MASTER THESIS

ADAPTATION TO EXOSKELETON BALANCE ASSISTANCE

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15-03-2024

Abstract

Balance impairment is particularly prevalent among populations with mobility limitations, including elderly individuals, stroke survivors, and patients with incomplete spinal cord injuries. Ankle plantar and dorsiflexion play crucial roles in maintaining balance along the anterior-posterior axis. While exoskeletons have demonstrated efficacy in facilitating walking, integrating balance control has also shown its importance. Investigating adaptation to external perturbations and devices provides valuable insights for improving exoskeleton balance assistance mechanisms.

In this study, adaptation to exoskeleton balance assistance was examined using a device that offered plantar flexion assistance to counteract anterior perturbations based on center of mass (COM) velocity. Electromyography recordings from the gastrocnemius medialis, soleus, and tibialis anterior muscles of the left leg, along with COM displacement measurements, were obtained from four participants undergoing a nine-trial walking experimental protocol.

Findings revealed that two participants adapted their plantar flexion muscles by reducing muscle effort. However, one participant exhibited inconclusive results due to controller inconsistency, while another showed different outcomes which can be explained by a different walking pattern. Notably, participants did not exhibit adaptation in COM displacement.

It is concluded that healthy adults can adapt to exoskeleton balance assistance by reducing plantar flexor muscle activity. This study can be used as a starting point for future research into adaptation to exoskeleton balance assistance.

Contents

1 Introduction

1.1 Problem definition

Falling due to balance loss can cause an increase in injuries, fear of falling and a decrease in community engagement. Balance loss is especially prevalent in populations where the mobility is impaired. Examples of these populations are elderly people, stroke survivors and patients with incomplete spinal cord injuries[1–4]. Therefore, it is very important to explore innovative solutions to improve mobility and reduce fall risk in these populations.

1.2 Human balance

Studies on human balance have shown that the ankle joint plays an important role in controlling human balance[5–7]. Human balance and falling due to balance loss is a three dimensional problem. However, in literature it is often split in to balance in the anterior-posterior direction and balance in the medio-lateral direction. Three strategies are used to remain balanced during walking. These are the hip strategy, ankle strategy and foot placement strategy. Hip strategy is mostly used in combination with the other strategy and is used for large perturbations. Foot placement strategy is important in staying balanced in the medio-lateral direction. Balance in the anterior-posterior direction can also be adjusted with the ankle strategy[7]. The ankle strategy uses muscle strength around the ankle joint to adjust the center of pressure within the length of the foot. This can lead to an increase or decrease in plantar flexion in the foot. This ankle strategy can be used to counteract unexpected perturbations in anterior-posterior direction. An increase in plantar flexion moment in the leading foot can decrease a forward velocity when a perturbation in the anterior direction is applied[8]. An ankle exoskeleton that can provide plantarflexion or dorsiflexion can help counteract a loss of balance.

In healthy adults, the muscles in the lower leg that contribute to plantar flexion moment in the foot are the gastrocnemius and the soleus[9]. The tibialis anterior located at the front of the lower leg contributes to dorsiflexion of the foot.

1.3 Exoskeletons

Exoskeletons have shown to be successful in providing assistance during walking[10–13]. However, these exoskeletons do not incorporate balance assistance. This results in patients still being dependent on their upper body mobility and on the use of crutches[14].

For exoskeleton control strategies, recent studies showed that balance can be implemented by incorporating center of mass (COM) feedback[8, 14, 15]. The effectiveness of incorporating the COM kinematics was shown with a standing controller by Emmens et al.[14]. They used an ankle-foot exoskeleton to investigate this balance controller. The torque that the exoskeleton provided was based on the COM kinematics and showed the same profiles as humans do when they are being perturbed. In addition, they observed decreased activity in the soleus muscle and increased activity in the tibialis anterior muscle during forward perturbations compared to a controller without COM kinematics.

Bayon et al. created an ankle-exoskeleton controller that assisted in counteracting forward perturbations during walking by providing plantar- and/or dorsiflexion[8]. The assistance provided was based on the change in COM velocity. They found that this controller significantly reduced muscle effort in soleus muscle and gastrocnemius medialis compared to no assistance when forward perturbations were applied at right heel strike.

Afschrift et al. developed a controller that supported walking balance and reduced muscle activity[15]. This controller was also based on COM velocity. They compared this controller to controllers without COM feedback. They specifically looked at the muscle activity response and COM displacement after perturbations. Less soleus muscle activity was needed when the controller with COM feedback was used compared to the controllers without COM feedback. The COM displacement was also smaller when the COM feedback controller was used. This suggests that implementing COM feedback in a balance controller requires less additional muscle activity to regain balance. The differences between the studies by Bayon et al. and Afschrift et al. were that the controller in the study by Bayon et al. only commands assistance when a perturbation is detected and otherwise does not command any assisting torque to the ankle. Within the study by Afschrift et al. the exoskeleton always delivered 30 % of the ankle joint moment that was computed by a neuromuscular model.

1.4 Adaptation

Most studies show immediate effects of balance assistance, but there has been an increasing push to better understand how people learn to use these devices[16]. The way people adapt to external perturbations or exoskeleton walking assistance is now a topic of interest. Investigating adaptation is becoming more important because it can help in providing new insight for developing control of the exoskeletons and how to use them.

Adaptation to external perturbations has been investigated by exposing people to repeated perturbations and it is shown that adaptation only occurs when it improves stability during walking. Cajigas et al. exposed participants to robot-induced perturbations during walking[17]. The perturbations were defined as changing the foot trajectory during swing phase in either step length or step height. The participants were constrained at the pelvis. The motor adaptation was studied by repeating blocks of 80 consecutive perturbations. They concluded that there was motor adaptation in the step length direction but not in the step height direction. This data showed that maintaining stability during walking was the underlying principle that caused the adaptation in step length and not in step height, since step height changes did not challenge balance. It showed that maintaining balance is prioritized over energy cost and walking pattern preservation.

Data by Schmid et al. suggested that people use their muscles to prevent their COM position to vary too much in anterior-posterior direction, when they are unfamiliar with the perturbation[18]. The participants stood on a platform that moved to the front and back continuously. Two different speeds of this movement were used in addition to two conditions which were eyes opened and eyes closed. They investigated the adaptation in terms of center of pressure, COM displacement and electromygraphy (EMG) data for the soleus (SOL) and tibialis anterior (TA) muscles. They support the conclusion of Cajigas et al. that maintaining stability is prioritized over energy cost.

Pavol and Pai found that the chance of losing balance is strongly related to the position and velocity of the participant's COM[19]. They applied repeated perturbations where the participants were perturbed during a sit to stand task. The platform underneath their feet made the participants slip. This underlines the importance of controlling our COM to remain balanced.

Other studies have investigated the adaptation to ankle exoskeletons by exposing people to exoskeleton assistance and measuring muscle activity and COM kinematics[20][21]. Gordon and Ferris used an ankle exoskeleton that was controlled by the SOL muscle activity magnitude [21]. The study showed that the participants already reduced their SOL muscle activity after 30 minutes of walking in the assisted exoskeleton.
Kao et al. used a similar muscle driven ankle exoskeleton to investigate whether participants' muscle activity changed during assisted walking versus unassisted walking[20]. They collected EMG data of gastrocnemius medialis(GM), gastrocnemius lateralis(GL), SOL and TA in the left lower leg and showed significant reduction in SOL muscle activity after 30 minutes of assisted walking. The other 3 muscles showed similar activation magnitudes compared to their baseline unassisted walking. In addition, they concluded that the time it took to adapt to the exoskeleton assistance increased, when the provided assistance increased.

Effects of training time on the adaptation to exoskeleton assistance is also investigated by Poggensee and Collins[12]. They looked at metabolic costs to determine adaptation. They showed that giving the user the appropriate time to gain expertise in walking with the exoskeleton can improve the effect of assistance and further reduce metabolic cost. They showed that within their study it took participants 109 minutes to become experts in walking with the exoskeleton.

Adaptation to exoskeleton assistance has also been assessed using EMG activity. Acosta-Sojo and Stirling investigated adaptation in muscle activity of the TA and GM muscles when walking in a powered ankle exoskeleton[22]. They showed a significant reduction in GM activity when walking 10 minutes in an assisting exoskeleton. They also showed that it is important to take into account that people respond differently to walking in an exoskeleton. This is important for future development so people have the time to learn how to walk with an exoskeleton, which supports the conclusion of Poggensee and Collins..

1.5 Knowledge gap

In summary, it is known that exoskeletons can provide assistance during walking and that including balance assistance in the control strategy of exoskeletons is necessary. In addition, exposing people to repeated external perturbations is often used to measure adaptation. Also, people can adapt to external devices in terms of muscle activity and COM kinematics. However, there have not been studies that investigated these aspects together. This led to the following research question for this thesis: How do healthy adults adapt to exoskeleton balance assistance?

To investigate this 4 participants walked on a treadmill in an ankle exoskeleton that assisted their balance. During walking they were repeatedly perturbed with a known magnitude to make them lose balance. The exoskeleton only assisted them when they lost balance and then provided the desired assistance to regain balance. Specifically, one primary objective and 2 secondary objectives were formulated to set up the protocol to obtain an answer on the research question.

The primary objective was to analyze how repeated exposure to exoskeleton balance assistance affected the muscle activity and COM displacement. This would show if adaptation occurred over time for one magnitude perturbations. Secondary objective (1) was to assess whether the effects in response to one magnitude perturbation would generalize to smaller magnitude perturbations. Secondary objective (2) was to assess whether there were after-effects and what they looked like in terms of muscle activity and COM displacement. This would show what is necessary to regain balance when the participant is not assisted by the exoskeleton anymore. The COM displacement would show if there is a big difference in how balanced a person is and the muscle effort would show if and how the lower leg muscles are used to stay balanced.

For all three objectives, hypotheses were formulated. The hypothesis for the primary objective was that the muscle effort of the GM and SOL muscles would decrease over time. It was expected that the TA muscle effort would also decrease, but less than GM and SOL. The COM displacement was also expected to decrease over time, since it would indicate a more balanced walking pattern. For secondary objective (1) it was hypothesized that the effects observed for the primary objective would generalize to the smaller magnitude perturbations. The expectation regarding the after effects was that there would be after effects that would show an increase in muscle activity once the exoskeleton balance assistance was turned off. This is hypothesized because if the participants would adapt, the muscles would need to substitute the work that the exoskeleton did when it was assisting. The COM displacement was hypothesized to be bigger once the exoskeleton was not assisting anymore compared to when it was assisting.

2 Methods

In this study participants walked on a treadmill with ankle exoskeletons during unexpected perturbations during a single session. Four able-bodied people (age: 24 $\pm$ 2 years , height: 1.76 $\pm$ 0.12 m, body mass: 75.75 $\pm$ 11.62 kg; mean $\pm$ SD) were included in this study. The group of participants consisted of 2 males and 2 females. The participants who were included did not have any known disabilities regarding maintaining their balance or during walking. During the experiments, the participants wore a safety harness to prevent them from falling. The participants provided a written informed consent. A local ethical committee at the University of Twente approved the experimental protocol.

2.1 Experimental setup

The participants walked on a force-instrumented, split-belt treadmill (MotekMedical, Culemborg, Netherlands) with an exoskeleton on both feet. They were also attached to a pusher device(Moog, Nieuw Vennep, Netherlands) that applied unexpected perturbations. The experimental setup is shown in Figure 1. Each part is described in more detail below.

2.1.1 External perturbations

During the experiment, participants experienced perturbations induced by a pusher device, resulting in the loss of balance. This pusher device was attached at the back of a soft brace that the participant wore around the pelvis(Fig. 1). All perturbations were applied in anterior direction at left heel strike and lasted for 250 ms. Subsequent perturbations occurred within the following 6 to 12 gait cycles. The magnitude of each perturbation was set at either 16% or 12% of the participant's body weight (BW), depending on the trial. Participants were not informed of the exact number of perturbations, but were given an approximate duration for the trial.

2.1.2 Balance Assistance

For balance assistance, participants wore a bowden cable-driven ankle exoskeleton on both feet. The mechanical design and low-level control are detailed in Meijners et al.[23]. Figure 1 shows the setup of the exoskeleton. This exoskeleton could only provide plantar flexion assistance.

The control of the ankle exoskeleton was based on a previously developed controller by Bayon et al. [8]. The goal of the controller was to detect when the user loses balance and to be able to command the correct amount of torque to compensate for the loss of balance. The final controller consisted of two main parts to achieve this goal.

The first part was to detect a perturbation and the second part was to provide the desired ankle torque to counteract the perturbation. For the detection part, the COM velocity was used. This COM velocity was the calculated mean of the position of the two front markers on the brace worn around the pelvis. The detection part of the controller was based on the error $\dot{e}$.

$$\dot{e} = \dot{x} - \hat{\dot{x}} \tag{1}$$

where $\dot{x}$ was the current COM velocity and $\hat{\dot{x}}$ was the predicted COM velocity which was the mean of the preceding 5 unperturbed gaitcycles. A perturbation was detected if the error in the current gait cycle was bigger than the averaged $\dot{e}$ range over the preceding 5 unperturbed gait cycles.

$${}^{m}\dot{e} > P\frac{1}{5}\sum_{i=m-5}^{m-1}(\max_t\{{}^{i}\dot{e}(t)\} - \min_t\{{}^{i}\dot{e}(t)\}) \tag{2}$$

with P the threshold that was previously determined by Bayon et al. at 1.5, m the current gaitcycle, i indicating which gait cycles are used to calculate the mean for the current gait cycle and t the local time[8].

The second part calculated the desired ankle torque when a perturbation was detected. The desired ankle torque is calculated by

$$\tau_l = KM\sqrt{gl_{com}}\dot{e}(t)\frac{F_l(t)}{F_l(t)+F_t(t)} \tag{3}$$

where K and g are always the same since K is a dimensionless factor that was previously determined by Bayon et al. at $K = 0.43$ and g is the gravitational acceleration $g = 9.81m/s^2$. The variables M and l_{com} are determined for each participant individually where M represents the participant's mass in kg and l_{com} is determined by $l_{com}[m] = 1.24 * l[m]$ with l the leg length measured from trochanter to lateral malleolus on the same side. These metrics are used to scale the support to the specific participant. The error and the ground reaction forces are used to scale to the amount of torque at that point in time that is needed to counteract the perturbation. The subscripts l and t represent the leading and trailing foot, respectively. The torque that was calculated was applied to the leading foot because providing plantar flexion in the leading foot counteracts the perturbation in anterior direction. Since all perturbations happened at left heel strike, only the left foot received plantar flexion assistance.

The assistance stopped when the left foot moved behind the COM position. This was incorporated into

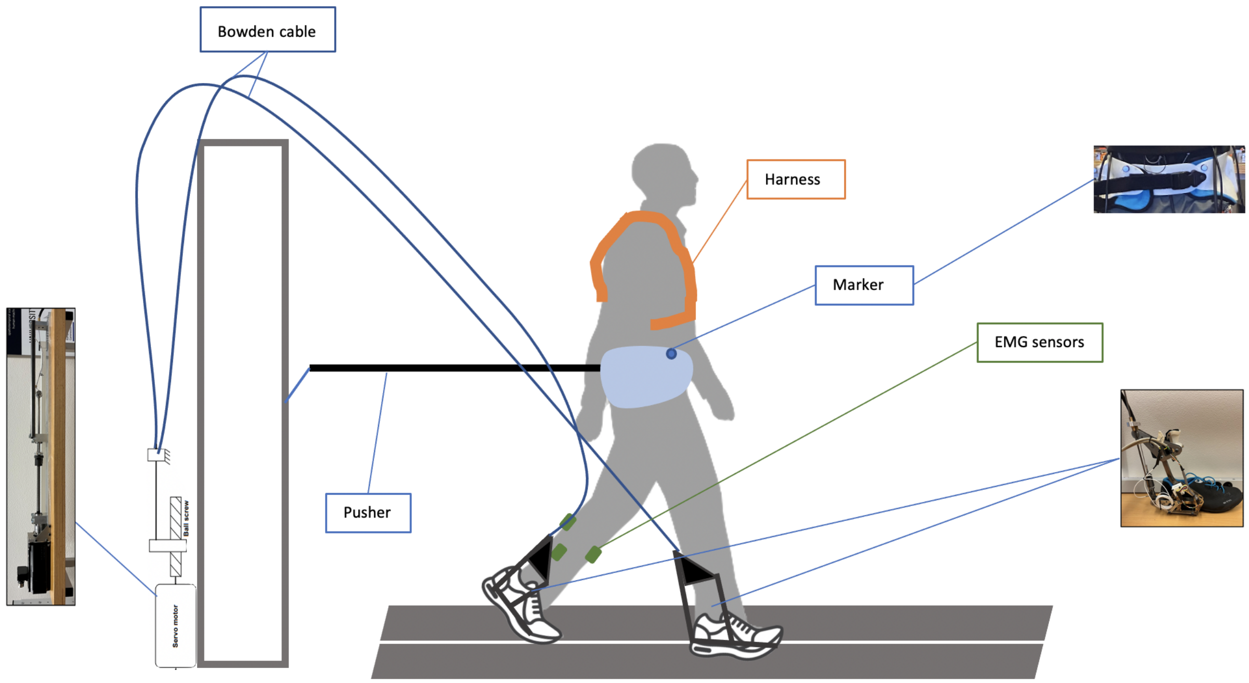


Figure 1: Experimental setup. The participant wore a harness which was attached to the ceiling. A soft brace is worn around the pelvis and shows one representative front marker on the illustration. The picture shows a front view of the brace with the two front markers. The pusher device is attached at the back of the brace to apply the perturbations. Two exoskeletons are shown with the bowden cabled that attached the shoes to the motors. Both the pusher device and the motor of the exoskeleton were mounted on a large pillar that stood securely on the ground.

the controller to prevent the exoskeleton from providing assistance when the foot is behind the COM, as this would lead to the participant gaining velocity in the anterior direction.

Two additional checks were implemented to prevent the controller from falsely detecting a perturbation. As mentioned above, two motion capture markers were used to estimate the COM position and velocity. If one of these two markers was obstructed which caused the marker not to be seen by the cameras, perturbation detection was turned off. Additionally, the controller waited 2 seconds before it could detect a new perturbation. This was implemented to prevent false positives from occurring when it was still a result from the perturbation before.

2.2 Experimental Protocol

At the beginning of the experiment, the participant was familiarized with all the equipment in the laboratory. Additionally, an introduction of the experiment was given including what was expected of the participant. Next, the participant's age, height, leg length and mass were acquired. Following this, all sensors were applied to the participant and the participant stepped on the treadmill where the exoskeletons were put on. The participant was instructed to walk with one leg on each belt and to cross the arms in front of the chest to prevent obstruction of markers. Furthermore, they were instructed to try and stay centered in anterior-posterior(AP) direction on the treadmill and to rely on the exoskeleton for support. Despite these specific instructions, they were told that it was most important to walk as normally as possible and that they should respond in the way that felt normal to them. The walking speed of the treadmill for each participant was determined by

$$speed = 0.63 * \sqrt{l}[m/s]$$

where l is again the length between the trochanter and the lateral malleolus[6]. The used velocity is slow, this is chosen as it is assumed that people with motor deficits will be more likely to walk at a slow walking speed.

The complete protocol consisted of a baseline walking trial followed by 9 experimental trials. Figure 2 shows the complete experimental protocol including the amount and magnitude of perturbations applied. During the first and last trial, the participants walked on the treadmill and received 8 pushes of a magnitude of 16 % BW without assistance from the exoskeleton. In the remaining trials, the exoskeleton assisted the participant if the controller detected the loss of balance. During each trial, the participant did not speak to prevent affecting walking behaviour. In between trials the treadmill was slowed down and small breaks were optional. The trials comprising 8 and 40 perturbations each lasted approximately 3 and 8 minutes, respectively. In total, participants engaged in treadmill walking for approximately 50 minutes.

2.3 Data collection

As previously mentioned, assistance was consistently provided to the left foot throughout the experiment. It was assumed that both legs would have similar responses within able-bodied people. For that reason, the electromyography (EMG) sensors were only applied at the left lower leg. Three EMG sensors were used to acquire muscle activity data of the gastrocnemius medialis (GM), soleus (SOL) and tibialis anterior (TA) muscles[24]. These sensors were placed using the directions of SENIAM[25]. The EMG data were sampled at a frequency of 2000 Hz.

A lower body markerset with 35 markers was used to capture the kinematics of the participant. A real-time connection was made between the control of the pusher-device, the exoskeleton and the treadmill and the qualisys motion capture system. This real-time connection connected the positions of the front markers that were used to calculate the COM velocity as described in section 2.1.2. The simulink system collected data at 1000 Hz.

The force plates in the treadmill collected ground reaction forces of each belt. With these ground reaction forces and the moments, the center of pressure (COP) was calculated. The COP was used to stop providing assistance.

2.4 Data processing

All data was processed with MATLAB(R2023, Mathworks). EMG raw data were filtered with a 4^{th} order butterworth bandpass filter with cutoff frequencies of 20 Hz and 350 Hz. After the bandpass filter, a notch filter was used to remove interfering noise from the power line source at 50 Hz. After the notch filter, the signal was rectified and finally a lowpass filter with a cutoff frequency of 3 Hz was used to get the EMG envelope. The maximum value of the baseline walking trial was used to normalize the data. The normalized data was used to analyze the results.

2.5 Outcome Metrics

The outcome metrics used to assess the objectives were muscle effort and COM displacement. The controller performance was assessed to assess whether the controller performed consistent for all participants. The areas and calculations are indicated in Figure 3. The controller performance was assessed with perturbation detection accuracy, amount of false positive and torque tracking performance. Only the detected perturbations were included for assessing the objectives. Additionally, the exoskeleton torque impulse is computed. This is computed to assess whether the desired torque to regain balance based on the controller changes over time.

The effort of each muscle was calculated by integrating the EMG data over the first 1000 ms after the onset of the perturbation. This was calculated for each perturbation of every trial. This time interval was chosen because it included the main response to perturbations as shown in Figure 3.

The COM position was estimated by calculating the mean of 2 front markers and 2 back markers on the pusher brace. The displacement is defined as the difference between the COM at the onset of perturbation and the maximum value of COM position in the next 2000 ms. This was chosen because it included the maximum displacement due to the controller. These values were all in AP direction.

For the primary objective, the change in muscle effort and COM displacement over time was investigated. This included Pre16, Post16 and the training trials. This showed whether there was adaptation in terms of muscle effort or COM displacement.

To assess secondary objective (1) Pre12 and Post12 were used to assess if similar trends were observed in these trials compared to the ones with 16 % BW. This would show if a training period with a known magnitude also caused adaptation in response to a different magnitude of perturbation.

To assess secondary objective 2, the change in muscle effort and COM displacement is investigated for the Post16 compared to Unassisted Post16. This shows the last assisted trial with 16 % BW so the participant is familiar with the exoskeleton and it is assumed that if there has been adaptation that it has occurred by now. This is used to assess whether the participants used more muscle activity once the balance assistance is removed as well as how much COM displacement changed. The COM displacement indicated whether the participant was more balanced and how much muscle effort ensured this balance.

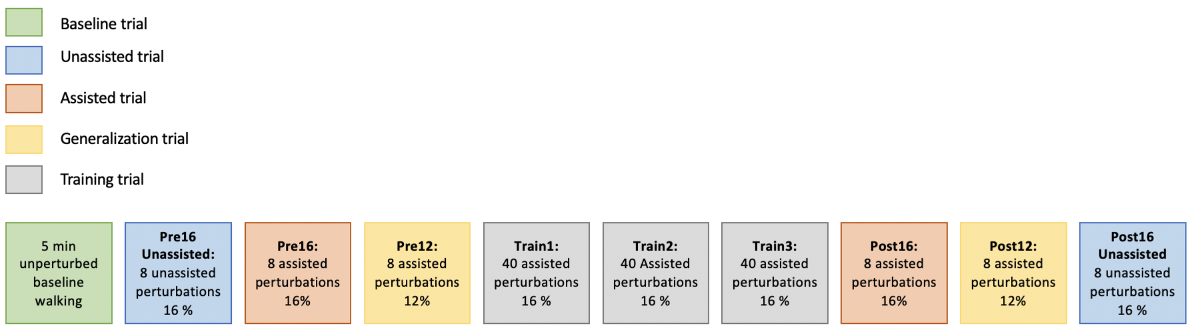


Figure 2: Flowchart of experimental protocol including baseline walking trial, two unassisted trials, two assisted trials with 16 % bodyweight perturbations, two assisted trials with 12 % bodyweight perturbations and three training trials with 16 % bodyweight perturbations

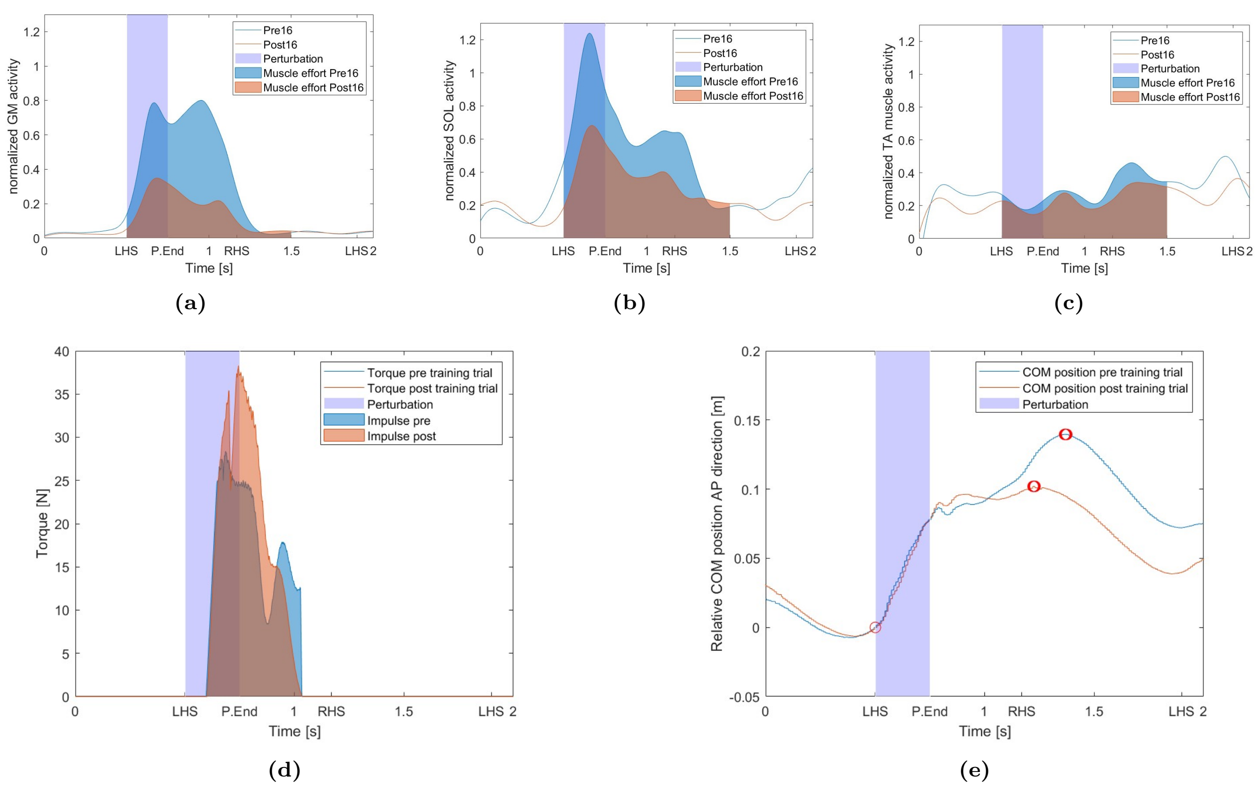


Figure 3: Methods for computing the outcome metrics for one perturbation of Pre16 and one perturbation of Post16 for P01. The muscle effort is calculated by integrating 1000 ms after perturbation onset for GM(a), SOL(b) and TA(c). The exoskeleton torque impulse is calculated by integrating the torque applied over the time that it is commanded (d) and COM displacement is defined as the difference between max value of 2 s after perturbation onset and COM position in AP direction at perturbation onset (e). Gait cycle events are indicated for left heel strike (LHS) and right heel strike (RHS) and the end of perturbation is indicated by P.End

3 Results

3.1 Controller Performance

In order to interpret the results correctly, the controller performance was assessed. This showed whether the participants had similar experiences with the controller or if there are differences that need to be taken into account. The controller performance was assessed with the accuracy of perturbation detection, the amount of false positives and the torque tracking performance, which is reported as percentage of commanded torque impulse that was actually delivered by the exoskeleton. Table 2 shows all these computed values for each participants.

There were three reasons why a perturbation was not detected. These were marker obstruction, a falsely detected perturbation happening in the 2 seconds before an actual perturbation or the threshold for detection was not exceeded. Table 1 presents the percentages of each reason for each individual participant. In summary, the main reason for no detection was not exceeding threshold for three participants(P01, P03 and P04) and false positives for one participant(P02). In Appendix A.1 examples are shown for a normally detected perturbation, a falsely detected and for an undetected perturbation due to not exceeding the threshold. In addition to the figures, more explanation regarding undetected perturbations including ideas for controller improvements are included.

Table 2 shows that the controller was consistent for P01, P02 and P03 regarding torque tracking performance. Only for P04 the percentage deviated with almost 20 % throughout the trials. In addition, the accuracy for perturbation detection was highest for P02. P03 experienced the most false positives throughout the experiment and P04 the least amount.

Table 1: Percentages of reasons for undetected perturbations. For the remaining 30 % of undetected perturbations for P01 the controller was off.

	Marker obstruction[%]	**False positive[%]**	**Threshold not exceeded[%]**
P01	5	17.5	47.5
P02	11	67	22
P03	0	32	68
P04	0	13	87

Table 2: The amount of detected perturbations over applied perturbations, amount of false positives and the torque tracking performance expressed as percentage of commanded torque impulse that was actually delivered by the exoskeleton for all participants. For each participant, the total known and applied perturbations is mentioned in the table caption. The total amount of perturbations is not equal for all participants, because the controller had to be stopped due to malfunctioning or the motion capture system had to stop due to capacity problems. P03 is missing Post12 because it was not conducted, due to malfunctioning controller. Train3 of P03 and Train1 and Train3 of P04 are missing due to unreadable files.

(a) P01, 106/146 detected, 39 False positives

Trial	**Detection**	**False Positives**	$\frac{provided}{commanded}$ [%]
Pre16	7/8	1	82
Pre12	5/8	0	80
Train1	23/38	11	80
Train2	33/40	15	80
Train3	28/36	3	80
Post16	6/8	7	80
Post12	4/8	2	80

(b) P02, 119/128 detected, 55 false positives

Trial	**Detection**	**False Positives**	$\frac{provided}{commanded}$ [%]
Pre16	8/8	2	88
Pre12	8/8	1	89
Train1	15/16	3	88
Train2	38/40	17	89
Train3	34/40	26	89
Post16	8/8	4	89
Post12	8/8	2	89

(c) P03, 70/104 detected, 80 false positives

Trial	**Detection**	**False Positives**	$\frac{provided}{commanded}$ [%]
Pre16	5/8	8	87
Pre12	5/8	13	90
Train1	23/40	15	87
Train2	31/40	34	89
Train3	-	-	-
Post16	6/8	10	89
Post12	-	-	-

(d) P04, 51/66 detected, 3 False positives

Trial	**Detection**	**False Positives**	$\frac{provided}{commanded}$ [%]
Pre16	7/7	0	92
Pre12	4/8	0	91
Train1	-	-	-
Train2	30/35	1	73
Train3	-	-	-
Post16	6/8	0	85
Post12	4/8	2	73

3.2 Adaptation

Figure 4 shows the change in averaged GM activity for P01. It shows that every next trial with 16 % BW perturbations the average GM muscle activity has decreased. For the remaining muscles and participants the figures are shown in appendix A.2.

The muscle effort is calculated for each perturbation that is applied as described in the methods. Figure 8 shows these values for each perturbation applied to P01 in chronological order. The undetected perturbations are also included in this plot to assess if participants responded differently when the exoskeleton unexpectedly did not assist their balance. Overall, the muscle effort for the undetected perturbations throughout the trials did not show a different pattern than the ones that were detected. Only the SOL muscle effort for P03 showed a different pattern for the undetected perturbations compared to the detected perturbations, as shown in Figure 5. The muscle effort for the undetected perturbations was for most perturbations bigger than for the ones that were detected. For the remaining participants these figures are shown in Appendix A.3.

The mean and standard deviation across each trial for each participant summarizing the results on the adaptation objective are shown in Figure 9. The majority of subjects showed a reduction in plantar flexion muscle activity over time. However, there was variability in subject responses. In addition, the majority of subjects showed no change in TA muscle activity over time. Again, there is subject variability. P03 showed a decreased TA activity, from an initial bigger muscle activity. Furthermore, P04 shows a different muscle activation distribution compared to the other participants. The GM and SOL muscle activity contribute approximately the equally effort, where the other participants show a bigger contribution from the SOL muscle compared to GM.

The majority of subjects showed small increased COM displacement over time. P03 showed a small decrease in COM displacement over time. P03 also had the largest displacement for all trials compared to the other participants.

Figure 6 shows the desired torque that was calculated by the controller. For the majority of participants the desired torque did not change over time. Only for P03 it increased over time, and for P01 it decreased over time.

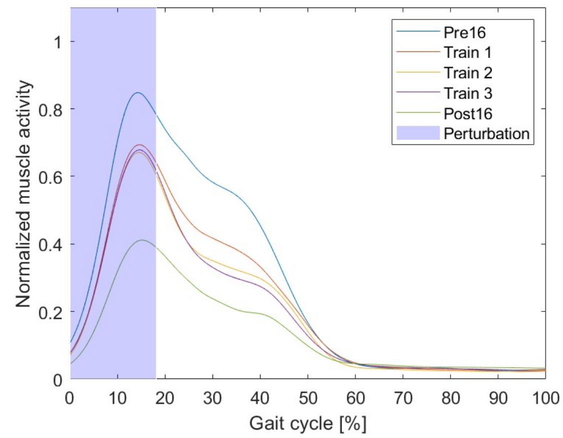


Figure 4: Average Muscle activity for P03 across all perturbation for Pre16, Train1, Train2 and Post16 for GM. Gait cycle starts and ends at left heel strike. Perturbation duration is indicated by shaded area.

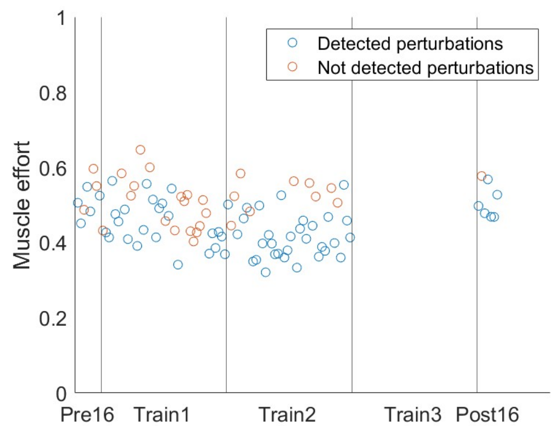


Figure 5: Muscle effort for each perturbation for SOL muscle of P03 in chronological order.

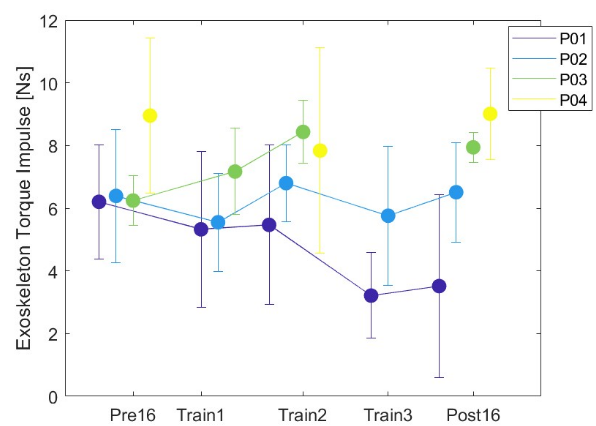
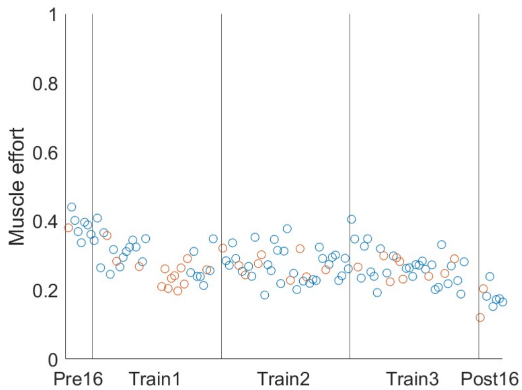
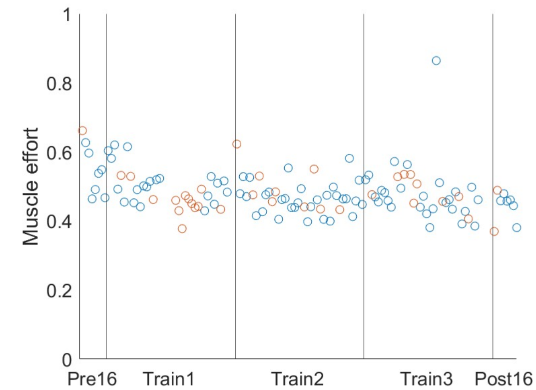


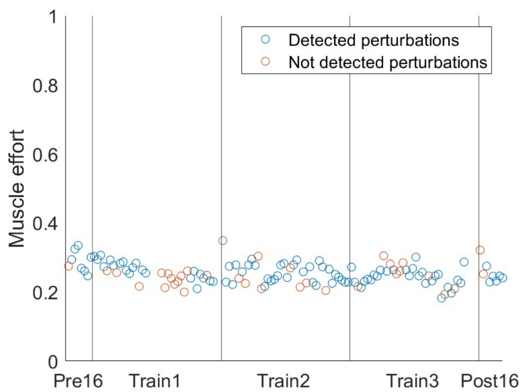
Figure 6: Mean and standard deviation of assistance provided by the exoskeleton across Pre16, Train1, Train2, Train3 and Post16 for all participants.



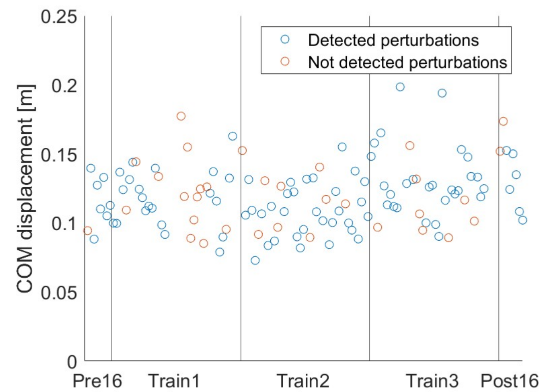
(a) GM



(b) SOL



(c) TA



(d) COM

Figure 8: Muscle effort of GM(a), SOL(b) and TA(c) and COM displacement (d) for each 16 % BW perturbation for P01 in chronological order.

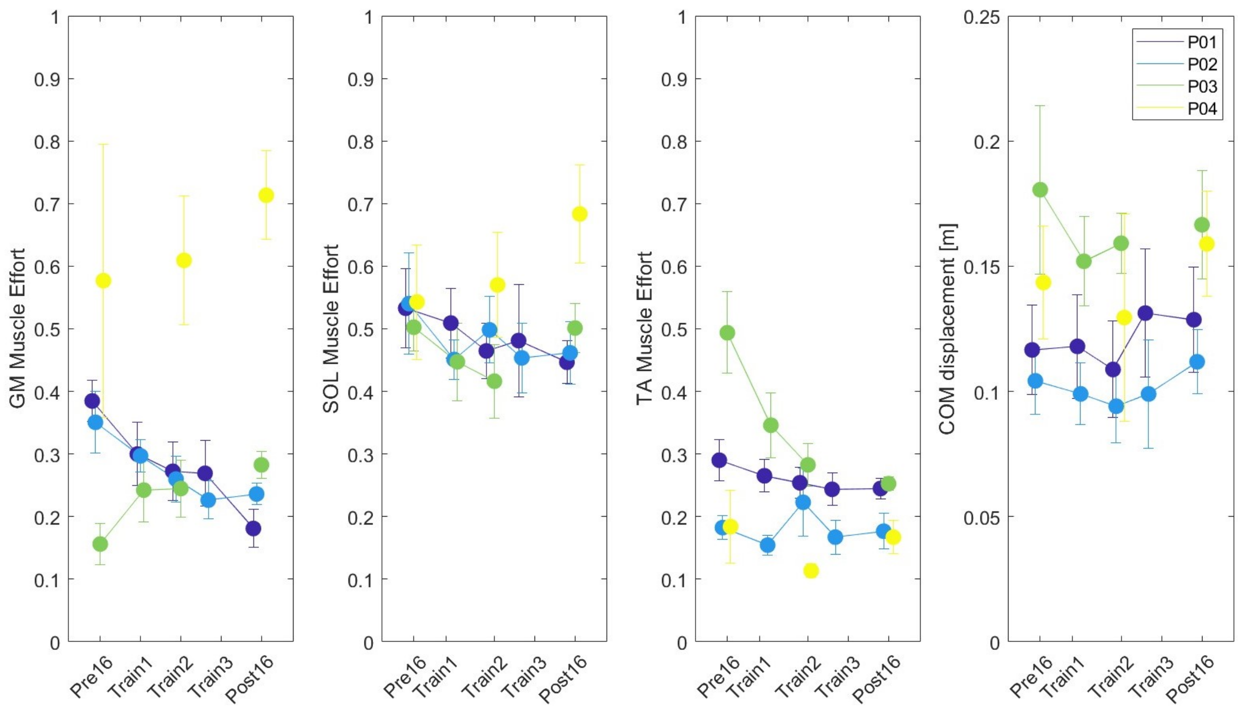


Figure 9: Mean and standard deviation of GM, SOL, TA muscle effort and COM displacement (from left to right) across each trial for all participants for Pre16, Train1, Train2, Train3 and Post16. P03 is missing Train3 and P04 is missing Train1 and Train3, due to unreadable files.

3.3 Generalization

Figure 11 shows the results to assess secondary objective 1. The majority of subjects showed results in line with the results for the 16 % BW perturbations. P03 is not included here, since the Post12 trial was not conducted as mentioned before.

All participants showed a decreased muscle effort in the Post12 trial compared to the Pre12 trial. The results for P04 were not in line with this participant's results for the 16% BW perturbations in terms of muscle effort. Figure 9 showed increased muscle effort over time, where now a decreased muscle effort is observed.

The results for COM displacement are in line with the results for the 16 % BW perturbations. Only P02 showed a small decreased COM displacement in the Post12 trial compared to the Pre12 trial, which is different than the observed small increase over time for the 16 % BW perturbations.

The torque impulse did not change over these 12 % BW perturbations, which is also in line with the exoskeleton torque impulse for the adaptation trials. The only difference in this metric is that the amount of exoskeleton torque impulse in general is lower than for the 16 % trials.

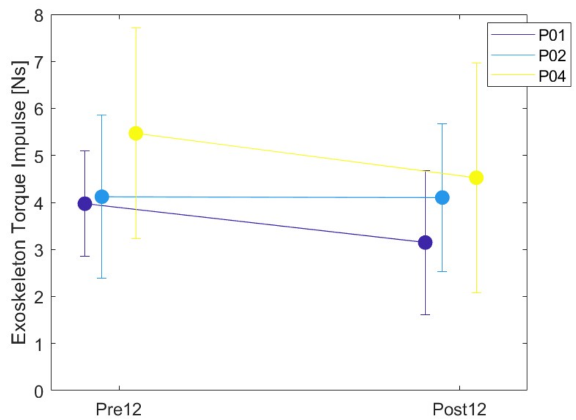


Figure 10: Mean and standard deviation of assistance provided by the exoskeleton across Pre12 and Post12 for P01,P02 and P04.

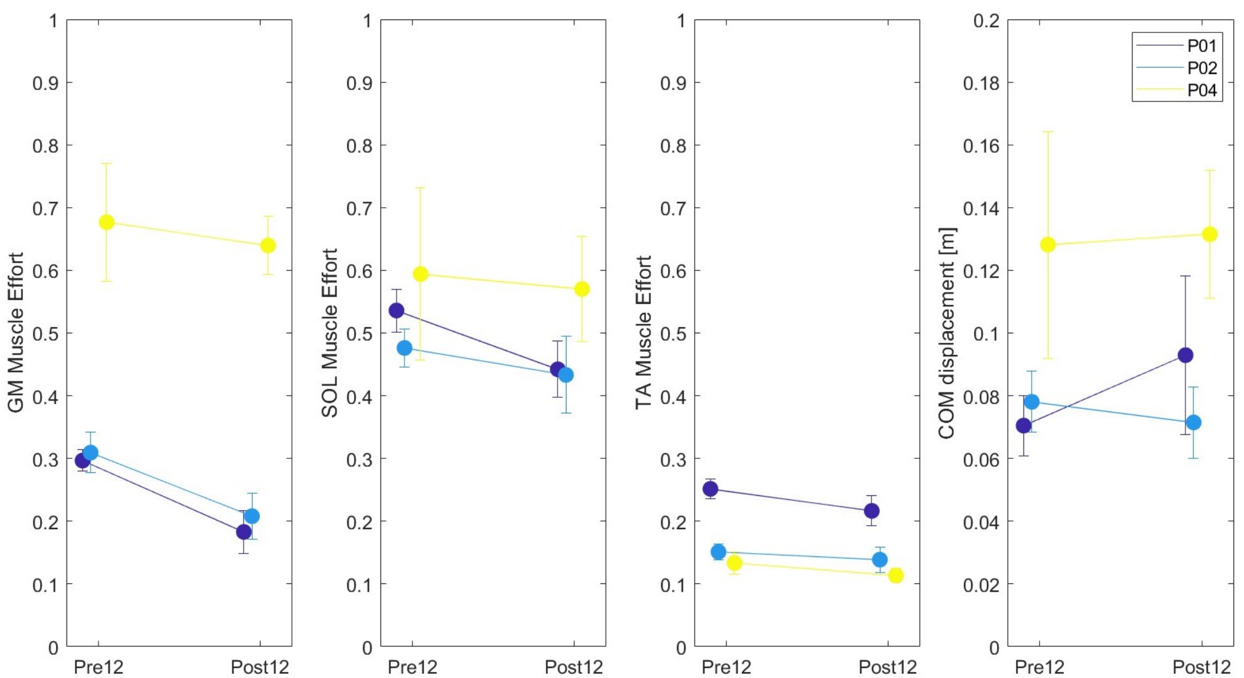


Figure 11: Mean and standard deviation of GM, SOL, TA muscle effort and COM displacement (from left to right) across Pre12 and Post12 for P01,P02 and P04.

3.4 De-training/After effects

Figure 12 shows the results for assessing secondary objective (2). All subjects responded in different ways. P01 showed increased muscle effort for the GM and SOL muscles during unassisted compared to assisted and no change in TA muscle. P02 showed increased GM muscle effort, no change in SOL and decreased TA muscle effort during unassisted compared to assisted. The other two participants(P03 and P04) showed decreased muscle effort for the GM and SOL in unassisted compared to assisted. They both showed no change in TA muscle effort.

The COM displacement increased for all four participants.

3.5 Additional gait cycle metrics

This section was added to look into differences between the participants. During the experiment it seemed clear that P04 had a longer stride duration than the remaining three participants. Therefore, this is calculated to see if it proved the assumption. Table 3 shows the average gait cycle duration calculated from left heel strike at perturbation onset to the next left heel strike for every trial and all participants. It shows that the average gait cycle duration for P04 was the highest.

Table 3: Mean and standard deviation of gait cycle duration across all detected perturbations throughout the experiment, measured from left heel strike at perturbation onset to next left heel strike for each participant.

Participant	Average gait cycle duration ±SD[s]
P01	1.44 ± 0.05
P02	1.26 ± 0.04
P03	1.27 ± 0.10
P04	1.80 ± 0.21

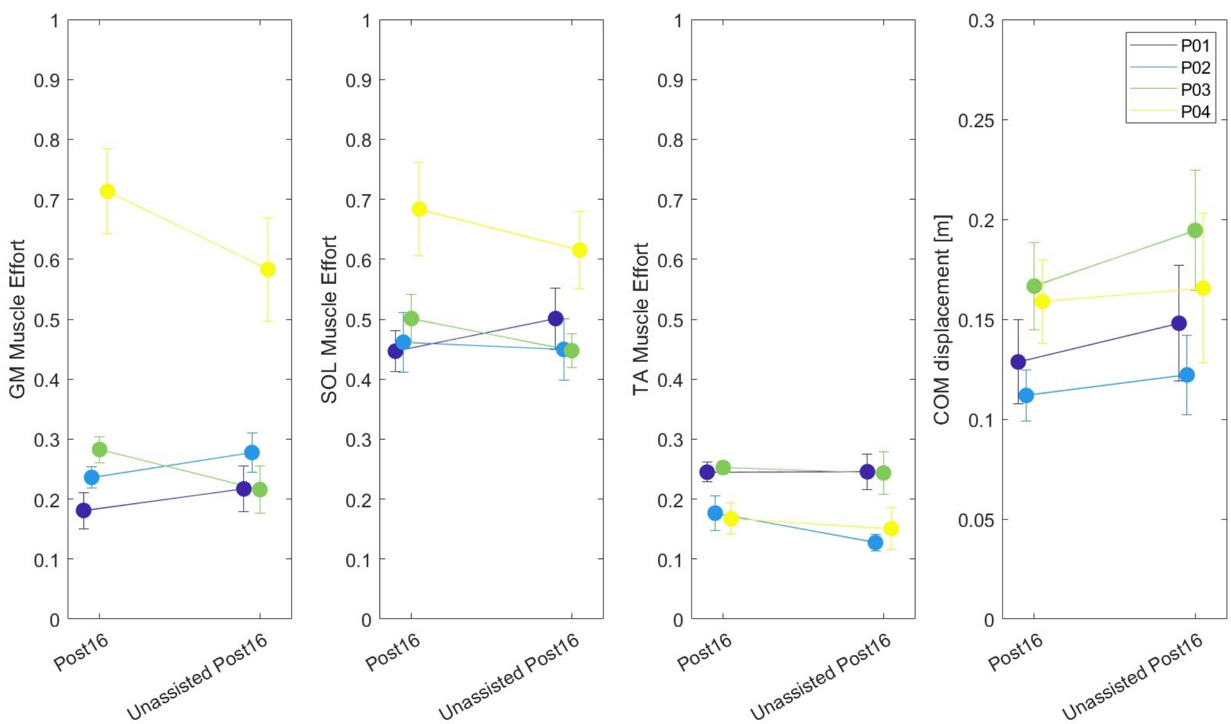


Figure 12: Mean and standard deviation of GM, SOL, TA muscle effort and COM displacement (from left to right) across Post16 and Unassisted Post16 for all participants.

4 Discussion

4.1 Summary results

4.1.1 Controller

The performance of the controller showed variability across participants, thereby influencing the study outcomes. The main reason for undetected perturbations was that the threshold for detection was not exceeded. Before the experiments were conducted, the controller was adjusted to perform the best in terms of detection. An explanation for the inconsistency could be that the controller was adjusted mostly based on two people before the actual experiments. They were more used to receiving the perturbations and therefore might have responded differently. This should be investigated further to prevent it from affecting the results. Suggestions for improving the controller performance are further discussed in Appendix A.1, since it was not the main objective for this thesis.

4.1.2 Adaptation

People can adapt to exoskeleton balance assistance by decreasing their plantar flexor muscle activity over time. P01 and P02 showed that the plantar flexor muscle activity can be reduced while remaining balanced, which is in line with what was expected. It is even shown that for P01 less exoskeleton balance assistance was needed in the last supported trial where there was a slightly larger COM displacement than in the first supported trial. This suggests that this participant learned to walk more efficiently and allowed a bigger COM displacement since it might not be affecting the balanced position. This indicates that the participant relies on the exoskeleton assistance and does not control the COM position[18]. P02 also showed a small increase for COM displacement over time, so this supports a similar trend. During Train1 of P02 after 15 perturbations, the rod that applied the perturbations at the pelvis broke off. The trial was stopped immediately but after attaching a new rod, the participant wanted to go on with the experiment. This explains the increase of muscle effort at Train 2 for both the SOL and TA, since this event could have caused the participant to stiffen up her muscles when expecting perturbations again. It is shown that the muscle effort went back to the level of muscle effort where it was before the rod broke off after the second training trial. This shows that the participant did rely on the controller and adapted to the balance assistance by decreasing plantar flexor muscle effort over time.

As mentioned before, the controller did not show consistency for all participants. P03 was affected most by the inconsistent perturbation detection, which is why these results are more difficult to interpret. The inconsistent controller could have caused the participant not to rely on the exoskeleton to provide the desired assistance to remain in a balanced position. This assumption is supported by the figures where all detected and not detected perturbations are plotted together. The SOL muscle showed a that the muscle effort was slightly higher for a few of the perturbations, which shows that the participant was disrupted in learning to walk with the exoskeleton. This shows that this participant experienced a different experiment than the other participants. For example P01 did not show different muscle effort values for the perturbations that were not detected, which could mean that P01 relied more on the exoskeleton to assist and did not adjust muscle effort for it.

The results of P04 do not support the conclusion that adaptation occurs by decreasing plantar flexor muscle activity, which can not be linked to inconsistency of the controller. However, there could be multiple reasons why this participant did not show adaptation. Often, the SOL muscle is the biggest contributor in counteracting perturbations in anterior direction [15], which is contradicted by the results of P04. Acosta-Sojo et al. suggested that individual differences between people can affect how people adapt to external devices[22]. In addition to the different distribution of muscle contribution to counteract perturbation, P04 also had a longer stride duration during the experiment. This indicates a different timing of the perturbation, which could indicate a difference in stability between participants. This might have led to the perturbations not challenging the stability of the participant in a way that needed adaptation[17].

The results showed clear responses in muscle effort as said before, but the COM displacement did not show clear changes throughout the experiments. Vlutters et al. investigated muscle responses and COM velocity and showed that the muscle activity showed more adaptation than the COM velocity[7]. The current work suggests that this is also the case for the COM displacement.

4.1.3 Generalization

The decreased plantar flexor muscle activity found for the one magnitude perturbations generalizes to a smaller magnitude of perturbations. It is shown that for the participants with a clear result of adaptation, the smaller magnitude perturbations showed the same trend which is a decrease in plantar flexor muscle effort after the training period. This indicates that a training period does not have to be for all different magnitude of disturbances, because it can affect responses to a different magnitude than the one during training period. However, this evidence is not very strong since perturbation detection accuracy was not high for all generalization trials. Detection for the generalization trials was around 50 % for P01 and P04 which is why these results are not very strong. In this case it could be that the perturbations did not cause a big enough loss of balance to detect it. For P02 all the perturbations were detected, which makes these results the most reliable. As pointed out in the results, the exoskeleton torque impulse is lower than for the 16 % BW trials, which makes sense since the perturbation is smaller so less assistance is needed to regain balance.

4.1.4 After Effects

After effects can also indicate whether a participant has adapted to the exoskeleton balance assistance or not. In this case the results for the after effects support the indications for the general adaptation trials. This is supported strongly again by P01 and P02 since they showed an increase in muscle effort in the unassisted Post16 trial compared to the assisted Post16 trial. This indicates that the participants need more muscle effort to remain balanced while they are perturbed. This is what was expected if a participant adapts to the balance assistance. It is shown that especially for P01 the GM muscle and the SOL muscle showed increased muscle effort once the balance assistance was removed. The participant needs to compensate for the assistance that was provided by the exoskeleton.

For P04 the opposite results are observed. This supports the conclusion that this participant did not adapt to the exoskeleton balance assistance. These results indicate that the participant was not relying on the exoskeleton to help him but might have been counteracting the assistance. Once it is removed, the muscles showed decreased muscle activity, which did not affect the COM displacement.

The increased COM displacement for P01 and P02 could indicate that without the exoskeleton the participants were less balanced even though they increased their muscle activity. This also supports that they adapted to the exoskeleton balance assistance.

Since the controller did not detect all perturbations that it should have, these unexpected unassisted perturbations can be compared to the expected unassisted perturbations(Unassisted Post16 trial). The unexpected unassisted perturbations did not show a clear pattern different from the ones that were detected throughout the assisted trials for the muscle of most participants. This was only the case for the SOL muscle for P03 as discussed before. This indicates that the other participants relied on the exoskeleton to assist. When it did not, they afterwards did not change their control strategy as shown by the assisted perturbations afterwards.

4.2 Limitations

The perturbations in this study were all applied at left heel strike, meaning the participant was still in double stance during the perturbation that was meant to make the participant lose balance. However, during double stance people are more stable than during single stance. This instant for perturbation onset was chosen because this would increase the time that the exoskeleton could provide assistance, since the exoskeleton could only provide plantar flexion assistance. Plantar flexion would only improve balance after anterior direction perturbation when applied to the leading leg. Applying perturbation at left heel strike instead of right toe off, which is at the end of double stance phase, would increase the time that the left leg is in front of the COM position so increase the time of assisting balance. However, it could be that when perturbations were applied at toe off thus in single stance, people would have adapted more since they would need more assistance to counteract the perturbation to regain balance. This also rises the question again whether the perturbations disturbed the balance enough to make people adapt[18]. However, as Bayon et al. already showed, the balance assistance assisted best when the perturbation was applied at heel strike instead of toe off[8]. Since the research question was to assess whether adaptation occurs when balance is assisted, heel strike seems the best time instant of applying perturbation for this study.

The participants in the work by Kao et al. were exposed to exoskeleton assistance for 60 minutes in total and showed adaptation after the first 30 minutes[20]. The two participants in this thesis showed adaptation after the complete protocol, with no clear point in time where it reached a steady state. Therefore, it is not known if more exposure would lead to bigger reductions of muscle effort.

Poggensee et al. also say people need time to learn to walk with exoskeletons to become experts and really adapt to it[12]. In their study they showed that it took approximately 109 minutes to adapt to the exoskeleton assistance. This indicates that the protocol used in for this thesis could have been too short to show adaptation, since it was approximately 50 minutes. However, Poggensee et al. studied adaptation to walking assistance which is less invasive than perturbing people and seeing how they react to that. Prolonging the duration of the perturbation protocol could potentially introduce additional undesirable factors, such as fatigue or discomfort at the site of perturbation.

Fatigue is a common factor to affect results[26]. In adaptation studies it is important to have a long enough protocol to be able to assess adaptation, but it still should be taken into consideration that fatigue can affect the results. In this case it is assumed that it did not affect the results since the participants walked on the treadmill for approximately 50 minutes. All participants were young healthy adults, so it can be assumed that they are able to walk in this setting without getting tired. In addition, the participants did not give any signs of getting tired by the protocol.

The experiments were conducted on young adults, but as described before these do not belong to the populations that would eventually be using an exoskeleton that assists balance. However, testing exoskeleton balance assistance on healthy adults tells us if and how healthy people adapt to it. This includes information on how healthy adults change their internal model to walk the best way. This indicates what people prioritize and this information can be used to design new or improve existing exoskeletons or balance control strategies to improve gait stability for people with impaired gait and or stability. This study showed that the plantar flexor muscle activity can be reduced when balance is assisted with an ankle exoskeleton that provides plantar flexion assistance. This could improve the walking ability of people with balance impairments since it would cost them less muscle effort.

4.3 Future recommendations

This thesis has given some insight into adaptation to exoskeleton balance assistance. It has also shown that there multiple factors that play a role when studying this adaptation. For future work, it would be interesting to see if more participants show the same results that these few participants have shown. A bigger set up of the experimental study would show more information on the adaptation of healthy people to exoskeleton balance assistance.

Furthermore, a more robust controller would increase the chance of getting a clear result and interpreting the results more easily. Improving the robustness of the controller could be done by making the threshold that was now previously determined subject specific as mentioned before. Adjustments that could be made to the controller used for conducting these results are discussed in the appendix.

Another way of improving the controller could be to use an inertial measurement unit (IMU) to estimate the COM velocity. In addition, an IMU could be used in a more realistic setting instead of a laboratory since it does not depend on motion capture systems and cameras. Westerdijk et al. have already shown that IMUs can be used to capture COM displacement during walking[27].

During this experiment, also kinematic data was acquired. It was beyond the scope of this thesis to analyze it, but this would be a recommendation to look at in the future. Gordon and Ferris looked at ankle angle during the assisted period and showed that this people also adapt in terms of this ankle angle after 24 minutes[21]. It would be interesting to see if adaptation also occurs for balance assistance for the ankle kinematics.

Emmens et al. investigated biological ankle torque for assisted walking versus unassisted walking[14]. This would be an outcome metric that could also indicate adaptation, because it would show how much ankle torque is exactly substituted by the exoskeleton.

5 Conclusion

The goal of this thesis was to assess whether healthy adults adapt to exoskeleton balance assistance. The results showed that healthy adults can adapt to exoskeleton balance assistance by decreasing their plantar flexor muscle activity. There are a few factors that are important to take into account when studying the adaptation to exoskeleton balance assistance. In this thesis the controller did not work exactly as planned, which created some difficulties. In addition to this practical part, another important thing to take into account for future research are individual differences. When more people are analyzed, the importance of this factor will also be more clear, but it has already been indicated in this small group of participants.

The study done for this thesis was able to combine knowledge of several studies done on human balance, exoskeleton balance assistance and adaptation to exoskeleton assistance. This thesis can be used as a starting point for investigating the adaptation to exoskeleton balance assistance further.

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A Results

A.1 Controller Performance

Examples are shown for normal detected perturbations and two examples of not detected perturbations. Both examples where the perturbation was not detected show why they were not detected. Figure 14 shows an example of the perturbation not being detected due to the $\dot{e}$ not exceeding the threshold. Figure 15 shows an example of the perturbation not being detected due to a falsely detected perturbation before the actual perturbation. A suggestion for solving the not exceeding threshold would be to use a baseline walking trial with walking to adjust the controller threshold. A code could be used to quickly run the first trials through and see what the detection accuracy would be. This case you would have a participants specific threshold. For P03 a lot more perturbations would have been detected if the threshold was put at 1.2 for example. However, it is a delicate balance since this participant already had 80 false positives included.

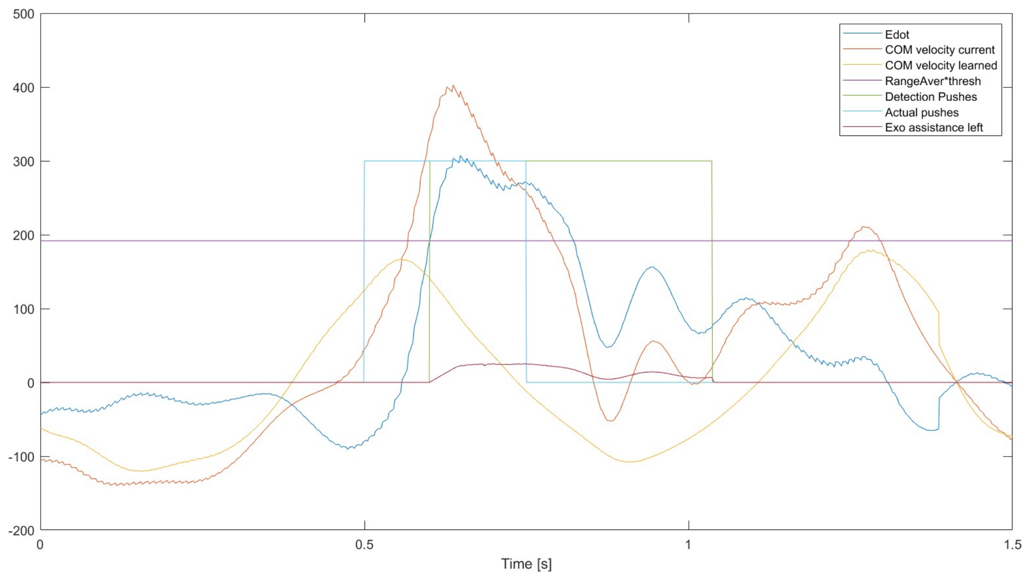


Figure 13: Example of detected perturbation Pert 16 of P01 during Train2

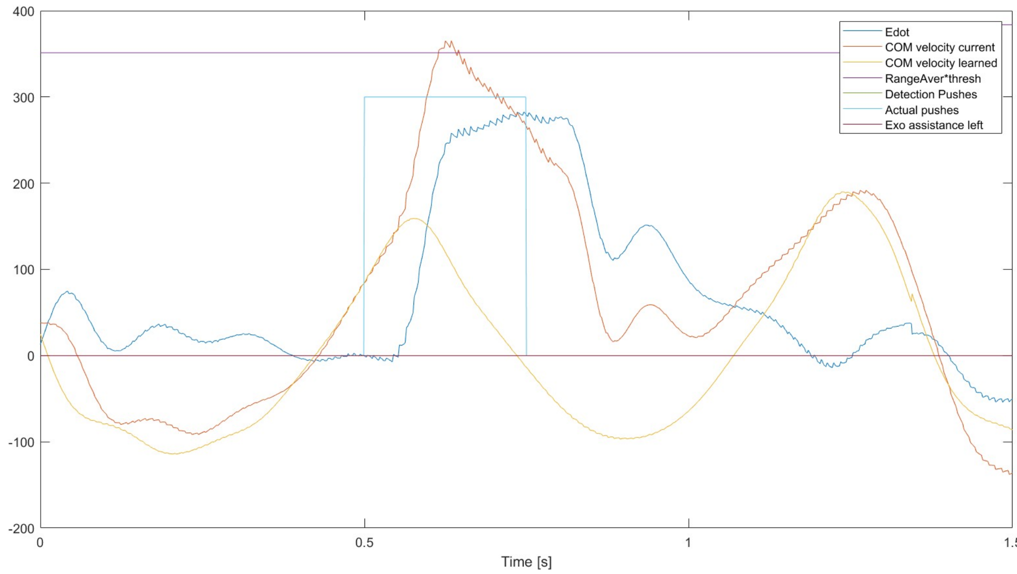


Figure 14: Example of undetected perturbation due to not exceeding threshold. Pert 8 of P01 during Train2

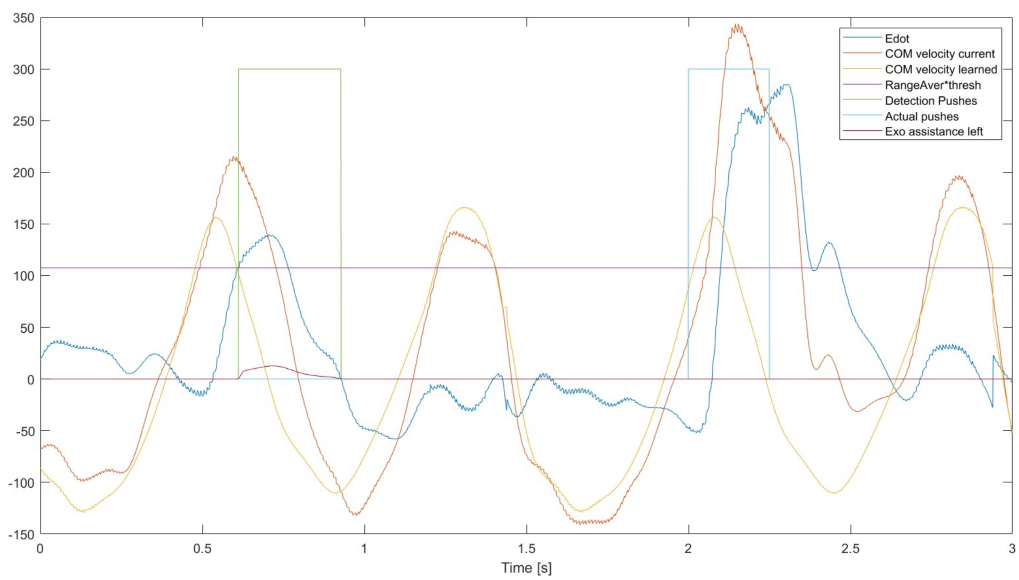
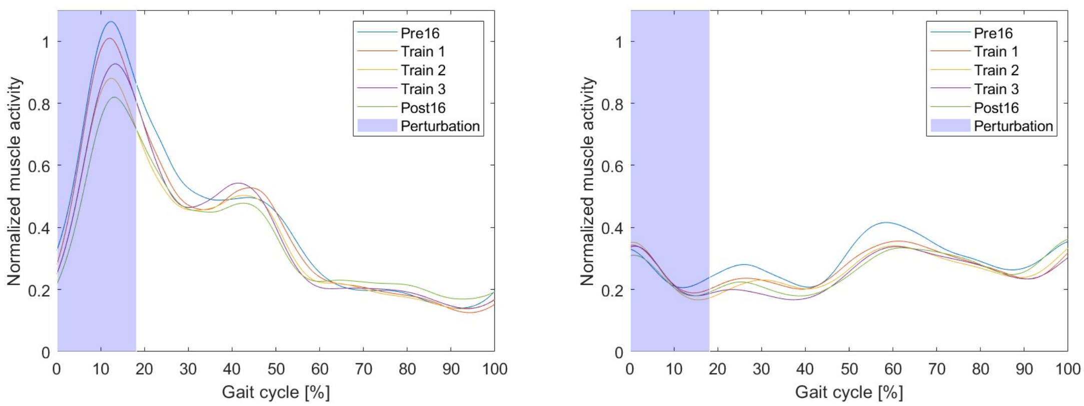


Figure 15: Example of falsely detected perturbation before actual perturbation happened. Pert 33 of P01 during Train2

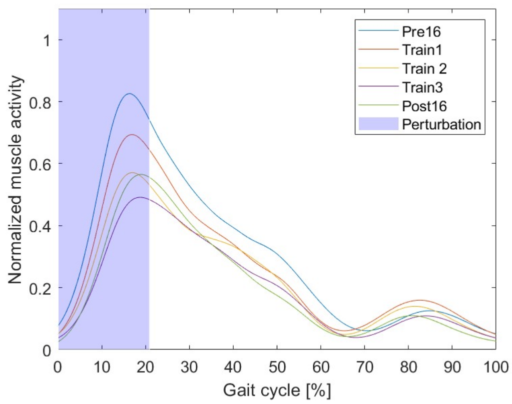
A.2 Time series for muscle activity



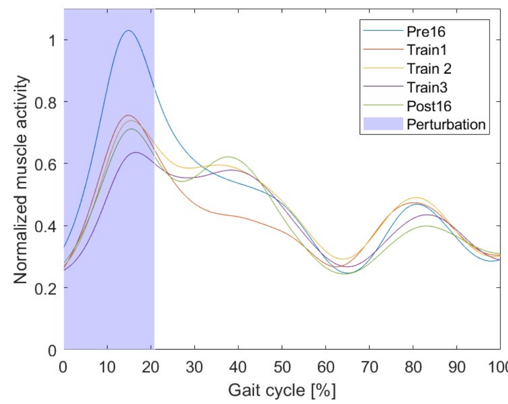
(a)

(b)

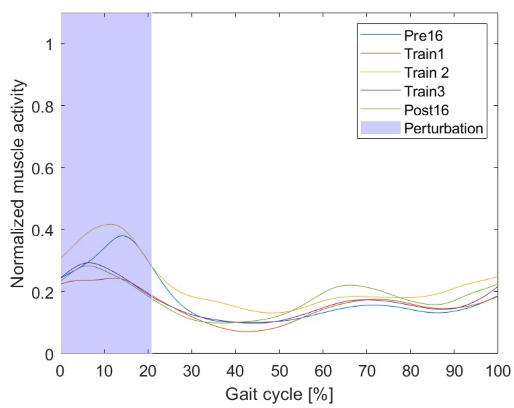
Figure 16: Average Muscle activity for P01 across all perturbation for Pre16, Train1, Train2, Train3 and Post16 for (a)SOL and (b) TA. Gait cycle starts and ends at left heel strike. Perturbation duration is indicated by shaded area.



(a)

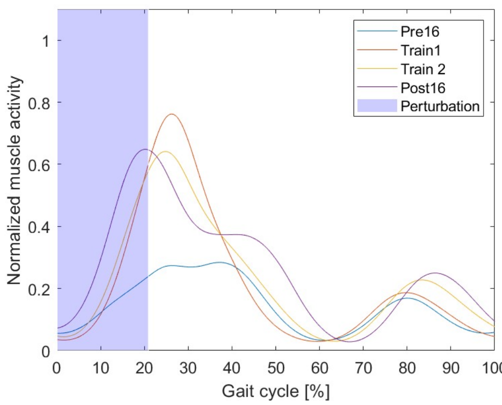


(b)

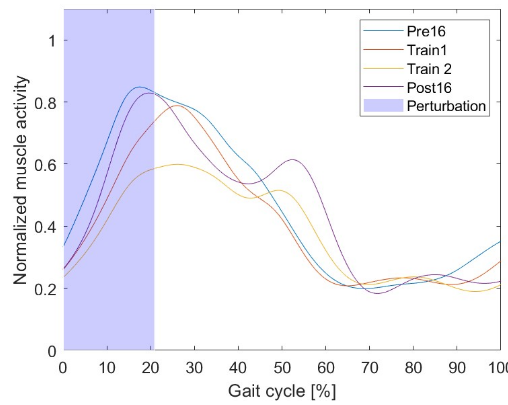


(c)

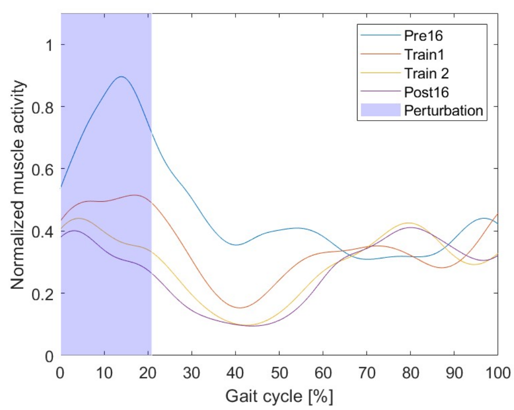
Figure 17: Average Muscle activity for P02 across all perturbation for Pre16, Train1, Train2, Train3 and Post16 for (a)GM, (b)SOL and (b) TA. Gait cycle starts and ends at left heel strike. Perturbation duration is indicated by shaded area.



(a)

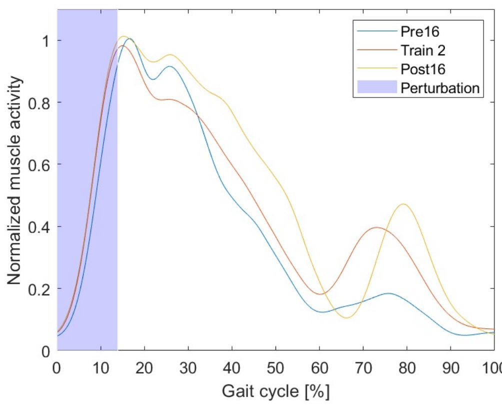


(b)

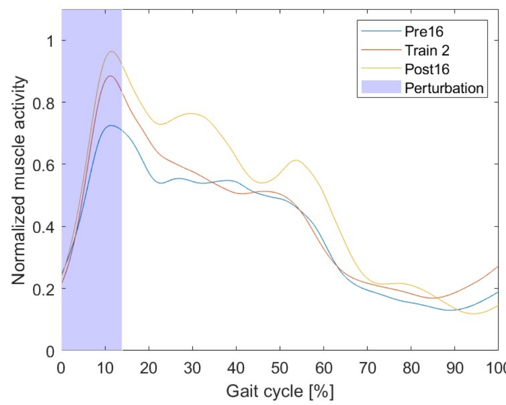


(c)

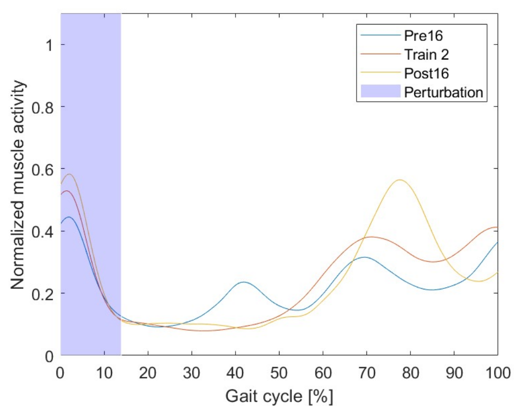
Figure 18: Average Muscle activity for P03 across all perturbation for Pre16, Train1, Train2 and Post16 for (a)GM, (b) SOL and (b) TA. Gait cycle starts and ends at left heel strike. Perturbation duration is indicated by shaded area.



(a)



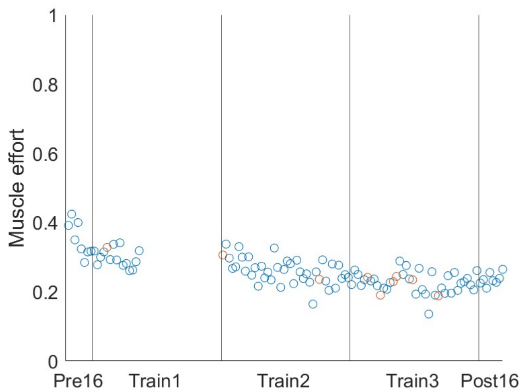
(b)



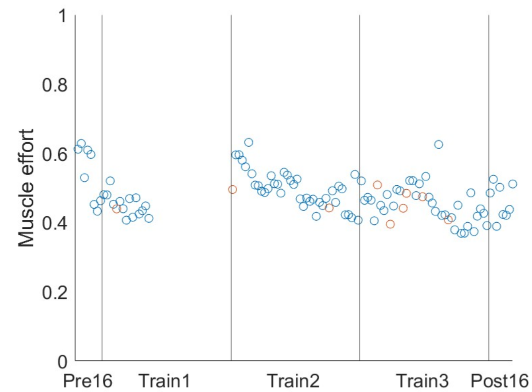
(c)

Figure 19: Average Muscle activity for P04 across all perturbation for Pre16, Train2 and Post16 for (a)GM (b)SOL and (c) TA. Gait cycle starts and ends at left heel strike. Perturbation duration is indicated by shaded area.

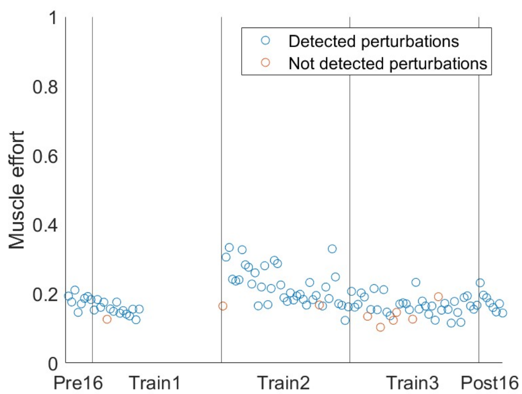
A.3 All perturbations chronologically plotted



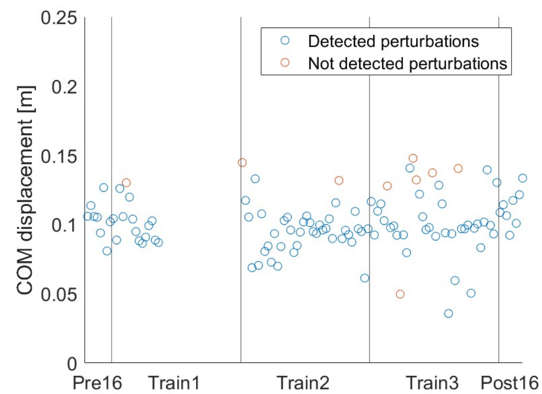
(a) GM



(b) SOL

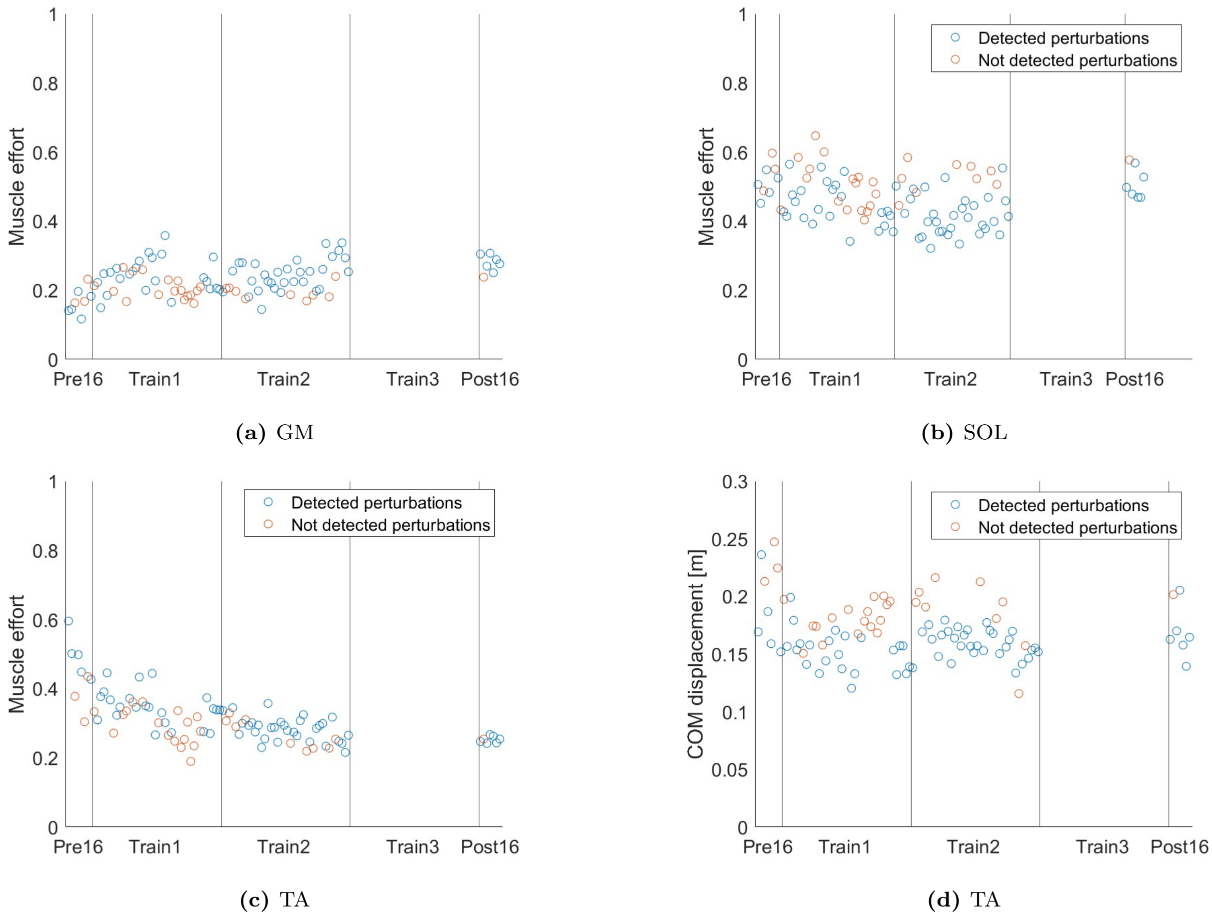


(c) TA



(d) TA

Figure 20: Muscle effort for each 16 % BW perturbation plotted for P02 for GM(a), SOL(b) and TA(c)



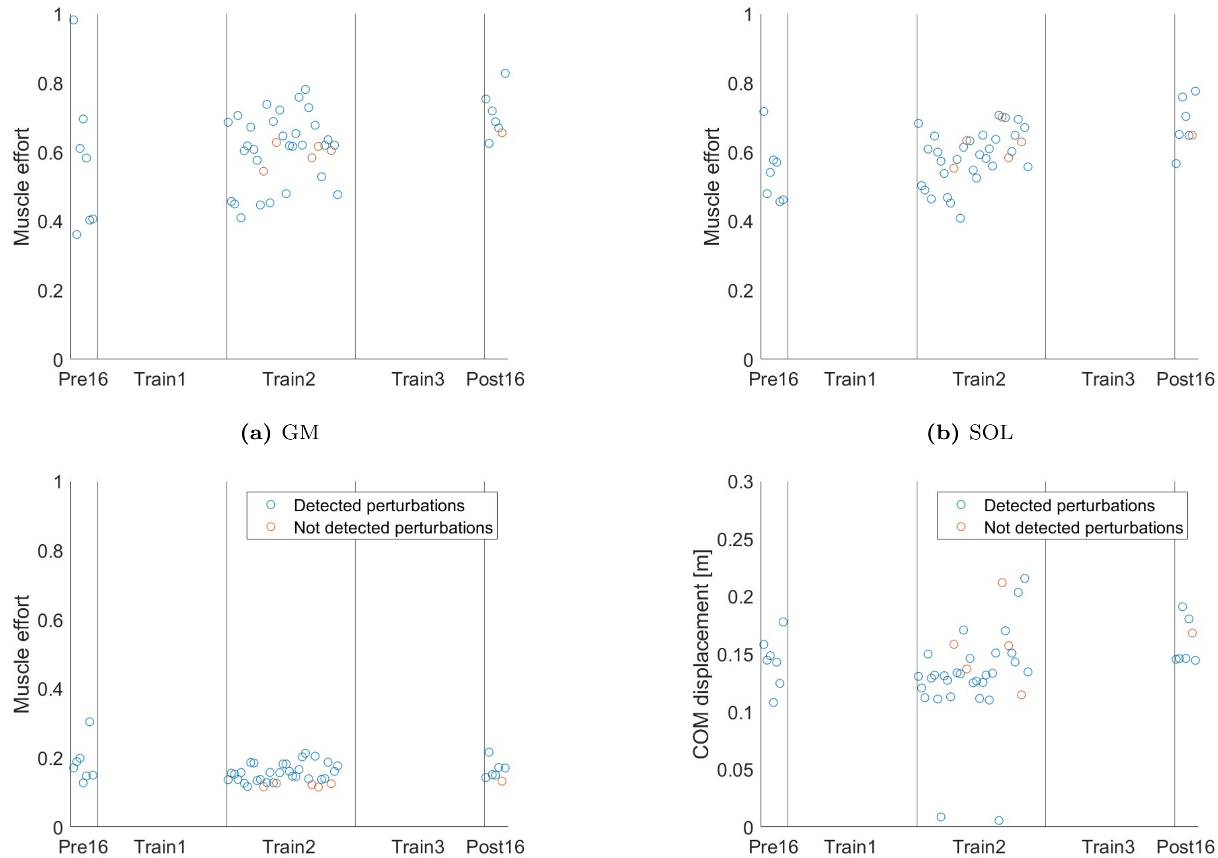
(a) GM

(b) SOL

(c) TA

(d) TA

Figure 21: Muscle effort for each 16 % BW perturbation plotted for P03 for GM(a), SOL(b) and TA(c). Train3 excluded, because it is not known which perturbations were detected.



(a) GM

(b) SOL

(c) TA

(d) TA

Figure 22: Muscle effort for each 16 % BW perturbation plotted for P04 for GM(a), SOL(b) and TA(c). Train1 and Train3 excluded because it was not known which perturbations were detected.