Reactive Balance Recovery during Perturbed Walking in Incomplete Spinal Cord Injury Patients

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Abstract

Introduction The incomplete spinal cord injury patient group experiences many falls each year. Previous research indicates that these frequent falls are caused by an impaired reactive balance strategy. However, there is a lack of research into the reactive balance strategy in the iSCI patient group. The goal of this research is to study the reactive balance control responses of individuals with iSCI compared to age- and gender-matched participants via characterizing muscle activity responses after unexpected pushes.
Methods The reactive balance strategies of iSCI participants were assessed using linear push and pull perturbations at the pelvis in anteroposterior and mediolateral direction during treadmill walking at comfortable walking speed. Five iSCI participants and five age- and gender-matched able-bodied participants were included. The muscle onset, muscle effort, Center of Mass (COM) velocity and the effort-COM velocity relation were assessed.
Discussion and Conclusion On the contrary to the expectations, the COM velocity was not more affected for iSCI participants. One of the iSCI participants exhibited delayed muscle onset relative to controls for the m. Soleus and the m. Gastrocnemius Medialis onset in the stance leg after a push forward at left toe-off, but no clear delayed onset was found for other conditions. For the muscle effort, the iSCI participants showed a weaker modulation for the effort of the m. Rectus Femoris and the m. Biceps Femoris over increasing perturbation magnitude and increasing COM velocity. A limitation of this study is the small participant population. Therefore the results of this research can not be translated as a conclusion for the whole iSCI population.

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Contents

1 Introduction

The occurrence of falls in the incomplete spinal cord injury (iSCI) patient group is a major problem[1]. Approximately 78% of the iSCI patient group experiences at least one fall each year[2], affecting both the mental and the physical health of the patient. Incomplete spinal cord injury affects the sensory information flow[3] of the patient and the ability to control the muscle activity that is involved in maintaining balance during gait. This affects the quality of ambulation and increases the risk of falling[4]. The disruption of the sensory information flow and muscle weakness in this patient group are an important reported cause for the loss of balance and the occurrence of falls[1][4][5][6]. Falls can result in serious health consequences, such as fractures, sprain or dislocation of joints. The high fall rate and the possibility of injuries after a fall can lead to embarrassment, frustration, and even fear of falling[4][7], which decreases the amount of walking activity. In addition, the embarrassment and the fear of falling can lead to a decrease in participation in society and community and affect the physical and emotional well-being of the patient[8].

To regain balance after an unexpected perturbation, the reactive balance response is activated. Even though the balance strategies[9][10] during normal walking are altered by the iSCI patient group using a slower walking speed or smaller step length to compensate for the impaired sensorimotor information flow, there is a high percentage that experiences at least one fall each year[11]. To mitigate these falls during walking and improve the overall quality of life of the iSCI patient group, knowledge is needed on how the balanced gait is affected by the iSCI. A better understanding of where the reactive balance responses are lacking to regain balance after unexpected balance loss during walking in this patient group can contribute to the development of new technologies that improve balance during gait.

In healthy gait, encountered perturbations are opposed using the reactive balance response. The reactive balance response has different strategies: the ankle strategy[12], hip strategy[13], or the stepping strategy[14]. The stepping strategy is often considered as the primary strategy[15] for recovering balance during gait, especially with perturbations[16] in mediolateral (ML) direction. If the balance during gait is perturbed in the ML direction, the center of mass (COM) is shifted from the base of support (BOS) in the direction of the perturbation force. The stepping strategy[17][18] alters the step length and step width in order to position the foot, which is the BOS, such that the center of pressure (COP) is positioned posterior and medial to the extrapolated center of mass (XCOM) to prevent falling over. When using the hip strategy[13], the COM is shifted back above the BOS by rotation of the upper body using flexion or extension of the hips. This strategy becomes best visible when the stepping and ankle strategy possibilities are blocked. Vlutters et al., examined the preference for hip strategy or stepping strategy during perturbed walking, using pin shoes to block the ankle strategy[15]. They found that there is a low priority for the hip strategy to counteract both mediolateral and anteroposterior perturbations during walking. The last strategy, the ankle strategy[16], is the modulation of the ankle torque to counteract mainly perturbations in the anteroposterior (AP) direction.
Muscle activation responses are needed to adjust the ankle torque, rotate the hip or make a balance correction using the stepping strategy.

While the reactive balance responses in the iSCI group have not been examined much, healthy participants have been studied on reactive balance responses. Vlutters et al. applied unexpected pelvis perturbations in mediolateral and anteroposterior direction during walking at the instance of toe-off[19]. These pelvis perturbations were applied using a hip brace which was horizontally connected to a motor and were applied at a magnitude of 8, 12, and 16% of the participant's body weight while walking on a dual-belt treadmill. Muscle responses were collected using EMG sensors and kinematic data was collected using a 12-camera Motion Capture System. From this study the knowledge was obtained that muscle activation[19] scales with perturbation magnitude and direction. Depending on the perturbation direction and magnitude, multiple actions using muscle contractions and joint torques are needed to recover balance after perturbation and return to the unperturbed gait cycle. Following ML perturbations[19], a clear onset of the Gluteus Medius was detected for both the swing and the stance leg and a clear onset of the Tibialis Anterior of the stance leg was detected. For the AP perturbations in the stance leg, a clear onset of the Tibialis Anterior was detected for the pulls backwards and a clear onset for the Gluteus Medius with the forward pushes. For the stance leg, increased co-contraction was shown, increasing with the magnitude of the perturbations. Supra-spinal neural[19] structures seem to be involved since there was latency detected for the muscle activation after the perturbation onset.
In addition to muscle onset, the muscle activation magnitude to counteract the perturbation increased with increasing perturbation magnitude. This indicated that increasing muscle effort is needed to counteract perturbations with increasing magnitude.

During walking, sensory information is continuously transformed to descending motor commands that adapt the leg muscles' activity. Afschrift et al. showed that the feedback from COM kinematics during perturbed walking using platform translations not only described the reactive ankle muscle activity but also the reactive ankle muscle activity [20]. This suggests that in able-bodied individuals, the central nervous system uses feedback from the COM kinematics to control the reactive muscle activation to maintain a stable gait after perturbation.

Although the reactive balance is important for fall prevention after an unexpected perturbation, there is a lack of research into the reactive balance recovery of the iSCI patient group during perturbed gait. As the afferent and efferent pathways within the nervous system are damaged, producing adequate muscle force as reactive balance response can be affected. In the research of Thigpen et al, the reactive balance response of iSCI patients was assessed using a platform that can rotate and translate unexpectedly[21]. Here, the ankle muscles m. Tibialis Anterior and the m. Gastrocnemius Medialis of the iSCI patient group shows a delayed onset of electromyographic (EMG) activity[21] compared to healthy participants.
Aurora et al. investigated the balance responses after an unexpected slip and found that iSCI patients had a limited ability to increase their margin of stability (MoS), as well as a smaller m. Soleus activity around 200-500 ms after the slip perturbation compared to the healthy subjects[11]. Also, they found a difference in the activation pattern of the trailing foot, with the healthy control group activating the m. Tibialis Anterior first and the iSCI patient activating the m. Gastrocnemius Medialis first. Additionally, a decrease in the ankle plantar

flexor muscle activity at the terminal stance was observed.
In the research of Chan et al.[22], the reactive step responses in iSCI patients in a lean-and-release test were studied. From this, the knowledge is obtained that iSCI patients need more steps to recover from perturbed balance[23].

Previous research provides evidence that the balance response is impaired in the iSCI patients. However, only Aurora et al. examined the reactive balance to perturbations (slips) during walking. The reactive balance was examined in the AP direction and only the lower leg muscle responses were measured. Since slip perturbations are not the only perturbation types iSCI patients might encounter, knowledge about the muscle response of the leg to perturbations is not complete. The muscle activation[24] reflects the reactive responses to correct for perturbations while keeping a stable and controlled gait. Therefore, it is beneficial to obtain a better understanding of how the leg muscles are activated in iSCI patients to recover balance after an unexpected perturbation during gait.

The goal of this study is to obtain knowledge about the reactive balance strategies of iSCI participants and where this strategy lacks to regain balance after unexpected perturbations compared to able-bodied individuals. As the balance response for the iSCI individuals might be impaired due to inadequate muscle activation[1][4][5][6], it is expected to find a delayed onset of muscle activation compared to the able-bodied group. In addition, it is expected to see modulation of faster muscle onset and higher effort with increasing perturbation magnitude for the able-bodied participants[19], but less modulation for the iSCI individuals. Even though the pro-active balance strategies improve the gait stability of the iSCI individuals, it is expected that they are more affected by the direct mechanical effects of the perturbations and show high COM velocities compared to the able-bodied group. As a higher COM velocity indicates more effect of the perturbation on the balance, it is expected to see an increasing muscle effort to counteract the perturbation with increasing COM velocity. This modulation is expected to be less for the iSCI individuals compared to the able-bodied individuals.

2 Methods

In this study, the reactive balance strategy of 5 individuals with iSCI and 5 able-bodied individuals was assessed. A similar setup[19] as Vlutters et al. will be implemented, using a hip brace to apply pelvis perturbations in mediolateral and anteroposterior direction during walking at comfortable walking speed on a dual-belt instrumented treadmill[19]. As a measure of how fast the muscles are activated after perturbation, the muscle onset after perturbation is examined. In addition, the leg muscles produce the force to counteract the perturbation force. Therefore, the effort of the reactive muscle response to counteract the pushes is calculated. Since the reactive ankle muscle activity during perturbed walking can be described by the COM kinematics[20], the relation between the effort and COM velocity after perturbation is examined as well.

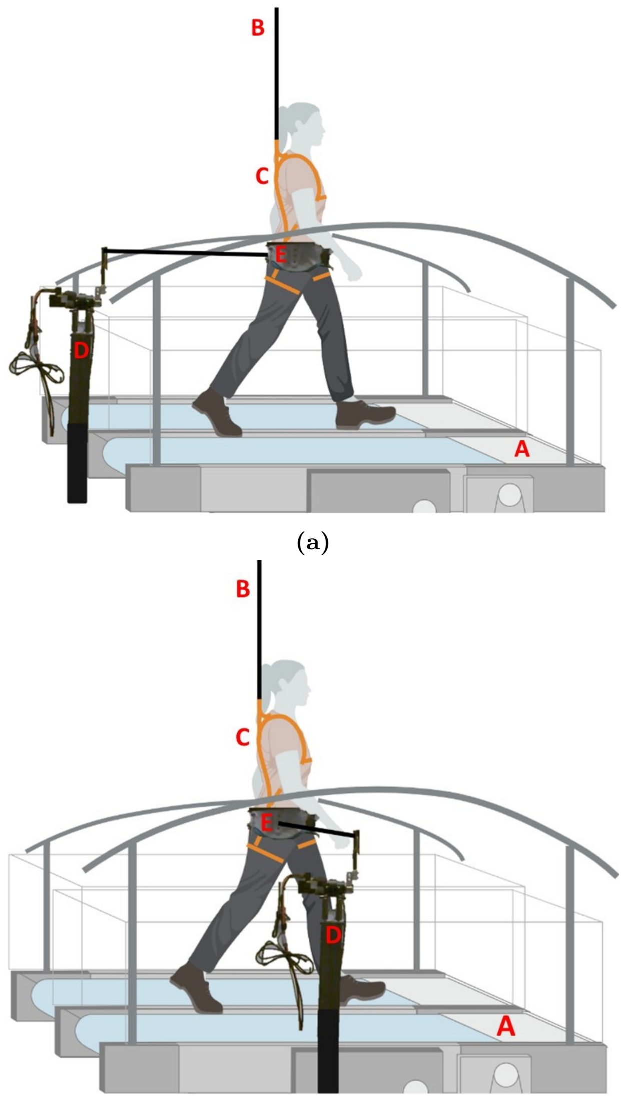

Figure 1: The participants walked on the dual-belt treadmill (A) with safety attachment to the ceiling (B) using a harness (C). The motor(D) is horizontally connected to the hip brace(E), which provides pushes in forward and backward direction (Figure 1a) or pushes in leftward and rightward direction (Figure 1b) at the pelvis.

2.1 Participants

This study was an on-site study with five incomplete spinal cord injury patients, with American Spinal Injury Association Impairment Scale (AIS) C or D, and five age- and gender-matched able-bodied participants. The patients needed to be at least 12 months after surgery, to ensure a stable neurological condition. In addition, all the participants($\geq$ 18 years) needed to be able to walk independently for at least 5 minutes without a walking aid and the walking speed inclusion was between 0.4 m/s and 1.0 m/s. The included age- and gender-matched participants should not have any musculoskeletal health problems. All experimental protocols were approved by the University Ethics Committee under application no. 230048, and all participants gave their written informed consent. A table with the demographics of the participants is shown in Appendix A.1.

2.2 Experimental Setup

In this experiment, the reactive walking balance was assessed during walking on an indoor treadmill using perturbations at the pelvis. Figure 1 depicts the experimental setup. The treadmill on which the participants walked at a slow walking speed, was a dual-instrumented belt treadmill (MotekForce Link, Culemborg, Netherlands). During the trials, the participants' treadmill walking was perturbed by perturbations in ML and AP direction, received at the pelvis provided by two motors (SMH60, Moog, Nieuw-Vennep, The Netherlands). One motor was placed behind the participant. The second motor was placed on the right side of the participant, adjacent to the treadmill. The motors were positioned such that a horizontal connection could be made by an aluminium rod via ball joints from the motors to the hip brace (Distrac Wellcare, Hoegaarden, Belgium), worn by the participant. The applied perturbations were controlled using an admittance controller, which minimized interaction forces during normal walking, as described in van der Kooij et al (2022)[25]. The torque and position data were recorded and implemented in Simulink (Matlab 2016b, Mathworks, US) and Twincat. During the whole experiment, the participant wore a safety harness that was attached to the ceiling to prevent injury if falling.

2.3 Experimental Protocol

The five individuals with iSCI (C/D) and five healthy age-matched subjects were explained what was going to happen during the trial and gave written informed consent. The length, leg length and the weight of the participant were measured. The leg length was used to calculate the desired slow treadmill speed as 0.63 m/s scaled to the square root of the leg length of the participant, to normalize for leg length. The experiment started with completing the Activity-specific Balance Confidence scale (ABC) test, which contained questions about their confidence in maintaining balance during (daily life) tasks in and outside their house. In this test, the participant had to give a percentage (0 - 100%) of how confident they were to maintain balance during a specific task, see Appendix A.2. For each participant, the mean confidence percentages can be found in Appendix Table A6. After finishing the ABC test, adjustment of the Pelvis Perturbator was made to place the perturbator at the correct height for the participant, if needed.

Next was the familiarization with the treadmill. If the walking speed on the treadmill was uncomfortable for the participant, the comfortable walking speed was measured and noted, but the trials were all performed at the slow walking speed of the normalized 0.63m/s.
The skin was prepared for the EMG measurements by shaving the leg hair (if needed) and using the skin prep gel from NuPrep. For each of the locations on the leg, the skin was shaved and cleaned. The EMG sensor placements are described in Section 2.4. In preparation for the motion capturing, the marker clusters and single markers were placed.
Prior to the perturbed walking trials, baseline walking for both the anteroposterior and the mediolateral direction was measured for 1 minute in which the motors were in zero impedance mode. Between these baseline walking trials, six blocks of 5-minute perturbed treadmill walking were recorded, containing 12 perturbations each, see Figure Appendix A.3, Figure A.11 and Appendix A.4 for the protocol overview. In the six pusher conditions, three of these condition blocks contained mediolateral pushes and three of the blocks contained anteroposterior pushes. The order in which the mediolateral or anteroposterior condition trials were performed was randomized and therefore differs for each participant. The 12 pushes were divided into three magnitudes: 8% (low), 12% (mid) and 16%(high) of the body-weight. Four perturbations of each magnitude were applied in each trial: two at left toe off (one push, one pull) and the other two at right toe off (again one push, one pull). Between the trials, the participant was allowed to take a rest for as long as they needed. After the perturbations trials, baseline walking was measured again, to enable a post-measurement check for the occurrence of muscle fatigue or check if the hip brace had moved.

2.4 Data Collection

The electromyographic data was captured using the Bagnoli Desktop EMG system (Delsys, Natick, MA USA). Bipolar surface electrodes were attached to the subject's skin according to the recommendations for the surface electromyography (sEMG) sensor locations of Seniam. Only the right leg muscle activations were measured, see Figure 2. The muscle activity of the Tibialis Anterior (TA), Gastrocnemius Medialis (GM) and Soleus (SOL) were measured, as these muscles are responsible for dorsi- and plantar flexion of the foot and adjusting the ankle torque during walking. In addition, the activity of the Biceps Femoris (BF) and the Rectus Femoris (RF) were measured, as these muscles cause knee flexion and extension. Lastly, the Adductor Longus (AL) and the Gluteus Medius (GLM) were measured because these cause hip adduction and abduction during walking. A 12-camera motion capture system (Oqus 600+, Qualisys, Göteborg, Sweden) was used to measure kinematic data. Eight marker clusters, containing four markers each, were placed on the upper arms, forearms, upper legs and lower legs. Single markers were placed on the seventh cervical vertebra, Sternum, right and left Clavicle, Acromium, Lateral and Medial Humerus Epicondyle, lateral and medial wrist, Anterior and Posterior Superior Iliac Spine, Trochanter, Lateral and Medial Femoral Epicondyle, Medial Malleolus, Calcaneus and on the 1st and 5th Metatarsal. In addition, single markers were placed on the brace: left and right, Anterior and Posterior Superior Iliac Spine and on the motor-rod connection. The EMG was collected at 2 kHz for all the participants, except iSCI participant 1 (P01) was collected at 1 kHz. The motion capture data was collected at 100 Hz and the sensor data of the motors was collected at 1 kHz. The data types were synchronized using the Qualisys Track Manager (QTM) software.

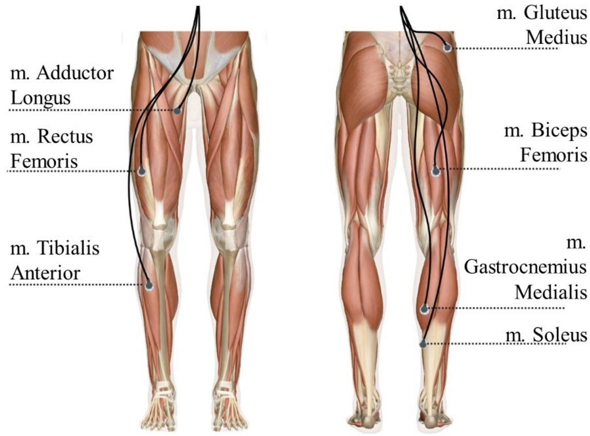

Figure 2: EMG placement on right leg.

2.5 Data Processing

The marker data was labelled with the QTM software. The QTM files were converted to MATLAB files, to enable further data processing in MATLAB version R2022B. Since human movement is in the frequency range of 0 to 20 Hz, the marker data signal was filtered with a 4th-order lowpass Butterworth filter with a cut-off frequency of 20 Hz[16]. From the raw EMG data, the linear envelopes were obtained. First, the offset of the raw signal was subtracted. Second, the lower frequency noise, like movement artifacts and the higher frequency noise was filtered out using a bandpass filter with cutoff frequencies 10 Hz and 500 Hz. A notch filter was used at 50 Hz to filter out the background noise, such as motor vibrations and electromagnetic radiation. After this, the EMG signals were rectified. Lastly, the EMG envelopes were formed using a 2nd-order lowpass filter with a cutoff frequency of 20 Hz. The EMG voltages were normalized to the maximum of the median stride during baseline walking.

During the trials, pushing the participants caused stepping over on the dual belt. This caused the ground reaction forces (GRF) to be less accurate to use for heel strike and toe-off detection. Therefore, the stance phase and swing phase were detected from the motion capture data, by using the data of the foot marker as in Zeni et al.[26]. As all the data was synchronized, the timestamps of the toe-off left and right were saved and used to extract the gait cycles from the toe-off to the next toe-off in the EMG data.
From the extracted unperturbed gait cycles, the outliers were detected and removed. This was done using percentiles because standard deviations might be affected by the outliers. The EMG data was interpolated over 100 samples and from this data, the interpolated EMG data percentile arrays were calculated for 25% and 75%. This means, that for each of the 100 samples, the 25% and 75% values of the gait cycles are calculated. This creates two arrays of 100 samples with the gait cycle values at the percentiles during the gait cycle trajectory. The range was calculated between the two percentile arrays and used for upper- and lower-bound calculation. If the EMG signal went outside one of the two bounds for at least 4 of the 100 samples, the signal was considered an outlier. The

lower-bound was set as 1.5 times the range array subtracted from the 25th percentile array and the upper-bound as 1.5 times the range array added to the 75th percentile array, see Figure 3. The detected outliers are flagged and removed from the unperturbed gait cycle structure. From the remaining gait cycles, the mean gait cycle and standard deviation were calculated for each muscle of the participant.

From the extracted perturbed gait cycles, the muscle onset was examined as a measure of how fast the muscles were activated after perturbation. The onset time was determined as the absolute timestamp after perturbation onset at which the normalized EMG voltage of the muscle deviated from the mean of the normal (unperturbed) strides during perturbation trials with at least three times the standard deviation of the unperturbed EMG, see Figure 4. Initially, the unperturbed strides during baseline walking were used for the normal strides, but these gave a lot of onset detection between 0-25ms (Appendix A.5, Figure A.14 and Table A7). This is physiologically not possible [27], but was caused because the perturbed signal was higher in normalized EMG value than the baseline walking for some muscles, see Appendix Figure A.15, A.16 and Table A8. This was probably because the participants knew perturbations were coming, which caused already some tension in the muscles for stiffer walking as a proactive balance strategy to counteract the perturbations that were coming. Thus, instead of the unperturbed baseline walking, unperturbed strides during perturbed walking were used. The unperturbed strides were the detected strides right before perturbation onset. On these strides, the same outlier detection that was used for the baseline walking was used to detect outliers. From this, the mean and the standard deviation were calculated to find the mean+3std threshold. When the onset was detected between 0 - 25ms, the onset was removed from the results. In addition, if the onset was detected above 500 ms after perturbation, the onset was not included in the results because such a delayed onset doesn't capture the reactive balance response [19][28].

As the leg muscles produce the force to counteract the perturbation force exerted by the pusher, the magnitude of the reactive muscle response to counteract the pushes was calculated. The integral of each muscle signal was calculated over the time between perturbation onset until 500ms after perturbation to capture the reactive response[19][28]. The muscle effort to counteract the perturbations was calculated by subtracting the integral of the unperturbed baseline walking (integral over toe-off to 500ms after toe-off) from the surface area of the perturbed signal.

The COM velocity was estimated using the position of the four hip markers (left and right for anterior and posterior Superior Iliac Spine). The position of these markers was differentiated to velocity. The COM velocity was then calculated as the mean of the velocity from the four hip markers, at the end of the perturbation duration (150ms) to see the direct mechanical effect of the push on the COM velocity.
To see if there is any modulation over the different perturbation magnitudes for the calculated muscle onset, muscle effort and COM velocity, the linear regression was calculated for each of the participants. The independent variable was the perturbation magnitude in the percentage of body weight and the dependent variables were the calculated muscle onset, muscle effort and COM velocity.

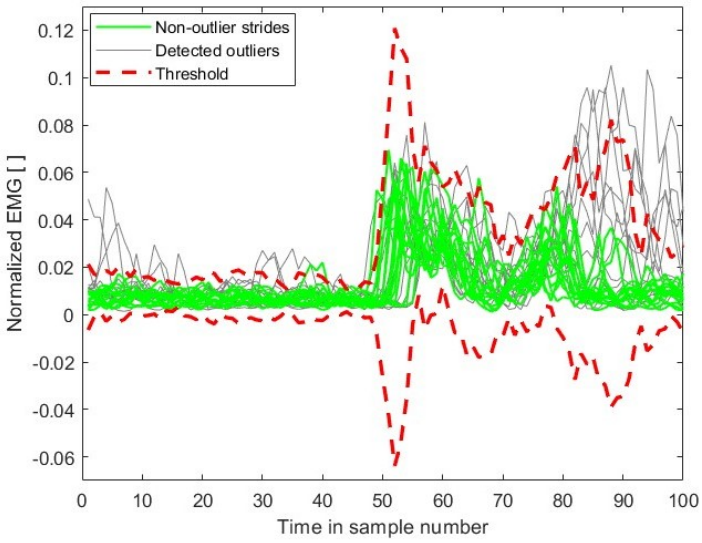

Figure 3: Outlier detection from the TA of the stance leg an iSCI participant. The green gait cycles are considered non-outliers, while the grey gait cycles cross the calculated upper- and lower-bound (red) for more than 4 samples.

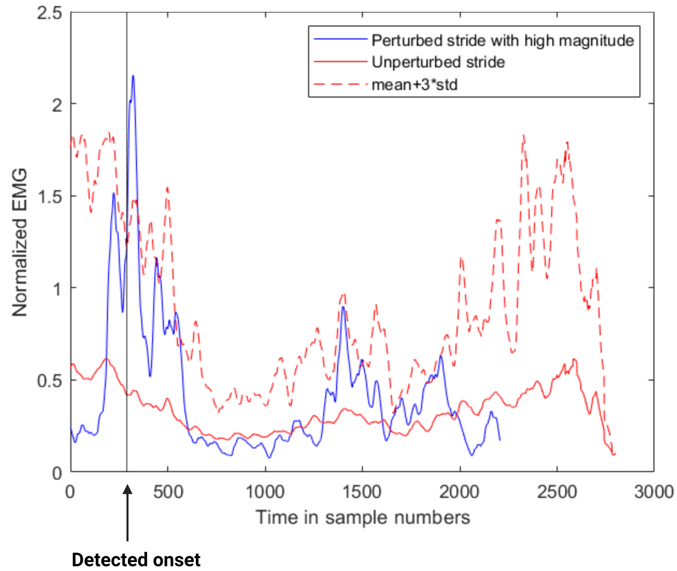

Figure 4: Onset detection method. Visible is the perturbed stride (blue line) from the TA muscle in the stance leg after a pull backwards with the highest magnitude. If the stride crosses the threshold (red dashed) 3std of the mean unperturbed walking stride (red line), the onset was detected (black vertical line).

3 Results

The reactive balance responses of a group of five iSCI patients and a group of five age- and gender-matched able-bodied participants were assessed following perturbations in AP and ML direction while walking on a dual-belt treadmill. The reactive EMG response was characterized using muscle onset and muscle effort for perturbations in four different directions (AP, ML) and three different magnitudes. In addition, the effect of the perturbation on the COM velocity and the muscle effort - COM velocity relation was analysed.

As there were a lot of perturbation conditions measured, this study focused only on a few perturbation conditions. This study focused on the perturbation conditions in the AP direction because there were some differences visible in the assessed EMG parameters for some of the muscles in the anteroposterior direction. In the result sections, the assessed parameters of the muscle activity of the TA, SOL, RF and BF are shown, as these were expected to give a clear muscle burst after perturbations in the AP direction. The focus was on the stance leg (left toe-off) since there were stronger expectations for the stance leg in anteroposterior perturbation direction. The conditions that were shown are the SOL and the BF after perturbation in forward direction, and the TA and the RF after perturbation backwards direction. For muscle onset, the GM after a push forward showed a clear difference for some of the iSCI patients compared to the able-bodied group and is therefore included in the results as well. In addition, for the muscle effort, the GLM effort after perturbation in a backwards direction was also included because this result showed a difference between the two groups. Unfortunately, the data collection of one of the patient participants (P04) is excluded from the analysis because this participant needed to hold the handrails to withstand the perturbations. This affects the reactive balance responses in the legs and is therefore left out. In addition, for one of the able-bodied participants (C02) the GLM signal during the perturbation trials was hardly visible, compared to the baseline walking and the GLM signal of the perturbation trials. Therefore, the GLM data of C02 is excluded from the analysis as well.

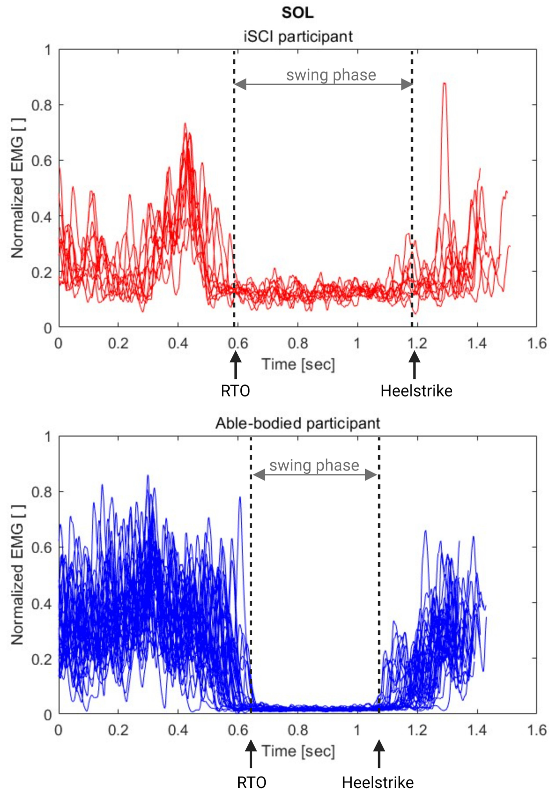

Figure 5: SOL envelopes of the right leg from midstance to midstance during unperturbed walking of P05 (red) and age- and gender-matched participant C05 (blue).

3.1 Representative Gait Cycles

From the baseline walking, the gait cycles were detected from toe-off to toe-off for both the left and the right foot, using the foot marker data. The gait cycles of the perturbed walking were detected from toe-off to toe-off as well. The obtained muscle envelopes of the SOL and the TA of iSCI participant P05 and age- and gender-matched able-bodied participant C05 during unperturbed baseline walking are shown in Figure 5 and 6. The gait cycles of the muscles of P05 and C05 showed similar activation patterns during the gait cycle phases. This was visible for all the iSCI participants and age- and gender-matched participants.

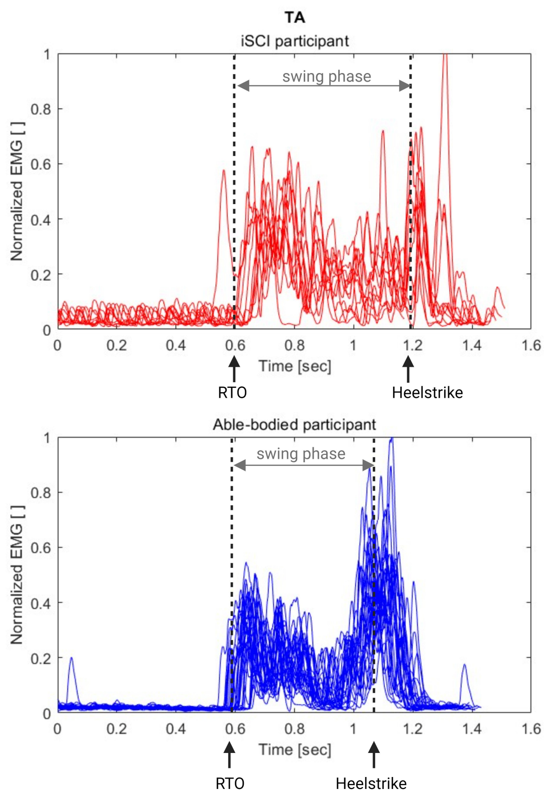

Figure 6: TA envelopes of the right leg from midstance to midstance during unperturbed walking of P05 (red) and age- and gender-matched participant C05 (blue).

3.2 Muscle Onset

The muscle onset was calculated as the timestep at which the perturbed EMG signal exceeded the mean EMG signal of the normal strides of perturbed walking plus three times the standard deviation of the normal strides. In this section, the results of the muscle onset detection are shown for the SOL and BF after a push in the forward direction and for the TA and RF after a push in the backward direction. An overview of other muscle onset results is depicted in Appendix A.6.

3.2.1 Muscle Onset after push forward

After a push in the forward direction, P01 and P02 showed a delayed onset of the SOL compared to all the participants of the able-bodied group, see Figure 7. In addition, P01 showed a delayed onset of the GM compared to four of the able-bodied group, see Appendix A.19.

Three of the four iSCI participants and one out of five of the able-bodied group showed faster SOL onset for higher perturbation magnitude. In addition, two of the iSCI participants and two of the able-bodied participants showed negative modulation of BF onset timing for increasing perturbation magnitude, see Figure 7.

3.2.2 Muscle Onset after pull backward

After a pull in the backward direction, C01 showed delayed TA onset for mid and high perturbation magnitude compared to all the other participants of both groups. As shown in Figure 7, faster onset detection of the TA was seen for P01, compared to all participants of the able-bodied group. In addition, it was visible that P01 showed also faster RF onset for the mid and high perturbation magnitude compared to the participants of the able-bodied group.

Four of the five able-bodied participants and two of the three iSCI participants showed negative onset modulation over increasing perturbation magnitude. For the RF onset, all of the iSCI participants and four of the able-bodied participants showed negative modulation of onset over increasing perturbation magnitude. This modulation was shown weaker for able-bodied participants compared to the three iSCI participants, see Table 1.

Table 1: Range of regression slopes [min max] intra-participant groups of the muscle onset after perturbation in the AP direction at LTO.

Muscle	Patient	Able-bodied
SOL	[-0.0398 0.0311]	[-0.0147 0.0546]
BF	[-0.1270 0.1312]	[-0.0512 0.0097]
TA	[-0.0205 0.0022]	[-0.0128 0.0992]
RF	[-0.0796 -0.0177]	[-0.0332 0.0141]

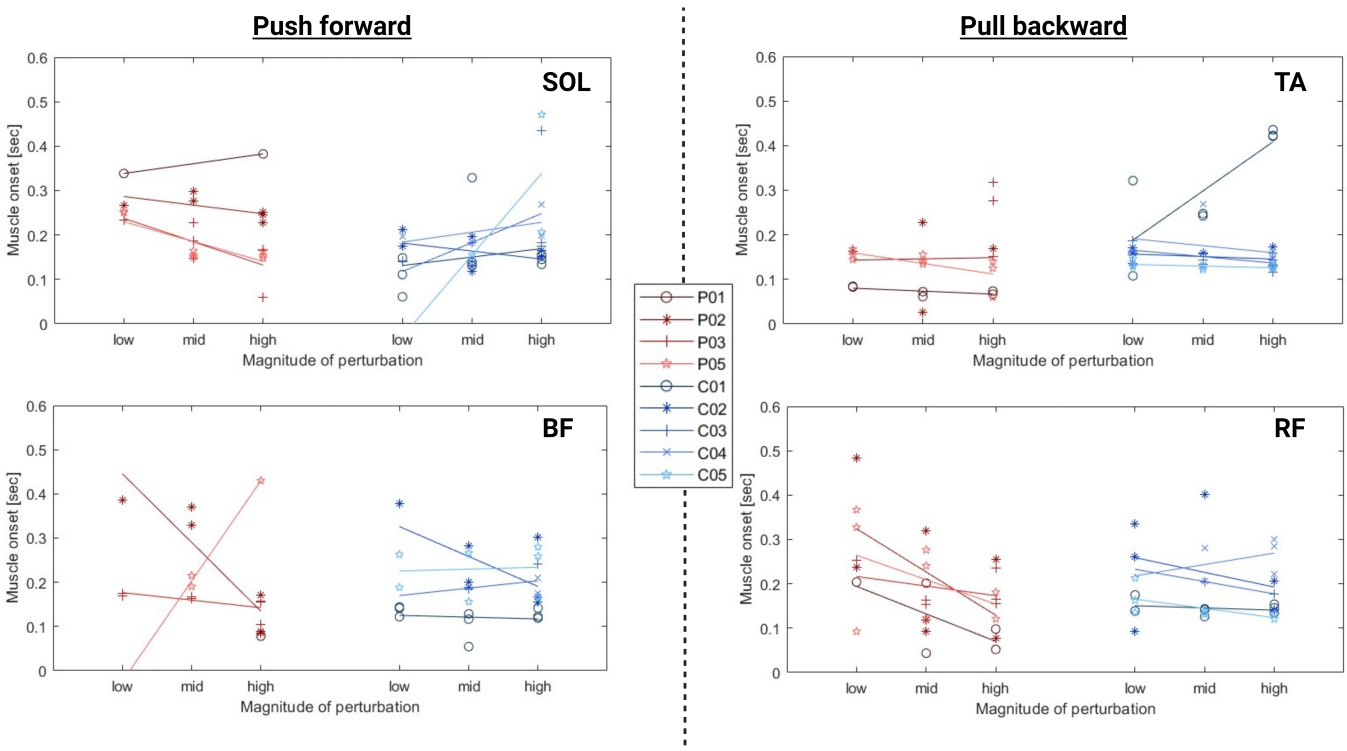

Figure 7: The muscle onsets and the linear regression lines over the different magnitudes of the perturbations at left toe-off for the SOL, BF, TA and RF at left toe-off for both patients (red) and able-bodied (blue) participants.

3.3 Muscle Effort

The effort was calculated as the difference between the integrated perturbed EMG signal from perturbation onset to 500 ms after and the integrated mean baseline walking EMG signal from toe-off to 500 ms after. In this section, the muscle effort results are shown for the SOL and BF after a push in the forward direction and for the TA and RF after a push in the backward direction. An overview of other muscle effort results is depicted in Appendix A.7.

3.3.1 Muscle Effort after push forward

The muscle effort of the SOL, showed a positive modulation of increasing effort over increasing magnitude of the perturbation in the forward direction for three of the iSCI participants and two of the able-bodied participants, as is visible in Figure 8. The BF showed a positive modulation of effort over increasing perturbation magnitude for two of the iSCI participants and all of the able-bodied participants.

As is visible in Figure 8 and Table 2, the BF effort modulation is weaker for all of the iSCI participants compared to the BF effort modulation of four out of five of the able-bodied participants. For the SOL effort, similar range modulation over increasing perturbation was shown (Table 2).

3.3.2 Muscle Effort after pull backward

After a pull in the backward direction, the TA of three iSCI participants and all the able-bodied participants showed positive effort modulation over increasing perturbation magnitude, see Figure 8. For the RF effort, positive modulation over increasing perturbation magnitude was visible for all the participants of the iSCI and able-bodied group.

The modulation of the TA effort of two iSCI participants was shown weaker than for all the able-bodied participants, see Figure 8 and Table 2. Three participants of the iSCI participant group showed weaker modulation of the RF effort than four participants of the able-bodied group.

Table 2: Range of regression slopes [min max] intra-participant groups from the muscle effort to counteract the AP perturbations at LTO.

Muscle	Patient	Able-bodied
SOL	[-0.0001 0.0247]	[-0.0112 0.0275]
BF	[-0.0003 0.0178]	[0.0171 0.0520]
TA	[-0.0001 0.0167]	[0.0014 0.0195]
RF	[0.0036 0.0287]	[0.0043 0.0272]

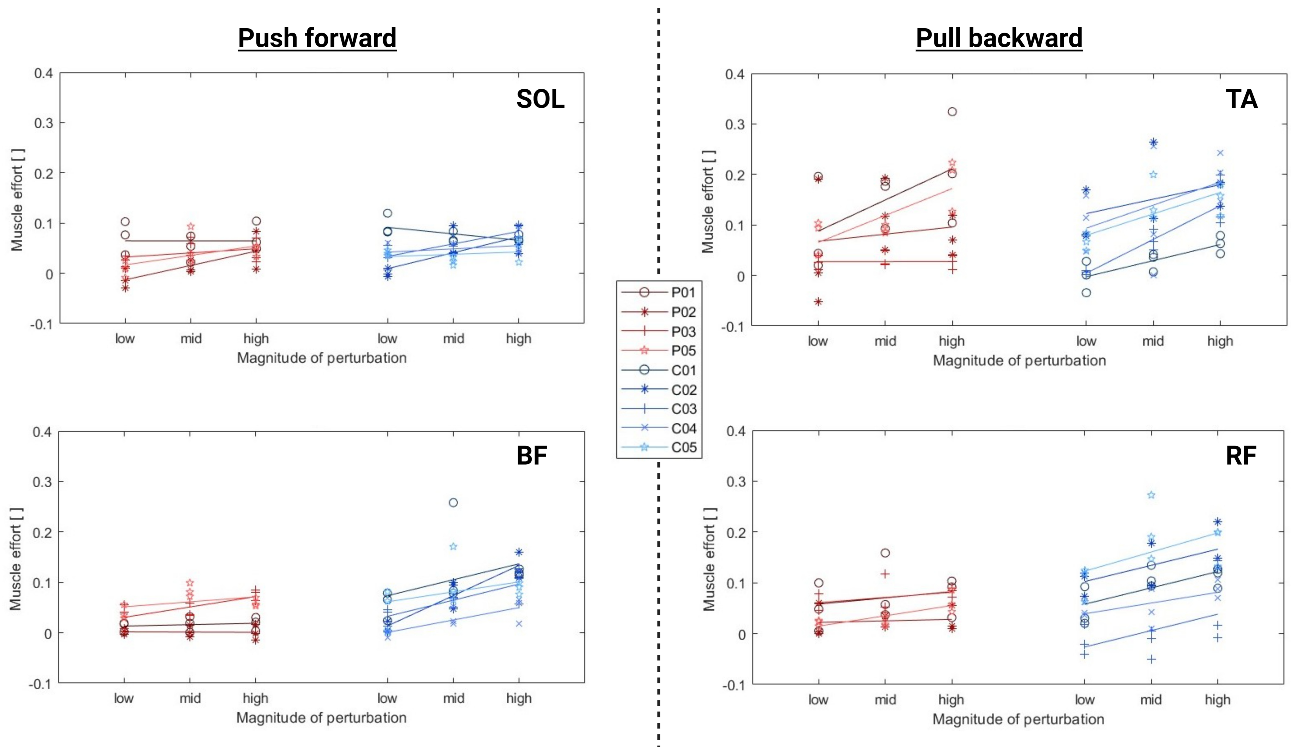

Figure 8: The muscle effort and the linear regression lines over the different magnitudes of perturbations at left toe-off for the SOL, BF, TA and RF for both patients (red) and able-bodied (blue) participants.

3.4 COM Velocity

The COM velocity was calculated using the position of the hip brace markers at 150 ms after perturbation initiation. The four individuals with iSCI showed higher positive COM velocities in the AP direction after a push forward than the COM velocities of three of the able-bodied individuals (Figure 9). In addition, the same three able-bodied participants showed negative COM velocities for the low and the mid perturbation magnitude, while the iSCI patients did not. The linear regression line showed approximately the same modulation patterns for both groups (Table 3). After a pull backwards, the participants showed negative COM velocities (Figure 9). For this condition, the iSCI group and the able-bodied group showed similar COM velocity values and similar negative modulation over the magnitudes (Table 3). An overview of the mean COM velocity for each of the conditions per group is depicted in Appendix A.8 Table A9.

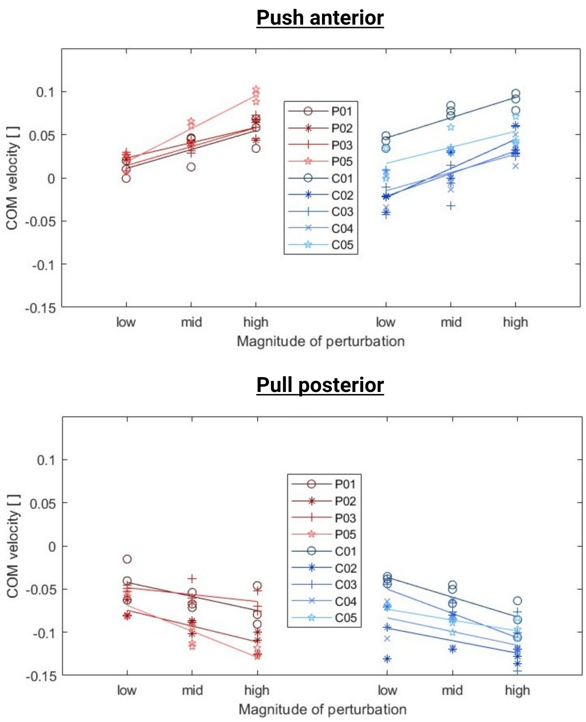

Figure 9: COM velocity of the patients (red) and the able-bodied (blue) participants after a perturbation in AP direction at left toe-off, calculated at the end of the push duration. The walking speed in the AP direction was 0.63 m/s normalized to leg length.

Table 3: Range of regression slopes [min max] intra participant groups from the muscle effort to counteract perturbation at LTO.

Condition	Patient	Able-bodied
Push anterior	[0.0153 0.0334]	[0.0162 0.0294]
Pull posterior	[-0.0260 -0.0069]	[-0.0243 -0.0109]

3.5 Effort - COM Velocity Relationship

Lastly, the muscle effort and COM velocity relation was analysed. This was done by calculating the linear regression for each of the participants between the calculated muscle effort and the COM velocity at 150 ms after perturbation onset. In this section the effort - COM velocity relations are discussed for the SOL and BF after perturbations in the forward direction and for the TA and RF after perturbations in the backward direction. An overview of other muscle effort - COM velocity relations than discussed in this section, is depicted in Appendix A.9.

3.5.1 Effort - COM Velocity Relationship after push forward

In Figure 10 is visible that four of the five able-bodied participants and two of the four iSCI participants showed positive modulation of increasing SOL effort over increasing COM velocity after perturbation. The BF effort showed positive modulation for all the able-bodied participants and three of the iSCI participants.

Two of the three iSCI participants that showed positive modulation of SOL effort over increasing COM velocity, showed weaker modulation than two participants of the able-bodied group. Three of the four iSCI participants showed weaker BF effort modulation over increasing COM velocity compared to all the participants of the able-bodied group, see Table 4.

3.5.2 Effort - COM Velocity Relationship after pull backward

After a perturbation in the backward direction, three iSCI participants and all able-bodied participants showed a positive modulation of increasing TA effort over increasing COM velocity in the backward direction, see Figure 10. The RF effort showed positive modulation of increasing COM velocity for all the able-bodied participants and three of the iSCI participants.

Apart from iSCI participant P01, the iSCI participants showed weaker TA effort modulation compared to the three able-bodied participants. Two of the iSCI participant group showed a weaker modulation of RF effort compared to four of the able-bodied group.

Table 4: Range of regression slopes [min max] intra participant groups for the effort - COM relation [min max] after a perturbation in the AP direction.

Condition	Patient	Able-bodied
Push anterior SOL	[-0.0013 0.0210]	[-0.0129 0.0321]
Push anterior BF	[-0.0001 0.0175]	[0.0136 0.0425]
Pull posterior TA	[-0.0758 0.0013]	[-0.0595 -0.0198]
Pull posterior RF	[-0.0258 -0.0074]	[-0.0359 -0.0069]

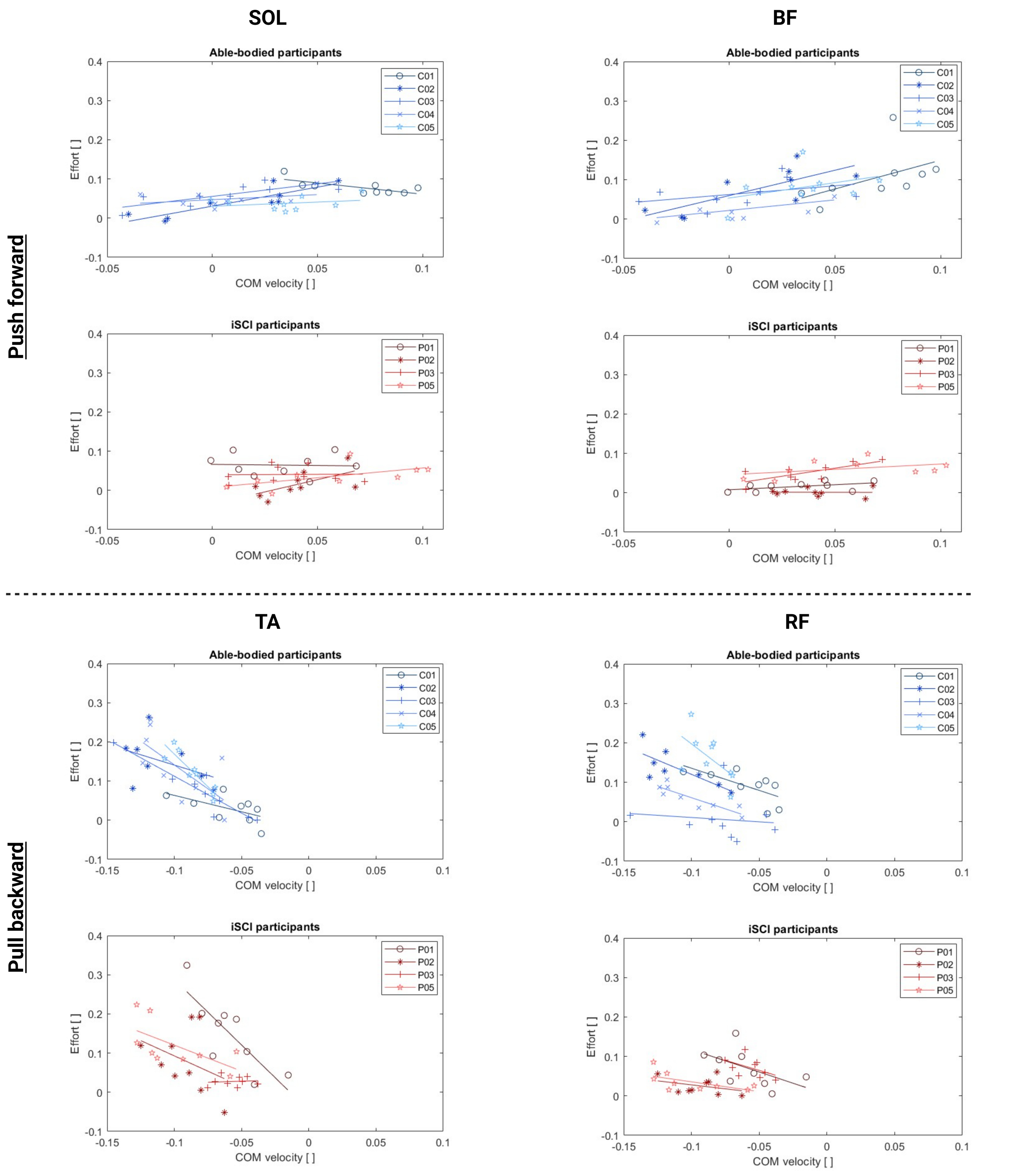

Figure 10: Effort - COM velocity relation for the able-bodied group (blue) and the iSCI patient group (red) after AP perturbations.

4 Discussion

In this chapter, the obtained results are discussed using previous research and the expectations for each of the assessed parameters. In addition, the limitations of this study and future work to obtain knowledge about the reactive balance strategy of iSCI participants are discussed.

4.1 Activity Balance Confidence scale

The participants of the able-bodied group showed confidence in their balance during activity as they all scored between 97% and 100% as a mean percentage. However, the iSCI participants showed more heterogeneity and less confidence with a mean percentage score range between 63% and 94%, see Appendix Table A6.

4.2 Muscle Onset

It was expected to see a delayed onset for the leg muscles of the iSCI participants compared to the muscle onsets of the able-bodied participants, as was seen for the GM and the TA in Thigpen et al.[21]. Aurora et al.[11], also reported a delayed onset for the TA of iSCI participants group but this was not found significant. Our results showed a delayed SOL onset for two of the iSCI participants (P01 and P02) compared to the able-bodied participant after a push in the forward direction. These two participants showed lower confidence in their balance during the ABC test compared to 7 of the other participants. In addition to the delayed SOL onset, P01 showed a delayed onset for the GM.
The onset results differ slightly from the study of Thigpen et al.[21] since no delayed TA onset was found. However, Thigpen et al. studied perturbations while standing instead of walking. Therefore, the findings of these two studies might not be comparable. Another expectation regarding the muscle onset was to see an earlier muscle onset detection for higher perturbation magnitude, as was seen for able-bodied subjects in the study of Vlutters et al.[19]. In this research, this was shown in the muscles of the participants after a pull in the backward direction, but not for all the participants after a push in the forward direction. It is unknown what caused this result to be different from Vlutters et al.[19].

4.3 Muscle Effort

It was expected to see an increase in effort over increasing perturbation magnitude, as was seen for muscle activity in other studies[9][19][21]. In addition, weaker modulation was expected to be visible for the iSCI patient group. For the upper leg muscles RF and BF, the results were in agreement with the expectation. After a push in the forward direction, a weaker BF effort modulation was seen over increasing magnitude in most of the iSCI participants compared to the able-bodied participants. In addition, after a pull backwards weaker effort modulation was seen for the RF in most of the iSCI participants. The iSCI participants showing the weakest BF and RF effort modulation were the same iSCI participants who showed delayed SOL onset after perturbations in the forward direction. Moreover, the iSCI participant that showed the strongest RF and BF modulation, scored the highest percentage on the ABC scale.

4.4 COM Velocity

As iSCI participants were expected to be less able to restore balance using the reactive strategies after perturbation, it was expected to see more effect on the COM velocity of the iSCI group than on the able-bodied group. The iSCI participants showed higher COM velocity at the end of the push forward, compared to three of the able-bodied group. This agrees with the expectation. However, for the COM velocity after a pull backwards, this was not seen. Further, for both groups, there was an increased COM velocity of increasing perturbation magnitude. A possible explanation for this is that the iSCI might already be using proactive balance strategies because the participants know the pushes or pulls will come. This might minimize the effect of an unexpected perturbation on the COM velocity.

4.5 Effort - COM Velocity Relation

In agreement with the expectation for the muscle effort over perturbation magnitude, it was expected to see an increase in muscle effort for increasing COM velocity as well. An increased COM velocity means a balance loss to counteract using muscle effort[20]. In addition, the expectation was to see less muscle effort modulation for the iSCI group, as the effort was also expected to be less modulated for the iSCI group with increasing perturbation magnitude. Apart from a few cases, the results agreed with the expectation that the effort was positively modulated over increasing COM velocity. In addition, the results showed a weaker positive RF and BF effort modulation for the iSCI group compared to the able-bodied group.

4.6 Limitations

This study included a small iSCI population. Therefore, no statistical tests were applied to the results. The results may indicate where to focus on in the reactive balance responses and might work as a study concept but do not give a reliable conclusion.

In addition, a drawback of the study is that the perturbations are not completely unexpected. Even though the perturbations happen at random time instances at toe-off and in random order, the participants know that the perturbations will be coming. As perturbations are expected, people tend to use proactive balance strategies to reduce the effect when the perturbation occurs, such as stiffer walking or smaller steps [29]. In the 'real world' walking scenarios, perturbations are mostly unexpected. Unfortunately, this can not be prevented in this research method.

In some cases, the muscle onset was immediately detected at the first time instance (Figure A.14). A lot of immediate onset detection was found when comparing the perturbed strides to the baseline walking. Because this was seen, the method was adjusted to compare the perturbation strides to the normal strides in the perturbation trials to overcome the problem of comparing a more relaxed baseline walking strategy to perturbation strides with having a proactive strategy, which is two different walking scenarios. Proactive balance strategies were characterized by previous research[9][10]. It is known that iSCI patient adjusts their walking speed, step length and stiffness to prevent balance loss during walking[9][10]. After adjusting the method, there were still some immediate muscle onsets detected but fewer than using the baseline walking, see Table A7.

This study focused only on the results in the anteroposterior direction because the results in the mediolateral direction did not show outstanding differences between the iSCI participants and the able-bodied participants. This could be because in the mediolateral direction, mostly for the high perturbation magnitude, stepping over was often seen as a strategy but not necessarily for the other perturbation magnitudes. This makes finding a relation between the parameters and the perturbation magnitude more difficult.

4.7 Future Work

As the goal of this research was to obtain knowledge about the reactive balance strategies of iSCI participants and where this strategy lacks to regain balance after unexpected perturbations, some future work can be done to improve this research method and obtain more knowledge that might be useful to improve the quality of life of the iSCI patient group.

First, continuing this research with a larger population would be beneficial. There were interesting results found for the muscle effort to counteract the perturbation in anteroposterior direction. Since the participant group was small, the findings are not necessarily applicable to the whole iSCI patient group. It could be interesting to see if similar results are found for larger populations. In addition, including a larger population enables statistical analysis. Using linear mixed models, it can be found if the weaker muscle modulations in the iSCI participants are significant.

In this study, effort values can not be compared intra and inter-participant groups because of the EMG normalization method. For future research, it is recommended to have a good maximum voluntary contraction (MVC) method, to enable effort magnitude comparisons between participants and obtain more knowledge about the reactive balance strategy in iSCI patients. When possible, use a strapping configuration to ensure isometric contraction. Use MVC positions as proposed by Peter Konrad [30].

Different balance strategies were not taken into account in this study. To obtain a better understanding of the different reactive balance strategies in the iSCI patient group, it would be good to make a distinction between the strides in which the balance strategy was stepping over and in which strides the torque adjustment strategy was used.

Since the COM velocity results did not indicate that the iSCI participants were more affected by the perturbations, additional analysis on the difficulty of the iSCI participants to withstand the perturbations could be beneficial to understand if the research outcomes were causing the participants to be more affected or not. This could potentially be done by estimating the COM displacement or by estimating the COM velocity at the heel strike after perturbation.

5 Conclusion

The goal of this study is to obtain knowledge about the reactive balance strategies of iSCI participants and where this strategy lacks to regain balance after unexpected perturbations compared to able-bodied individuals. In this study, a delayed onset was found in two of the iSCI participants in the m. Soleus and in one of the iSCI participants in the m. Gastrocnemius Medialis after a push in the forward direction. The muscle effort modulation of the RF and BF in the stance leg to counteract the perturbations of different magnitudes was found weaker for the iSCI participants compared to the age- and gender-matched able-bodied participants' effort modulation. In addition, a weaker RF and BF effort modulation in the stance leg over increasing COM velocity was found for the iSCI participants compared to the able-bodied participants. As the participant population is very small in this study, the results are not reliable for the whole iSCI population, but this study could be used as a study method example for future research. In addition, the results could be indicative of results using a larger iSCI population when studying the reactive balance strategies in iSCI participants during walking.

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A Appendix

A.1 Demographics of the Participants

Table A5: The demographic parameters of the participants.

Participant	Participant tag	Gender	Age [years]	Length [m]	Mass [kg]	Leg length [m]
Patient 1	P01	F	45	1.5	76.3	0.8
Patient 2	P02	M	70	1.79	80	0.98
Patient 3	P03	F	70	1.78	70.7	0.97
Patient 4	P04	M	74	1.87	86.6	1.04
Patient 5	P05	M	42	1.78	80.5	0.94
Control 1	C01	F	51	1.77	83	0.96
Control 2	C02	M	68	1.83	87.9	0.99
Control 3	C03	F	74	1.66	76	0.92
Control 4	C04	M	73	1.85	100	1.02
Control 5	C05	M	53	1.84	86.2	1.02

A.2 ABC survey

The Activities-specific Balance Confidence (ABC) Scale©

Instructions: For <u>each</u> of the following activities, please indicate your level of balance confidence by choosing one of the points on the scale below from 0% to 100%.

If you do **not currently do the activity**, try and imagine how confident you would be if you had to do the activity. If you **normally use a walking aid to do the activity or hold onto someone**, rate your confidence as if you were using these supports. If you have any questions, please ask the administrator.

0%	10	20	30	40	50	60	70	80	90	100%
No Confidence										Completely Confident

"How confident are you that you can maintain your balance and remain steady when you....

1. walk around the house? _____%
2. walk up or down stairs?_____%
3. bend over and pick up a slipper from the front of a closet floor? _____%
4. reach for a small can off a shelf at eye level? _____%
5. stand on your tip toes and reach for something above your head? _____%
6. stand on a chair and reach for something?_____%
7. sweep the floor?_____%
8. walk outside the house to a car parked in the driveway?_____%
9. get into or out of a car?_____%
10. walk across a parking lot to the mall?_____%
11. walk up or down a ramp?_____%
12. walk in a crowded mall where people rapidly walk past you?_____%
13. are bumped into by people as you walk through the mall?_____%
14. step onto or off of an escalator while holding onto a railing?_____%

15. step onto or off an escalator while holding onto parcels such that you cannot hold onto the railing?_____%

16. walk outside on icy sidewalks?_____%

The Activities-specific Balance Confidence (ABC) Scale©

Acknowledgment: Dr. Myers must be acknowledged as the primary developer and copyright holder of the ABC Scale (using the statement above) in all publications, clinical manuals, or other materials reporting on the use and results of this scale. If you requested and received her permission to use or translate this scale, you should report this in your publications.

Permission and Cost: The print version of the scale may be reproduced for student training, research and clinical practices in which therapists and assistants use the scale to assess fewer than 1000 patients per year. **In all other cases**, including: translation into other languages than English, other modifications to the scale itself and/or instructions, use in clinical trials, for commercial or marketing purposes, or in larger scale practices (1,000+ patients per year) and/or electronic record keeping, **permission must be obtained** by the researcher or institution by contacting amyers@uwaterloo.ca. Costs may apply.

Administration: The ABC Scale can be **self-completed** in about five minutes using the paper version, electronically (e.g, touch screen) or by interview. The full instructions must be given on the scale itself (as shown on page 1) or via a cover sheet or letter. A contact should be provided should respondents have questions. Individuals must be capable of understanding the instructions and should not be influenced by others (family, friends or clinicians).

It is important not to use the terms “falling” or “fear of falling” when administering the ABC. In the late 1990’s (Myers, 1999), we modified the rating directive from confidence in doing each activity “without losing your balance or becoming unsteady” ***to** confidence in **“maintaining your balance and remaining steady**”*. The latter is more positive, affirmative and action oriented (i.e., people may recover their balance from a trip, slip or change in position).

They should picture themselves doing each of the activities at home and in their community **(not in a clinical setting)** and in a bipedal, upright position (as opposed to sitting in a chair to bend over, e.g., item 3). If their home does not have stairs, they should consider places they visit (e.g., family, friends) with stairs. It is assumed most stairs have one or two hand railings.

Scoring: Total the ratings (possible range = 0 to 1600) and divide by 16 (or the number of items completed; minimum of 12) to get each person’s ABC score. If a person qualifies his/her response to items #2, #9, #11, #14, or #15 (e.g., "up" versus "down"), use the **lowest**

confidence rating of the two (as this will limit the entire activity). **Total scores can be computed if a person answers <u>at least 12 of the 16 items</u>** (Myers et al., 1998).

To examine **change**, the scale must be administered at least twice (e.g, pre/post therapy) and scores compared. Do <u>not</u> simply ask clients if their confidence has increased or decreased.

<u>Please cite the following 3 references on the development of the English ABC Scale:</u>

1. Powell LE & Myers AM. The Activities-specific Balance Confidence (ABC) Scale. *J Gerontol Med Sci* 1995; 50 (1):M28-34.
2. Myers AM, Powell LE, Maki BE et al. Psychological indicators of balance confidence: Relationship to actual and perceived abilities. *J Gerontol Med Sci* 1996; 51A: M37-43.
3. Myers AM, Fletcher PC, Myers AH & Sherk W. Discriminative and evaluative properties of the Activities-specific Balance Confidence (ABC) Scale. *J Gerontol Med Sci* 1998; 53A: M287-M294. **This article includes <u>benchmarks</u> for interpreting ABC scores in various populations (e.g., active older adults, home care clients, patients undergoing hip and knee replacement).*

Note: If you cannot obtain these articles, email amyers@uwaterloo.ca.

Book: Myers, AM. <u>Program evaluation for exercise leaders</u>. Human Kinetics, 1999. Contains outcome measures, including the ABC Scale with the modified rating directive.

Another key article is the one by Moore et al. (2018) which recommends the ABC Scale as a **core outcome measures** for adults with neurological conditions undergoing rehabilitation. *This article reviews the empirical evidence and suggested cut-off scores for the ABC Scale.*

Moore JL, Potter K, Blankshain K, Kaplan SL, O'Dwyer LC, Sullivan JE. A core set of outcome measures for adults with neurological conditions undergoing rehabilitation: A Clinical Practice Guideline. *J Neurologic Phys Ther.* 2018; 43(3): 174-220.

<u>Target Audience:</u>

The ABC Scale is intended for <u>ambulatory</u> (with or without use of walking aids and/or occasional personal assistance) <u>community dwelling</u> older adults (OAs), as well as persons with balance related disorders. It is <u>not</u> intended for residential living seniors (e.g., nursing homes). Another tool, the AFC scale, developed for this audience is referenced below.

Pearce NJ, Myers AM, Blanchard RA. Assessing subjective fall concerns in residential living seniors: Development of the Activities-specific Fall Caution Scale. *Arch Phys Med Rehabil.* 2007: 88: 724-731.

Blanchard RA, Myers AM, Pearce NJ. Reliability, construct validity, and clinical feasibility of the Activities-specific Fall Caution Scale for residential living seniors. *Arch Phys Med Rehabil.* 2007: 88: 732-739.

<u>Additional notes:</u>

In addition to client background (age, sex, education) and relevant clinical information, you should document use of walking aids, fall history (especially recurrent falls) and driving status.

The ABC Scale has been used to examine the effectiveness of various types of interventions (e.g., fall prevention programs, balance training, rehabilitation) with different populations. It has also been translated into dozens of languages. Search the published literature and use the information above to determine if the authors administered and scored the scale as intended.

Table A6: Mean percentages obtained in the ABC questionnaire.

Participant	Mean percentage ABC
P01	65%
P02	73%
P03	63%
P04	94%
P05	83%
C01	98%
C02	99%
C03	99%
C04	97%
C05	100%

A.3 Experimental Protocol Flow

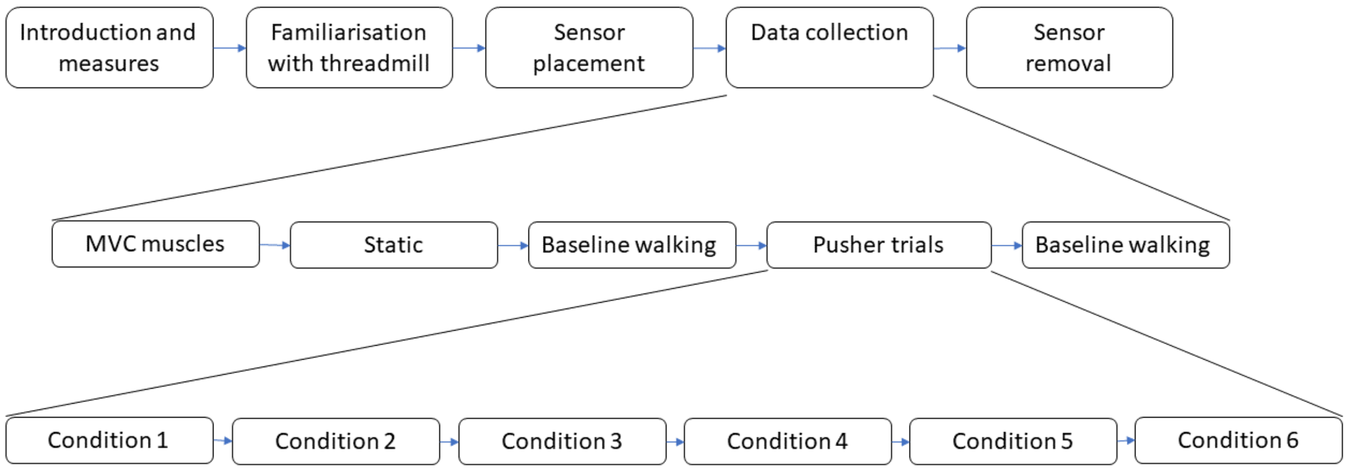

Figure A.11: Diagram of the experimental work flow.

A.4 Perturbation Conditions

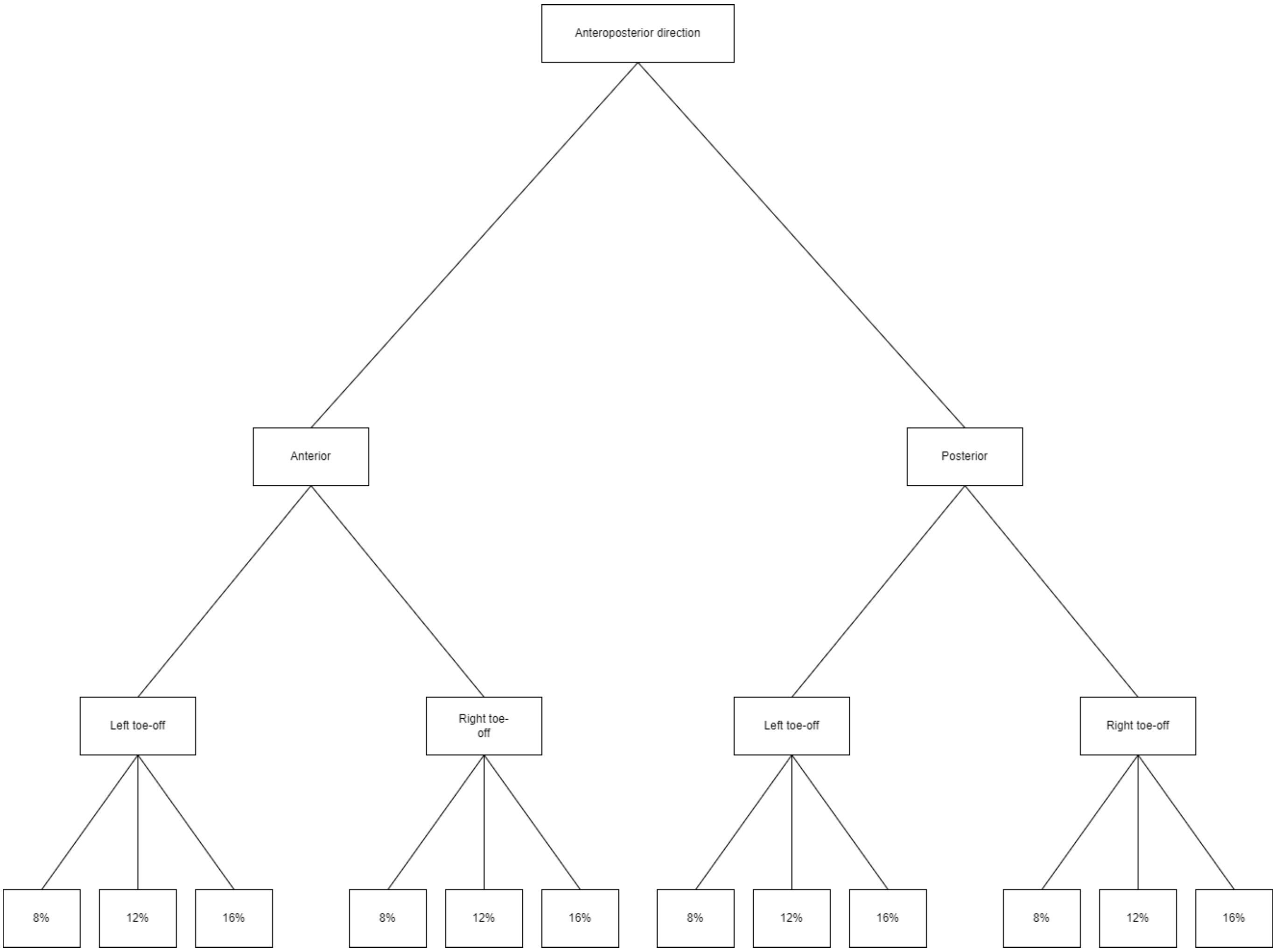

Figure A.12: Perturbation conditions with motor attachment in anteroposterior direction. The diagram shows all the directions, perturbation instances and magnitudes during 1 trial in AP direction.

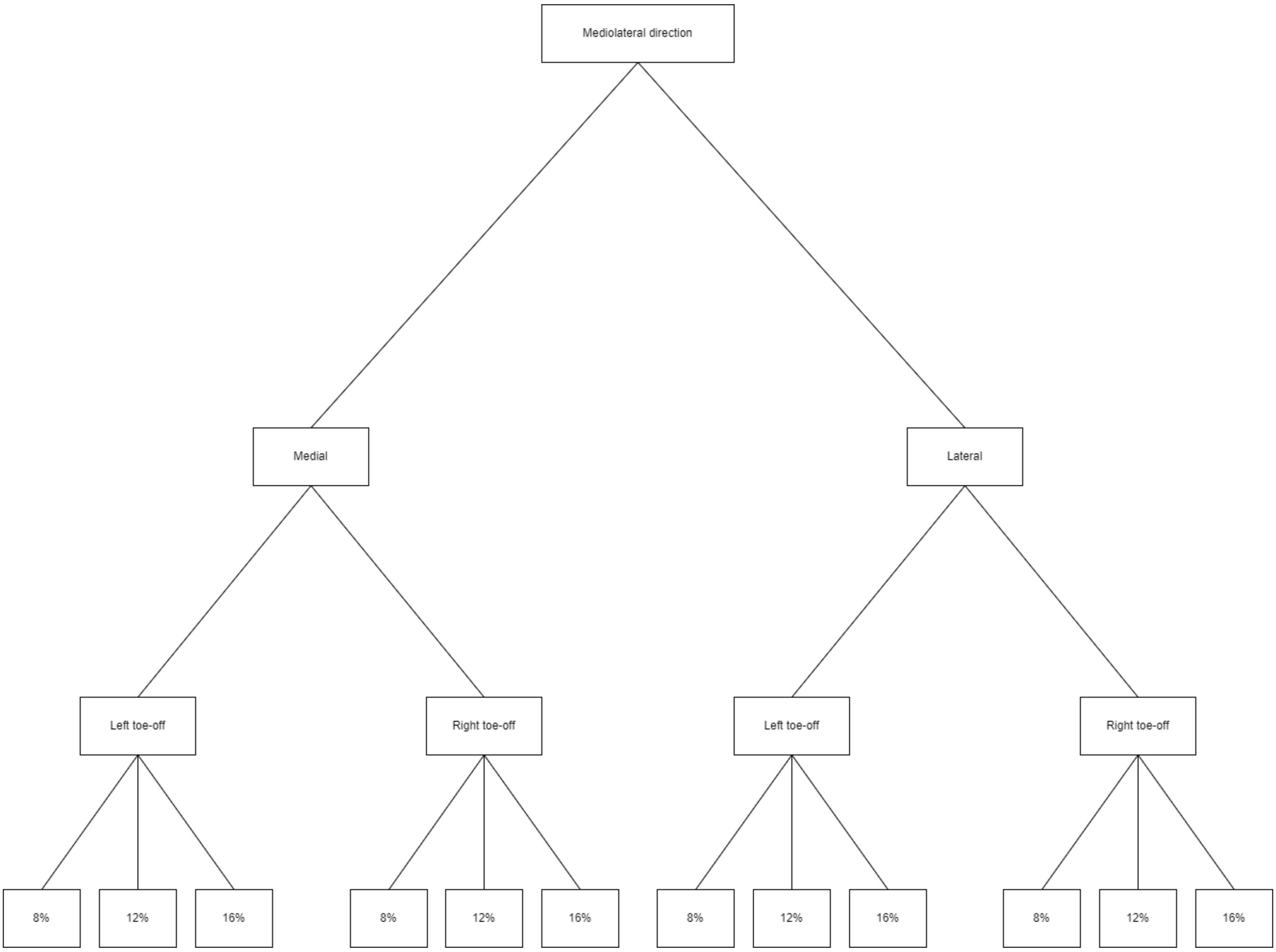

Figure A.13: Perturbation conditions with motor attachment in mediolateral direction. The diagram shows all the directions, perturbation instances and magnitudes during 1 trial in ML direction.

A.5 Muscle onset detection method

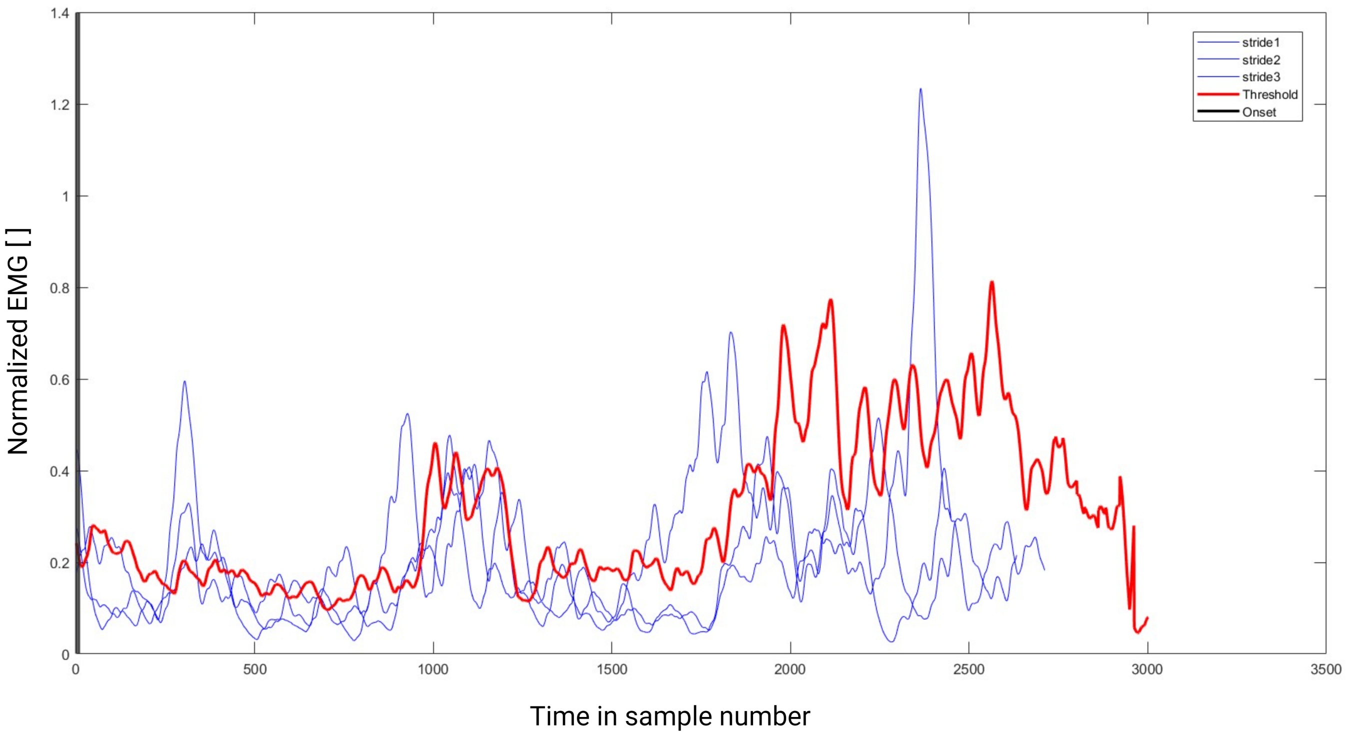

Figure A.14: Visible are three perturbed strides in the AL of P05. All perturbed strides have an immediate muscle onset detection (black vertical line) at the first time instance.

Table A7: Given are onset values (between brackets) and the amount of onset values that were detected between 0-25ms. These values were deleted from the results.

	Using baseline walking	Using normal strides perturbed walking
C01	113 (out of 501)	2 (out of 419)
C02	33 (out of 323)	25 (out of 262)
C03	56 (out of 476)	2 (out of 304)
C04	71 (out of 485)	10 (out of 333)
C05	43 (out of 488)	8 (out of (426)
P01	143 (out of 493)	50 (out of 301)
P02	82 (out of 502)	3 (out of 346)
P03	73 (out of 495)	6 (out of 333)
P05	106 (out of 502)	12 (out of 393)

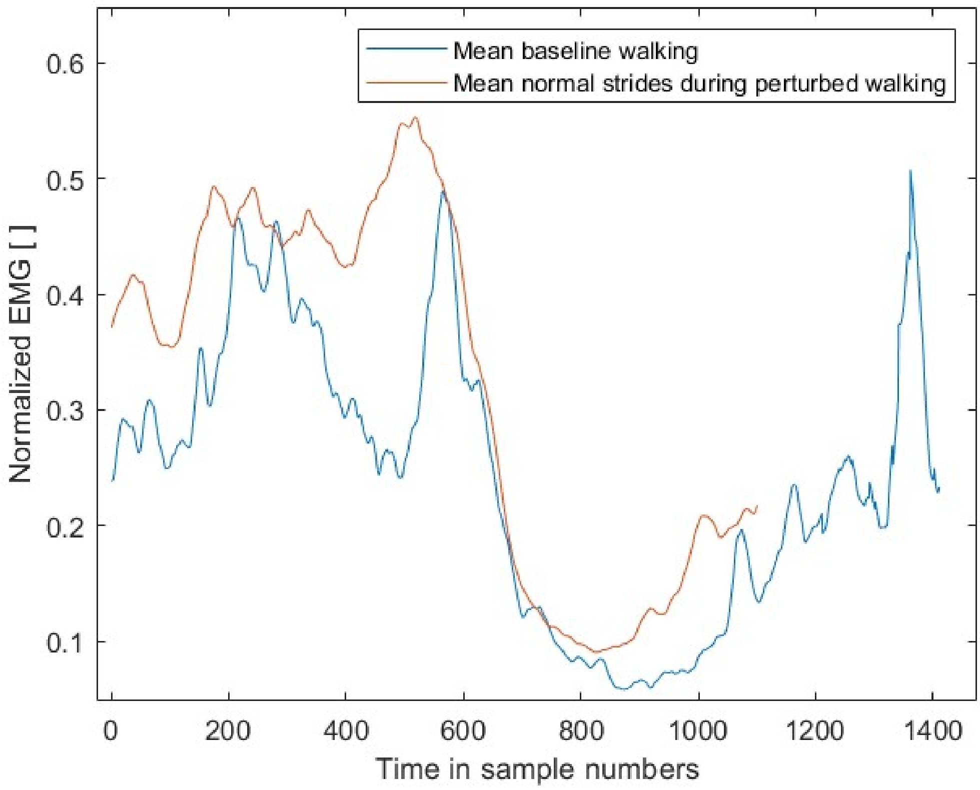

Figure A.15: The mean baseline gait cycle of the P01 SOL with the mean of the normal strides during the perturbed walking trials.

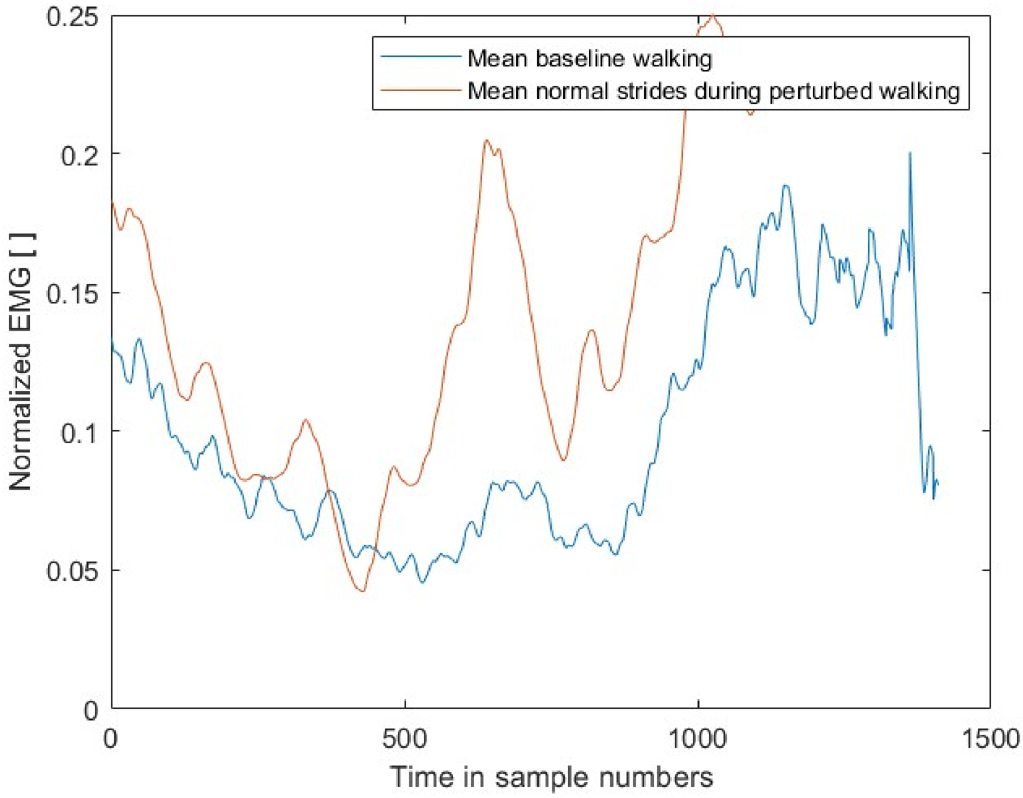

Figure A.16: The mean baseline gait cycle of the P01 RF with the mean of the normal strides during the perturbed walking trials.

Table A8: The number of muscles having higher initial values for the EMG signal of the mean normal strides during perturbed walking compared to the mean of the baseline strides.

Participant	Number of muscles with higher EMG
P01	6 from 7
P02	5 from 7
P03	5 from 7
P05	6 from 7
C01	5 from 7
C02	3 from 6
C03	3 from 7
C04	5 from 7
C05	3 from 7

A.6 Muscle onset

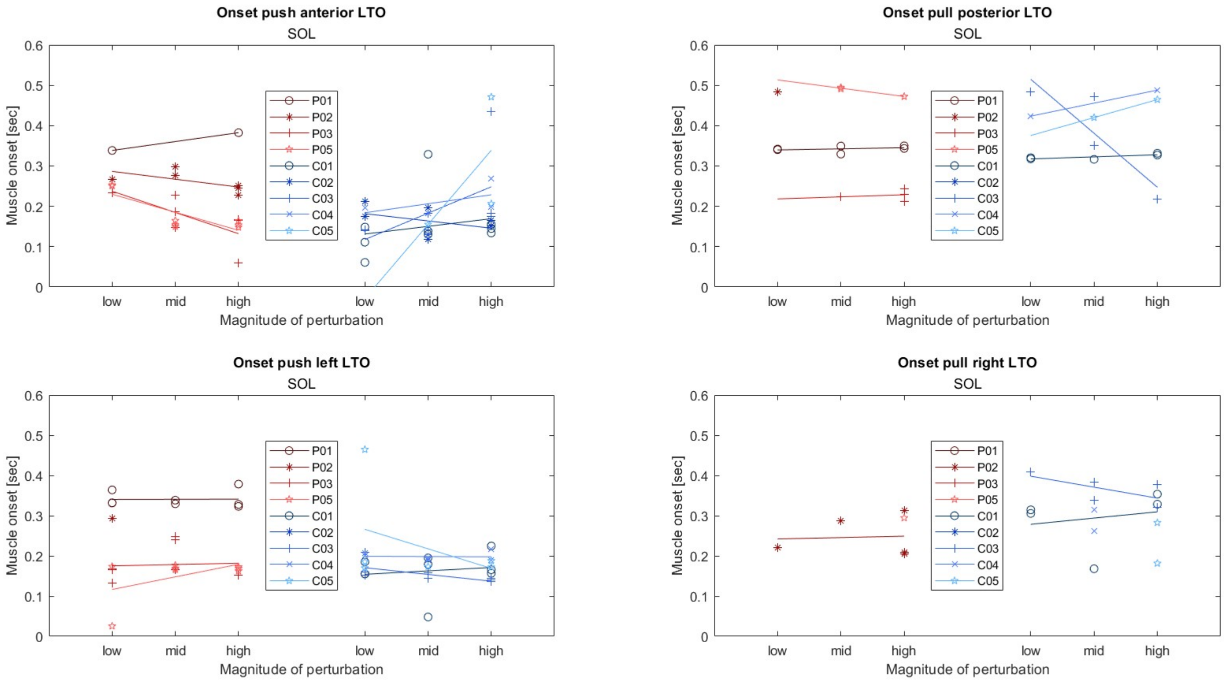

Figure A.17: M. Soleus onset after perturbation at left toe-off.

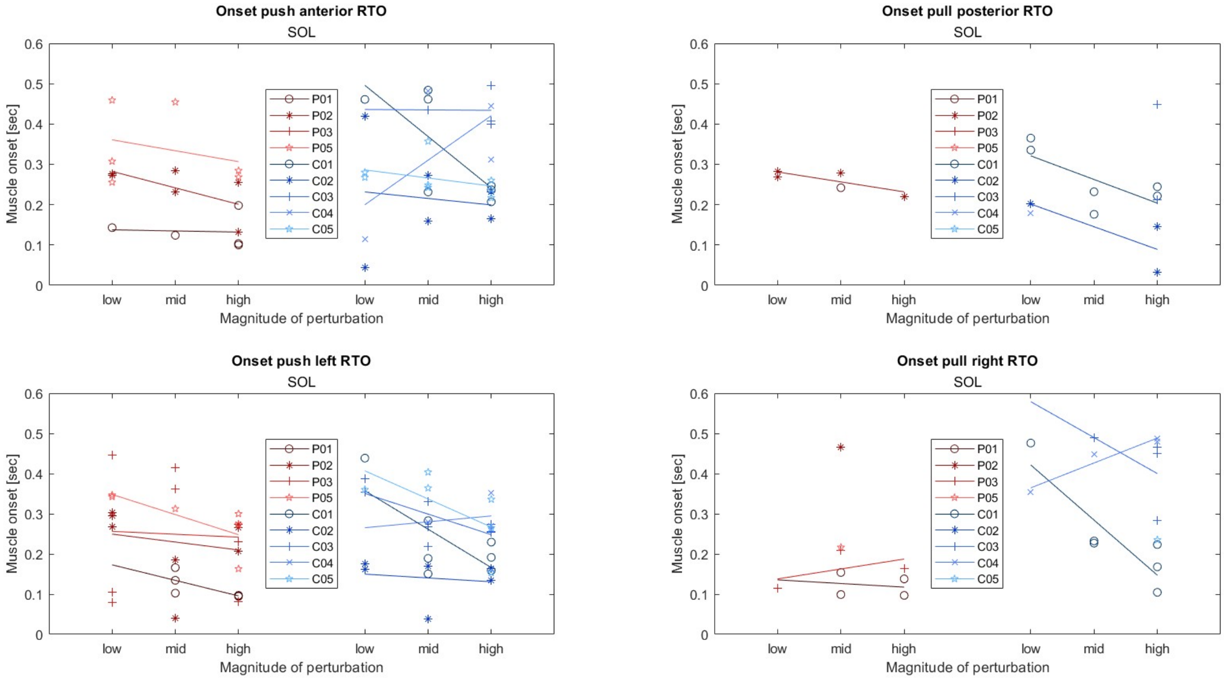

Figure A.18: M. Soleus onset after perturbation at right toe-off.

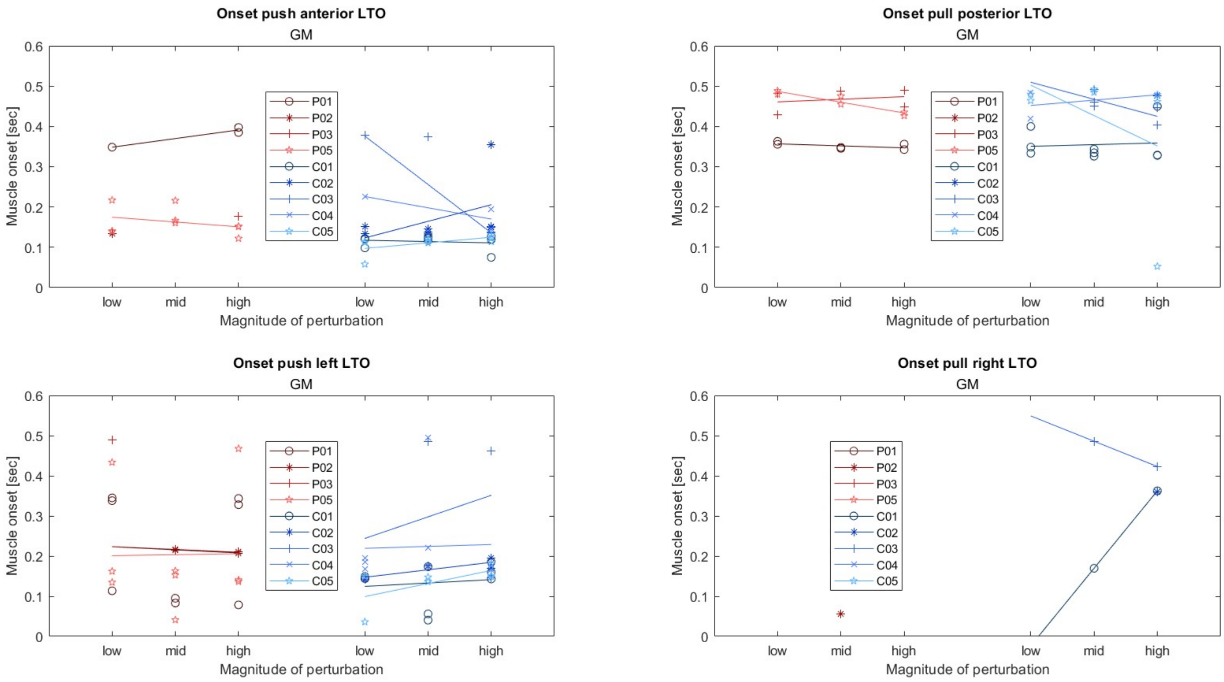

Figure A.19: M. Gastrocnemius Medialis onset after perturbation at left toe-off.

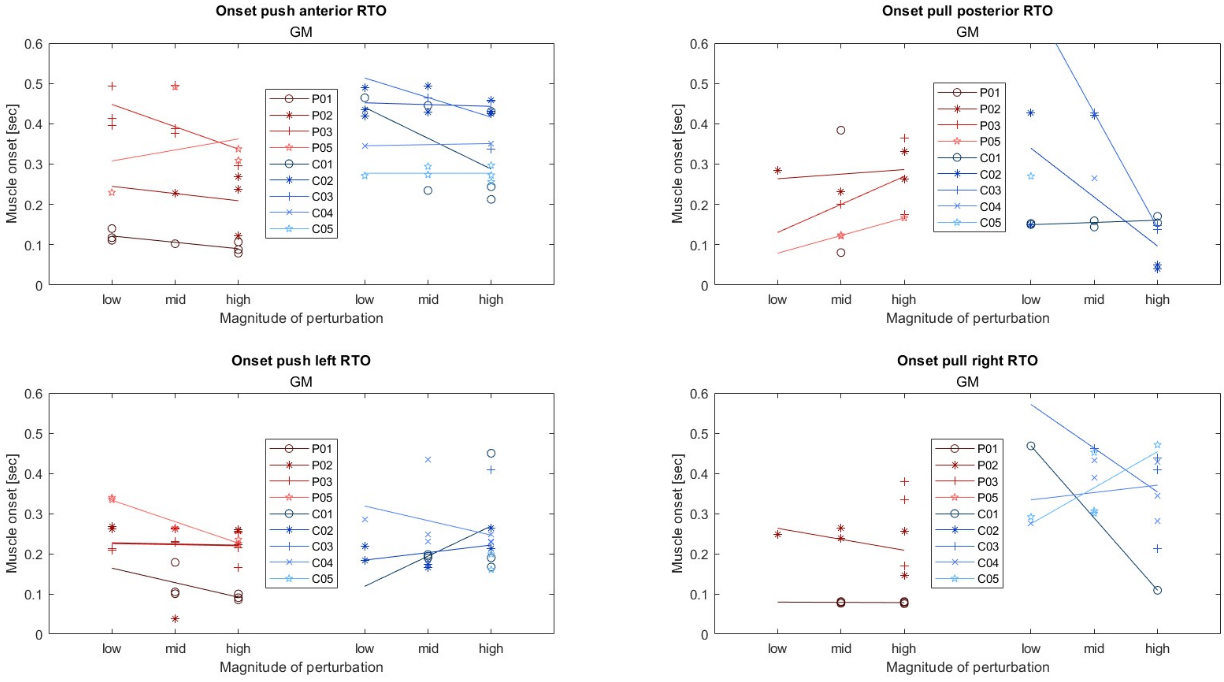

Figure A.20: M. Gastrocnemius Medialis onset after perturbation at right toe-off.

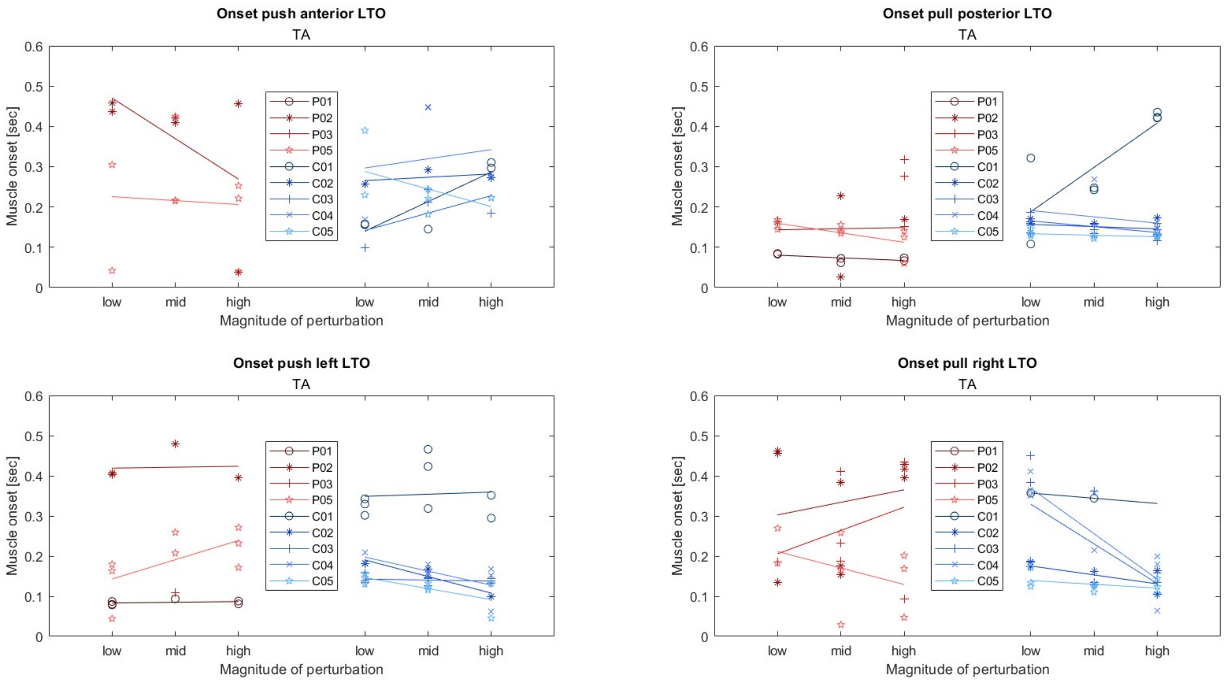

Figure A.21: M. Tibialis Anterior onset after perturbation at left toe-off.

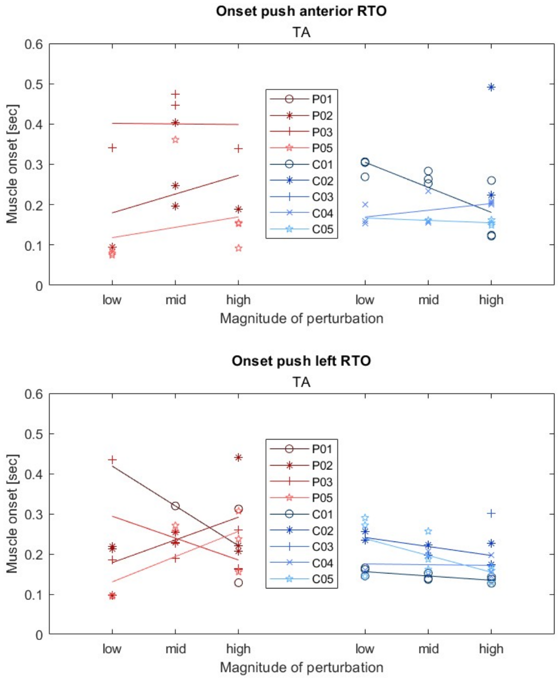

Figure A.22: M. Tibialis Anterior onset after perturbation at right toe-off.

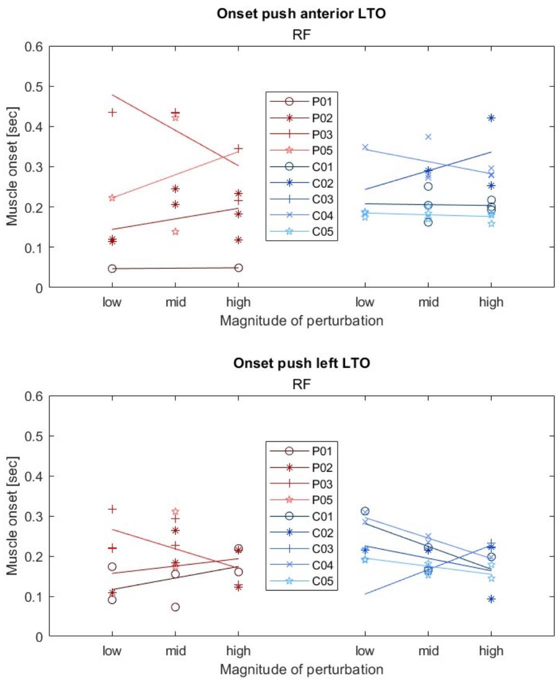

Figure A.23: M. Rectus Femoris onset after perturbation at left toe-off.

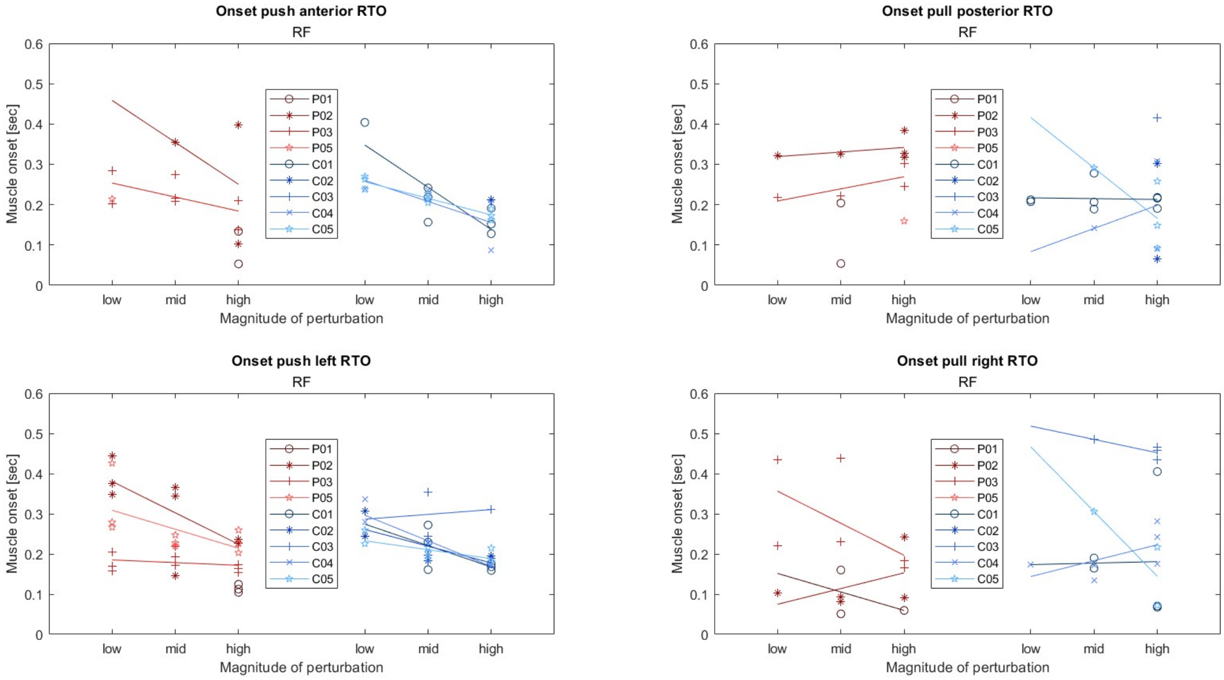

Figure A.24: M. Rectus Femoris onset after perturbation at right toe-off.

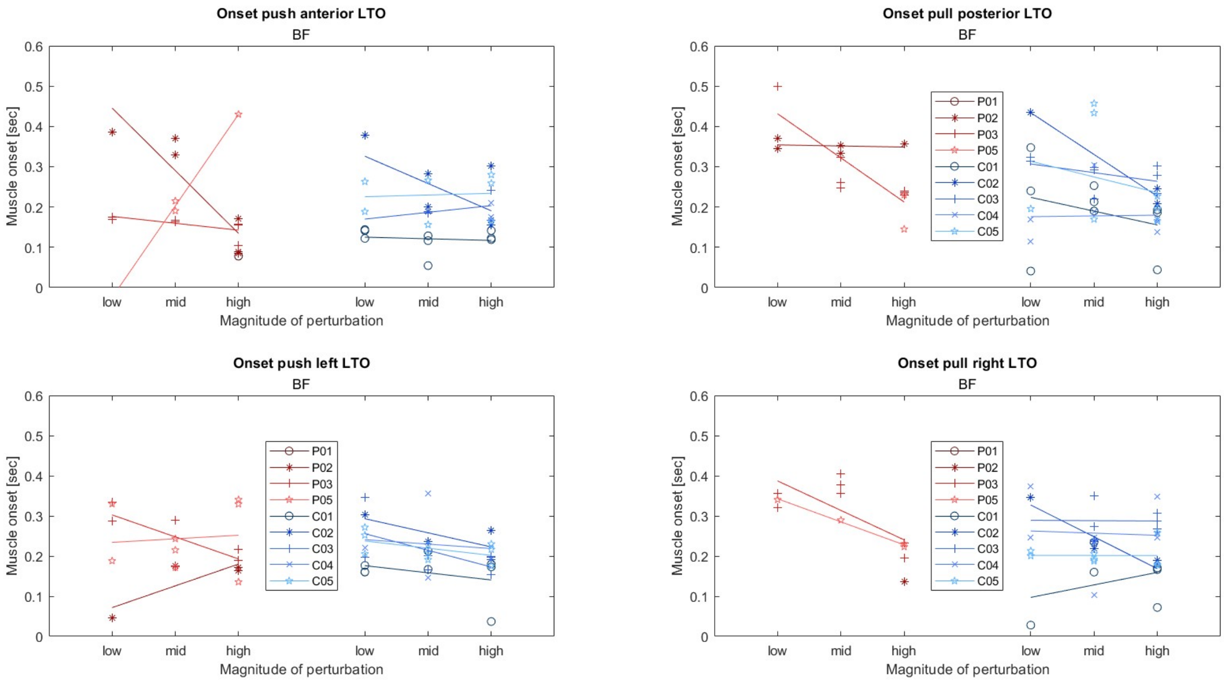

Figure A.25: M. Biceps Femoris onset after perturbation at left toe-off.

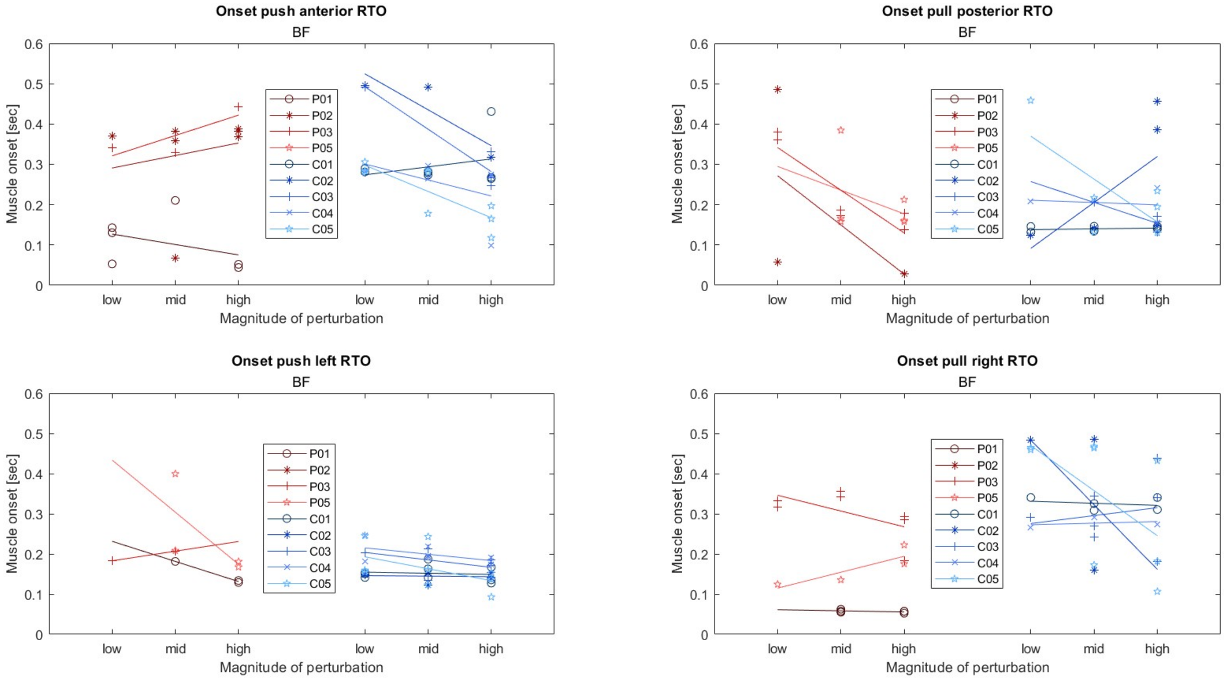

Figure A.26: M. Biceps Femoris onset after perturbation at right toe-off.

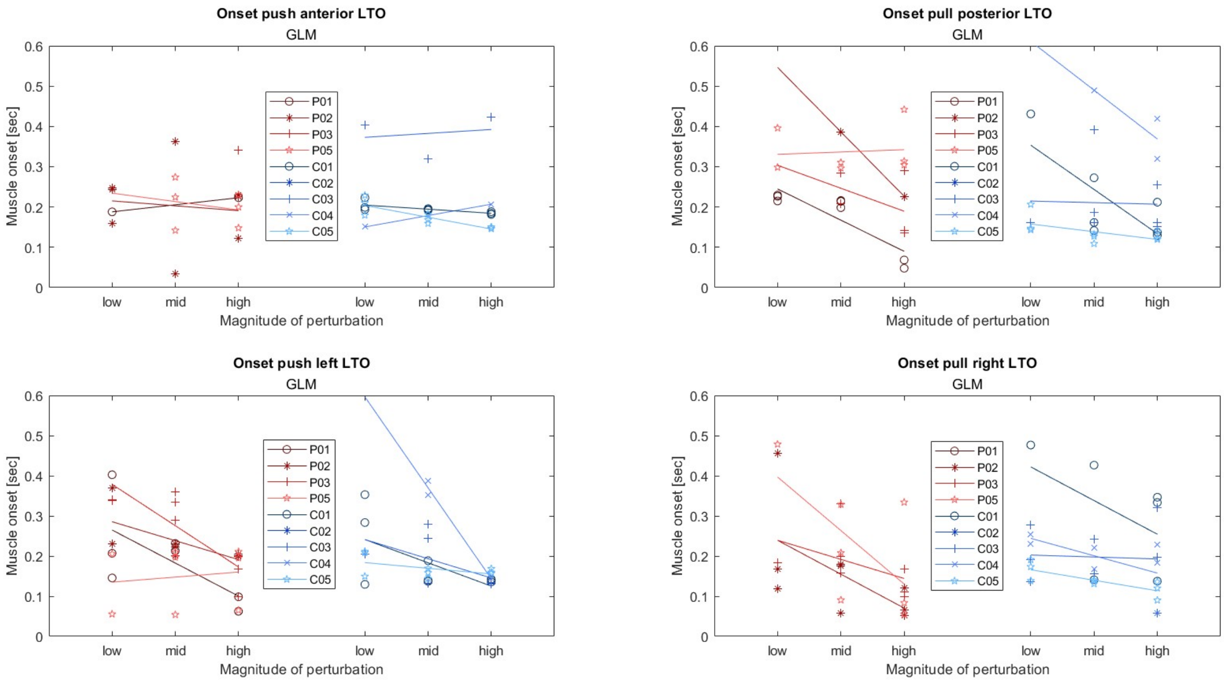

Figure A.27: M. Gluteus Medius onset after perturbation at left toe-off.

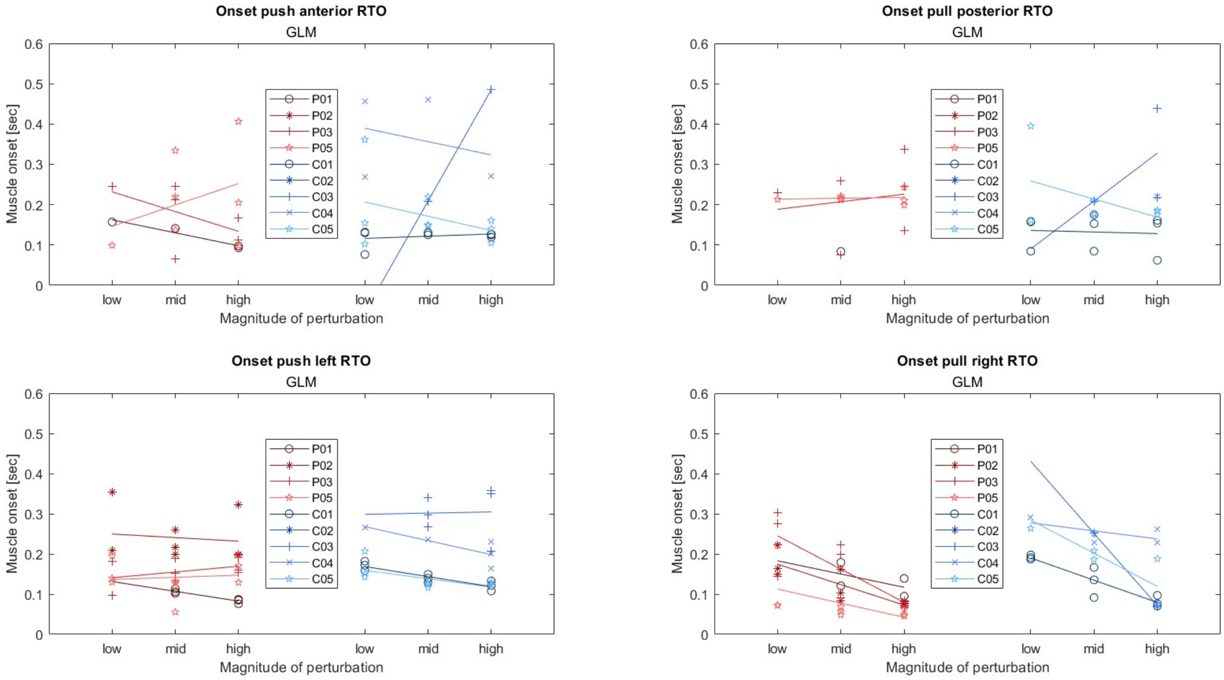

Figure A.28: M. Gluteus Medius onset after perturbation at right toe-off.

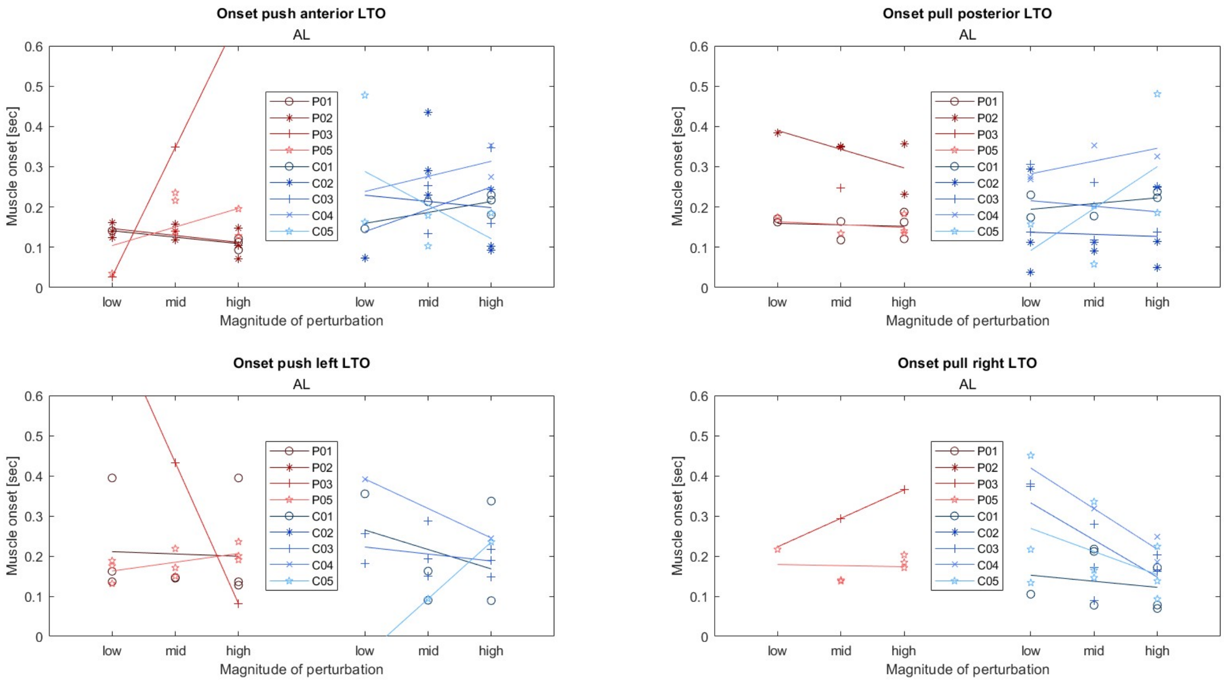

Figure A.29: M. Adductor Longus onset after perturbation at left toe-off.

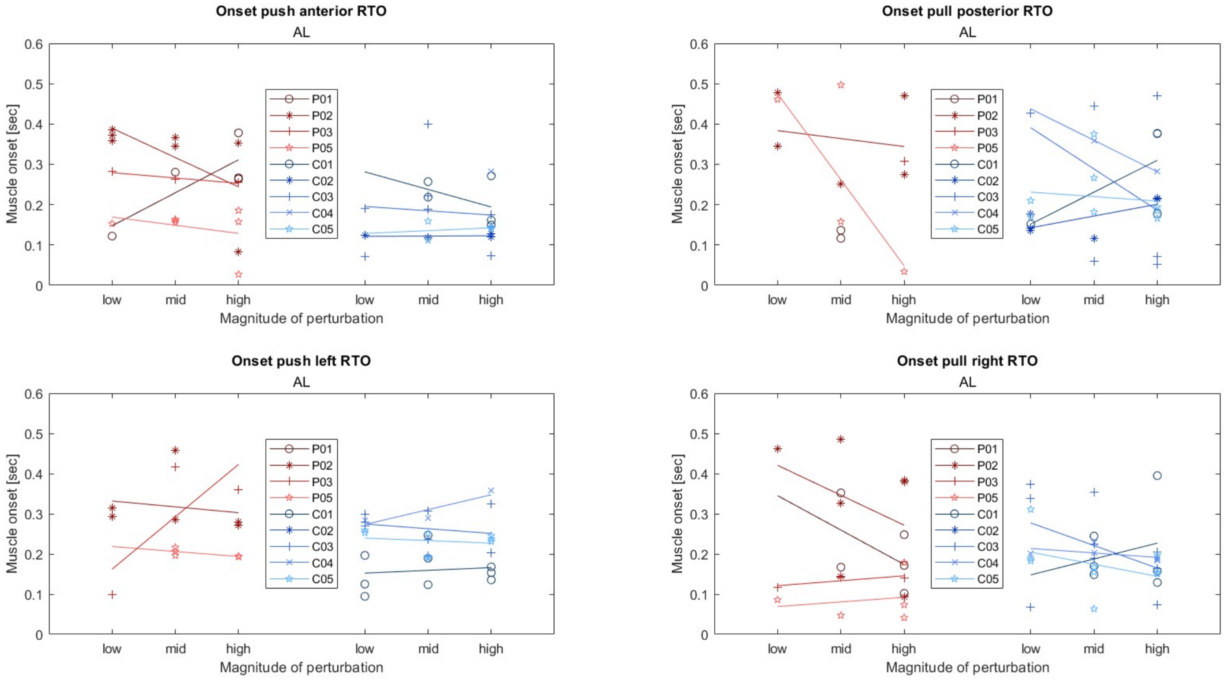

Figure A.30: M. Adductor Longus onset after perturbation at right toe-off.

A.7 Muscle effort

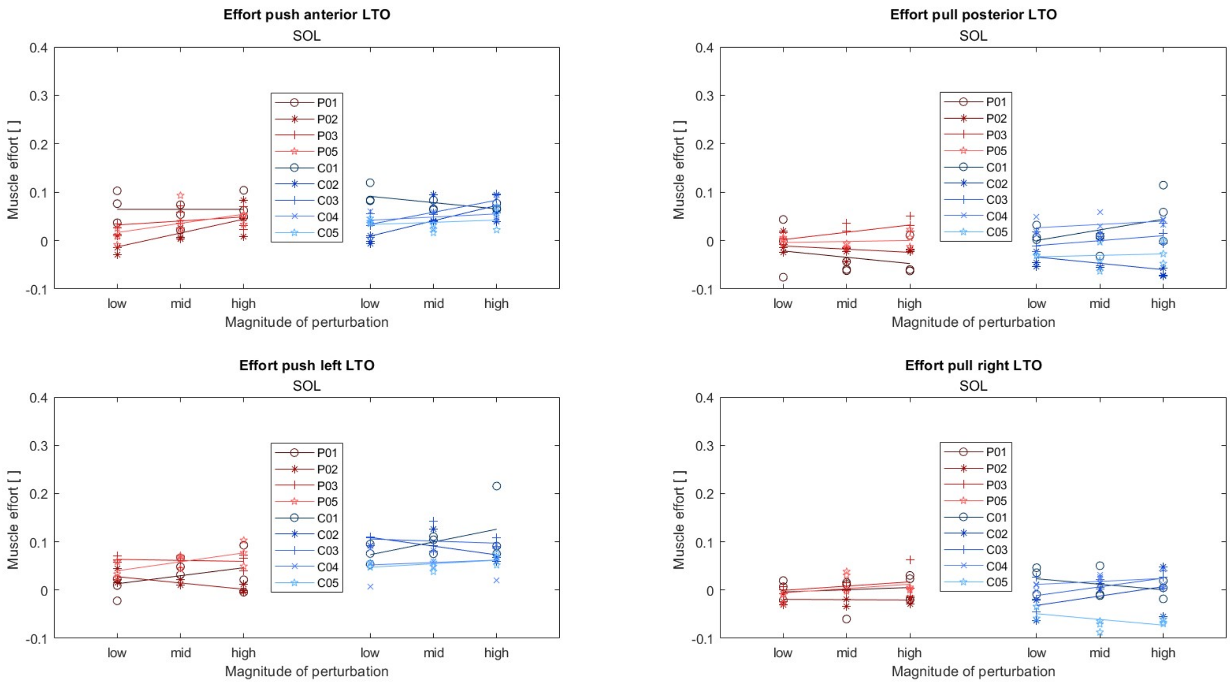

Figure A.31: M. Soleus effort after perturbation at left toe-off.

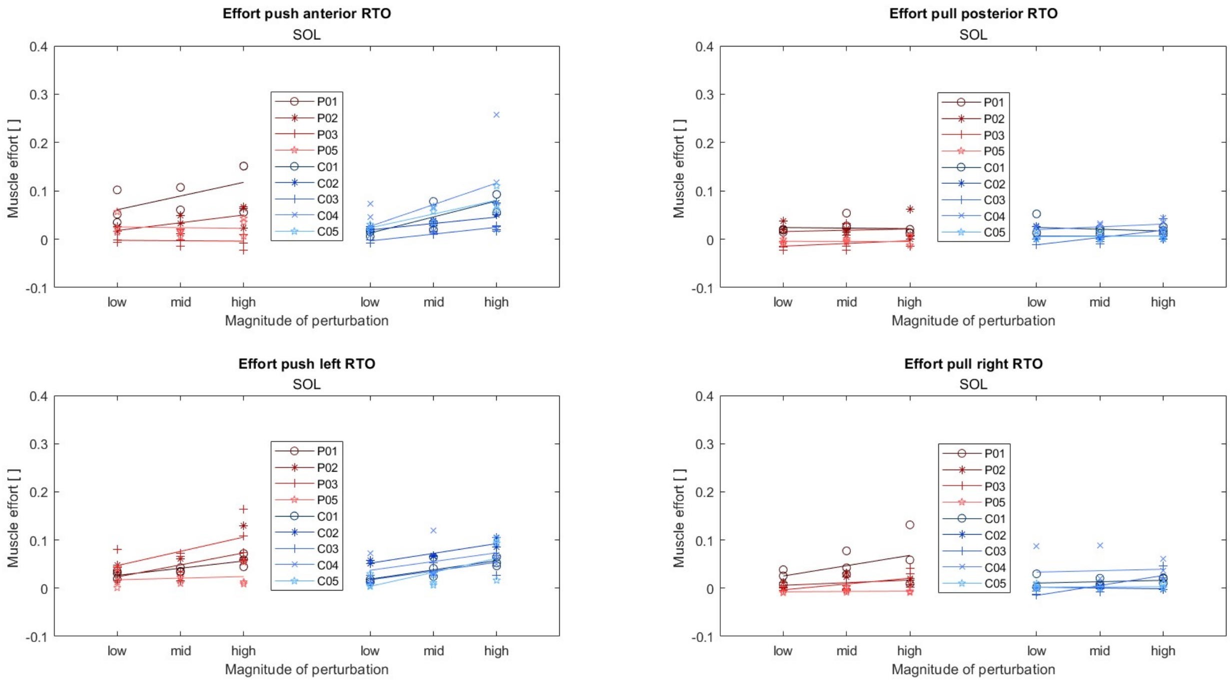

Figure A.32: M. Soleus effort after perturbation at right toe-off.

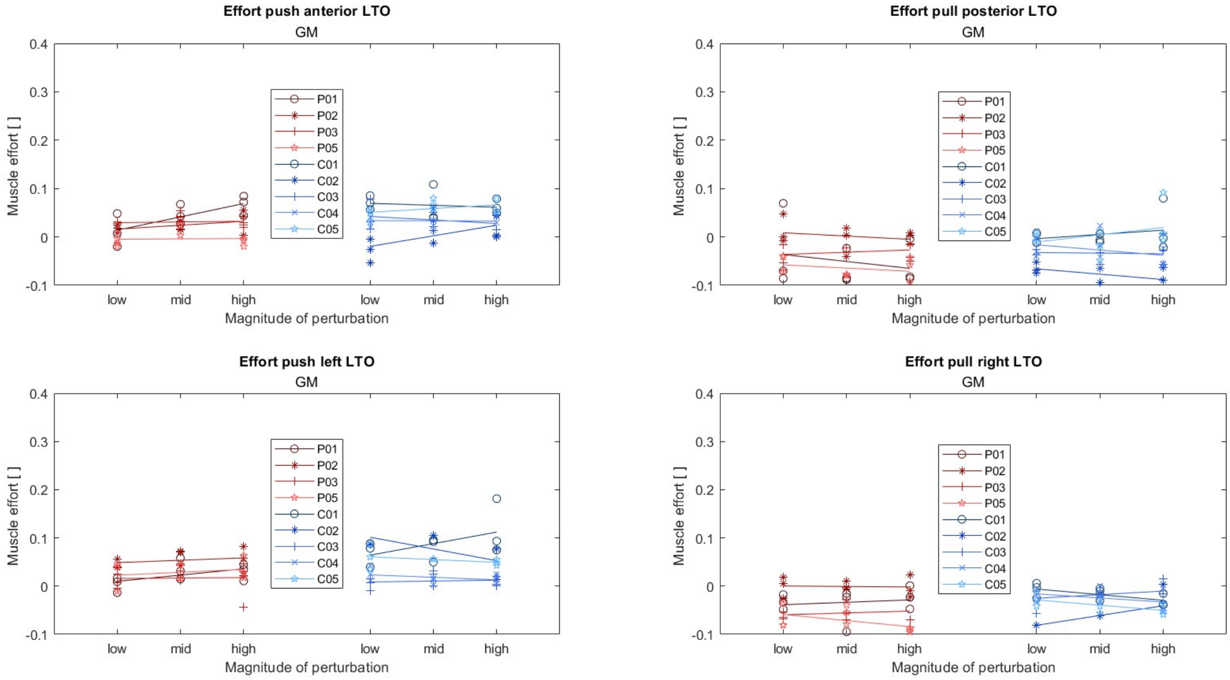

Figure A.33: M. Gastrocnemius Medialis effort after perturbation at left toe-off.

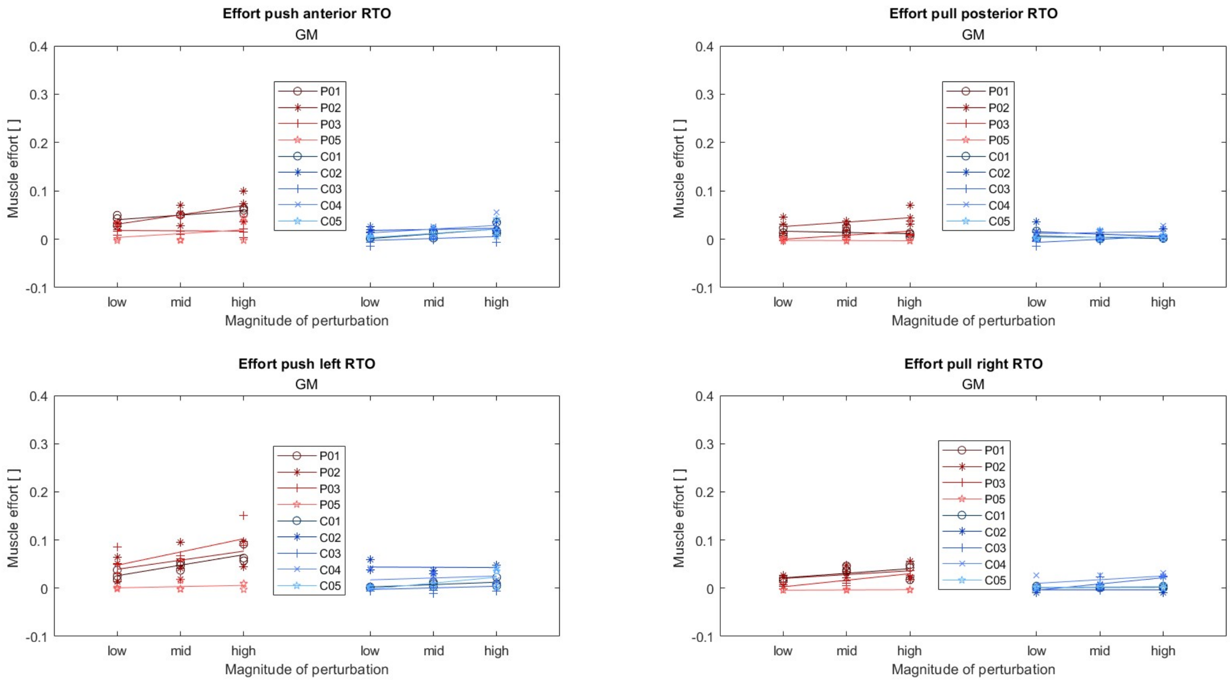

Figure A.34: M. Gastrocnemius Medialis effort after perturbation at right toe-off.

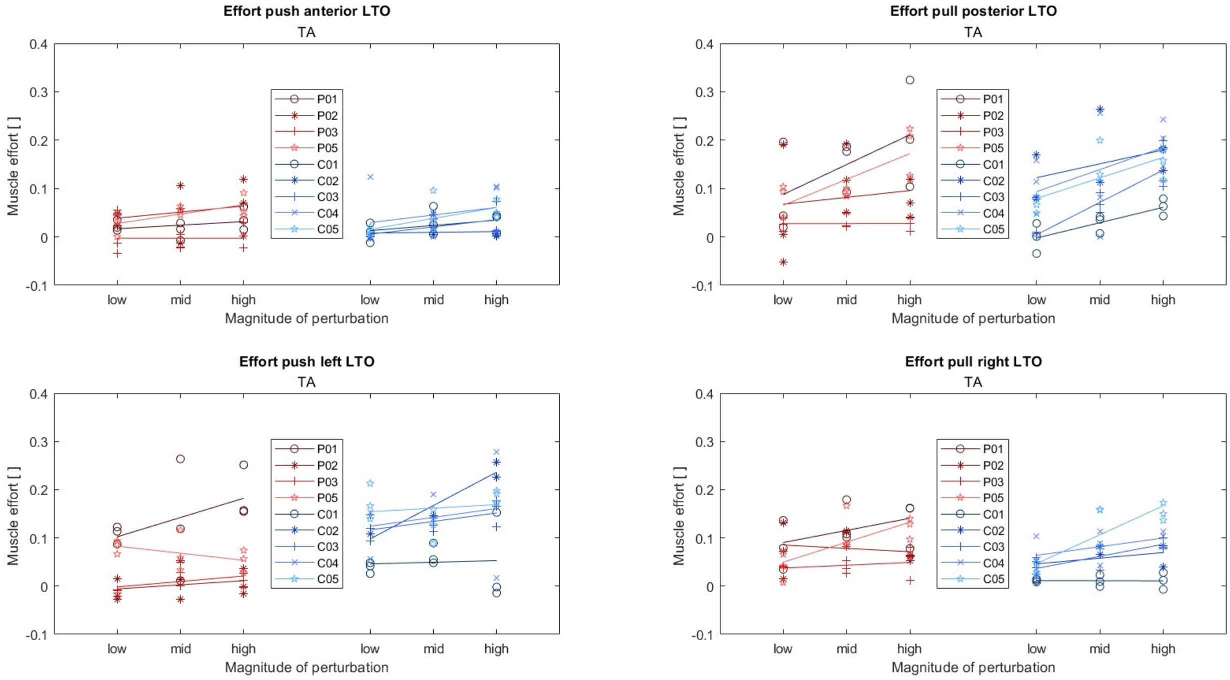

Figure A.35: M. Tibialis Anterior effort after perturbation at left toe-off.

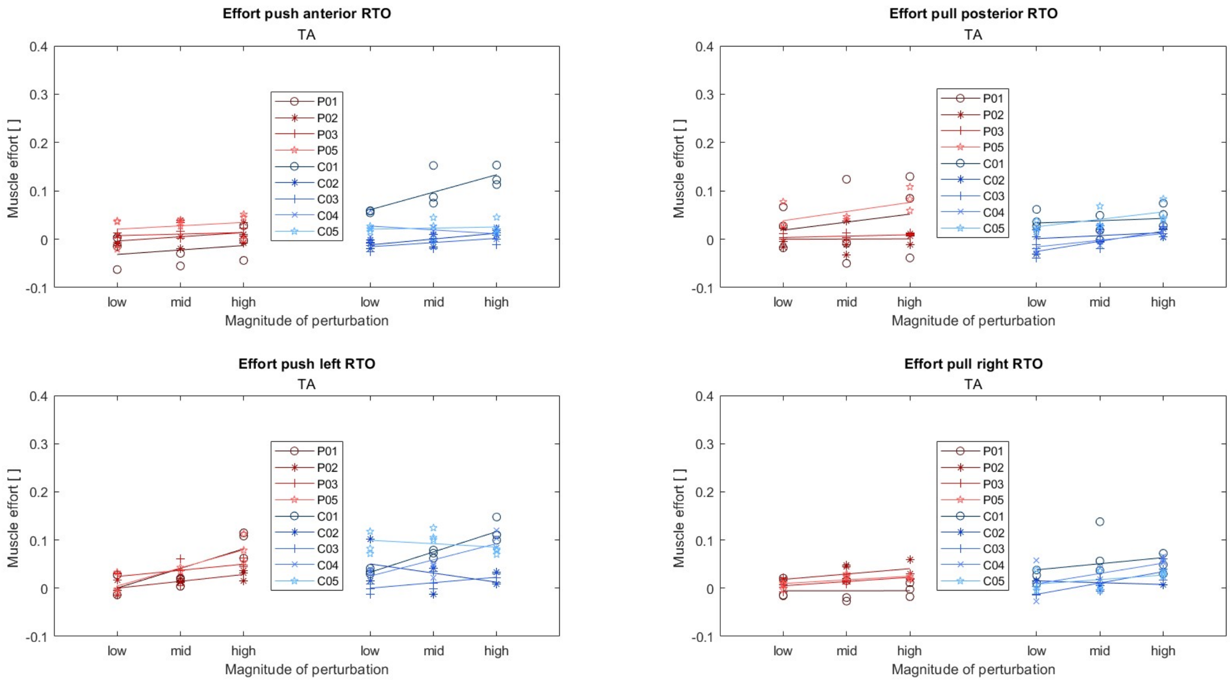

Figure A.36: M. Tibialis Anterior effort after perturbation at right toe-off.

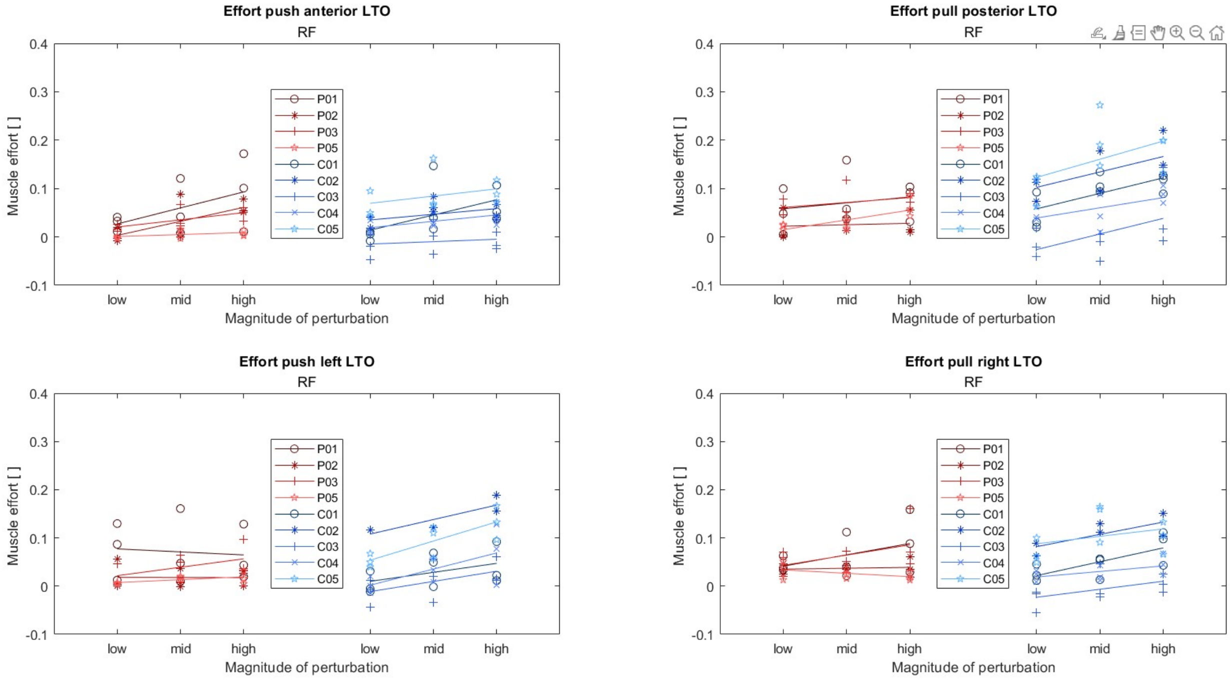

Figure A.37: M. Rectus Femoris effort after perturbation at left toe-off.

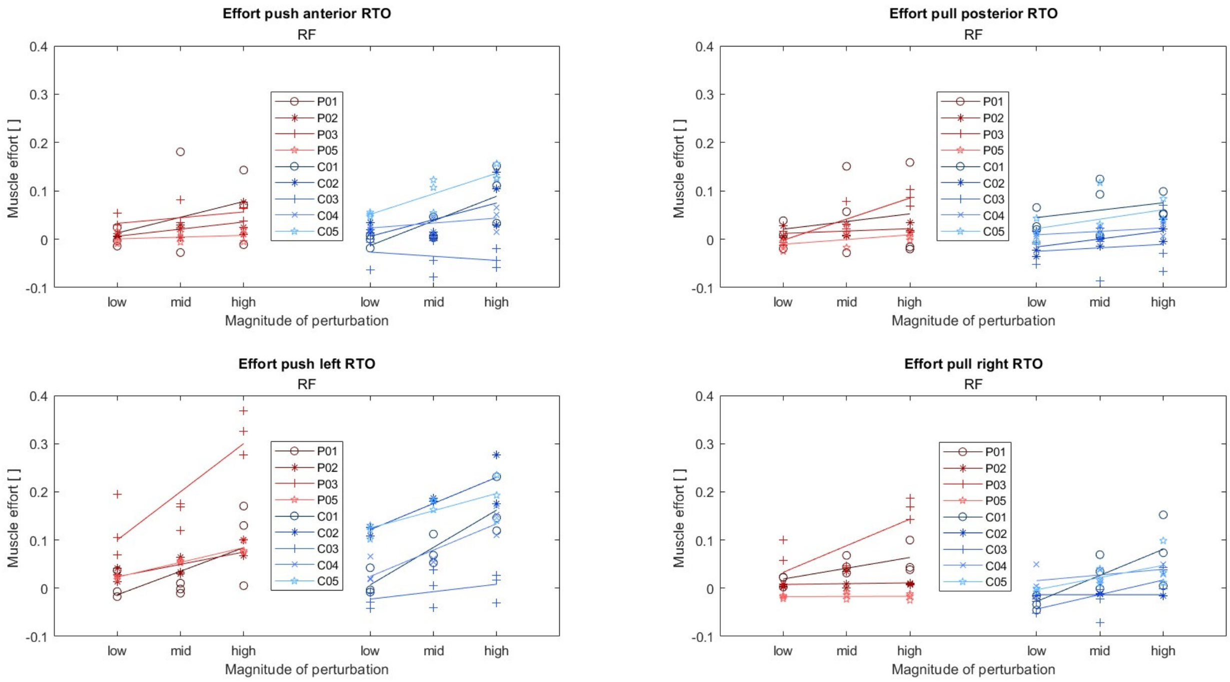

Figure A.38: M. Rectus Femoris effort after perturbation at right toe-off.

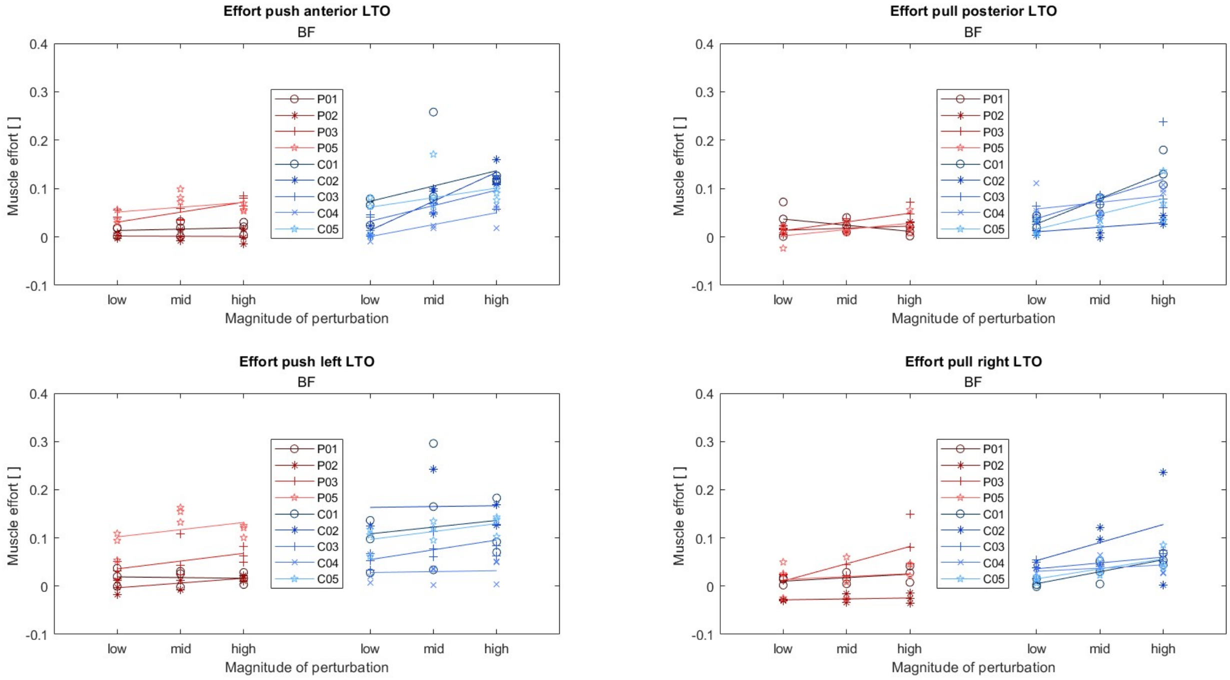

Figure A.39: M. Biceps Femoris effort after perturbation at left toe-off.

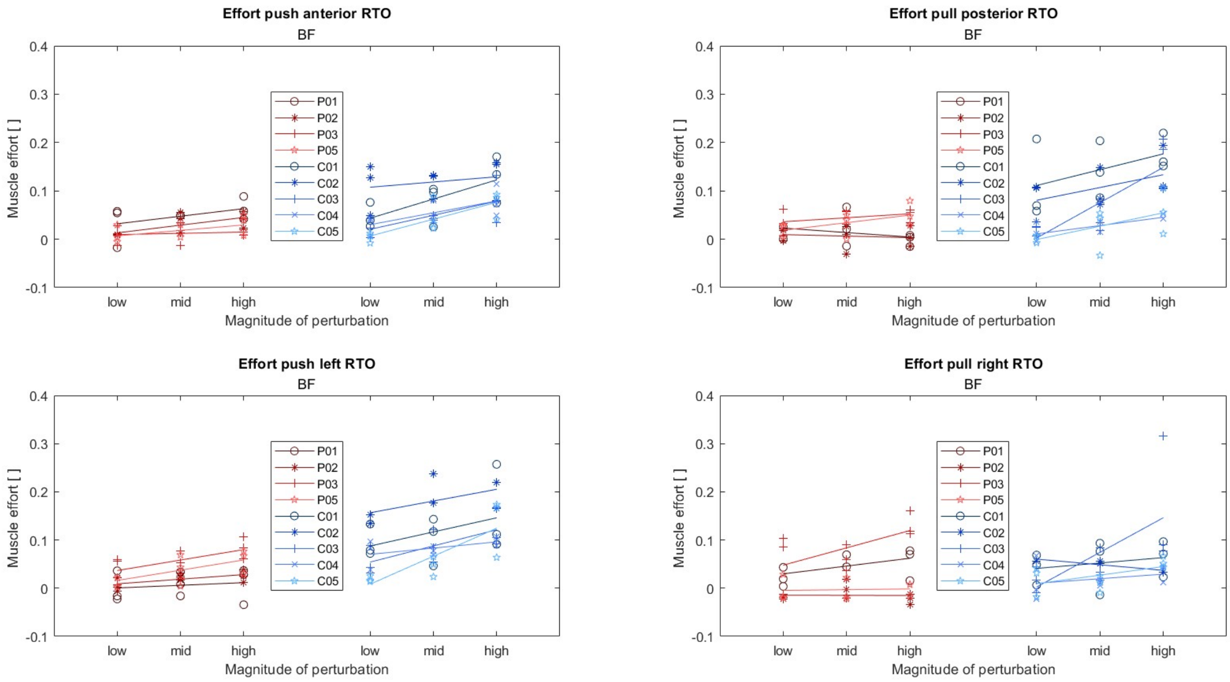

Figure A.40: M. Biceps Femoris effort after perturbation at right toe-off.

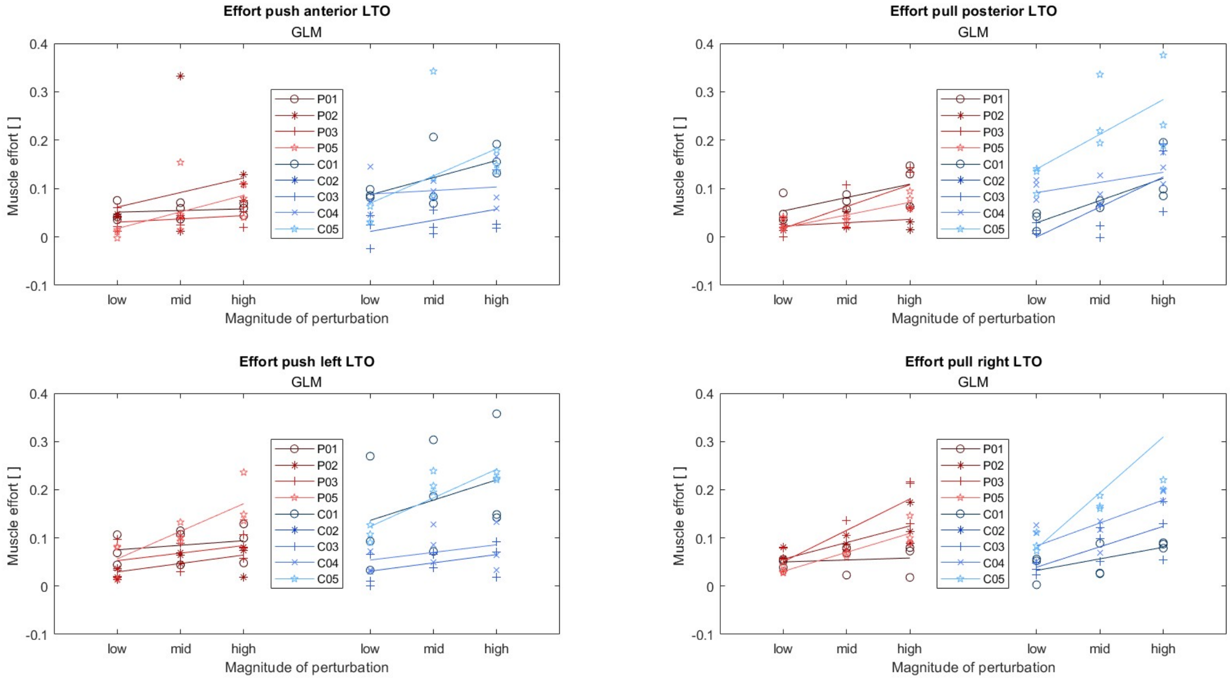

Figure A.41: M. Gluteus Medius effort after perturbation at left toe-off.

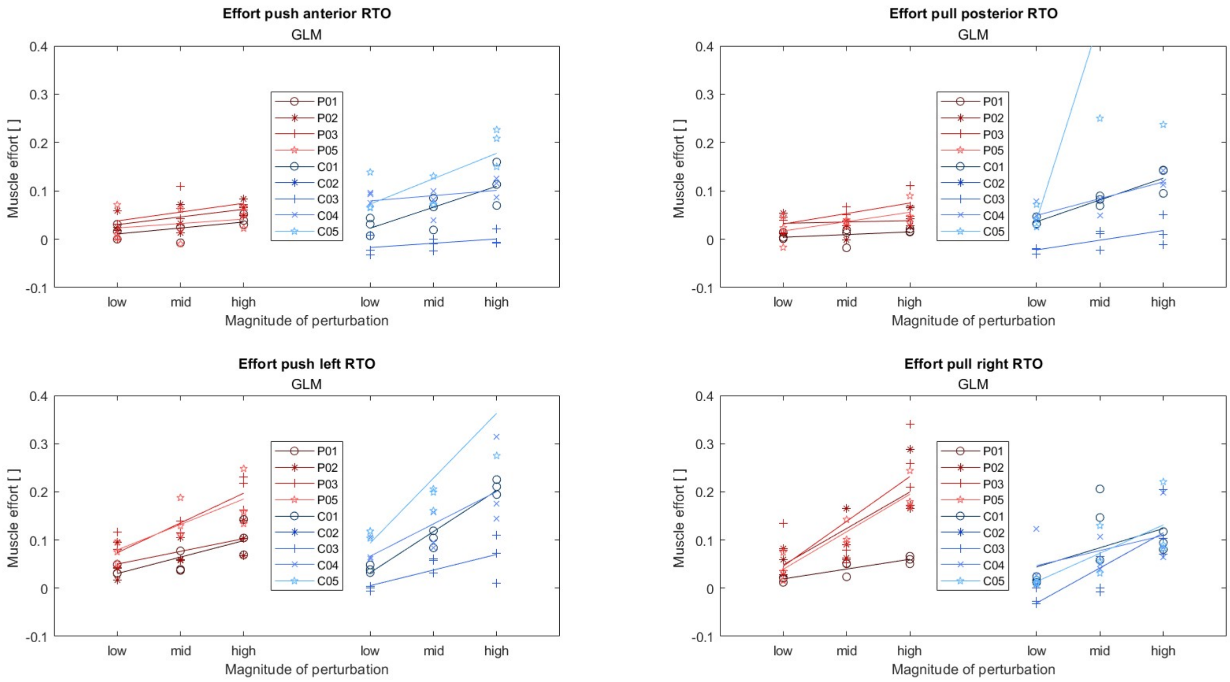

Figure A.42: M. Gluteus Medius effort after perturbation at right toe-off.

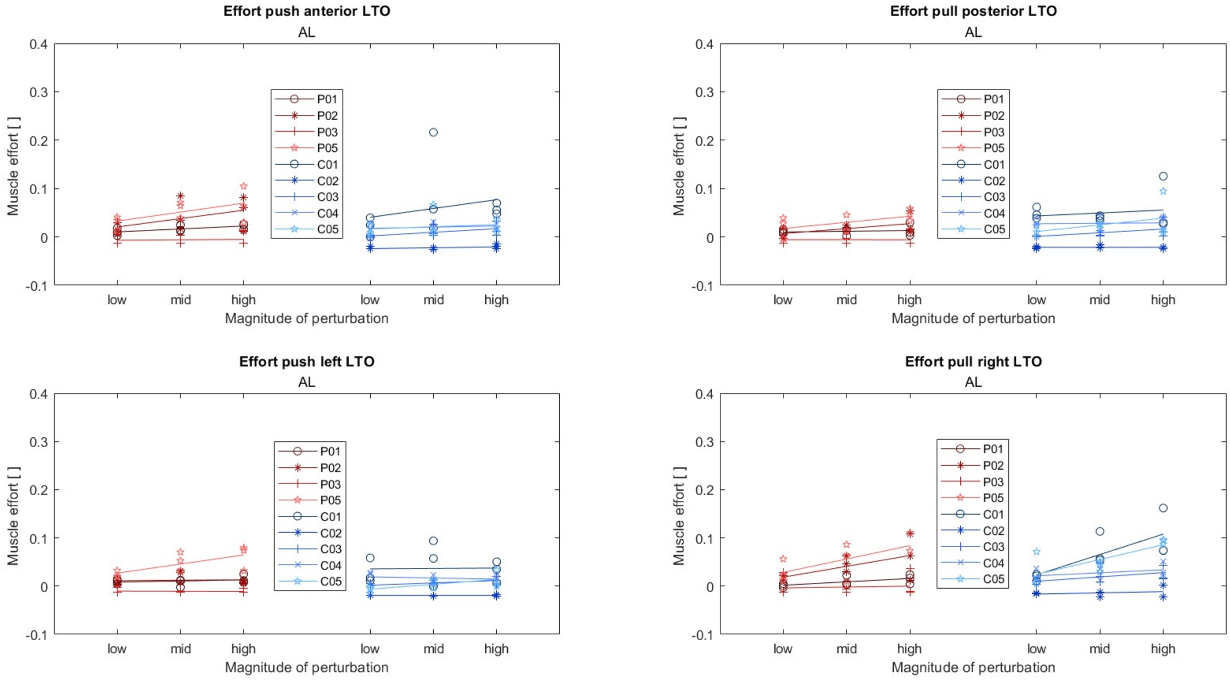

Figure A.43: M. Adductor Longus effort after perturbation at left toe-off.

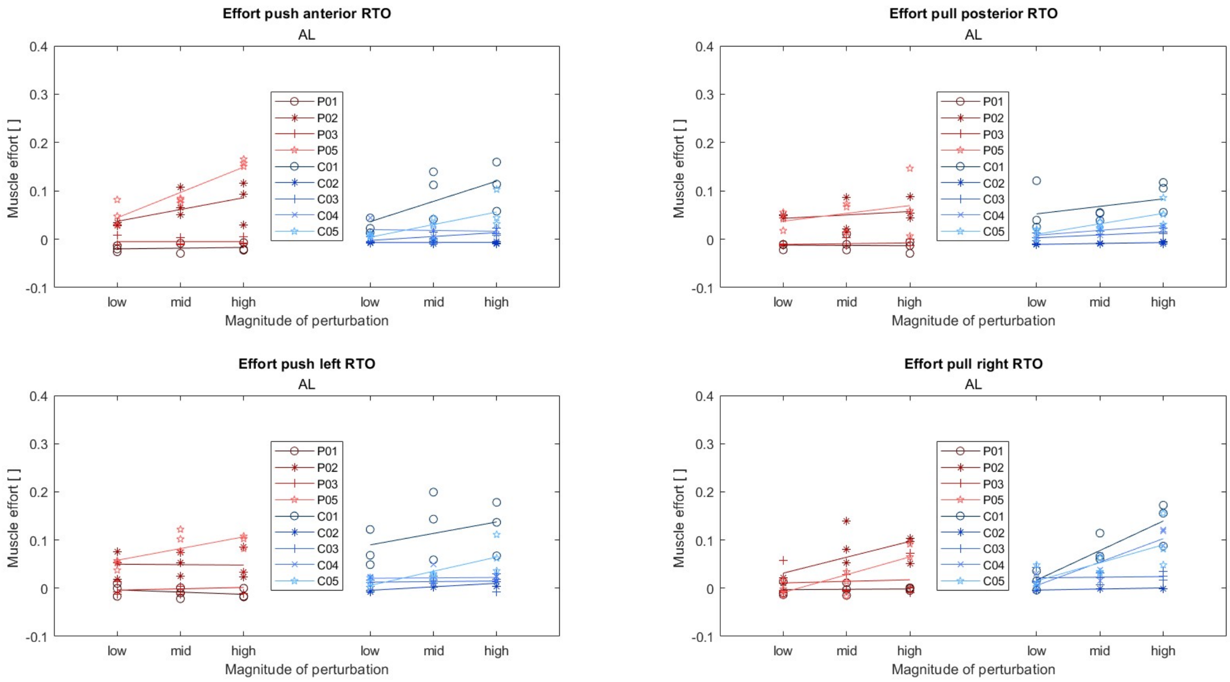

Figure A.44: M. Adductor Longus effort after perturbation at right toe-off.

A.8 Center of mass velocity at different conditions

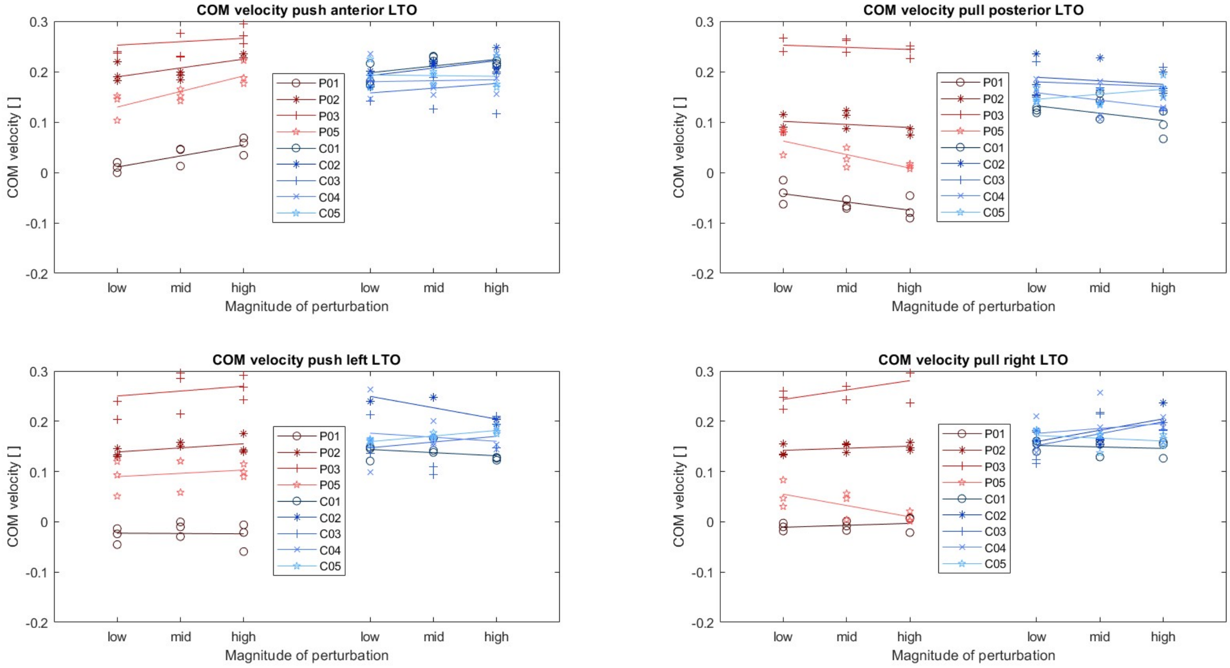

Figure A.45: Center of mass velocity after perturbations duration. Push initiation at left toe-off.

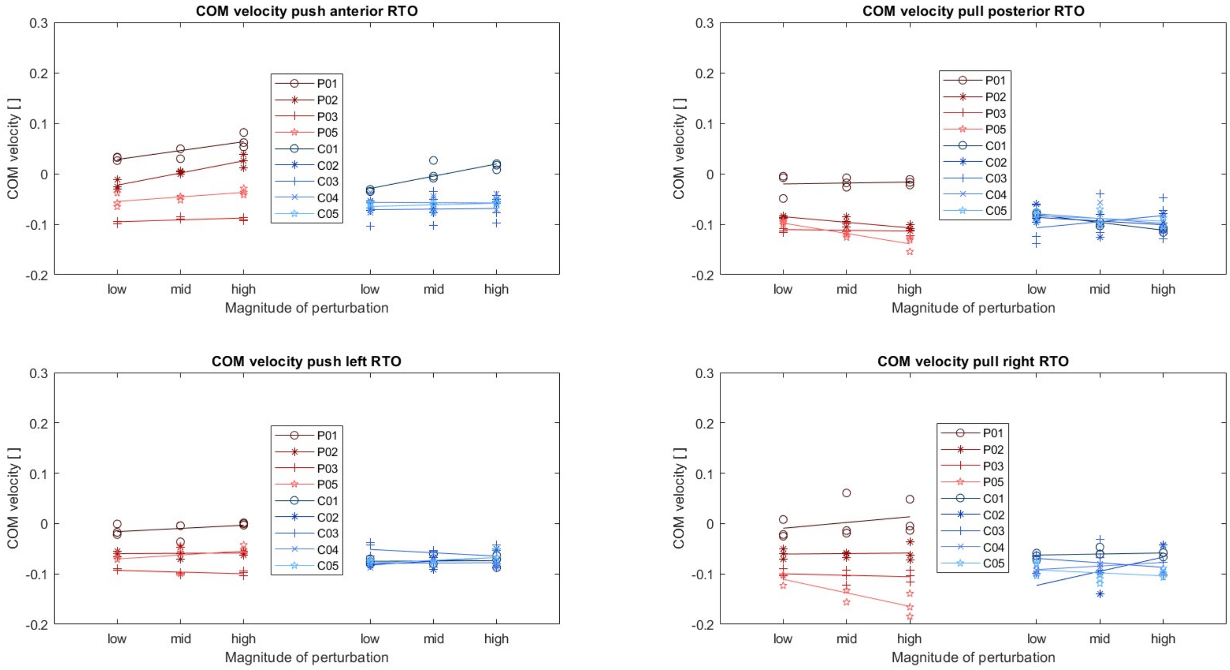

Figure A.46: Center of mass velocity after perturbations duration. Push initiation at right toe-off.

Table A9: The center of mass velocities for the different conditions at the end of the perturbation.

Perturbation initiation side	Perturbation type	Magnitude	Normalized COM velocity mean (std) Control	Normalized COM velocity mean (std) Patient
LTO	**Push anterior**	**Low**	0.0008 (0.0287)	0.0167 (0.099)
		Mid	0.0249 (0.0357)	0.0411 (0.0139)
		High	0.0503 (0.0253)	0.0668 (0.0211)
	Pull posterior	**Low**	-0.0697 (0.0279)	-0.0570 (0.0191)
		Mid	-0.0805 (0.022)	-0.0797 (0.0243)
		High	-0.1074 (0.231)	-0.0934 (0.0292)
	Push left	**Low**	-0.0418 (0.0256)	-0.0290 (0.0139)
		Mid	-0.0419 (0.0332)	-0.0272 (0.0169)
		High	-0.0387 (0.0304)	-0.0251 (0.0146)
	Pull right	**Low**	-0.0361 (0.0299)	-0.0281 (0.0157)
		Mid	-0.0323 (0.0277)	-0.0228 (0.0178)
		High	-0.0283 (0.0231)	-0.0219 (0.206)
RTO	**Push anterior**	**Low**	0.0468 (0.0225)	0.0284 (0.0093)
		Mid	0.0680 (0.0189)	0.0514 (0.0128)
		High	0.0862 (0.0183)	0.0745 (0.0252)
	Pull posterior	**Low**	-0.0209 (0.0238)	-0.0427 (0.0212)
		Mid	-0.0402 (0.0310)	-0.0552 (0.0346)
		High	-0.0578 (0.0259)	-0.0691 (0.0433)
	Push left	**Low**	0.0073 (0.0254)	-0.0146 (0.0151)
		Mid	-0.0725 (0.0215)	0.0077 (0.0187)
		High	-0.0699 (0.0273)	0.0055 (0.0113)
	Pull right	**Low**	-0.0017 (0.0229)	-0.0193 (0.0132)
		Mid	-0.0002 (0.0284)	-0.0118 (0.0253)
		High	0.0064 (0.0334)	-0.0194 (0.0253)

A.9 Muscle effort - COM velocity relation

A.10 After push forward

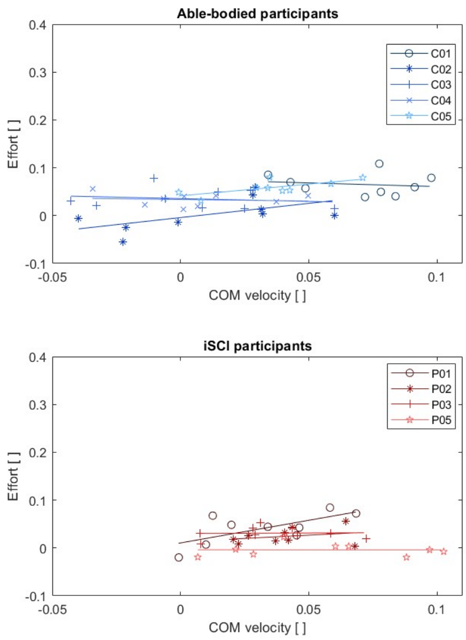

Figure A.47: Effort - COM relationship for the GM after perturbation at LTO in forward direction.

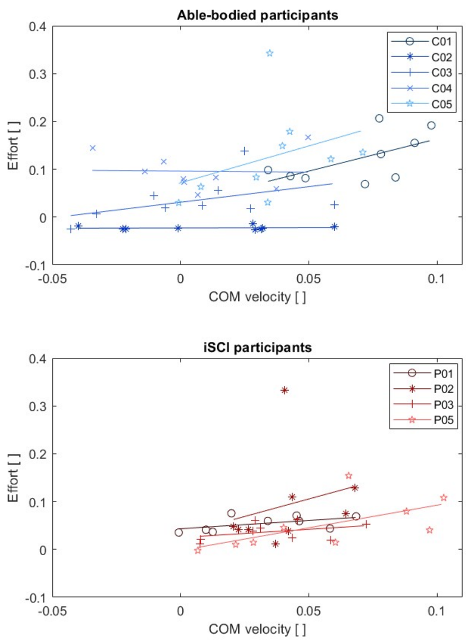

Figure A.48: Effort - COM relationship for the GLM after perturbation at LTO in forward direction.

A.11 After pull backward

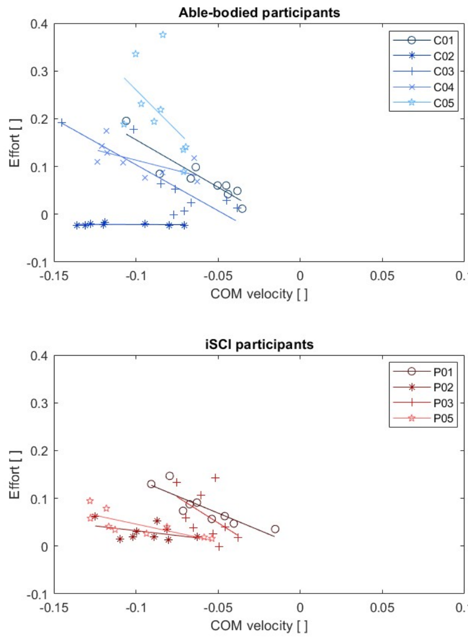

Figure A.49: Effort - COM relationship for the GLM after perturbation at LTO in backward direction.

A.12 After push leftward

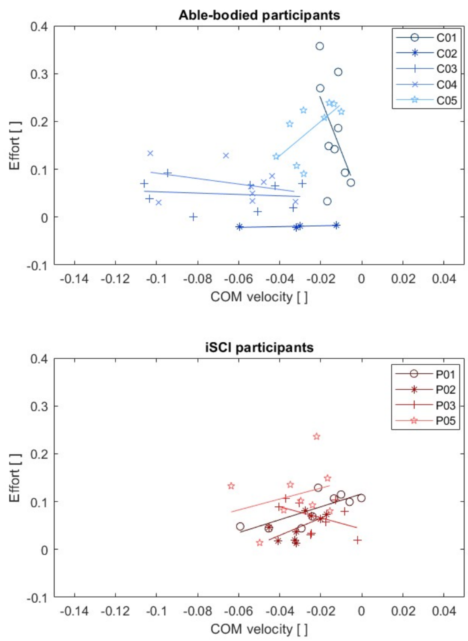

Figure A.50: Effort - COM relationship for the GLM after perturbation at LTO in leftward direction.

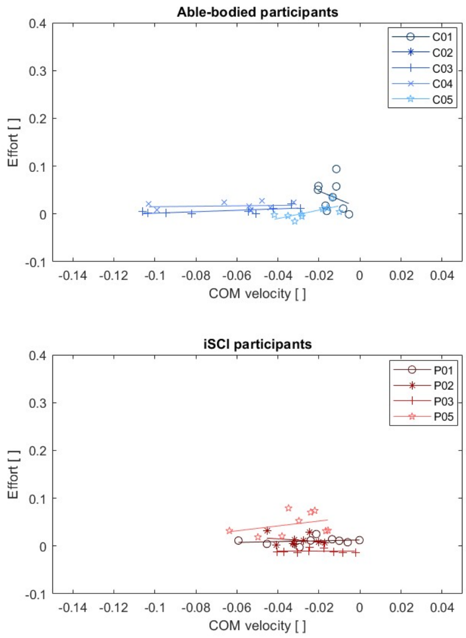

Figure A.51: Effort - COM relationship for the AL after perturbation at LTO in leftward direction.

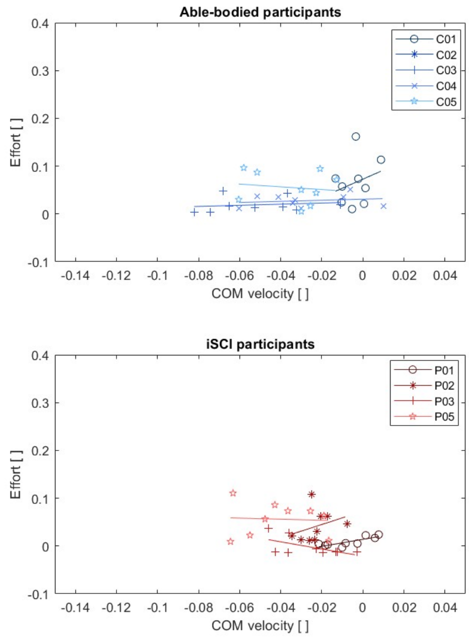

Figure A.53: Effort - COM relationship for the AL after perturbation at LTO in rightward direction.

A.13 After pull rightward

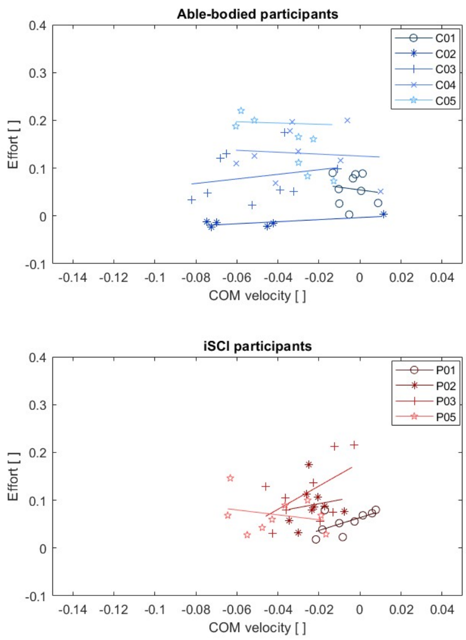

Figure A.52: Effort - COM relationship for the GLM after perturbation at LTO in the rightward direction.