place within Young Investigator Award*.

List of figures and tables

following lean-and-release (Overground trip perturbation), all participants were exposed to one trip-like overground gait perturbation (gait speed matched to the treadmill speed at 1.4 $m \cdot s^{-1}$) induced using a method as for treadmill-based trip perturbations. A safety harness was worn during all tasks to prevent contact of any part of the body with the ground (except for the feet).

Figure 14: Margin of stability (MoS, top) and base of support (BoS, bottom) during for unperturbed baseline walking (B, BASE) and the first (T1) and eighth (T8) trials of treadmill-based perturbation ~~exercise~~ practice for low (n = 10) and high (n = 10) perturbation magnitude groups. Data for T1 and T8 is shown at left foot touchdown after perturbation (PERT) and the subsequent right foot touchdown after recovery step (REC). ~~Please n~~ Note that PERT at T1 (9.9 ± 6.9 cm) and T8 (10.4 ± 8.5 cm) showed quite similar mean values for the high perturbation magnitude group. Values are presented as means with SD error bars. ‡: sig. different to BASE at T1 and T8 for low and high ($p < 0.05$); *: sig. different between T1 and T8 for low and high ($p < 0.001$); †: sig. different between low and high at T1 and T8 ($p < 0.05$).

Figure 15: Margin of stability (MoS, top) and base of support (BoS, bottom) during lean-and-release and overground trip transfer tasks for low (n = 10) and high (n = 10) perturbation magnitude groups as well as controls (n = 10; CTRL). Data is shown for cable release or left foot touchdown after gait perturbation (PERT) and subsequent right foot touchdown for recovery step (REC) for all groups and unperturbed baseline walking (BASE) only for the overground task. Note that the BoS at PERT equalled zero for all groups during lean-and-release and hence is not shown. Values are presented as means with SD error bars. ‡: sig. different to BASE ($p < 0.001$); #: sig. different to CTRL ($p < 0.05$).

Figure 16: Sagittal plane joint angle kinematics for the low perturbation magnitude group (n = 10) and statistical parametric mapping (SPM) analyses of the ankle, knee, and hip joint during the entire swing phase (toe-off to touchdown, 0-100 %) of the recovery step for the eighth treadmill-based trip (TRM), lean-and-release (LRT), and overground trip (OVG). *1st row*: Joint angle comparison between all three tasks via means ± standard deviation (bold lines and shaded areas) across participants. *2nd row*: SPM one-way repeated measures ANOVA

average over the control group (no practice; n = 10) at touch-down of the recovery step for each of the transfer tasks.

Table 1: Intraclass correlation (ICC [3,1]) coefficients with 95 % confidence intervals (CI_{95}) and range of root mean square errors (RMSE; in cm) for margin of stability at release (MoS_{RL}) and at touchdown (MoS_{TD}) as well as the base of support at touchdown (BoS_{TD}) between the measurements (Baseline, Post 0.5 h and Post 48 h) for the maximal and single lean angle protocols (n = 43 and n = 26 respectively).

List of abbreviations

η_p^2	Partial Eta Squared (effect size)
°	Degree of joint angles
Base	Baseline/unperturbed walking
BoS	Base of support
BoS_{TD}	Base of support at touchdown
cm	Centimetres
CoM	Centre of mass
CTRL	Control group
F	Value for F-statistic
h	Hour(s)
Hz	Hertz
ICC	Intraclass correlation
kg	Kilogram
LRT	Lean-and-release task
m	Metre
m/s or $m \cdot s^{-1}$	Metre per second
MoS	Margin of stability
MoS_{RL}	Margin of stability at cable release
MoS_{TD}	Margin of stability at touchdown
ms	Milliseconds
N	Newton
n	Sample size
OVG	Overground trip perturbation
p	probability value
Pert	Timepoint of perturbation
r	Correlation coefficient
Reco1 or Rec	Touchdown first recovery step after perturbation
RMSE	Root mean squared error
SD	Standard deviation
SPM	Statistical parametric mapping
t	Value for hypothesis test statistic via Student’s t-test
TD	Touchdown
TOST	Two one-sided test
Trial 1 and 8	First and eighth treadmill-based trip perturbation
TRM	Treadmill-based trip perturbation
TRM_{high}	Higher treadmill-based trip perturbation magnitude
TRM_{low}	Lower treadmill-based trip perturbation magnitude
X_{CoM}	Extrapolated centre of mass

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1. Introduction

Human locomotion daily faces a variety of perturbations to stability. In the event of stability loss, the central nervous system must be capable of executing motor responses that are adequate to the particular perturbation (e.g. a trip situation) in order to maintain postural integrity and prevent falling. If the characteristics of the perturbation (e.g. magnitude or environment) change, the system will need to adjust its motor response to a different stability constraint. However, such adjustment might be readily achieved if the perturbations elicited some shared motor output for stability recovery, which would decrease the complexity of motor control adjustments and hence generalise specific fall-resisting skills to various situations of challenged stability. To address these assumptions and extend the knowledge of fall resilience, the current thesis focused on different perturbation paradigms that elicit a similar stability recovery response and examined adaptation and transfer of motor control in adults across the lifespan. This introduction will provide a brief overview of the prevalence and incidence of falls, followed by an evidence-based review of motor output to recover stability, their assessment, adaptability, and transfer across perturbations.

1.1. Prevalence of falls

Although falls are observed across the entire adult life span, their prevalence increases with age (Peel et al., 2002; Schumacher et al., 2014; Talbot et al., 2005). Approximately every tenth adult aged over 40 years, and every fourth to every third aged 65 or older falls at least once a year (Bergen et al., 2016; Palumbo et al., 2016; Rapp et al., 2014; Schumacher et al., 2014), with a disposition assumed for recurring falls in later years if a fall occurred earlier (Schumacher et al., 2014). A person who experiences one or more falls often suffers from morbidity (e.g. hip fractures and undifferentiated bone fractures) and immobility, as well as mortality (Burns & Kakara, 2018; Rubenstein, 2006; Terroso et al., 2014). Such situations result in enormous costs for health care systems (Florence et al., 2018), but not only physical, also psychological consequences, such as post-fall anxiety syndromes, leading to lack of confidence during activities and substantially reducing the quality of daily living

(Stenhagen et al., 2014; Talbot et al., 2005; Yardley et al., 2002). The strong association between ageing and falls is underlain by multifactorial changes (e.g. of biological, behavioural, and environmental natures) or pathologies such as declines in auditory, vision, vestibular function, musculoskeletal properties and function, proprioception and touch sensitivity that may degrade execution of motor responses during locomotion in the context of perturbed stability (Paraskevoudi et al., 2018; Rubenstein, 2006; Sturnieks et al., 2012; Terroso et al., 2014). With respect to the established likelihood of declines in gait and stability control in the older population, there is a need to understand fall initiation and how to cope with it.

1.2. A biomechanical perspective on incidence of falls

Evidence for the incidence of falls confirms these occurring majorly during locomotion (e.g. due to slips or trips; Chrenshaw et al., 2017; Talbot et al., 2005). About 20% of all indoor falls and 60% of all outdoor falls in older adults were reported to result from a slip or stumbling (Luukinen et al., 2000). With regards to observational research carried out in long-term care centres, most falls were observed after stability loss due to an incorrect shift of body mass or external hazards (Robinovitch et al., 2013; Yang et al., 2018), with 30% of those resulting from a trip causing sudden loss of stability in the anterior direction (Yang et al., 2018). Even in younger adults (18-35 years of age), the distribution of falls incidence seems similar, with the main causes reported being slips or trips (Heijnen & Rietdyk, 2016). Putting these prominent challenges to postural integrity and their consequences of stability loss and falls into a mechanical perspective, any perturbations can cause excursions of the body's centre of mass towards or beyond the limits of stability, i.e. the boundaries of the base of support (the area underneath and between the feet; Woollacott & Shumway-Cook, 1996). Note that, particularly during locomotion, not only the position but also the velocity of the centre of mass must be considered to quantify stable or unstable body configurations (Hof et al., 2005; Pai & Patton, 1997). A frequently used parameter that accounts for such spatial and temporal components of the centre of mass in relation to the boundaries of the base of support is based on an inverted pendulum model of the body, i.e. the margin of stability (Hof, 2008; Hof et al., 2005). The model incorporates the horizontal distance between the boundaries of the base of support and the extrapolated centre of mass which is defined by the sum of the horizontal position of the centre of mass and its horizontal velocity in relation to the eigenfrequency of the inverted pendulum, i.e. the square root of gravitational acceleration divided by the pendulum length (ankle joint axis of rotation to the centre of mass). The model relies on the assumption that excursions of the centre of mass are inevitable but relatively small in relation to a pendulum length that is constant during unperturbed human locomotion (Hof et al., 2005). In the event of external perturbations (e.g. trips or slips), excursions of the centre of mass are no longer purely caused by the movement of the whole-body due to locomotion. Centre of mass changes in these situations are likely to be larger and may not be retained within the limits of stability (i.e. negative margin of stability) for subsequent motor actions (e.g. step; Bruijn & van Dieën, 2018). This substantially increases the degree of instability and hence the risk of a fall. In order to cope effectively with stability loss, the central nervous system must execute appropriate motor output to restore the desired motion state of the centre of mass, i.e. stable body configuration.

1.3. Stability recovery responses to perturbations

Motor control during perturbed stability is suggested to rely on either continuous sensory input (an immediate reactive feedback-driven response to the perturbation; Bierbaum et al., 2011; MacLellan & Patla, 2006; Marigold & Patla, 2002; Pai et al., 2003) or on prior knowledge of the perturbation or environmental context (proactive adjustments; Bhatt et al., 2006; Bierbaum et al., 2010; Bohm et al., 2012; McCrum et al., 2016; Shumway-Cook & Woollacott, 2007). In either case, the extent of perturbation-induced centre of mass displacement in relation to the limits of stability has been shown to determine subsequent musculoskeletal responses. In theory, there are three ways to maintain a stable body configuration (Hof, 2007; Figure 1). If the body faces minor perturbations, counter-rotation of body segments (e.g. arm or trunk movements opposing the direction of perturbation) or grasping external objects (if available) could be sufficient to regulate stability. However, in the context of larger perturbations leading to high displacements of the centre of mass, as likely induced by trips or slips, moving the centre of pressure in the direction of perturbation, e.g. by performing a step increasing the limits of stability and controlling trunk movements, has been identified as the main recovery response to stability loss most effectively during locomotion (Hof et al., 2005; Maki & McIlroy, 2006; Pavol et al., 2001; Wang et al., 2017; 2020). These stepping responses have been shown to deteriorate in older compared to younger adults or in fallers compared to non-fallers (Alissa et al., 2020; Okubo et al., 2021). Bearing in mind that such important reactive stepping actions would depend on the perturbation-induced direction of centre of mass displacement (e.g. slip situations initialise posterior stability loss that requires a motor response by means of a backward step; Martelli et al., 2017), the work for this thesis will focus on perturbations that elicit anteriorly directed stepping responses.

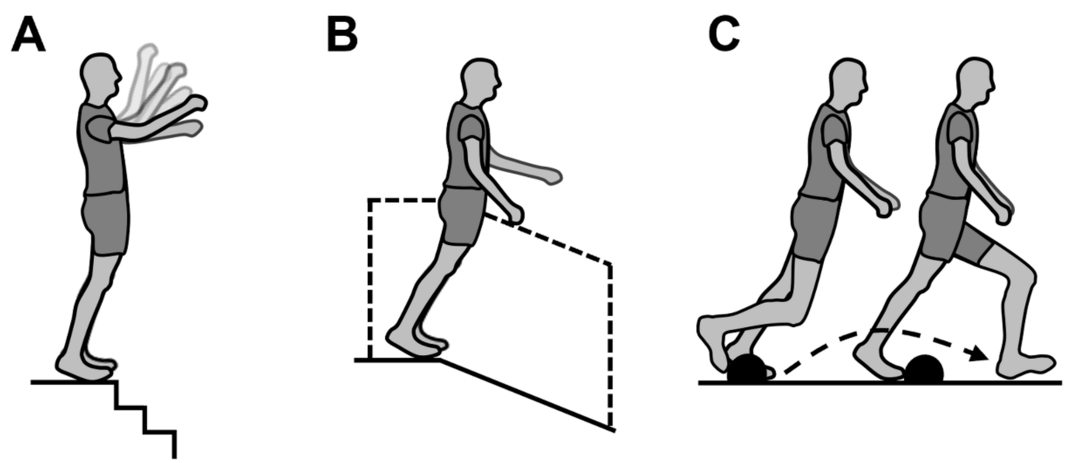

Figure 1: Three ways of maintaining stability (adapted from Hof, 2007); (**A**) Counter rotation of segments around the centre of mass, (**B**) Applying external forces other than the ground reaction force, e.g. by holding on to a handrail; (**C**) Moving the centre of pressure with respect to the vertical projection of the centre of mass, e.g. by stepping.

1.4. Assessment of recovery responses and performance

Whether for clinical settings or research, effective assessment methods must exist that attempt to match the nature of unexpected real-world perturbations and aid evaluation of the quality of stability recovery responses. The *lean-and-release* task, being a well-established method to identify individual deficiencies in stability recovery performance after sudden anterior stability, has often been used for adults over a wide age range (Arampatzis et al., 2008; Carty et al., 2012; 2015; Karamanidis & Arampatzis, 2007; Karamanidis et al., 2008). During the task, participants are released from a static forward-inclined position (their centre of mass lies outside the boundaries of the base of support, an unstable state) and encouraged to regain stability with a single forward step. Findings notably indicate deterioration in stability performance in older compared to younger populations, i.e. insufficient length of step, prolonged stepping time and often inability to fully recover within a single step (Karamanidis & Arampatzis, 2007; Karamanidis et al., 2008; Carty et al., 2015; Okubo et al., 2017; Werth et al., 2021), with such measured deterioration predicting future fall incidence (Carty et al., 2015). Depending on the experimental objective or design, two main lean-and-release task protocols have been used frequently. One consists of gradual trial-by-trial increase in displacement of the centre of mass to determine the maximal lean angle for recovery, i.e. the maximum displacement from which a person can recover with a single step (Aragão et al., 2011; Arampatzis et al., 2011; Bohm et al., 2020; Hamed et al., 2018). With regards to this form of the testing procedure, previous studies have pointed out that repetition of this simulated forward-fall elicits immediate improvement in recovery stepping performance (Carty et al., 2012; Ringhof et al., 2019), and that even small changes in stepping responses affect the ability to recover stability with a single step (Carty et al., 2012) i.e., the criterion to determine the maximal lean angle. This could bias the reliability of the testing procedure in the context of repeated measurement designs (e.g. pre-post testing involved in interventions for stability performance), considering that such task adaptation may vary between targeted populations as well as assessment timepoints.

An alternative protocol uses only one or more predefined lean angles, usually following none or a few practice trials (Carty et al., 2012; 2015; Karamanidis & Arampatzis, 2007; Karamanidis et al., 2008; Mademli et al., 2008). This protocol plays an important role in examining population-related and/or intervention effects on stability performance in a more time-efficient manner, as opposed to the maximum lean angle protocol. Yet it remains of concern whether stability performance is affected by even a few task repetitions. In fact, no systematic study has previously been carried out to confirm reliability of the assessment of stability recovery performance using either of the two protocols, including involvement of adults over a wide age range (**objective for Study 1**).

1.5. Adaptation in recovery responses and performance due to practice

Next to adequate and reliable assessment methods of recovery from stability perturbations, it is essential to develop intervention programs targeting improved stability control as well as determining their effectiveness on fall resistance. Over the years, a variety of practice-based approaches have been evaluated that incorporated interventions on stability, muscle strength/power, cardio-vascular endurance, flexibility, or mobility, altogether addressing fall mitigation in older and/or pathological populations (Cameron et al., 2018; Chang et al., 2004; Guirguis-Blake et al., 2018; Hopewell et al., 2018; Sherrington et al., 2019). In this regard, practice paradigms that involve challenges to stability have been shown to be amongst the most effective approaches (Hamed et al., 2018, Sherrington et al., 2019, Sibley et al., 2021). In addition to those approaches targeting general improvements in physical functioning and health, evidence over the past two decades favours the use of stability interventions that a) incorporate sudden perturbations large enough to induce stability loss that would lead to a fall (failing a sufficient motor response), and b) specifically simulate situations of perturbed locomotion in the real world. Such interventions could produce yet greater increases in the effectiveness of fall resistance induced by practice (Grabiner et al., 2014). This seems reasonable given that fall-resisting skills (such as reactive stepping actions after perturbations) are directly addressed rather than performance-related parameters that only have presumptive association with fall risk (Grabiner et al., 2014).

In this regard, extensive research has been conducted for which the participants are exposed to several stability challenges during locomotion that mimic real-world situations of trips or slips, i.e. practised trips or slips (Karamanidis et al., 2020). Whether targeting the improvement for slip or trip resilience, outcomes of these studies are consistent in indicating that even single sessions of repeated slip- or trip-like perturbations (as opposed to conventional long-term interventions) have the potential to elicit acute and retainable adaptations in stability control for adults across the lifespan (Bhatt & Pai, 2009; Bhatt et al., 2012; Bohm et al., 2015; Epro et al., 2018b; König et al. 2019b; Lee et al., 2018; Liu et al., 2017; McCrum et al., 2018; Wang et al., 2011; Yang et al., 2013). This suggests that repeated exposure to large perturbations leads to procedural learning of internal models for the sensorial prediction of recurrent stability challenges and hence increases facilitation of effective and efficient motor responses (Izawa et al., 2008).

It remains of great interest to understand the neurophysiological mechanisms that elicit adaptive changes in stability control and its facilitation due to practised exposure to perturbations. It is suggested that information during adaptation of motor responses is processed at both spinal and supraspinal levels and may involve hardwired reflexive actions. Dietz and colleagues (1985) investigated the lower extremities and indicated supraspinal control of the lower limb's recovery responses in stance, i.e. the vestibular system seems to play a role in control via a polysynaptic spinal reflex pathway. Later, Bolton (2015) summarised cortical processes that contribute to stability recovery responses, along with cortico-basal ganglia and cortico-cerebellar loops (also Jacobs & Horak, 2007). Accordingly, several studies have examined changes at brain level that may explain the establishment or improvement of motor control skills. For instance, studies using transcranial magnetic stimulation on the cerebellum or primary motor cortex indicated that the motor cortex is involved in motor learning (Hardwick et al., 2013) and retention (Galea et al., 2011). In contrast, the cerebellum seems especially important for adaptation of newly learned locomotor patterns (Jayaram et al., 2011). The latter study incorporated a perturbation-practice protocol with participants adopting a new locomotor pattern in response to perturbed treadmill gait. This research revealed that locomotor adaption in humans is strongly correlated with depression of cerebellar excitability, specifically related to a reduced inhibition of Purkinje cells that are responsible for motor control.

However, according to the findings of Jakob & Horak (2007), suggesting that cerebral cortex involvement is proportional to the increase in latency of stability recovery response, one may argue that unexpected perturbations require even faster motor control mechanisms. In this regard, both animal and infant studies pointed to spinal locomotor circuits, previously modified by experiences required for adaptation, are also involved when responding to perturbations (Lam et al., 2003; Zhong et al., 2012; Martinez et al., 2012). Further, it is postulated that such adapted spinal locomotor circuitry responds to continuing changes in environment in an automatic manner, motor responses from milliseconds to minutes being partly or quite independent of supraspinal areas (Zhong et al., 2012). With the optimisation of responses for coping with perturbations in the longer term (retention or transfer) involving spinal and cortical levels (Jakobs & Horak 2007; Bolton 2015; Zhong et al., 2012), it seems reasonable to assume that practising motor recovery responses to perturbations could manifest neurophysiological adaptations that may be retained for recurring or even novel challenges to stability.

1.6. Transfer of stability recovery responses

In fact, repetitive execution of stability recovery responses has been shown to result in more effective coping with perturbations under altered conditions (transfer), e.g. from practising with treadmill-based slip-like perturbations to unpractised overground slips, or from simulated slips on a moveable platform to a non-practised slip on an oily surface (Bhatt & Pai, 2009; Grabiner et al., 2012; Lee et al., 2018; Parijat & Lockhart, 2012; Wang et al., 2011; Yang et al., 2013). Thus the neuromotor system seems capable to transfer fall-resisting skills given a common ground of contextual sensory feedback between two perturbations in different environments (Bhatt & Pai, 2008; Bhatt et al., 2013; Patel & Bhatt, 2015). This strengthens the potential for laboratory-based perturbation practice paradigms to interfere productively with falls that occur in daily life. Nevertheless, if the context of perturbation differs, the system needs to adjust its adapted motor response to unpractised task constraints to achieve positive transfer. Such might be readily achieved if the types of perturbations elicit some degree of shared stability responses for recovery. The objective of a previous study was to test transfer of adapted stability control from repeated treadmill-based gait perturbations to a lean-and-release task in healthy young, middle-aged, and older adults (König et al., 2019b). Transfer between tasks was assumed on the basis that they are characterised not only by a similar consequence of the perturbation itself (anterior displacement of the centre of mass) but also by similarities in subsequent stability recovery response (anterior increase in base of support by rapid stepping actions). However, despite revealing adaptive changes in stability control from repeated gait perturbations for all three age groups (young, middle-aged, and older adults), there was no enhanced stability performance during the lean-and-release transfer task after treadmill-based perturbation practice, with comparison made to age-matched controls who did not undergo prior practice (König et al., 2019b). These results indicate that even slight differences between perturbations could mean that desired motor adaptations would be deteriorated, if not redundant. Thus, exploring essential characteristics of perturbation-induced motor responses is crucial in promoting adaptation and transfer of fall-resisting skills (Harper et al., 2021; **objective of Study 2**).

Previous studies of our group (Epro et al., 2018b; König et al., 2019b; McCrum et al., 2018; Süptitz et al., 2013) incorporated a perturbation paradigm consisting of a single session in which participants faced eight separate treadmill-based gait trip-like perturbation trials. These few trials were sufficient to elicit remarkable adaptation and retention effects in trip-resistance in adults young and old. However, there is evidence that the design of perturbation practice paradigms is an essential factor affecting the transfer of fall-resisting skills. It has been shown recently that sufficient repetition of practised perturbations is required for both facilitation of long-term adaptive changes and enhancement of transfer in fall-resisting skills (König et al., 2019a; Lee et al., 2018). However, not only the number but also the magnitude of perturbations that induce motor error could be a crucial factor (Karamanidis et al.

2020). In the context of slip-perturbation paradigms it has already been shown that transfer of adapted stability control requires exposure to practised perturbation of appropriately high magnitude (Liu et al., 2016) – yet even higher than magnitudes which were shown to elicit adaptations during practice (Pai et al., 2014). In addition, there is evidence that adaptations in controlling slip-perturbations can be transferred to perturbations of even higher magnitudes (Patel & Bhatt, 2015). This suggests that increased perturbation magnitude during practice could lead to greater adaptations (Jayaram et al., 2011) and at the same time enhance transfer. Note, however, that those studies focused only on slip-like perturbations and incorporated investigations of transfer for which the contextual information of motor error was the same (Patel & Bhatt, 2015) or slightly altered (Liu et al., 2016). Furthermore, it is worth mentioning that results from modelling and experimental studies indicate a nonlinear relationship between the sizes of error feedback and of adaptation (Wei & Kording, 2009), suggesting that larger motor errors would not necessarily mean greater adaptation or greater transfer of motor skill adaptations. The effect of perturbation magnitude for short-term treadmill-based gait-trip practice on the transfer of adapted recovery response to different task constraints remains unknown (**objective of Study 3**).

It is crucial to understand factors that enable the neuromotor system to adapt and transfer stability control between different postural perturbations and task constrains, and hence to shed new light into the generalisation of fall-resisting skills in daily life (Harper et al., 2021; Karamanidis et al., 2020). The outcomes of this thesis may inform researchers and clinicians as to how to integrate and apply task-specific perturbation paradigms to both the assessment and practice of fall resilience.

2. Aims

This thesis examined control characteristics of spatiotemporal stability in adults across the lifespan and for different perturbation paradigms, all requiring reactive stepping responses in the anterior direction to cope with stability loss. The overall aim of this work was to investigate factors that elicit or limit adaptation and transfer of stability recovery responses. A **first study** served to examine two commonly performed clinical protocols for fall risk evaluation (forward lean-and-release task): whether these assessments of stability recovery responses are reliable within and between days and in adults aged from young to old. A **second study** examined to what extent non-practised recovery responses for a lean-and-release task and for a different perturbation paradigm (treadmill-based trip-like perturbation) are associated and predictable one to the other, similarly in adults aged from young to old. Based on existing literature it was hypothesised that recovery performance in one task had limited predictive power for performance of the alternate task. A subsequent **third study** incorporated the practice of treadmill-based gait perturbation. It elaborated on the transfer of adapted recovery responses (using different perturbation magnitudes) to non-practised stability loss for a lean-and-release task and an overground trip perturbation, in young to middle-aged adults. It was hypothesised that increased perturbation magnitude during practice would elicit or even enhance adaptation and transfer of stability recovery responses.

These three studies are presented separately in the following chapters as pre- or post-edited versions published in peer-reviewed journals, with the citation style amended to the format of this thesis following permission granted by the respective journal.

3. First study | Stability recovery performance in adults over a wide age range: A multicentre reliability analysis using different lean-and-release test protocols

Journal of Biomechanics (2021, v. 125, p. 110584; DOI: 10.1016/j.jbiomech.2021.110584).

3.1. Abstract

The ability to effectively increase the base of support is crucial to prevent from falling due to stability disturbances and has been commonly assessed using the forward-directed lean-and-release test. With this multicentre study we examined whether the assessment of stability recovery performance using two different forward lean-and-release test protocols is reliable in adults over a wide age range. Ninety-seven healthy adults (age from 21 to 80 years) were randomly assigned to one out of two lean angle protocols: gradual increase to maximal forward-lean angle (maximal lean angle; n = 43; seven participants were excluded due to marker artefacts) or predefined lean angle (single lean angle; n = 26; 21 participants needed to be excluded due to multiple stepping after release or marker artefacts). Both protocols were repeated after 0.5h and 48h to investigate intra- and inter-session reliability. Stability recovery performance was examined using the margin of stability at release (MoS_{RL}) and touchdown (MoS_{TD}) and increase in base of support (BoS_{TD}). Intraclass correlation coefficients (confidence intervals at 95%) for the maximal lean angle and for the single lean angle were respectively 0.93 (0.89-0.96) and 0.94 (0.89-0.97) in MoS_{RL}, 0.85 (0.77-0.91) and 0.67 (0.48-0.82) in MoS_{TD} and 0.88 (0.81-0.93) and 0.80 (0.66-0.90) in BoS_{TD}, with equivalence being revealed for each parameter between all three measurements ($p < 0.01$). We concluded that the assessment of stability recovery performance parameters in adults over a wide age range with the means of the forward lean-and-release test is reliable, independent of the used lean angle protocol.

3.2. Introduction

Falls are often caused by stability disturbances and remain a global health issue that majorly affects older but also middle-aged adults and often lead to severe health conditions, or even death (Burns & Kakara, 2018; Peeters et al., 2018; Terroso et al., 2014; Stenhagen et al., 2014). It is therefore important using reliable assessments for the recovery performance after stability disturbances to identify individual deficiencies or to classify the effectiveness of acute or long-term interventions on stability recovery performance.

The increase in base of support (BoS), i.e., to control the centre of mass (CoM) within the BoS, is one of the main motor responses to recover stability after disturbances (Hof, 2007). Stability recovery performance can be determined using the margin of stability concept (MoS; Hof et al., 2005) that provides information about the position of the CoM considering its velocity (extrapolated CoM; X_{CoM}) in relation to the boundaries of the BoS, where the X_{CoM} being outside the BoS represents an unstable state of the body, and vice versa. To identify individual deficiencies in stability recovery performance after sudden stability loss in the anterior direction, the *lean-and-release* test has often been applied on adults over a wide age range (Arampatzis et al., 2008; Carty et al., 2012; 2015; Karamanidis and Arampatzis, 2007; Karamanidis et al., 2008). These studies revealed clear deficits in stability recovery performance along with the inability to recover stability with a single step due to an insufficient (slow and low) increase in BoS with ageing. This age-related decline could even be associated to future falls in community-dwelling older adults (Carty et al., 2015; Okubo et al., 2017). Furthermore, the test has been applied to monitor acute effects of muscle-fatigue (Mademli et al., 2008; Walsh et al., 2011) or training interventions (Aragão et al., 2011; Arampatzis et al., 2011; Bohm et al., 2020) on stability recovery performance as well as to examine the inter-task transfer of acquired fall-resisting skills from gait trip-like perturbation training (König et al., 2019b).

Lean-and-release test protocols often differ depending on the study design. Several studies investigated the maximal lean angle from which one can recover with a single step (Aragão et al., 2011; Arampatzis et al., 2011; Bohm et al., 2020; Hamed et al., 2018). Repeated exposures to sudden stability loss in the anterior direction however seem to lead to immediate improvements in recovery stepping performances (Carty et al., 2012; Ringhof et al., 2019). Even small differences in stepping responses possibly evoked by task repetition could affect the ability to recover stability with a single step (Carty et al., 2012) i.e., the criterion to determine the maximal lean angle. Thus, it remains unclear yet whether such protocol is reliable. An alternative protocol is a sudden anterior stability loss from one or more predefined lean angles usually following a few practice trials (Carty et al., 2012; 2015; Karamanidis & Arampatzis, 2007; Karamanidis et al., 2008; Mademli et al., 2008). A recent study conducting consecutive exposures to anterior stability loss on young adults from a single lean angle indicated to an appropriate consistency and no day-to-day differences in the increase in BoS and in the MoS measured at 500ms after touchdown (Ringhof et al., 2019). However, the authors revealed less reliability for the assessment of task demand (lean angle) and MoS at the instant of touchdown. When furthermore considering several trial repetitions (intra-session) and the focus on one age group only (young adults), yet it remains unclear whether the assessment of stability recovery performance using only one exposure to sudden stability loss from a predefined lean angle is reliable across adults over a wide age range.

With this multicentre study we asked whether the assessment of main parameters used to determine stability recovery performance i.e., MoS at release, MoS and BoS at touchdown, conducting the lean-and-release test is reliable in adults over a wide age range (21 to 80 years; *n* = 97). We separately investigated two lean-and-release test protocols i.e., determination of the maximal lean angle from which a participant is still able to recover stability with a single forward step (maximal lean angle protocol) and a single exposure to stability loss from a predefined lean angle (single lean angle protocol).

3.3. Methods

3.3.1. Participants and experimental design

The study took place at three laboratories (Humboldt-Universität zu Berlin, German Sport University Cologne, and Robert-Bosch-Hospital in Stuttgart). A total of 97 adults ranging from 21 to 80 years of age were investigated. They were healthy and moderately physical active (e.g., regular weekly exercise). Exclusion criteria were any neurological or musculoskeletal injuries or impairments of the lower limbs limiting movement. After providing written informed consent, participants were randomly assigned to either a maximal lean angle protocol, or a single lean angle protocol (maximal or single for Berlin: n = 18 or 15; Cologne: n = 15 or 17; Stuttgart: n = 17 or 15). Both protocols were repeated once within a single session (Baseline and Post 0.5 h) and after two days (Post 48 h) to determine intra- and inter-session reliability. At all measurement timepoints, participants wore the same pair of their own non-slippery sports/leisure shoes. The study was approved by the respective local ethical committees (approval numbers for Cologne: 141/2017; Berlin: EA/082/15; Stuttgart: 266/2016MP2) and met all requirements for human experimentation in accordance with the Declaration of Helsinki (World Medical Association, 2013).

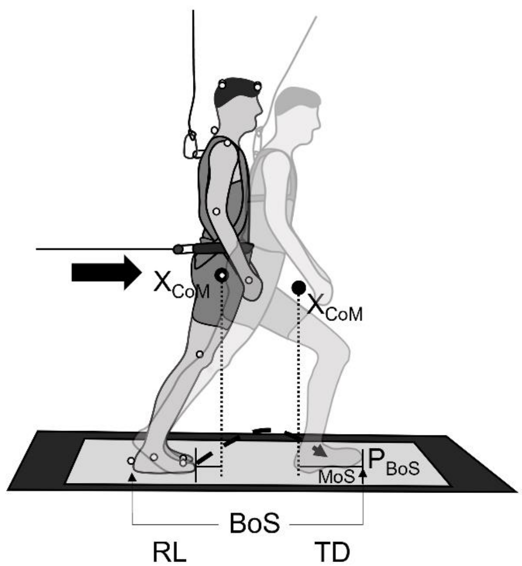

Figure 2: Illustration of the lean-and-release test. At the initial position (release, RL; dark-grey figure) the extrapolated centre of mass (X_{CoM}) with velocity equal to zero lies outside the anterior boundary of the base of support (P_{BoS}) and lies inside P_{BoS} following a successful recovery stepping phase at touchdown (TD; light-grey figure) after release. The margin of stability (MoS) is defined as the difference between the P_{BoS} and X_{CoM}, with positive versus negative values for the MoS equal to the X_{CoM} being located inside versus outside the P_{BoS}. Note that the illustrated supporting cable for the static-inclined positioning attached on the bellybutton-/pelvis-level was used in Stuttgart and Cologne, respectively, whereas it was attached on chest-level in Berlin.

3.3.2. Determination of the maximal forward-lean angle

Participants were always protected by a safety harness connected to an overhead track, allowing for full range of motion in anterior and lateral directions while preventing contact of the body with the ground (except for the feet). While standing on a force plate mounted in front of a second one (1080 Hz, 60 x 90 cm, Kistler, Winterthur, Switzerland or 1000 Hz, 40 x 60 cm, AMTI, MA, USA: depending on the laboratory) with their feet in parallel at hip-width and flat on the ground, participants were set in a forward-inclined position via an inextensible horizontal cable attached to a belt around the participant's pelvis (Karamanidis et al., 2008; Figure 2) and at the other end either to a custom-built pneumatic-driven brake-and-release system in Cologne (Karamanidis & Arampatzis, 2007), to a wall-mounted rail incorporating a snap-shackle-release system in Stuttgart, or to a wall-mounted electromagnet in Berlin (Hsiao & Robinovitch, 1999). The level of cable attachment differed between laboratories i.e., chest-level in Berlin versus level of pelvis/umbilicus in Cologne and Stuttgart. The initial lean angle (23 ± 2 % of body mass for participants ≤ 36 years, and 10 ± 2 % for ≥ 43 years accounting for the task demand in relation to the participants' age; Karamanidis & Arampatzis, 2007; Madigan, 2006) was controlled via a load cell (depending on the laboratory either custom-made 0-1 kN, or Megatron 0-5 kN; MEGATRON Elektronik GmbH & Co. KG, Munich, Germany) incorporated into the horizontal cable. Without any warning, the cable was suddenly released, randomly between 10 to 30 seconds. The lean angle was increased gradually by 3% if the participants were able to recover stability with a single step as instructed priorly (Karamanidis & Arampatzis, 2007). If the participants needed more than one step or a safety harness support (> 20 % of body mass determined by a second load cell incorporated into the harness suspension cable, i.e., multiple stepping; Karamanidis et al., 2008; Cyr & Smeesters, 2009), this trial was repeated. The measurement was terminated if the participants needed more than one step to recover stability in two consecutive trials. The last lean angle linked to a successful single step recovery was defined as the maximal lean angle. Please note that there were no prior practice trials performed for all measurement time points.

3.3.3. *Exposure to stability loss from a predefined forward-lean angle*

Safety assumptions, measuring equipment, procedure for the initial placement of the participants and task instructions matched the maximal lean angle protocol (see *Determination of the maximal lean angle* and Figure 2). Participants were released only from a single predefined lean angle corresponding to 23 ± 2 % of body mass. The forward-lean angle was chosen according to our previous results showing older adults still being able to recover stability from lean angles of approximately 20 % of body mass (Karamanidis et al., 2008). Only at Baseline but not for Post 0.5 h and Post 48 h, all participants performed three prior practice trials at 20 ± 2 % of body mass to familiarise with the task.

3.3.4. *Data collection and processing*

To quantify stability recovery performance for the two protocols, reflective markers were tracked via an optical motion capture system using ten infrared cameras (120 or 250 Hz, depending on the laboratory; Nexus; Vicon Motion Systems, Oxford, UK). The markers defined the foot, shank, thigh, trunk, upper and lower arm, hand, and head (Bierbaum et al., 2013; Figure 2; see *Appendix* for a detailed description). Two events were identified for both test protocols: (a) release of the supporting cable determined by a 50 % reduction in the leaning force signal provided by the incorporated load cell via a synchronised analogue TTL signal, and (b) foot touchdown of the recovery step determined via the vertical ground reaction force of the second force plate (threshold ≥ 5 N). The anterior MoS was determined at cable release (MoS_{RL}) and foot touchdown (MoS_{TD}) of the recovery step, calculated in accordance with Hof and colleagues (2005) as the differences between the extrapolated CoM (X_{CoM}) in the anterior direction and the anterior boundary of the BoS (P_{BoS}; see Figure 2). Segment masses and CoM locations were calculated based on the data reported by Dempster et al. (1959) and the position of the whole body's CoM in the 3D space was calculated according to Winter (1979), using a custom-made MATLAB script (2020b, MathWorks®, Natick, MA, USA). The BoS at touchdown (BoS_{TD}) i.e., the distance between the anterior and posterior boundaries of the base of support, was determined using the vertical projection of a heel marker of the trailing foot and the tip of the shoe of the recovery foot (Figure 2), considering the distance of a metatarsal marker to the anterior boundary of the shoe (measured during preparation).

3.3.5. Statistics

For both protocols, three measurement trials (Baseline, Post 0.5 h and Post 48 h respectively) were included. Normality and variance homogeneity of anthropometrics (body height, body mass, body mass index) and the analysed stability control parameters (MoS_{RL}, MoS_{TD}, BoS_{TD}) were checked using the Shapiro-Wilk test and Mauchly's sphericity test ($p > 0.05$). Body heights, body masses and body mass indexes of all participants separated by protocols and centres were statistically compared using separate one-way ANOVAs or Kruskal-Wallis tests, with Bonferroni-adjusted or Mann-Whitney-U *post hoc* tests performed in the presence of significant main effects. Potential differences in all analysed stability control parameters between repeated measurements were examined using separate (for both the maximal and single lean angle protocol) one-way repeated measures analyses of variance (ANOVA) with trials as within-subject factor. In the presence of significant main effects, Bonferroni-adjusted *post hoc* tests for pairwise comparison were performed to locate potential differences. Two-way mixed model intraclass correlation coefficients (ICC, absolute agreement, and single measures) over all trials were calculated, with confidence intervals at 95 % (Koo & Li, 2016). ICC were defined as "poor" (< 0.50), "moderate" (0.50-0.75), "good" (0.75-0.90) and "excellent" (> 0.90) to interpret reliability (Portney, 2020). Root mean square errors (RMSE) were computed to determine the average dispersion of the observed trial from the previous one and reported as a range between all trials. To argue for the absence of an effect being large enough to state a significant discrepancy between trials, two one-sided tests (TOST) for equivalence were performed. According to Lakens (2013, 2017), the difference between dependent trial means, and respective confidence intervals at 95 %, were tested with a standardised lower (Δ_L) and upper (Δ_U) bound of equivalence based on Cohen's *dz* that was calculated from current raw data. The level of significance was set at $\alpha = 0.05$. All statistical and non-statistical analyses as well as descriptive computations were performed using SPSS Statistics (v26, IBM, Chicago, IL, USA) and MATLAB (2020b, MathWorks®, Natick, MA, USA).

3.4. Results

3.4.1. Maximal lean angle

Due to significant marker artefacts during the measurements, data of seven participants were excluded hence 43 participants (19 females; 29 to 77 years) were considered for the statistical analyses. Body height, body mass and body mass index did not significantly differ between the centres (Berlin: 172.9 ± 9.3 cm, 75.0 ± 11.6 kg and 25.0 ± 2.9 kg/m^2; Cologne: 172.3 ± 13.1 cm, 75.1 ± 15.8 kg and 25.0 ± 2.7 kg/m^2; Stuttgart: 172.9 ± 7.7 cm, 77.3 ± 9.4 kg and 25.9 ± 2.5 kg/m^2). Regarding the MoS_{RL}, an excellent ICC of 0.93 (CI_{95} [0.89-0.96]; Table 1) was computed over all trials, with RMSE ranging between 3.3 and 5.1 cm. Although the TOST at 95 % confidence revealed an effect statistically different from zero for the MoS_{RL} between Baseline and Post 48 h (Figure 3), there was no significant difference when using a one-way ANOVA, neither between Baseline and Post 0.5 h or Post 0.5 h and Post 48 h (Figure 4). Moreover, TOST showed statistically significant ($p < 0.01$) equivalence between all pairs of trials. Furthermore, there were no significant differences in the MoS_{TD} between all trials, showing good ICC of 0.85 (CI_{95} [0.77-0.91]), RMSE ranging between 4.9 and 7.1 cm, and significant equivalence ($p < 0.001$). Good reliability (ICC of 0.88; CI_{95} [0.81-0.93]) was also revealed for the BoS_{TD} whilst an absence of significant differences, with RMSE ranging between 7.4 and 8.4 cm, and statistical equivalence ($p < 0.001$) between all trials.

Table 1: Intraclass correlation (ICC [3,1]) coefficients with 95 % confidence intervals (CI_{95}) and range of root mean square errors (RMSE; in cm) for margin of stability at release (MoS_{RL}) and at touchdown (MoS_{TD}) as well as the base of support at touchdown (BoS_{TD}) between the measurements (Baseline, Post 0.5 h and Post 48 h) for the maximal and single lean angle protocols (*n* = 43 and *n* = 26 respectively).

		Maximal lean angle (*n* = 43)		**Single lean angle** (*n* = 26)	
	Measurements	**ICC [CI_{95}]**	**RMSE (cm)**	**ICC [CI_{95}]**	**RMSE (cm)**
MoS_{RL}	Baseline vs. Post 0.5h	0.96 [0.93-0.98]	3.3	0.95 [0.90-0.98]	1.8
	Post 0.5h vs. Post 48h	0.93 [0.88-0.96]	4.0	0.95 [0.90-0.98]	1.8
	Baseline vs. Post 48h	0.90 [0.81-0.94]	5.1	0.92 [0.84-0.97]	2.3
MoS_{TD}	Baseline vs. Post 0.5h	0.85 [0.74-0.92]	6.0	0.70 [0.43-0.85]	5.1
	Post 0.5h vs. Post 48h	0.90 [0.83-0.95]	4.9	0.66 [0.38-0.83]	5.1
	Baseline vs. Post 48h	0.80 [0.66-0.89]	7.1	0.66 [0.39-0.83]	5.8
BoS_{TD}	Baseline vs. Post 0.5h	0.91 [0.83-0.95]	7.4	0.82 [0.63-0.91]	6.7
	Post 0.5h vs. Post 48h	0.88 [0.78-0.93]	7.8	0.80 [0.61-0.91]	7.3
	Baseline vs. Post 48h	0.85 [0.74-0.92]	8.4	0.78 [0.57-0.89]	8.2

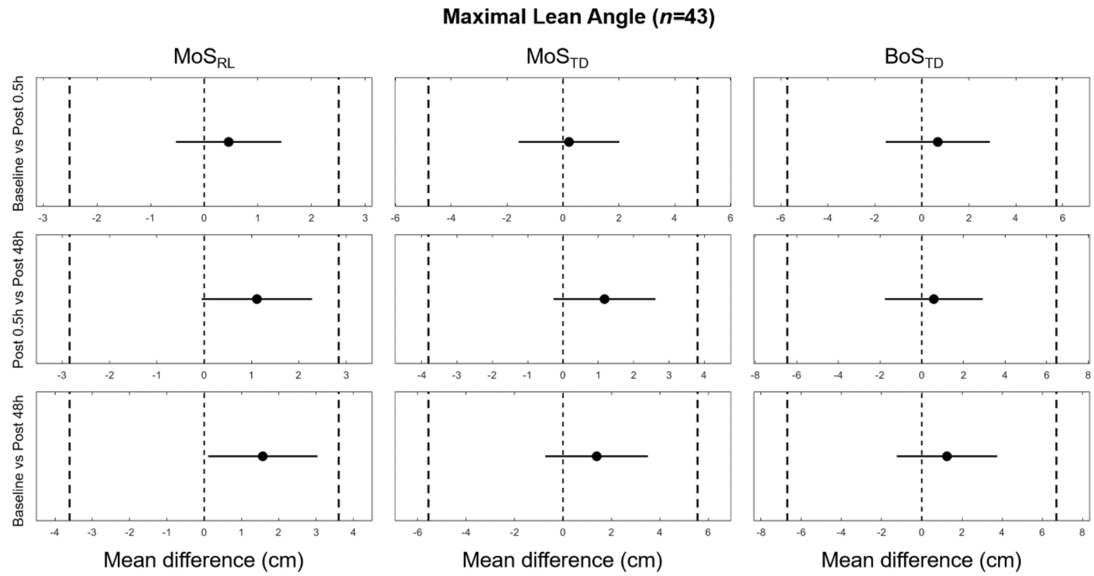

Figure 3: Two one-sided equivalence in margin of stability at release (MoS_{RL}) and at touchdown (MoS_{TD}) as well as the base of support at touchdown (BoS_{TD}) between the trials (Baseline, Post 0.5 h and Post 48 h) for the maximal lean angle protocol (*n* = 43). The thin vertical dashed line refers to the zero-effect size, with lower as well as upper equivalence bounds are illustrated by the thick vertical dashed lines, the black dot equals the mean differences between the trials, and 95 % confidence intervals are represented with the black horizontal lines.

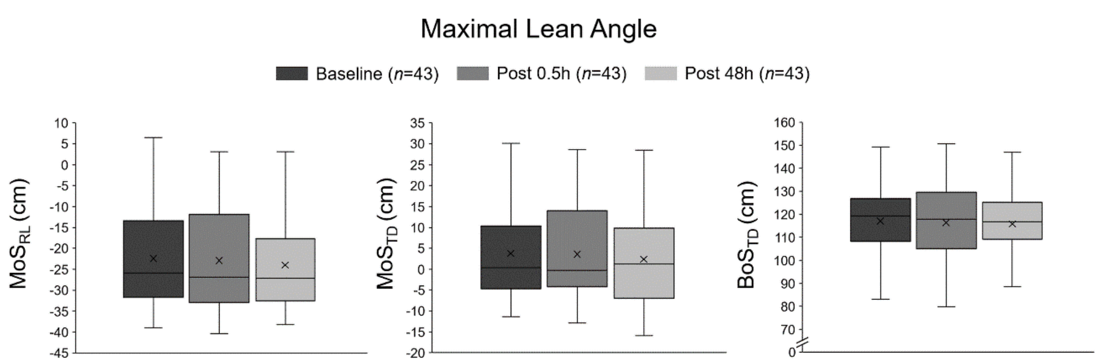

Figure 4: Margin of stability at release (MoS_{RL}) and at touchdown (MoS_{TD}) as well as the base of support at touchdown (BoS_{TD}) between the trials (Baseline, Post 0.5 h and Post 48 h) for the maximal lean angle protocol (*n* = 43). Results are presented as boxplots (mean, median and interquartile range between 25th and 75th percentile along with minimum and maximum values).

3.4.2. Single lean angle

Data of ten participants could not be analysed appropriately and were excluded from the statistical analysis. Only body mass but neither body height nor body mass index of the included 41 participants (15 females; 22 to 70 years; Berlin: 172.2 ± 6.7 cm, 68.7 ± 9.2 kg and 23.1 ± 1.9 kg/m^2; Cologne: 175.7 ± 8.1 cm, 79.8 ± 11.8 kg and 25.8 ± 2.6 kg/m^2; Stuttgart: 172.4 ± 7.0 cm, 80.9 ± 19.2 kg and 27.1 ± 5.4 kg/m^2) significantly differed between Berlin and both Cologne (p = 0.020) as well as Stuttgart (p = 0.020). Please note that 13 out of 15 adults (61-70 years) required multiple steps during the single lean angle protocol at all measurements. Thus, those data were non-statistically observed and excluded from further processing as we investigated continuous variables which are affected differently between single and multiple stepping responses. When considering the data of adults (21 to 60 years; n = 26) who were able to successfully recover stability with a single step, there were no significant differences in MoS_{RL} between all trials (Figure 5), with an excellent ICC of 0.94 (CI_{95} [0.89-0.97]; Table 1) and RMSE ranging between 1.8 and 2.3 cm. The MoS_{TD} neither differed between all trials, showing a moderate ICC of 0.67 (CI_{95} [0.48-0.82]) and RMSE ranging between 5.1 and 6.8 cm. The BoS_{TD} showed good reliability (ICC of 0.80; CI_{95} [0.66-0.90]), with no differences between all trials and RMSE ranging between 6.7 and 8.2 cm. For MoS_{RL}, MoS_{TD} and BoS_{TD}, TOST revealed significant (p < 0.01) equivalence between all trials (Figure 6). We furthermore observed 13 out of 15 older adults who were not able to recover stability with a single step from the single pre-defined lean angle (single lean angle protocol) in all trials revealing an overall consistency of recovery stepping behaviour i.e., multiple stepping.

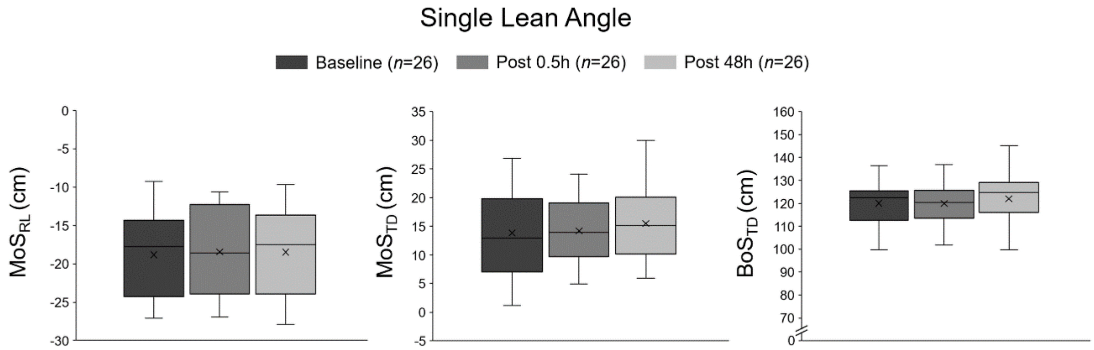

Figure 5: Margin of stability at release (MoS_{RL}) and at touchdown (MoS_{TD}) as well as the base of support at touchdown (BoS_{TD}) between the trials (Baseline, Post 0.5 h and Post 48 h) for the single lean angle protocol (*n* = 26). Results are presented as boxplots (mean, median and interquartile range between 25th and 75th percentile along with minimum and maximum values).

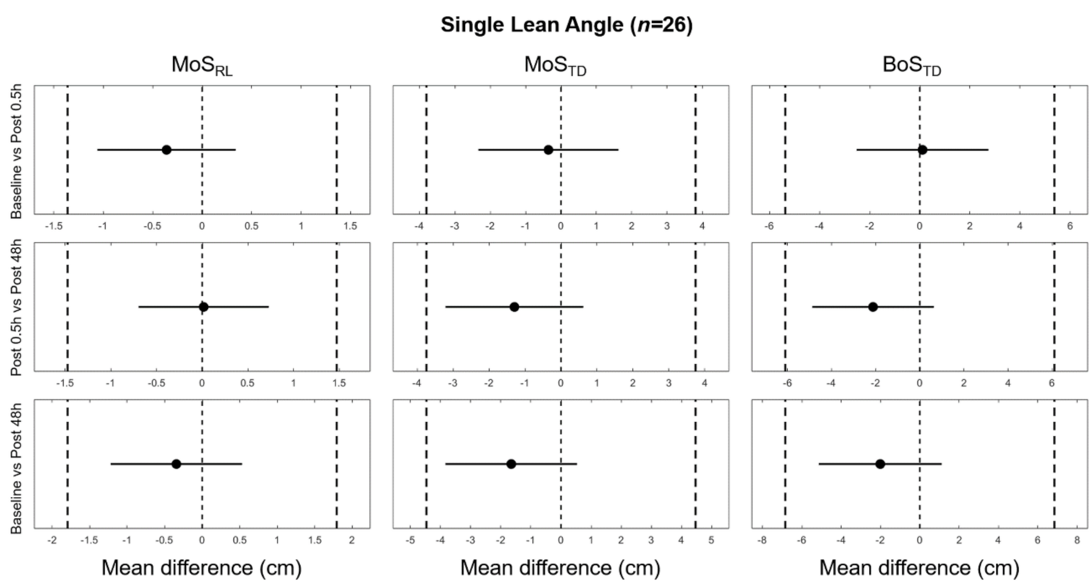

Figure 6: Two one-sided equivalence in margin of stability at release (MoS_{RL}) and at touchdown (MoS_{TD}) as well as the base of support at touchdown (BoS_{TD}) between the trials (Baseline, Post 0.5 h and Post 48 h) for the single lean angle protocol (*n* = 26). The thin vertical dashed line refers to the zero-effect size, with lower as well as upper equivalence bounds are illustrated by the thick vertical dashed lines, the black dot equals the mean differences between the trials, and 95 % confidence intervals are represented with the black horizontal lines.

3.5. Discussion

In this multicentre study we examined whether the assessment of stability recovery performance parameters in adults over a wide age range is reliable if conducting a maximal forward-lean angle approach as well as a single exposure to sudden anterior stability loss from a predefined lean angle. For both lean angle protocols, we revealed statistically appropriate consistency and equivalence with the absence of any relevant differences in all analysed parameters. The results indicate that the lean-and-release test is a reliable assessment to potentially identify individual deficiencies or to classify the effectiveness of acute or long-term interventions on stability recovery performance.

3.5.1. Maximal lean angle

With the MoS_{RL} considered as the main criterion, the maximal lean angle protocol has often been used as a standardised assessment method to identify age-related deficiencies or intervention effects on stability performance. Previous studies reported an improved stability performance in older adults following several months of stability and/or strength training (more negative MoS_{RL} ranging on average from 2.8 to 6.6 cm at post compared to pre intervention) i.e., they were able to successfully recover stability with a single step from a more inclined and unstable position (Arampatzis et al., 2011; Bohm et al., 2020; Hamed et al., 2018). The current study revealed excellent reliability (ICC of 0.93) for the MoS_{RL} in adults over a wide age range (29 to 77 years), with lower differences (1.1 cm on average) compared to the intervention studies, and overall significant equivalence between all trials. This strengthens the outcomes of previous findings demonstrating differences, indicating that those were caused by the conducted intervention rather than biased by task adaptation or drawbacks related to the reliability of measurements.

We terminated a measurement after two consecutive failures of single step recovery occurred. Respectively, some participants might have learned due to task repetitions (Carty et al., 2012; Ringhof et al., 2019) with the same lean angle. However, we demonstrated excellent reliability in the MoS_{RL} using the same procedure for all measurements indicating that the identified individual maximal lean angle can be postulated as an ultimate task demand to test recovery performance in a reliable manner. This was further supported by good consistency (ICC ranging from 0.85-0.88), non-significant differences and equivalence between all trials revealed for the stability performance at touchdown (BoS_{TD} and MoS_{TD}) i.e., the ability to recover stability from similar maximal lean angles at all trials always came along with similar step lengths and control of the CoM in relation to the BoS. Thus, we state that using the maximal lean angle protocol is a reliable assessment to determine the maximal capability of stability recovery performance in adults over a wide age range.

Due to low and unequal sample sizes of and between different age groups we did not consider an age-related contribution to the reliability results for the maximal lean angle protocol. However, when pooling all participants above 43 years (n = 22) according to the chosen single lean angle protocol that accounted for a lower initial task demand for such age cohorts, we found significant ($p < 0.001$) and good ICC for all analysed parameters (range between 0.86 and 0.90). Moreover, for this sub-pool of adults above 43 years we revealed an average RMSE over trials of 6.3 cm in the MoS_{TD}, that was similar to the average error between trials of all participants under 36 years (6.7 cm; n = 21). Thus, we believe that pooling all participants for the analyses did not cause a bias related to the current reliability and that age had no relevant effect on the main outcomes of the current study.

3.5.2. Single lean angle

Our results revealed excellent reliability in the MoS_{RL} and overall equivalence, indicating that the lean angle was effectively controlled across all trials (mean and RMSE on average for all trials: -18.6 cm and 2.0 cm). It is important to note that the standard deviations of the MoS_{RL} were rather high (on average for all trials: 5.8 cm) assuming a higher inter-subject variability potentially caused due to heterogeneous body configurations (body height may influence MoS_{RL}). However, since body heights were homogenous between the participants, the high standard deviation can rather be explained by laboratory-related differences in the attachment level of the supporting cable and incorporated load cell respectively i.e., a more proximal attachment (level of umbilicus or higher) required the participant to lean more forward and led to lower values in the MoS_{RL}. Such a higher demand on stability recovery performance caused mostly all older adults (*n* = 13 out of 15; 61 to 70 years) failing to recover stability with a single step during the single lean angle protocol in all trials. Nevertheless, since all younger and middle-aged adults were able to recover with a single step, the demand seemed to be appropriate for these age populations and furthermore the assessment was highly reliable. In contrast to our findings, Ringhof and colleagues (2019) recently revealed poor between-session reliability for the demand on stability recovery performance (measured in degrees) following exposure to stability loss from a single predefined lean angle (15 % of body mass) in young adults. These results may be difficult to compare with the current outcomes as we used a different parameter for the task demand i.e., MoS. However, when considering that the MoS_{RL} is mainly used to interpret the demand on stability recovery performance, we confirmed this was reliably assessable among adults over a wide age range.

Since the single lean angle protocol requires a constant demand on stability recovery performance within each execution (i.e., a pre-defined MoS_{RL}), the MoS_{TD} and BoS_{TD} have commonly been used as the main criteria to determine stability performance. A recent study indeed showed an improved stability performance (MoS_{TD}) after a single trial repetition in younger and middle-aged adults, without any prior practice trials performed at Baseline (ΔMoS_{TD} on average for both age-groups: 3.8 cm, $p < 0.01$; RMSE: 7.8 cm; = 27; König et al., 2019b). In contrast to those findings the current study did not reveal significant differences between the means of all trials in the MoS_{TD}, particularly of Post 0.5 h versus Baseline (ΔMoS_{TD}: 0.4 cm, with a comparably lower RMSE of 5.1 cm on average respectively). This could be explained mainly by the constant BoS_{TD} between trials revealed in the current study. As our participants performed three practice trials prior to the Baseline measurement, we cannot confirm the absence of task adaptations which might have been occurred due to immediate task repetition during Baseline. But we proved the current protocol to be a reliable assessment approach without performing any further practice trials prior to Post 0.5 h and Post 48 h. Yet a control group is required essentially to exclude bias caused by rapid adaptation due to consecutive repetition of an unpractised stability task (König et al., 2019b).

The different levels of cable attachment i.e., chest versus umbilicus versus pelvis, led to different stability demands according to one standardised percentage of body mass. Thus, it might have caused a drawback to determine reliability of the single lean angle protocol and stability performance in older adults in a standardised manner. However, although their demand on stability (MoS_{RL} on average for all trials: -20 cm; *n* = 15) might have contributed to an inability to recover stability with a single step, multiple stepping was observed to be consistent during all trials for 13 out of 15 older adults, indicating to no functionally relevant learning due to task repetition and hence to a reliable assessment of stability performance that has previously been shown to predict future falls (Carty et al., 2015). Nevertheless, to overcome any influence of different cable-attachments we postulate considering the initial state of body configuration with the means of the MoS instead of relying solely on the percentage of body mass for the assessment of stability recovery performance using a single lean angle approach.

We concluded that the assessment of stability recovery performance parameters in adults over a wide age range using the forward lean-and-release test is reliable, independent of the used lean angle protocol. Our results further strengthen the use of an exposure to stability loss from a single predefined lean angle, as this protocol being less time-consuming and less demanding could especially be beneficial to test stability recovery performance in clinical settings.

3.6. Acknowledgments

This multicentre study was financially supported by the Bundesministerium für Bildung und Forschung (BMBF).

4. Second study | The ability to increase the base of support and recover stability is limited in its generalisation for different perturbation tasks

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4.1. Abstract

The assessment of stability recovery performance following perturbations contributes to the determination of fall resisting skills. This study investigated the association between stability recovery performances in two perturbation tasks (lean-and-release versus tripping). Healthy adults (12 young: 24 ± 3 years; 21 middle-aged: 53 ± 5 years; 11 old: 72 ± 5 years) were suddenly released from a forward-inclined position attempting to recover stability with a single step. In a second task, all participants experienced a mechanically induced trip during treadmill walking. To assess dynamic stability performance, the antero-posterior margin of stability (MoS), the base of support (BoS), and the rate of increase in BoS were determined at each foot touchdown (TD) for both tasks. Only weak to moderate correlations in dynamic stability performance parameters were found between the two tasks ($0.568 > r > 0.305$, $0.001 < p < 0.04$). A separation of participants according to the number of steps required to regain stability in the lean-and-release task revealed that multiple- (more than one step) compared to single-steppers showed a significantly lower MoS at TD ($p = 0.003$; $g = 1.151$), lower BoS at TD ($p = 0.019$; $g = 0.888$) and lower rate of increase in BoS until TD ($p = 0.002$; $g = 1.212$) after release. Despite these profound subgroup differences in the lean-and-release task, no differences between multiple- and single-steppers were observed in the stability recovery performance during tripping. The results provide evidence that the ability to effectively control dynamic stability following a sudden stability disturbance in adults across a wide age range is limited in its generalisation for different perturbation tasks.

4.2. Introduction

Daily-life locomotion is a challenging task. While walking on slippery or uneven paths, crossing over obstacles lying on the ground or managing to pass along narrow walkways, one faces countless situations that can disturb movement, requiring the neuromotor system to adjust its motor output for coping with external perturbations (e.g. a trip), control stability or avoid falls. Although falls are observed among adults of all ages, their incidence increases with aging contributing to the most prominent cause for injuries, hospitalisation or even death among the elderly population (Burns & Karkara, 2018; Hoskin, 1984; Terroso et al., 2014). Therefore, assessing and understanding stability recovery responses in adults of various ages is highly relevant to reduce or even avoid forward falls and related injuries at old age (Karamanidis et al., 2020; Lee et al., 2020; McCrum et al., 2019).

To maintain stability during walking, the central nervous system needs to ensure a continuous interaction between perceptual information and motor responses (Scott, 2004). Human locomotion requires the combination of multiple sensory information originating from somatosensory, vestibular, and visual systems, together with the coordination of numerous skeletal muscles. When experiencing an unexpected trip during locomotion, a change in the relation between the centre of mass (CoM) and the base of support (BoS) is observed, with the CoM moving closer to the edge of the BoS. This change leads to a significant decrease in the margin of stability (MoS) compared to unperturbed walking causing an unstable body configuration (König et al., 2019b; McCrum et al., 2016). Hence, in order to increase the MoS and efficiently counteract a forward fall, a relatively long and rapid anterior step is required (Karamanidis et al., 2020). Given that older as well as middle-aged compared to younger adults require on average more steps to regain a stable MoS following a sudden stability loss (Karamanidis & Arampatzis, 2007; Pai et al., 2010), large focus has been placed on developing testing paradigms to evaluate stability recovery responses following sudden stability loss. Various studies have investigated human stability recovery performance and the ability to increase effectively the BoS in the anterior direction following externally induced stability perturbations using an unexpected release from a forward inclined position, i.e. the lean-and-release task

(Bolton & Mansour, 2020; Carty et al., 2015; McCrum et al., 2019; König et al., 2019b; Thelen et al., 1997; Wojcik et al., 1999). Previous research has demonstrated that future fall risk in older populations can be predicted by the recovery stepping behaviour observed in such lean-and-release tasks (Carty et al., 2015; Mansfield & Maki, 2009). Süptitz and colleagues (2013) reported that following a sudden gait-trip perturbation, older in comparison to young adults show a decreased capacity to rapidly and effectively increase their BoS, indicating to a higher fall risk. This could explain why older adults often require multiple steps to regain their stability during trip-like perturbations.

The ability to increase effectively the anterior BoS is an essential skill to regain stability control in a lean-and-release task (Karamanidis et al., 2008) as well as during tripping (Epro et al., 2018a; McCrum et al., 2019). Besides, it has been reported that a significant increase in BoS of the recovery step following an anterior stability loss in both tasks can be observed when compared to unperturbed walking (König et al., 2019b). Although critical task parameters (e.g. muscle activity patterns, muscle-tendon-unit lengths and body dynamics) may differ possibly due to different body configurations, and the static or dynamic nature, both tasks involve perturbations being large enough to cause unstable body configurations which require similar stability recovery responses (i.e. increase in anterior BoS due to rapid stepping) crucial for safe locomotion and fall prevention in everyday life. Thus, one might suggest a link between the reactive stepping performances in these tasks. Regarding this, a recent study showed no inter-task transfer of fall-resisting skill adaptations from short-term practice of treadmill gait-perturbations to a lean-and-release task (König et al., 2019b), suggesting only a limited generalisation of improved fall-resisting skills. Nevertheless, the aforementioned study focused on the transfer of adaptations acquired during practised gait-perturbations on the treadmill to a lean-and-release task, rather than on an association of stability recovery performance between the two tasks. Up to date, literature is still lacking information regarding the association of the capability to regain stability effectively and rapidly between the lean-and-release task and tripping-task in adults of various age. This could be of great interest for clinical settings regarding the evaluation of dynamic stability performance in aging adults. Therefore, the present study aimed to examine the relationship between the stability recovery performance during lean-and-release task and a tripping-task on a treadmill among adults across a wide age range (*n* = 44; 24 to 72 years). In addition, it was investigated whether there are differences in treadmill tripping performances between single- and multiple-steppers observed in the lean-and-release task. It was hypothesised that stability recovery during a lean-and-release task is not a valid measure to appropriately predict tripping recovery performance.

4.3. Methods

4.3.1. Participants and experimental design

A total of 44 healthy adults of various ages (24-72 years) participated in this study. Participants were not eligible to perform the experiments if they were suffering from any movement limiting neurological or musculoskeletal impairments or diseases of the lower limbs. After an initial briefing, all participants provided their informed consent. In the first stability recovery task, all participants were unexpectedly released from a static forward-inclined position (lean-and-release task). Following this, they were exposed to an unexpected trip-like perturbation while walking on a treadmill at a given speed. To ensure safety, participants were secured by a full-trunk safety harness attached to an overhead track allowing antero-posterior and medio-lateral movements but preventing any contact of the body with the ground (except for the feet). The present study was approved by the ethics committee of the German Sport University Cologne (ethical approval no. 141/2017) and was conform to all requirements for human experimentation in accordance with the Declaration of Helsinki.

4.3.2. Lean-and-release task

Participants' stability recovery performance was evaluated using a lean-and-release task, that has been described in previous studies (Karamanidis et al., 2008; König et al., 2019b). Briefly, the participants were standing on a force plate (1080 Hz, 60 x 90 cm: Kistler, Winterthur, Switzerland) with their feet in parallel and flat on the ground (Figure 7). They were gradually inclined in the forward direction and held by a custom-built pneumatic brake-and-release system via a horizontally running inextensible Teflon cable connected to a belt around the pelvis (Karamanidis & Arampatzis, 2007). The targeted inclination matched an angle corresponding to a value of 23 ± 2 % body mass and was controlled with the means of a load cell implemented in series with the supporting cable. The exact forward lean was chosen according to previous results of the reduced ability of older adults to regain stability within a single recovery step from cable loads of more than 23 % body mass (Karamanidis et al., 2008). Once any anticipatory movement was attenuated (i.e. antero-posterior and medio-lateral body mass shift corrections, checked real-time via cable load and ground reaction forces) the supporting cable was released without any further notice after an arbitrary period between 10-30 seconds. Prior to the measurement, participants were previously instructed to try regaining a stable stance with a single recovery step after being released using the limb of their choice (Madigan & Lloyd, 2005). To guarantee novelty of the task, no prior practice trials were performed. According to previous findings (Karamanidis & Arampatzis, 2007), stability recovery performance was categorised into two stepping behaviours, i.e. single stepping versus multiple stepping. Participants were defined as 'single-steppers' if they needed only one step to recover stability or if a follow-up step of the contralateral limb did not exceed the anterior displacement of the recovery limb's foot. Consequently, participants were defined as 'multiple-steppers' if they required any additional step of the recovery limb or needed a safety harness support, i.e. more than 20% of body mass observed via a second load cell integrated into the harness suspension cable (Cyr & Smeesters, 2009).

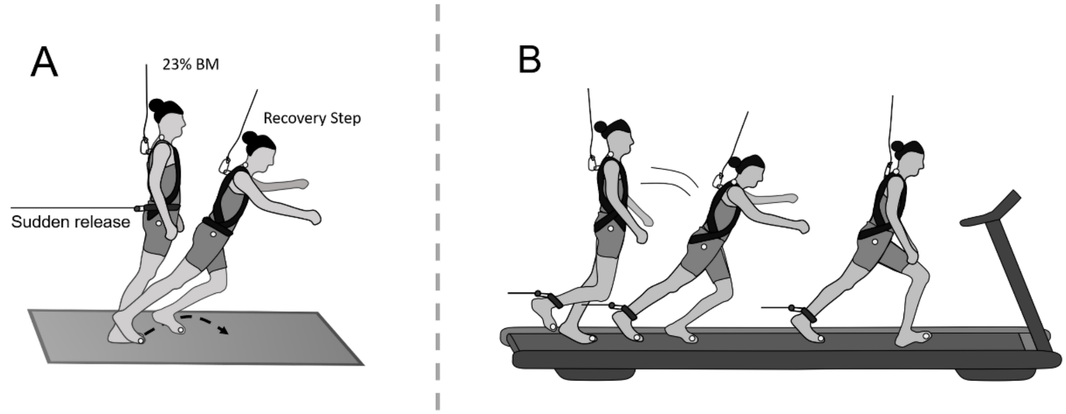

Figure 7: Schematic illustration of the experimental set-ups. **(A)** Lean-and-release-task. Participants were released once from a forward-inclined position. Lean angles were normalised to the participants' body mass (23 % of body mass) ensuring to standardise the level of stability loss. **(B)** Tripping-task during treadmill walking. Participants were exposed to a trip while walking on a treadmill. The trip was induced using a custom-built pneumatic driven cylinder system. In an event of a fall, an overhead safety harness prevented the participant's body (except the feet) from touching the treadmill belt. White circles represent the five retroreflective markers attached to anatomical landmarks used to evaluate the spatiotemporal stepping characteristics during both tasks.

4.3.3. Single exposure to a trip-like perturbation during treadmill walking

The tripping-task used in the current study has been conducted previously (Epro et al., 2018a; König et al., 2019b). The protocol started with the participants walking unperturbed on a treadmill (pulsar 4.0; h/p/cosmos, Nussdorf-Traunstein, Germany) at a standardised speed of 1.4 m/s for four minutes followed by a baseline measurement (25 stride cycles of walking). Subsequently, they were exposed to an unexpected trip-like perturbation induced using a custom-built pneumatic cylinder system and encouraged to continue walking afterwards (Figure 7). Throughout one entire swing phase, the perturbation (restraining pull) was applied using a strap attached to the right ankle connected via a Teflon cable to the perturbation device. Although participants received prior information about the task, they were not able to anticipate the onset and removal of the perturbation. All participants were invited to familiarise only with unperturbed treadmill walking 4-7 days prior to the measurement day. To guarantee novelty of the task, no exposures to treadmill perturbations were performed prior to the actual measurement.

4.3.4. Data collection and processing

To determine the CoM trajectories and dynamic stability control during the two tasks, a reduced kinematic model was used (Süptitz et al., 2013). Five retroreflective markers were attached to anatomical landmarks (seventh cervical vertebra, both greater trochanters and forefeet of the left and right legs, respectively; see *Appendix* for a detailed description) and tracked via a 10-camera optical motion capture system (120 Hz; Nexus 2.6.1; Vicon Motion Systems, Oxford, UK). Three-dimensional coordinates of the markers were smoothed using a fourth-order digital Butterworth filter with a cut-off frequency of 20 Hz (Epro et al., 2018a). Foot touchdown (TD) of the recovery step in the lean-and-release task was determined as the moment at which the vertical ground reaction force measured by a second force plate (1080 Hz, 60 x 90 cm; Kistler) exceeded a threshold value of 20 N. For the tripping task, TD was defined as the impact peak of an analogue signal acquired using 2-D accelerometers (± 50 g, 1080 Hz; model ADXL250; Analog Devices, Norwood, MA) positioned on the tibia of each leg (Süptitz et al., 2013). The antero-posterior margin of stability (MoS) was calculated as the difference between the anterior boundary of the base of support (BoS) and the extrapolated centre of mass (XCoM), which includes both the position and the velocity of the CoM. The MoS and BoS were assessed at each TD during unperturbed, perturbed, and the first six recovery steps following the perturbation (Epro et al., 2018a), as well as at TD of the first recovery step during the lean-and-release task (Karamanidis et al., 2008). The BoS was calculated as the distance between the toe markers of the trailing and stance limb at TD for both tasks. Furthermore, the rate of increase in BoS during the lean-and-release task was calculated as the ratio between the BoS at TD and the swing time until TD of the first recovery step.

4.3.5. Statistics

Normal distribution of all variables was confirmed by Lilliefors-corrected Kolmogorov-Smirnoff tests ($p > 0.05$). To examine the relationship between the lean-and-release task and the tripping task performance across participants, Pearson product-moment correlation coefficients were computed for the MoS, the BoS, and the rate of increase in BoS. Since younger adults are not representative of high fall risk, subgroup comparisons (single-steppers versus multiple-steppers) regarding dynamic stability during the lean-and-release task as well as during the tripping-task were performed including only middle-aged and older adults. Independent samples *t*-tests were used to examine differences between single-steppers and multiple-steppers in the MoS, the BoS, and the rate of increase in BoS for the lean-and-release task. Subgroup comparisons for the tripping-task were performed using separate two-way mixed-measures ANOVAs with factors subgroups (single- versus multiple-steppers) and events (perturbed and the following six recovery steps) for the MoS and the BoS. In case of significant main effects or interactions, Duncan's post-hoc corrections were applied. The level of significance was set at α=0.05 and effect sizes were calculated using Hedges' g and partial eta square (η_p^2). Effect sizes were considered small (η_p^2= 0.01; *r* = 0.1; g = 0.2), medium (η_p^2 = 0.06; *r* = 0.3; g = 0.5), or large (η_p^2 = 0.14; *r* = 0.5; g = 0.8). To identify age-related differences in the MoS, the BoS, and the rate of increase in BoS amongst the three age-groups (young, middle-aged, old) during the lean-and-release task, separate one-way ANOVAs were used. Separate two-way mixed-measures ANOVAs were used to detect age-related differences in the MoS, and the BoS during the tripping-task, with age-group (young, middle-aged, old) and events (perturbed and the following six recovery steps) as factors. Differences in age, body height and mass as well as physical activity between the three age groups were analysed using separate one-way ANOVAs. In cases of significant main effects or interactions, Duncan's post-hoc tests were applied. All statistical and non-statistical analyses were performed using Statistica software (Release 10.0; Statsoft Inc., Tulsa, OK, USA) and MATLAB (2020b, MathWorks®, Natick, MA, USA).

4.4. Results

4.4.1. Association of stability performance between tasks

There were statistically significant correlations between stability recovery performances (MoS and BoS at TD and rate of increase in BoS until TD) of the lean-and-release task and the tripping-task (MoS and BoS at TD of the first recovery step). Although significant, weak to moderate correlations were found between the MoS at TD during tripping and the MoS at TD of the lean-and-release task ($r_{44} = 0.568$, $p < 0.001$; Figure 8) as well as between the BoS at TD during tripping and the BoS at TD during the lean-and-release task ($r_{44} = 0.305$, $p = 0.044$; Figure 9). Similarly, there was a significant correlation between the BoS at TD of the lean-and-release task and its respective rate of increase in BoS until TD ($r_{44} = 0.600$, $p < 0.001$). Furthermore, a significant correlation was detected between the BoS at TD during tripping and both the MoS at TD ($r_{44} = 0.411$, $p = 0.006$; Figure 9) as well as the rate of increase in BoS at TD ($r_{44} = 0.357$, $p = 0.017$; Figure 9) of the lean-and-release task. No statistically significant correlations were found between the MoS at TD during tripping and both the BoS at TD, or the rate of increase in BoS until TD during the lean-and-release task (Figure 8).

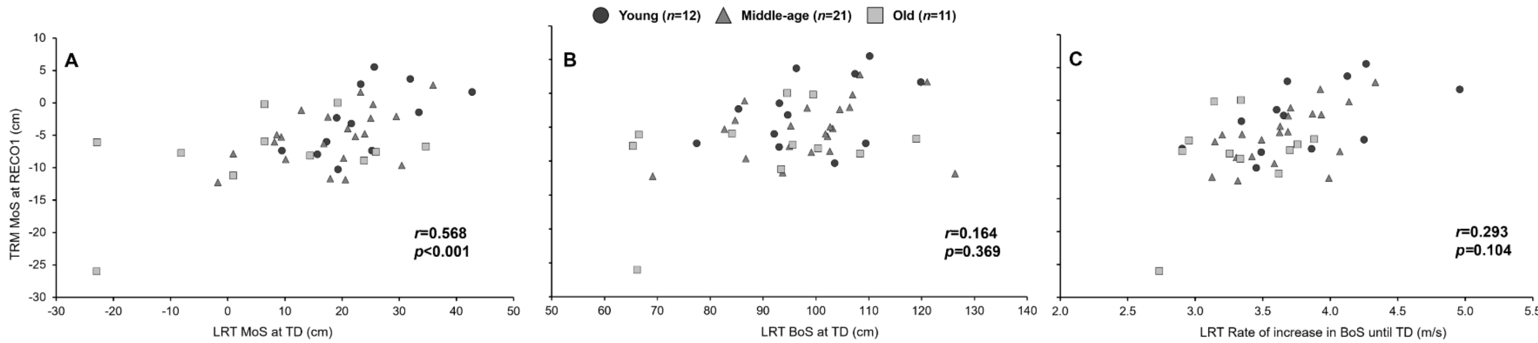

Figure 8: **(A)** Relationship between the margin of stability (MoS) of the first recovery step of the tripping task on the treadmill (TRM) and the MoS at foot touchdown (TD) during the lean-and-release-task (LRT). **(B)** Relationship between the MoS of the first recovery step of the TRM task and the base of support (BoS) at foot TD during the LRT. **(C)** Relationship between the MoS of the first recovery step of the TRM task and the rate of increase in BoS until foot TD during the LRT.

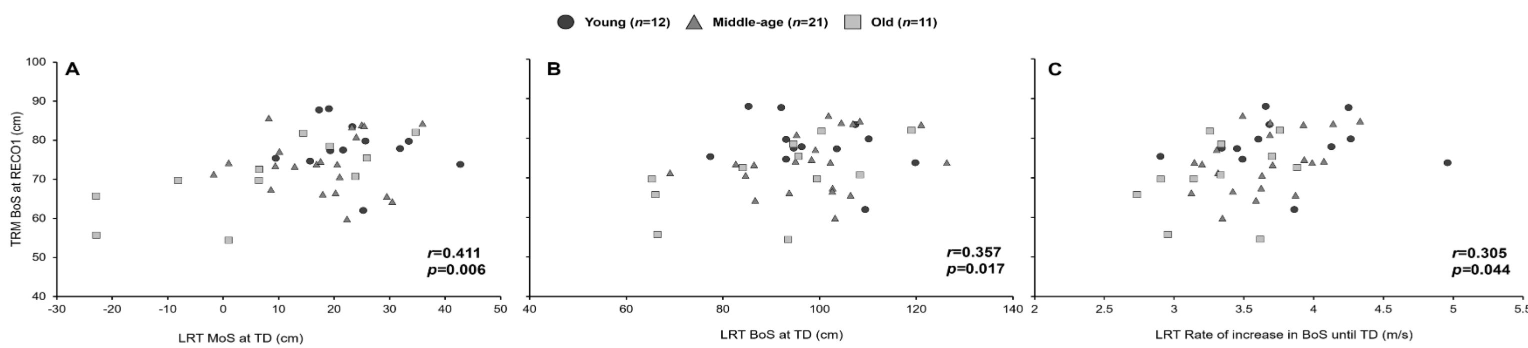

Figure 9: **(A)** Relationship between the base of support (BoS) of the first recovery step of the tripping task on the treadmill (TRM) and the margin of stability (MoS) at foot touchdown (TD) during the lean-and-release-task (LRT). **(B)** Relationship between the BoS of the first recovery step of the TRM task and the BoS at foot TD during the LRT. **(C)** Relationship between the BoS of the first recovery step of the TRM task and the rate of increase in BoS until foot TD during the LRT.

4.4.2. Single- and multiple-stepper subgroup comparison

Eighteen out of 44 participants were determined as multiple-steppers following sudden stability loss in the lean-and-release task (none of the young, 40 % of the middle-aged and 90 % of the old adults). Since younger adults are not representative of high fall risks, only middle-aged and older adults were included in the subgroup comparisons [single-steppers (n = 14) versus multiple-steppers (n = 18)] for dynamic stability control. Multiple- compared to single-steppers showed significantly lower MoS at TD [$t(30) = 3.228$, $p = 0.003$, $g = 1.151$], lower BoS at TD [$t(30) = 2.49$, $p = 0.019$, $g = 0.888$], as well as lower rates of increase in BoS until TD [$t(30) = 3.352$, $p = 0.002$, $g = 1.212$] during the lean-and-release task, with no significant differences in the MoS at release. There were no significant differences between multiple- and single-steppers in the MoS as well as the BoS at TD accounting for the steps from perturbation to the sixth recovery step during tripping (Figure 10). There was a statistically significant event-effect in the MoS and BoS of consecutive steps [$F(6,180) = 150.408$, $p < 0.001$, $\eta_p^2 = 0.834$; $F(6,180) = 105.152$, $p < 0.001$, $\eta_p^2 = 0.778$] independent of the subgroups. Post-hoc analysis revealed a higher MoS in the first four recovery steps ($p < 0.001$) and a significantly higher BoS in the first three recovery steps ($p < 0.001$), when comparing one step to the following one.

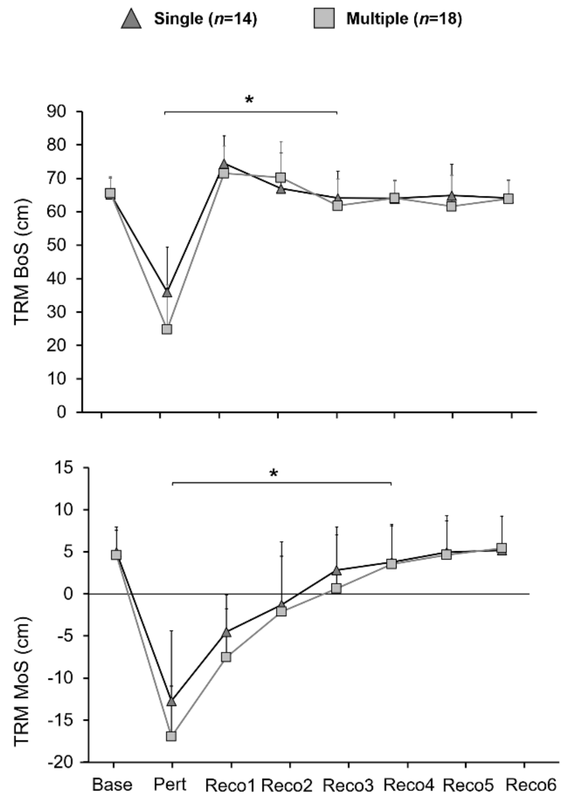

Figure 10: **(A)** Base of support (BoS) and **(B)** margin of stability (MoS) during the tripping task on the treadmill (TRM) for single- (*n* = 14) and multiple-steppers (*n* = 18). Data is shown for baseline walking (Base), at touchdown of the perturbation (Pert) as well as for the six recovery steps following the perturbation (Reco1 - Reco6) for the two subgroups. Values are presented as means with SD error bars. *: significant different BoS (first three recovery steps) and MoS (first four recovery steps) when comparing two consecutive steps ($p < 0.001$).

4.4.3. Age-related effect on stability performance

There was a statistically significant effect in age [$F(2,41) = 312.42$, $p < 0.001$, $\eta_p^2 = 0.934$] between the three analysed groups: young: 24 ± 3 years; middle-aged: 53 ± 5 years; older: 72 ± 5 years. Body height (176 ± 8 cm vs. 173 ± 11 cm vs. 170 ± 9 cm), body mass (70.8 ± 11.6 kg vs. 74.8 ± 12.7 kg vs. 73.3 ± 12.8 kg) and physical activity (6.2 ± 2.4 h/week vs. 6.6 ± 4.6 h/week vs. 6.5 ± 2.5 h/week) did not significantly differ between the three age groups. Regarding the MoS at TD of the recovery step in the lean-and-release task, there was a significant age effect [$F(2,41)$

= 5.279, p = 0.009, η_p^2 = 0.205; Figure 11], with older adults showing a lower MoS compared to young (p = 0.002) and middle-aged (p = 0.028) adults (Figure 11). The rate of increase in BoS showed a statistically significant age effect [$F(2,41)$ = 3.896, p = 0.028, η_p^2 = 0.159], with lower rates of increase in BoS for older compared to young adults (p = 0.007; Figure 11). No differences between groups in the BoS at TD were found.

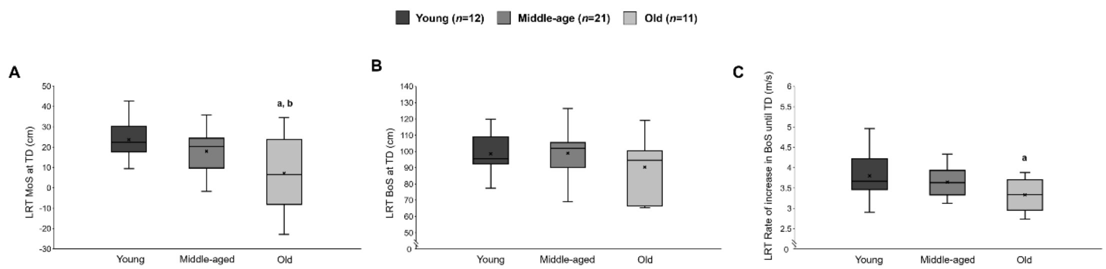

Figure 11: **(A)** Base of support (BoS) and **(B)** margin of stability (MoS) at foot touchdown (TD) and the **(C)** rate of increase in BoS until foot TD during the lean-and-release task (LRT). Results are presented as boxplots with the mean (line), median (x) and interquartile range between 25th and 75th percentile along with minimum and maximum values) for all three age-groups [young (n = 12), middle-aged (n = 21) and older adults (n = 11)]. a: old statistically different to young (0.002 < p < 0.007); b: old statistically different to middle-aged (p = 0.03).

Following the applied trip-like perturbation while walking, the MoS at TD of the perturbed step was on average -12.8 ± 9.4, -13.5 ± 7.5 and -18.2 ± 6.1 cm for young, middle-aged, and older adults, respectively. The analysis of the MoS at TD of the perturbed and following six recovery steps revealed a significant age effect [$F(2,41)$ = 3.74, p = 0.030, η_p^2 = 0.154]. Post-hoc tests revealed that older compared to young adults had a significantly (p = 0.008) lower MoS at TD (Figure 12). Although not reaching a statistical significance, there was a tendency (p = 0.053) for a lower MoS at TD during tripping in older compared to middle-aged adults. Additionally, there was a significant event effect [$F(6,246)$ = 196.35, p < 0.001, η_p^2 = 0.827] in the MoS at TD. Post-hoc analysis revealed a significantly higher MoS at TD in the first four recovery steps, when comparing two consecutive steps (p < 0.001). Regarding the BoS at TD, there were significant age [$F(2,41)$ =11.75, p < 0.001, η_p^2 = 0.364] and event [$F(6,246)$ = 101.93, p < 0.001, η_p^2 = 0.713] effects following the trip. Post-hoc analysis for age revealed a lower BoS at TD in older compared to the young (p <

0.001) and middle-aged adults (p = 0.011). Furthermore, middle-aged compared to young adults showed a lower (p = 0.013) BoS at TD following the trip (Figure 12). Post-hoc analysis of the event-effect showed a significantly higher BoS at TD for the first three recovery steps, when comparing two consecutive steps (p < 0.001).

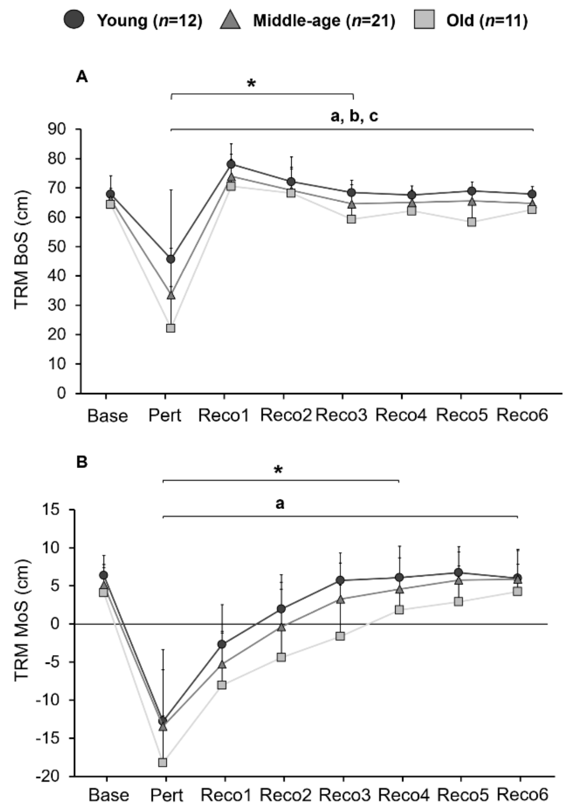

Figure 12: **(A)** Base of support (BoS) and **(B)** margin of stability (MoS) during tripping task on the treadmill (TRM). Data is shown for baseline walking (Base), for touchdown at perturbation (Pert) as well as for the six recovery steps following the perturbation (Reco1 - Reco6), in young (n = 12), middle-aged (n = 21) and older adults (n = 11). Values are presented as means with SD error bars. a: old significantly different to young (0.001 < p < 0.008); b: old significantly different to middle-aged (p = 0.011); c: middle-aged significantly different to young (p = 0.013); *: significant different BoS (first three recovery steps) and MoS (first four recovery steps) when comparing two consecutive steps (p < 0.001).

4.5. Discussion

The present study aimed to examine the association between the stability recovery performances in a lean-and-release task and a tripping-task during treadmill walking among adults of various ages. In addition, it was investigated if separating the participants into subgroups according to their stability recovery behaviour in the lean-and-release task (single- vs. multiple-steppers) reveals differences between the subgroups' recovery behaviour in the tripping-task. Whilst there were significant correlations between the lean-and-release and the tripping task, those were mainly weak to moderate with only up to one third of explained variance and heterogeneous in terms of statistical significance. Moreover, despite clear differences in the lean-and-release performance, single- and multiple-steppers demonstrated similar stability control when recovering from tripping. The combined pattern of results hence indicates limited generalisation of stability recovery performance between both tasks.

Previous research has shown that stability recovery performance in a lean-and-release task is a good predictor of future fall risk among older adults (Carty et al., 2015). The current study revealed like earlier studies (Carty et al., 2015; Thelen et al., 1997; Wojcik et al., 1999) a gradual age-related deterioration in the ability to recover stability with a single rapid step following a sudden stability loss as well as diminished recovery performance in tripping. It is widely accepted that the ability to increase the BoS rapidly and effectively in the anterior direction is an essential component of dynamic stability control (Carty et al., 2012) and represents one main motor response to recover stability following a sudden forward fall (Karamanidis et al., 2008) or trip (Epro et al., 2018a; McCrum et al., 2018). Since this recovery response is evoked in a similar manner in a lean-and-release task and during tripping, an association of the stability recovery performances between the two tasks could be expected. The current study however revealed no consistent pattern in the results, with only some of the correlations showing significant but weak correlations, indicating that the changes in MoS and BoS of the tripping-task are not related to the stepping behaviour of the lean-and-release task (Figure 8). Moreover, comparisons between single- and multi-steppers indicated differences only in stability recovery performance during the lean-and-release task but not in tripping recovery during walking on the treadmill. Despite an enhanced ability to rapidly increase the BoS and control stability in single- compared to the multiple-steppers during the lean-and-release task, there were no group-related differences in the recovery performance following a trip-like perturbation (BoS, MoS at TD from perturbed and the following six recovery steps). Thus, correlations and subgroup analyses did not indicate a functionally relevant association in stability recovery performances between both tasks.

Even though an important attribute of the neuromotor system is the capacity to transfer skills from one task to another, up to date literature is still lacking knowledge regarding the topic of inter-task transfer. It is suggested that the lean-and-release task and the tripping-task share similar stability recovery responses (König et al., 2019b), i.e. to increase the BoS rapidly and effectively in the anterior direction. To support this task similarity, an additional analysis for the BoS at TD during baseline walking, first recovery step during tripping as well as during the lean-and-release task was performed. Results showed significantly higher ($p < 0.001$) values in the first recovery step for the tripping-task and lean-and-release task compared to baseline walking (BoS at TD during baseline walking: 66 ± 5 cm; gait perturbation: 74 ± 8 cm; lean-and-release task: 97 ± 14 cm; $p < 0.001$). This confirms previous observations stating that an effective anteriorly increase in the BoS is required to recover stability following a sudden large perturbation as in the two tasks investigated, hence strengthening the assumption of a shared stability recovery responses (König et al., 2019b). However, the present study was unable to prove functionally relevant associations between the stability recovery performance of both tasks, suggesting that stepping recovery in a lean-and-release task seems not to be a valid measure to predict the recovery performance after tripping. These results are supported by earlier findings reporting no inter-task transfer of fall-resisting skills from an unexpected trip-perturbation to a sudden release from a forward-inclined position (König et al., 2019b). In contrast, previous studies found positive transfers of adaptations between different tasks using similar perturbation methods, i.e. slipping evoked by platform translation to untrained walking over a slippery surface (Bhatt & Pai, 2009). These opposing findings suggest that generalisation of stability recovery skills from one task to another might be possible but seem to be limited if factors beyond common recovery responses differentiate perturbation responses in motor tasks sharing the same main stability recovery response.

Previous research has shown that different biomechanical demands or perturbations elicit distinct ‘task-specific’ motor components, even between highly similar tasks, e.g. mechanically induced perturbations during standing on a stable or unstable platform (Chvatal & Ting, 2013; Munoz-Martel et al., 2019). Despite that in the current study the MoS at TD of the perturbed step during treadmill tripping (on average for all analysed subjects: -0.15 ± 0.09 m) matched the MoS at the time point of release during the lean-and-release task (-0.14 ± 0.08 m), differences in task difficulty cannot be ruled out entirely as a contributing factor to the low correlations. Although sharing a similar stability control response, the absolute values of the magnitude of increase in the BoS were approximately 1.3 times higher for the lean-and-release task compared to the tripping task, which may have at least partly been induced by the lean-and-release task being more challenging due to its task-specific requirement to regain stability using a single step. Nevertheless, the current findings revealed significant age-related differences in recovering from tripping, whereas multiple- versus single-steppers in the lean-and-release task demonstrated no differences during tripping. Thus, whilst both tasks clearly demonstrated challenges on dynamic stability control, limited transfer and generalisation cannot be explained only based on the weak or moderate inter-task correlations but further on the sub-group comparisons (single vs. multiple steppers). Although not in the scope of the current study, a possible explanation for the lack of generalisation for recovery performances to different perturbation tasks may lay beyond the similarities in spatiotemporal stepping characteristics. Thus, it cannot be excluded that neuromotor control required to recover stability might differ between the two deployed tasks in the current study, i.e. they share only a limited number of muscle synergies possibly affecting the small associations between recovery performances. Regarding this, although the lean-and-release as well as tripping task are frequently considered for the investigation of stability recovery performance, they should not be used interchangeably in clinical settings.

It is important to note that the current study addressed only the antero-posterior components of dynamic stability control since both tasks consist of anteriorly induced perturbations. One might argue that the medio-lateral stability could have played a role potentially affecting the current results. However, when analysing spatiotemporal components (medio-lateral directed increase in BoS and velocity of the CoM at TD), both parameters were in absolute terms multiple factors lower than the antero-posterior components for each task respectively (increase in BoS on average during LRT and TRM for the medio-lateral versus antero-posterior direction: 0.01 ± 0.06 m versus 0.97 ± 0.14 m, and 0.05 ± 0.15 m versus 0.74 ± 0.08 m; velocity of the CoM at TD on average: 0.19 ± 0.45 m/s versus 1.28 ± 0.22 m/s, and 0.12 ± 0.06 m/s versus 1.36 ± 0.17 m/s, respectively). Thus, we are confident that the effects of the medio-lateral stability during the anteriorly directed perturbations used in the current study were less functionally relevant compared to the antero-posterior components.

4.6. Conclusion

In conclusion, no functionally relevant associations were identified between unpractised recovery performances following a sudden stability loss from a static forward-inclined position and a trip during treadmill walking. Moreover, alike previously performed studies the current results showed deteriorations in the ability to recover from unexpected stability perturbations with aging. Thus, the current study provides evidence that the ability to increase the BoS and effectively recover from stability perturbations deteriorates with aging and is limited in its generalisation for different perturbation tasks in adults across a wide age range.

4.7. Acknowledgments

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5. Third study | Differences in motor responses to stability perturbations limit fall-resisting skill transfer

Scientific Reports (2022, v. 12; DOI: 10.1038/s41598-022-26474-7).

5.1. Abstract

This study investigated transfer of improved stability recovery performance to unpractised perturbations. Thirty adults (20-53 years) were assigned equally to three treadmill walking groups: groups exposed to eight trip perturbations of either low or high magnitude and a third control group that walked unperturbed. Following treadmill walking, participants were exposed to stability loss from a forward-inclined position (lean-and-release) and an overground trip. Lower limb joint kinematics for the swing phase of recovery steps were compared for the three tasks using statistical parametric mapping and recovery performance was assessed by analysing margin of stability and base of support. The perturbation groups improved stability (greater margin of stability) over the eight gait perturbations. There was no group effect for stability recovery in lean-and-release. For the overground trip, both perturbation groups showed similar enhanced stability recovery (margin of stability and base of support) compared to controls. Differences in joint angle kinematics between treadmill-perturbation and lean-and-release were more prolonged and greater than between the two gait perturbation tasks. This study indicates that (i) practising stability control enhances human resilience to unpractised perturbations, (ii) which is not necessarily dependent on the perturbation magnitude but (iii) may be limited by differences in motor response patterns between tasks.

5.2. Introduction

Human locomotion daily faces a variety of perturbations to stability that provoke adjustments to maintain postural integrity and avoid falls. It has been suggested that the central nervous system monitors and corrects motor responses based on the prediction of sensory consequences of perturbations (Robinson, 1975). This must however be accurate to the nature of perturbation, and hence motor control is constantly refined based on error-feedback information (Diener et al., 1988; Shadmehr, 2017). In mechanical terms, the system commands an internal representation of the centre of mass (CoM) in relation to the base of support (BoS) based on prior experience (Conditt et al., 1997; Pai & Iqbal, 1999). If exposed to perturbations which lead to excursion of the CoM beyond the boundaries of the BoS [a state of instability (Gill et al., 2019; Hof, 2008)], such information will be received, and appropriate motor responses follow to regain the desired state of the CoM, i.e. a stable body configuration. Given such capability of neuromotor processing, recovery responses adapted from practised exposure to perturbations could enhance coping with altered forms of the practised perturbation (Bhatt & Pai, 2008, 2009; Grabiner et al., 2012; Patel & Bhatt, 2015). Based on these assumptions, developing stability control through repeatedly perturbed locomotion has been recognised as an important paradigm for acquisition of general skills for resisting falls in daily life (Harper et al., 2021; Karamanidis et al., 2020).

Extensive research studies have attempted to mimic real-life situations of postural threats during locomotion and practised these (e.g., trips or slips). They indicate that single sessions of repeated perturbation practice can elicit acute and retainable improvements in stability control in adults across the lifespan (Bhatt & Pai, 2009; Epro et al., 2018b; König et al. 2019b; Lee et al., 2018; Wang et al., 2011; Yang et al., 2013). Transfer of such stability improvements induced by practice to altered forms of the practised perturbations has been revealed previously (Bhatt & Pai, 2009; Grabiner et al., 2012; Lee et al., 2018; Wang et al., 2011; Yang et al., 2013), and reported to reduce falls incidence in daily life of community-dwelling adults (Pai et al., 2014; Rosenblatt et al., 2013) - although not for all falls. Those results are in line with the general assumption that such skill transfer relies on a common ground of contextual sensory feedback information between practised and non-practised perturbations, which would only require the system to fine tune or modify the adapted motor response (Bhatt & Pai, 2008; Patel & Bhatt, 2015; Bhatt et al., 2013). If the characteristics of the perturbation (e.g. magnitude or environment) changed, the system would need to adjust its motor response to a different stability constraint to achieve positive transfer. Such might be readily achieved if the types of perturbations elicit some degree of shared stability responses for recovery. In our recent studies we could not, however, show functionally relevant associations (Bosquée et al., 2021) nor performance transfer from repeated treadmill-based gait perturbations (König et al., 2019b; König et al., 2022) to a clinical fall-risk assessment in the form of a lean-and-release task (Carty et al., 2015). Performance transfer failed to occur even though both tasks shared the same direction of perturbation (anterior) and the same stability recovery response (a rapid anterior recovery step after stability loss). We suggested distinctive synergistic control of muscles as a potential factor limiting transfer of motor skill adaptations between the two tasks (König et al., 2022). Nevertheless, since practised stability control via repeatedly perturbed locomotion has been confirmed to be beneficial ~~to~~ in resistance to practised perturbations, transfer most likely relies on other factors yet to be examined.

The design of perturbation practice paradigms might play a central role in adapted skill transfer. One crucial factor could be the perturbation magnitude affecting the applied motor error during practice. There is evidence that adaptations in stability control to treadmill-based perturbations can be transferred to an even higher magnitude for a situation for which the contextual information of the motor error is the same (Patel & Bhatt, 2015). Accordingly, practising at a certain perturbation magnitude could elicit a calibration of motor output to different perturbation magnitudes, potentially even to different perturbation tasks. Another study indicated that transfer of adapted stability control requires exposure to practised perturbation of appropriately high magnitude (Liu et al., 2016) – yet even higher than magnitudes which are sufficient to elicit adaptations during practice (Pai et al., 2014). Thus, for the perturbation magnitude used in our treadmill-based perturbation paradigms the given perturbation magnitude might have been too low to provoke transfer, hence it can be suggested that an increased perturbation magnitude during practice could lead to greater adaptations (Jayaram et al., 2011) and hence enhance performance transfer. In contrast to this, modelling and experimental studies have argued that there is a nonlinear relationship between the sizes of error feedback and of adaptation (Wei & Kording, 2009), suggesting that greater motor errors would neither mean greater adaptation nor greater transfer of motor skill adaptations.

Therefore, in addition to our previous transfer investigations (König et al., 2019b; König et al., 2022), this study aimed to examine whether an increased magnitude would elicit or even enhance transfer of adapted recovery performance during short-term practising on treadmill-based gait-trips to a non-practised stability loss from a static forward-inclined position (lean-and-release task) and to an overground trip. Both transfer tasks were chosen as they share a similar motor response to recovery stability (increase in the BoS in the anterior direction by stepping) as that required for treadmill-based gait-trips. It was hypothesised that exposures to higher perturbation magnitudes during practice would lead to more pronounced transfer effects in unpractised stability perturbation tasks.

5.3. Methods

5.3.1. Participants and experimental design

Thirty healthy and moderately physically active adults (20-53 years of age) were recruited. Exclusion criteria were any neurological or musculoskeletal injuries or impairments limiting locomotion. After informed consent was obtained from all participants, they were randomly assigned to three groups of equal size. Two groups were exposed to eight successive trip-like perturbations while walking on a treadmill, with perturbations applied at higher perturbation magnitude or lower magnitude (TRM_{high} group, *n* = 10, three females, averages and standard deviations of age, body height and body mass: 24.7 ± 4.9 years, 1.77 ± 0.08 m, 81.1 ± 16.6 kg; TRM_{low} group, *n* = 10, four females, 29.0 ± 8.5 years, 1.75 ± 0.17 m, 70.9 ± 13.7 kg). A control group walked unperturbed on the treadmill for a similar duration (approximately 20 minutes) to the other groups (CTRL group, *n* = 10, one female, 32.9 ± 10.4 years, 1.80 ± 0.084 m, 82.7 ± 13.8 kg). Afterwards, a single trial of each of two non-practised transfer tasks took place in the same order for all participants, i.e. at fist stability recovery after sudden release from a static forward-inclined position (lean-and-release task), followed by stability recovery after a trip-like perturbation while walking over a flat surface (overground trip). There were short rests of 10 minutes between all tasks. Participants wore their own non-slippery leisure/sports shoes throughout all measurements. They were protected by wearing a safety harness connected to an overhead track that allowed for full range of motion in anterior-posterior and medio-lateral directions but prevented contact of any part of the body with the ground (except for the feet). Measurements were reviewed and approved by the ethics committee of the School of Applied Sciences at London South Bank University (approval ID: SAS1826b) and met all requirements for human experimentation in accordance with the Declaration of Helsinki (World Medical Association, 2013).

5.3.2. Trip-like perturbation practice

The trip-like perturbation paradigm has been used in previous studies (Epro et al., 2018a; 2018b; König et al., 2019a; 2019b). Four to seven days prior to measurements, all participants were familiarised with unperturbed treadmill walking. Participants walked on a treadmill (Valiant 2 sport XL; Lode B.V., Groningen, The Netherlands) at a standard speed (1.4 $m \cdot s^{-1}$). A Teflon cable and ankle strap connected each of a participant’s ankles to a custom-built pneumatically driven perturbation device located behind the treadmill (Figure 13). The strap created a negligible resistance of less than 3 N. Following four minutes of walking (Karamanidis et al., 2003), recordings of twelve consecutive steps served to determine stability control during unperturbed walking (Epro et al., 2018b). As the participants continued to walk, eight trip-like perturbations were induced unexpectedly, with each successive perturbation being followed by variable washout periods (2-3 minutes) of unperturbed walking (König et al., 2019b; Epro et al., 2018b). The perturbations were induced by the experimenter activating a pneumatic cylinder using a hand trigger connected to the perturbation device. A restraining force was thereby applied to the left limb via a Teflon cable and ankle strap during mid-stance phase of the right foot to standardise an interruption to motion of the left limb during its mid-swing (i.e. anterior velocity of the lateral malleolus equalled zero).

The restraining force was released at touchdown of the left foot to allow for continuity in walking after the perturbation. The subsequent anterior increase in the BoS using the contralateral right leg was defined as the recovery step. One group (TRM_{low}) was perturbed in a manner that has previously been shown to improve retainable fall-resisting skills (100 N restraining force), with a rise time of ~20 ms (Epro et al., 2018a). Another group (TRM_{high}) was exposed to an increased perturbation magnitude (140 N, rise time ~20 ms). Although participants were informed of being perturbed at some points during walking and were encouraged to continue walking, the onset and removal of the resistance was applied without any immediate warning. The CTRL group walked unperturbed at the same standard speed (1.4 $m \cdot s^{-1}$) for a similar period of time as the perturbation groups.

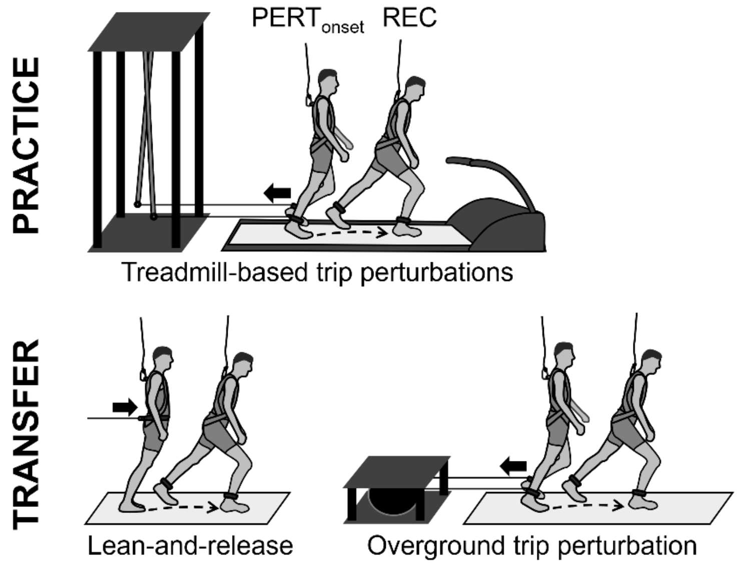

Figure 13: Schematic illustration of the practised and the two transfer stability perturbation tasks. The practised task consisted of eight successive trip-like gait perturbations on the treadmill. Perturbations were induced using a custom-built pneumatically driven cylinder system at unexpected timepoints during a swing phase of the left leg ($PERT_{onset}$) eliciting subsequent touchdown (PERT) followed by a recovery step with the right leg (REC). In the first transfer task after treadmill-based practice (Lean-and-release), all participants were released from a forward-inclined position once only. Lean angles were normalised to the participant's body mass (33 % of body mass). In the second transfer task following lean-and-release (Overground trip perturbation), all participants were exposed to one trip-like overground gait perturbation (gait speed matched to the treadmill speed at 1.4 $m \cdot s^{-1}$) induced using a method as for treadmill-based trip perturbations. A safety harness was worn during all tasks to prevent contact of any part of the body with the ground (except for the feet).

5.3.3. Lean-and-release transfer task

This task was operated according to previous studies (Karamanidis & Arampatzis, 2007; Karamanidis et al., 2008). Participants were forward-inclined with their feet placed flat and at hip-width on the first of two force platforms mounted in series (1080 Hz, 60 x 90 cm, Kistler, Winterthur, Switzerland; Figure 13). The inclination was maintained by means of an inextensible, horizontally running supporting cable attached to a belt around the participant's pelvis and at the other end to a custom-built pneumatically driven brake-and-release system. The inclination was set with 33 ± 2 % of participant body mass as measured by a load cell incorporated into the supporting cable (Karamanidis & Arampatzis, 2007; Karamanidis et al., 2008). During task instruction participants were asked to choose the right leg to recover stability after release using a single step onto the second force platform. To initiate the perturbation the supporting cable was suddenly released within 10 to 30 seconds after the participant was stabilised in the starting position. Note that there were no prior practice trials for this task to ensure an unpractised task condition.

5.3.4. Overground trip transfer task

Participants walked at a standard speed (1.4 $m \cdot s^{-1}$) on a custom-built flat wooden walkway (8 m length, 1.2 m width), with a Teflon cable and ankle strap attached to both ankles (Weber et al., 2022). The cables were in turn attached to a custom-build pneumatically driven brake-and-release device located behind the walkway (Figure 13). Walking speed was monitored live via an optical motion capture system that recorded a reflective marker located on the seventh cervical vertebra (16 infrared cameras operating at 120 Hz; Miqus2, Qualisys, Gothenburg, Sweden). Once participants arrived at the end of the walkway, they were guided back to the initial position to prevent tangling of the Teflon cable. Thus only one direction was considered for measurements. Following familiarisation with this walking, recordings of three consecutive forward walking trials (in total 12 steps as for the *Trip-like perturbation practice*) served to determine stability control during movement on the walkway. Subsequently, a single trip-like perturbation was induced randomly within the subsequent five to ten forward-walking trials, when the standard speed was consistently reached (Weber et al., 2022). As for *Trip-like perturbation practice*, the perturbation was operated by means of a hand trigger connected to the perturbation device and evoked by a breaking action of the Teflon cable on the left leg. This occurred during mid-stance phase of the right leg; the brake was released at touchdown of the left foot. Note that the strap created a negligible resistance of less than 3 N during unperturbed walking. The subsequent anterior increase in the BoS using the contralateral right leg was defined as the recovery step. Although participants were informed that their walking would be perturbed at some point and they were encouraged to continue walking after perturbation, the onset and removal of the resistance was applied without any immediate warning. Similar to the *Lean-and-release transfer task* there were no prior practice perturbation trials, ensuring an unpractised task condition.

5.3.5. Data collection and processing

In addition to that on the seventh cervical vertebra, further eight reflective markers were tracked via the optical motion capture system. These were placed on both greater trochanters, lateral epicondyles of the femur, lateral malleoli, and the tips of the big toe (see *Appendix* for a detailed description). Three-dimensional coordinates of the markers were smoothed using a fourth-order digital Butterworth filter with a cut-off frequency of 20 Hz. To assess the state of stability, the anterior margin of stability (MoS) was calculated in accordance with Hof and colleagues (Hof et al., 2005) as the difference between the extrapolated CoM in the anterior direction and the anterior boundary of the BoS (front toe marker) at foot touchdown. Extrapolated CoM was calculated as the sum of the position of the CoM (the average of left and right trochanter) and the average velocity of the CoM and C7 in relation to the square root of gravitational acceleration by reference leg length (Süptitz et al., 2013). The BoS was defined as the distance between the anterior boundary of the leading and trailing feet (i.e., the difference between the projections of the two toe markers). Foot touchdowns during locomotion were determined using different approaches depending on the motor task analysed. For treadmill walking, impact peaks of two 2D accelerometers (1080 Hz; ADXL250; Analog Devices, Norwood, MA, USA) placed over the tibia of each leg were used (Süptitz et al., 2012) as the treadmill that was used did not incorporate force plates. For overground walking, the vertical position and acceleration of the heel and toe markers were employed (Maiwald et al., 2009). For the lean-and-release task, using force platform data, touchdown was defined by the time at which the vertical ground reaction force exceeded 20 N. Foot toe-off was estimated using the local maximum in the vertical acceleration of the toe marker in relation to its minimum vertical position (Maiwald et al., 2009) for all tasks.

During unperturbed walking on both treadmill and overground, stability was determined as the averaged MoS and BoS across six consecutive foot touchdowns of both legs. The state of instability (MoS and BoS) at the time of perturbation during both walking tasks was identified at touchdown of the perturbed left foot (after resistance was applied). In the lean-and-release task the state of instability was determined at the release of the supporting cable (50 % reduction in the leaning force recorded by the incorporated load cell). Stability recovery performance was evaluated at foot touchdown of the recovery step for each task (right leg). In addition, sagittal plane joint angles at the ankle, knee and hip were calculated for the swing phase of the recovery step (take-off until touchdown normalised to 101 points for each task and participant) for all three tasks. Subsequent analyses of kinematics served to further examine generalisation of motor output in recovery from the several stability perturbations, which would support to critical discussion of the initial assumption that performance transfer would be possible if recovery characteristics (i.e. increase in BoS by stepping) were similar. Accordingly, data from the eighth trial of treadmill-based perturbations, as well as from the unique trials of both lean-and-release and walkway trip were used.

5.3.6. Statistics

Parametric assumptions for both parameters (MoS, BoS) were checked and confirmed using Shapiro-Wilk tests ($p > 0.05$). Possible differences between practice groups and controls in age, body mass and body height were examined using separate one-way analyses of variance (ANOVA). For the treadmill task, only the first (Trial 1) and eighth (Trial 8) perturbations were considered for the analysis of adaptive changes in stability control, as these represent the first (unpractised) and practiced performances. To assess the effect of perturbation magnitude on stability, two-factor ANOVA were computed for both MoS and BoS separately, and for Trials 1 and 8, with factors event (levels: unperturbed walking, perturbation) and group (levels: TRM_{low}, TRM_{high}). In addition, two-factor ANOVA, also with factors event (levels: unperturbed walking, recovery step after perturbation) and group (levels: TRM_{low}, TRM_{high}), were computed for the BoS, separately for Trials 1 and 8 serving to assess changes in treadmill walking in response to sudden perturbations.

Adaptive changes due to practice were assessed using separate two-factor ANOVA with factors trial (levels: Trial 1, Trial 8) and group (levels: TRM_{low}, TRM_{high}) for both MoS and BoS at perturbation and subsequent recovery step.

Potential transfer of stability control adaptations from perturbation practice to the performance in the lean-and-release task and overground trip was assessed by comparing MoS and BoS at perturbation and subsequent recovery step for the three groups (TRM_{low}, TRM_{high} and CTRL) using separate one-way ANOVA. To evaluate the effect of exposure to an overground trip on stability, separate two-factor ANOVA with factors event (levels: unperturbed walking, perturbation) and group (levels: TRM_{low} vs. TRM_{high} vs. CTRL) were used. In addition, a two-factor ANOVA with factors event (levels: unperturbed walking vs. recovery step after perturbation) and group (levels: TRM_{low} vs. TRM_{high} vs. CTRL) was computed for the BoS to assess changes in overground locomotion. In cases of significant main effects, Bonferroni *post-hoc* corrections were applied. In addition to stability analyses for evaluation of performance transfer between tasks, sagittal plane joint angle kinematics of ankle, knee, and hip joint angles for the recovery step were compared using statistical parametric mapping (SPM) open-source code SPM1d (version M.0.4.8, www.spm1d.org). One-way repeated measures ANOVA from the three tasks (eighth treadmill perturbation, lean-and-release, overground trip) was applied to the kinematic data for each of the three joints. A statistical parametric map SPM{F} was created by calculating the conventional univariate F-statistic at each point of the entire swing phase of the recovery step. If SPM{F} crossed a threshold corresponding to 0.99, *post-hoc* SPM{t} maps were calculated for each of the three pairwise comparisons. When the SPM{t} map crossed the critical threshold, a significant difference ($\alpha = 0.01$) was found between the examined pair of trials. Furthermore, for the entire swing phase of the recovery steps, root mean square errors (RMSE) were computed for the three joints (°) to determine the averaged difference in absolute magnitude observed for lean-and-release task as well as for overground trip kinematics from those during the eighth treadmill-based trip. All analyses were performed using SPSS Statistics (v27, IBM; Chicago, IL, USA) and MATLAB (2020b, MathWorks®, Natick, MA, USA) and if not stated otherwise, statistical significance was set at $\alpha = 0.05$.

5.4. Results

5.4.1. Stability control for treadmill-based perturbation practice

At Trial 1 as well as Trial 8 there was a significantly lower MoS at perturbation compared to unperturbed walking in both practice groups ($p \leq 0.001$), with TRM_{high} compared to TRM_{low} showing a 2.4-fold lower ($p < 0.001$) MoS at Trial 1 ($F = 86.49$, $p < 0.001$; Figure 14), and a 7.8-fold lower MoS ($p < 0.001$) at Trial 8 ($F = 56.31$, $p < 0.001$; Figure 14). The BoS at perturbation compared to unperturbed walking was lower ($p < 0.001$) for both trials in both practice groups, whilst a 2.9-fold lower ($p = 0.003$) BoS at Trial 1 ($F = 14.97$, $p = 0.001$; Figure 14), and a 3.6-fold lower BoS at Trial 8 ($F = 23.04$, $p < 0.001$; Figure 14) were revealed for TRM_{high} compared to TRM_{low}. For the eighth perturbation, both perturbation groups showed on average a more positive MoS (more stable state of stability) at perturbation compared to the first perturbation ($F = 20.79$, $p < 0.001$; Figure 14). The MoS at touchdown of the subsequent recovery step after the eighth perturbation was higher than for the first one ($F = 62.03$, $p < 0.001$; Trial 8 of 1.3 ± 3.4 cm vs. Trial 1 of -5.3 ± 3.3cm) for both practice groups, with a slightly lower MoS overall (Trial 1 and Trial 8; $F = 4.17$, $p = 0.044$) for TRM_{high} compared to TRM_{low}. Whilst there were no differences between practice groups, nor between perturbation trials (Trial 1 vs. Trial 8), the BoS at touchdown of the recovery step after perturbation was higher compared to unperturbed walking at Trial 1 ($F = 7.23$, $p = 0.015$) as well as at Trial 8 ($F = 5.92$, $p = 0.026$).

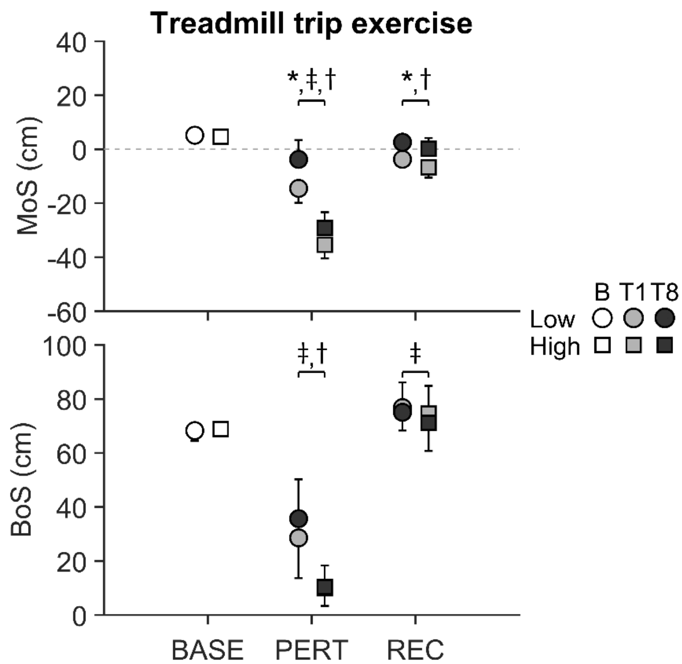

Figure 14: Margin of stability (MoS, top) and base of support (BoS, bottom) during for unperturbed baseline walking (B, BASE) and the first (T1) and eighth (T8) trials of treadmill-based perturbation ~~exercise~~ practice for low (n = 10) and high (n = 10) perturbation magnitude groups. Data for T1 and T8 is shown at left foot touchdown after perturbation (PERT) and the subsequent right foot touchdown after recovery step (REC). ~~Please n~~ Note that PERT at T1 (9.9 ± 6.9 cm) and T8 (10.4 ± 8.5 cm) showed quite similar mean values for the high perturbation magnitude group. Values are presented as means with SD error bars. ‡: sig. different to BASE at T1 and T8 for low and high (p < 0.05); *: sig. different between T1 and T8 for low and high (p < 0.001); †: sig. different between low and high at T1 and T8 (p < 0.05).

5.4.2. Transfer of practised stability control to unpractised perturbations

The state of instability caused by the perturbation in the lean-and-release task (MoS at release of the supporting cable) did not differ amongst the three groups (TRM_{high}, TRM_{low}, CTRL; F = 1.47, p > 0.05; Figure 15). The analysis of potential treadmill-based transfer in stability performance resulted in no significant effects for MoS or BoS at touchdown of the recovery step after release neither between the practice groups nor between these and the control group (Figure 15). With respect to the overground trip, the perturbation caused a lower (F = 822.14, p < 0.001) MoS compared to unperturbed walking, with no significant differences amongst the three groups for neither of the two events (Figure 15). Furthermore, the BoS at timepoint of the perturbations compared to unperturbed walking was lower ($F = 549.37$, $p < 0.001$) in all three groups, with no significant group effects at neither of the two events (Figure 15). However, there was a significant group effect for touchdown of the recovery step after perturbation ($F = 9.24$, $p < 0.001$). *Post-hoc* tests showed a significantly higher MoS for both TRM_{low} ($p = 0.001$) and TRM_{high} ($p = 0.008$) compared to CTRL, with no further effect of the ~~exercised~~ practised perturbation magnitude (TRM_{low} vs. TRM_{high}) on transfer performance (Figure 15). Whilst all groups showed a significant higher BoS at touchdown of the recovery step compared to unperturbed walking ($F = 153.20$; $p < 0.001$), there was a group effect for the BoS at touchdown of the recovery step ($F = 6.35$; $p = 0.005$). *Post-hoc* tests indicated significance for TRM_{low} ($p = 0.006$) and for TRM_{high} ($p = 0.048$) for the BoS compared to CTRL. Note that there were no significant ($p > 0.05$) group effects for age, body height, and body mass.

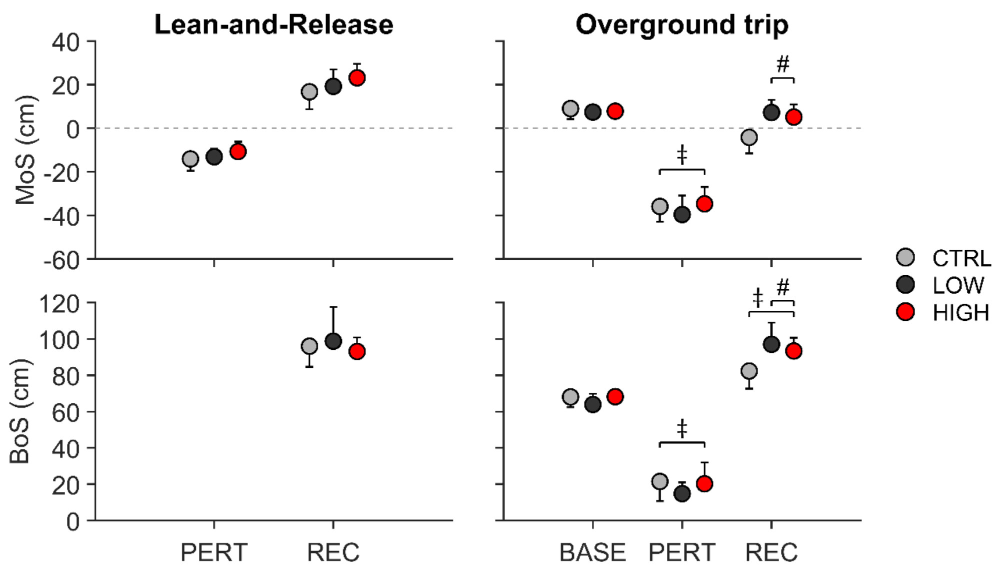

Figure 15: Margin of stability (MoS, top) and base of support (BoS, bottom) during lean-and-release and overground trip transfer tasks for low ($n = 10$) and high ($n = 10$) perturbation magnitude groups as well as controls ($n = 10$; CTRL). Data is shown for cable release or left foot touchdown after gait perturbation (PERT) and subsequent right foot touchdown for recovery step (REC) for all groups and unperturbed baseline walking (BASE) only for the overground task. Note that the BoS at PERT equalled zero for all groups during lean-and-release and hence is not shown. Values are presented as means with SD error bars. ‡: sig. different to BASE ($p < 0.001$); #: sig. different to CTRL ($p < 0.05$).

SPM analyses for the TRM_{low} group computed separately for each joint showed significant main effects for ankle [0-11 % of swing phase (*p* = 0.007), 33-55 % (*p* = 0.002)], knee [0-100 % (*p* < 0.001)] and hip [25-100 % (*p* < 0.001)] across the three perturbation tasks (eighth treadmill trip vs. lean-and-release vs. overground trip; Figure 16). *Post-hoc* tests [SPM{*t*}] on the knee and hip joint angle kinematics revealed that the lean-and-release task compared to the treadmill trip was recovered with significantly more knee as well as hip flexion for most of the swing phase [knee: 10-99 %, i.e. 89 % in total, *p* < 0.001; hip: 20-100 %, i.e. 80 %, *p* < 0.001]. Furthermore, recovering from the lean-and-release task compared to the treadmill trip involved significantly higher dorsiflexion (*p* = 0.001) between 0-14 % of the swing phase. In comparison to this, the overground trip compared to the treadmill trip differed only for 27 % of the swing phase in knee (45-59 % and 86-99 %, *p* = 0.003), 43 % in hip (57-100 %, *p* < 0.001) and 2 % ankle joint (0-2 %, *p* = 0.010).

A similar trend for inter-task differences in kinematics was found for the TRM_{high} group (Figure 17). Recovering from the lean-and-release task differed in total for 40% of the entire swing phase in knee (28-68 %, *p* < 0.001), 49 % in hip (51-100 %, *p* < 0.001), and 35 % in ankle (0-35 %, *p* < 0.001) joints compared to the treadmill task. In contrast, overground compared to treadmill trip showed only differences for the knee and the ankle joints at the initiation of the swing phase (ankle: 0-13 %, *p* = 0.005; knee: 0-17 %, *p* = 0.003; Figure 17). Furthermore, the overground trip compared to the lean-and-release task showed ~1.5 to 2-fold lower differences in absolute size in knee and hip joint angle kinematics for the swing phase of the recovery step from the treadmill trip, and that independent of the exercise practice group [RMSE for the overground trip vs. the lean-and-release task compared to the treadmill trip in knee (TRM_{low}, 15 vs. 30 ° and TRM_{high}, 11 vs. 20 °) and in hip (TRM_{low}, 14 vs. 21 ° and TRM_{high}, 8 vs. 17 °)].

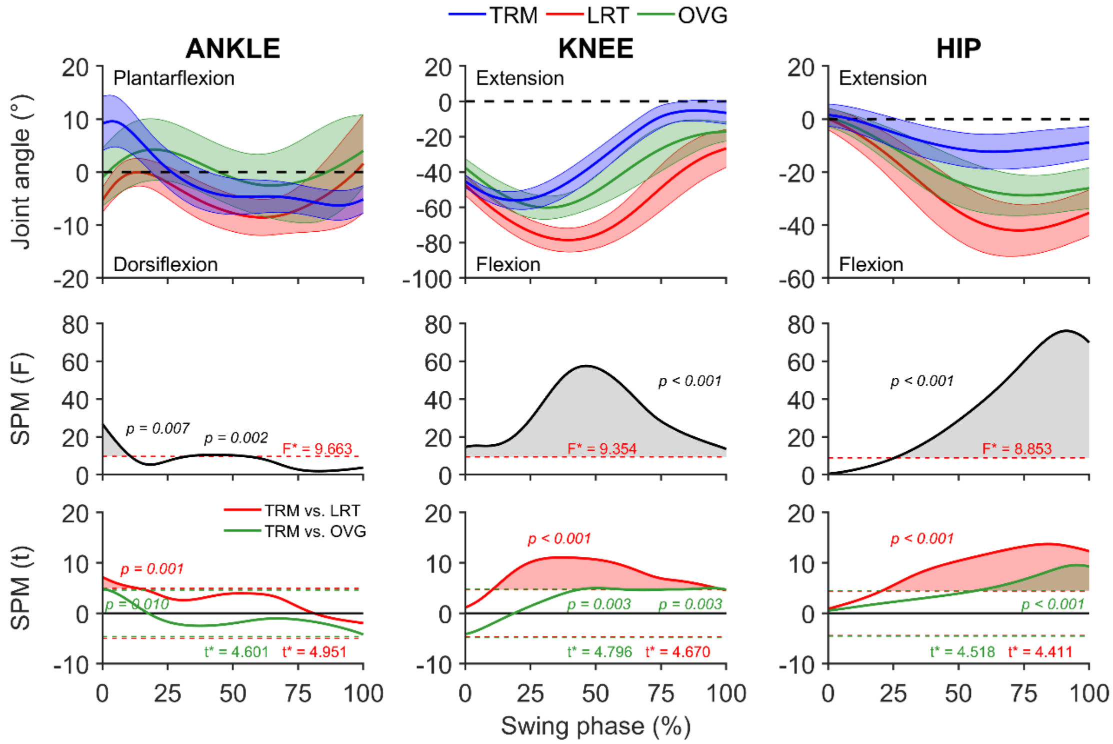

Figure 16: Sagittal plane joint angle kinematics for the low perturbation magnitude group (*n* = 10) and statistical parametric mapping (SPM) analyses of the ankle, knee, and hip joint during the entire swing phase (toe-off to touchdown, 0-100 %) of the recovery step for the eighth treadmill-based trip (TRM), lean-and-release (LRT), and overground trip (OVG). *1st row*: Joint angle comparison between all three tasks via means ± standard deviation (bold lines and shaded areas) across participants. *2nd row*: SPM one-way repeated measures ANOVA [SPM(*F*)] and univariate *F*-statistic (*F**) with significant threshold at 99 % confidence (dashed, red line) with task as factor (TRM, LRT, OVG). The shaded grey areas indicate to significant differences between the three tasks. *3rd row*: post-hoc tests [SPM{*t*}] comparing pairs of independent joint angle curves (i.e. TRM vs. LRT, red; TRM vs. OVG, green). *t*-statistic (*t**) with significant threshold at 99 % confidence is shown with dashed, red/green lines, and shaded red/green areas indicate to significant differences between the respective pairs of tasks.

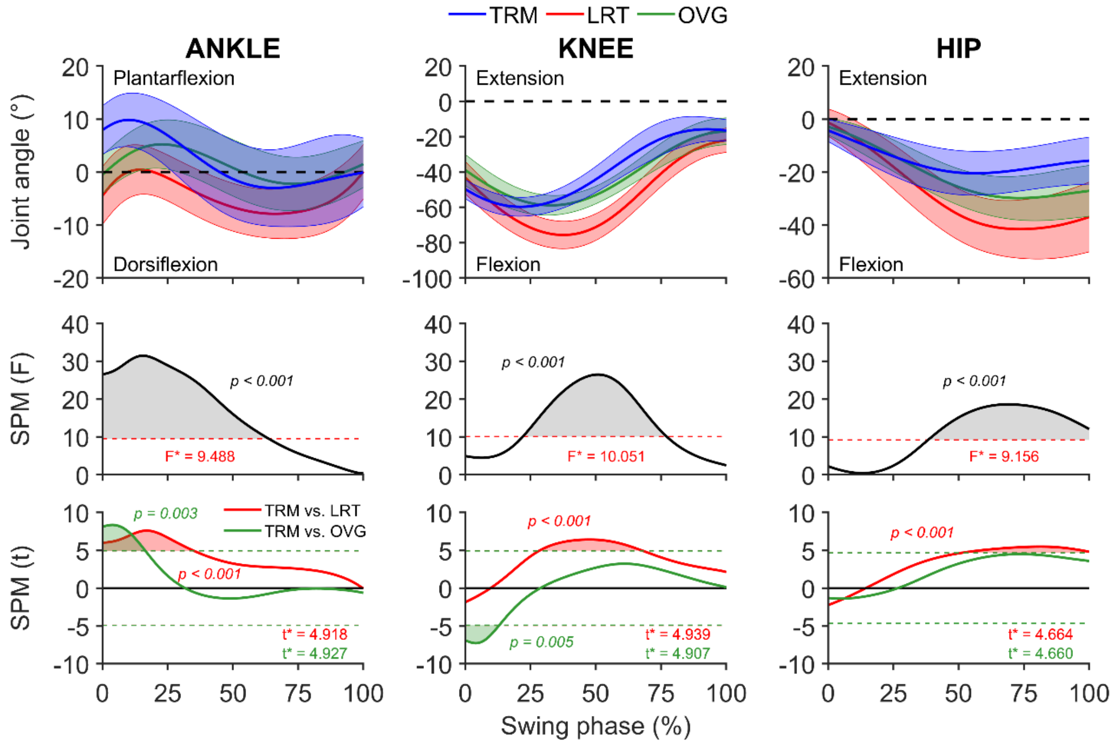

Figure 17: Sagittal plane joint angle kinematics in high perturbation magnitude group (n = 10) and statistical parametric mapping (SPM) analyses of the ankle, knee, and hip joint during the entire swing phase (toe-off to touchdown, 0-100 %,) of the recovery step for the eighth treadmill-based trip (TRM), lean-and-release (LRT), and overground trip (OVG). *1st row*: Joint angle comparison between all three tasks via means ± standard deviation (bold lines and shaded areas) across participants. *2nd row*: SPM one-way repeated measures ANOVA [SPM(F)] and univariate F-statistic (F^*) with significant threshold at 99 % confidence (dashed, red line) with task as factor (TRM, LRT, OVG). The shaded grey areas indicate to significant differences between the three tasks. *3rd row*: post-hoc tests [SPM{t}] comparing pairs of independent joint angle curves (i.e. TRM vs. LRT, red; TRM vs. OVG, green). t-statistic (t^*) with significant threshold at 99 % confidence is shown with dashed, red/green lines, and shaded red/green areas indicate to significant differences between the respective pairs of tasks.

5.5. Discussion

Factors that elicit or limit transfer of stability control adaptations have not yet been thoroughly investigated. The current study tested transfer potential for an established perturbation magnitude (MoS_{low}) previously shown as being sufficient to elicit acute refinements of motor responses (Epro et al., 2018a; 2018b; König et al., 2019a; 2019b), and further examined the influence of an increased perturbation magnitude (MoS_{high}) on fall-resisting skill adaptations performance transfer to other tasks. The general assumption for positive transfer from practised skills to unpractised tasks could be confirmed with both practice groups adapting stability control (skill refinement) from a single session of repeated treadmill-based trip perturbations. Moreover, recovery performance to a non-practised overground trip after treadmill-based practice was enhanced compared to controls indicating that treadmill-based perturbation paradigms have the potential to mitigate fall risk during overground tripping. Concerning the TRM_{low} group, it is worth noting that even though the state of instability observed for treadmill-based gait perturbations was 2.7-fold lower compared to the unpractised overground trip (MoS at timepoint of perturbation on average: -14.5 cm vs. -39.6 cm), participants showed enhanced stability recovery performance after overground tripping compared to the control group. These findings match previous evidence revealing positive transfer of adaptations from practising with lower to higher (transfer) perturbation magnitudes (Patel & Bhatt, 2015). Thus it may be suggested that the central nervous system is capable of rapidly calibrating the required motor action to cope with higher magnitudes of perturbation purely based on information from prior exposure to lower magnitude (Horak & Diener, 1994), i.e. the repertoire of prior experience (repeated increase in BoS).

Contrarily to further hypothesised, an upwards manipulation in practised perturbation magnitude did not lead to enhanced transfer performance in an overground trip, confirming that greater motor errors do not necessarily lead to greater adaptation or inter-task transfer of adaptative changes. As expected, an increased perturbation magnitude during practice led to greater motor errors (lower MoS and BoS in relation to unperturbed walking) for both the first (unpractised) and eighth treadmill trip trials, indicating a generally higher demand on the neuromotor system to execute appropriate recovery responses. Interestingly, although the absolute change in MoS as well as in BoS between the perturbation and subsequent recovery step was higher in TRM_{high} compared to TRM_{low}, it resulted in only slightly different states of stability at recovery step touchdown across trials (Figure 14), and in no group differences in BoS at touchdown of the recovery step. These results suggest that the central nervous system can appropriately calibrate motor responses to the specific perturbation magnitude for the first (non-practised) perturbation, and only fine tunes such for subsequent perturbations. The absence for any magnitude effect on MoS or BoS for recovery touchdown, however, does indicate that recovery responses were executed to the minimum required (i.e. positive MoS) to preserve continuity of walking post recovery. Nevertheless, independent of the practice group the BoS for recovery was higher compared to unperturbed walking for both Trials 1 and 8 during practice. This highlights that exposure to perturbations *per se* elicits an increase in BoS higher than for unperturbed walking to control stability at best, which is potentially a crucial factor for inter-task transfer performance. This would explain the significantly higher BoS at touchdown of the recovery step after perturbation in both practice groups compared to the control group, contributing to achieve a positive MoS (i.e. control of CoM within the boundaries of the BoS) and hence a stable state after recovering from a threat of falling.

In our previous studies we did not detect any transfer of treadmill-based trip resilience to the recovery performance in a lean-and-release task (König et al., 2019b; König et al., 2022) and provided evidence that this might be explained by differences in the task-specific neuro-muscular control of motor output (König et al., 2022). The current study aimed to test the likelihood that a variation in perturbation magnitude during treadmill-based practice (higher when compared to our previous studies) could interfere with transfer performance. Since neither of the practice groups (TRM_{low} nor TRM_{high}) differed to controls in recovering stability after release, it can be suggested that a potentially increased excitability of the motor cortex by an upward manipulation of the perturbation magnitude during practice seems redundant for transfer of trip-resisting skills. Furthermore, looking at the joint kinematic patterns, extensive differences were found of the recovery limb between the eighth treadmill gait trip (the adapted stability performance) and lean-and-release recovery for both low- and high-magnitude perturbation groups. Although both perturbation recoveries started with a flexion in knee and hip joints and were followed by dorsiflexion to allow foot clearance of the swing limb (Eng et al., 1994; König et al., 2022), for both tasks a subsequent knee extension and simultaneous hip extension would serve to further increase step length to establish an upright posture at touchdown. Contrarily, the swing phase for lean-and-release as compared with treadmill perturbation showed a later onset of subsequent knee extension, not only in relative but also in absolute terms (lean-and-release, TRM_{low}, 99 ± 17 ms; TRM_{high}, 92 ± 10 ms vs. treadmill, $TRMl_{ow}$, 69 ± 11 ms; TRM_{high}, 53 ± 16 ms) independent of the magnitude group (F = 53.02, $p < 0.001$). The swing phase was notably characterised by significantly higher flexion of both knee and hip joints for a substantial proportion of the entire recovery step (ranging from ~40 to 89 % across joints and exercise groups (Figures 16 and 17). Such differences in kinematic patterns of an anterior step between tasks may be related to a different nature of initialisation of perturbations. Even though there was similar or even lower MoS at time of perturbation (depending on practice group) between lean-and-release and treadmill trip (indicating a similar/lower state of instability caused by the perturbation), the body is more inclined anteriorly in a lean-and-release task to achieve the initial instability, given that the velocity of the CoM at release is ~ zero. This explains, for the lean-and-release task, both a more dorsiflexed ankle configuration at the beginning of the swing phase caused by the greater initial lean angle and the requirement for higher as well as prolonged knee and hip flexion in order to extend swing leg's foot clearance and eventually increase the BoS. Our data further indicate to differences in task continuity beyond the touchdown of perturbation recovery which might influence recovery steps. Next to the establishment of an upright posture until recovery touchdown after perturbation, the preparation for subsequent weight acceptance and push-off phases is crucial for treadmill-based perturbations since continuity in walking is not only desired but required. This would rather be hindered by a higher hip or knee flexion for treadmill touchdowns as opposed to recovery after lean-and-release with a single step to maintain stable stance after touchdown. When considering the comparison of the two gait perturbations (treadmill vs. overground trip), kinematic differences were either absent (i.e. for the knee joint of the high perturbation magnitude group) or occurred for only ~2-43 % of swing phase (across joints and practice groups) and were generally lower in absolute magnitude (RMSE across groups and joints on average up to 15 °) as opposed to the comparison of treadmill trip and lean-and-release (on average up to 30 °). Altogether, and in line with the findings of König and colleagues (2022), it may be suggested that the degree of contextual task differences lead to different temporal frameworks for the single motor response (i.e. increase in BoS by anterior stepping). Thus investigation of time courses of motor responses may be essential for understanding success or failure of transfer of trip resilience from one task to another.

5.6. Conclusion

In conclusion, this study confirmed that repeated exposure to treadmill-based gait perturbations leads to rapid adaptive changes in stability recovery performance and demonstrated that such perturbation practice paradigms have the potential to elicit transfer of stability recovery performance to non-practised overground trip-like perturbations. However, higher perturbation magnitudes do not necessarily seem to influence transfer performance. Transfer of stability recovery responses and performance from one task to another may partly be subject to the degree of similarity in recovery motor responses between perturbations.

6. Main findings and discussion

Within the scope of integration and application of task-specific perturbation paradigms in both the assessment and practice of fall resilience it is crucial to understand factors enabling the neuromotor system to transfer stability control between different postural perturbations. This thesis focused on different perturbation paradigms that elicit similar stability recovery responses and examined adaptation and transfer of motor control, with the perspective to broaden knowledge about the effectiveness and efficiency of assessment, improvement, and transfer of fall-resisting skills. The following chapters are separated by investigated topics (Studies 1, 2 and 3) and provide insight into the main findings and conclusions derived from this work. Limitations of the studies and practical implications for future investigation will be discussed subsequently.

6.1. Predictive validity in assessment of stability control

The first study of this thesis aimed to examine two commonly used lean-and-release task protocols (maximal vs. predefined, single lean angle) and tested for intra- and inter-session reliability of stability performance parameters on healthy participants from young to old. With a multicentre design chosen to account for a wide range of age groups, multiple investigators, as well as varying methodologies, the study revealed consistency and equivalence between repeated assessments of stability performance within and between sessions using either task protocol, and hence can be used reliably to identify individual performance deficiencies or to classify effectiveness of interventions for stability recovery enhancement. However, previous findings indicated no transfer of practice-induced improvement in trip-resilience during sudden anterior stability loss from a static forward-inclined position (König et al., 2019b). There was therefore reason to doubt that stability control after anteriorly induced perturbations can be generalised. Hence clinical assessment methods, such as the lean-and-release test, would be limited in evaluating and certainly explaining performance for different perturbations. Thus the objective of the second study was to identify to what extent the lean-and-release test explains and can be used at all to assess fall resilience to trip-like perturbations. We therefore examined the relation of common recovery performance parameters between two tasks.

In general, it is important to note that, irrespective of the task, both perturbations caused greater challenges to stability in older compared to younger adults, indicating an age-related deterioration in the ability to increase the base of support rapidly and effectively in the anterior direction (König et al., 2019b; Karamanidis & Arampatzis, 2007; Karamanidis et al., 2008; Werth et al., 2021). This strengthens the assumption that both tasks are characterised not only by a similar result of perturbation itself (anterior displacement of the centre of mass) but also by similarities in subsequent stability recovery responses (anterior increase in base of support by rapid stepping actions) impacting age-dependent stability performance to a similar extent. However, statistical computation on the relation of all analysed stability parameters (margin of stability, base of support, and rate of increase in base of support), including all age-groups, revealed a maximum of only one third of explained variance for the alternative task. Furthermore, we separated participants into subgroups according to their stability recovery behaviour in the lean-and-release task (single- vs. multiple-steppers; Karamanidis & Arampatzis, 2007; Karamanidis et al., 2008; Werth et al., 2021) and analysed subgroup recovery performance during the trip-like perturbation. Although there were significant age-related differences in recovery from the trip-like perturbation, age-matched multiple- versus single-steppers observed during the lean-and-release test demonstrated no differences in stability performance for the gait trip task. Thus, we were able to establish limited transfer of stability recovery not only based on the weak or moderate inter-task correlations but also on sub-group comparisons. Based on these results, we concluded that the lean-and-release test and the treadmill-based trip-like perturbation should not be used interchangeably in clinical settings.

6.2. Determining factors for adaptation and transfer of stability recovery responses

The third study investigated whether increased perturbation magnitude used for practice of stability recovery responses would yet affect transfer to a non-practised lean-and-release test and a non-practised trip-like perturbation during walking overground. Strikingly, improvements in stability performance from a single session of repeated treadmill-based trip perturbations, were revealed in both practice groups (low and high perturbation). Moreover, recovery performance to a non-practised overground trip after treadmill-based practice was enhanced compared to controls indicating that treadmill-based perturbation paradigms have the potential to mitigate fall risk for overground tripping. However, an upwards manipulation in practised perturbation magnitude did not lead to enhanced transfer performance in an overground trip. This confirms previous findings from modelling studies (Wei & Kording, 2009) that greater motor errors do not necessarily lead to greater adaptation or inter-task transfer of adaptive changes.

Bearing in mind the findings obtained in the second study as well as those of König and colleagues (2019b), one may question why the lean-and-release test was chosen again to test for transfer potential of practised trip resilience. The reasons for this are twofold and based on the overall aim of investigating factors that potentially elicit or limit transfer of recovery responses between perturbations. Firstly, we previously failed to detect transferred stability control between the lean-and-release test and both non-practised trip-like perturbations as well as practised trip-resilience using an established perturbation magnitude previously shown being sufficient to elicit acute refinement of motor responses (Epro et al., 2018a; 2018b; König et al., 2019a; 2019b). Therefore, we aimed to examine whether increased perturbation magnitude might influence fall-resisting-skill transfer. Secondly, by including an overground trip in the investigation we were able to compare stability performance during treadmill-based trip perturbations with both the lean-and-release test and a perturbation of slightly altered context compared to the practised one. In this way we increased the chance of detecting similarities or differences in recovery responses, expanding knowledge of transfer potential between perturbations.

In this third study neither gait-trip practice groups showed enhanced stability recovery after release compared to controls. Thus increased excitability of the motor cortex by increase in perturbation magnitude during practice seemed redundant for trip-resisting skill transfer. With regards to similarities or differences in recovery stepping responses, we analysed and compared lower limb kinematics for ankle, knee, and hip joints between perturbations using statistical parametric mapping. There were extensive differences between the eighth treadmill gait trip (adapted stability performance) and lean-and-release recovery for both low- and high-magnitude perturbation groups (ranging between 40 and 89% across joints and practice groups). In contrast, when comparing the two gait perturbations (treadmill vs. overground trip), kinematic differences were either absent or occurred only between 2 and 43% of swing phase (across joints and practice groups). Such differences of an anterior step between tasks may be related to the different initiations of perturbations (static for lean-and-release; during locomotion for gait trip), or indicative of differences in task continuity (stabilise within a single step for lean-and-release; continue walking for gait trip) beyond the immediate touchdown for perturbation recovery. Accordingly, we maintain that trip-resilience cannot be appropriately evaluated by means of the clinical lean-and-release assessment method.

To investigate support for the conclusion stated above, we additionally determined transfer in relation to the degree of similarity in kinematic motor response patterns for all participants who experienced gait trip perturbation practice (n = 20; Figure 18). Therefore, differences between task motor response patterns for both transfer tasks (i.e. lean-and-release and overground gait-trip compared to the eighth treadmill-based gait-trip) were calculated as the individual root-mean-square difference (averaged based on the knee and hip joint angle kinematics) for the entire swing phase of the recovery steps. Transfer potential was calculated as the difference between an individual's margin of stability at touch-down of the recovery step for both transfer tasks (all participants of both practice groups; low and high perturbation magnitude) and an average of the control group (no practice; n = 10). These calculations (Figure 18) confirm that the potential for transfer of stability performance from one task to another is partly subject to the similarity in recovery motor response patterns between perturbations. This, in line with the most recent findings of our group (König et al., 2022), provides evidence that recovery responses for a lean-and-release test and a trip-like perturbation have underlying different neuro-muscular activity patterns. However, it must be pointed out that our analyses provided in Figure 18 also indicate that transfer of adapted stability recovery responses to unpractised perturbations cannot be explained purely by similarities in kinematic responses, suggesting that motor transfer may yet be characterised by high inter-individual variability and underlying multifactorial dependencies.

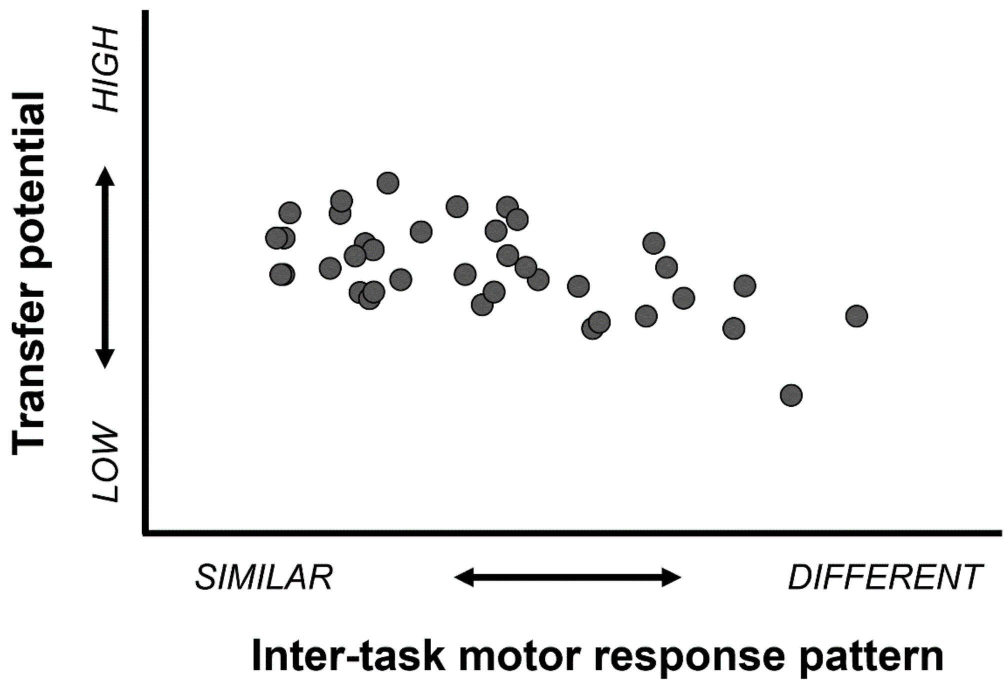

Figure 18: Illustration of the relationship between the extent of similarity in motor response patterns and transfer potential. Each point represents data for an individual participant (*n* = 20) who underwent repeated treadmill-based perturbation practice investigated in Study 3. Inter-task motor response pattern is expressed as the individual root-mean-squared difference calculated for the entire swing phase of the recovery step for both transfer tasks, i.e. lean-and-release as well as overground gait-trip, compared to the eighth treadmill-based gait-trip. Transfer potential is expressed as the difference between the individuals' margin of stability (all participants of both practice groups) and an average over the control group (no practice; *n* = 10) at touch-down of the recovery step for each of the transfer tasks.

6.3. Limitations

6.3.1. Lean-and-release method

For the first study using the lean-and-release test in different research centres to examine the reliability of stability performance parameters, the attachment of the supporting cable to the body differed between the three research centres. This was found to affect standardisation of the static inclined position prior to cable release if determined via a pulling force equivalent to a predefined percentage of body mass. In particular, when performing the single lean angle protocol, participants had to lean more anteriorly (more unstable body configuration at task initiation, i.e. cable release) for a cable secured at chest-level compared to pelvis-level, and hence older participants were not able to subsequently recover stability with a single step. This was a drawback in analysing the desired stability performance parameters (margin of stability and base of support at touchdown of recovery steps) in a standardised manner, as these are affected differently between single and multiple stepping responses. Thus, when determining the lean angle via a monitored pulling force, one should ensure that the attachment is in accordance with the methods provided within respective studies. Since the initial percentage of body mass was set lower when performing the maximal compared to the single lean angle protocol (10% compared to 23%), it was yet possible identifying the maximal lean angle which could be recovered with a single step and analysing respective stability performance even in older adults. However, based on our current study design, we cannot exclude that for different populations (the frail, the elderly), an attachment at chest level equivalent to 10% of body mass would have been too demanding in the first place. To improve standardisation between repeated measurements in future investigations, alternative methods to determine the lean angle should be examined and it should be noted that defining the initial position prior to release via angle measurement in degrees has been shown to be less reliable (Ringhof et al., 2019). Instead, given that the margin of stability is widely used to assess stability recovery performance after cable release, one might also consider using it to position participants at the desired static inclination.

On a similar note, our evidence-based approach to determine the initial lean angle via predefined percentages of individual body mass does not consider body height. Thus large variations in the position of the CoM within and between subject groups would affect the MoS measured at release (i.e. more anterior position of the vertical projection of the CoM in relation to the BoS for participants characterised by a higher position of the CoM compared to those, matched to body mass, but with lower position). However, even if we assume differences of 5 cm, the variation in the vertical projection of the CoM will be less than 1.5 cm when considering the analysed lean angles. Such a difference would be small in relation to the MoS (i.e. due to the incorporation of eigenfrequency for calculation of the extrapolated CoM) and hence not functionally relevant considering the forward-inclined leaning angles used for this PhD. Accordingly, no significant group-differences (if applicable) in the MoS at cable release were detected between in any of the three presented studies.

Regarding the transfer analysis for treadmill-based gait trips to an unpractised lean-and-release task (Studies 2 and 3), one might argue that, especially for younger adults, the chosen predefined inclination (23% or 33% of body mass) was not challenging enough to identify any transfer effects. In contrast, a maximal lean angle approach would have been more suitable. However, one must firstly bear in mind that a significant number of middle-aged and older adults (40% of the middle-aged and 90% of the older) were, although instructed to do so, not able to recover from such predefined lean angles with a single step (Study 2). Secondly, both study designs (Studies 2 and 3) were conceived to assess transfer of recovery performance to non-practised perturbations. As described in the first study, the determination of the maximal lean angle would have required several exposures to sudden anterior stability loss that would have potentially elicited task adaptation. Finally, the margin of stability prior to the recovery step (challenge on stability control) was similar between tasks (on average for all analysed subjects in Study 2: -0.14 ± 0.08 m at timepoint of cable release; -0.15 ± 0.09 m at perturbed step), which altogether indicates that a maximal lean angle approach would not have significantly strengthened the reported outcomes.

6.3.2. *Analyses of stability criteria*

Throughout the three studies, a simplified kinematic model was used to calculate the main parameter determining stability performance, i.e. the margin of stability, where the anteroposterior position of the centre of mass is defined by the trochanter markers and its velocity calculated from both trochanter markers and a marker placed on the seventh cervical vertebra. This is based on the work of Süptitz and colleagues (2013) who validated such a reduced kinematic model when testing it against a twelve-segment, full-body model (26 markers; on average $r = 0.90$, $p < 0.01$], finding no differences in analysed parameters (extrapolated centre of mass and margin of stability) between the two approaches. Note that this was done for the assessment of recovery performance during unperturbed and perturbed treadmill walking for adults across the lifespan, using a similar treadmill-based perturbation paradigm like that for the present thesis. With regards to the determination of dynamic stability control during overground locomotion, a recent study (Havens et al., 2018) showed that using simplified models can lead to overestimates in the margin of stability. However, if the motion of the trunk was included, the differences in the estimated centre of mass motion and margin of stability between the simplified model and full-body model were considerably decreased for overground walking. Note that throughout the current three studies the applied reduced kinematic model incorporated the trunk.

In a similar context, the inverted pendulum model itself might have partially been invalid for our experimental designs. Other than for unperturbed walking, the pendulum length (distance between axis of rotation and centre of mass) may not remain constant during perturbations due to changes in lower limb joint angles. This may result in an alteration of pendulum mechanics. However, McCrum and colleagues (2014) as well as Süptitz and colleagues (2013) reported no substantial pendulum length changes during treadmill-based gait trip-perturbations, and it can hence be suggested that the current perturbation paradigms did not cause any drawbacks for calculation of stability performance parameters.

6.3.3. Study population

Finally, it is important to mention that the third study was performed only on young- and middle-aged adults and we cannot therefore draw any conclusion about transfer to non-practised overground trips in older adults or different populations. Combining the work of our group (König et al., 2019b; 2022) with our findings regarding factors eliciting or limiting transfer of stability performance across perturbations (the second and third studies) allows us to develop thinking further. We postulate that the ability to recover stability and moreover transfer of improved stability control across various anteriorly directed perturbations rather depends on the specificity of contexts and constraints related to respective motor tasks. Nevertheless, to endorse both the current findings and those provided earlier (König et al., 2022), we need to elaborate on neuromuscular control (muscle synergies) during overground trips.

6.4. Practical implications

The outcomes of this thesis shed new light on factors enabling the neuromotor system to transfer stability control between various postural perturbations. This may have a major impact on the conceptualisation and implementation of future frameworks aimed at assessing and mitigating fall risk.

6.4.1. Assessment of fall risk

Being a well-established assessment method to assess stability performance after unexpected perturbations for the past decades, the lean-and-release test has been conducted on cohorts of various ages and health conditions. Findings notably indicate deterioration in stability performance after a sudden loss of anterior stability as predicting future falls in older populations (Carty et al., 2015). They also indicate positive effects elicited by stability interventions. For the first study incorporated of this thesis, we confirmed that the forward lean-and-release test is a reliable assessment of stability recovery performance parameters in adults over a wide age range, independent of the lean angle protocol used. Our results further strengthen use of exposure to stability loss from a single predefined lean angle, as this protocol is quicker, less demanding and could be especially beneficial for testing stability recovery performance in clinical settings. Nevertheless, when conducting a single lean-angle protocol, the choice of task familiarisation needs careful consideration. In line with previous studies (Aragão et al., 2018; Arampatzis et al., 2008; Carty et al., 2012; Karamanidis and Arampatzis, 2007; Mademli et al., 2008), the participants in the current study practised three times with a slightly lower inclination prior to the first trial that was measured. Given that previous findings (König et al., 2019b) indicated potential single-trial adaptations in this assessment task, performing a few practice trials prior to the first measured exposure to sudden anterior loss of stability is highly recommended if pre-post designs (repeated assessments) are used. Based on the current study that revealed that assessed stability performance parameters are reliable without performing any further practice trials prior to the second (post 0.5h) and third test (post 48h), follow-up measurements in such pre-post designs would not be affected by potential adaptations to further assessments.

6.4.2. Practice and transfer of stability recovery responses

In line with several previous studies of our team (Epro et al., 2018b; König et al., 2019b; McCrum et al., 2018; Süptitz et al., 2013), the treadmill-based trip-like perturbation paradigm used throughout the second and third studies has once again been shown to be effective and efficient (~25 minutes) in eliciting adaptive changes in stability performance. A major finding of the current thesis is that such task-specific practice has the potential to elicit transfer to unpractised overground trips. Interestingly, one particular practice perturbation magnitude, as successfully established for healthy adults across the lifespan (e.g. Epro et al., 2018b; König et al., 2019b) and for stability-impaired patients (McCrum et al., 2018), seems to be sufficiently similar in outcome for both adaptation and transfer to higher magnitudes. Hence this perturbation magnitude may be suitable for various populations (the frail, those with pathologies) with limited tolerance of higher perturbation doses and magnitudes in both clinical and research settings. We believe we have established that assessment of common stability performance parameters is reliable using a clinical assessment method (lean-and-release) and that it can be used in the future to assess recovery performance after anterior perturbations in a variety of population groups, specifically the ability to effectively increase the base of support. We also confirmed previous evidence (Rosenblatt et al., 2013; Kiss et al., 2018) that, although the motor response to recover stability loss is used in a variety of anterior perturbations, this does not mean *per se* that it is transferable. Based on the current evidence it seems promising to focus on specific perturbations (e.g. trips or slips), reflecting the most common causes of fall incidence, in order to improve motor outputs responsible for recovering stability. However, such task-specific studies should not be seen as the sole orientations for the most effective mitigation of falls, as the latter would imply that all types of perturbations that a person could possibly be exposed to in daily life must be practised. Since the control of motor responses seems to partly influence transfer, it may be assumed that practising different perturbation tasks – that do not necessarily mimic specific challenges to stability in daily life – eliciting similar recovery response patterns could, in turn, also elicit transfer. Thus more research is required to understand the factors underlying transfer of learned motor skills to unpractised, unexpected challenges of stability and hence to establish adequate fall prevention assessments and programs that effectively cover real-world scenarios (Harper et al., 2021).

6.5. Conclusions

This thesis provides evidence that practising stability control during a single session of repeated treadmill-based gait-trip-like perturbations enhances human resilience to a non-practised overground trip. Such enhancement does not appear to be influenced by an increase in perturbation magnitude during practice but is rather elicited by the extent of similarity in motor response patterns between tasks of perturbed locomotion. In contrast to this, we confirmed that, despite similarity of recovery execution, operation of the stability recovery responses might yet be specific to the nature of the task, which could be one factor limiting transfer between anteriorly directed perturbations.

7. Appendix: data collection/processing across studies

7.1. Study 1: Full body kinematic model approach

To determine the CoM dynamics in Study 1 the researchers across all study centres used a full body marker set consisting of 21 retroreflective markers. The marker locations defined specific body segments, i.e. foot = calcaneus, lateral malleolus, and second metatarsal bone; shank = lateral malleolus and lateral epicondyle of the femur; thigh = trochanter and lateral epicondyle of the femur; trunk = 7th cervical vertebra and the trochanter; upper arm = acromion and lateral epicondyle of the humerus; lower arm = lateral epicondyle of the humerus and hand; hand = the centre of the radius-styloid process and ulna-styloid process; head = the circumference defined bilaterally, anterior and posterior (Fig. 2). Body segments were subsequently compared with existing reference data from cadaver bodies by Demster and colleagues (1959), and thereby segmental masses. The locations of each of the segment CoMs, and the body CoM were calculated using a customised MATLAB script. The BoS was determined as the distance between the heel marker of the trailing limb and the anterior boundary of the recovery foot. As we only used one marker placed on the shoe (at the level of the second metatarsal bone), we additionally measured the circumference of the shoe prior to assessments. We could therefore post-process the distance from the marker to the anterior boundary of the recovery foot and thereby account for the entire BoS used to control stability after sudden anterior perturbation.

Based on these components of dynamic stability, the MoS was defined as the difference between the anterior boundary of the BoS and the extrapolated CoM, calculated in accordance to Hof and colleagues (2005) from

$$\mathrm{X}_{CoM} = P_{CoM+}\frac{V_{CoM}}{\sqrt{\frac{g}{L}}};$$

with

P_{CoM} the anteroposterior component of the vertical projection of the CoM to the ground (from the average of the left and right trochanter markers);

V_{CoM} the anteroposterior velocity of the CoM;

g gravitational acceleration; and

L the reference leg length (defined by the distance between the trochanter and the centre of the lateral malleolus).

7.2. Studies 2/3: Reduced kinematic model

For Studies 2 and 3, we used a reduced marker set model (5 retroreflective markers) to examine stability control for all tasks (gait perturbations during treadmill and overground walking, lean-and-release). The reduced marker model was time-efficient, and we mitigated problems such as multiple marker loss and artefacts (especially during gait trips in pilot studies) that accompanied use of either marker clusters or full-body marker sets. This model to assess CoM dynamics has been revealed as being sensitive enough to detect e.g. adaptation phenomena in MoS due to repeated practice (Epro et al., 2018b; König et al., 2019b), retention phenomena over a period of 1.5 years (Epro et al., 2018b; König et al., 2019b), as well as age-related deterioration in stability performance (e.g. Bosquée et al., 2021; König et al., 2019b) in both - treadmill gait perturbation as well as lean-and-release tasks. Moreover, we found that the application of this reduced model is reliable for unperturbed walking (for analyses of 12 consecutive steps/subject on the treadmill as well as three consecutive overground walking trials; 12 steps per subject in total). Thus for both studies we decided to use only five markers to determine stability control dynamically - placed on the seventh cervical vertebra and bilateral trochanters (defining the trunk) and centres of the big toes (defining the boundaries of the BoS). The extrapolated CoM was calculated based on Süptitz and colleagues (2013) using

$$\mathrm{X}_{CoM} = P_{CoM+} \frac{\frac{1}{2}(V_{CoM} + V_{C7}) + |V_{BoS}|}{\sqrt{\frac{g}{L}}};$$

with

P_{CoM} the anteroposterior component of the vertical projection of the CoM to the ground (average of left and right trochanter markers);

V_{CoM} the anteroposterior velocity of the CoM;

V_{C7} the anteroposterior velocity of the C7 marker – accounting for trunk kinematic;

V_{BoS} the anteroposterior velocity of the BoS;

g gravitational acceleration; and

L the reference leg length.

V_{BoS} was calculated from the average velocity of the toe markers during the stance-phase of unperturbed treadmill walking to account for the belt velocity). This was redundant for lean-and-release and overground trials. L was defined by the distance between the right trochanter and the centre of the right lateral malleolus and was measured prior to the assessments.

In contrast to Study 1, the BoS was defined as the distance between the anterior boundaries of the leading and trailing feet (i.e. the difference between the projections of the two toe markers). Note that we used four additional markers in Study 3 (i.e. bilateral lateral femur epicondyles and lateral malleoli), which were required for collection and processing of joint angle kinematics for recovery steps for all perturbation tasks.

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