The dynamic response of leg muscles to perturbations in the sagittal plane during stance

Bachelor Thesis

Filip Cengic | 1668072
Department of Biomechanics
Institute of Sports Science
Technical University of Darmstadt

TECHNISCHE
UNIVERSITÄT
DARMSTADT

Filip Cengic
Student-ID: 1668072
Course of studies: Sports and Computer Science

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Submitted: March 14, 2015

Advisor: Dr. Christian Rode

Prof. Dr. André Seyfarth
Institute of Sport Science
Department of Human Science
Technical University Darmstadt
Hochschulstraße 1
64289 Darmstadt
Germany

Solemn Declaration

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Abstract

Designing robotic devices like exoskeletons requires a high amount of adaptational skills. In order to control the actuation of an exoskeleton, the biomechanics of human anatomy and movement need to be fully understood. Because there are many possible combinations of motor unit actions, which produce a desired net moment, it is assumed that the central nervous system uses specific activational patterns. With the results of the experiment, the change in muscle activity can be predicted considering the simple inverted pendulum model. This could be helpful when designing exoskeletons, e.g. the controller of the robot can be fed by input data extracted from experimental results. Based on this knowledge robotic movement can be adapted to sudden perturbations during stance.

1. Introduction

The ability to adjust postural control is of major importance in daily life (Fransson, 1998). Whether lying, standing or sitting, gravity is working against us. Often individual muscles are not activated voluntarily. Equilibration and posture control is an automated process, which works even when there is no anticipation of movement by the human. The total state of equilibrium does not exist. People are constantly exposed to perturbations, both at rest and in motion. Whether there is a hole in the pavement or an increased stair step, in such situations, they usually have no time to react voluntarily. In this case, the human reflectory system plays an important role. Here, the sensory information comes from somatosensory receptors, such as muscle spindles, tendon organs, joint receptors, cutaneous mechanoreceptors, nociceptors, and thermal sensors (Brooke & Zehr, 2006).

As part of the B.A.L.A.N.C.E[1] project, the effect of unexpected horizontal perturbations during stance on the organization of automatic postural responses was studied in the human subject. Moore et al. (1988) investigated, that postural responses to unexpected perturbations during stance are automatic and highly stereotyped in humans (Diener et al., 1984; Nashner, 1977; Nashner et al., 1979). They exposed subjects to unexpected horizontal translations of the support surface, while oriented at several different angles with respect to the platform motion. The responses to horizontal translations of the support surface in the anterior-posterior direction (*A-P direction*) involve activation of particular muscle groups with distinctive amplitude and latency relationships. For unexpected perturbations in the A-P direction, the ankle strategy is used to exert torque at the ankle and hip strategy is used when the torque at the ankle is insufficient to correct stance (Moore et al., 1988, p. 648). It is shown, that ankle muscles are used as the primary postural control mechanism in response to sagittal plane perturbations (Runge et al., 1999).

The response to perturbations can either result in displacement of body segments against external forces, like leaning against gravity, or muscle force generation with upright trunk (Winter, 1995). According to Rode and Seyfarth (2014, p. 2) it is possible to decompose postural control from axial force production. Certain combinations of torques applied at different joints can result in forces perpendicular to the leg axis (*perpendicular forces*) in the sagittal plane, which are used to balance the upper body (Hoitz, 2014, p. 8).

[1] The Balance Project is a interdisciplinary project funded by the European Union. It aims at creating an exoskeleton that provides balance support for humans.

This has led to the hypothesis, that biarticular muscles are responsible for the compensation of horizontal forces, whereas monoarticular muscles act predominantly in axial direction. Existing standing simulations show that the concept of biarticular muscles can be applied as well, if specific subtasks are fulfilled by the vastii (body support), rectus femoris and hamstrings (postural control), and the synergistic action of all biarticular muscles (Rode & Seyfarth, 2014, p. 6).

In upright position, the center of mass is high up, the support surface of the body small and the ankle stiffness low. Standing on two legs represents a much greater challenge, as the four-legged stance of animals. The human being is often not aware of this challenging sensory-motor performance, it is like the musculoskeletal system operates as its own. Another publications showed an increased contribution by biarticular muscles to the compensation of horizontal forces. Maus et al. (2010) studied the walking model compared to an inverted pendulum into detail. They assume a virtual pivot point (*VPP*) above the center of mass as an external support to maintain balance during walking. In human walking, torques extending the hip are required in the first part and torques flexing the hip are required in the last part of stance. These torques reduce horizontal forces. Another purpose of biarticular muscles is the coupling of torques and therefore the transfer of energy, e.g. the M. gastrocnemius transfers work produced by the knee extensors to the ankle joint (Bobbert & van Soest, 2000, p. 50).

2. The B.A.L.A.N.C.E project

The experiments evaluated in this work are part of the European B.A.L.A.N.C.E project. This project is part of the 7th Framework Program (*FP7*) of the European Commission, which has a total budget of over € 50 billion. The goal of B.A.L.A.N.C.E is to realize an exoskeleton that improves postural balance performance of humans, targeted at users facing balance challenging conditions or suffering from a lack of ability to walk. In the context of biomechanics, postural balance describes the ability to control the Center of Mass (*CoM*) in relationship to the base of Support. In quiet stance the task is to keep the CoM safely within the base of support, reducing the movement of the center of pressure to a minimum, while the task during walking is to successfully integrate postural adjustments with locomotor strategies to maintain safe gait (Shkuratova et al., 2004).
The exoskeleton, which will be developed within the EU project, should cover two main use cases:

1. Worker support
2. Assistive device

First use of exoskeletons is to support and augment human motor functions. People would no longer need to carry heavy loads over a large distance. Such an exoskeleton could be used in neuro-rehabilitation, as well as for military purposes.

In addition to worker support, such an exoskeleton could be used as an assistive device in order to help people that suffer from mild health conditions during movement. This makes elderly users the primary subjects in this group. Through appropriate intervention and rehabilitation risk of falls may be reduced (Hur, 2010, p. 3). But also post-stroke patients, who suffer from hemiparesis, meaning an asymmetry in functionality of both legs, could use such an assistive device during neurorehabilitation. Thus, lower extremity exoskeletons for these groups should fulfill two important functions. One is to support progress in walking and the second is to maintain a safe gait pattern to prevent falls. According to the B.A.L.A.N.C.E consortium (2013, p. 7), the exoskeleton should reduce the metabolic cost of walking, which effectively will increase the mobility range of the patients.

3. Theory

About 1,5 million years ago, the species *Homo erectus* **arose during evolution, which was similar to the modern human, with respect to its feet construction, as well as its upright form and bipedal locomotion (Bennet et al., 2009; Nigg et al., 2000, p. 161). Nevertheless, these extinct species struggled with the same problems like people nowadays. For the coordination of movement the central nervous system (***CNS***) with its control mechanisms plays an important role. The CNS takes over the control of nerve impulses, which arrive at the motor end plate and cause the contraction. In cooperation with the brain the CNS makes intelligent decisions, which and how many muscles will be stimulated. According to Sweigard (1974, p. 174) another important factor that contributes to equilibrium is taking correct posture:**

"Posture in the standing position is a dynamic phenomenon in which the amount and extent of muscle work, and the wear and tear on the skeletal framework and its joints and ligaments, depend largely on the efficiency of the neuromuscular coordination engaged habitually in maintaining upright balance."

The ideal postural alignment of the human body accords with the principles of mechanical balance, as closely as its structural design will allow. Unfortunately, no alignment of the human skeleton with its many bones, so varied in size and design, can result in mechanical balance. The human body needs muscle power and ligamentous support at joints to maintain its upright equilibrium (Sweigard, 1974, p. 183).

3.1. Inverted Pendulum Model

All human movement is performed in a gravitational environment, which makes posture and balance a complicated task regulated by human visual, vestibular and somatosensory system (Gaerlan, 2010, p. 8). In response to slow velocity, or small amplitude surface displacements, the body behaves like a flexible, inverted pendulum, displacing the center of mass while keeping the trunk approximately aligned with the lower extremities (Henry et al., 1998, p. 1). Nashner and McCollum (1985) make distinctions related to the perturbation intensity. According to Mille (2003, p. 3) with small perturbations, the body tends to behave like a simple inverted pendulum, whereas with larger amplitude it behaves more like a double-inverted pendulum.

While applying perturbation (chapter 4.1.2), an increased muscle activity can be recorded at severe points in time. After each perturbation, the involved body segment needs to swing back to its anatomical rest position. During this phase increased muscle activation can be recorded, which we designate as a secondary effect. Such secondary effects can be observed better, when adding a second pivot point (i.e. hip joint) to the ankle. Thus, it is desirable to use double inverted pendulum biomechanics in modelling human balancing behaviour (Hettich, 2011, p. 2). However, because of complexity issues the single inverted pendulum model (Fig. 1) is used within this work, in order to keep the validation process simple and understandable (chapter 5.3).

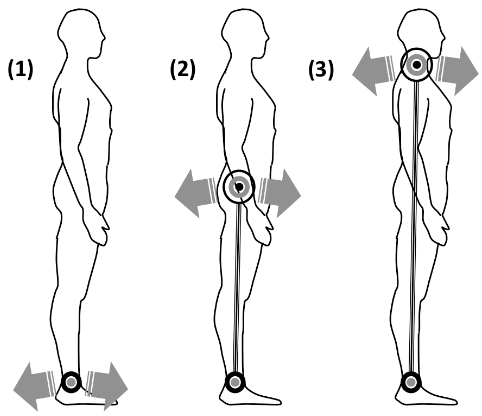

Figure 1: Inverted pendulum model for perturbation trials at ankle (1), hip (2) and shoulder (3)

3.2. Control instances of the CNS

The response signals for muscle contraction are connected via neural pathways, where all sorts of information are sent, e.g. bending the little toe requires a plurality of muscles. Gray (1973) states: "*It is seldom possible for a person to make a single muscle contract at will*." Our brain would be overwhelmed within the shortest time, if movement patterns were based on voluntary muscle activation. Not single muscles, but movement programs are stored in the central nervous system, e.g. generalized motor programs (Roth, 1989; Wiemeyer, 1982, 1994; Wollny, 1993). Thus, the coordination of muscle action to produce a desired movement and to stabilize the skeletal structure, is patterned in the CNS (Sweigard, 1974, p. 146).

In order to plan, control and execute voluntary movements, the cerebral cortex plays an important role. At the top of the command chain lays the sensorimotor area of the cerebral cortex. The fraction of the cerebral cortex controlling each part of the body is by no means proportional to the size of that part, e.g. the thumb needs 10 times more cortical area for its control, as the whole thigh (McCahon, 1984, p. 141). The following table (Table 1) gives an overview about the motor centers, which are responsible for postural control (Gollhofer, 2009, p. 175):

Table 1: Responsible motor centers for postural control

Motor Centres	Function
Spinal cord	Fastest and easiest processing of afferent information
Brainstem	Sensorimotor integration
Cerebellum	Coordination of movement
Basal ganglia	Postural flexibility
Motor Cortex	Planning, control, and execution of voluntary movement

The spinal cord is the fastest and easiest processing structure of afferent information. Here, reflectory signals are processed without passing through the brain. The brainstem is composed of the extended spinal cord (*medulla oblongata*), pons and midbrain. In early studies, mammals were able to compensate perturbations via reflectory mechanisms of the spinal cord/brainstem without the involvement of higher centers (Magnus, 1924; Sherrington, 1906). In the neural network *Formatio reticularis* the sensory information from the vestibular, proprioceptive and the visual system runs together and can be integrated into motor commands (Gollhofer, 2009, p. 176). Perception of imbalance by visual, proprioceptive, tactile and vestibular analyzers is an essential prerequisite to perform adequate compensatory movements, in order to restore the equilibrium (Gollhofer, 2009, p. 172). Nevertheless, Fritzpatrick et al. (1994) showed that upright position with simultaneous exclusion of visual, vestibular and cutaneous sensors is also possible.

The cerebellum is responsible for the coordination of movements, as well as for the coordination of agonist and antagonist activity (Diener & Dichgans, 1992). Postural control is specifically adapted to the perturbing situation. Nashner (1976) showed, that patients with cerebellar lesions often are not capable of performing this type of motor learning (Gollhofer, 2009, p. 177f.). If the cerebral cortex is removed entirely, the reflectory system of the spinal cord is still in function. It is almost normal with respect to many common motor functions, but it's not possible to learn new skills (McCahon, 1984, p. 141).

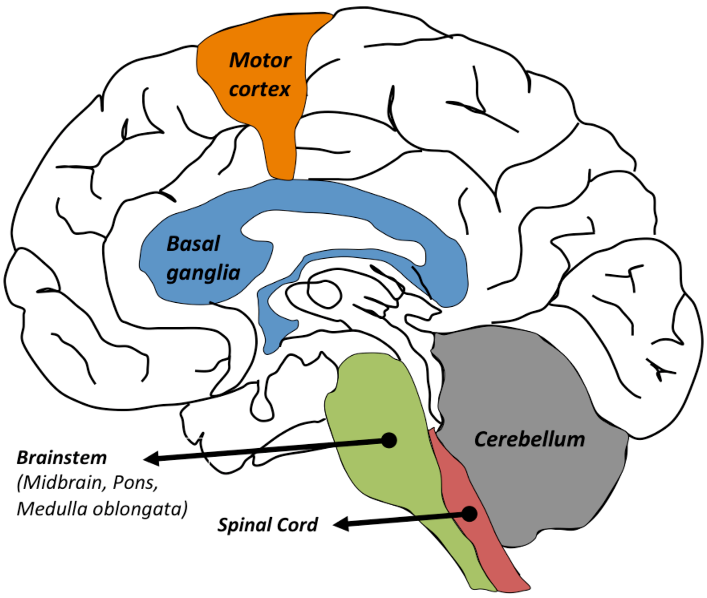

Figure 2: Centers of the human brain (mod. after mult-sclerosis.org)

The spinal cord, brain stem and sensorimotor cortex guarantee along with the basal ganglia and the cerebellum (Fig. 2) that a continuous flow of information from the sensorium can be used to control the movement. This flow involves information, such as the position of the body and the limbs in space, the length and tension of the muscles as well as the moving speed. Each of the three control instances is provided independently by sensory information, so they have their own influence on movement control. Furthermore, each instance is responsible for specific aspects of motor control. Thus, the system of movement control has both, a hierarchical and a parallel organization structure (Schubert, 2009, p. 21).

3.2.1. Onset latency

Due to the efferent pathways in the human *CNS*, the action potential arrives delayed at the muscle. Prior muscle activation, action potentials need to cross the synaptic gap. From the generation of an action potential, up to the activation of a motor unit, it takes a while (Stein et al., 2000). The speaking is of onset latency. Simply said, this describes the length of time for a muscle transition from the state "not activated" to "activated". Several processes contribute to this delay, e.g. it depends on the nerve conduction, synaptic transmission, generation of muscle action potential, conduction of the potential along the muscle fibre and down into the transverse tubules, release of $Ca2+$ from the sarcoplasmatic reticulum, binding of the $Ca2+$ ions to troponin molecules, or even on a conformational change in the troponin molecule that moves the tropomyosin molecules and allows the binding of myosin heads to actin molecules to begin (Mannard & Stein, 1973, p. 164).

3.3. Group innervation and muscle coordination

A muscle never executes a complete movement as an isolated element (Bernstein, 1935, p. 83). Duchenne (1959) confirms Bernstein's point of view concerning muscle coordination: "Isolated action of the muscle is not in the nature of things." The key to understand muscle coordination is to find the contributions of individual muscles to movement.

In order to execute motor programs correctly, the muscles work either together, or against each other (Donskoi, 1975, p. 54). According to the antagonistic principle, the skeletal muscle acting on a joint is matched by other skeletal muscle producing an opposite action on the same joint. At first sight this may sound paradoxical, e.g. while the coactivation of hamstrings and M. rectus femoris. Such coactivation would result in hip and knee extension, due to the different moment arms between the muscles at hip and knee joint. At the knee joint the moment arm of the M. rectus femoris is larger than of the hamstrings while the opposite is true for the hip joint. Thus, the power liberated in the hamstrings will predominantly appear as an increase in net power in the knee joint as long as the M. rectus femoris generates a larger force than the hamstrings (Van Ingen Schenau, 1989, p. 318). Another feature, which take over biarticular muscles, is a lower contraction velocity. If hip extension and knee extension occur simultaneously, the shortening velocity of the biarticular hamstrings/M. rectus femoris is lower than of the monoarticular hip/knee extensors (Van Ingen Schenau et al., 1990, p. 641). This indicates, that the muscles are operating in a more favourable region of their force-velocity relationship.

3.4. Postural control

Postural control is achieved by the interaction of many components and includes the integration of visual, somatosensory, and vestibular system information as well as the execution of appropriate motor commands (Winter, 1995; Maeda et al., 2013). Without the involvement of the muscles only passive postures are maintained, e.g. lying on the ground or in the water (Donskoi, 1975, p. 187). In case of active postures, the system of mutually movable limbs becomes rigid by the tension of the muscles. A human kind is able to maintain balance, but also in case of unexpected perturbations to restore it. The significance of balance in biomechanical context, compared to a rigid physical body, is not the existence of specific laws of mechanics to living systems, but the intelligent use of it (Donskoi, 1975, p. 193). In situations where the sizes of the postural disturbances change, an adaptation process is necessary by adjusting the timing and amplitude of their postural muscle responses (Dietz et al., 1989; McIlroy, & Maki, 1993). Depending on the age, this process may last longer or shorter (Lin et al., 2002, p. 37).

3.4.1. Postural response to perturbations

To counteract disturbances of equilibrium, the CNS must activate appropriate muscles depending on the perturbation and biomechanical preconditions (Gollhofer, 2009, p. 174). The response reaction of each muscle is unique, since the amplitude and timing relationships between the muscles are different for each of the perturbation directions (Moore et al., 1988, p. 653). It has been suggested that certain pre-programmed activation patterns exist that are initiated depending on the perturbation (Nashner & McCollum, 1985). Due to the numerous types of perturbations and the multitude of different compensatory reactions, it is unlikely that such automatic postural responses in the form of detailed movement trajectories exist (Schmidt & Lee, 1999). The control of low surface oscillations (below 1 Hz) is mainly attributed to the vestibular system, whereas for rotary displacements of the support surface vestibo-spinal mechanisms play a role (Allum & Pfaltz, 1985). In contrast, the somatosensory system is increasingly used to compensate translational displacements.

3.4.2. Purpose of biarticular muscles

Each muscle is unique and provides a unique contribution to a certain motor task (Kuo, 2001, p. 11). Bobbert and Soest (2000) investigated the hypothesis whether the control of a system with biarticular muscles is simpler and more flexible than the control of a system with only monoarticular muscles. In total, 12 muscles (4 monoarticular) are involved in the production of a torque at the knee joint, of course muscles are activated in a sequential order. Some muscles need to work harder to achieve movements that do not lie in pulling direction (Diedrichsen, 2009, p. 3). Almost all muscles of the body attach closely to joints. Due to the short lever arm, contraction at a higher force application is saved. This is known as the golden rule of muscle mechanics (Donskoi, 1975, p. 44). One of the main functions lays in the coupling movement in the joints that are crossed by biarticular muscles, e.g. the M. gastrocnemius couples knee extension to plantar flexion. This mechanism allows a transport of power from knee to ankle joint (Van Ingen Schenau, 1989, p. 313). According to Rode & Seyfarth (2014, p. 2) postural control might be decoupled from axial force production. Certain combinations of torques applied at different joints can result in perpendicular forces in the sagittal plane, which are used to balance the upper body.

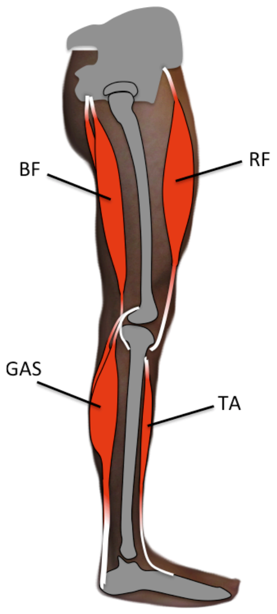

Figure 3: Four leg muscles (BF = M. biceps femoris; RF = M. rectus femoris; GAS = M. gastrocnemius; TA = M. tibialis anterior)

4. Methods

Within this investigation 9 male sports students participated in the experiment. The participants were in a healthy condition and did not suffer from any diseases or restrictions in their mobility at the time of the measurements. The mean age, body mass and height of the participants were 24.3 ± 1.9 years, 77.3 ± 7.8 kg and 182.1 ± 7.5 cm.

The experiments described in this work are designed to answer more questions than approached in this thesis. This work focuses on the dynamic change in activity of particular muscles. The EMG signal is further examined in this work, to see if the muscle is turning on or off immediately after the perturbation.

4.1. Experimental Design

A subject standing on both legs generates torques at each leg equally. While applying a force at the ankle, the affected leg has to be lift off the ground. This assures, that the standing leg is able to compensate all torques. Furthermore, it was perturbed at the hip joint and at the neck in antero-posterior direction. The intensity of each perturbation (10 N, 20 N, 30 N) was estimated randomly.

4.1.1. Perturbing zones

At ankle, hip and shoulder severe perturbations were applied, in both anterior and posterior direction. In order to determine the influence of the knee joint angle on the response action of particular muscles, the subject had to withstand the perturbation with stretched (180°), semi-flexed (155°) and flexed (140°) knees. It was also perturbed in vertical direction, again with different knee joint angles. The force application was applied in horizontal, as well as in vertical direction, to show how leg function can be decoupled into axial and perpendicular function. Axial leg function support the body weight, whereas perpendicular leg function support balance or acceleration/deceleration in locomotion. The selection order of the perturbing zones was chosen randomly, so that the subject could not adapt to the perturbation. Thus, effects of accommodation and habituality can be avoided.

4.1.2. Duration of a trial

One trial at a specific body zone (e.g. ankle anterior) lasted 100 seconds. During this time, 6 perturbations were applied, which respectively had a duration of 15 seconds (Fig. 4). Before and after each perturbation there was a short break from 5 seconds to reattach the rope. In addition, 5 seconds were added at the end of each trial in order to avoid unpredictable events.

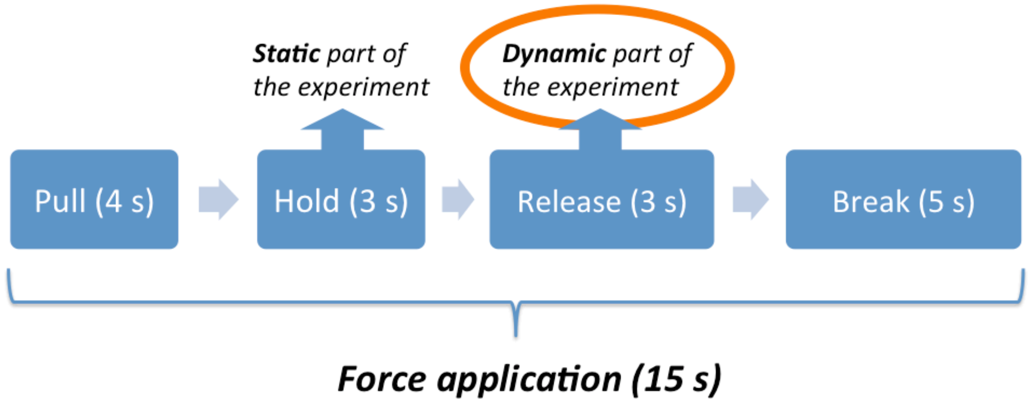

Figure 4: Force application process

As seen in Fig. 4, the force was increased until the desired force has been reached within the first 4 seconds. In order to investigate the static muscle behaviour (Hoitz, 2014), the subject had to counteract the force at the same time to keep the balance. After 3 seconds, external force caused by cord tension was aborted instantly. During this time the dynamic muscle behaviour can be analyzed into detail.

4.1.3. Presumption on muscle activation

In this study it is assumed that 10 muscles play an important role for the balancing behaviour of the human. Depending on force application different muscles are responsible for the compensation of the external perturbation. The following table shows the predicted response reaction of each muscle (Table 1).

Table 2: Presumption on highly (de)activated muscles (light blue: decrease in activity; dark blue: increase in activity)

	L_BF	L_TA	L_SL	L_ST	L_RF	L_VM	L_VL	L_GT	L_GM
AAE	→	→	→	↑	→	→	→	→	↓
APE	→	→	↑	↓	↑	→	→	→	↑
HAE	→	↑	→	→	→	→	→	→	↓
HPE	→	→	↑	→	→	→	→	→	↑
SAE	↓	↑	↓	↓	↑	↑	↑	↓	↓
SPE	↑	↓	↑	↑	→	↓	↓	↑	↑

Based on the muscle work during the static part of the experiment (chapter 4.1.2), predictions were made whether the muscle is going to increase or decrease in activation. Within this work the single inverted pendulum model (chapter 3.1) is used to make these predictions. However, problems arise in the simple pendulum model, when predictions were made for perturbations at the ankle. Since the lever arm is nearly equal to zero, the body mass is hardly excited to vibrate. The force is compensated at the ankle. The pendulum is not caused to oscillate, thus, only secondary effects can be reported at the ankle. This distinguishes ankle perturbation from other perturbations. When perturbing at the ankle zone, muscles work from left and right leg against each other, e.g. while the left hamstrings are working right side, the M. rectus femoris remains active left side. This applies to both biarticular muscles, as well as to monoarticular muscles.

In the multiple inverted pendulum model secondary effects must be integrated within the prediction of muscle activity, e.g. while applying forces at the hip in posterior direction. Here, the M. soleus goes on at first, but in the back swing phase M. gluteus maximus needs to be activated. In the double pendulum model many muscles remain inactive, but in time of the back swing phase they become active, e.g. the hamstrings during trial HAE (Hip Anterior Extended).

4.2. Experimental Setup

As part of the experiment dynamometric, kinemetric and electromyographic recordings were captured, all triggered by a common start signal.

4.2.1. Dynamometric recordings

In order to trigger the electromyographic recordings, the ground reaction force (*GRF*) in x-, y- and z-direction with a KISTLER force plate (*9260AA*) and a frame rate of 1000 Hz has been measured. This is to ensure that the time lag between the start times is small compared to the measurement frequency (less than 1 ms for force measurements) and that the measuring systems remain synchronized within the measurement.

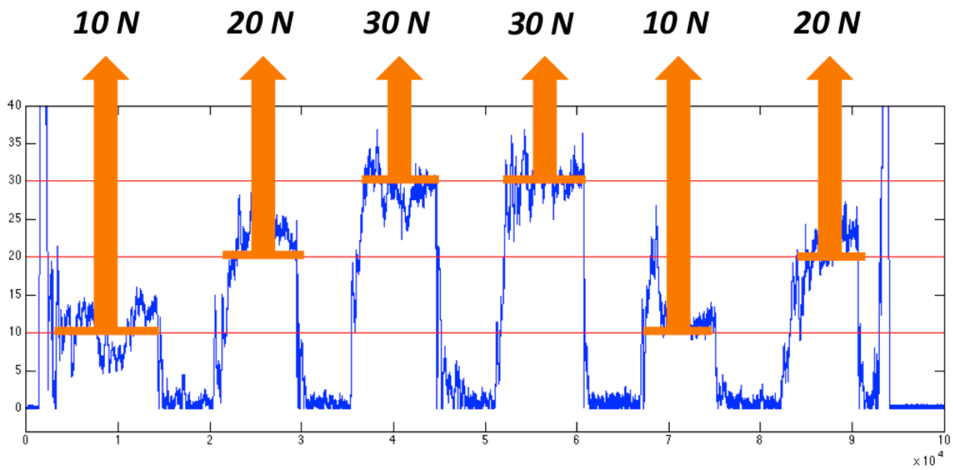

Figure 5: Measured GRF in the sagital plane (HAE)

As shown in Fig. 5, some challenges must be faced in order to retrieve necessary information from the dynamometric recordings. For a human being it is easy to determine the applied force intensity, however, a computing machine needs to struggle and perform certain algorithms to estimate the correct intensity. In addition, there is no guarantee that the penknife indicates the exact traction due to the long shortening path. Furthermore, holding a force steadily to 100% is virtually impossible.

4.2.2. Kinemetric recordings

Beside the electromyographic and dynamometric measurements, kinematic data was recorded using the motion capturing *Oqus 5* system (Qualisys) at a 250 Hz sample rate. A total number of 15 markers on each subject and 1 additional marker attached to the cord were put, in order to estimate the point in time when releasing force (chapter 4.3). More information about the location and number of the placed markers can be retrieved in Table 3.

Table 3: Kinemetric marker placement

Location		Number
Forehead	LHEA/RHEA	2 (left/right)
Shoulder	LSHO/RSHO	2 (left/right)
Neck	CS07	1
Hip	LTRC/RTRC	2 (left/right)
Knee	LKNM/RKNM; LKNL/RKNL	4 (left/right; medial/lateral)
Ankle	LANM/RANM; LANL/RANL	4 (left/right; medial/lateral)
Rope	Rope	1

4.2.3. Electromyographic recordings

In order to capture the muscle activity (*EMG*) a Delsys Bagnoli Desktop EMG system was used. EMG data were gathered from 15 muscles, 10 on the left (standing) leg and 5 on the right (hanging) leg at 2000 Hz sample rate.

According to the European recommendations for surface electromyography (Hermens et al., 1999, p. 40-53) the bipolar EMG-electrodes were put on 10 different surface muscles (Fig. 6). The electrodes were placed over the muscle belly, where the signal can be detected best. Within this study, the 10 major muscles involved in balance maintenance were chosen. The functions of the muscles are listed in Table 4.

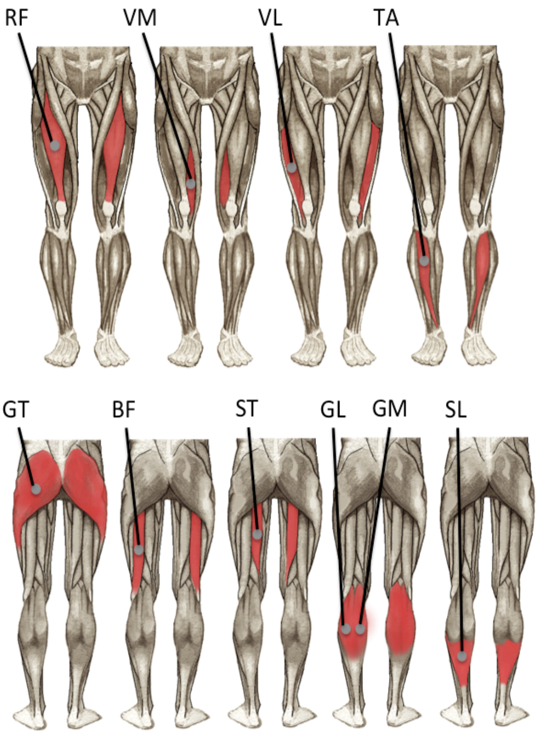

Figure 6: Involved muscles during the trials

Table 4: Functions of lower limb muscles (Hermens et al., 1999, p. 40-53)

Muscles	Function
M. rectus femoris (RF)	Extension of the knee joint and flexion of the hip joint.
M. vastus medialis (VM)	Extension of the knee joint.
M. vastus lateralis (VL)	Extension of the knee joint.
M. tibialis anterior (TA)	Dorsiflexion of the ankle joint and assistance in inversion of the foot.
M. gluteus maximus (GT)	Extends, laterally rotates and lower fibres assist in adduction of the hip joint. The upper fibres assist in adduction. Through its insertion into the iliotibial tract, helps to stabilise knee in extension.
M. biceps femoris (BF)	Flexion and lateral rotation of the knee joint. The long head also extends and assists in lateral rotation of the hip joint.
M. semitendinosus (ST)	Flexion and medial rotation of the knee joint. Semitendinosus also extends and assists in medial rotation of the hip joint.
M. gastrocnemius medialis (GM)	Flexion of the ankle joint and assists in flexion of the knee joint.
M. gastrocnemius lateralis (GL)	Flexion of the ankle joint and assists in flexion of the knee joint.
M. soleus (SL)	Plantar flexion of the ankle joint.

4.3. Procedure

Human studies require special care to protect the interests and rights of the participating subjects. All participants must be informed about the aims, procedure and risks of a trial and agree to the procedure. An important part of this process is to clarify the preparative actions when placing EMG electrodes on the muscle surface.

4.3.1. Preparative actions

The EMG measured signal can be influenced by a variety of factors (Gruber et al., 2009, S. 132), e.g. muscle anatomy, characteristics of single motor units, movement-related factors, physical or technical factors. A preventive arrangement to reduce confounding variables is the adequate preparation of the contact junction, this includes:

- Skin shaving
- Degreasing/cleaning by alcohol
- Correct application of the electrode
- Putting on the electrodes parallel to the muscle fibres direction

4.3.2. MVC measurements

A major drawback of any EMG analysis is that the microvolt scaled values depend strongly on the given measuring conditions. One way to eliminate the confounding variability is the normalization of an amplitude value to a reference value, e.g. the innervation value of a maximum voluntary contraction (*MVC*). The major effect of all normalization routines is to rescale the amplitude from microvolts to percent of a selected reference value. It is important to understand that the amplitude normalization does not change the shape of the EMG curve, but only the Y-axis scale (Konrad, 2005, p. 29).

Before starting the experiment, static MVC contraction values for each muscle were recorded. The resulting MVC innervation level serves as a reference level for all subsequent measurements. In order to get valid results, it is important to fixate all involved body segments. After an initial warm-up, the subject in the MVC test position is prompted within 3-5 seconds to increase the force to the maximum and to hold it just as long. Afterwards the subject relaxes 3-5 seconds and repeats this cycle at least once after a 30- to 60-second pause. The order of exercises should subject-wise change to avoid systematic errors (Konrad, 2005, p. 29).

The major advantage of the MCV normalization is the creation of a uniform and valid reference. This allows a direct quantitative comparison of EMG values between individuals. Group values and normative data can be determined and verified statistically (Konrad, 2005, p. 32).

4.3.3. Application of forces

The force application process (Fig. 7) was carried out using a cord connected to a penknife, which was attached to the subject's body. The other end of the cord was connected to a crossbow-like trigger. This mechanism enabled a sudden force release, in order to investigate the dynamic behaviour of the involved muscles. In addition, forces were applied in vertical direction by holding weights (1 weight = 5 kg) in both hands, which the subject had to drop after 3 seconds. The horizontal forces were applied once with 10 N, 20 N and 30 N, whereas the vertical ones were 100 N, 200 N and 300 N. To avoid possible anticipations or accommodation effects of the subject, this procedure was repeated.

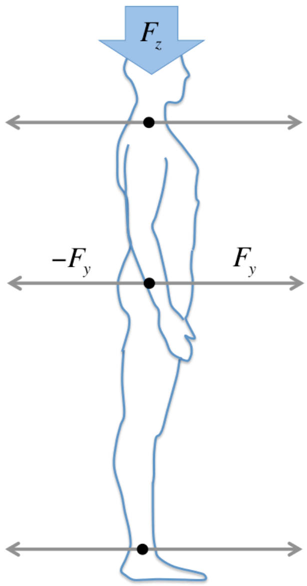

Figure 7: Locations and directions of applied forces (mod. after Hoitz, 2014, p. 20)

4.4. Data processing

In the following several methods on data processing will be considered into detail.

4.4.1. Dynamometric processing

The perturbing times vary within the trials, which could be due to human imperfection. In order to get the proper perturbing times, the dynamometric recordings were taken into account. These recordings in antero-posterior direction (Fig. 7) in the sagittal plane provide information about the applied force. It has been derived with respect to time. The start/end points of each perturbation could be estimated precisely when the first derivative of the smoothed force curve exceeded the value of 0,5. Based on this formulation the gradient of the force curve is sufficiently high in this case, so that a sufficient force increase or decrease can be assumed. This results in 12 windows (window size: 1 s), which describe the times when forces are applied or no forces are applied, respectively. Those 12 windows (Fig. 8) form the basis for further EMG analysis.

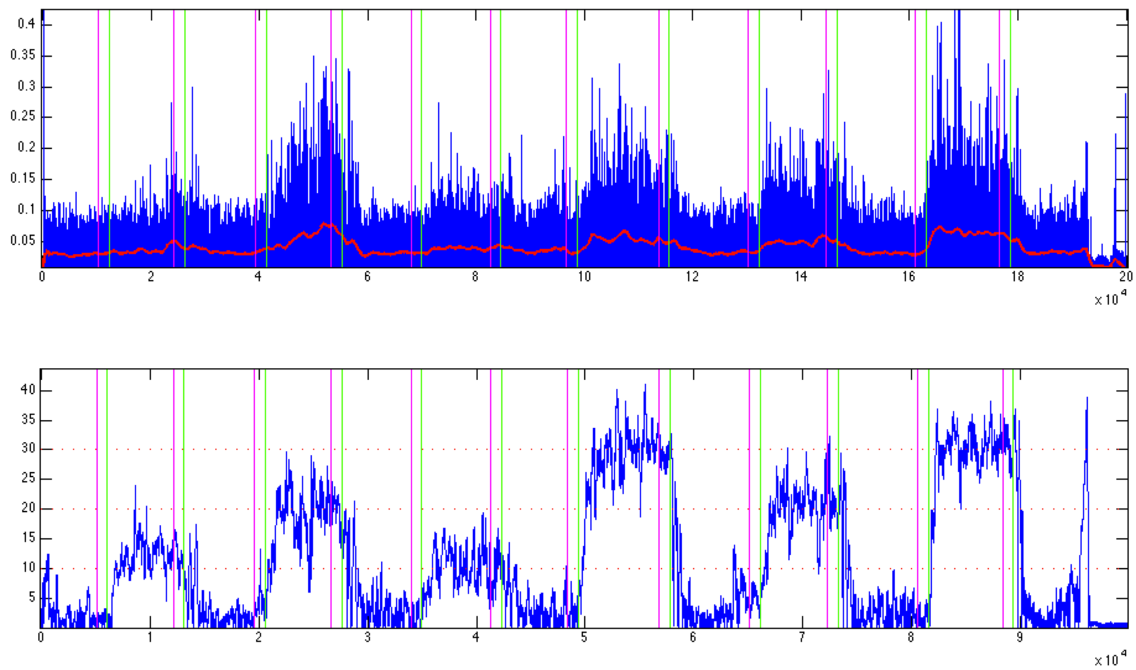

Figure 8: Definition of perturbing times based on dynamometric recordings

The graphs in Fig. 8 are purely illustrative. In the upper graph all EMG samples are presented along the x-axis. With a total number of 200.000 samples and a frequency of 2000 Hz, we get a time period of 100 s. Along the y-axis the normalized EMG activation is shown. In the lower graph the curves of the force plate during perturbations in the sagittal plane are plotted against all 100.000 samples recorded with a corresponding frequency of 1000 Hz. The curve hills have their high points at

10, 20 resp. 30 N. The vertical lines drawn in magenta show the initial times of each perturbation and the green lines the related end times.

4.4.2. EMG processing

Due to the sensitive nature of EMG recordings, the signal is affected from various external noises or artefacts, respectively. In order to remove the unwanted signal components, the processing part is divided into different tasks:

1. General Filtering
2. Rectification
3. Normalization
4. Smoothing

In order to find appropriate cutoff frequencies, several comparative studies were taken into account. Henry et al. (1998, P. 34) bandpass filtered the EMG signal within the frequency range of 75-2.000 Hz. Prior to filtering they pre-amplified (x 5000-10.000) the EMG signal. The dynamics of activation takes place within the range of a few Millivolts. Hence, signal amplification should be handled with caution, in order not to amplify unwanted interference components. Nashner (1977, p. 15) also bandpass filtered the EMG signal within the range of 10-10.000 Hz, after the signal was full-wave rectified. In the reactive balance experiments investigated by Ferber et al. (2002, p. 240) all raw EMG analogue signals were pre-amplified (x 7000), analogue filtered (20-7000 Hz), and then converted into digital signals sampled at 1200 Hz. Based on different approaches related to own experience, a second-order Butterworth filter (Fig. 9) within the frequency range of 10-150 Hz was used. The high cutoff frequency (150 Hz) is low compared to the values suggested by the aforementioned authors, but in this study most of the frequency power lays within this range.

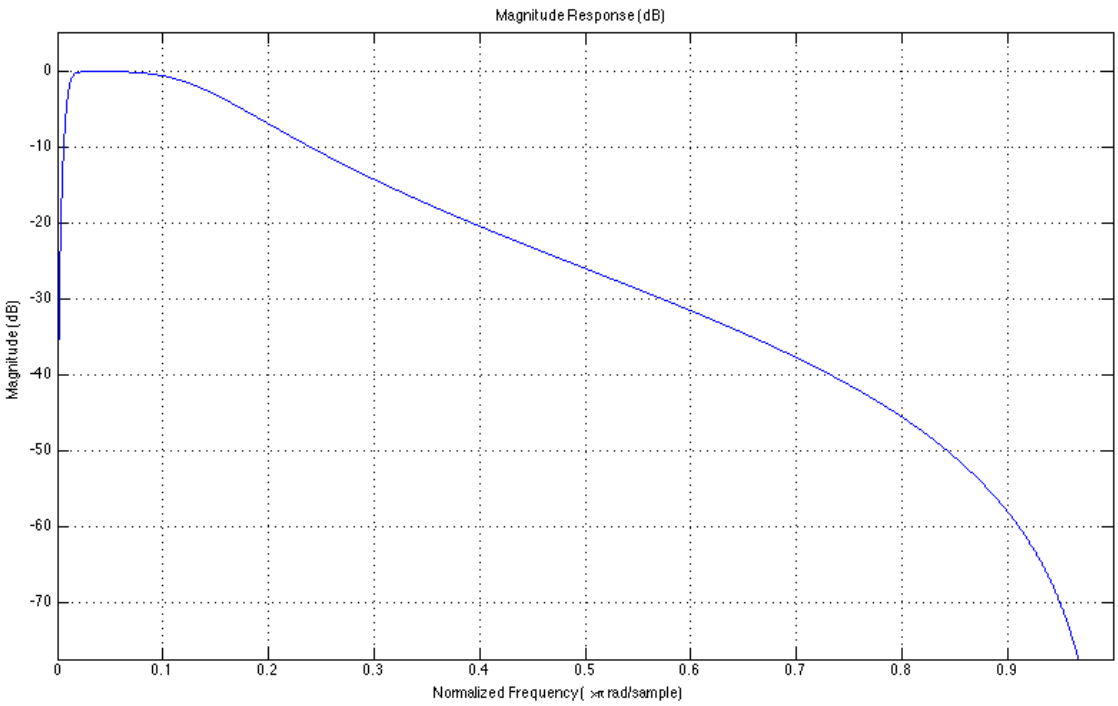

Figure 9: Second-order Butterworth filter in MatLab (cutoff frequencies 10-150 Hz)

Prior to full-wave rectification, a 50 Hz notch filter has been applied.
For normalization issues the muscle activity was set in relation to the maximum voluntary contraction of each muscle. At least, the **M**oving **A**verage (*MAV*) algorithm has been performed in order to smooth the EMG curves and make them more readable. Hereby a window size of ¼ s was chosen.

4.4.3. Estimating dynamic muscle activity

An important task of data processing is to estimate the dynamic muscle behaviour immediately after releasing the applied force. A fraction of a second is enough to determine which muscles are responsible for dynamic balancing issues. The activation behaviour for 10 muscles of the left leg was predicted hypothetically. By means of the experiment is to show whether the activation hypotheses are correct.
The hypothesis validation process is based on 2 values (*val1*, *val2*), which are set into relation. First value *val1* describes the averaged muscle activity 2 s before releasing the force instantly, the other *val2* is the mean of all samples recorded half a second after force release. If *val2* is smaller compared to *val1*, the muscle is about to turn off while performing a certain movement. Conversely, this means an

increase in activity of the affected muscle. If there is no change in muscular activity (i.e. activational change < 2 %), the muscle is not involved in the movement.

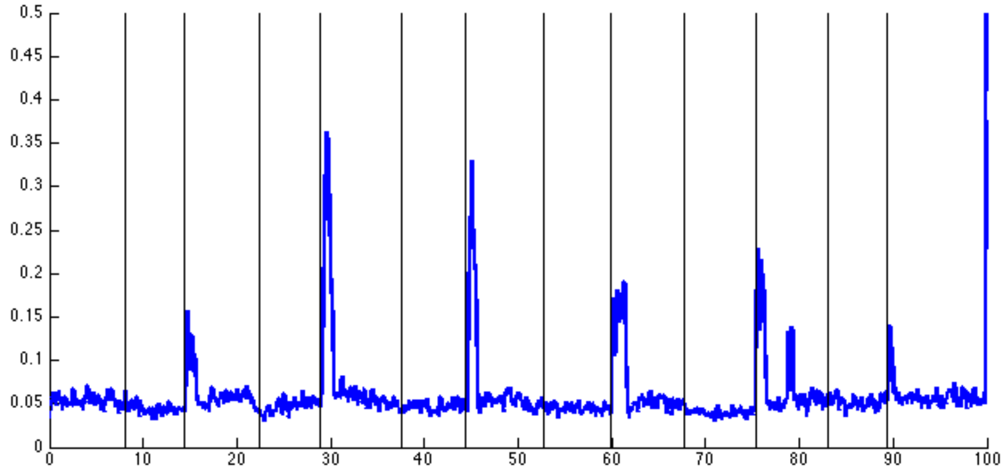

Figure 10: EMG recording of M. gastrocnemius lateralis during trial SPE

Upper Fig. 10 shows the EMG recordings of M. gastrocnemius lateralis while perturbing at the neck in posterior direction. In theory it seems to be easy capturing the peaks, which occur immediately after force release. Practically it looks different, e.g. the length of time for each peak varies from perturbation to perturbation. Likewise, it isn't possible to take the high point of each peak and set in comparison with the static value. The EMG signal is stochastic in nature, i.e. a raw EMG curve may not be reproduced in its exact form a second time. This is related to the respective active set of detected motor units, which are constantly changing. If several motor units that are very close to the electrodes are firing at the same time, a peak might occur, caused by the electrical superimposition of single motor units (Konrad, 2005, p. 10). Therefore only half a second was taken into consideration when analyzing muscle behaviour immediately after force release. The change in muscle activity is low, but there is a trend in direction, i.e. either the signal goes up, down or it stays at the same level. In order to estimate the trend, a threshold value needs to be defined. In this study, the parameter is set to 0,02 (*2 %*). A change of muscle activation more than 5% is considered as a major change. A value less than 2 % is regarded as maintaining muscle activity. This threshold parameter indicates the muscular transition from state "not activated" to "activated" and vice versa.

5. Results

In the following (Fig. 11-16) the results of the experiment are presented. In total there are 6 diagrams, each diagram represents a statistically evaluated trial with 9 subjects, i.e.:

- Ankle Anterior Extended (AAE)
- Ankle Posterior Extended (APE)
- Hip Anterior Extended (HAE)
- Hip Posterior Extended (HPE)
- Shoulder Anterior Extended (SAE)
- Shoulder Posterior Extended (SPE)

In this investigation trials were taken into account, where the subjects have their knees stretched. A further study could show the influence of the knee angle on the activation of individual muscles. In addition, no differentiation was made between the perturbation intensity (10, 20, or 30 N) and the corresponding muscle response reaction.

5.1. Activity matrix

In Fig. 11-16 the normalized change in muscle activity is plotted among the y-axis. A normalization means, that the muscle activity of each subject was set into relation to its MVC value. Starting from an activational change in activation higher than 2%, the change is stated has significantly high. In the following the matrix (Table 5) all values are presented as decimal numbers.

Table 5: Activity matrix for all muscles

	L_BF	L_TA	L_SL	L_ST	L_RF	L_VM	L_VL	L_GT	L_GM
AAE	1,87	3,86	–0,37	5,38	0,52	0,71	0,47	0,55	–0,32
APE	–1,42	1,31	3,68	–1,74	2,08	0,33	0,29	0,00	2,59
HAE	–0,63	3,56	–0,42	0,62	1,38	1,13	1,55	0,30	–1,71
HPE	0,37	–0,48	4,83	0,82	–0,48	–1,01	–1,54	–0,25	4,55
SAE	–0,19	4,82	0,87	1,27	3,29	3,80	6,87	1,32	–2,02
SPE	4,35	1,96	16,17	5,68	–0,69	–1,17	–2,40	0,22	8,16

Surprisingly, the response reaction of the left M. gastrocnemius lateralis (L_GL) is passive during all trials. There is no significant change in the activation observed. Thus, within this work the M. gastrocnemius lateralis is neglected.

5.2. Normalized response activity

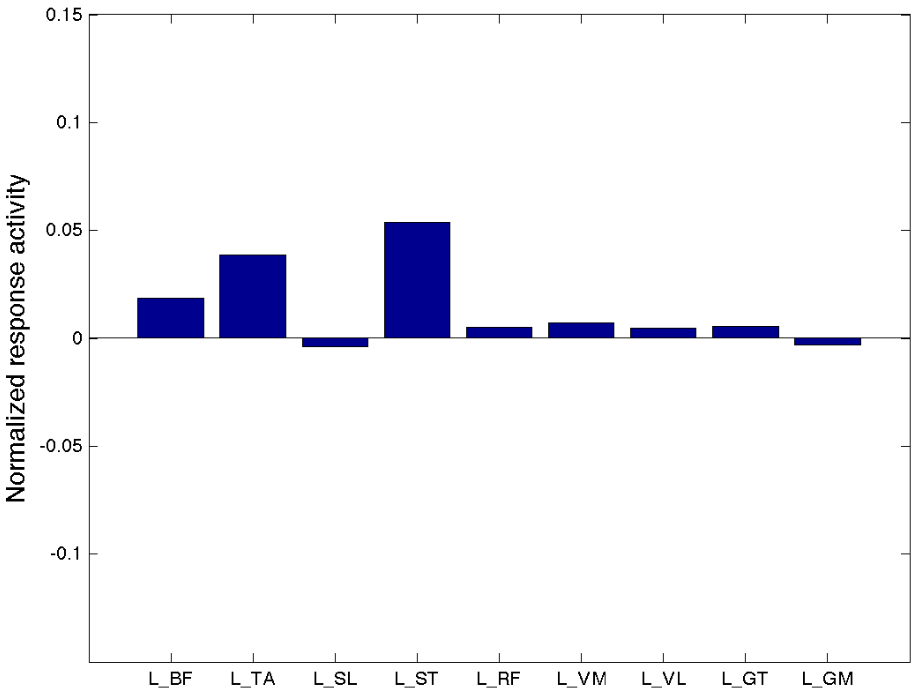

Figure 11: Normalized response activity during trial AAE

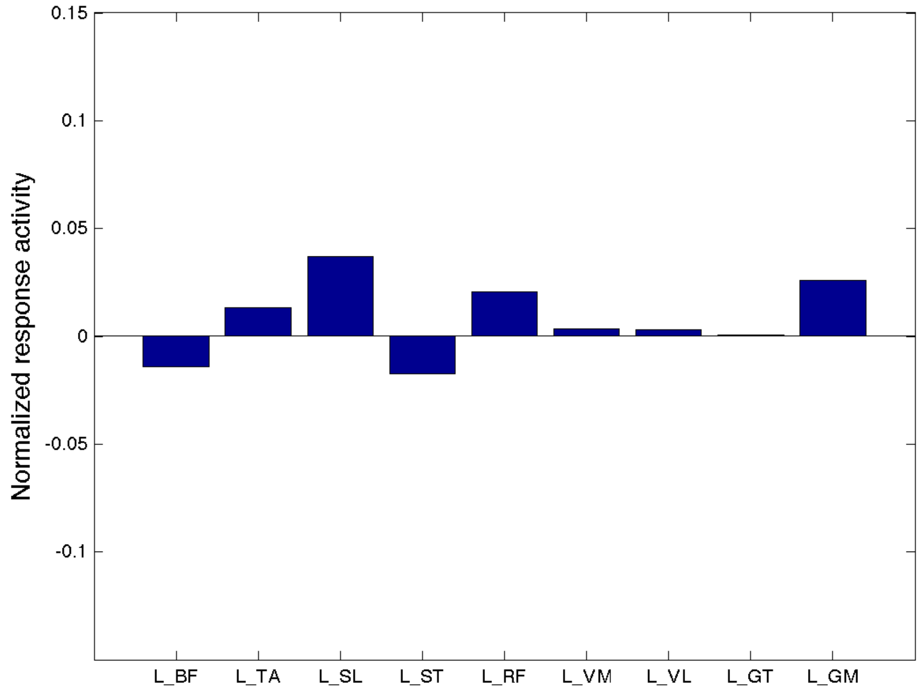

Figure 12: Normalized response activity during trial APE

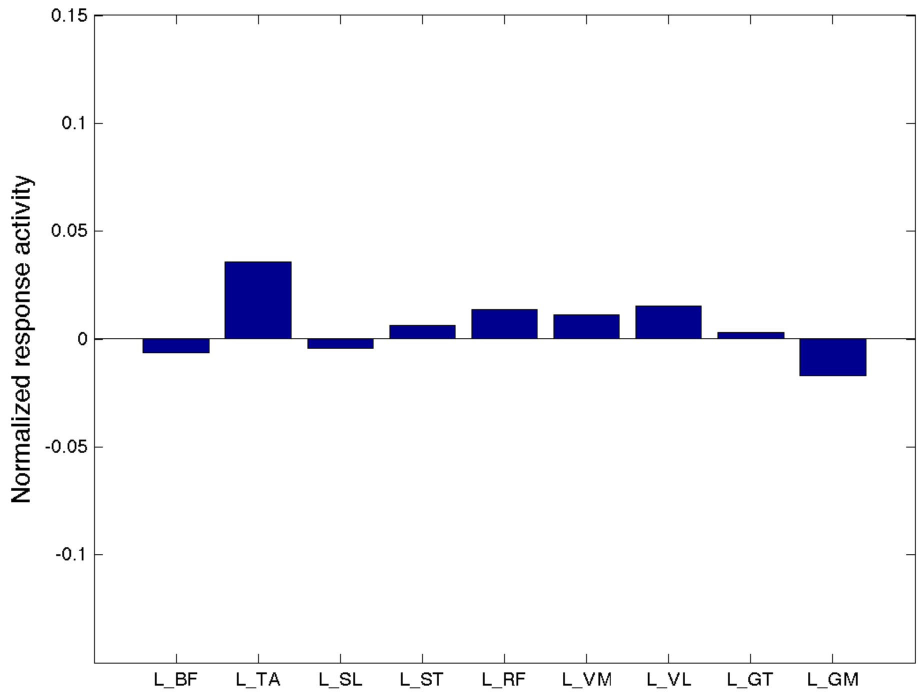

Figure 13: Normalized response activity during trial HAE

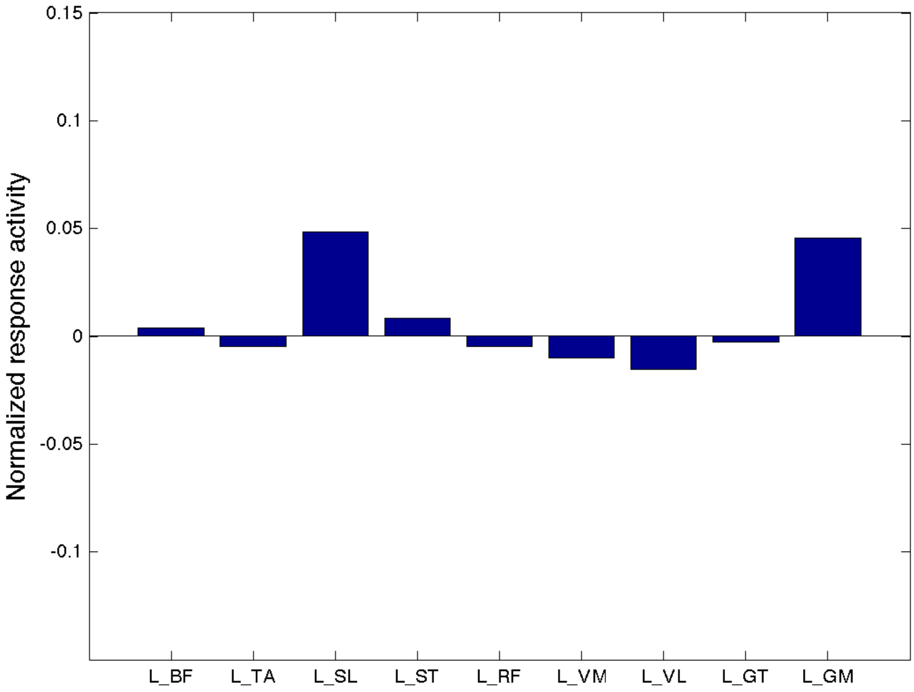

Figure 14: Normalized response activity during trial HPE

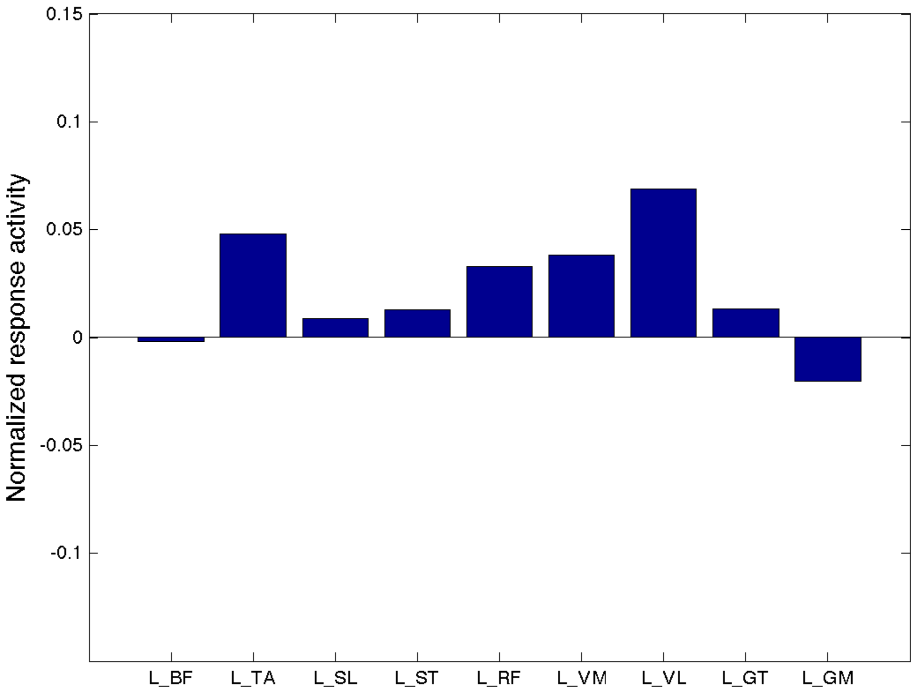

Figure 15: Normalized response activity during trial SAE

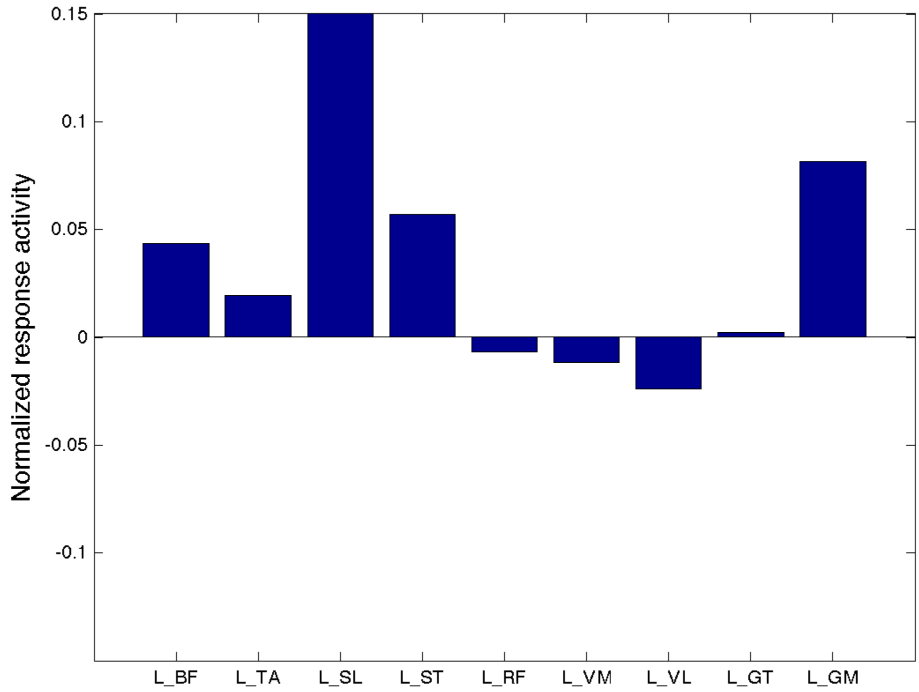

Figure 16: Normalized response activity during trial SPE

5.3. Hypothesis Validation

Referring to the previewed assumptions made in chapter 4.1, it can be seen, that in 11 cases the assumed response activity was estimated incorrectly (Table 6). The other 43 estimations (cells with green fillings) were correct.

Table 6: Validation table of hypothesized assumptions (expected value in brackets)

	L_BF	L_TA	L_SL	L_ST	L_RF	L_VM	L_VL	L_GT	L_GM
AAE	→	↑ (→)	→	↑	→	→	→	→	→ (↓)
APE	→	→	↑	→ (↓)	↑	→	→	→	↑
HAE	→	↑	→	→	→	→	→	→	→ (↓)
HPE	→	→	↑	→	→	→	→	→	↑
SAE	→ (↓)	↑	→ (↓)	→ (↓)	↑	↑	↑	→ (↓)	↓
SPE	↑	→ (↓)	↑	↑	→	→ (↓)	↓	→ (↑)	↑

The predictions are virtually correct for M. biceps femoris. In the trial SAE, a drop of muscle activity is assumed, but according to the measurement the activational behaviour remains constant. Although the activational trend is negative (Fig. 15), the threshold value of 2% could not be reached. An increased muscle activity is

recorded in M. tibialis anterior during the trial AAE (Fig. 11). In case of perturbations in posterior direction applied at the shoulder, the tibial muscle is supposed to switch off. Also the left M. soleus shows a constant activity while perturbing at the shoulder girdle, although it would need to turn off, likewise M. semitendinosus. The predictions for the left M. rectus femoris and M. vastus lateralis are consistently correct. According to the results of measurement, the activational state of the M. gluteus maximus is continuously constant.

6. Interpretation

Basically, the predictions are consistent with the measurement results. In 11 of 54 cases, the predictions are not as accurate as supposed. Most mispredictions occur during the trials in the shoulder area, where 11 out of 18 cases were predicted correctly. The deviations between the measured result and presumptions are not severe, i.e. a higher activation is measured where the muscle would have had to remain inactive. But never increased muscle activation was measured, where a decrease in activation was assumed. It is difficult to draw conclusions about the dynamic behaviour of muscles, while perturbing at the ankle. Considering the simple inverted pendulum model (chapter 3.1), the lever arm is relatively small, so that only secondary effects may occur theoretically. More descriptive statements can be drawn up from trials at the hip and shoulder. It is considered that the relationship between perturbations in the anterior and posterior direction is inverse (Table 2). One can say, if the activational state of a particular muscle within the trial SAE (*Shoulder Anterior Extended*) goes up, this goes down in trial SPE (*Shoulder Posterior Extended*). Only for the M. rectus femoris this inverse relationship is not applicable. Within the trial SAE an increased muscle activity was measured, whereas no significant change in activation within trial SPE is assumed. These assumptions have been confirmed for the M. rectus femoris by experimental data.
From the results it can be concluded, that the three knee extensor muscle heads of M. quadriceps femoris (L_RF, L_VM, L_VL) work synergistically when perturbations occur in posterior direction. Similarly, the M. gastrocnemius medialis continuously shows an increased muscle activity when perturbing in posterior direction. This works synergistically with M. soleus. The monoarticular tibial muscle shows a continuously increased muscle activity during perturbations in anterior direction. In posterior direction no change in activity is observed. When perturbing at the shoulder girdle posteriorly, the hamstrings (L_BF, L_ST) show a significantly high change in activation. The left gluteus maximus is continuously inactive or no change in activity is recorded, respectively.

With the results of the experiment, the change in muscle activity can be predicted considering the simple inverted pendulum model. This could be helpful when designing exoskeletons, e.g. the controller of the robot can be fed by input data extracted from experimental results. Robotic movement can be adapted to sudden perturbations during stance.

7. Discussion

According to the investigation of Hoitz (2014), biarticular muscles respond first to external perturbations. Those results are based on the static part of the experiment. In this study the dynamic effects of monoarticular and biarticular are taken into account. The hypothesis validation (chapter 5.3) has shown, that in almost 80 % (i.e. 43 cases) the predicted values match with the experimental values. In the 11 (~20 %) incorrectly estimated cases, there are no major mispredicts.

Referring to previous Fig. 16, the burst amplitude of the left M. soleus is much higher than of other muscles. In order to get an answer to this question, a deeper look into the EMG values of the affected subject might help (Fig. 17).

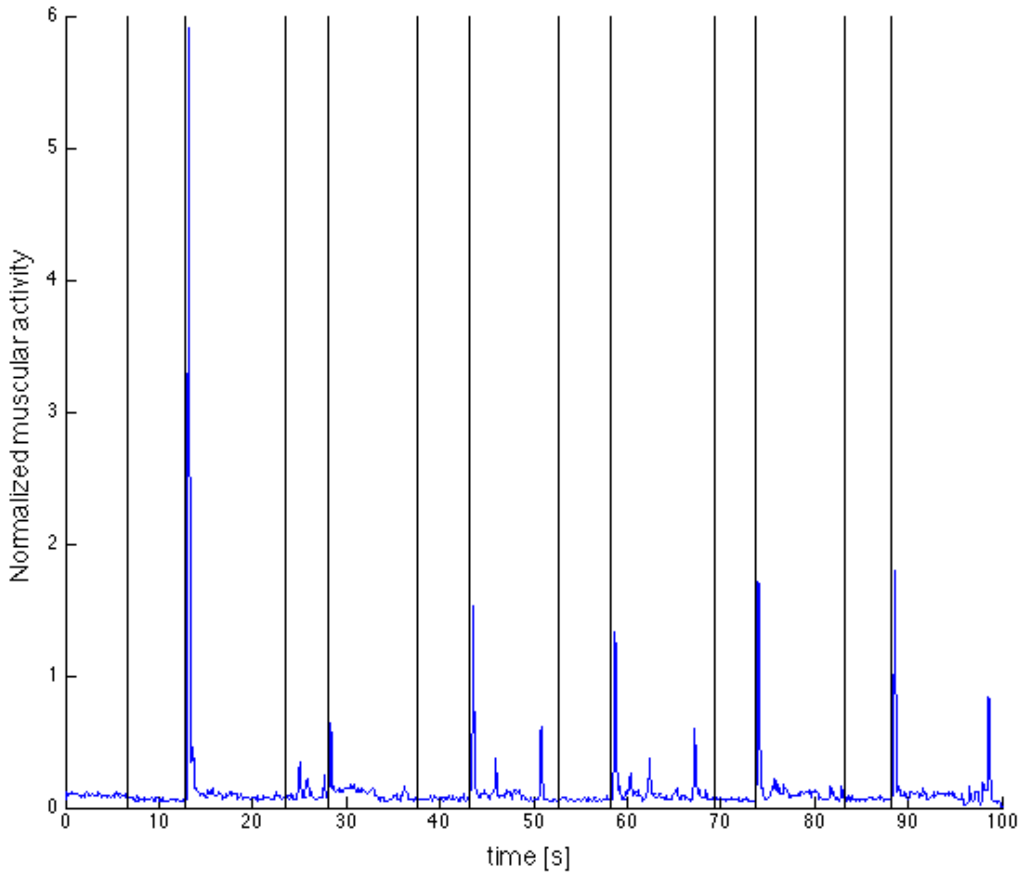

Figure 17: Normalized muscular activity of a subject during trial SPE

In Fig. 17, EMG values above 1 ("bursts") mean, that muscle activation is increased beyond the maximum voluntarily deployable activation value. Thus, the value 1 is

the maximum voluntary contraction considering a single subject. There is a particular reason why the activation values result in such burst amplitudes, which exceed the *MVC* value. When a muscle is activated, all motor units are addressed at once. The muscle has to get sorted before it can do work. The EMG signal is stochastic in nature, this is due to the fact that the proportion of activated motor units changes constantly. The remaining question is why the first peak is much higher than the other 5 bursts. A possible explanation could lie in the placement of the electrode. A possible answer could be that at the time of the first perturbation many motor units are recruited next to the electrode, so the amplitude peak is greater, than in following perturbations. In addition, EMG signals are usually measured within the range of a few in millivolts. Therefore, special care should be taken with the gaining factor, in order not to disrupt the signal.
In any case, the change in muscle activity is very small, since not always the recruited motor units are activated close to the surface electrode. At a change of more than 2% in activation, statements can be made about the activational behaviour of certain muscles. A change in muscle activation of more than 5% is considered as a major change. The calculation of muscle activity after each perturbation can be looked up in chapter 4.4.

From the results of this study it is not apparent, what is the difference between monoarticular and biarticular muscles. It is proved that various muscles, that go over the same joint, work synergistically. It is also apparent from this study, which muscles are involved during a certain perturbation. To find out, if biarticular muscles respond first to horizontal perturbations, the time window after each perturbation must be analysed into detail. Due to a certain inconsistency of the EMG signal, difficulties turned out that make a distinction between monoarticular and biarticular muscles hardly possible. Other studies (Van Ingen Schenau, 1994, p. 495) demonstrated, that monoarticular muscles appear to show simple flexor or extensor activation patterns closely related to the required joint displacements. Furthermore, biarticular muscles often exhibit complex multiple bursts of activity aimed at a fine-regulation of the distribution of net moments over the joints of the limb. These findings suggest that monoarticular muscles are primarily responsible for the generation of force and work, whereas biarticular muscles control the direction of external forces by regulating the distribution of the net moments across the joints (Van Ingen Schenau, 1994, p. 503).

This work can be seen as a continuation of the work of Hoitz (2014). Hoitz (2014) considered in his work the static behaviour of the corresponding muscles. Within this work a dynamic activation pattern is provided in order to make muscle activation predictable for sudden perturbations.

8. Future Work

As mentioned before, the experiments described in this work are designed to answer more questions than approached in this thesis, e.g. it would be interesting to determine the relationship between perturbations and increased muscle activity. Referring to Fig. 8 in chapter 4.4.1 it can be seen that there is a correlation, but it is unknown, whether this correlation is linear, cubic, or at least non-linear. It is also possible, that the knee angle has an influence on the recruitment of several motor units while applying external forces at the ankle, hip or shoulder. For example, it could be determined whether the involved muscles are stressed more when the subject has flexed or extended knees.

As part of this work no differentiation has been made between the applied force intensities, i.e. 10, 20 and 30 N. One hypothesis might be that force intensity has no influence on the selection of the involved muscles. However, it has an influence on the activational behaviour of individual muscles. In other words, at an increased force intensity the muscle activation amplitude rises. The cause of the increased activation dynamics could lie in the recruitment of several motor units or in an increased firing frequency. To determine the frequency at which individual motor units fire, the Matlab Wavelet toolbox can be used. In addition, wavelets are ideal for the suppression of signal noise. Standardized algorithms for decomposition of EMG signal guarantee an automatic filtering process. With this approach, it might be possible to make statements about the dynamic behaviour of monoarticular and biarticular muscles. Existing works that deal with wavelet analysis in EMG processing were contributed by Phinyomark et al. (2011) and Reaz et al. (2006).

As part of the B.A.L.A.N.C.E experiment forces have been applied as well in vertical direction. The next step would be to evaluate this data and compare the results with the results gained from horizontal perturbation experiments. Thus, the hypothesis can be confirmed that postural control might be decoupled from axial force production (Rode & Seyfarth, 2014). It would be interesting to know what is role of monoarticular and biarticular muscles during synergetic cooperation of a certain movement. Taking tracked kinematic data into account, one can find out how the subject behaves during each trial. During static contraction work of the muscular system the attitude of the subject is to be maintained. In order to verify that the subject does not oppose against gravity with its own body mass, kinemetric data can be used.

Another question would be to find out, if voluntary muscle contraction capacity is a limiting factor in the performance of balance-related tasks. A comparative work is already available from Lin (2002, p. 43). In previous studies, investigators have found significant correlation between the strength of the lower extremity and the performance of functional tasks involving balance control (Ferrucci et al., 1997; Topp, Mikesky, & Thompson, 1998). Finally, it makes sense to extend the simplified single pendulum model, either in a double inverted pendulum or triple inverted pendulum model. Of course, corresponding predictions for first and secondary effects in the back swing phase must be made.

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