Ipsilateral sensory feedback in soleus EMG activity during stair ascent

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Abstract

Background and aim: A constant adaptation of the locomotor output from sensory afferent feedback ensure efficient and stable level gait, especially in a variable environment. The sensory feedback involvement during more complex motor task, like stair climbing, are not yet known. The present study aims to investigate the involvement of the sensory afferent feedback contribution to the locomotor motoneuronal drive and to the corrective stretch reflex during stair climbing. Subjects climbed unrestrained a seven step perturbation stair apparatus. On random trials, the fourth step was rotated downward (50 mm, x g acceleration), which could either cause an vertical perturbation or a unload perturbation mid-stance. Horizontal perturbation of the stepping surface could be moved inward in the parasagittal plane with respect to level ground. Three protocol was tested, where 9 subject completed the Vertical and Horizontal perturbation, and 4 subject finished the unload perturbation. The vertical perturbation revealed a positive correlation between the elicited soleus EMG response and the ankle velocity at the perturbation onset. An significantly elevated medium-latency reflex was seen as an consequence of the horizontal perturbation in the soleus amplitude. The induced unload responses did not reveal any meaningfully result and are to be investigated further.

The interaction between sensory afferents and the central nervous system during human locomotion is well-documented [27]. According to Nielsen and Sinkjaer [26], the interaction is generally split into two fundamentally different types: 1; Sensory afferents are integrated in the internal commands for driving motor-neurons [2, 18, 36]. Type 2; the essential information about the error between intended and executed movement from sensory afferents in the presence of disturbances [6, 11, 15, 42] is used in a feedback loop to achieve disturbance rejection through corrective reflexes [27].

However, despite this knowledge, the extent of and finer details concerning the interaction between sensory afferents and the central nervous system are not yet fully known. Most current studies have examined the interaction during walking on a smooth level surface [2, 36], and thus it is still largely unknown to which extent, if any, the sensory feedback is involved in other human behavioral patterns.

This paper aims to examine this, by studying the role of sensory afferents on corrective reflexes during a more complex functional task in the form of climbing a staircase. During the ascend, exact foot placement and proper leg swing are of higher priority compared to level walk, and thus stair climbing requires greater lower-limb joint demands [8, 24], higher energy cost due to added vertical translation [4, 39], and offers a more challenging dynamic balance situation [20]. Based on these dissimilarities, it is assumed that ascending a staircase requires altered muscle contributions and sensory feedback. The following sections concerns sensory feedback and its use in reflex-based disturbance rejection during locomotion.

As mentioned previously, one crucial function of reflexes during locomotion is that of disturbance rejection. During leg movement e.g. walking it may happen that the swinging leg collides with an obstacle, the standing foot slips, or other nonpredictable disturbances occur. Corrective reflexes describe the feedback of such an unexpected sensory afferent [26]. These corrective reflexes are, by the stochastic timing of their cause, not anticipated by the central nervous system and aim to error-correct the ongoing movement to avoid harmful circumstances such as falling [26]. An example of such a corrective reflex might be the soleus stretch response activated by rapid dorsiflexion of the ankle during level walking. This dorsiflexion response has been demonstrated through an external mechanical perturbation [34, 43].

Through experiments it has been shown that during walking the soleus stretch response consists of three different components: 1; Short-Latency Response (SLR) mediated mainly by velocity sensitive group Ia [16, 23, 38], 2; spinal Medium-Latency Response (MLR) presumably mediated by length sensitive group II [[16, 32] and force-sensitive group Ib afferents [18], and 3; Long-Latency Response (LLR) suggested to be transcortical or subcortical responses [23, 40, 41]. These experiments thereby suggest that a corrective reflex is mediated both at cortical and spinal origin during walking on a level surface. The SLR has surprisingly not been observed during experiments where test subjects are exposed to sudden, rapid acceleration of the walking surface using a treadmill [7, 12, 13, 25], which is assumed to elicit a more natural reflex response compared to the external mechanical dorsiflexion. Thus, the short-latency reflex might not occur during normal locomotion and, therefore, not reflect an autogenetic corrective stretch reflex.

On the contrary during hopping, an early burst can be observed in the EMG from the soleus muscle starting about 45 ms after touch-down, Dyhre-Poulsen et al. [14], Zuur et al. [44]. The early EMG burst, when advancing or delaying the touch-down without the subject's knowledge, advance or delay the burst [44]. The soleus EMG burst during hopping about 45 ms after touch-down suggests that the group Ia afferent might mediate the EMG response. However, this reflex may be expectedly and can therefore not be a response considered as an unexpected sensory error. This claim is in line with the finding that monkeys trained to land on a solid surface demonstrated muscle activity arising already approximately 38 ms before landing, even though the height was varied [19].

Contributions from sensory feedback to the motoneuronal drive during walking have been proven by unloading the ankle extensor muscle. af Klint et al. [2] unloaded the ankle extensor by decreasing muscle load due to a drop of supporting ground, and Sinkjær et al. [36] similarly implied a rapid shortening of ankle extensors muscles through an external mechanical perturbation. The unloading cause a significant depression in the electromyographic (EMG) activity, with a latency of 60ms [2, 36]. Observed depression latency implies that the sensory feedback acts as a feedback to the motoneuronal drive during walking. As Petersen [30] finds, the minimum time for a transcortical stretch reflex in the tibialis muscle is likely to be over 75 ms. The distance from the tibialis to the cortex is equal to the distance from soleus to the cortex [44].

This leaves the question to which degree corrective reflexes are spinal mediated and whether the short latency stretch reflex contributes to the corrective reflex. Currently, most of the studies on sensory afferents in human locomotion have focused on the role of sensory contribution during level walking. It is still, however, unclear to what extent sensory feedback is involved in generating human locomotion in other functional tasks such as during stair climbing and the exact involvement of sensory afferents during corrective reflexes is neither well documented.

This study therefore aims to examine whether spinal circuits are involved in generating locomotion activity during stair climbing and aim to investigate whether a natural disturbance of the ankle joint by a horizontal/vertical perturbation of a stair steps might elicit an autogenetic corrective response. With the purpose to investigate the afferent feedback, a staircase apparatus is constructed with the option of controlling one of the steps on the stairs. This step is denoted as the perturbation-step, which should be able to deliver well defined horizontal inward and vertical downward translational movements. The resulting perturbation will cause a change in the walking pattern of a test subject, which can be used to investigate the electro-physiological and bio-mechanical features of human stair climbing.

I. Method

Thirteen able-bodied subjects are recruited for the first two protocols; vertical and horizontal perturbation. 10 subjects were recruited for the third unload protocol. Subjects were recruited from the local university's staff and students with no known history of neuromuscular disorders. Subjects gave their informed consent prior to the experiment.

For the first two protocols, four subjects were excluded from the test sample size of thirteen test-subjects due to complications. One subject was excluded due to mechanical issues with the perturbation apparatus. Two subjects were excluded due to insufficient electrical recordings caused by sweat obstructing the EMG recordings. Further, one subject was unable to follow the study protocol (subject stopped prior to the induced perturbation) and was therefore excluded. This resulted in a total sample population of eight subjects in the two first protocols.

Four subjects completed the third protocol, but further trials were terminated due to mechanical issues with the test setup.

A. Apparatus and instrumentation

All data presented was recorded from the left leg. Subjects were instrumented with bipolar surface EMG electrodes (NeuroLine 720, AMBU A/S, Denmark) with an individual ground on the soleus (SOL) and tibialis anterior muscles of the left leg. The resulting perturbation was presumed to change the muscle activity in SOL, but as it cannot be ruled out that the induced ankle plantar flexion could produce an increased activity in the tibialis anterior (TA) muscle, that through reciprocal inhibition would reduce the SOL muscle activity, the muscle activity of tibialis anterior (TA) was also monitored throughout the trials. The EMG signals were amplified and band-pass filtered (10–1 kHz) using an analog filter and custom-built amplifiers. Electrodes were placed over the muscle belly, in line with the muscle, with an inter-electrode distance of 3-cm between electrode centres. The placement of the EMG electrode was in accordance with the guideline provided by Seniam [33]. Foot-strike was recorded using a force sensitive resistor (Interlink FSR, LuSense) placed under the left shoe-sole (under the palm of the foot). Left angle position was recorded using a surface mounted goniometer (model SG150; Biometrics, Gwent, UK). The goniometer was placed over the ankle joint and taped with double-sided tape to the skin of the ankle and the shoe heel. The goniometer output was calibrated to reflect the true joint angle using a video recording of one subject's stair gait.However raw data was used when comparing data across subjects.

EMG and kinematics were recorded with lightweight custom-built wiring for the Data Acquisition (DAQ) system. The subject was never obstructed by the wires. The data was recorded using Mr. KICK 2 software [37], with a sampling frequency of 2 kHz and saved for later process-

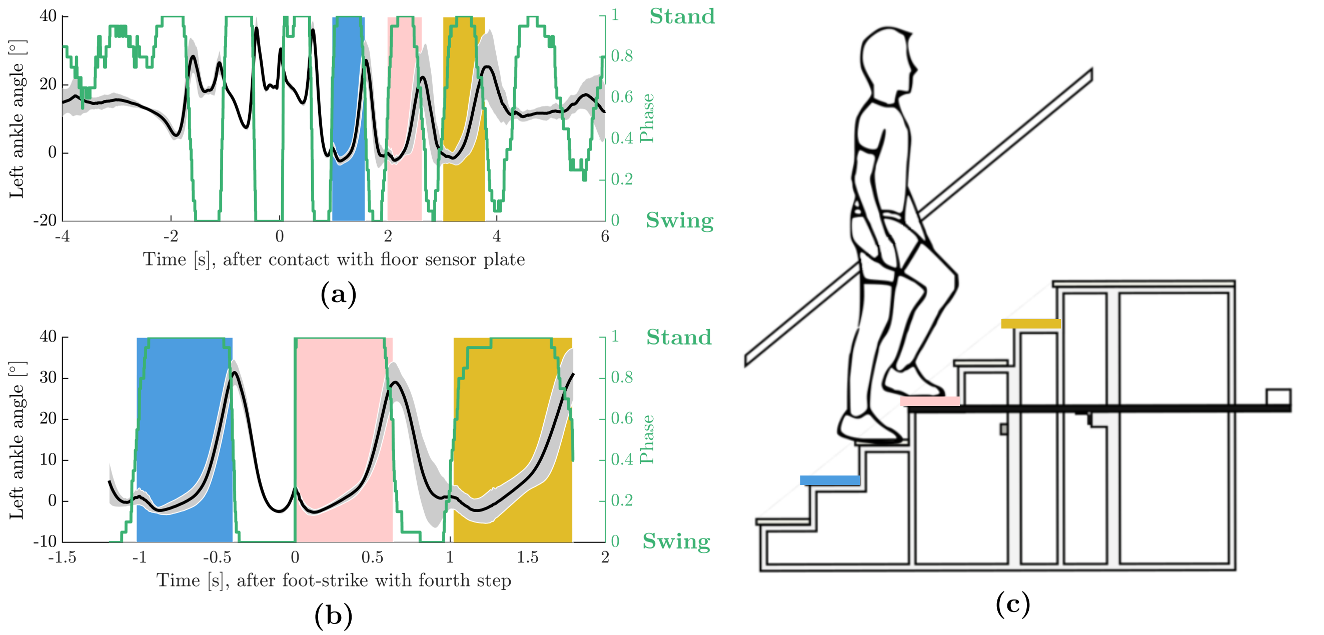

Figure 1. (a) Average left ankle trajectory from a representative subject during a complete trial. Black line, average represent the average ankle position (± 1 SD). Yellow line; average foot-constant and lift-off. Each step is marked with a patch color. These patches only represent an average estimate of where the stand-phase begins and ends due to the subject walked with varied speed (a) Same data as from figure A, but data is aligned with foot-strike of the fourth step. (c) Staircase apparatus. where each step is marked in the ankle trajectory with the same color as the patches of (a) and (b).

ing. A video camera (Sony α-57, AVCHD progressive) was used to record foot placement of two subjects on the fourth stair tread. The camera was set up so to point orthogonally to the walking direction of the test subject. The horizontal foot placement could thereby be obtained along the walking direction.

1. *Device to induce reflexes*

Vertical and horizontal perturbations were imposed by a custom-built staircase apparatus [5]. The entire system consists of a seven step staircase with a platform located in prolongation of the seven steps. The fourth stair step was constructed to perform either a vertical or horizontal perturbation, based on the system setup. An overview of the staircase apparatus can be seen on Figure 1(c), where the steps making contact with the left leg are marked grey, red and yellow; the red marked step (fourth step) is the perturbation step. Both the horizontal and vertical perturbations are produced through a rotary movement mechanism. A solenoid actuator can elicit the perturbation by disengaging a supporting locking mechanism under the stair tread. The perturbation step is only upheld at a pivoting joint when the locking-mechanism is disengaged and thereby enables the perturbation step to pivot downward to a specified position when stepped upon by a test subject.

The vertical perturbation did not produce a perfect vertical displacement but a 2.6° forward pitch was also generated from a 50 mm downward displacement (see Figure 2). The horizontal perturbation was achieved

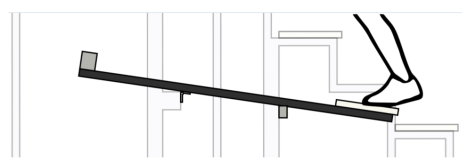

Figure 2. A vertical perturbation. The step surface can translate 50 mm downward when the fourth step is disengaged. The magnitude of the perturbation can be adjusted by moving the stop-block.

through the same mechanical mechanism as in the vertical perturbation, but only a 100 mm section of the step surface could be moved in a downward direction [See Figure 3]. The inner section of the step surface sustained its position during horizontal perturbation. The step surface did not translate along the horizontal axis, but produced a similar effect as an actual horizontal translation.

The vertical and horizontal perturbations were intended to be elicited prior to foot contact with the perturbation step. A custom-built laser configuration was therefore used to detect the foot-position at the end of the swing phase of the transition from the second to the fourth step. This enabled the perturbation to be elicited right before the laser was obstructed and disengaged the locking mechanism prior to foot contact with the perturbation step. The trigger mechanism that activates the perturbation was manually setup to each subject and hence the perturbation was elicited a few milliseconds prior to foot contact with the perturbation step.

During the unload protocol, the vertical perturbation mechanism was utilized but with a different trigger mechanism than with the laser. A custom-built force sensitive platform was placed upon the perturbation step. The unload perturbation could therefore be activated exactly at a specific time after the onset of the foot strike with the fourth step.

Noise masking was achieved by activating a secondary set of solenoid actuators (connected to nothing) when the primary solenoid actuator (connected to the locking-mechanism) was not activated; thereby a mechanical sound would occur regardless of a perturbation being induced or not. In addition, subjects wore noise-cancelling headphones (Sony, wh-1000xm4) to mask the possibility of differentiating between the sounds of the primary and the secondary solenoid system.

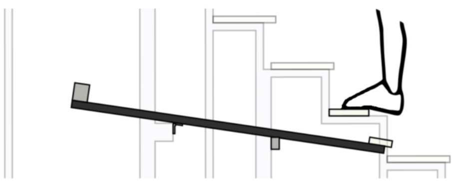

Figure 3. A horizontal perturbation. The perturbation is produced by translating a 100 mm section of step surface downward.

B. Protocols

A comfortable walking speed and step size were determined for each individual subject to imitate/simulate the natural gait pattern of the subject. The floor was marked with four steps prior to the stair apparatus, so that when the subject would start walking with his or her right leg, the fourth step would be approximately at the center of a force plate (OR6-5; Advanced Mechanical Technologies, Westminster, CO). The vertical force measure (Fz) from the force plate was used to detect a foot strike and acted as a trigger for a 10-second data acquisition (DAQ) system. Each sweep consisted of 4 seconds prior and 6 seconds after a trigger event (see Figure 1(a)). After the fourth step the subject began climbing the seven steps of the stair apparatus. The subject stopped after standing with both legs on the platform at the top. After each trial the subject made a turn, walked back to the floor marking, and continued without delay with the next trial. Subjects were allowed to view the stair as they approached, but were instructed to look at a mark on the opposing wall when walking on the stair apparatus. This was done to ensure that the head orientations were consistent throughout the experiment and a perturbation would be indistinguishable visually from control trials.

All subjects were instructed and trained for about 5 minutes to walk on the stair apparatus. Subjects were instructed to touch down on the fourth step (perturbation step) with the left foot. Subjects received enough practice time to complete a sweep without looking at the stair step and maintain a uniform gait speed. Subjects did not practiced with perturbations in order to record initial effects of the imposed perturbation and measure the adaption characteristics of the gait.

1. Protocol: Vertical and horizontal

The vertical and horizontal protocol consisted of four sub-protocols as illustrated in the flowchart 4. All subjects started by completing 20 baseline sweeps, where no perturbation would occur of either type. This baseline session was denoted as the pre-baseline protocol. The purpose of the pre-baseline protocol was to obtain a reference signal before any adaptive mechanisms from the perturbation protocols could influence the gait pattern of the test subject.

Subjects were divided so that half the test population would initially complete 20 randomized vertical perturbations in the first protocol, and subsequently complete 20 randomized horizontal perturbations in the second protocol. The other half would complete the protocols in the opposite order as illustrated in the flow-chart on Figure 4. This protocol order was implemented in order to investigate adaptive strategies from each perturbation type unaffected by a prior adaptive mechanism caused by a specific perturbation session. Finally the session was completed with post-protocol, consisting of 20 baseline sweeps.

During the vertical and horizontal trials perturbations were imposed on random trials (probability of 33% and no two consecutive perturbation were possible). Subjects walked until ≥ 20 successful perturbation-trials were recorded which amounted to a total of around 70 trials per perturbation session. Each perturbation session amounted to around 50 trials with no perturbation. These trials are denoted as vertical-control trials and horizontal-control trials respectively.

During the vertical perturbation session a perturbation trial resulted in a 50 mm downward displacement of the step surface. This perturbation depth was chosen in order to minimize the resulting pitch/slope from the vertical perturbation. Surface slope within 3° was reported undetectable by [2]. The horizontal perturbation session elicited a 100 mm displace of step surface nosing. This horizontal perturbation was chosen in accordance with a previous study of a foot placement analysis.

2. Protocol: Unload

On a random subset of the trials the vertical perturbation was activated during the stand phase. The movement of the perturbation step was initiated at a preset latency after foot strike corresponding to mid stance; 300 ms. This produced the stair equivalent of the unload response in the soleus muscle, i.e., a short-latency depression in the muscle activity following an unloading of the muscle–tendon unit. The perturbations were presented randomly (probability of 25%, and no two consecutive perturbations were possible) to prevent subject anticipation. Data was acquired until 20 trials of each perturbation were recorded.

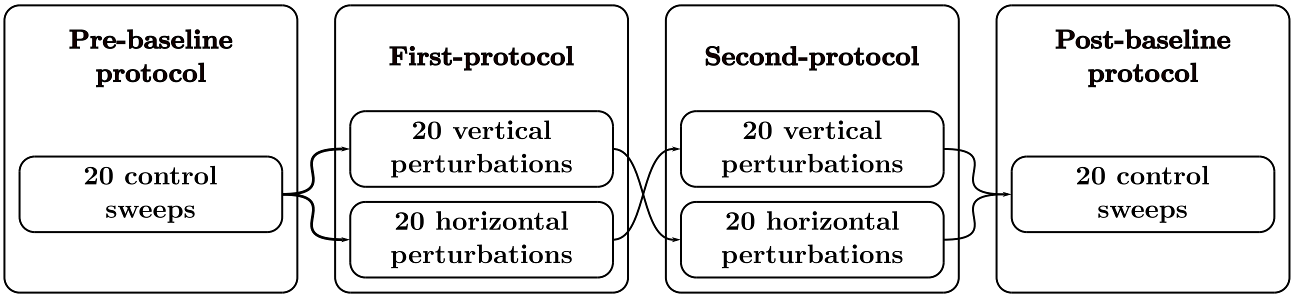

Figure 4. Vertical and horizontal study protocol. The pre-baseline protocol is followed by either a vertical protocol or a horizontal protocol, denoted as the first protocol. The perturbation type during the second protocol is the type not used to the first protocol. The study protocol ends with a post-protocol session.

C. Data analysis

Data analysis was carried out off-line. Individual trials were removed by means of visual inspection of the ankle trajectory if they deviated significantly from the average. The EMG records were rectified and low-pass filtered at 40 Hz (first-order Butterworth filter) to extract an amplitude envelope. The filter order was chosen in order to investigate a transient effect.

Individual records of each sweep were averaged to create a single set of records per condition and subject. The EMG activities were normalized because of high inter-subject stride speed variability (Second step stride duration across all subject; mean 1068 ms $\pm$ 189 ms, outlier 892 ms). EMG data was normalized by dividing soleus and tibialis activity by the average EMG peak of each sweep during the stand-phase of the second step. This was done for each protocol to adjust for inter-subject and intra-subject speed variability between protocols.

1. *Vertical Perturbation Onset*

The vertical perturbation onset (PO) was defined as when the fourth step was tread upon and the perturbation mechanism had moved down to its designated position (50 mm downward displacement). The ensemble average of the ankle trajectory after foot-strike can be seen in Figure 8A for a single representative subject. From this ankle trajectory an initial registration of the foot-strike (FS) can be seen in both control and perturbation data occurring at the same time, however the perturbation onset occurs in this specific subject around 50 ms later. This effect is expected; the force of the subject foot is moving the perturbation step down as intended, until stopped after the 50 mm downward displacement. The travel time causes the delay between vertical perturbation and vertical control trials.

The perturbation onset is marked in Figure 8A as a later local ankle peak in the ankle trajectory. This local ankle peak correlates with the defined perturbation onset. This has been verified in a subsequent test. An FSR sensor was placed on the stair-apparatus stop-block (see Figure 5B). The sensor module could then be used to indicate when the stop-block made contact with the perturbation step at its designated 50 mm position. In

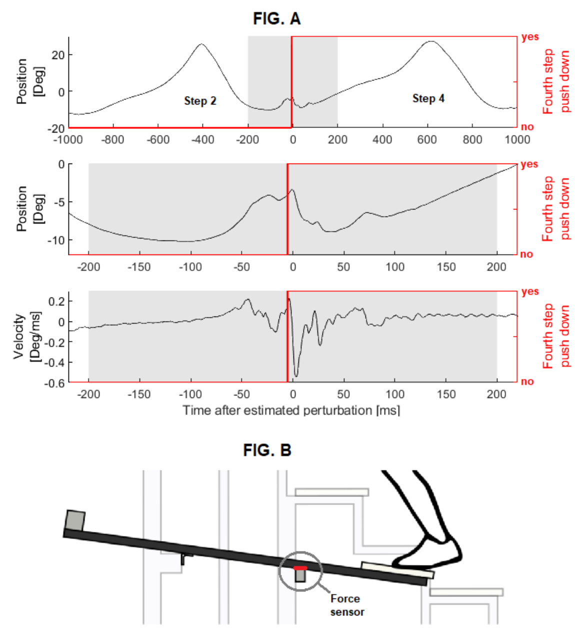

Figure 5. Perturbation onset. [A] Data from single subject (n=15) aligned with an estimation of the perturbation onset (local ankle peak). Top and middle plot; average left ankle position (Dorsalfleksion and Plantarflexion). Bottom plot; left ankle velocity. Red line; fourth step makes contact with the stop block, the actual perturbation onset. Gray shaded region illustrates the same time range across plots. [B] FSR sensor module placement.

Figure 5A an ensemble average of a single subject left ankle trajectory and the average sensor module detection of the perturbation onset is displayed. The data is aligned such that the local ankle peak is time zero. A discrepancy of 6 ms $\pm$ 0.5 ms from the local peak and the actual perturbation onset can be seen in Figure 5A. Furthermore, the perturbation onset is also correlated with the beginning of the largest ankle velocity change in the dorsal direction.

2. *Vertical perturbation*

In order to investigate the effect of the vertical perturbation on locomotor muscle activity, an ensemble average was computed from the time of the perturbation onset.

The outcome of the vertical perturbation resulted in two general forms of ankle trajectories, denoted type 1 and type 2 sweeps (see Figure 7). Type 1 sweeps resulted in a foot strike with step four at approximately the same time as foot strike during vertical control trials. On the contrary, foot strike was delayed in the type 2 sweeps with respect to control data and type 1 sweeps. The stepping speed did not appear altered when comparing to the prior step on the staircase.

This is assumed due to an altered timing of obstructing the laser beam, which initiate the activated perturbation actuators. The type 2 perturbation is assumed not to be delayed in respect to the perturbation onset, but rather an delayed foot strike with the fourth step. This assumption is backed up by the fact that the stepping cycle on the second step was not significantly altered. An earlier obstruction of the laser beam would result in the perturbation step being released earlier by the mechanism. Thus, the step would be lower when the foot strikes it compared to the type 1 sweep.

The type 1 vertical perturbation follows the same ankle trajectory course as presented in Figure 5. The perturbation onset can therefor be estimated in the type 1 perturbation sweeps by aligning with the local peak, denoted PO in Figure 8A.

The local ankle peak is calculated by manually locating in each type 1 perturbation sweep a rough estimation of the PO position. Following, the position at which the ankle velocity data switches to move in the negative is determined, i.e. the ankle begins to stop the plantarflexion movement transition to perform a dorsal-flexion. Similarly the control data is aligned with the local ankle peak around the foot-strike (see Figure 8B).

The type 2 perturbation onset is estimated by aligning the ankle trajectory during the second step's stand-phase with type 1 perturbation. A 30% to 70% window of the ankle-trajectory during stand-phase of step 2 was obtained for each of the type 2 perturbations. This window is subsequently used in a cross-correlation analysis with the average ankle trajectory of the type 1 perturbation. The maximum point of the cross-correlation is used as the estimate of the lag between each type 2 sweep and the average type 1 sweep. This lag is then subtracted from each type 2 sweep to align with the perturbation of the type 1 sweeps. (see Figure 6). The control data is aligned with the type 1 perturbation using the same method (see Figure 8C)

The method for obtaining the ensemble average about the perturbation onset is summarized in general terms in the following:

1. Align data with the foot strike of the fourth stair step by using the data from the left foot-FSR sensor (see Figure 8A)
2. Manually sort perturbation sweeps according to perturbation-type; type 1 and type 2 (see Figure 7)
3. Align type 1 perturbation with local peak corresponding to perturbation onset (see Figure 8B)
4. Use cross-correlation analysis of previous steps to find offset in vertical control and perturbation type 2 data. Align data with perturbation type 1 sweeps. (see Figure 8C)

3. *Horizontal and Unload perturbation*

In order to investigate the effect of the horizontal and Unload perturbation on locomotor muscle activity, an ensemble average was computed from the time of the perturbation onset. The horizontal perturbation onset was defined as when the ankle trajectory begin to deviate from the control sweeps. The Unload perturbation onset was defined 300 ms after foot strike with the stepping surface.

D. Statistical analysis

No transform was found that made the variable normally distributed. Thus, nonparametric Wilcoxon signedrank tests were used to test for differences between groups and the significance levels were Bonferroni corrected. All tests were conducted at a significance level of $P = 0.05$.

II. Results

All subjects' averaged EMGs and ankle positional data from the control trials are shown superimposed on the averaged recordings with perturbation trials. Only trials were the EMGs and positional data matched in the period prior to the perturbation onset were included in the analysis.

A. Vertical perturbation during stair climbing produced an increased soleus activity

Figure 8 shows the responses from a representative subject in whom the vertical perturbation was randomly induced prior to stepping on the fourth step. The subject's walking speed was approximately 0.6 s/step. The EMGs and position data from the ankle joint in each graph were aligned with vertical perturbation trials (thin black lines, n=20) and perturbation onset, as defined in section 1. In trials where a vertical perturbation was imposed, the zero time mark indicates that the perturbation mechanism has moved to its designated position (50 mm downward), and a dorsal ankle flexion was induced. The vertical-control trials (thick grey lines, n=59) are shown superimposed, where the prior ankle trajectory is aligned with the vertical-perturbation trials, as described in section I C 1. The red region marks the standing phase of the fourth step as measured from vertical-perturbation trials. This region did not reflect the standing phase of the superimposed vertical control trials.

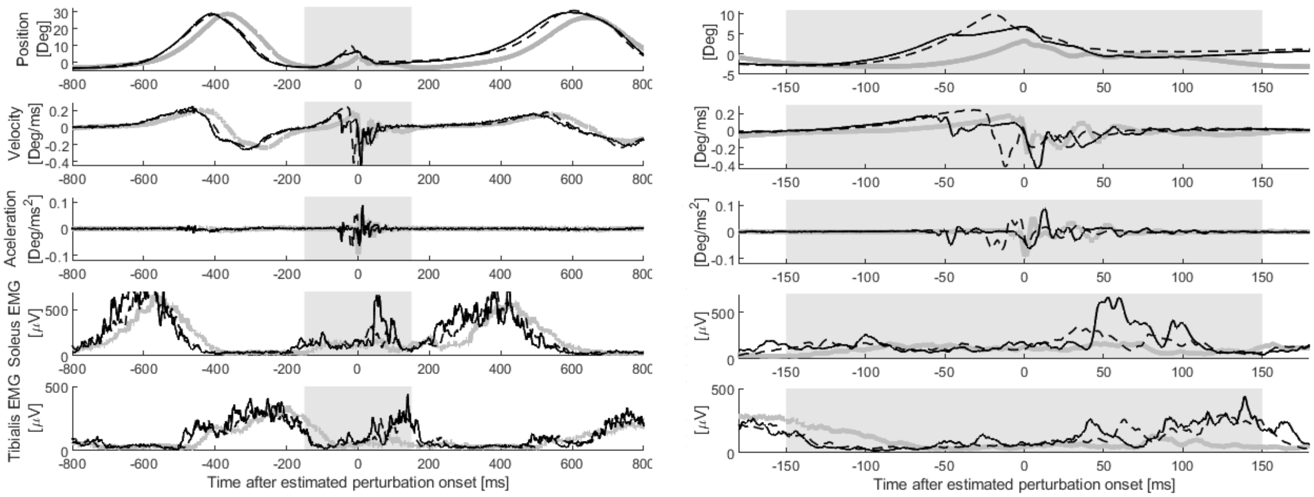

Figure 6. Plots of average angular position, angular velocity and angular acceleration of the angle as well as average EMG results from soleus and tibiallis for a single test subject. Left hand: Full test, right hand: zoom in at perturbation onset. Black line: type 1 vertical perturbation; black stippled line: type 2 vertical perturbation; grey line: control. Grey patch is the same time-period across all plots.

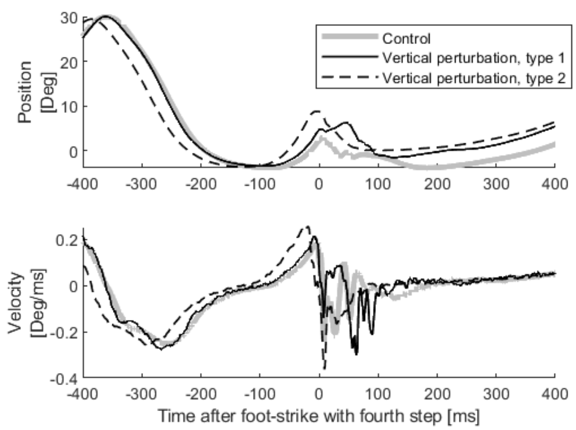

Figure 7. Vertical perturbation types. Data from single subject aligned with foot-strike of fourth step. Top plot; average left ankle angular position (Dorsalfleksion and Plantarflexion). Bottom plot; left ankle angular velocity.

The imposed stretch was reflected as a sudden dorsalflexion, which peaked at 0.5 $^{\circ}/ms^2$. All recorded data before the foot strike matched the recorded data from vertical control. A small depression in the soleus and tibialis anterior EMG could be observed both with an onset around -12 ms. Six out of nine subjects displayed a decreased amplitude in soleus and tibialis anterior EMG activity before the perturbation onset compared to the vertical control trials.

After the imposed perturbation a distinct soleus burst could be observed at approximately 36 ms with a peak activity at 44 ms. Seven out of nine subjects exhibited a soleus burst following the perturbation, with a mean estimated onset around 42 $\pm$ 5 ms. Similarly, a burst in tibialis anterior can be observed in Figure 8C with an onset latency of 37 ms. Six out of nine subject recordings resulted in a distinct EMG burst in tibialis anterior with an onset 45 $\pm$ 14 ms. Only two test subjects demonstrated the two types of perturbations described in section I C 1. The data from one of these subjects is displayed in Figure 6. The small soleus burst in type 2 vertical perturbation can be observed occurring 21 ms prior to the type 1 vertical perturbation.

1. Velocity dependency of soleus response during vertical perturbation

A analysis was carried out investigating the correlation between the soleus EMG amplitude and the ankle angular positional data. The Soleus EMG amplitude data was normalized, and the background muscle activity was subtracted using the control trial data. The average amplitude for a window from 40 to 70 ms after the perturbation onset was used as dependent factors. For the independent factors based on the positional data a window from -15 ms to 15 ms relative to the perturbation onset was used. The average positional data was then subtracted based on the average angular velocity calculated using finite difference methods on the positional data. The a potential correlation found was between the average soleus EMG and the average angular velocity, which is shown in Figure Figure 10, where each color of the data points corresponds to a single test subject. Data were only extracted from type 1 vertical perturbation. A linear model was fitted to the data points giving the linear relation $y = 7.57\,x$, where y and x is the average soleus EMG and the average angular velocity respectively. The

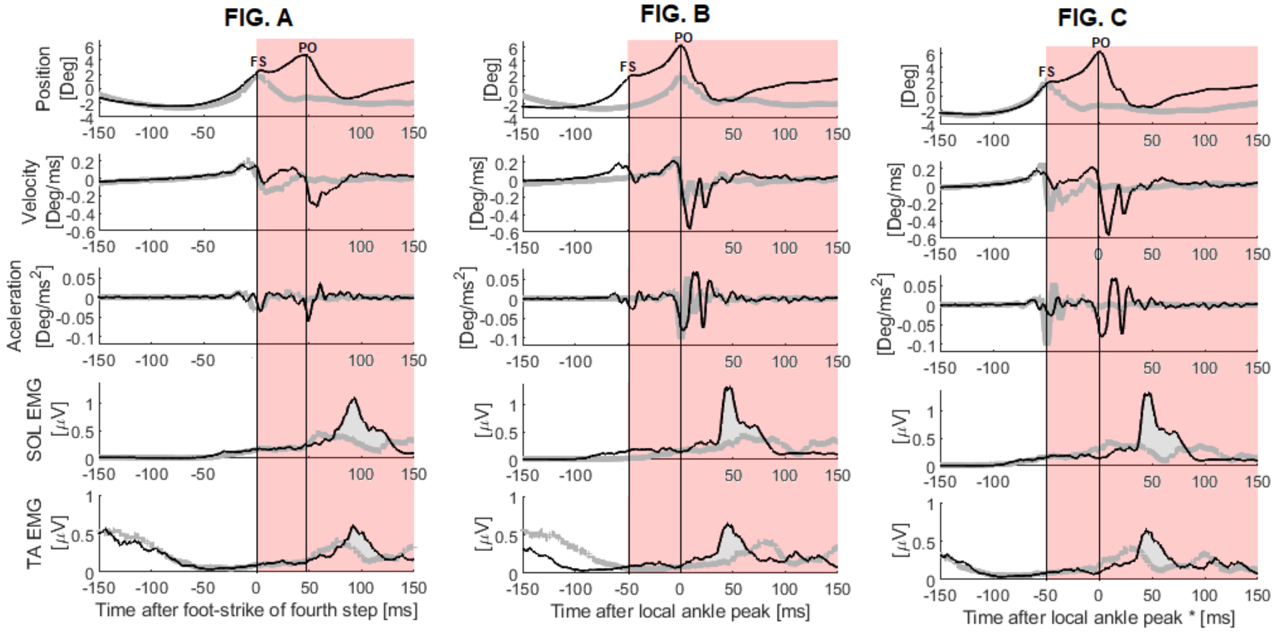

Figure 8. Overall step in the process for obtaining the ensemble average round the perturbation onset (PO) for the vertical perturbation. Data from single representative subject. **Black-line** average perturbation trials. (n=20). **Gray-line** average control data (n=40). **Red-region** stand phase.

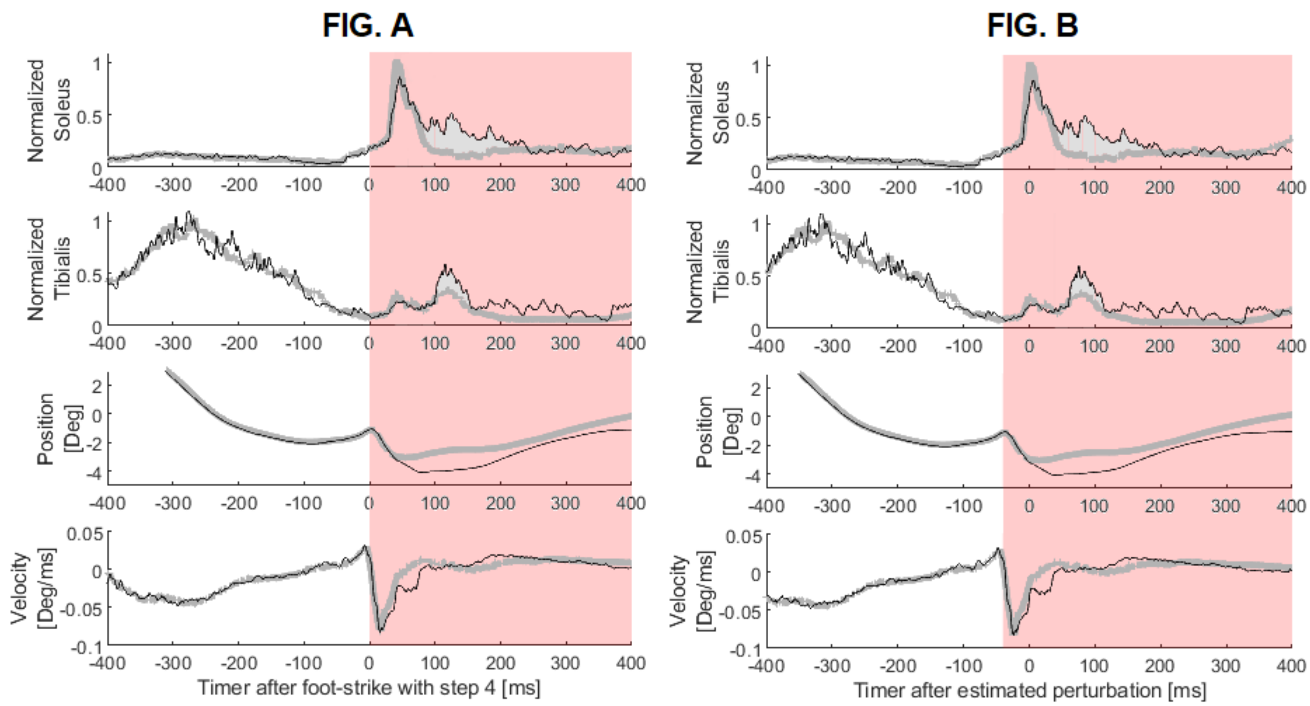

Figure 9. Overall process for obtaining the ensemble average round the perturbation onset (PO) for the horizontal perturbation (0 deg equals standing position). Data in figure A is aligned with foot strike of the fourth step. Data in figure b is aligned with the estimated perturbation onset.

R-squared value of the linear regression is 12.3%, meaning 12.3% of the relationship can be explained by the linear equation.

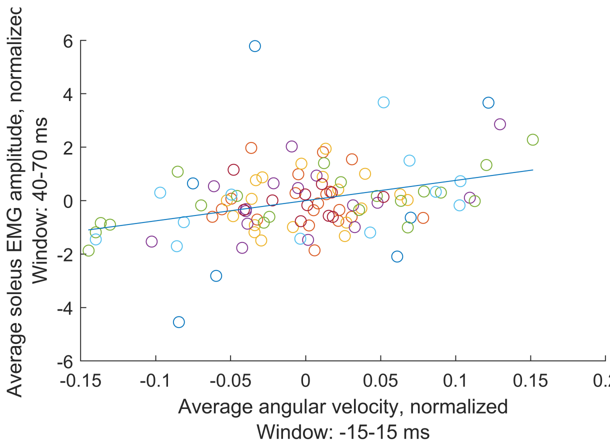

Figure 10. Scatter plot and linear regression for average soleus EMG amplitude and average angular velocity. Each color of the data points corresponds to a separate test subject.

B. Horizontal perturbation caused increased soleus activity

Figure 9B shows the responses from one subject where the horizontal perturbation was randomly induced prior to stepping on the fourth step. At time zero the ankle trajectory of horizontal perturbation trials begins to deviate from horizontal control data. The data acquisition as obtained when aligned with the fourth step is illustrated in Figure 9A.

In each graph, the normalized muscle activities, ankle joint position, and velocity recording in the control situation (thick grey lines, n=59) and perturbation (thin black lines, n=20) are shown superimposed. The red region indicates the approximate stand phase of the four steps as measured from foot strike to lift off.

The horizontal perturbation caused an increase in the dorsal flexion of the ankle. All subjects, except one, experience an increased dorsal flexion. The subject who did not display any increase in dorsal flexion following the perturbation was noted to be toe walking and excluded from further data analysis. All kinematic and EMG prior to the horizontal perturbation matched the horizontal control. Approximately 40 ms after perturbation onset, a marked increase in the soleus EMG activity was present in 4 out of 8 subjects.

Similarly, an increase in the tibialis anterior EMG acitivity was present around 60 ms after the perturbation onset. Not all subjects had this tibialis anterior burst, only 2 out of 8 subjects exhibited this muscle response.

The magnitude of the EMG response between the horizontal perturbation and control trials were investigated, the effect of which sensory afferents contribute to the observed EMG facilitation. The area under the curve of the EMG recordings was calculated in two windows: The

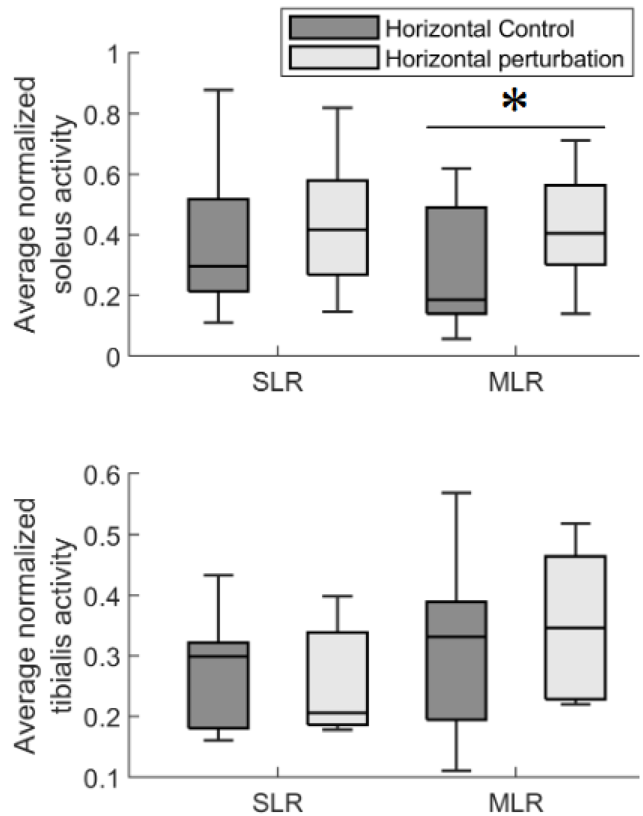

Figure 11. Box plots for average, normalized soleus and tibialis activity, both SLR and MLR, comparing horizontal control and perturbed trails. Only average normalized soleus MLR show statistical significant difference.

windows were defined as 39-59 ms and 60-80 ms. These windows are similarly defined by an article by Thompson et al. [40], and the 39-59 ms window represents the short-latency response (SLR onset 39 ± 2 ms [16]) and the medium latency response in 60 - 80 ms (MLR response peak around 78 ± 6 ms.Grey et al. [16]). The Wilcoxon signed-rank test was used to compare the magnitude of the response between the defined windows. The analysis revealed that the soleus MLR window in the horizontal perturbation was significantly [$p < 0.05$] larger than the horizontal control trials. (See Figure 11). No significance was found for any other conditions.

C. Unload experiment during stair climbing

Four subjects completed the unload protocol before the protocol was terminated due to after mechanical problems, which hindered further investigations. Four subjects completed the protocol; one completed 200ms and 400ms unload perturbation, and the remaining three completed a 300ms unload perturbation.

Only in 1 of 4 subjects did the perturbation step drop produce soleus activity that deviated from the control EMG within the 35 to 70 ms window following unload perturbation. This unload-perturbation was elicited 400

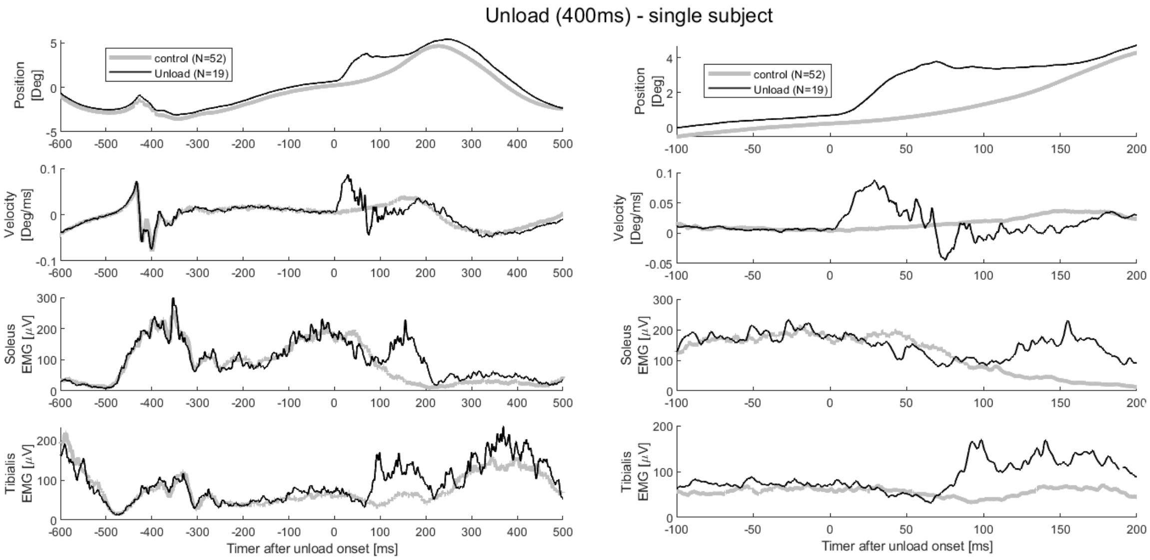

Figure 12. Ensemble average data for single test subject in unload protocol. Plots of angular position, angular velocity, soleus EMG, and tibialis EMG data. Left hand: full data set; right hand: zoom in near perturbation onset. Grey line: control trials; black line: unload trials.

ms after the foot strike with the fourth stair step. Across all subjects, a change in ankle trajectory started 66 $\pm$ ms after the perturbation onset. Only in one subject did depression in soleus EMG follow after unload-perturbation, as shown in Figure 12. The subject was walking at about 0.57 step/s and the traces of the perturbed trials (thin black lines) are shown superimposed with the unload-control trace (thick gray line). The platform was rapidly lowered in this subject at a present latency of 432 ms, when including the mechanical delay measured from foot-strike. A reduction of the SOL EMG amplitude was observed in a single subject, which was afterwards followed by a strong facilitation. (See figure 12, shaded areas).

The onset of the perturbation step movement was determined by a secondary test, in order to measure the mechanical delay after the programmed trigger event. The system displayed a consistent mechanical delay of 32 ms $\pm$ 3. Soleus EMG activity began to deviation from control trial around 24 ms and more sharply depression at around 37 ms. No soleus EMG amplitude decrease were observed at unload perturbation times 200 and 300 ms. However, a strong facilitation of the tibialis anterior EMG activity was observed in all subjects with onset of 66ms $\pm$ 13 ms.

III. Discussion

The aim of this study was to better understand the spinal networks involvement during more complex locomotor tasks than level walking and to elicit a perturbation induced by the body's own weight, in order to reflect natural responses rather than artificially created responses. The responses to the perturbations applied in this study are the consequences of many kinds of sensory input, some of which are probably inhibitory and some excitatory on the motor neurones of the investigated muscles.

A. Methodological limitations

One important issue in the experiments presented in this paper is the unknown effect of walking pace. To better mimic the natural gait of the test subjects, the participants were allowed themselves to choose a moderate pace. This naturally led to a variation in walking pace between participants and sweeps. The walking pace can primarily influence the vertical-perturbation trials, where an altered walking pace can change the perturbation onset due to an altered trigger time of the laser beam and a change in the position of the fourth step (as illustrated in Figure 7).

During an early pilot experiment, a metronome was used to help control the walking pace of the participants. Test subjects were instructed to attempt to synchronize their gait with the clicking frequency of the metronome. All participants in this pilot experiment described the attempt of synchronization as very distracting and it leading to an unnatural walk, especially during the transition from walking on flat ground to ascending the staircase. Thus, no further attempts to control the walking pace of the participants was made. Several other studies [1, 3], examining sensory feedback during locomotion, have similarly chosen to let the test-subject walk with no pace control.

The subject's head position was stabilized by instructing the subject to focus on a point on the opposite wall. Several subjects found not looking at their feet during the stairs climb to be very abnormal; Especially the transition from level walking to stair climbing was very difficult. The subjects focused on the opposite wall in order to exclude visual clues about the perturbation; therefore, they were allowed to look at their gait but focus on the wall marking before treading the perturbation step. An attempt to stabilize the head also had the purpose to rule out that the balance organ in the ear affected the result. A head mounted accelerometer might have been able to rule out the balancing organ for further investigation.

IV. Vertical perturbation

The vertical perturbation is presumed to elicit a stretch reflex of the ankle exterior as a consequence of a sudden dorsal flexion of the ankle trajectory. This dorsal flexion is caused by stopping the downward displacement after moving 50 mm. Seven out of nine subjects exhibited a soleus burst following the defined perturbation onset with a group mean latency of 42 $\pm$ 5 ms. Similarly, soleus stretch latency responses were observed by Sinkjaer et al. [34] during walking with an onset latency of 42 $\pm$ 3 ms. Contrary to the data by Sinkjaer et al. [34], no facilitation of the tibialis anterior activity was observed following the dorsal perturbation. In the present study, a tibialis anterior burst was observable in six out of nine subjects, at an onset latency of 45 $\pm$ 14 ms. When the ankle joint was dorsally perturbated during walking, the tibialis anterior muscle should be depressed due to reciprocal inhibition when the agonist was activated [9, 10]. In studies where a treadmill acceleration was imposed that caused a dorsal flexion, it did result in a facilitation of the tibialis anterior, but at a later onset at 65-75 ms. A larger population size should have been present if something more affirmative can be concluded.

The stretch elicited by the vertical perturbation was due to the subject's own weight suddenly causing dorsiflexion when the perturbation step was stopped at its designated position (50 mm down). This design caused a systemic difference in the ankle trajectory obtained from two of the subjects. These subjects displayed type 1 and type 2 perturbation, as defined in section I C 2. The onset of the soleus burst did differ between the two perturbation types in both subjects. Therefore it cannot be ruled out that the initial foot strike (FS) in type 1 perturbation caused the soleus burst instead of the dorsiflexion. If no difference in the soleus burst was present, then the defined perturbation onset (PO) might be validated as being the correct perturbation onset. However as the foot strike noted at time -50 ms in type 1 perturbation occurred prior to the defined perturbation onset, it cannot be ruled out as the origin of the soleus burst. If so, the resulting soleus burst cannot be excluded from being of a cortical origin due to transmission time of 84 ms. In a future study, a positional sensor mounted on the stair step can shed light on this issue and be used to define the exact perturbation onset.

Concerning the previously discussed origin of the soleus activity, a significant relationship was found between the ankle speed at the defined perturbation onset and soleus burst at a later 40 to 70 ms. According to Grey et al. [16], Houk and Rymer [17] the perturbation should theoretically facilitate the velocity sensitive Ia receptor. Therefore linear regression might suggest that the resulting facilitation in soleus amplitude is induced by dorsal perturbation. The linear regression could thereby suggest that the EMG burst are mediated from a spinal origin.

A cause of interest is the mapping of the reflex feedback pattern during non-standard walking locomotive activities and its potential regarding rehabilitation for patients with locomotor challenges. It has been shown by Wolpaw [42], that a protocol that bases reward on the size of reflex response can induce a change in spinal reflex pathways. This protocol is known as Operant Conditioning and can create a multi-site plasticity in the central nervous system. The multi-site plasticity is explained by a negotiated equilibrium model, where all behaviors in the behavioral repertoire are influenced by the plasticity of the spinal network and by the plasticity in the brain. Thus, the conditioning of a behavior will have a cascade-like effect on all behavioral patterns of the subject as the system as a whole is pushed towards a new negotiated equilibrium point. The more natural reflex response induced on the stair-case could potentially be superior to the mechanical perturbations described in Wolpaw [42]. Thus, the rehabilitation of the reflex response trained on the stair-case could likely have a beneficial effect on the entire behavioral repertoire according to the negotiated equilibrium model. The protocol will perhaps be challenging to implement and standardize with people with walking difficulties. Therefore the system setup should be simplified if this idea is to be pursued.

V. Horizontal perturbation during stair climbing

The intention of the horizontal perturbation was to induce a sudden stretch of the ankle exterior, by removing the outer section of the support surface. No significant difference was obtained when comparing the horizontal perturbation and control trials in a window from 39-59 ms. However a significant difference was obtained when comparing a window from 60-80 ms in the soleus muscle activity between horizontal perturbation and control trials. This provide an indication that a spinal reflex was present in the soleus response. The reflexes during walking are highly regulated and peaked during the stance phase, assisting maintaining a upright position of the body. The reflexes are depressed during the swing phase when they would oppose ankle flexion. This has been demonstrated by the reflex excitability of the H-reflex [31] and the mechanically elicited stretch reflex [34].

An article by Lorentzen et al. [22] were they investigated the sensory feedback during toe walking. In one of their protocol, the supporting ground was suddenly dropped right after ground contact. Their intention was to induce a unload responses of the of the soleus muscle activity. This can be indirectly compared which this study, where dorsal flexion was induced right after foot contact by the test subject's own weight. The 8 cm drop by Lorentzen et al. [22], must produce a dorsal flexion when the foot collides with the floor after the drop. The soleus EMG activity remained unaltered until around 120 ms after the drop of the platform. This is in contrast to the observed responses of this study, where a significant difference in soleus EMG where observed that could be of spinal origin. No spike where observed Lorentzen et al. [22] vertical force data when the foot collides with the ground.

Only 4 out of 8 subjects had a distinct soleus burst after 40 ms. This might be caused by the variance in the subjects' gait pattern, e.g., one subject was noted to be toe walking, and, therefor was unaffected by the perturbation. Similarly, the center of gravity could potentially explain the differences in observed motor responses. If the subject's center of gravity was in front of the left foot instead of to behind, it would result in different torque values around the ankle. Furthermore, the foot placement would also change the properties of the induced stretch. This problem could be overcome by either a simultaneous video analysis of the foot position or a laser array that could indicate the foot's placement within a desired margin of error.

VI. Unload Perturbation during Stair Climbing

In contrast to previous studies where the ankle extensor was unloaded during level walking [1, 3, 18, 35, 36] or toe-walking Lorentzen et al. [22], this study aimed to elicit a unload response during stair climbing. Ten subjects were intended to complete the unloading of the ankle extensor during stair climbing. Only four subjects completed the unload protocol due to mechanical issues with the test setup.

The drop in muscle activity following an unload perturbation have showed an onset latency that was too long for the velocity sensitive Ia afferent and too short for the length-sensitive group II afferents to contribute to the initial depression. The load-sensitive group Ib afferent was assumed to mediate the initial unload effect af Klint et al. [3]. The unload responses onset is defined when a marked decreases in vertical forces are estimated. It was outside the scope of this study to incorporate a force sensitive sensor that could sample the vertical forces during the fourth step of the staircase (the perturbation step). In another article, where an unload response was induced during level walking, the perturbation onset caused a change in the ankle trajectory at 12 ms to 18 ms after the drop in vertical forces [3]. An accelerometer was used to measure the mechanical delay and it is assumed that the vertical forces decreased as a consequence of the downward motion. The ankle trajectory began to deviate from control data around 13 ms in this study, which is within the same range as observed by Af Klint et al. [1], af Klint et al. [3].

One subject did display a decrease in the soleus amplitude following the perturbation onset. The estimated depression occurred after 37 ms, thereby have too short a latency to explain the depression onset as mediated by load-sensitive afferent, and only the velocity-sensitive Ia afferent could have mediate the observed response. This would not be consistent with the conclusions derived from Af Klint et al. [1], af Klint et al. [3]. If the result can be replicated in more test subjects, then an ischaemic nerve block can be induced to investigate the influence of velocity-sensitive Ia contribution to the EMG response [36].

A noticeable difference between level walking and stair ascent is the movement direction of the ankle during the stride phases [21]. When level walking, the angle will naturally, during the mid-stand phase, perform dorsiflexion in order to create a forward sway of the body. Oppositely, the mid-stand phase of stair ascent is characterized by the ankle performing a plantarflexion. The soleus muscle activity will have increased amplitude in both locomotor activities during mid-stance. The reciprocal inhibition would probably not be the resulting factor causing no depression in the stair climbing trials as observed. A more significant moment of force will affect the ankle during stairs ascent compared to level walking [28]. It has been shown that unload responses can be observed during toe-walking [22]. Therefore, the missing response in this study cannot be directly coherent with the increased moment of the force around the ankle nor due to increased co-contraction.

The unloading perturbation of the ankle exterior was followed in all subjects by long latency facilitation in the tibialis anterior with an onset latency of 67 $\pm$8 ms, similarly to the result by Sinkjær et al. [36], where an external mechanically plantar-flexion perturbation was applied to the ankle. The result demonstrated long facilitation of tibialis— a distinct burst in the tibialis anterior EMG with an onset latency of 75 ms. The tibialis burst found by Sinkjær et al. [36] was assumed to be a transcortical reflex loop Petersen et al. [29], Sinkjær et al. [36]. Interestingly, no increase in tibialis anterior was reported during unloading perturbation on level walking with a platform drop Af Klint et al. [1], af Klint et al. [3].

Currently, this study has left more problems unanswered than addressed. It is the belief that a lot of valuable information in the future can be extracted, but the mechanical setup and the protocol needs further development.

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