BACHELOR THESIS

THE ADAPTATION OF BALANCE OF ABLE-BODIED HUMANS WHEN EXPOSED TO BALANCE PERTURBATIONS

Kylar de Boer

BIOMECHANICAL ENGINEERING
BIOMECHATRONICS AND REHABILITATION TECHNOLOGY

EXAMINATION COMMITTEE
dr. E.H.F. van Asseldonk
dr. M.E. Eveld
dr. H. Wang

DOCUMENT NUMBER
BE - 947

UNIVERSITY OF TWENTE.

12-07-2023

Summary

Ankle-exoskeletons are devices that can be used for balance assistance. To optimize the usage of these exoskeletons, it needs to be known how people adapt to exoskeleton balance assistance. Before it is possible to do that, it first needs to be known how people adapt their balance in general, without exoskeleton assistance, to external perturbations. Therefore this study investigates the adaptation of balance to unexpected forward pushing perturbations. Four participants received random anterior pushing perturbations at toe-off while walking. The perturbations had a magnitude of 12% of the body weight for four training trials of 40 perturbations. Before and after the training trials there was a trial of 8 perturbations of the same magnitude. To look for generalization of adaptation there were also two trials, pre- and post-training, with 16 perturbations including 8 perturbations with a magnitude of 8% and 8 perturbations with a magnitude of 16% of the bodyweight. The centre of mass (COM) velocity at heelstrike after the perturbation did not change significantly when comparing pre-and post-training trials. There was also no significant adaptation of balance when looking at the generalization of adaptation. It can be concluded that there is no adaptation of balance for the perturbations of this study when looking at the COM velocity.

Samenvatting

Enkelexoskeletten zijn apparaten die gebruikt kunnen worden voor het assisteren van balans. Om het gebruik van deze exoskeletten te kunnen optimalizeren is het nodig om te weten hoe mensen zich adapteren wanneer ze de exoskeleten gebruiken. Voordat dit mogelijk is, moet eerst worden onderzocht hoe mensen hun balans in het algemeen adapteren. Daarom word in dit onderzoek de adaptatie van balans van onverwachte voorwaartse duw perturbaties onderzocht. Vier proefpersonen ontvingen random anteror duw perturbaties op het moment van toe-off terwijl ze liepen. De perturbaties hadden een kracht van 12% het lichaamsgewicht voor vier training-trials. De training-trials hadden elk 40 perturbaties. Voor en na de training-trials was er nog een trial met 8 perturbaties met dezelfde kracht als de training-trials. Om te kijken of er ook generalisatie van de adaptatie plaatsvindt zijn er ook twee trials met 16 perturbaties waarvan 8 met een kracht van 8% en 8 met een kracht van 16% het lichaamsgewicht. De snelheid van het zwaartepunt van het lichaam tijdens heelstrike veranderde niet significant tussen pre- and post-training trials. Er was ook geen sprake van een significant verschil tussen de pre- and post-training trials van de generalisatie trials. De conclusie van dit onderzoek is dat er geen adaptatie van balans is voor voorwaartse duwperturbaties wanneer er gekeken wordt naar de snelheid van het zwaartepunt.

Contents

1 Introduction

Exoskeletons are devices that are used for many different applications within the health sciences. They are already useful as a tool to help with gait rehabilitation after an incomplete spinal cord injury [1]. Exoskeletons can also be used to improve muscle forces in patients with brain conditions like Parkinson's disease [2] and stroke [3, 4]. Another use for exoskeletons is assisting with lifting and lowering tasks by reducing lower back loading and hip extensor torque [5].

1.1 Exoskeletons for balance assistance

People with neuromuscular impairments (e.g., individuals with spinal cord injury, stroke survivors, or the elderly) often have trouble with walking and maintaining balance. These types of conditions result in a higher number of falls as a result of balance loss [6, 7]. To lower the number of falls it is a possibility to wear an exoskeleton for balance assistance. For example, a lower limb exoskeleton has been shown to improve the balance of people with incomplete spinal cord injuries while standing [8] and while walking without external perturbations [9]. There are also ankle exoskeletons that can assist with balance. A study from Afschrift et al. investigated multiple controllers for an ankle exoskeleton for balance assistance after unexpected perturbations. The ankle exoskeleton was effective in supporting balance for pushing perturbations as evidenced by a reduction of the forward centre of mass (COM) movement and a reduction of muscular activity to counter the pushing perturbation. The ankle exoskeleton was also effective in supporting balance for pulling perturbations because of a reduction of the backward centre of mass movement and a reduction in muscle activity [10]. A study by Bayon et al. developed a controller for an ankle exoskeleton that assists balance by providing assistive torques that counteract the reaction to forward perturbations in the anterior direction. This resulted in reduced muscle activity of the soleus, gastrocnemius medialis, and the gastrocnemius lateralis [11].

1.2 Adaptation to exoskeletons

While using the exoskeleton for longer periods of time, the user gets used to the exoskeleton. This results in some form of adaptation. A study from Poggensee and Collins found that after only twelve minutes of walking with an ankle exoskeleton using generic assistance, the metabolic cost was already reduced by approximately 10%. After four hours of training the metabolic cost reduced by approximately 28%. The adaptation reached a steady state around 109 minutes of training [12]. So it is already known that the human body adapts to exoskeleton walking assistance. However, no research is done about how humans adapt when walking with an ankle exoskeleton that assists balance during perturbations (i.e., exoskeleton balance assistance). Insight into the extent of this adaptation makes it possible to develop better controllers for these exoskeletons. It is also useful for developing training regimes for the usage of the exoskeletons.

1.3 Adaptation to balance perturbations

To be able to know how people adapt to exoskeleton balance assistance, it is first important to know how people adapt to balance perturbations without this exoskeleton. Adaptation to perturbations without wearing an exoskeleton can then serve as a baseline before studying adaptation to perturbations while wearing an exoskeleton.

Multiple studies investigate balance and the adaptation of balance to unpredictable or predictable balance perturbations.

A study from Wang et al. considered how young adults adapt their balance when exposed to tripping perturbations. There were eight tripping trials where the participant received one unannounced tripping perturbation. They found that the COM velocity at heelstrike reduced when comparing the last trip with the first trip [13].

A study from Schmid et al. investigated how people adapt their balance while they are standing on a platform that continuously moves in a sinusoidal fashion, with two different frequencies, along the sagittal plane. With both the high and low frequency the muscle activity of the tibialis anterior and soleus spiked at the beginning of the perturbation. After that, the activity quickly reduced until it reached a steady state. The amplitude of the oscillations of the COM remained invariant with the high-frequency tests. For the low-frequency tests the COM cyclic displacement was approximately the same as for high-frequency in the beginning, but increased after the first few cycles. This can be explained by the phenomenon that the body starts to passively sway with the movement of the perturbation instead of fighting against it [14]. A study by Bhatt et al. investigated the adaptation of balance when people experience tripping and slipping perturbations. The protocol for the experiment is shown in Figure 1. For the slipping perturbations there was adaptation of balance denoted by fewer falls and less backward balance loss after the training. This adaptation was in the form of a bigger forward COM velocity after the slipping perturbation to counteract the backward balance loss. For the tripping perturbation there was also adaptation of balance denoted by fewer falls and less balance loss after the training. This adaptation was in the form of a smaller forward COM velocity after the tripping perturbation [15].

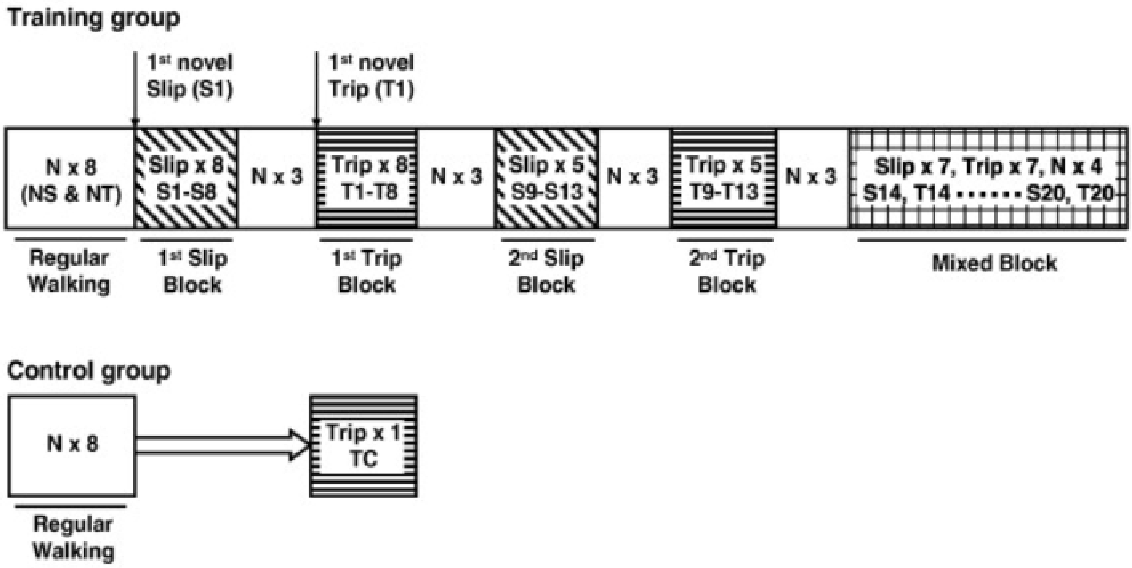


Figure 1: The perturbation training protocol used by Bhatt et al. The participants of the training group had four trials where they experienced tripping or slipping perturbations and one trial with mixed perturbations. In between these trials there were trials without any perturbations. The participants adapted to the repeated exposure to slips and trips shown by fewer falls in the Mixed training after the training blocks [15].

Wang et al. investigated the adaptation to obstacle-induced trips during gait. The study had two purposes. The first purpose was to see if an obstacle-induced trip training regime reduced trip-induced falls during gait. The second purpose is to examine the retention of the induced effects of this training on an immediate basis. The training protocol that was used during the experiments is shown in Figure 2. The participants showed proactive and reactive adaptation to tripping balance perturbations. After training the number of falls after the perturbation decreased drastically. The adaptation is also visible in the reduced forward COM velocity, larger toe clearance, and reduced forward rotation at the end of the recovery step. Additionally, the reactive stability, trunk angle, and recovery step length were all high for the first perturbation and then decreased for the later perturbations [16]. This effect is called the 'first trial effect' in this study.

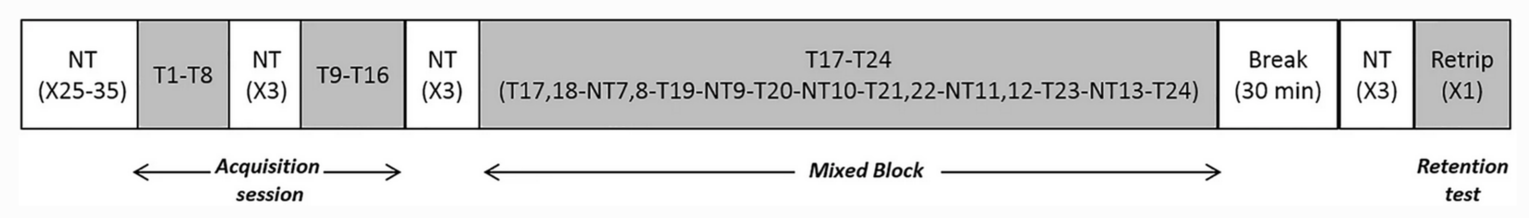


Figure 2: The perturbation training protocol used by Wang et al. The participants of the training group had multiple trials where unexpected trips were induced. The participants adapted to the repeated exposure to trips, shown by fewer falls in the Mixed block when compared to the first block of trips [16].

1.4 Knowledge gap

While there are multiple studies that investigate the adaptation of balance to various types of perturbations, to date there are no studies that look at the adaptation of balance while walking and experiencing unpredictable anterior perturbations applied to the COM. It is important to investigate these forward pushes while walking because exoskeletons can be used in the future to assist with these types of perturbations. In order to study adaptation to exoskeleton balance assistance in the future, it is first important to look at the adaptation of balance while walking and experiencing unpredictable perturbations without exoskeletons. Therefore the research question for this work is: How does a person adapt their balance when they are exposed to repeated unexpected COM perturbations in the anterior direction during gait without external assistance?

To be able to study the adaptation of balance to forward perturbations it is first important to understand what recovery strategies the body uses when reacting to these perturbations. According to Leestma et al. there are two recovery strategies when the COM is perturbed. The first strategy is the stepping strategy. This strategy is primarily used when experiencing large perturbations. When such a perturbation is detected the placement of the foot is changed to enable a large change in the centre of pressure relative to the centre of mass. This causes the ground reaction force lever arm to increase. The second strategy is the ankle strategy. This strategy is primarily used when experiencing small perturbations. Modulating the ankle torque of the stance leg causes a small shift in the centre of pressure and the balance is regained [17]. The COM state, centre of pressure and the torque around the ankle are thus important parameters when maintaining balance. The COM state can be used to measure if people maintain their balance. So to measure balance loss, the COM state is a useful metric. The COM state can be displayed as the COM position or the COM velocity. The COM velocity is the change of COM position and thus tells something about how fast the COM moves. This metric is used in many articles and is also for that reason chosen as the main outcome metric of this study [13, 14, 15, 18].

1.5 Objectives

To answer the research question, this study has three objectives, one primary objective and two secondary objectives.

The primary objective is to assess to what extent people adapt their balance to unexpected forward perturbations. Multiple studies found a reduction in forward COM velocity after exposure to repeated trip perturbations [13, 15, 16]. The direction the body moves to during a trip is similar to a pushing perturbation in the anterior direction. It is thus hypothesized that the COM velocity post-training will reduce during the experiment.

The first secondary objective is to look at the first trial effect during the experiment. It is possible that the adaptation already happens during the first perturbations of the experiment.

Wang et al. found that the reactive stability was the highest at the first tripping perturbation and decreases at the later tripping perturbations [16]. Participants thus reacted strongly to the first perturbation and less strongly to the later perturbations. It is thus hypothesized that the COM velocity of the first perturbation will be the highest and the remaining perturbations result in a lower COM velocity.

The second secondary objective is to look at the transfer of adaptation with perturbations of different magnitudes than the main perturbations. The perturbations during the experiment have a specific magnitude. It might happen that possible adaptation to these perturbation transfer to perturbations of a different magnitude. A study by Bhatt et al. investigated the adaptation of balance to both slipping and tripping perturbations. They found that people also adapt their balance when exposed to two different types of perturbations. They found that the COM velocity decreases during training when exposed to the tripping perturbations [15]. The tripping perturbation is in a direction similar to the forward-pushing perturbation. It is thus hypothesized that the COM velocity post-training will also decrease when exposed to perturbations of a different magnitude.

2 Methods

2.1 Participants

Four able-bodied participants took part in this study. Two of these participants were male and two were female (age 21±0.6 yrs, height 180±8 cm, and mass 73.4±15 kg). The setup and experiment have been approved by the university ethics committee. All of the participants were given information about the experiment beforehand and signed a consent form.

2.2 Pusher device

During the experiments, participants walked on a dual-belt, force-instrumented treadmill (custom Y-Mill, Motek medical, Culemborg, The Netherlands). To prevent falling there are handrails at the side of the treadmill and the participants wear a safety harness which is attached to the ceiling. To induce perturbations there is a pusher device (Moog, Nieuw-Vennep, Netherlands). This device is positioned posterior to the participant behind the treadmill to push in the anterior (forward) direction. The pusher device is attached to a brace that the participant wears around their hips (i.e., approximately at the participants' COM). The pusher device gives perturbations with a strength that is proportioned to the participants' weight. During this experiment perturbations of 150 ms duration with a strength of 8%, 12%, and 16% of the participants' body weight are given. The experimental setup is outlined in Figure 3.

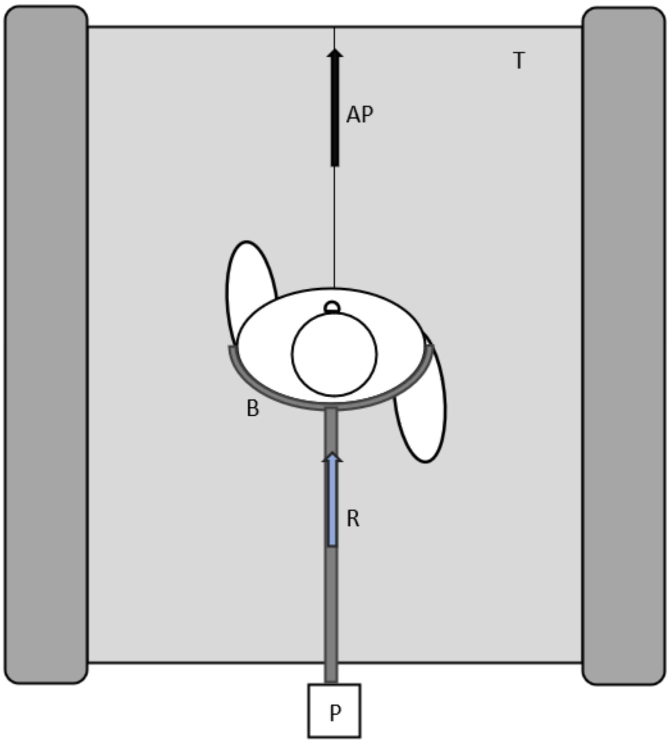


Figure 3: The experimental set-up. A pusher device is used to give perturbations to a participant that walks on a dual-belt, force-instrumented treadmill. B, pelvic brace; P, pusher device; R, a rod that connects the pusher device with the pelvic brace; T, dual-belt force-instrumented treadmill.

2.3 Protocol

Before the start of the experiment the participants are given a short introduction and explanation of the experiment. They are informed that they are going to walk on a treadmill at a slow pace and that during this walking they will experience small, random pushes of magnitudes that are not intended to induce falls. They are instructed to try to walk as normally as possible throughout all the trials. A general timeline of how long everything, including the preparations, is going to take is provided. Then an explanation of the placement of all the sensors and markers is given. Additionally, before every trial an estimate of the time they are walking is given.

First, the participant's mass, height, and leg length are measured. The leg length is the length from the base of the foot until the greater trochanter. Second they are instructed to walk on the treadmill for approximately one minute to familiarize themselves with the walking speed of $0.63\sqrt{l}$ m/s where l is the subject's length of the leg.

Third, EMG sensors are placed on the appropriate muscles. When the sensors are placed correctly the maximum voluntary contractions (MVCs) are measured. Fourth, the markers for the motion capture are placed on the body. To be able to process the data, the participant first needs to perform some tasks before the actual experiment begins. First, there is a static measurement where the participant stands still with all the markers visible without wearing the brace around the hips. This measurement is repeated while wearing the brace. Lastly, the participant is asked to walk two steps forward and two steps backward. With this last measurement, a model is made that automatically labels the markers while recording. Lastly, the harness is put on.

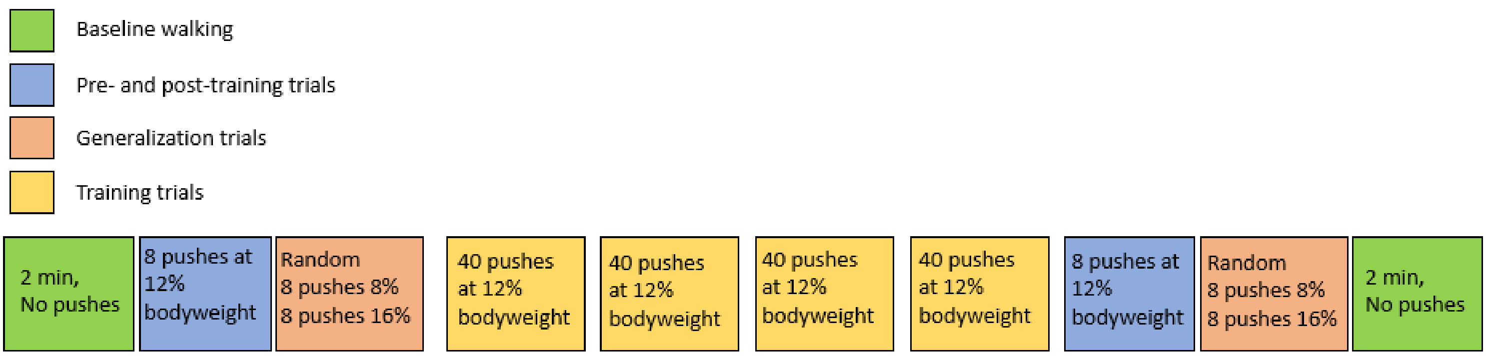


Figure 4: Protocol for the experiments. There are two baseline walking trials before and after the perturbation trials. Before the training trials of 40 perturbations with a strength of 12% body weight, there are two trials. The first trial is a trial with 8 perturbations with a strength of 12%. The second trial is a generalization trial with two times 8 perturbations with strengths of 8% and 16%.

To see if the participants adapt their balance to perturbations, a training period is necessary where they experience a substantial amount of perturbations. Before and after these training trials there are smaller trials that measure their responses before and after they possibly adapted their balance. The protocol is outlined in Figure 4. Before the trials the participant walks for two minutes without perturbations. After that, there is a pre-training trial where there are eight perturbations with a strength of 12% of the mass of the participant. The perturbations are randomly induced, every 6 to 12 steps, at toe off position of either the right or left foot. Toe-off is the moment only the toes of a foot touch the ground, right before taking the foot off the ground. After that, there is a pre-generalization trial where there are eight perturbations with a strength of 8% and eight perturbations with a strength of 16%. The strength is decided on a semi-random basis. Every perturbation it is decided whether the magnitude is going to be 8% or 16%. If the past two perturbations were already of the same strength, the next perturbation is automatically of the other strength. After these first two trials the participant gets a small break of approximately two to three minutes. Then it is time for the training trials. There are four training trials where the participant receives 40 perturbations with a strength of 12%. After every condition, the participant gets a break of two to three minutes. When the four training trials are finished the next trial is the post-training trial with eight perturbations of 12%. After that, there is the post-generalization trial with semi-random eight perturbations of 8% and eight of 16%. Lastly, the participant again has to walk for two minutes without perturbations being induced. After this, the harness, sensors, and markers are removed and the participant is done with the experiment.

2.4 Data collection

To measure the kinematics of the participants an 8-camera motion capture system and two video cameras are used (Qualisys, Göteborg, Sweden). The motion capture system makes use of 73 markers that are located on the bony parts and body segments of the body and the pusher device. The necessary markers for the calculations of the estimated COM are the markers on the right and left anterior superior iliac spine and the markers on the right and left posterior superior iliac spine. Muscle activity is also measured with electromyography sensors (Bagnoli, Delsys, Natick, MA, USA). These sensors are placed on the soleus, gastrocnemius, and tibialis anterior. In this study, the forces that are measured by the force plates and the muscle activity are not used. These were still measured because they are being used in possible future studies using the data of these experiments.
The kinematic data were collected at 100Hz with the motion capture cameras and Qualisys Track Manager software, QTM (Qualisys, Göteborg, Sweden). The analog data from the force plates and EMG was collected at 2000Hz. This data was also collected with QTM and is synced with the kinematic data. The pusher data was collected at 1000Hz in Twincat.

2.5 Data processing

With the QTM software the markers were labeled and gaps of missing labels were interpolated with relational gap-filling. The data was then processed and synchronized in MATLAB 2023a (Mathworks, Natick, MA, USA). The marker data is filtered with a fourth order low-pass Butterworth filter with a cutoff frequency of 20Hz. The data of the force plates are filtered with a lowpass Butterworth filter of the fourth order with a cutoff frequency of 40Hz. The data is parsed into individual perturbations, from 0.5 seconds before the perturbation until 2 seconds after the perturbation.

With the data the COM is estimated by taking the mean of the position of the markers on the right and left anterior superior iliac spine and the marker on the right and left posterior superior iliac spine. The derivative over time of the COM position is calculated for all trials to get the velocity of the COM. The moment when the heel hits the ground after the perturbation (heelstrike) is determined and the COM velocity at heelstrike is then calculated. The generalization trials consist of perturbations with two magnitudes, 8% and 16%. These trials are first split up into the two different strengths. The median, 1^{st} quartile, 3^{rd} quartile, minimal value, and maximum value are calculated to make a boxplot. To examine if the differences between the pre-and post-training trials for the normal and generalization trials are significant, a paired t-test is used.

3 Results

To determine if people adapt their balance when exposed to repeated forward-pushing perturbations, three objectives are being examined. The primary objective is to evaluate to what extent people adapt their balance to unexpected forward perturbations. The first secondary objective is to evaluate the first trial effect during all the trials. The second secondary objective is to evaluate the generalization of the adaptation. During the experiments with Participant 2 there were technical issues with the equipment, which caused the data to be unusable. The training trial data of Participant 1 were not saved during the experiments; therefore, the COM velocities from the training trials are not reported for Participant 1.

Participants' COM velocity initially responded in phase with the perturbation, then increased after the perturbation ended to reach a maximum around the point of heelstrike. This trend can be seen in Figure 5, which shows the COM trajectories for perturbations of Participant 1's pre- and post-training trials.

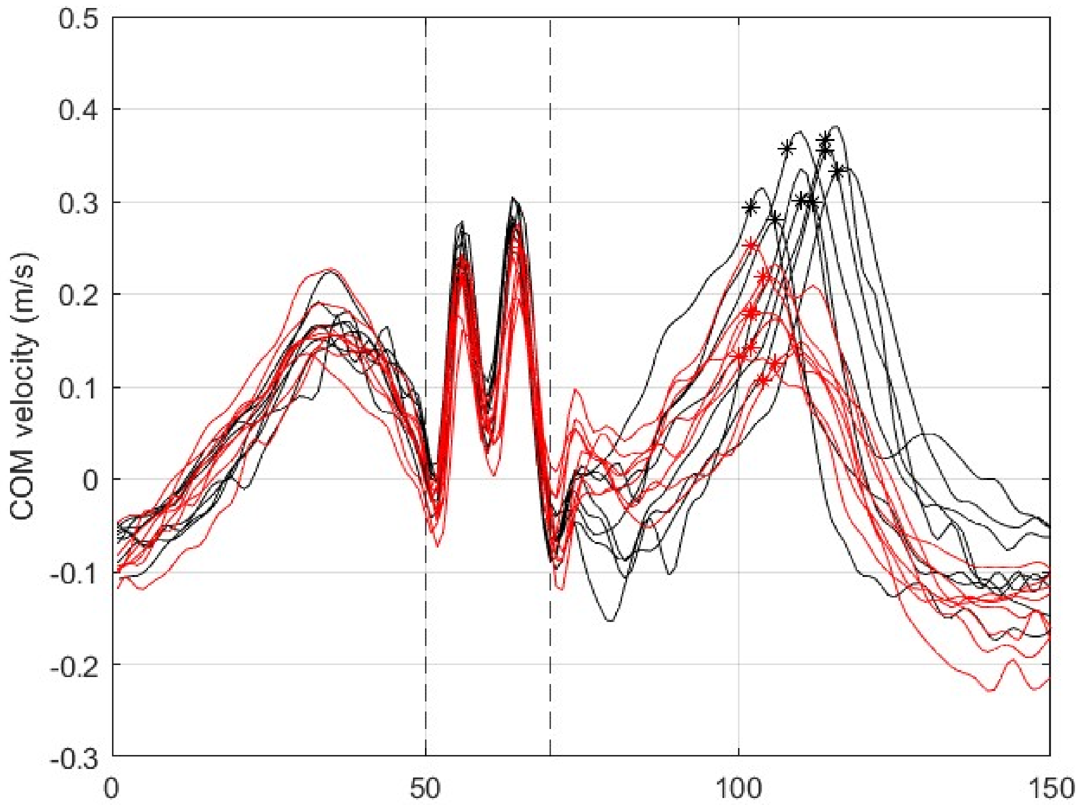


Figure 5: The COM velocity during a perturbation of participant one. The black lines are the velocities during the pre-training trial and the red lines are the velocities during the post-training trial. The asterisks denote the instance of heelstrike after the perturbation. The left vertical line at 50 is the start of the perturbation. The left vertical line at 70 is the end of the perturbation.

3.1 General adaptation

Overall, there was no inter-participant trend in COM velocity at heelstrike when comparing pre-training to post-training trials. As shown in Figure 6, the median COM velocity increased by 3.7% from pre- to post-training trials for Participant 3 and by 23.4% for Participant 5. In contrast, the COM velocity decreased by 49.5% for Participant 1 and by 22.4% for Participant 4. The differences between the averages of participants 1 and 5 are significant with p-values of $4{\cdot}10^{-5}$ and 0.018 respectively. The differences of participants 3 and 4 are not significant with p-values of 0.44 and 0.18 respectively.

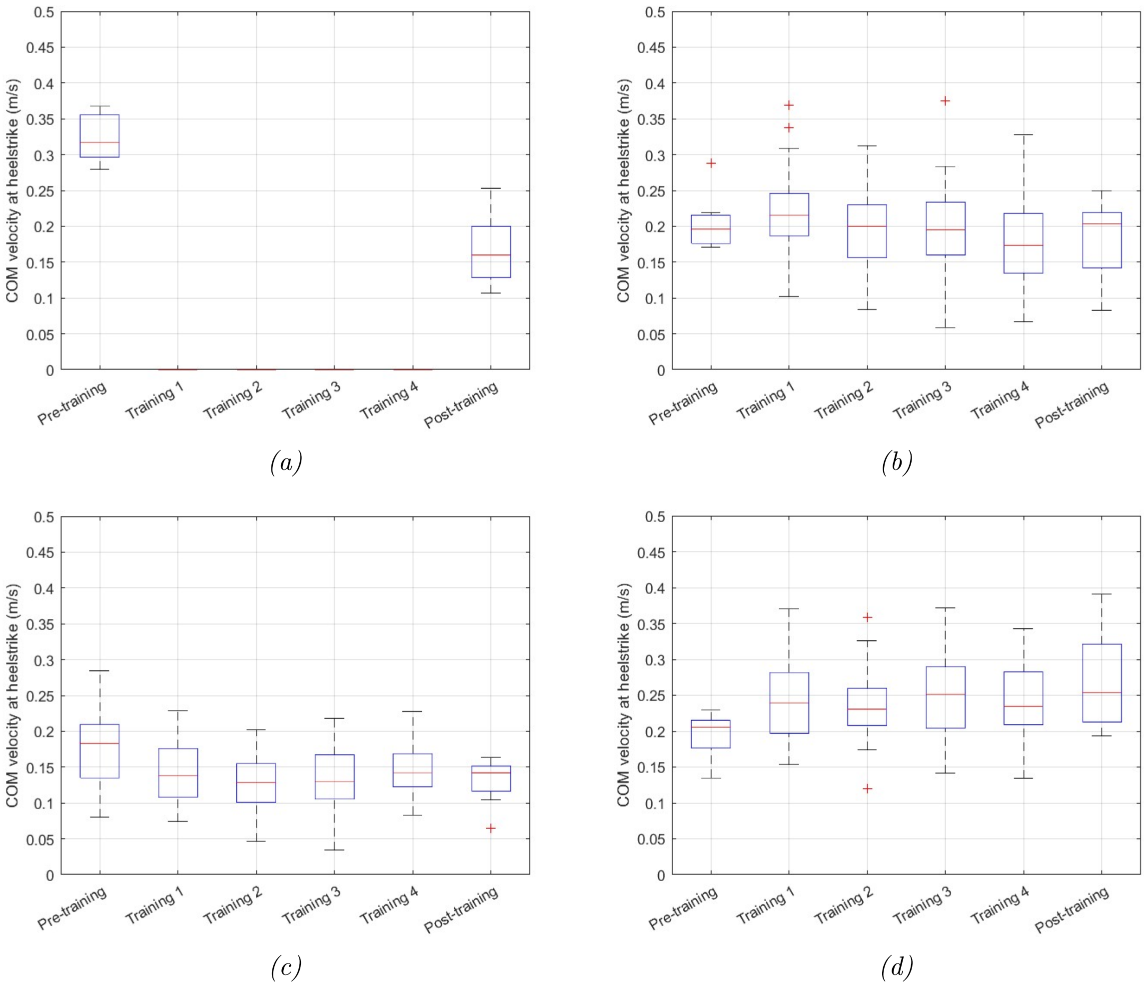


Figure 6: Boxplots of the COM velocities at heelstrike for all trials with a perturbation magnitude of 12%. (a) is Participant 1, (b) is Participant 3, (c) is Participant 4, and (d) Participant 5.

3.2 First trial effect

When considering the COM velocity at heelstrike for all perturbations for all trials with a perturbation magnitude of 12%, there is no clear downward or upward trend visible, as shown in Figure 7. Additionally, for all the participants and trials the first few perturbations are not higher when comparing those with the later perturbations.

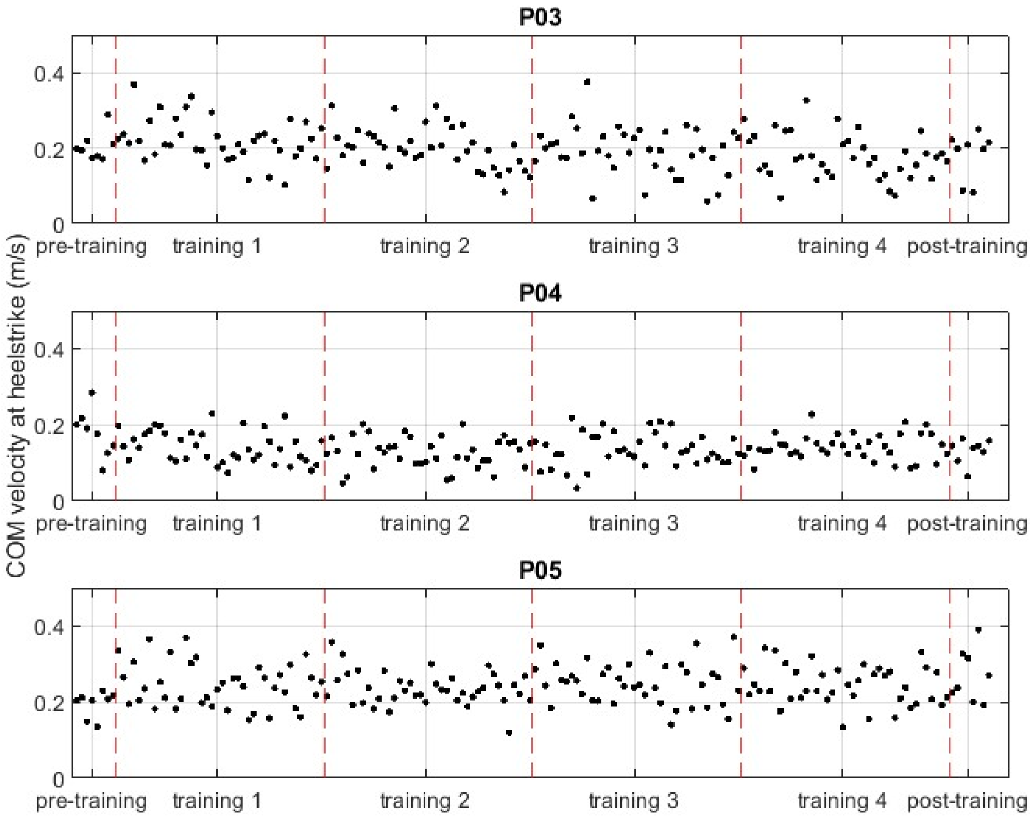
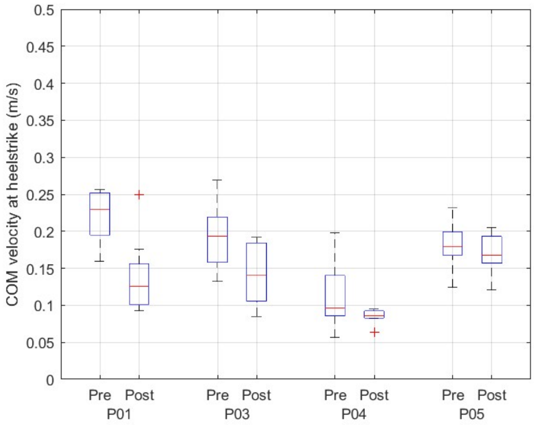


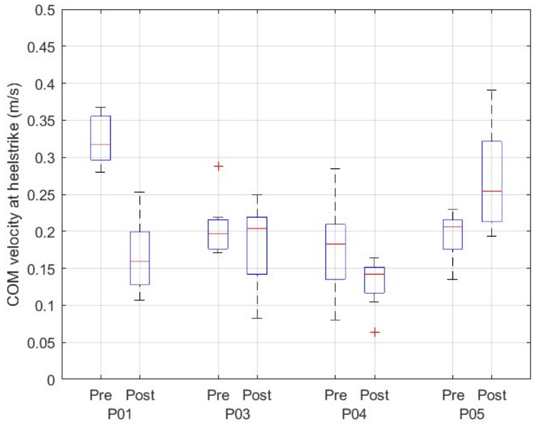
Figure 7: The COM velocity at heelstrike for all perturbations of the trials with a perturbation magnitude of 12%.

3.3 Generalization of adaptation

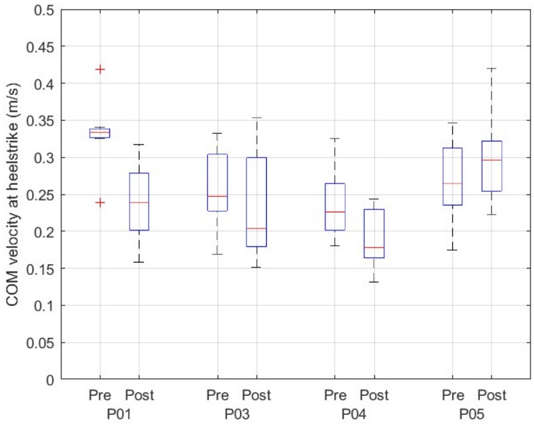
The median of the COM velocity from the generalization trials with a perturbation magnitude of 8% decreases for all participants when comparing the post-training trial with the pre-training trial. As shown in Figure 8a, for Participant 1 it decreases by 45.3%, for Participant 3 by 26.9%, for Participant 4 by 10.9%, and for Participant 5 by 6.3%. The differences between the averages were significant for participants 1 and 3 with a p-value of 0.0027 and 0.018, respectively. The differences were not significant for participants 4 and 5 with a p-value of 0.16 and 0.34, respectively. For the generalization trials with a perturbation magnitude of 16% the COM velocity decreases for participants 1, 3, and 4 and increases for participant 5. As shown in Figure 8c, for Participant 2 it decreases by 28.2%, for Participant 3 by 17.5%, for Participant 4 by 21.3% and for Participant 5 it increases by 12.2%. The differences between the averages were significant for participants 1 and 4 with a p-value of 0.03 and 0.074, respectively. The differences were not significant for participants 3 and 5 with a p-value of 0.53 and 0.22, respectively.



(a)



(b)



(c)

Figure 8: Boxplots for the generalization trials. The boxplots are grouped per participant. (a) Trials with a perturbation magnitude of 8%, (b) Pre-and post-training trials with a perturbation magnitude of 12%, (c) Trials with a perturbation magnitude of 16%.

4 Discussion

The research question of this paper is: How does a person adapt their balance when they are exposed to repeated unexpected pelvic perturbations in the anterior direction during gait without external assistance? By investigating three objectives this question is answered. The primary objective investigated to what extent people adapt their balance to unexpected forward perturbations. The first secondary objective investigated first-trial effects within trials. The second secondary objective investigated whether there is a transfer of adaptation (generalization) when exposed to perturbations with different magnitudes.

The hypothesis for the primary objective was that the COM velocity post-training would reduce compared to pre-training values. For Participant 1 this is true, as the COM velocity at heelstrike decreased significantly by 49.5%. For participant five the opposite occurred, in which the COM velocity at heelstrike increased significantly by 23.4%. The other two participants did not exhibit a significant decrease or increase. So only one of the participants showed a significant decrease in COM velocity after training (Figure 6). Therefore, the hypothesis for the primary objective is not supported by this work; it thus cannot be said that the COM velocity decreases post-training.

The hypothesis for the first secondary objective was that the COM velocity would be higher after the first perturbation compared to subsequent perturbations (i.e., a first-trial effect). When looking at the COM velocities at heelstrike for each perturbation (Figure 7) the spread of the COM velocities of the first perturbations are not higher than the COM velocities of the last perturbations. Instead, the COM velocity generally keeps a steady level throughout the trials. Therefore, the hypothesis for the first secondary objective is not supported by this work; it thus cannot be said that the COM velocity will be higher with the first perturbation and then decrease for the later perturbations.

The hypothesis for the second secondary objective was that the COM velocity post-training would also decrease when exposed to perturbations of a different magnitude than the magnitude of the training trials. The generalization trials were split up into the perturbations with a magnitude of 8% and 16%. For the 8% magnitude, all four participants show a decrease in COM velocity at heelstrike, but only two of those are significant. For the 16% magnitude, three of the four participants show a decrease in COM velocity, but only two of those are significant. One participant shows an increase in COM velocity. As only two out of four participants decreased their COM velocity significantly it cannot be said with certainty that the COM velocity decreases post-training at heelstrike with perturbations with a different magnitude, and therefore the hypothesis for the second secondary objective is not supported by this work.

There are several potential explanations for the results observed in this work. First, there may be no clear adaptation of balance because the perturbation was not threatening. The adaptation that was described in section 1.3 was for tripping perturbations. The perturbations often resulted in falls the first few times. The perturbations in this research are small pushes that do not pose a real threat of falling. It is thus possible that there was no clear adaptation because the body does not deem it important with the perturbations of this magnitude. Because the perturbations pose no threat of falling, the body can also relax to the perturbation. Schmid et al. discovered that the body starts to passively sway with the movement of perturbation [14]. It may be that this phenomenon also occurred during this study. The low magnitude can also explain why there is no first trial effect visible. The first perturbation was not strong enough to induce a large decrease for the later perturbations. The low magnitude can lastly also explain how there is more adaptation to perturbations of different magnitudes. Because the magnitude is different the mind sees it as more important and reacts more to it.

Second, even though there is no adaptation of balance to unexpected perturbations of this magnitude considering COM velocity, there may be an adaptation to other metrics. Multiple parameters change when someone experiences perturbations. For example, ankle torque after a

perturbation to counteract the additional forward velocity from the perturbation may change. The centre of pressure also shifts in the lateral direction of the perturbation. The muscle activity also changes after a perturbation [19]. Thus these metrics may change over time when participants are exposed to perturbations. It is also possible that adaptation is visible in the effort it takes to recover the balance loss. It is already found that the effort decreases while wearing an ankle-exoskeleton [11]. Lastly, the recovery strategy that is used can change as a form of adaptation.
Third, there are differences between participants when looking at the adaptation of balance to perturbations. Participant 1's COM velocity decreases in the course of the trials for all magnitudes of perturbations, while Participant 5's COM velocity increases for two out of three magnitudes. This may be caused by the usage of different recovery strategies, as described in section 1.4, that individual participants use. Additionally, the athleticism of a participant, the comfort level during experiments, and other individual characteristics can also influence the adaptation [20].

To examine the findings of this research there should be more studies in the future that investigate the balance adaptation to unexpected forward-pushing perturbations. To find out if there is no adaptation to perturbations of the magnitude used in this study because there is no threat of falling, future studies can investigate adaptation to stronger pushes. It is possible to let one group of participants experience pushes that make the participants fall at first and one group of participants experience perturbations of a lower magnitude. To find out if there is no adaptation with the COM velocity at heelstrike, future studies can investigate other parameters like muscle activity, centre of pressure, or effort. To find out if there is no adaptation because of individual differences between participants future studies can investigate personal characteristics. For example, investigating the differences between athletic and nonathletic people, or investigating the differences between different recovery strategies. The differences between male and female participants can also be investigated. This study consisted of only 4 participants. To be able to draw a stronger conclusion future studies can be done with more participants.

5 Conclusion

The purpose of this study was to assess the adaptation of balance by answering the following general question: How does a person adapt their balance when they are exposed to repeated unexpected pelvic perturbations in the anterior direction during gait without external assistance? This question was explored considering three objectives. The primary objective was to assess to what extent people adapt their balance when looking at the COM velocity. The outcome for this objective is that there is no adaptation measured by a decrease or increase of COM velocity after training. The first secondary objective was to determine if there was a first trial effect. This was not the case for the COM velocity. The second secondary objective was to assess whether there was a generalization of adaptation to perturbations of different magnitudes. There was also no significant adaptation of the COM velocity for perturbations of a different magnitude. With this knowledge, it can be said that humans do not adapt their balance with their COM velocity. The goal of this study was to use the knowledge gained for future studies with ankle exoskeletons. As there is no adaptation found in this study it does not need to be taken into account as of now when looking at adaptation while wearing an ankle exoskeleton.

Bibliography

[1] Gil-Agudo Megía-García Pons JL, Sinovas-Alonso I, Comino-Suárez N, Lozano-Berrio V, et al. Exoskeleton-based training improves walking independence in incomplete spinal cord injury patients: results from a randomized controlled trial [Article]. Journal of NeuroEngineering and Rehabilitation. 2023;20(1). doi:10.1186/s12984-023-01158-z.

[2] Romanato M, Fichera F, Spolaor F, Volpe D, Sawacha Z. Could an Exoskeleton-Driven Rehabilitation Treatment Improve Muscle Forces Generation in PD? - a Pilot Study [Book chapter]. Lecture Notes in Computational Vision and Biomechanics. 2023;38:36 – 49. doi:10.1007/978-3-031-10015-4_3.

[3] Calabrò RS, Naro A, Russo M, Bramanti P, Carioti L, Balletta T, et al. Shaping neuroplasticity by using powered exoskeletons in patients with stroke: a randomized clinical trial [Article]. Journal of neuroengineering and rehabilitation. 2018;15(1):35. doi:10.1186/s12984-018-0377-8.

[4] Bortole M, Venkatakrishnan A, Zhu F, Moreno JC, Francisco GE, Pons JL, et al. The H2 robotic exoskeleton for gait rehabilitation after stroke: Early findings from a clinical study Wearable robotics in clinical testing [Article]. Journal of NeuroEngineering and Rehabilitation. 2015;12(1). doi:10.1186/s12984-015-0048-y.

[5] Huysamen K, de Looze M, Bosch T, Ortiz J, Toxiri S, O'Sullivan LW. Assessment of an active industrial exoskeleton to aid dynamic lifting and lowering manual handling tasks. Applied Ergonomics. 2018;68:125–131. doi:https://doi.org/10.1016/j.apergo.2017.11.004.

[6] Stolze H, Klebe S, Baecker C, Zechlin C, Friege L, Pohle S, et al. Prevalence of Gait disorders in hospitalized neurological patients [Article]. Movement Disorders. 2005;20(1):89 – 94. doi:10.1002/mds.20266.

[7] Saunders L, Dipiro N, Krause J, Brotherton S, Kraft S. Risk of fall-related injuries among ambulatory participants with spinal cord injury [Article]. Topics in Spinal Cord Injury Rehabilitation. 2013;19(4):259 – 266. doi:10.1310/sci1904-259.

[8] Emmens A, Van Asseldonk E, Masciullo M, Arquilla M, Pisotta I, Tagliamonte NL, et al. Improving the Standing Balance of Paraplegics through the Use of a Wearable Exoskeleton. vol. 2018-August; 2018. p. 707 – 712. doi:10.1109/BIOROB.2018.8488066.

[9] Font-Llagunes JM, Lugris U, Clos D, Javier Alonso F, Cuadrado J. Design, control, and pilot study of a lightweight and modular robotic exoskeleton for walking assistance after spinal cord injury [Article]. Journal of Mechanisms and Robotics. 2020;12(3). doi:10.1115/1.4045510.

[10] Afschrift M, Asseldonk EV, Mierlo MV, Bayon C, Keemink A, van der Kooij H, et al. Assisting walking balance using a bio-inspired exoskeleton controller. bioRxiv. 2022. doi:10.1101/2022.10.19.512851.

[11] Bayón C, Keemink AQL, van Mierlo M, Rampeltshammer W, van der Kooij H, van Asseldonk EHF. Cooperative ankle-exoskeleton control can reduce effort to recover balance after unexpected disturbances during walking [Article]. Journal of NeuroEngineering and Rehabilitation. 2022;19(1). doi:10.1186/s12984-022-01000-y.

[12] Poggensee KL, Collins SH. How adaptation, training, and customization contribute to benefits from exoskeleton assistance [Article]. Science Robotics. 2021;6(58). doi:10.1126/scirobotics.abf1078.

[13] Wang TY, Bhatt T, Yang F, Pai YC. Adaptive control reduces trip-induced forward

gait instability among young adults. Journal of Biomechanics. 2012;45(7):1169–1175. doi:https://doi.org/10.1016/j.jbiomech.2012.02.001.

[14] Schmid M, Bottaro A, Sozzi S, Schieppati M. Adaptation to continuous perturbation of balance: Progressive reduction of postural muscle activity with invariant or increasing oscillations of the center of mass depending on perturbation frequency and vision conditions [Article]. Human Movement Science. 2011;30(2):262 – 278. doi:10.1016/j.humov.2011.02.002.

[15] Bhatt T, Wang TY, Yang F, Pai YC. Adaptation and generalization to opposing perturbations in walking [Article]. Neuroscience. 2013;246:435 – 450. doi:10.1016/j.neuroscience.2013.04.013.

[16] Wang Y, Wang S, Bolton R, Kaur T, Bhatt T. Effects of task-specific obstacle-induced trip-perturbation training: proactive and reactive adaptation to reduce fall-risk in community-dwelling older adults [Article]. Aging Clinical and Experimental Research. 2020;32(5):893 – 905. doi:10.1007/s40520-019-01268-6.

[17] Leestma JK, Golyski PR, Smith CR, Sawicki GS, Young AJ. Linking whole-body angular momentum and step placement during perturbed human walking [Article]. Journal of Experimental Biology. 2023;226(6). doi:10.1242/jeb.244760.

[18] Van Asseldonk EHF, Koopman B, Van Der Kooij H. Locomotor adaptation and retention to gradual and sudden dynamic perturbations; 2011. doi:10.1109/ICORR.2011.5975379.

[19] Vlutters M, van Asseldonk EHF, van der Kooij H. Lower extremity joint-level responses to pelvis perturbation during human walking [Article]. Scientific Reports. 2018;8(1). doi:10.1038/s41598-018-32839-8.

[20] Guzelbulut C, Suzuki K, Shimono S. Singular value decomposition-based gait characterization [Article]. Heliyon. 2022;8(12). doi:10.1016/j.heliyon.2022.e12006.