**KU LEUVEN**

**GROEP BIOMEDISCHE WETENSCHAPPEN**

FACULTEIT BEWEGINGS- EN REVALIDATIEWETENSCHAPPEN

# Medio-lateral stability during walking controlled by response of stance and swing leg gluteus medius

door Inti Vanmechelen

masterproef aangeboden tot het behalen van de graad van Master of Science in de lichamelijke opvoeding en de bewegingswetenschappen

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prof. dr. I. Jonkers, promotor

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Opgesteld volgens de richtlijnen van *Journal of Neurophysiology*.

**Woord vooraf**

Graag wil ik in de eerste plaats Maarten Afschrift bedanken voor zijn medewerking in het tot stand komen van dit eindwerk. Zonder zijn bijdrage en heldere, geduldige antwoorden op al mijn vragen zou dit resultaat niet mogelijk geweest zijn. Ook wil ik graag prof. Ilse Jonkers en prof. Friedl De Groote bedanken voor hun constructieve feedback met betrekking tot inhoud en vormgeving van het resultaat. Tevens zou ik graag de KULeuven bedanken voor de opportuniteit dit onderzoek uit te voeren. Wat betreft taalgebruik wil ik graag Lieze Hardy, Kalp Patel en Fiona Ritchie bedanken voor hun feedback betreffende grammaticale correcties. Daarnaast zou ik graag mijn ouders bedanken, die me de voorbije jaren doorheen mijn Universitaire curriculum onvoorwaardelijk gesteund hebben en me alle kansen hebben aangeboden om een studierichting naar keuze te volgen.

Genk, 17/05/2017 I.V.

**Algemene situering**

In dit onderzoek wordt er gekeken naar de respons van de gluteus medius spier wanneer subjecten verstoord worden tijdens het wandelen. Onderzoek in dit domein kadert binnen een ruimer doel om meer inzicht te verwerven in de manier waarop spieren reageren na evenwichtsverlies. Aangezien een toename van spieractiviteit een belangrijke manier is om het evenwicht te herstellen, kan een verhoogd inzicht in de werking van de spiercontrole een belangrijke factor zijn in het onderzoek naar het herstellen van het evenwicht. Evenwichtscontrole –of het gebrek daaraan- is niet zelden een oorzaak van vallen, vooral in de oudere populatie.

Spieractiviteit gaat tevens samen met voetplaatsing (Hof et al. 2010, Hof and Duysens, 2013) en er is reeds aangetoond dat subjecten verschillende strategieën kunnen toepassen om het evenwicht te herstellen. De strategieën die gebruikt worden tijdens staan zijn daarbij niet noodzakelijk dezelfde als de strategieën die toegepast worden tijdens het wandelen. In wandelen is de voetplaatsing na een perturbatie een belangrijke factor om te bepalen welke strategie door een bepaalde proefpersoon wordt toegepast. Voetplaatsing is tevens een belangrijke indicator voor het activeren van bepaalde spieren, aangezien eerder al werd aangetoond dat spieractiviteit een sterke predictor is voor voetplaatsing (Rankin, Buffo and Dean, 2014). Deze kennis kan bijdragen tot het creëren van een draagvlak voor het toepassen van strategieën voor het herstellen van het evenwicht in combinatie met specifieke spieractiviteit.

Verder hebben we in deze studie onderzocht of het mogelijk is om de spieractiviteit van de gluteus medius spier te reconstrueren op basis van informatie van het massacentrum van het lichaam. Een verplaatsing van het massacentrum en voetplaatsing kunnen waardevolle informatie bieden indien deze afwijken van wat men van een gezond subject zou verwachten. In dat geval kan het traject van het massacentrum meer inzicht bieden in welke spier geactiveerd wordt, of welke spier vermeden wordt in het geval van spierzwakte. Zo kunnen inzichten bijvoorbeeld gebruikt worden in de ontwikkeling van biomimetische controle van een exokselet om evenwicht te ondersteunen bij spierzwakte. Deze controle kan op basis van de center of mass kinematica de gluteus medius assisteren bij het controleren van het evenwicht bij mensen met spierzwakte.

**Referenties**

Hof, A. L. and Duysens, J. (2013) ‘Responses of human hip abductor muscles to lateral balance perturbations during walking’, *Experimental Brain Research*, 230(3), pp. 301–310. doi: 10.1007/s00221-013-3655-5.
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**Medio-lateral stability during walking controlled by response of stance and swing leg gluteus medius**

**Abstract**

The gluteus medius muscle has been shown to be an important contributor to medio-lateral stability during walking. How this muscle responds after a perturbation and how this response affects the consequent steps is currently investigated poorly in the literature. The main aim was to evaluate the role of stance and swing leg gluteus medius activity in medio-lateral balance control during walking. We hypothesize that feedback of center of mass position and velocity controls the response of stance and swing leg gluteus to the perturbation. Furthermore, a feedback model was created to reconstruct the muscle activity based on center of mass information (CoM) and musculo-tendon complex length (LMT) information. 18 subjects [age: 21 ± 2 STD] were placed on a treadmill and perturbations were applied in 7.5, 22.5, 37.5 and 52.5% of the gait cycle. Kinematics, foot position and EMG of the major lower limb muscles were collected, information on the location and velocity of the CoM was obtained by CoM and CoP information and changes in position and velocity of the LMT were estimated. This information was used to create feedback gains, where each set of feedback gains was based on i) CoM and ii) LMT information. 8 subjects showed significantly increased gluteus medius responses for the stance leg and 13 subjects for the swing leg. The stance leg gluteus medius muscle showed increased activity in 7.5, 37.5 and 52.5% of the gait cycle. The swing leg gluteus medius muscle was increased in all four perturbation timings. The muscle activity reconstruction based on CoM information was significantly better than the one based on LMT information for the stance leg, but not for the swing leg.

**Introduction**

Since poor balance control is an important precursor to falls or trips, it is crucial to understand the neuromechanics of balance control to improve the prevention of falls. Increasing our knowledge on balance control during walking extends the opportunities to intervene when balance control deteriorates. An appropriate intervention can lead to a reduced number of falls, and especially when dealing with the elderly, a smaller risk of hospitalization.

Balance or postural equilibrium (Selzer *et al.,* 2014) is defined as the control of the body's Center of Mass (CoM) with respect to the Center of Pressure (CoP) within the Base of Support (BoS). During standing, balance is achieved when the downwards projection of the CoM remains within the BoS (i.e. the possible range of the CoP (Winter, 1995)). During dynamic activities, balance is achieved when the position of the extrapolated center of mass, which consists of the position and the velocity of the CoM, remains within the BoS (Hof *et al.*, 2005). Therefore, the mechanisms of balance control can be divided into two parts: moving the CoP within the BoS and adapting the BoS by stepping. Due to the small range of the BoS and the high CoP, the body is inherently unstable (Maki and McIlroy, 1997). The control of the relationship between CoM and BoS in different activities is regulated by reactive and predictive balance control strategies, which are controlled by postural responses (Misiaszek, 2006). Postural responses can be differentiated from reflexes because they are mediated through the brainstem centers, while reflexes are controlled at the level of the spinal cord (Allum *et al.,* 1993). Postural responses are reactions that maintain the body in an upright position by adjusting the muscle tone (Batra *et al.*, 2011). Anticipatory postural responses activate muscles prior to a movement to compensate for the destabilization of the following movement, whereas automatic postural responses are defined as a muscular response following a perturbation (Wollacoot, 2008; Porges, 2009). The latencies for postural responses are longer than those for stretch reflexes, but shorter than those for voluntary movement (Allum *et al*., 1993). Furthermore, reactive balance control strategies can be divided in two distinct classes: the 'fixed-support' strategies and the 'change-in-support' strategies *(*Horak *et al*., 1989). For balance control during standing, two types of 'fixed-support' strategies can be used. The first type, which is also known as the 'ankle-strategy', acts to control imbalances by shifting the center of pressure in a fixed BoS (Maki and McIlroy, 1997). The second type is known as the 'fixed-support' hip-strategy, which activates the hip muscles in an attempt to regain balance by decelerating the body's CoM (Horak *et al*., 1989; Misiaszek, 2006).

When balance is perturbed, the muscles will react in order to prevent the body from falling. First, the alterations in the musculoskeletal system are caused by a perturbation and sensed by the visual, somatosensory and proprioceptive systems. Second, the feedback information is integrated in the nervous system and a signal is sent through the spinal cord to the muscles via the efferent nerve fibers (Moholkar *et al.*, 2009). Several studies (Nashner and McCollum, 1985; Horak and Nashner, 1986; Dietz *et al.*, 1989) suggested that postural responses were not just components of a number of muscles synergies, but a rather complicated process in which the direction of the perturbation is a crucial influencing factor (Carpenter *et al.* 1999). Multiple studies (Horak and Nashner, 1986; Horak *et al.,* 1990; Inglis *et al.*, 1994; Henry *et al.,* 1998) have investigated muscle response latencies after a perturbation in standing. Carpenter et al. distinguished the stretch reflex and balance-correcting responses of the different muscles, whereas Henry et al. (Henry *et al.,* 1998) did not differentiate the muscle responses (Carpenter *et al.,* 1999). They concluded that the 'balance-correcting' responses which stand under voluntary control of the central nervous system appear 120ms after a perturbation and subside 220ms after the disturbance of balance. These findings were in line with previous studies by Allum et al. (Allum *et al.,* 1993, 1994; Allum *et al.*, 1996) in which responses were divided into stretch responses (40-100 & 80-120ms), balance-correcting responses (120-220ms), secondary balance-correcting responses (240-340ms) and stabilizing reactions (350-700ms).

When a perturbation increases from 'weak' to 'strong', the applied strategies to regain balance shift from non-stepping ('fixed-support') strategies to stepping strategies. As a stepping strategy, the 'change-in-support' strategy has a high capability to adequately defend the body against a loss of balance for two reasons. Firstly, the 'change-in-support' strategy implies that the BoS is extended due to a displacement of the foot. This creates more margin for the CoM to displace without exceeding the borders of the BoS. Secondly, the fact that a change in support has taken place, increases the distance between the CoP and the CoM (Misiaszek, 2006). Hof et al., discovered that subjects apply 'fixed-support' strategies and 'stepping strategies' when controlling for imbalance during walking (Hof *et al., 2010).* The neuromotoral system responds rapidly by applying the 'ankle strategy' which is the fastest strategy, but covers a displacement of no more than 2 cm of the CoP. This displacement is accomplished by a medial or lateral movement by the CoP under the foot's sole (Hof and Duysens, 2013). The 'stepping strategy' is slower, but is able to cover a CoP displacement up to 20 cm. Furthermore, the stepping strategy is often used in combination with a shortening of the stance phase to make the corrective step earlier (Hof and Duysens, 2013).

Several researchers (Dietz *et al.*, 1986; Schillings, Van Wezel and Duysens, 1996; Rankin *et al.,* 2014) have obstructed the foot in the swing phase to investigate the muscle responses to the applied perturbation. Rankin et al. applied perturbations to the swing leg and measured the associations between medio-lateral foot placement and several measures of gait kinematics and muscle activity (Rankin *et al.,* 2014). They found that swing-phase gluteus medius activity is related to medio-lateral foot placement in perturbed and unperturbed walking. An increase of gluteus medius activity was associated with a more lateral foot placement and an increase in CoM velocity and acceleration. Interestingly, not only the swing leg showed increased activity, but the contralateral stance leg also showed increased gluteus medius activity in the subsequent stance and swing phase after the initial lateral perturbation. In a similar experiment, the gluteus medius' response to medio-lateral perturbations was modulated during the gait cycle by introducing perturbations in stance (Hof and Duysens, 2013). An increase in hip abductor activity of the swing leg was found when the push to the left was given at 4, 15 and 24% of the stance phase. The gluteus medius' response was mainly found between 100-150ms and 170-250ms with maximal amplitude during early double stance. Contrary to previous studies, no short-latency (less than 100ms) responses were found. The 100-150ms responses resemble the 'primary balance correcting responses' as described by Allum et al. (Allum *et al.*, 1994).

As can be concluded from the cited literature, the gluteus medius' contribution to medio-lateral stability in walking is higher than previously accepted and we might be underestimating its importance in situations deviating from stable gait patterns.

Recently, different studies have shown that muscle responses after a perturbation of standing can be predicted using CoM-kinematics (Lockhart and Ting, 2007; Welch and Ting, 2008; Geyer and Herr, 2010; Song and Geyer, 2013). Lockhart & Ting reproduced muscle responses in cats from CoM-acceleration, velocity and displacement using a single-link inverted pendulum model (Lockhart and Ting, 2007). In a follow-up study on human subjects, Welch & Ting confirmed that muscle responses and CoM kinematics can be predicted from delayed CoM position, velocity and acceleration in perturbed standing (Welch and Ting, 2008). Furthermore, Herr and Geyer showed that human steady-state walking can be simulated using a model including muscle length, velocity and force feedback (Geyer and Herr, 2010). Medio-lateral balance control was achieved by a 3D feedback controller based on CoM feedback and hip angle & angular velocity to control the hip abduction torque (Song and Geyer, 2013). The models used so far have either focused on predicting muscle activity while perturbing standing or predicting Ground Reaction Forces in steady-state walking. A combination of both seems to be the missing part to complete the available information on sensorimotor control of muscles.

**Research question and hypothesis**

Although several studies showed that the gluteus medius muscle is of major importance for medio-lateral balance control during walking, it is still unclear how the balance correcting response of the stance phase gluteus medius is modulated during the stance phase. Furthermore, it is unclear which feedback system drives the response of the stance and swing leg gluteus medius. Therefore, the two main goals of the study are: (1) to describe the phase-dependent response of the lower limb muscles, more particularly the gluteus medius muscle, after a medio-lateral perturbation during different phases of the gait cycle. We hypothesize that the gluteus medius muscle will show a phase-dependent increase after perturbations in different phases of the gait cycle. (2) To predict the muscle activity of the gluteus medius muscle by using a feedback model consistent of feedback gains based on CoM displacement and velocity. Thereafter, the feedback is extended to include the (mono synaptic) feedback from the hip joint angle and angular velocity and the ability of both models to predict the muscle activity will be evaluated. Based on the high correlation between foot placement and gluteus medius muscle activity in the swing leg (Rankin, Buffo and Dean, 2014), we expect the feedback model to be applicable on both stance and swing gluteus medius. We hypothesize that a delayed-feedback model based on CoM information will be able to adequately reconstruct muscle activation of the gluteus medius muscle after a perturbation. We expect that the feedback model based on LMT information will not be able to accurately reconstruct muscle activity, since increased muscle activity after a perturbation is assumed to be mediated by automatic postural responses and not a local reflex.

Since we dispose of joint kinematics and EMG-signals, we can derivate CoM kinematics to investigate if the feedback model as described by Ting (Welch and Ting, 2008) will be applicable on medio-lateral perturbations during walking. Since this model has not yet been applied on perturbed walking, our research will be a crucial contribution to gain further insights in the feedback model and the influence of CoM kinematics on muscle control during walking in unstable circumstances.

**Materials and methods**

*Data collection.* The data was collected on 18 healthy subjects [age: 21 ± 2 STD] without movement disorders. First, the subjects walked for 1 minute on the treadmill without perturbations at 1.1 m/s. Second, the first out of three sessions of 48 perturbations started. The perturbation was applied during the left stance phase when walking at a speed of 1.1 m/s and consisted of a treadmill acceleration or deceleration, or medial and lateral platform movement. The applied perturbation had 3 magnitudes: small, medium and large. A small perturbation in the medio-lateral direction was a platform movement of 2cm, medium 3.5cm and large 4.5cm (figure 1). All results presented are the responses to a platform translation to the left during the left stance phase. The following perturbation was applied when the subjects reached steady state walking (i.e. the standard deviation of the last 5 strides times is below 0.05s). The perturbation was introduced in different phases of the gait cycle: at 7.5%, 22.5%, 37.5% and 52.5% after left heel contact (figure 2). The subjects had 5 minutes rest between each perturbation session. In the following results, we will use the term ‘gait cycle’ for the action between the left heel contact and the next left heel contact.

Electromyography (EMG), motion capture and ground reaction forces were measured during the adaptation and the perturbation sessions. The EMG signals of 8 lower limb muscles in both legs were measured (). The raw EMG data (100 Hz, Noraxon) was filtered with a 50 Hz Notch filter to remove mains power, followed by a 4$^{th}$ order Butterworth bandpass filter between 20 and 400 Hz and a 4$^{th}$ order Butterworth filter which rectified and low pass filtered the signal with a cutoff frequency of 15 Hz to create a linear envelope. Motion capture was measured in Vicon (200Hz), with 36 markers on different anatomical landmarks. Ground reaction forces were measured through 2 different force plates (100 Hz), one under each foot.

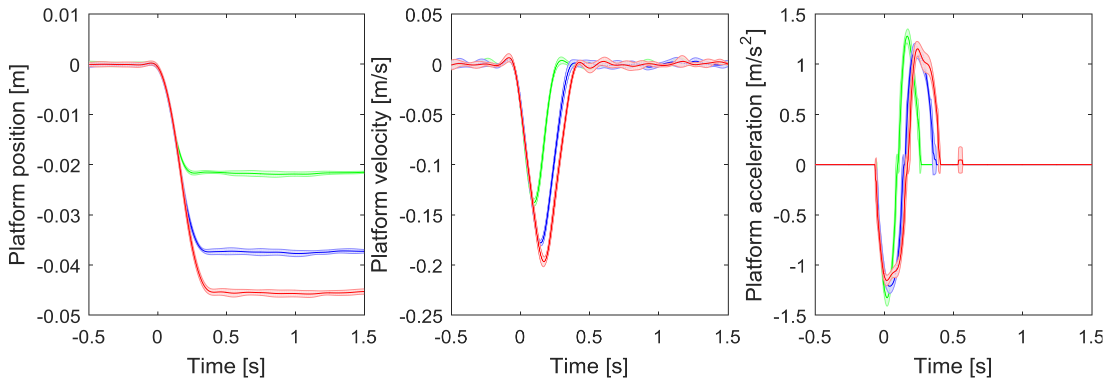



**Figure 1: Platform displacement, velocity and acceleration**

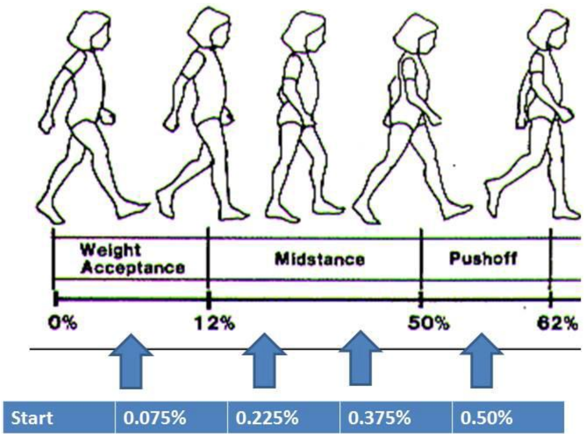



**Figure 2: Perturbation timings**

The experimental data of each individual trial was then analyzed in OpenSim (Delp *et al.*, 2007). First, the generic model (Gait23dof54m) was scaled to the subject-anthropometry. Second, the scaled model was used with measured 3D marker coordinates to compute the joint kinematics using an inverse kinematics procedure (De Groote *et al.*, 2008). Further data processing was performed in MATLAB. The change in length of the musculo-tendon complex was obtained with the Muscle analysis tools in OpenSim.

*Data processing*. To assess foot placement, normal step width was taken as the mean step width during the unperturbed trials. For the perturbed trials, an increase in step width represents more outward foot placement and a decrease in step width agrees with a more inward foot placement. The increase in muscle activity in response to the perturbation with respect to normal walking was calculated by subtracting the mean muscle activity during reference walking from the muscle activity after a perturbation for each stride. The reference EMG-value was taken $EMG_{mean}$+/- 2*$EMG_{STD}$ to avoid including normal EMG activity in the perturbed signal. To generate a time-dependent response activity of the gluteus medius muscle, the difference between the reference EMG data during normal walking and the EMG data during perturbed walking was integrated for three time periods. The chosen time periods were based on previous literature and were as follows: F1 = 0-100ms, f2 = 100-300ms, f3 = 300-800ms. The muscle activity of the stance and swing leg gluteus medius muscle was obtained by calculating the overall mean of all subjects for all trials.

To extract the necessary components for the construction of the feedback model, the feedback gains were calculated from CoM position and velocity feedback (Ting et al., 2007, 2008). The value CoM-Pfoot (foot position) was calculated for the reference walking and perturbed walking and subsequently, the difference of $(COM - Pfoot)_{perturbed}$ and $(COM - Pfoot)_{reference}$ was calculated. To exclude all CoM movement and change in foot position that was not a response to the perturbation, the reference CoM-Pfoot position was taken $(COM - Pfoot)_{mean}$+/- 2*$(COM - Pfoot)_{STD}$. Both CoM and LMT position and velocity information and EMG activity were selected based on the left heel strike timing. These values provide us with the necessary information to calculate the feedback gains based on CoM & Pfoot, CoM velocity and EMG-activity. A neural delay of 100ms was implemented to account for the transmission time of the afferent and efferent information (tau). In agreement with Welch and Ting, feedback gains were calculated using:

$$EMG(t) = |k_1^+ \Delta COM\ (t - \lambda)| + |k_2^- \Delta COM\ (t - \lambda)| + |k_3^+ \Delta C\dot{O}M\ (t - \lambda)| + |k_4^- \Delta C\dot{O}M\ (t - \lambda)| \qquad \text{Eq. 1}$$

Where $\Delta COM$ is the change in CoM position, $\Delta C\dot{O}M$ is the change in CoM velocity, $k_1^+$is the positive position feedback gain, $k_2^-$is the negative position feedback gain, $k_3^+$ is the positive velocity feedback gain, $k_4^-$ is the negative velocity feedback gains and $\lambda$ is the time delay of 100ms. This system of equations (Eq. 1) was solved in a least squares sense for k1, k2, k3 and k4 during the automatic postural response (i.e. first 300 ms after perturbation).

A similar method was used to test if the change in muscle activity in response to the perturbation could be the result of a local reflex. To accept or reject this hypothesis, we investigated the change in length of the musculo-tendon complex of the gluteus medius muscle. As with CoM data, we have LMT position change and LMT velocity change, which enables us to calculate the feedback gains:

$$EMG(t) = |k_1^+ \Delta LMT\ (t - \lambda)| + |k_2^- \Delta LMT\ (t - \lambda)| + |k_3^+ \Delta L\dot{M}T\ (t - \lambda)| + |k_4^- \Delta L\dot{M}T\ (t - \lambda)| \qquad \text{Eq. 2}$$

Where $\Delta LMT$ is the change in LMT position and $\Delta L\dot{M}T$ is the change in LMT velocity. The time delay tau, $\lambda$, in this equation is 50ms, since a reflex is quicker than a postural response.

After calculation of both feedback values, we reconstructed the EMG signal by using the feedback gains i) based on $\Delta COM$ and $\Delta C\dot{O}M$ and ii) based on $\Delta LMT$ and $\Delta L\dot{M}T$. To verify if the signals reproduced by equations 1 and 2 matched the measured EMG signal, the least square error was calculated.

*Statistics and outcome measures.* A paired t-test was used to identify the subjects that showed an increase in gluteus medius activity in f2 for all perturbations. This was done for the left (stance) and right (swing) leg separately. To verify if the increase in muscle activity was dependent on the timing of perturbation, a wilcoxon signed rank test was used on the previously selected subjects.

To assess the accuracy of our feedback model to reconstruct muscle activity, we used the RMS method to compare the amount of error in each reproduced signal. The reconstructed signal was based on a separate set of feedback gains for the 8 subjects that showed increased EMG-activity in the stance leg in the first 300ms after the perturbation. Similarly, a set of feedback gains for the 13 subjects showing increased EMG-activity in the swing leg was calculated. Relative RMS-values were obtained by RMS(LMT)/RMS(CoM) to compensate for inter-individual variation in EMG-values. A signed wilcoxon test was then applied to see if the ratio RMS(LMT)/RMS(CoM) was significantly different from 1.

**Results**

*Stepping strategy*. Figure 3 shows the step width after a perturbation (compared to reference walking) for the different perturbation timings and magnitudes. The right foot is placed more outwards after perturbations at 7.5 and 22.5% of the gait cycle. For perturbations at 37.5 and 52.5%, there is no increased deviation of the right foot. At the first left heel strike after perturbation, the left foot is placed more outwards for perturbations at 7.5 and 22.5% of the gait cycle. For the perturbation at 52.5% of the gait cycle, the cross-over strategy is applied as the left foot is placed more inwards.

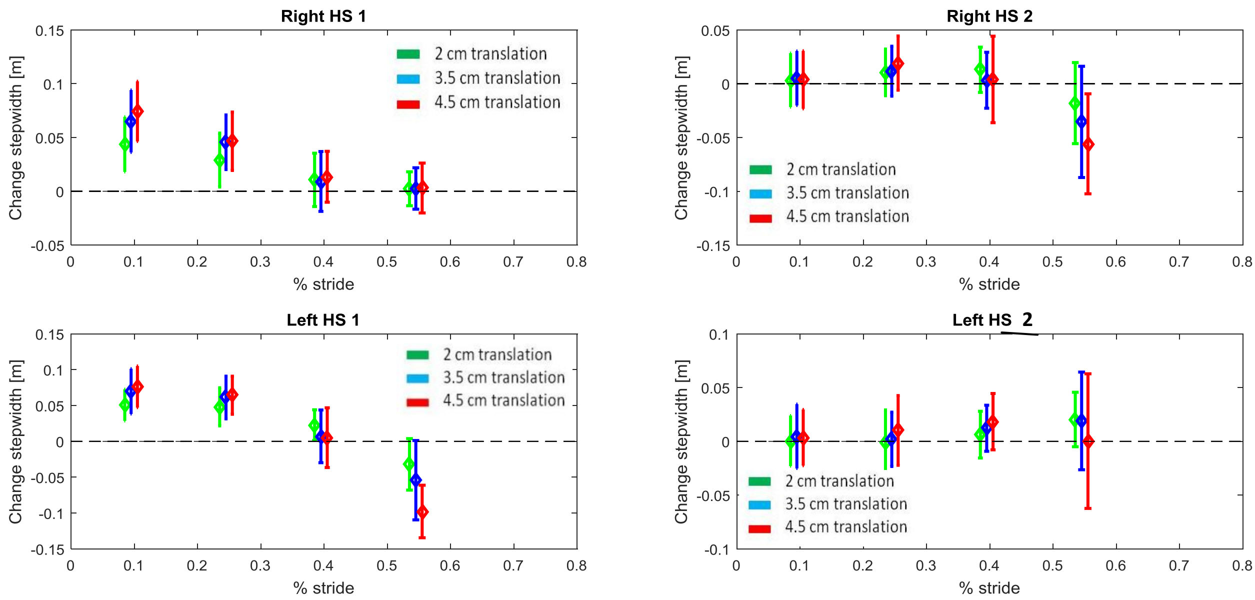



**Figure 3: Step width after the perturbations. Right HS 1 is the first heel strike right after perturbation for perturbations in 7.5, 22.5 and 37.5% of the gait cycle and right before heel contact for the perturbation at 52.5%. Left HS 1 is the first left foot contact after the perturbation. The colors represent the magnitudes of the perturbations, positive values represent a wider step width than normal, negative values a smaller one.**

*Increase of muscle activity*. Different perturbations cause the activity in multiple leg muscles to increase (appendix, figure 1). The significant increases ($p<0.05$) of activity in all measured muscles after a platform translation to the left are shown in table 1. A significant increase in left and right leg gluteus medius activity was found in the second interval, the third interval or both. No increases of activity were recorded during the first interval.

| | Tib L | Gastroc L | Vast L | Rect Fem L | Bic Fem Lat L | Bic Fem Med L | Soleus L | Glut Med L |
|---|---|---|---|---|---|---|---|---|
| 7,50% | f2 ↗ | f1 ↘ f3 ↘ | | | | | f3 ↘ | f2 ↗ f3 ↗ |
| 22,50% | f1 ↘ | f2 ↘ | | | | | f2 ↘ f3 ↘ | |
| 37,50% | f1 ↘ | | f1 ↘ f3 ↘ | | | f1 ↘ f2 ↗ | f1 ↘ f2 ↘ | f2 ↗ |
| 52,50% | | f1 ↘ | | | f3 ↗ | | | f2 ↗ |

| | Tib R | Gastroc R | Vast R | Rect Fem R | Bic Fem Lat R | Bic Fem Med R | Soleus R | Glut Med R |
|---|---|---|---|---|---|---|---|---|
| 7,50% | f1 ↘ f2 ↗ f3 ↘ | f2 ↗ f3 ↗ | f2 ↗ | f2 ↗ | f2 ↗ | f2 ↗ | f1 ↘ f2 ↗ f3 ↗ | f2 ↗ f3 ↗ |
| 22,50% | f1 ↘ f2 ↗ | f2 ↘ | f2 ↗ | f2 ↗ | f2 ↗ | | f1 ↘ f2 ↗ | f2 ↗ |
| 37,50% | f2 ↗ f3 ↗ | f2 ↗ f3 ↗ | f2 ↗ f3 ↗ | f2 ↗ f3 ↗ | f2 ↗ | | f2 ↗ | |
| 52,50% | f1 ↘ f2 ↗ f3 ↗ | f2 ↗ f3 ↗ | f2 ↗ | f2 ↗ | f2 ↗ | f1 ↘ f2 ↗ | f2 ↗ | f3 ↗ |

**Table 1: Increases of muscle activity for all muscles following the different perturbation timings. F1 (0-100ms), f2 (100-300ms) and f3 (300-800ms) are the time intervals in which the increases were measured. The percentages represent the timing of the onset of the perturbation (figure: see appendix).**

*Stance phase gluteus medius activity*. The muscle activity of the gluteus medius was only altered in the first step after perturbation (figure 4). The left stance phase occurs from 0 to 60% of the gait cycle, while the leg is in swing from 60 to 100% of the gait cycle. The increased activity occurs immediately after heel contact for the first perturbation shown in the figure. There is a marked increase in activity in the current step, but minimal increase in the following steps. Figure 5A shows the left gluteus medius activity of all subjects after the four perturbations, which is agreement with the results of table 1.

*Swing phase gluteus medius activity*. The right gluteus medius shows increased activation after a perturbation at 7.5, 22.5 and 52.5% of the gait cycle (figure 5B). At 37.5%, the perturbation is not followed by increased gluteus medius activity (table 1).

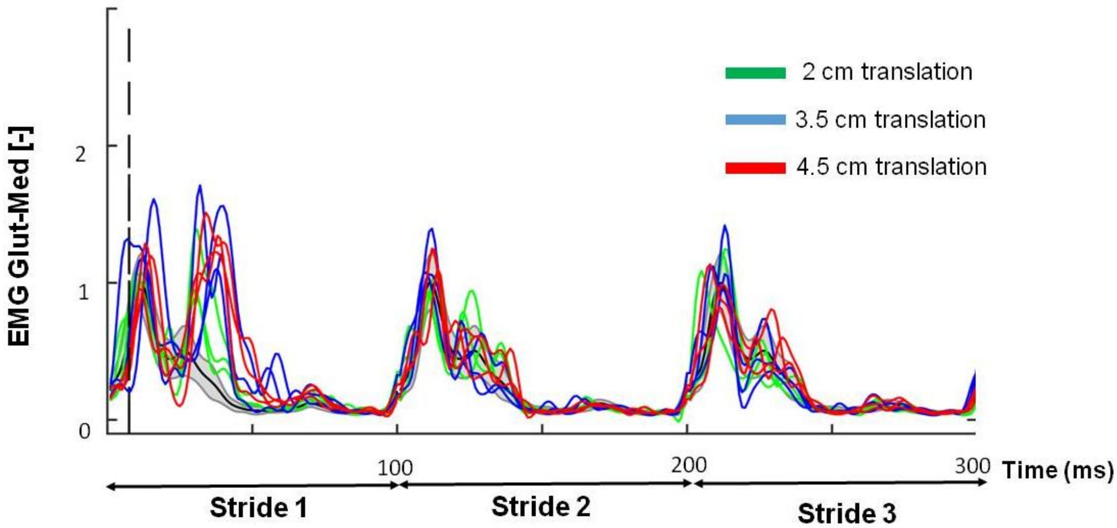



**Figure 4: Increase of the gluteus medius activity of the left stance leg after a perturbation in the first phase of stance for a single subject. The dotted vertical line represents the perturbation timing. The black line and grey area are mean muscle activity and standard deviation during normal walking. The colors represent the perturbation magnitudes.**

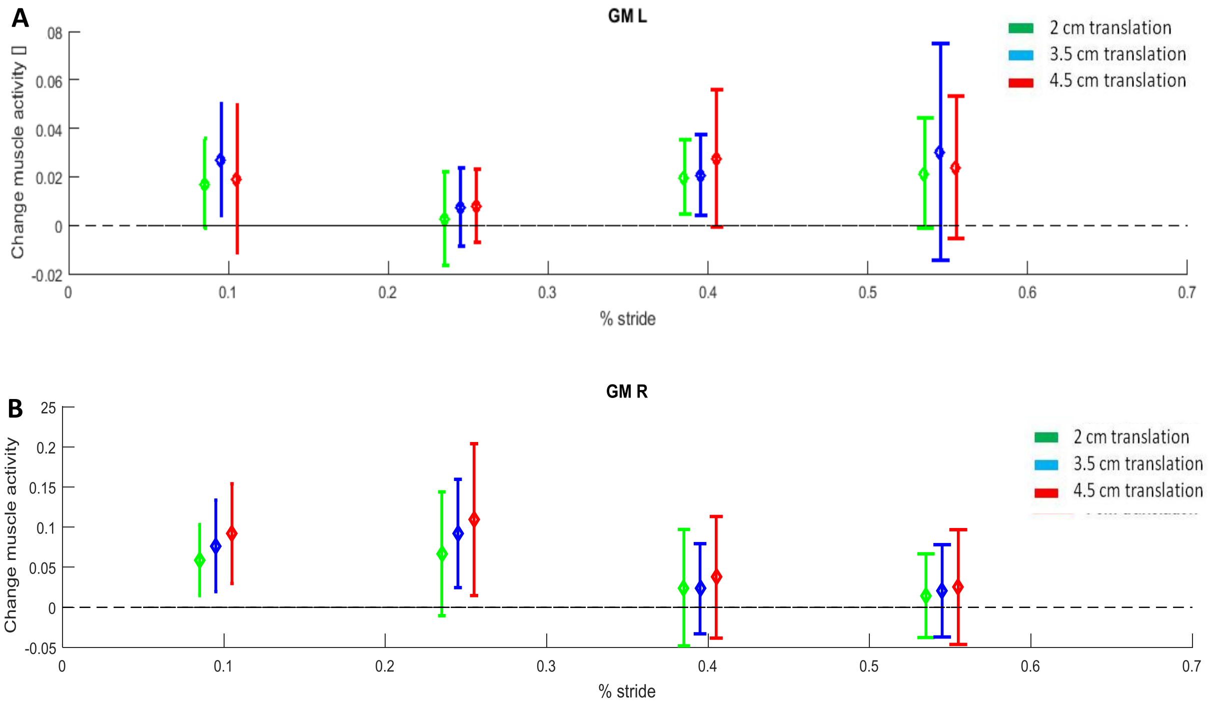
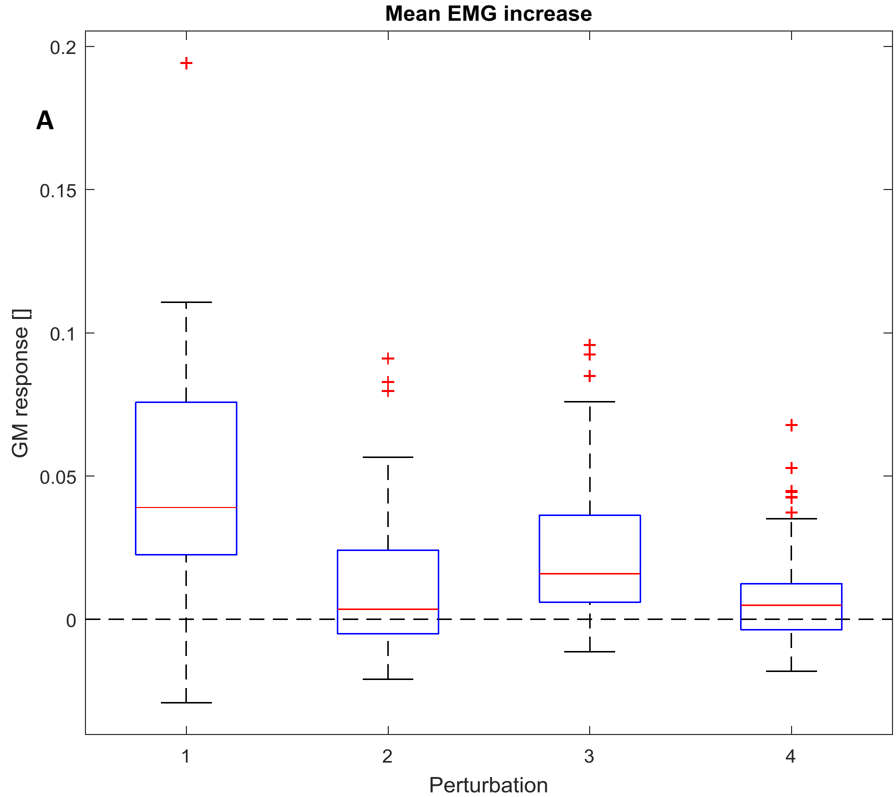



**Figure 5: Increase of muscle activity of the stance (A) and swing (B) leg gluteus medius for a perturbation of the left foot to the left. Data shown are mean increases and standard deviations for all subjects. The four timings shown represent the muscle activity after every perturbation. The colors represent the perturbation magnitudes.**

*Phase-dependent gluteus medius activity*. A significant increase in stance phase gluteus medius was found for the perturbations at 7.5%, 37.5% and 52.5% of the gait cycle for the selected subjects (figure 6A). For the swing leg, gluteus medius activity was significantly increased after all four perturbation timings (figure 6B) for the selected subjects. The figure in appendix (figure 2) shows the response of the right gluteus medius muscle after the 4 perturbation timings (7.5, 22.5, 37.5 and 52.5% of the left gait cycle).

*Feedback model*. Figure 7 shows both the measured and the reconstructed EMG signal in the first phase of the gait cycle (7.5%) for a representative subject. The RMS-values of the reconstruction based on CoM information were significantly smaller (p=0.0078) than the RMS-values based on LMT-information for the stance leg (figure 8A). For the swing leg, there was no significant difference between the muscle activity reconstruction for CoM and LMT information (figure 8B).



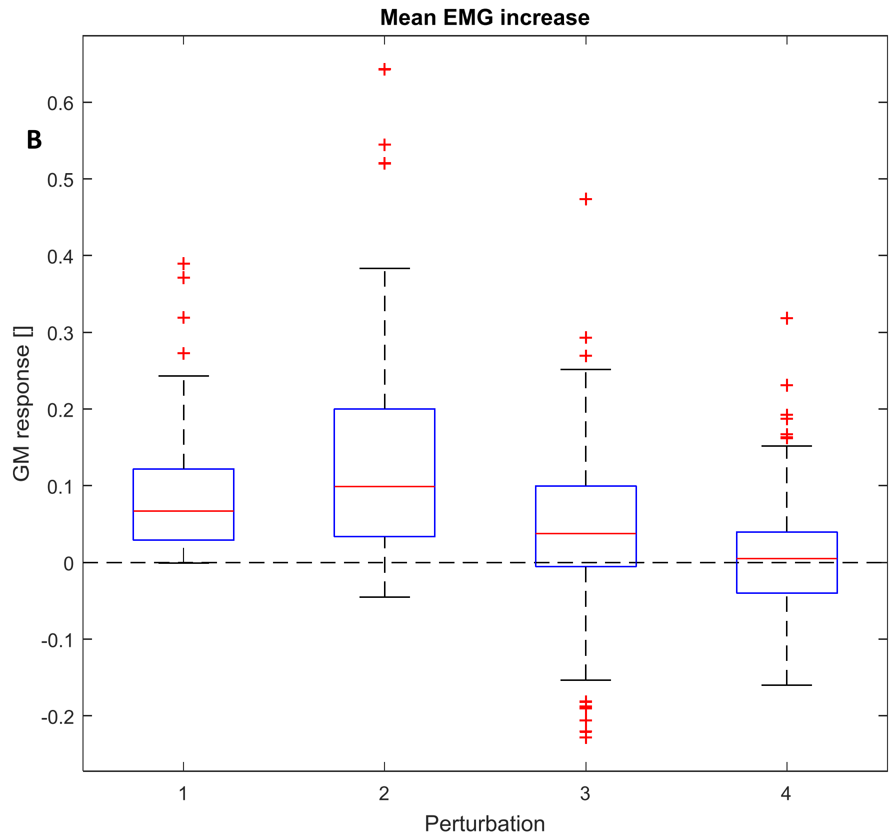




**Figure 6: Phasic response of the left (A) and right (B) gluteus medius activity for the selected subjects. The four events represent the timing of perturbation. The dotted line represents the background activity and the box plots represent means and standard deviations.**

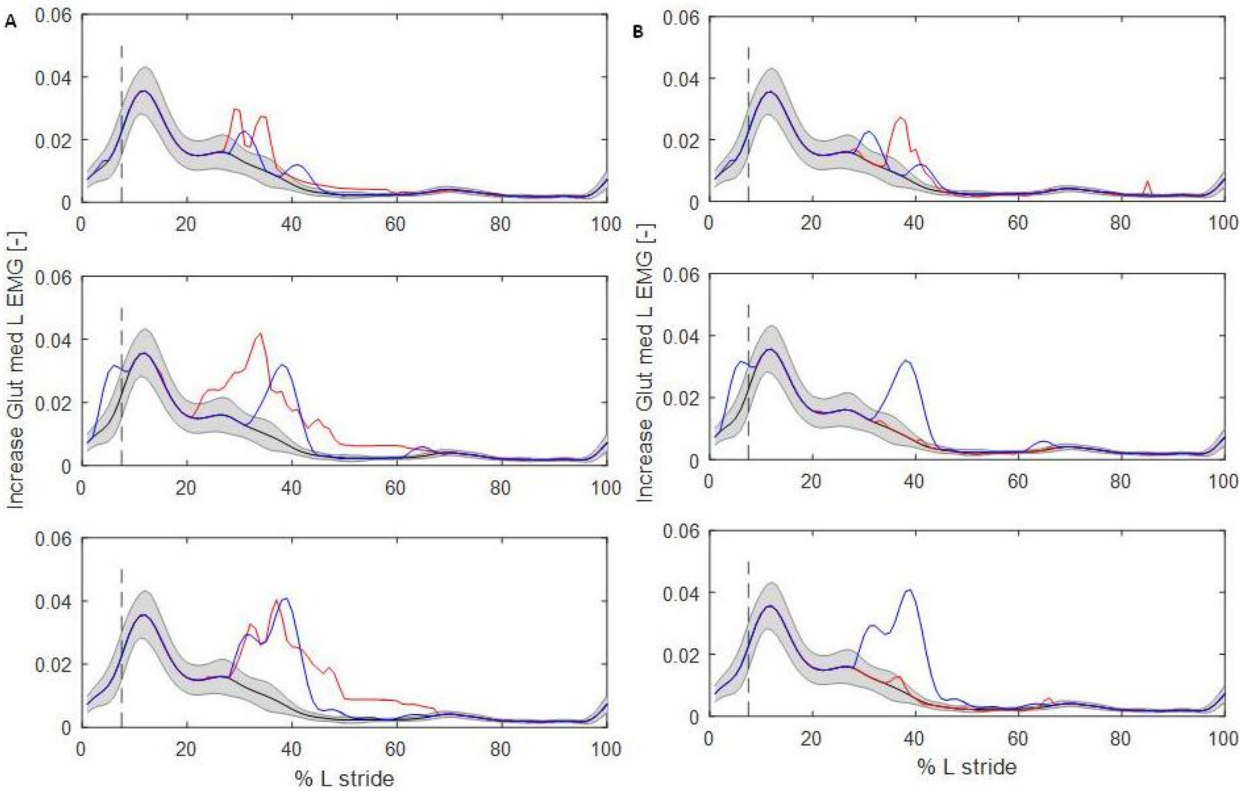
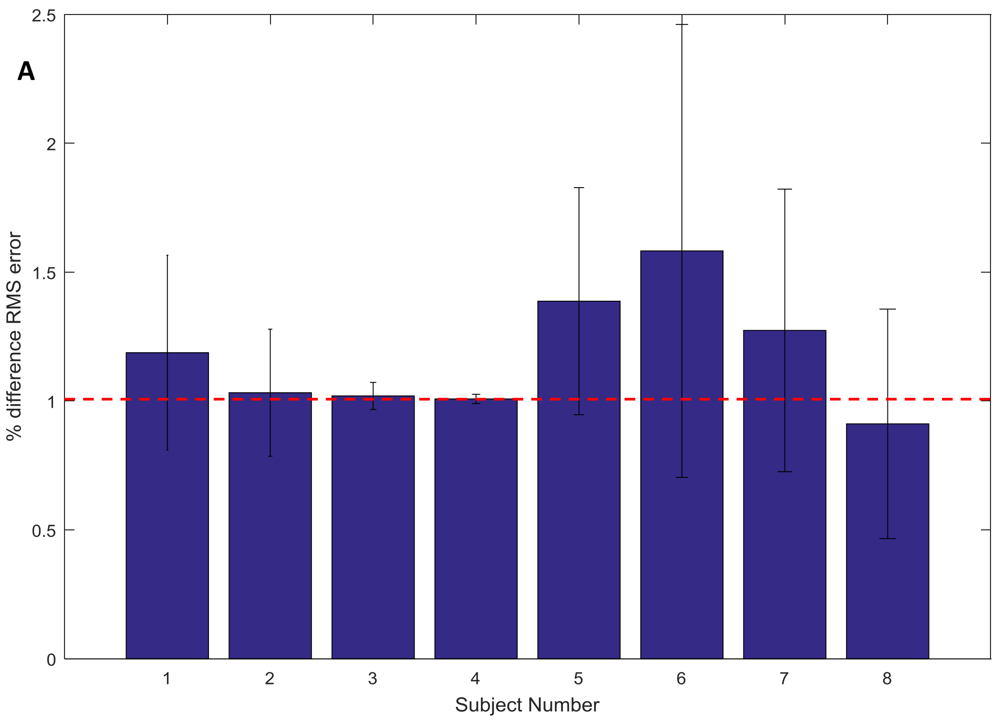



**Figure 7: Reconstruction of the EMG signal based on CoM (A) and LMT (B) feedback at 7.5% of the gait cycle. The blue line is the measured EMG signal, the red line is the reconstructed EMG signal. Mean and standard deviation of the unperturbed trials are grey. Top to bottom: small, medium and large perturbation.**



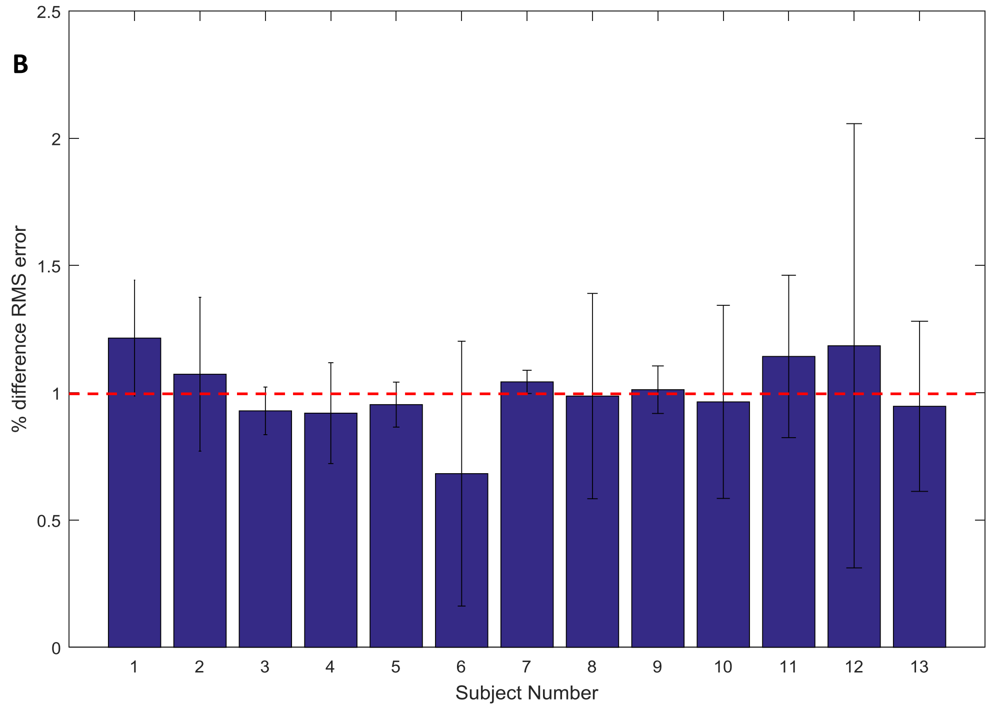



**Figure 8: Relative RMS values and standard deviations for the EMG reconstruction based on CoM and LMT information for all perturbations for the stance (A) and swing (B) leg.**

## Discussion

The main goal of this study was to understand the role of the gluteus medius muscle to control medio-lateral perturbations of gait, by investigating the changes in experimentally recorded EMG activity following a perturbation and its dependency on perturbation timing. Furthermore, we investigated to what extent CoM or LMT feedback could explain the experimental muscle activations using a feedback model.

The results for foot placement after the perturbations showed that subjects indeed use a ‘change-in-support’ strategy as described by Hof (Hof *et al.*, 2010) and Misiaszek (Misiaszek, 2006). For perturbations in 7.5 and 22.5% of the gait cycle, the right foot is placed more outwards to compensate for the loss of balance at the left foot. The subjects therefore extend their base of support, which is a perequisite for more stability (Hof *et al.,* 2010). For a perturbation in 52.5% of the gait cycle, the subjects are in double support when the perturbation is applied on the left foot, so the corrective step will be the next left step. This is confirmed by our results, as the first left heel strike is more medial than usual, in some cases over-crossing the right foot, causing a decrease in the base of support. For the second right heel strike after a perturbation at 52.5% of the gait cycle, the right foot is placed more medially than usual. This could be caused by the previous over-crossing step of the left foot. The right foot is placed at a relatively normal position, but since the left foot is more medial, the step width is smaller and the base of support is narrowed (figure 9).

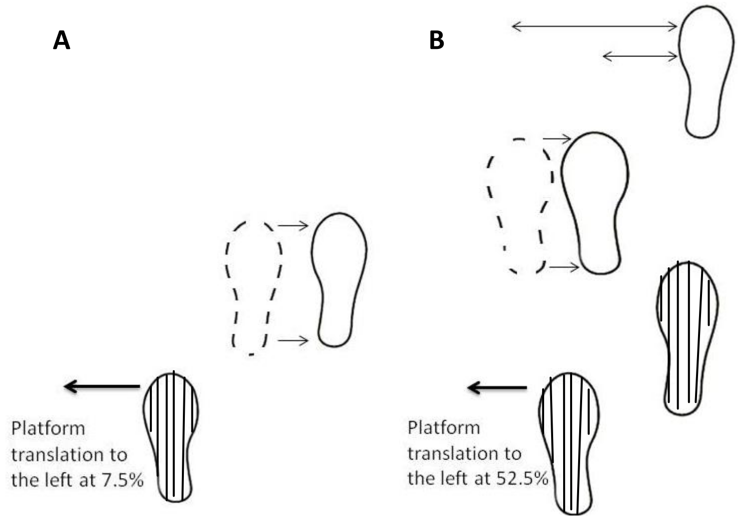



**Figure 9: Foot placement after a perturbation at 7.5% (A) and 52.5% (B). Hatched footsteps are the feet on the ground at the onset of perturbation.**

After a medio-lateral perturbation, muscle activity in the gluteus medius muscle of the stance and swing leg was significantly increased. This increase was present for all perturbation magnitudes but not for all perturbation timings. The subjects showing increased responses in 100-300ms after the perturbation were selected for subsequent analysis. The hypothesis that the gluteus medius muscle would present a phase-dependent increase after perturbations in different phases of the gait cycle was confirmed. In the stance leg, there was a significant increase in gluteus medius muscle activity in 7.5, 37.5 and 52.5% of the gait cycle while no significant increase was found at 22.5%. A possible explanation for the increase at 7.5% is the importance of the gluteus medius muscle for foot clearance of the swing leg in pre-swing, whereas the foot has already passed mid-swing at 22.5% of the gait cycle. For all subjects, a significant increase in swing leg gluteus medius activity was found in response to all perturbations except at 37.5%. Since the right heel strike takes place between 40 and 50% of the gait cycle, the perturbation at 37.5% is applied right before initial contact of the right foot. This would mean that the subjects had approximately 25-125ms to react to the perturbation, an interval, too short to be attributed to postural responses. Furthermore, the platform acceleration was lower in comparison with Hof. et al. (Hof and Duysens, 2013), which could result in a slower sensory detection of the perturbation and therefore a slower response. For the subsequent analysis with the selected subjects, muscle activity was significantly increased for all perturbation timings. This suggests that the increase in step width for the perturbations at 7.5% and 22.5% is mainly regulated by swing phase gluteus medius activity, since the right leg has sufficient time to reposition before the right foot touches the ground. These results are in agreement with Rankin et al. (Rankin *et al.*, 2014) who showed that swing-phase gluteus medius activity is the strongest predictor for medio-lateral foot placement in both unperturbed and perturbed walking.

In line with Hof, short response latencies were only studied to exclude any voluntary activity (Hof and Duysens, 2013). We found no increase in muscle activity in the first 100ms after the perturbation (table 1), which suggests that the observed activity to correct for the balance perturbation is initiated by the central nervous system and that the gluteus medius muscle activity is not part of a reflex response. These findings are in line with previous measurements of Allum (Allum *et al.*, 1993) and Hof (Hof *et al.*, 2010; Hof and Duysens, 2013). The increase of muscle activity in the second and third interval is an indication that the subjects manage their disrupted balance by automatic muscle activation. This activation is part of the balance-correcting response in which the gluteus medius attempts to minimize the loss of balance by adjusting its activation to prevent a fall. These results are in agreement with Hof (Hof and Duysens, 2013) and suggest that the observed balance-correcting responses are part of an automatic and involuntary control of balance.

Medio-lateral perturbations during gait can be compensated through adaptations within one stride. This implies that the subject corrects the imbalance within one step or adopts a change-in-support strategy for the right leg. Indeed, no increase of gluteus medius activity was visible in the second left step after the perturbation. Because the left foot is displaced after left heel contact, the subjects cannot use a 'change-in-support' strategy in their left leg. Because the BoS is already fixed, the only way the subjects can adapt is by shifting their CoP within the BoS, an action that requires activity of the gluteus medius muscle.

This study integrated a feedback model to evaluate the potential feedback signals contributing to the observed responses in terms of gluteus medius activity. It was hypothesized that the feedback model based on CoM information would accurately predict muscle activity of the gluteus medius muscle, whereas LMT information would not be a valuable source to predict muscle activation. Indeed, CoM feedback was able to reconstruct the response of the left gluteus medius muscle more accurately compared to LMT feedback (figure 8A). This was not the case for the swing leg gluteus medius muscle (figure 8B), which was in contradiction with our expectations and the results of Rankin et al. (Rankin *et al.*, 2014). These results suggest that the stance leg gluteus medius muscle relies on automatic postural responses rather than a reflex activity and would therefore be mainly driven based on CoM feedback. These findings suggest that the central nervous system uses information from the CoM to adjust the necessary muscle activation to a perturbation. This feedback model needs more validation, but these preliminary results indicate that it would be possible to predict muscle activity based on CoM information. This would enable us to estimate how much muscle activity is needed based on the kinematic information. Consequently, this model could be applied in rehabilitation by using devices (e.g. exoskeletons) that take over the lacking muscle function to adequately react in destabilizing environments.

*Limitations.* The results of this study need to be evaluated in the context of the following limitations. Since the calculations of the feedback model are derived from the increase of muscle activity after a perturbation, the statistics for validation of the model were based only on the subjects who showed an initial significant increase in muscle activity in the first 300ms after the perturbation. Since our subjects showed variable behaviour in response to the perturbation, the subjects were clustered in to groups, depending on their responses. It is to be expected that the feedback model will not be applicable to subjects who do not show increased activation after a perturbation. These subjects may use alternative methods to regain their balance, not necessarily including gluteus medius activation. Secondly, the calculation of our feedback gains was based on CoM position and velocity, whereas Whelch and Ting (Welch and Ting, 2008) used CoM position, velocity and acceleration. Since our collected acceleration data were qualitatively lower than the position and velocity data, we created the feedback model without acceleration information, which could possibly have influenced our results. However, Rankin et al. (Rankin *et al.,* 2014) showed that the gluteus medius muscle activity in swing was most strongly predicted by CoM position and less by CoM velocity and acceleration. To what extent this hypothesis is applicable to the stance leg is currently unknown. Thirdly, the information on the position and velocity change of the muscle fibers is based on the simulated musculo-tendon length rather than the simulated fiber length. This relies on the assumption that the tendon length is invariant. Therefore the dynamics of the tendon and its effect on the fiber length are ignored. This will influence the feedback information in the model and could possibly influence the results.

*Conclusion*. Overall, control of walking following a medio-lateral perturbation of the stance limb increases both stance and swing leg gluteus medius muscle activity. This increased activity is not part of a reflex, since it is not present in the first 100ms after the perturbation. For the stance leg gluteus medius, muscle activity could be reconstructed using information of the displacement and velocity of the CoM. Both these results imply that humans use postural responses rather than reflex activity to regain balance after a perturbation.

# Appendix

## 1. Populaire samenvatting

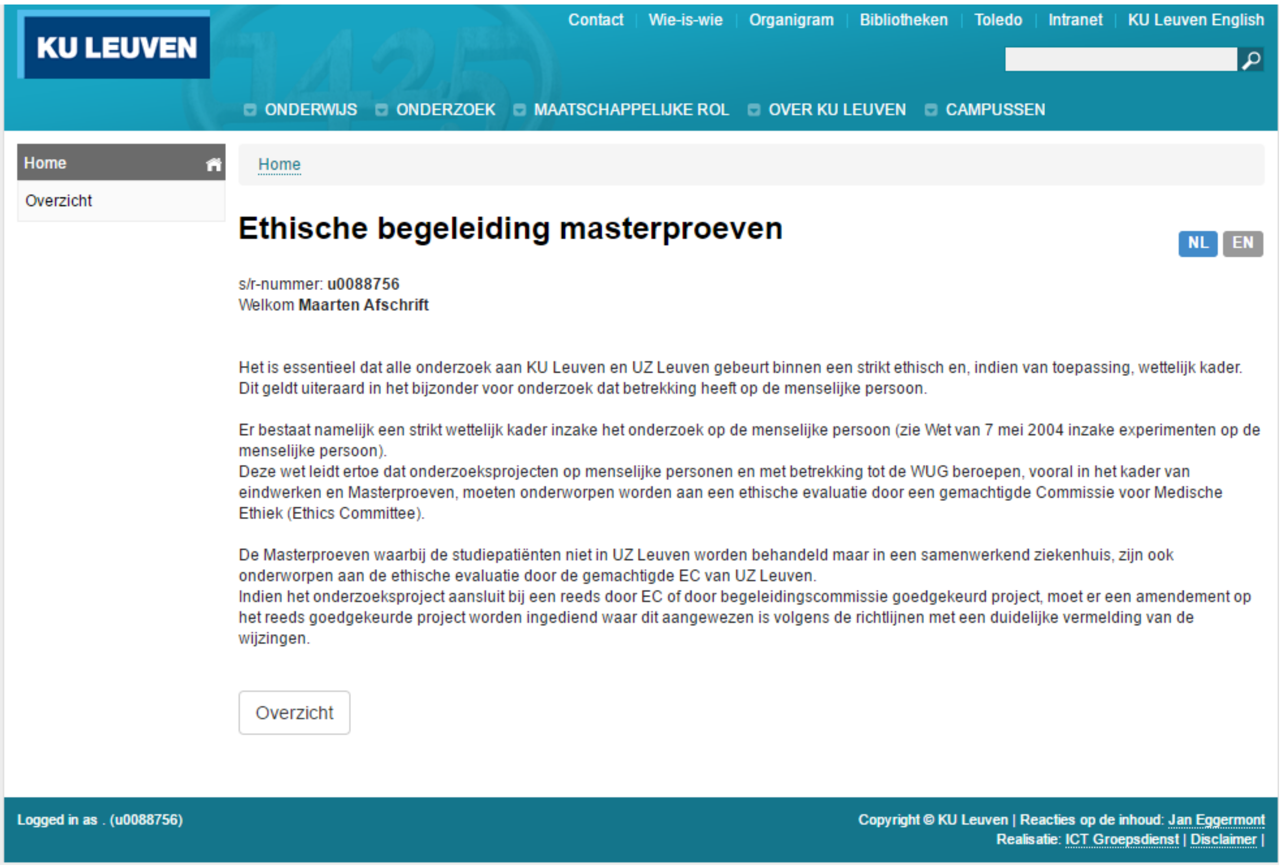
Deze studie focust op de manier waarop de gluteus medius spier reageert wanneer subjecten uit evenwicht gebracht worden tijdens het wandelen door een zijwaartse perturbatie. Onderzoek in dit domein kadert in een groter opzet om meer inzicht te verwerven in de manier waarop spieren reageren in een onstabiele omgeving.

Om dit te onderzoeken zijn 18 subjecten op een loopband geplaatst waarvan de linker- en rechterzijde apart kunnen bewegen. Perturbaties (oftewel een verstoring van het evenwicht) worden veroorzaakt door een verplaatsing van de linkerkant van de loopband naar links (lateraal) of naar rechts (mediaal). In het huidige experiment wordt er gefocust op de respons bij een verplaatsing van de loopband naar links. Deze perturbaties starten op verschillende momenten tijdens de gang cyclus: op 7.5, 22.5, 37.5 en 52.5%. De hypothese dat de activiteit van de gluteus medius spier afhankelijk is van de timing van de perturbatie wordt getoetst. Voorts wordt de spieractiviteit gereconstrueerd op basis van informatie van het massa centrum van het lichaam en veranderingen in het spier-pees complex van de gluteus medius. Er wordt verwacht dat de reconstructie op basis van de informatie van het massa centrum beter zal zijn dan de reconstructie gebaseerd op spier-pees complex informatie.

Tijdens de trials werd informatie over de spieractiviteit, kinematica en voetplaatsing verzameld. Uit deze informatie werd de spieractiviteit van de gluteus medius spier tijdens het normale wandelen berekend, deze dient in de trials met perturbatie als achtergrond activiteit. Verder wordt er getracht de spieractiviteit van de gluteus medius spier na een perturbatie te reconstrueren op basis van 'feedback gains'. Deze 'feedback gains' werden berekend op basis van de positie en snelheid van het massa centrum (CoM) en de verplaatsing en snelheid van de lengte van het spier-pees complex (LMT) van de gluteus medius. Voorts werd er een 'vertraging' (tau) in de formule verwerkt, die de neuronale transmissie van de spier naar de hersenen en vice versa, in rekening neemt. De 'feedback gains' die gebruikt worden in het feedback model zijn gebaseerd op de subjecten die een verhoging van spier activiteit vertonen in 0-300ms na de perturbatie.

Onze resultaten toonden aan dat 8 van de 18 subjecten een significant verhoogde spieractiviteit vertoonden in de linker gluteus medius en 13 van de 18 subjecten een significant verhoogde activiteit vertoonden in de rechter gluteus medius 100-300ms na de perturbatie. Voor de 8 subjecten met een verhoogde spieractiviteit is de gluteus medius significant verhoogd voor perturbaties tijdens 7.5, 37.5 en 52.2% van de gang cyclus. Voor de 13 subjecten met verhoogde spieractiviteit in het zwaaibeen was de gluteus medius actief in alle perturbatie timings. Voor de reconstructie van de spieractiviteit door middel van de feedback gains was de reconstructie gebaseerd op CoM informatie significant beter dan de reconstructie gebaseerd op LMT informatie voor het staan been, maar niet voor het zwaai been. Dit betekent dat het mogelijk zou zijn de spieractiviteit van het staan been te bepalen indien de CoM informatie gekend is, wat in de praktijk gebruikt kan worden om hulpmiddelen te creëren die een deel van de spierfunctie overnemen indien deze door de persoon in kwestie niet meer adequaat gegenereerd kan worden.

## 2. OBC Beslissing



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Welkom **Maarten Afschrift**

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Er bestaat namelijk een strikt wettelijk kader inzake het onderzoek op de menselijke persoon (zie Wet van 7 mei 2004 inzake experimenten op de menselijke persoon).
Deze wet leidt ertoe dat onderzoeksprojecten op menselijke personen en met betrekking tot de WUG beroepen, vooral in het kader van eindwerken en Masterproeven, moeten onderworpen worden aan een ethische evaluatie door een gemachtigde Commissie voor Medische Ethiek (Ethics Committee).

De Masterproeven waarbij de studiepatiënten niet in UZ Leuven worden behandeld maar in een samenwerkend ziekenhuis, zijn ook onderworpen aan de ethische evaluatie door de gemachtigde EC van UZ Leuven.
Indien het onderzoeksproject aansluit bij een reeds door EC of door begeleidingscommissie goedgekeurd project, moet er een amendement op het reeds goedgekeurde project worden ingediend waar dit aangewezen is volgens de richtlijnen met een duidelijke vermelding van de wijzigingen.

Overzicht

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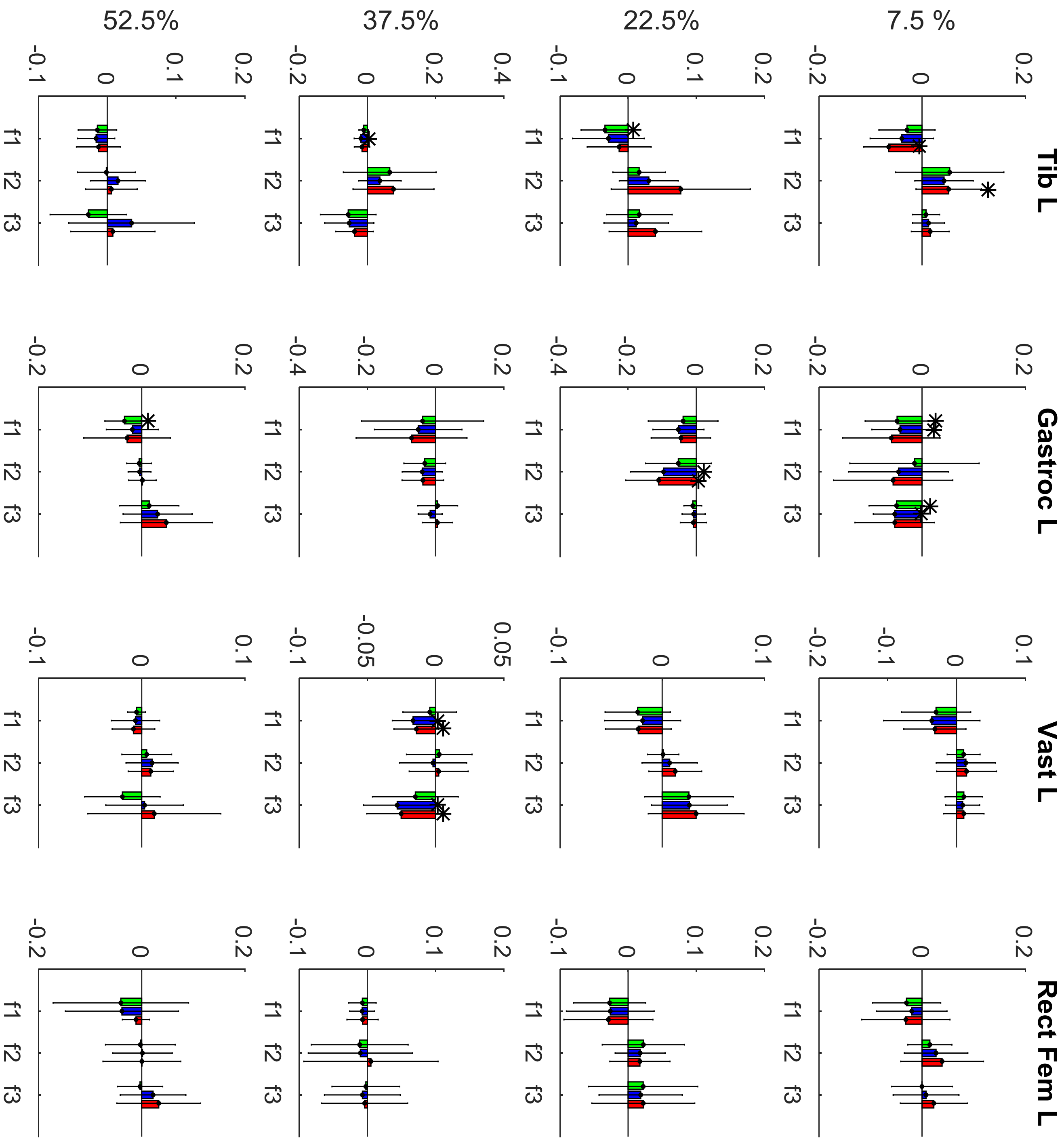
## 4. Text Appendices

### Muscle increases

Figure 1 shows the increases of the different muscles for a perturbation to the left for the left leg.



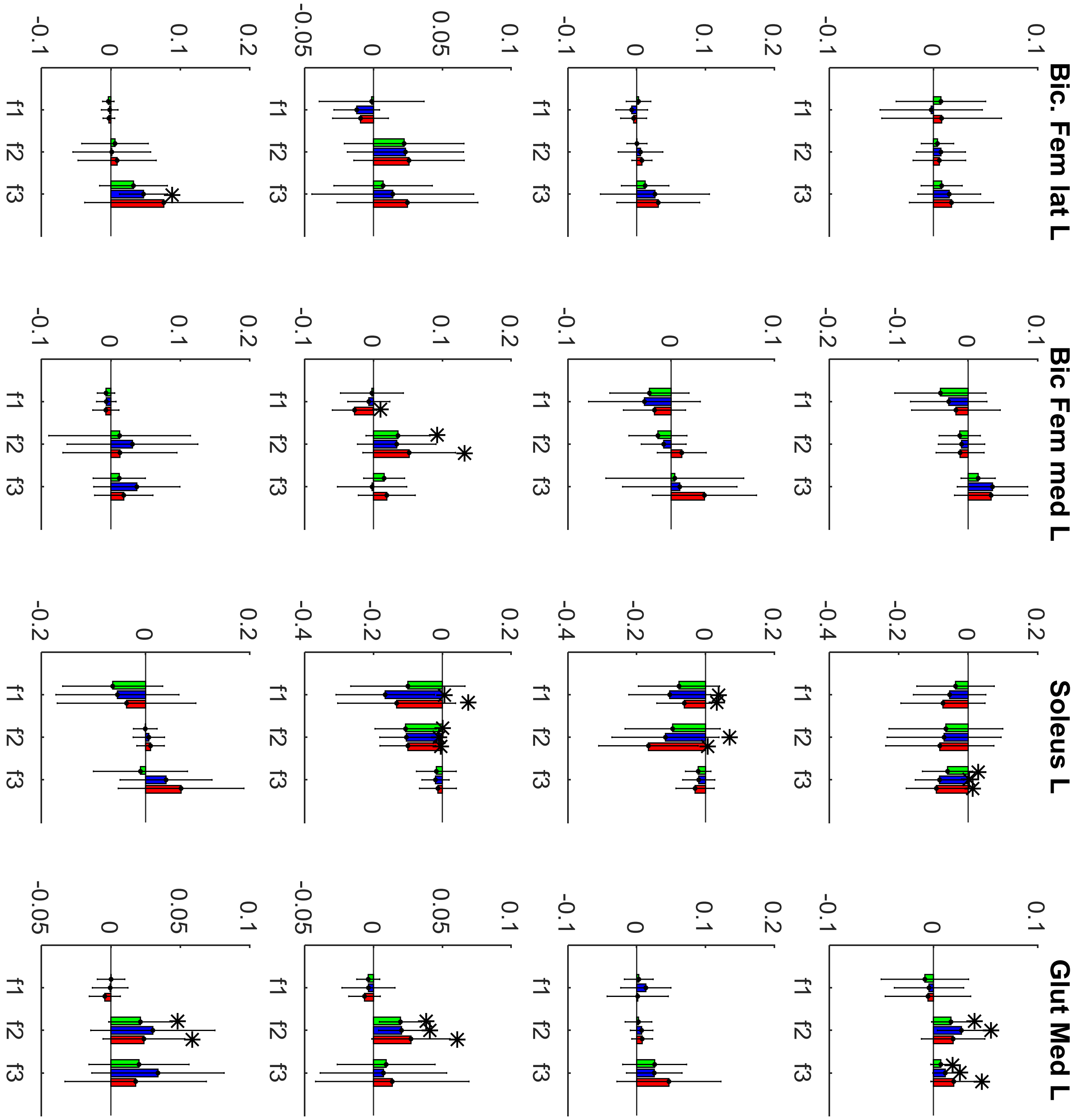



**Figure 1: Increase of muscle activity for the muscles in the left leg. Green is the smallest perturbation, blue the middle one and red the largest.**

**Phasic increase of muscle activity**

The figures below show a comparison of the phasic muscle activity of our results with the one from Hof (Hof and Duysens, 2013). Figure 2A shows the results of Hof, while figure 2B shows our results of the right gluteus medius muscle after the 4 perturbation timings (7.5, 22.5, 37.5 and 52.5% of the left gait cycle). In Hof's experiment, the subjects were pushed at the level of the waist. This means that when the subjects are pushed to the left while the right leg is in stance, the left leg will be the swing leg. In our experiment, a platform translation of the left foot to the left means that the foot is pulled to the left, which will cause the subject to fall to the right, making the right leg the swing leg. The results are similar, showing clearly increased activation between 35 and 55% of the gait cycle and no increase in 65% of the gait cycle.

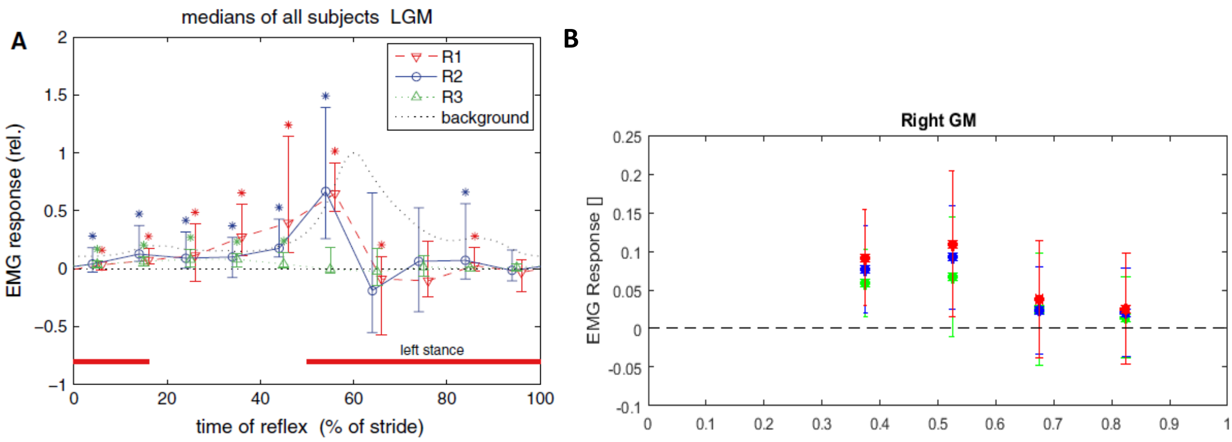



**Figure 2: A) Results of Hof (2013). Pushes are given every 10% of the gait cycle. B) Phasic increase of the swing phase gluteus medius muscle after integration. The colors represent the various perturbation sizes. The x-axis represents the gait cycle as a percentage of 100. The dotted line in the figure represents the normal EMG activity, while activity above this line can be considered increased activity.**