DESIGN AND EVALUATION OF ARTIFICIAL CONTROLLERS ASSISTING VOLUNTARY BALANCE PERFORMANCE IN PARAPLEGIA AND IN STROKE

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Abstract

The mobility impairment caused by a paralysis like a spinal cord injury or a stroke has, beside many other impacts, an influence on the transfer of signals between the muscles of the lower extremities and the brain. In a paraplegic person, this means that she or he can not stand without holding onto a support or standing in a standing frame while the impact on the ability to balance in a hemiplegic person can be less severe. Although the connection between the muscles and the brain is impaired by the injury, the muscles still retain the ability to contract if innervated.

This thesis describes control approaches which combine the remaining voluntary control of the paraplegic and stroke patients with the artificially controlled stimulation of the muscles of the paralysed limbs to aid the subject in balancing.

The aim was to develop new control approaches which would assist balance in paraplegic subjects and in stroke. To support standing in paraplegic subjects, the moment generated at the ankle using electrical stimulation of the shank muscles was integrated with the voluntary control of the upper body, resulting in the concept of **I**ntegrated **V**oluntary **C**ontrol (**IVC**). In the outer loop the ankle moment produced by the paraplegic subject due to his voluntary upper body movement was estimated using a mathematical model based on the inclination angles of upper and lower body. This estimated ankle moment was then compared with the actual moment applied at the force plates the subject was standing on, and an appropriate stimulation signal was applied to the paralysed shank muscles. Experimental evaluation initially involved four able bodied volunteers in which base line results with stiffness and stiffness-viscosity controllers using a rotating standing platform were obtained. This was extended to the paraplegic subject, where electrical muscle stimulation was used to generate the required ankle moment. The IVC concept was then evaluated with the paraplegic subject and compared to the base line results.

Due to the nature of the system and implied perturbation onto the control system controlling the posture of the paraplegic subject the known evaluation values (e.g. rise

time, steady state value, overshoot value etc.) are not suitable. Therefore, the variance of a time signal around its mean value was used as an evaluation value which allowed to compare the achieved performance of the paraplegic subject employing the new control approach with the stiffness and stiffness-viscosity controllers directly.

To assist balance in stroke patients, a new training approach was introduced based on the concept of integrating the voluntary abilities of the patient with mechanical balance support and sensory electrical stimulation. This concept was evaluated in a training program with one stroke subject which demonstrated the feasibility and potential balance improvement resulting from this approach.

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As in any big project the contribution and support of many people helped in completing this thesis. As there would be not enough space to name everybody I would like to mention only a few.

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Contents

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List of Abbreviations

ASIA	American Spinal Injury Association
ARX	autoregressive with exogenous inputs
ARMAX	Autoregressive-moving-average model with exogenous inputs
CCS	Cooperative Control Synthesis
CNS	Central Nervous System
CoM	Centre of Mass
CoP	Centre of Pressure
CP	characteristic polynomial
FES	Functional Electrical Stimulation
GRF	Ground Reaction Force
MRF	Multipurpose Rehabilitation Frame
OP	Operating Point
PRBS	Pseudo Random Binary Sequence
SCI	Spinal Cord Injury
SOL	soleus muscle
TA	tibialis anterior muscle
TENS	Transcutaneous Electrical Nerve Stimulation
TFL	tensor fascia latae muscle
VAS	vastus muscle

1. Introduction

In a person's development standing up and trying to keep the balance happens very often before the first words are spoken. During these early stages of personal development the task of keeping the balance takes all the concentration and usually the child is not able to do anything else other than trying to stay on the feet without falling over. But soon after the child has learned to keep the balance during standing and starts then to walk and to run this balancing task begins to become more and more subconscious so that one's concentration can be focused on the more complicated things we perform during standing.

Beside the possibility of having a wider range of being able to reach for things placed at some distance from the ground research has shown that standing plays an important role for several physiological processes as e.g. digestion, control of blood pressure etc.

1.1. Aim and Objectives

The focus of this thesis is to develop a new control approach controlling the posture in standing of paraplegics by combining the voluntary movement of the subject's upper body with an artificial posture control of the paralysed lower body by the means of **F**unctional **E**lectrical **S**timulation (**FES**). With this approach the objective is to let the subject work more with his/her upper body to counterbalance disturbances so less support of the artificial control is needed and to compare the results with measurements achieved with already known controllers to see any improvements in performance using this new approach. This concept is termed **I**ntegrated **V**oluntary **C**ontrol (**IVC**).

Another target of this thesis is to transfer the approach of combining voluntary and artificial control developed with paraplegic subjects to the use with stroke patients. The focus is here on sensory stimulation, targeting the hemiplegic subject's sensory pathways and encouraging the patient to use the appropriate muscle groups for certain perturbations applied during the training of balance. To verify the use of this approach the measured results are compared to measurements achieved with an approach of balance training without stimulation.

1.2. Thesis Outline

Chapter 2:

This chapter gives background information on the physiology of the spinal cord and the resulting impacts on the human body caused by a spinal cord injury. Furthermore, a very short overview of the causes for stroke with an overview of the applied rehabilitation approaches are given.

Chapter 3:

Chapter 3 explains the respective equipment used for the experiments with the paraplegic and hemiplegic subject and their controllers.

Chapter 4:

In order to being able to compare the results achieved with the new control approach with the measurements obtained with the known controllers an alternative evaluation method is used. Some background information of different evaluation methods to characterise the performance during standing is given. This leads to the motivation for the use of the alternative evaluation. The informational value of this method is shown by the evaluation of the standing performance of four healthy and one paraplegic subject using a stiffness and stiffness-viscosity feedback control approach. While the required ankle moment in healthy subjects was produced using a rotating platform, FES of the shank muscles was used in the paraplegic subject.

Chapter 5:

Chapter 5 develops the new control approach which incorporates the voluntary movement of the subject's upper body in combination with the artificial control controlling the moment at the ankle by the means of functional electrical stimulation. Experiments are conducted with one paraplegic subject employing this new control approach. The results are then compared with the performance achieved with stiffness and stiffness-viscosity controllers using the evaluation method explained in chapter 4.

Chapter 6:

Chapter 6 presents a balance training approach for stroke patients, combining sensory stimulation of the shank muscles with mechanical balance support to develop voluntary function through a training programme. First tests are conducted without stimulation and after these series sensory stimulation is added. In order to see the change in behaviour these results are compared using known evaluation values.

Chapter 7:

This last chapter concludes the results and gives an outlook on possible further objectives which might be worthwhile to be followed up.

1.3. Thesis Contribution

The contribution of this thesis consists of the following:

- Development of a new control approach which takes the voluntary movement of the upper body of a paraplegic subject into account for the control of the moment at the ankle. This approach is called **I**ntegrated **V**oluntary **C**ontrol (IVC).
- Development of a new balance training approach for the use with stroke patients.
- Development of an evaluation value which allows to compare control performances of different controllers
- Validation of this evaluation value using different stiffness and stiffness-viscosity feedback controllers which control the moment at the ankle of a paraplegic subject using **F**unctional **E**lectrical **S**timulation (FES).
- Comparing the IVC control approach with the stiffness and stiffness-viscosity feedback controllers using the newly developed evaluation value.

1.4. Publications

The results of chapter 6 were published with the Journal of Medical and Biological Engineering (JMBE) [1] and parts of it were presented at the 28th Annual International Conference of the IEEE Engineering in Medicine and Biology Society [2].

2. Background

For able-bodied people standing seems to be an easy task to perform as they are able to carry out different activities during standing without thinking about maintaining their balance. One only becomes aware of the importance and value of this ability when it is restricted due to illness or injury.

This introductory chapter gives a basic overview of the anatomy of the spine and explains some of the difficulties paraplegics face due to the injury of the spinal cord. Furthermore, the principles of functional electrical stimulation are described. As there were experiments performed with a stroke patient, some background information about the illness is presented as well, explaining some of the difficulties the stroke population faces in their everyday life. Finally, a concise literature review of the related research fields is given.

2.1. Spinal Cord Injury (SCI)

2.1.1. Physiology

The spinal cord forms the neural connection between the brain and the rest of the body allowing control signals being sent from and receptive signals sent to the brain. The spinal cord is protected by the vertebral column which is a flexible structure built out of 26 bones (see Figure 2.4 on page 8) grouped into

- 7 cervical vertebrae in the neck with 8 pairs of cervical nerves, abbreviated with the letter **C**,
- 12 thoracic vertebrae (**T**) that articulate with the 12 pairs of ribs,
- 5 lumbar vertebrae (**L**) of the lower back,
- 1 sacrum which is actually a fusion of 5 sacral vertebrae (**S**) (the fusion occurs from the age of late teens to early 20s), and

- 1 coccyx or “tailbone" which is a fusion of 4 coccygeal vertebrae.

The vertebral column bears the weight of the head, neck and trunk and transfers the weight onto the legs. Each vertebra has a vertebral arch which forms the vertebral foramen. All vertebral foramina together form the vertebral canal which encloses the spinal cord. Discs between the vertebrae act as shock absorbers.

The spinal cord consists of two substances, the grey and the white matter. The grey matter lies in the central part of the spinal cord and has, in cross section, a butterfly shape which varies at different levels. The white matter is organised around the grey matter as shown in figure 2.1.

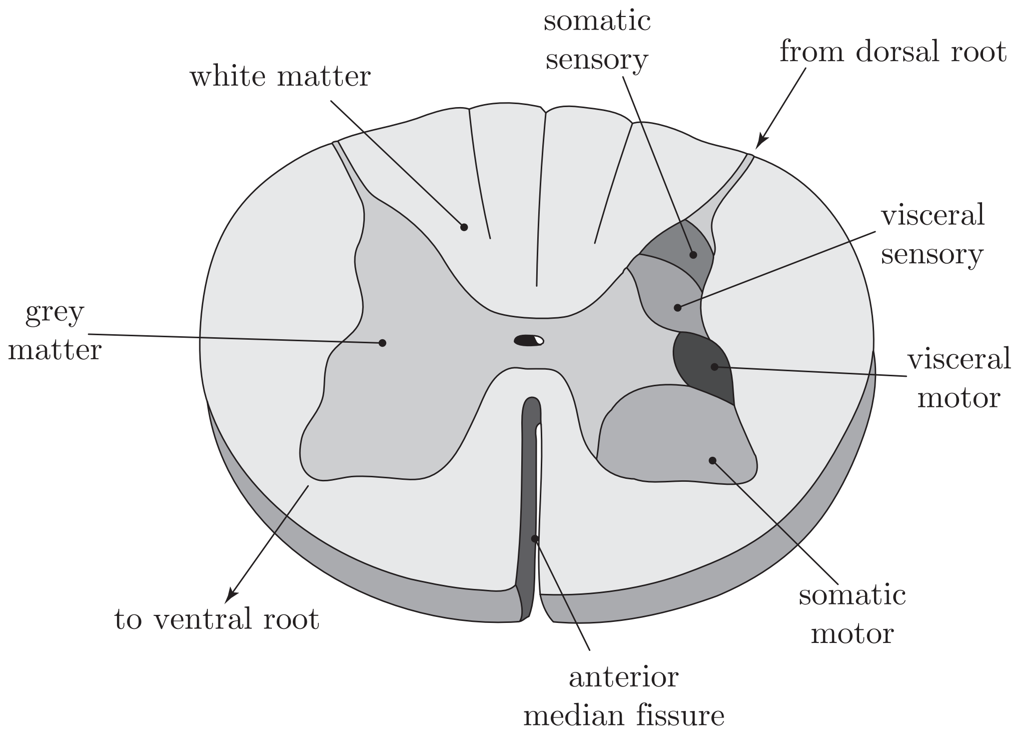


Figure 2.1.: The cross section of the spinal cord and its sectional organisation. *Adapted from [3].*

The white matter contains nerve bundles which are organised into ascending tracts carrying sensory information *to* the brain (*sensory pathway*), descending tracts carrying motor signals *from* the brain to the muscles (*motor pathway*), and short tracts carrying sensory or motor signals between segments of the spinal cord.

The grey matter contains the cell bodies of neurones which are organised into functional groups or nuclei. The sensory nuclei receive information from receptors in the body which is sent to the brain while the motor nuclei pass the information, coming from the brain, on to peripheral effectors (see figure 2.2).

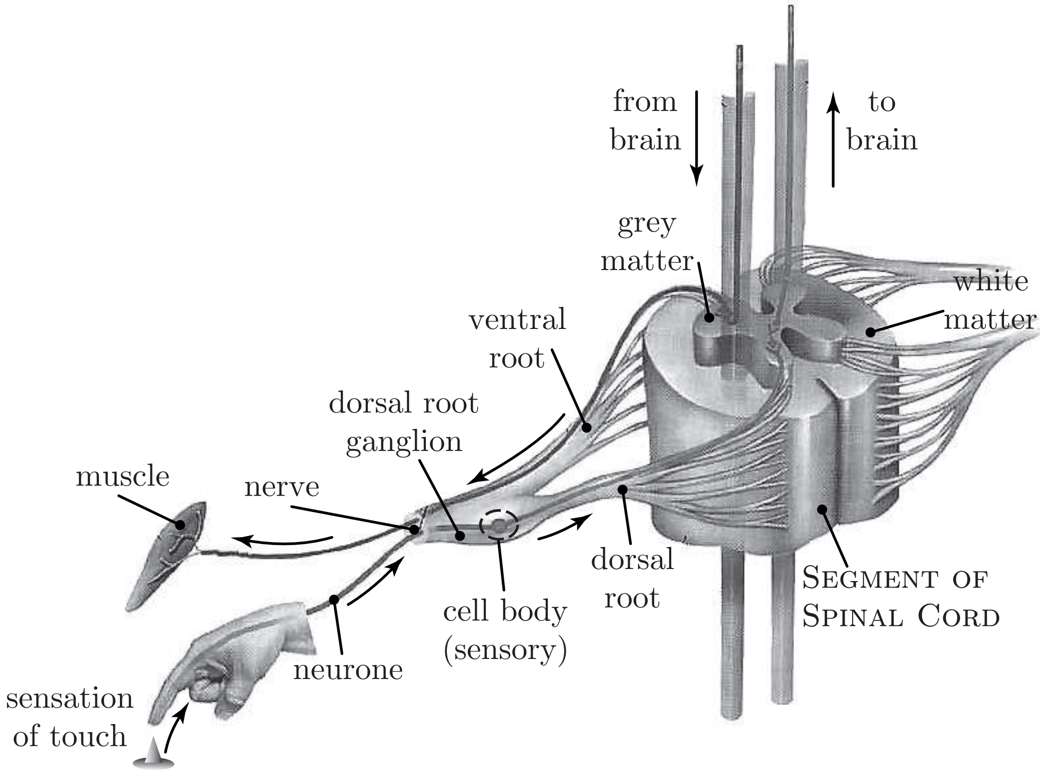


Figure 2.2.: Segment of the spinal cord and the information flow in the case of excitation. *Adapted from http://www.mascip.co.uk.*

Besides the sensory and motor pathways, the spinal cord also contains neural circuits which are mainly involved in reflexes and which can be activated without an input from the brain. The key feature of a reflex is that a particular stimulus leads to a fixed response which is very rapid and can not be controlled because the signal processing does not happen in the brain. These neural circuits are, if they remained intact after the injury, responsible for the spasms paralysed people often have.

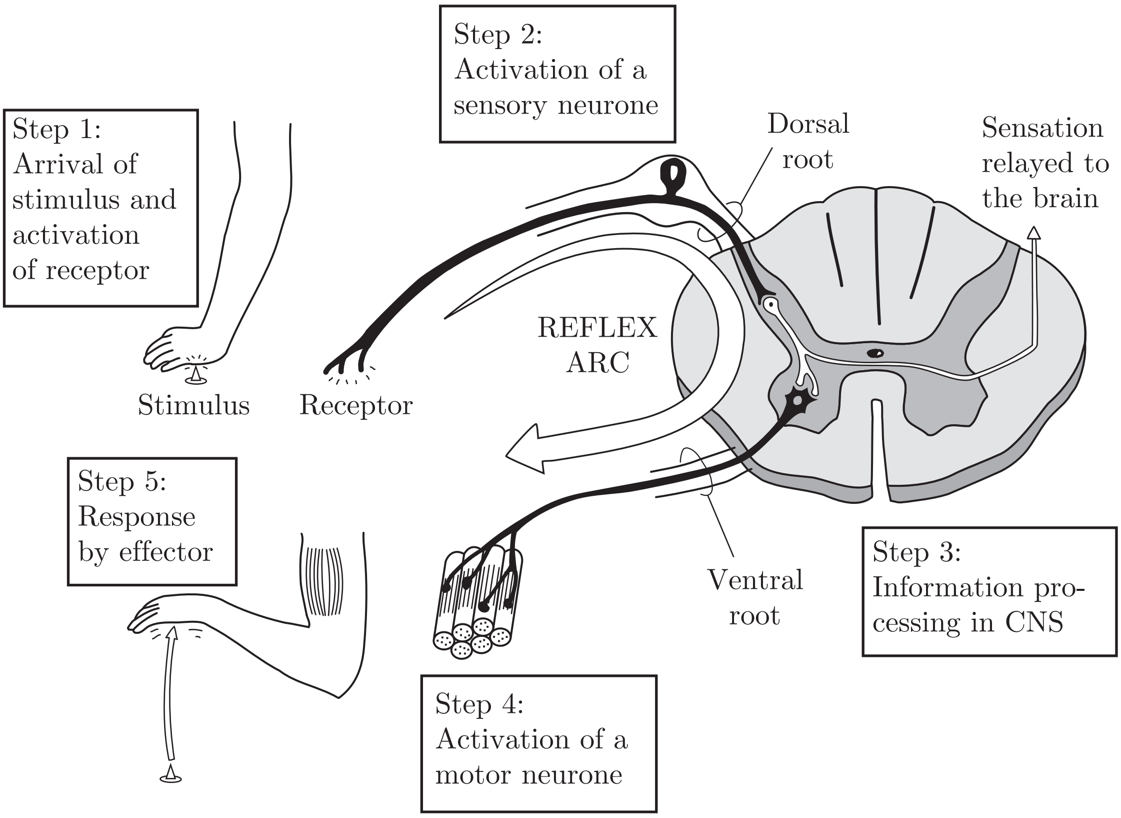


Figure 2.3.: Diagram of a reflex. *Adapted from [3].*

Figure 2.3 shows a schematic diagram explaining the reflex triggered by a stimulus. Applying, for example, a painful stimulus to a part of the body leads to the activation of pain receptors which respond to the stimuli that cause or accompany tissue damage (see “Step 1" in figure 2.3). The activation of receptors leads to an activation of sensory neurones (“Step 2") which deliver the information to the spinal cord. There the information is processed (“Step 3") and the appropriate motor neurones are activated (“Step 4") which carry the signal into the periphery. Finally, a peripheral effector responds to the stimulus, e.g. contraction of a muscle (“Step 5"). As figure 2.3 clearly shows, the brain is not involved in the processing of the information and therefore reflexes are fast, automatic responses to specific stimuli.

2.1.2. The Impact of Spinal Cord Injury

An injury of the spinal cord results in an interruption of the neurological pathways between the brain and the body parts innervated below the injury. Depending on the severity of injury, patients may retain sensation and/or mobility of the affected body parts. The dermatome chart shown in figure 2.4 depicts the different areas of skin and their associated nerves that branch off the spinal cord at each vertebra to illustrate the sensory effects of an injury.

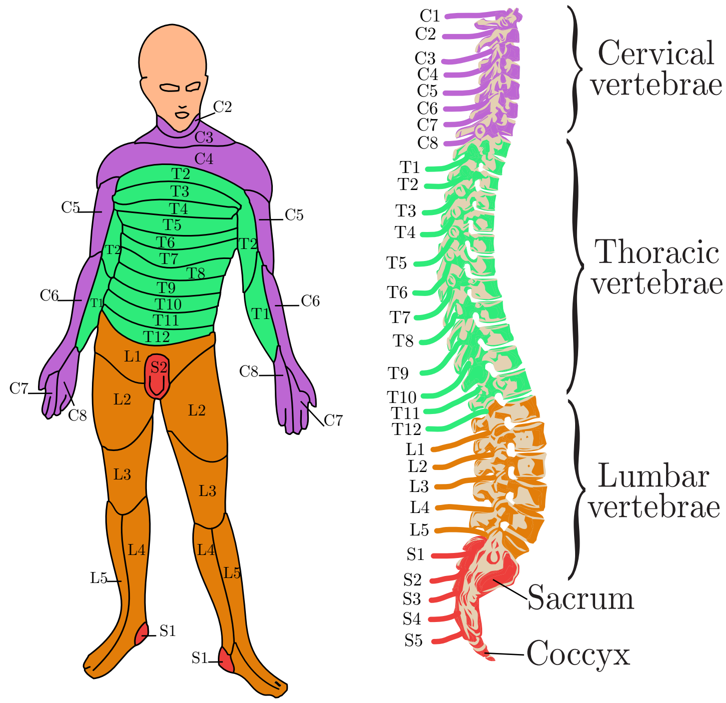


Figure 2.4.: The dermatomes of each spinal segment are located on particular regions of the body.

In most cases of spinal cord injury above the level of the lumbar vertebrae L1, the ability to use the leg muscles voluntarily is impaired or absent. As a consequence, paralysed people face several physiological problems [4].

2.1.3. Secondary complications of SCI

The human body is an adjustable system which adopts to loading of the cardiovascular and muscular system over a period of time. Due to the fact that the lower limbs are no longer used to the extent they were before a spinal cord injury, the skeletal and muscular systems adjust to this situation by reducing their bulk. Major bone loss occurs during the first 6 months after SCI and stabilises between 12 ± 16 months at two thirds of the original bone mass [5]. As a consequence, the bones are vulnerable to fracture [6, 7].

Because the leg muscles, comprising the biggest muscle groups in the body, are immobilised, spinal cord injured people usually have poor cardiopulmonary fitness. As another side effect, these people have a low energy expenditure which entails the risk of obesity [8, 9]. There are exercise recommendations for spinal cord injured people which help to regain fitness [10]. The suggestions range from different sports and activities for people who have retained upper extremity function to activities employing FES of the paralysed limbs (e.g. FES cycling [11]).

Besides the impact on the musculoskeletal and the cardiovascular systems, urinary tract infections and renal complications are common amongst spinal cord injured people [12, 13]. Another major physiological problem spinal cord injured people face is bowel dysfunction [14] and consequently haemorrhoidal problems [15, 16].

In patients with a lesion above the T5 level, the autonomic nervous system's regulation of the autonomous functions of the heart, digestive system, liver, gallbladder, lungs, urinary tract, and genital system is affected. Additionally, these patients face a disorder in the regulation of blood pressure. As a result, sudden and life threatening hypertension can occur. The triggers for this autonomic dysreflexia are stimuli originating from below the level of lesion, e.g. the abdominal or pelvic viscera, skeletal muscles or, most commonly, distension of the bladder. Sometimes it occurs in combination with bradycardia and cardiac dysrhythmia. As a result of the elevated blood pressure, affected people reported headaches, muscle spasms, paresthesia, shivering, pallor followed by flushing of the face, sweating in areas above and around the lesion, nasal obstruction, desire to void, anxiety, malaise and nausea and other symptoms. After removal of the stimulus the elevated blood pressure and the symptoms decline [17, 18].

Patients with a high injury (cervical region) face breathing problems as the intercostal and abdominal muscles are paralysed. Because of these breathing difficulties, quadriplegics are vulnerable to pneumonia and other lung diseases [19]. The recovery from these lung diseases is complicated and takes a long time as among other reasons the ability to cough and clear the respiratory tracts from secretions is also affected [20, 21]. Injury at the level of the cervical vertebrae C1-C2 results in ventilator dependency.

As a result of poor blood supply caused by unrelieved pressure, particularly over bony prominences, pressure sores evolve. Not only the skin but also the deeper tissue can be affected leading to serious health problems which can lead to fatal infections [4, 22].

In addition to all the complications mentioned, many spinal cord injured people suffer from chronic pain which has a major impact on quality of life [23–25].

Besides all these physical problems, spinal cord injured people face psychological problems like mood swings and depression as well [26]. To reduce the risk of depression Boekamp *et al.* [27] suggest the fostering of independence to encourage them to develop new sources of self-esteem and involve relatives maintaining a supportive relationship within the family.

It has been shown that passive standing can help to address many of the mentioned problems [28]. In section 2.4.2 a more detailed explanation of the benefits spinal cord injured people can gain from standing is given.

2.2. Functional Electrical Stimulation (FES)

2.2.1. Principles

The first known experiments using electrical current for stimulation were carried out by Luigi Galvani. In 1786 he obtained muscular contraction in a frog by touching its nerves with a pair of scissors during an electrical storm. He did various other experiments and published his findings in 1791 in his essay *De Viribus Electricitatis in Motu Musculari Commentarius* (Commentary on the Effect of Electricity on Muscular Motion) [29]. Respective reactions Galvani observed also with human muscle.

Due to an electric current pulse travelling from one electrode to another, an action potential in the innervated nerves is invoked causing the muscle to respond with a twitch. Consequently, rapidly repeated stimulation pulses allow the muscle to contract and by controlling the stimulation in terms of duration and intensity muscle function can be achieved. The use of electrical stimulation to allow functional muscle contraction is called **F**unctional **E**lectrical **S**timulation (FES) (see figure 2.5).

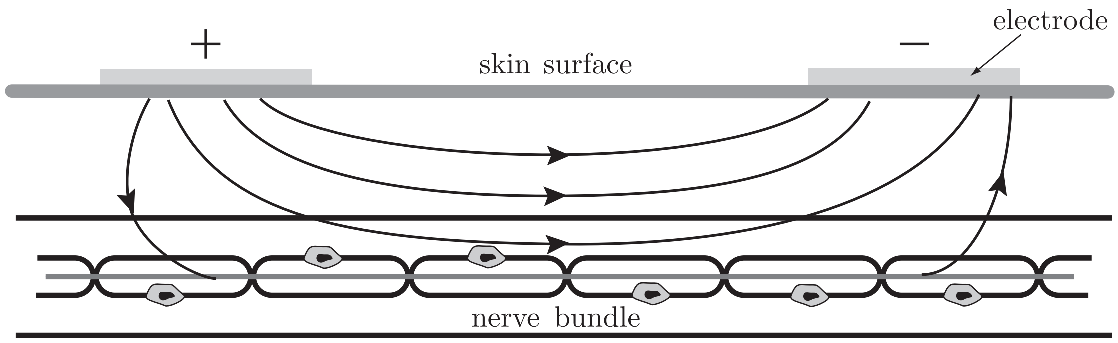
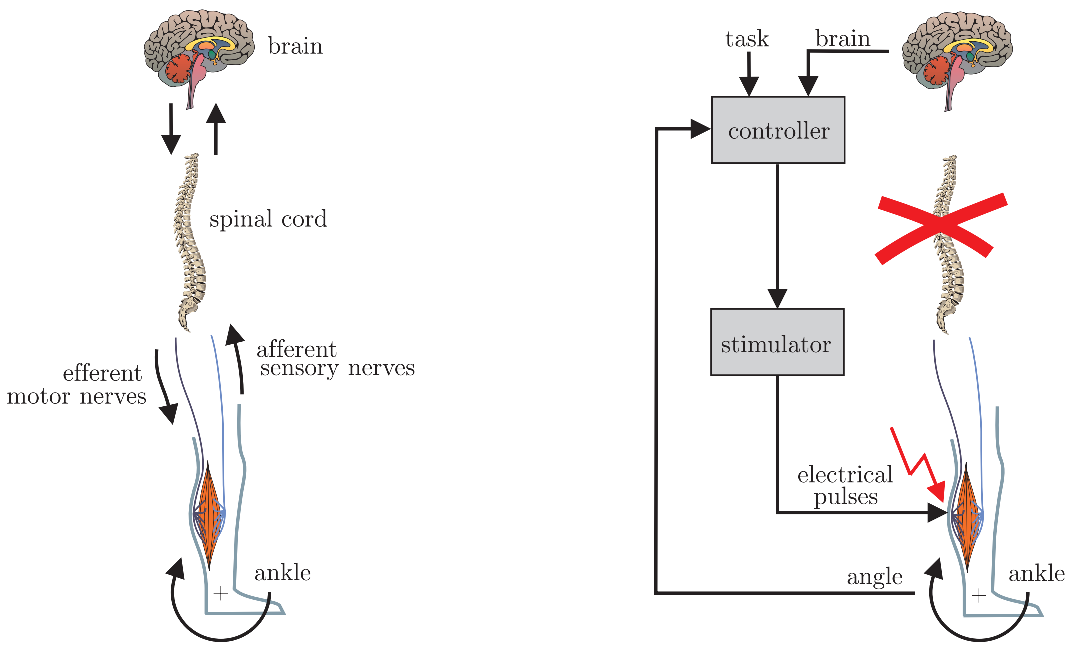


Figure 2.5.: Representation of current density in the tissue using surface electrodes. *Adapted from [30].*

Teeter *et al.* mention 18 different applications where FES has been used in rehabilitation with spinal cord injury [31]. FES has been reported to cause an increase in muscle mass and circulation and changes in muscle fibre composition [32]. Electrically stimulated cycling has been observed to improve fitness, lower-extremity circulation, and circulatory response to ischemia [33–35] as well as an increase in bone density [36]. Baldi *et al.* [37] were able to prevent muscle atrophy in acute spinal cord injured patients using FES.

In the case of a spinal cord injury the interrupted connection to the paralysed parts of the body can be bridged by the controlled use of FES as depicted in figure 2.6. In chapter 5 a new approach to control the amount of stimulation is explained.



(a) Motor function control in an intact individual. Nerve signals from and to the brain are sent via the spinal cord.

(b) Motor function control in a paraplegic using FES. The injury in the spinal cord is bridged by a controlling system.

Figure 2.6.: The principle of FES to restore motor function control in paraplegia with the example of plantarflexor stimulation. *Adapted from [38].*

2.2.2. Muscle fatigue due to functional stimulation

A typical muscle contains hundreds or even thousands of fibres that are arranged into functional groups called motor units. The individual fibres that make up a motor unit are innervated by a single α-motor neuron and vary in size as well as in the type of constituent muscle fibres. As a general rule, small motor units have small diameter axons, typically innervating muscle fibres that are of a slow phenotype, are fewer in number and are relatively fatigue resistant. In contrast, large motor units have large diameter axons and typically innervate larger numbers of faster muscle fibres that are more fatigable. The size principle of voluntary motor unit recruitment describes the progressive recruitment of small, typically slow motor units followed in order of increasing size to the larger, typically fast motor units [39]. Bickel *et al.* state that transcutaneous electrical stimulation activates these motor units in a different way than during a voluntary induced contraction [39]. Stimulation induced contractions are influenced by the placement of electrodes and stimulation parameters resulting in a reversal of the size principle, thus recruiting larger (fast) motor units prior to the slow ones. Bickel *et al.* believe this theory is based on two commonly agreed-upon findings: (1) axons of the larger motor units have a lower excitability threshold, and (2) data demonstrate increased fatigue with stimulation- versus voluntary- activation [39].

Furthermore, the rate of fatigue is dependent on the frequency of stimulation pulses applied to the muscle.

Jones *et al.* [40] carried out experiments using a stimulation signal with a frequency of 20 Hz and 80 Hz for 60 seconds, respectively. The authors observed with continuous stimulation at 80 Hz that the force decreased to about 80% of its initial value in the first 12 s, and then more rapidly, generally reaching less than 20% after 60 s. The maximum force generated by stimulating the nerve at 20 Hz was about 70% of the value achieved with the 80 Hz stimulation. At 20 Hz the force was maintained, or somewhat increased for the first 20 to 30 s and then gradually declined. After about 20 s more force was obtained when stimulating at the lower frequency. For contraction times longer than 35 s, the area under the force-time curve was greater when the muscle was stimulated continuously at 20 Hz than when stimulated at 80 Hz [40].

During intermittent stimulation, however, Matsunaga *et al.* [41] observed that low frequency stimulation signals fatigue the muscles quicker than high frequency stimulation. The authors used stimulation signals with a frequency of 20 Hz and 100 Hz stimulating the right quadriceps of healthy and paralysed subjects at certain periods of time for a few seconds during a 60 minute trial. The authors found that at the end of the 60 minute trial the achieved force was around 26% smaller than at the begin of the trial with a stimulation frequency of 20 Hz. When using the 100 Hz stimulation frequency the force was reduced only for about 18% and was significantly different from the results achieved with a stimulation signal frequency of 20 Hz. The authors suggested to use the intermittent stimulation with a high signal frequency for applications involving closed-loop control strategies for FES [41].

Eser *et al.* [42] applied intermittent stimulation during controlled FES-cycling using stimulation signals of 30, 50, and 60 Hz. The authors found significant differences in power output between measurements achieved with the stimulation frequency of 30 and 50 Hz as well as between the results achieved with 30 and 60 Hz.

With a stimulation frequency of 50 Hz the paralysed subjects achieved a 19.1% higher mean power output during 30 minutes cycling sessions if compared to the power output achieved with a stimulation signal frequency of 30 Hz. Results using 60 Hz stimulation signal show even an increase in power of 24.9% compared to the results applying the 30 Hz stimulation. These findings confirm the results of Matsunaga *et al.*

2.3. Stroke

2.3.1. Background

Although stroke mortality rates for men and women have decreased to about a third of the level they were in 1968 [43] it is still the most frequent cause of severe adult disability and the third most common cause of death in Scotland [44]. About 77% of all deaths caused by stroke occur within the population aged older than 65 years [43]. Of all people diagnosed with stroke approximately 80% survive the first month of injury. About 50% of all surviving stroke patients recover to the extent that they become fully independent [44].

A stroke occurs when an area in the brain is deprived of its blood supply and this area can not be supplied with the necessary oxygen and nutrients. The most common cause of strokes is a blood clot that closes off a blood vessel, preventing the blood from flowing freely. Such a clot could form either within a blood vessel of the brain (thrombosis) or build somewhere else in the body then travel to the brain to cause damage there (embolism). A further reason for a stroke could be arteriosclerosis where fatty deposits build up on the walls of the blood vessels and result in a reduced blood flow or in the worst case closing of the blood flow completely. Another cause can be high blood pressure causing the walls of blood vessels to weaken and eventually to break.

The severity of impact of a stroke depends on which part of the brain is cut off from the blood supply and how big the affected area is. Usually, the afflicted people face a paralysis of different body parts. If the left side of the brain is injured the right side of the body will be affected and vice versa. The paralysis caused by a stroke is called "hemiplegia". Due to the complexity of the brain an injury can lead to numerous different symptoms and consequences.

Risk factors of stroke

There are several risk factors which distinguish persons at high risk from persons at low risk for cardiovascular diseases. These are of various kinds: atherogenic personal attributes (e.g. serum cholesterol level, blood pressure, glucose intolerance); living habits that promote these traits (inactivity, overeating) or carry a direct cardiovascular risk like smoking; signs of preclinical cardiovascular disease (electrocardiographic abnormalities, cardiac enlargement or other evidence of impaired cardiac function) [45].

To assess the risk of a cardiovascular disease the most useful single factor for detecting persons at high risk of cardiovascular disease would be blood pressure, since a large body of evidence indicates that it is the most potent antecedent to the cardiovascular diseases, although it is not equally important for all of these diseases [45]. If using other single factors the efficacy of such an approach would vary with the characteristic chosen and the cardiovascular disease for which risk was being assessed.

Therefore, Kannel *et al.* developed a formula which indicates the probability of a cardiovascular disease to be manifested in a specified time period given a set of variables x_1, x_2,..., x_i measured at the first examination with the variables x_i being the risk factors mentioned above [45]. With this risk function Kannel *et al.* developed a tool which allowed to identify persons at high risk so as to focus attention on them and to avoid needlessly alarming persons at lower risk, because of a single stigma.

Use of stimulation in stroke patients

Clinical studies in patients with stroke revealed improvement in certain tasks while stimulating the respective body parts during rehabilitation exercises [46].

Some success has been reported using cyclic neuromuscular electrical stimulation and EMG-triggered neuromuscular stimulation of the wrist and finger extensors in patients with hemiparetic stroke. Stimulation triggered by reaching a minimal EMG threshold during voluntary wrist extension has, in small trials, increased the amplitude of the voluntary EMG signal of the affected hand, when coupled to bilateral movements compared with unilateral ones, and when coupled to bilateral movement-induced stimulation compared with unilateral movement-stimulation. Mechanisms include muscle strengthening from stimulation, sensory feedback to augment sensorimotor integration for the task at spinal and supraspinal levels, and perhaps an augmented cortical sensory drive for activity-dependent plasticity. Unlike most of the trials of EMG biofeedback and EMG-triggered stimulation of the 1980s and 1990s, these small studies reveal suggestive functional gains in use of the affected hand in a selected population [46].

Johannson *et al.* [47] carried out experiments using acupuncture combined with electrical stimulation. Acupuncture treatment was started 4 to 10 days (mean, 6.5) after stroke onset and continued twice a week for 10 weeks. Acupuncture was given on the paretic as well as the nonparetic side. Traditional Chinese acupuncture points were used, and a total of 10 needles were kept in place for 30 minutes each time. In addition to manual stimulation, electrical stimuli with a frequency of 2 to 5 Hz were given to four needles on the paretic side. The intensity of stimulation was such that a muscle contraction was obtained. The acupuncture and control groups received standard individual stroke rehabilitation treatment including daily physiotherapy and occupational therapy [47].

At the end of the study, i.e., 1 year after stroke onset, 18 of the 78 patients had died. Ten of the patients (five in each group) died from cardiac disease; two in each group from recurrent stroke. Pulmonary edema, pneumonia, intestinal gangrene, and intestinal hemorrhage were the cause of death in the remaining four patients. A significantly larger number of patients treated with acupuncture were living at home 3 months after stroke onset, and the difference remained up to 1 year. The number of days spent at the neurologic wards was the same for the two groups: 18.6 days for controls and 18.8 days for the acupuncture group (for the survivors, the figures were 17.3 and 17.2). However, control patients stayed longer in geriatric rehabilitation units and nursing homes (which in Sweden also have some rehabilitation facilities), making the mean number of days spent at hospitals and nursing homes 165.5 for the control group and 86.5 for those that received acupuncture. If only the survivors are counted, the figures are, respectively, 161.1 and 88.2 days. Patients given acupuncture reported a higher quality of life than the control group. The scores for energy, mobility, emotional reaction, and social isolation were significantly lower-indicating less problems-at 3 and 6 months [47].

2.3.2. Rehabilitation approaches

Due to the hemiplegia caused by stroke the affected patients face an impairment in balance during standing and walking. Therefore, stroke is considered to be one of the greatest risk factors for falls in older adults [48] leading sometimes to devastating complications [44]. To enhance the quality of life in this population group it is essential to employ rehabilitation methods to improve the ability to balance and reduce the risk of falling. Previous studies have shown improvement in functional balance [49, 50] and mobility [51–53] through exercise.

Marigold *et al.* [54] employed two different types of exercise with two groups of subjects to investigate the effects on fall-reduction in stroke in a more general way. The first group was engaged in performing an agility exercise programme challenging dynamic balance. This exercise regime used tasks where the subjects had to stand in various postures (e.g. tandem or feet apart, one foot stance, and weight-shifting) and cope with various challenges while walking (e.g. different step lengths and speeds, tandem walking, figure-eight walking, stepping up and over low risers, side stepping, crossover stepping, and stepping over obstacles). Additionally, the subjects had to perform sit-to-stand movements, rapid knee raises while standing, and react to perturbations during standing. The second group took part in a stretching/weight-shifting exercise programme focussing on slow, low-impact movements consisting of stretching and weight shifting. For the weight-shifting exercises tai chi-like movements and reaching tasks were used to encourage the subject to use the paretic lower limb more. The participants were asked to do the stretching of major muscle groups while standing on mats and on the floor. Moreover, the subjects had to get down on and up from the floor with the support of the instructors. Marigold *et al.* observed reduced falls in the group performing agility exercises compared to the stretching/weight-shifting group and suggest that exercise programmes for chronic stroke patients should include dynamic balance training with emphasis on multisensory and agility tasks.

Matjačić *et al.* [55] carried out a case study with one chronic stroke patient, applying novel dynamic balance training techniques during standing and stepping using a commercially available mechanical balance training device called BalanceTrainer which is commercially available (Medica Medizintechnik GmbH, Hochdorf, Germany). This standing frame secures the person standing in the frame and at the same time allows movement in the sagittal and frontal planes. A more detailed description of this device can be found in section 3.3 on page 37 and in [56]. In this study the subject had to carry out different tasks. Firstly, he was asked to lean in different directions. This was performed in parallel and tandem stances. Secondly, the therapist assisted the subject in carrying out functional tasks by bending the upper body in different directions and performing these actions again in the different stances. Finally, the subject was asked to perform exercises to train the various phases of a step. For this, the subject had to stand in parallel stance and train the push-off phase first. Later, the subject was asked to shift the body weight completely onto the front leg in order to train the swing phase. These different exercises train the hip extensors and abductors that control the posture of pelvis and trunk in the sagittal and frontal planes, respectively. As a result of this training the jerk in the hip moments was reduced, leading to less stress on the musculoskeletal system. Additionally, the subject gained more confidence in walking leading to a reduced fear of falling.

As a conclusion Matjačić *et al.* suggest that the work is continued and that the dynamic balance training be combined with electrical stimulation. This work was carried out in a joint project and is presented in chapter 6.

2.4. Research Context

One focus of this thesis is to combine the voluntary movement of the upper body of a spinal cord injured person with the artificial control of FES applied to the subject's lower limbs. To show the relevance of the research presented in this thesis, an overview of the literature dealing with general issues connected to standing and balancing is firstly given. This will lead to a description of the difficulties paraplegic patients face after their injury and the importance of standing.

Furthermore, a new control strategy is introduced which is designed to improve the performance of balance in spinal cord injured people. In order to be able to compare the newly developed control approach with other control strategies suggested in the literature an appropriate evaluation method is presented as well.

Additionally to the work with a paraplegic subject, results of a joint project involving a chronic stroke patient performing a new developed training regime are presented.

2.4.1. General overview

Standing in itself is a very easy task to fulfill for an able-bodied person although the human body is an unstable mechanical construction, with two-thirds of the body mass located two-thirds of body height above the ground [57]. For the analysis of standing the human body can be modelled as an inverted pendulum having, depending on the required accuracy of the model, one or several links [58–61].

To keep this unstable mechanical construction stable, a control system has to act constantly. In order to understand the way healthy people are able to maintain their balance so easily, researchers mainly studied two different experimental configurations: quiet and perturbed standing.

"Quiet Standing" approach

For several decades researchers have studied the human postural control system with the help of centre of pressure (CoP) trajectories measured during quiet standing. For the analysis of the posturographic data summary statistics were mainly used. In general, this approach did not allow for a physiologically meaningful interpretation. Therefore, Collins and De Luca proposed a different system which would enable interpretation of responses of humans during quiet stance [62]. For the experiments, the subjects stood barefoot in an upright posture and in a standardised stance for a series of five 90 second trials under eyes-open conditions. Between every trial the subject rested for two minutes. With their "stabilogram-diffusion" analysis the researchers could extract CoP parameters which could be directly related to the steady-state behaviour and functional interaction of the neuromuscular mechanisms underlying the maintenance of erect stance. Furthermore, they found that postural sway during quiet standing is indistinguishable from correlated noise and can be modelled as a system of bounded, correlated random walks [63]. From these findings Collins *et al.* stated that over short-term intervals, an open-loop control scheme is acting whereas during long-term intervals, a closed-loop control scheme is used.

Neurophysiologists point out three systems which are involved in balance and posture [64]:

- **vision:** used for planning locomotion and avoiding obstacles,
- **vestibular system:** measuring the linear and angular accelerations,
- **somatosensory system:** a complex of a multitude of sensors detecting the position and velocity of all body segments, their contact with external objects, and orientation in the gravitational field.

A large number of experiments have been devised to find out the contribution of each system in keeping the human body stable. In his paper Winter [57] proceeded in a different way in approaching the question of how human beings maintain stability in stance. He used a mechanical model of an inverted pendulum representing the human body and the centre of pressure as a control variable. To control the human body during quiet stance and small perturbations in the anterior-posterior direction, Winter stated that only the ankle plantar-/dorsiflexor muscles are responsible.

In later work Winter *et al.* [65] developed a control scheme to model balance in quiet standing. The author approximated the body by a 14-segment inverted pendulum model including shanks and feet (2 segments), thighs (2 segments), pelvis (1 segment), lower arms (2 segments), upper arms (2 segments), trunk (4 segments), and head (1 segment) to estimate the total body centre of mass. Moreover, it was assumed that muscles act as springs which cause the centre of pressure and the centre of mass (CoM) to move in phase as the body sways around the desired operating point. In this setting the CoP is controlled in the sagittal plane by ankle torque (plantarflexor/dorsiflexor muscles) and in the frontal plane by the hip abductor/adductor muscles. To validate their control scheme, the authors had ten adults stand quietly at three different stance widths for two minutes in each position with the eyes open. The measurements showed similar results to the predictions of the simulated control scheme. From their findings Winter *et al.* proposed that the restoration torque during quiet standing is set by the joint stiffness and is therefore passive. This work triggered a controversial discussion on whether only the stiffness of the ankle muscles is responsible for the stability of the human body during quiet standing or not.

Morasso *et al.* [66, 67] questioned the statement Winter *et al.* made in [65, 68]. They found that the actual stiffness value of the ankle muscles is far smaller than that theoretically needed. The authors backed up this statement with results of other researchers [69–73] concerning muscle and joint stiffness. Furthermore, Morasso *et al.* quoted other researchers [74–82] who looked into the sensory feedback of muscles even when the stress on the muscle is very small. From this development the authors came to the conclusion that whatever the input(s) to the central nervous system (CNS) might be and the ways it uses the information, there is too much indirect evidence that the stabilisation mechanisms and applied control action are multi-faceted and active rather than purely passive.

Loram *et al.* [83] agree with Morasso *et al.* that intrinsic ankle stiffness is inadequate for providing stability. The authors applied small perturbations similar to the size and velocity of the movements occurring during quiet stance. These perturbations did not cause any stretch reflex responses in the gastrocnemius and soleus muscles. On average, the measured intrinsic stiffness was about 91% of the needed stabilising value and stayed approximately constant. As this stiffness can not be neurally controlled in quiet standing the authors suggest the origin of the stiffness to be in the foot, Achilles' tendon and aponeurosis rather than provided by the activated calf muscles.

"Perturbed Standing" approach

With the "Perturbed Standing" approach a perturbation is applied from outside that puts the subject out of balance. For the realisation of perturbations different devices have been used. Perturbations have been applied mainly in three different ways:

- horizontal displacement of the base the subject is standing on as described in [61, 84–86]
- rotational displacement of the base the subject is standing on as described in [58, 87–89]
- application of forces at hip height using a special mechanical rotating frame as described in [90–94]

As these different perturbation approaches were also used to develop control strategies to control the posture of balance impaired people some of these approaches will be explained in a more detailed way later in section 2.4.3.

Van der Kooij *et al.* compared the methods which have been developed in recent years to identify and quantify balance control in humans [95]. The authors point out that the outcomes of the studies using the approach of "Quiet Standing" are "ambivalent and speculative or trivial". This is because the reported results depend on unmeasured internal disturbances and sensor noise which apply an unknown perturbation. They come to the conclusion that only the "Perturbed Standing" approach in combination with the correct identification method allows the separate identification of balance control and musculo-skeletal dynamics.

2.4.2. Importance of standing in paraplegia

Health benefits through exercise in paraplegic people have been reported in many papers (see e.g. [32, 96, 97]) but there are few investigations about the benefits of standing in spinal cord injury.

Kunkel *et al.* [98] did an investigation on the effect of standing on spasticity, contractures, and osteoporosis in paralysed males. The mean age of the subjects was 49 years and average time since injury was 19 years. To observe the impact on spasticity a clinical assessment was used. The degree of contracture was defined by the joint range of motion of the lower extremities and the change in bone density was assessed with a dual photon absorptiometer. The subjects stood on average 144 hours over a mean of 135 days.

Comparison of the scores for the clinical assessment and joint range of motion between the beginning of the study and the final measurements showed no significant difference. The bone density was measured in the lumbar spine and in the femoral neck. Comparing the results of the first and last assessment showed also no change in bone density but the subjects reported feeling healthier and 67% continued to stand.

From these findings the authors concluded that standing had no impact on the measured variables but it had a positive psychological effect on the participating subjects and no ill effects.

Eng *et al.* [99] presented the results of a survey focussing on the use of prolonged standing for people with spinal cord injury. For the study, the responses of 126 returned mailed survey questionnaires were evaluated. On average, the subjects stood for 40 minutes once a day, 4 days a week using either standing devices like a standing frame or braces in combination with an assisting device such as a walker. In a relatively short time (within a week) the subjects would experience benefits from prolonged standing but which would not last very long (only 1 day). Although some of the respondents reported negative effects from prolonged standing like increased pain, increased fatigue, breathing difficulties, or increased spasticity most of the respondents stated

- an overall improvement in their well-being,
- improved circulation as some observed reduced swellings in legs and feet,
- a reduced reflex activity as some experienced reduced muscle spasms,
- improved bowel and bladder function,
- improvements in self-care, digestion, breathing, and skin integrity,
- better sleep,
- decreased pain, and

- psychological benefits.

2.4.3. Control strategies

In order to make standing for spinal cord injured people more beneficial, the paralysed muscles of the lower limbs should be activated by means of FES. With the stimulation the patient would use his/her own muscles together with their own metabolic energy, the atrophied paralysed muscle would be restrengthened, and the blood flow in muscle and skin increased [28]. This would allow the impaired person to use their arms for functional tasks rather than supporting themselves during standing.

The first results in supported standing were achieved by the application of open loop controllers where a fixed stimulation current and a fixed pulsewidth provided contraction of the stimulated muscle groups [100]. The advantage of this approach is its simplicity since the system is easy to set up. As there are no sensors and advanced control algorithms involved, this method of control is not very flexible towards behavioural system changes like fatigue and other disturbances.

Jaeger was the first to introduce the concept of unsupported standing in paraplegia applying a closed loop control system to control the posture of paraplegics in standing. The idea of unsupported standing is to allow the paraplegic subject to stand without arm support. In his paper [101] he presented a simulation model of the ankle muscles as actuators providing the torque needed to keep the body in an upright position. The body was modelled as a single-link inverted pendulum and with an appropriate PID-controller Jaeger proved that the closed-loop system was stable under small perturbations.

Hunt *et al.* [102] took Jaeger's work a step further and were the first to prove that the stimulated ankle plantar flexors of a paraplegic subject were strong enough to keep the subject in an upright position. For the control, a cascade structure was used with the inner control loop controlling the moment produced by the ankle plantar flexors and an outer loop controlling the inclination angle of the body. This nested structure was chosen because of its robustness to changes in muscle properties and interference from spasticity. For every control loop a linear quadratic Gaussian controller was used. Again, the body was modelled as a single-link inverted pendulum with the rotation axis at the ankle. With this approach, the subject was able to stand for not longer than a minute before the ankle muscles became too fatigued to hold the body in an upright position; the system then became unstable.

In order to prolong the time of FES assisted standing in paraplegia Holderbaum *et al.* [103] therefore suggested an H_∞-controller which is more robust to plant uncertainties as they occur due to nonlinearity of the stimulated muscle. Holderbaum *et al.* modelled the body as a single-link inverted pendulum and used the same nested control loop structure as Hunt *et al.* With the suggested H_∞-controller the authors reported a standing time of 7 and a half minutes before the subject's leg muscles were too fatigued to keep the body in an upright position.

Gollee *et al.* [104] suggested another approach based on the control strategy suggested by Hunt *et al.* [102] in order to make the control of unsupported standing in paraplegia more reliable, consistent, and prolong the duration of standing. As the strength, the rate of fatigue, and spasticity of the paralysed muscles depend on the condition of the subject the author aimed to develop a control approach which would be robust enough to deal with these uncertainties and disturbances. As a result the authors used a nested structure as before [102] but instead of employing a Hammerstein model of the paralysed muscle, as Hunt *et al.* suggested, Gollee *et al.* identified a model of the muscle after performing a series of identification tests. Further the authors designed a controller using the standard pole placement method instead of using a LQG-controller [102]. With this solution the authors could show that the controller performance was consistent and robust under varying conditions.

To make the process of standing more natural Matjačić *et al.* [59] introduced a novel approach for unsupported standing in paraplegia. The idea was to combine the voluntary movement of the subject's upper body and an artificially controlled ankle stiffness. For this the body of the subject was modelled as a double-link inverted pendulum representing the upper body and the legs as the two links. The subject with whom the experiments were carried out stood in a special standing frame called the Multipurpose Rehabilitation Frame (MRF). With the MRF bracing the knee and hip joints mechanically, the subject is prevented from collapsing. With a hydraulic actuator at ankle height acting as an artificial ankle joint the applied stiffness was controlled. For the stiffness control a proportional controller was used allowing a satisfactory tracking performance of the reference torque. With this experimental setup the authors showed that unsupported standing with the integrated voluntary movement of the upper body was feasible.

Jaime *et al.* [105] presented a slightly different approach from Matjačić *et al.* The ankle stiffness was provided by FES of the ankle flexors as Hunt *et al.* suggested in their earlier work [106, 107]. For the control, a nested loop structure was used with the ankle moment controlled in the inner loop and the inclination angle of the lower body in the outer loop. For the design of the inner control loop the pole placement approach was used. With this work Jaime *et al.* showed that unsupported standing in paraplegia with the voluntary use of the upper body was possible.

Mihelj *et al.* [93] proposed a different control approach from Matjačić *et al.* The idea was to use an energy optimal controller in order to keep the body in an upright position with a minimum of energy used by the ankle joint muscles, approximating the body by a double-link inverted pendulum. The control design was based on minimising a cost function which estimated the energy of the muscles acting at the ankle joint by observing the position of the centre of pressure. The upright standing posture with minimal energy cost was achieved by keeping the centre of pressure close to the axis pointing vertically through the ankle joint. The simulation results were compared with the responses of an able-bodied person and showed a reasonable correspondence with natural human behaviour. In later work Mihelj *et al.* [94] applied this approach with a paraplegic subject and were able to keep the disabled person standing in the MRF in a stable posture.

2.4.4. Evaluation of standing

The main challenge which has to be taken into account for the control of paraplegic standing is the relatively rapid development of fatigue of the stimulated muscles. The main evaluation value that gives a clear indication of the quality of control is the duration of time the subject is able to stand before the muscles become too weak to keep the body in a stable position. Several authors defined different criteria allowing assessment of control performance.

Kralj *et al.* [108] introduced a fatigue index in order to give a prediction of functional endurance of stimulated leg muscles during standing of spinal cord injured subjects. The authors defined the ratio of decrease in muscle force during the first 30 seconds versus the initial value as a fatiguing index. For the determination of the fatiguing index the isometric knee joint torque was used. Applying the fatiguing index in experiments with several paraplegic subjects, the authors found it to be poor and not applicable for the prediction of the endurance of muscles.

Popović *et al.* [109] introduced a stability criterion to assess the standing condition of able-bodied people during standing, evaluating the centre of mass and the centre of pressure. The idea behind this criterion is to allow, for example, a spinal cord injured person to stand up, sit down and stand safely using a standing device without hand support. The authors defined three different stability zones using the centre of pressure (CoP) as the evaluation variable. The desired zone is that where the subject is stable and the CoP is located 99% of the time during quite standing. The second zone is the undesired zone where the subject is forced to change posture in order to maintain stability. The unstable third zone is the zone where the subject has to make a step in any direction in order to prevent falling. The authors developed a general model which allowed computation of the stability zone *a priori* and made it possible to assess the subject's stability using the CoP measurements.

Mihelj *et al.* [93] used an energy optimal controller where the designed algorithm satisfied a cost function based on an energy optimal behaviour during standing. For this method, the value of the cost function is used as the evaluation criterion.

2.5. Objectives

All of the outlined methods for evaluation of the performance in standing allow only a statement about the performance in conjunction with the control approach used. None of these evaluation methods can be used to compare them with other completely different control approaches. Therefore, in order to determine which control approach is more beneficial for the disabled subject in standing, a quality measure is needed to evaluate the control performance of the different control strategies. Since no work related to this topic could be found in the literature, a quality measure is developed allowing a clear judgement concerning why one controller is more beneficial than others (see chapter 4). This evaluation approach now allows the direct comparison of different control strategies and their control performance.

Two case studies are carried out applying the evaluation of standing to two different types of patients suffering from neurological disfunction. Both case studies are presenting new approaches to improve the balance of impaired subjects.

As the subject has no motor function in the lower limbs, the first case study deals with the problem of controlling the posture of a paraplegic subject during standing by applying functional electrical stimulation.

Jaime [38] presented an approach where the artificial control of the ankle stiffness by means of FES combined with the free voluntary movement of the subject's upper body would allow a paraplegic subject to be stable during standing. Further, Jaime showed that the ankle torque was essential to keep postural stability as the subject would lose stability if the stimulation would be switched off. For this approach the author regarded the voluntary movement of the subject's upper body as an unknown disturbance and was not considered in the control design. This leads to the problem that the needed amount of stimulation stabilising the subject might be too high and therefore the paralysed muscles fatigue quicker. Therefore, a new control approach is introduced which estimates the ankle torque that the voluntary movement of the subject's upper body would produce (see chapter 5). By subtracting this moment from the commanded ankle moment reference, only a reduced moment needs to be tracked. This will cause a smaller control action requiring a reduced stimulation which will therefore allow prolonged standing in paraplegic subjects. To show the difference in quality to the control Jaime [38] used in his work, the new quality measure will be used.

The second case study deals with improving the balancing skills of a stroke patient. Contrary to the approach taken for the experiments with a paraplegic subject, .the aim is to improve the remaining balancing function in this subject. Therefore, the stroke patient receives only sensory stimulation during perturbed stance (see chapter 6) while responding to perturbations applied to the standing frame.

The subject underwent two different periods of perturbation training, each lasting ten days. During the first period the subject was perturbed in eight different directions. During the second period the subject was also perturbed, but was assisted by sensory electrical stimulation of the soleus (SOL), tibialis anterior (TA), tensor fascia latae (TFL), and vastus muscles (VAS) in the impaired leg. After each period of training, an assessment was carried out to measure the forces the subject applied on the ground via two force plates and the EMG responses of the SOL, TA, TFL, and VAS muscles. The comparison was evaluated with the new developed evaluation method.

3. Experimental Equipment

This chapter gives an overview of the equipment used for the experiments and its functionality, namely the neuromuscular stimulator, the Multipurpose Rehabilitation Frame, and the BalanceTrainer.

3.1. The Neuromuscular Stimulator

For the experiments described in chapters 5 and 6, a stimulator ("Stanmore Stimulator") with eight channels was used. The stimulation is current controlled, monophasic, and charge balanced. The stimulator can be controlled from a computer via an RS232 interface and produces rectangular pulses with a pulsewidth of up to maximal 800 μs adjustable in steps of 2 μs with an accuracy of 0.5 μs. The maximal amplitude of 130 mA can be reached in 10 mA steps. For the experiments a sampling time of 50 ms was used (sampling frequency of 20 Hz) delivering the current to the muscles using round surface electrodes (PALS®) with a diameter of 5 cm. A more detailed description of the "Stanmore Stimulator" can be found in [110].

3.2. The Multipurpose Rehabilitation Frame (MRF)

3.2.1. General Description

As already mentioned in section 2.4.2, standing is very important in rehabilitation for spinal cord injured patients. Until a few years ago there were only static standing frames available. These standing devices secure the balance impaired patient with belts by locking the hip and knee joints which prevent the standing person from collapsing. However, these devices allow only static standing and are not suitable for balance training. Moreover, the use of the static standing frames is not very challenging and the balance impaired person may easily lose the interest in performing standing exercises.

In order to make the standing an active task Matjačić *et al.* [59] presented a standing frame which allows balance impaired subjects to use their upper body to balance actively during standing. The authors designed a standing frame mounted on joints aligned to the rotating axis of the ankles allowing a tilting movement in posterior/anterior direction (see Figure 3.1).

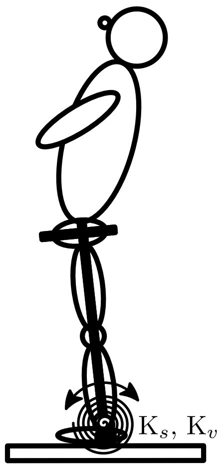


Figure 3.1.: New rotating standing frame with a supporting stiffness (K_s) and viscosity (K_v) produced by a hydraulic system.

To supply the subject with variable support, a hydraulic actuator is connected to the joint providing a certain stiffness (K_s) and viscosity (K_v) which can be adjusted very precisely. This means that the frame acts as an artificial ankle joint. The balance impaired person is secured as the hip and knee joints are locked mechanically. This prevents the subject from collapsing but at the same time movement of the lower body in the sagittal plane is possible as the frame tilts in a mechanically constrained range of $\pm 23°$. This puts the subject in a double-link inverted pendulum configuration with the head, the arms and torso being the upper and the legs the lower link. To study the nature of standing the hydraulic actuator can also be used to apply perturbations.

Matjačić *et al.* [90, 111] did further alterations to this Multipurpose Rehabilitation Frame (MRF) in order to make the balance task more realistic. A second joint and hydraulic actuator were added to support the standing subject with a certain stiffness for movements in the frontal plane. With these new alterations the frame could tilt not only in the sagittal plane but in the frontal plane as well.

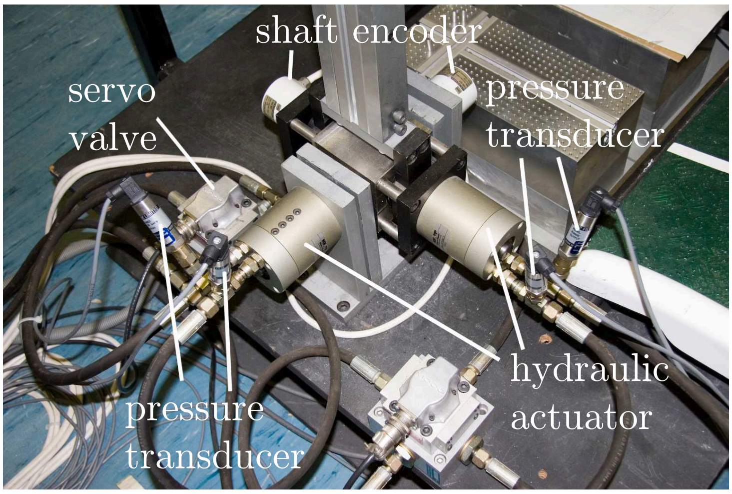


Figure 3.2.: Actively controlled two degree-of-freedom joint.

Figure 3.2 shows the two degree-of-freedom joint, which is actively controlled by two hydraulic actuators and two servo valves. In order to keep the impact of the weight of the frame minimal, aluminium profiles are used for several parts of the MRF. Furthermore, actuators are needed which have a good power/weight relationship, as one actuator will be moved by the other during action. Hydraulic actuators were therefore chosen, as these have a better power/weight relationship than electric motors.

The MRF can be operated in two different modes. In the first operating mode the subject stands on a rotating foot platform which is attached to the frame and immobilises the subject's ankle joint. The hydraulic actuators are now used to control the frame's lower joints. This operating mode can be used with able-bodied subjects simulating the situation of a spinal cord injured person, who is not able to activate the ankle voluntarily. Figure 3.3(a) shows an able-bodied subject standing on a rotating platform. Experimental results using the MRF in this operation mode are presented in chapter 4.

For the second operating mode the subject stands flat on two force plates with the subject's ankle joints providing the required support for balancing. Operating the MRF in this mode with able-bodied people as subjects, the natural mechanism of controlling upright standing and posture can be investigated. Using the MRF in this operating mode with spinal cord injured subjects would allow

1. the training of their balancing skills,

2. the investigation and test of artificial balance control approaches controlling the torque the hydraulic actuators apply onto the frame,

3. the investigation and test of balance control strategies regulating the intensity of FES applied to the shank muscles of the subject.

Figure 3.3(b) shows a neurological impaired subject standing in the MRF in this mode. Experimental results with a paraplegic subject using the MRF in the second operation mode are described in chapter 5.

(a) Able-bodied subject standing on a rotating platform.

(b) Paraplegic subject standing on force plates.

Figure 3.3.: Subjects standing in the MRF.

3.2.2. The Electro-Hydraulic Servo Circuit

Figure 3.4 shows the electro-hydraulic servo circuit of the MRF.

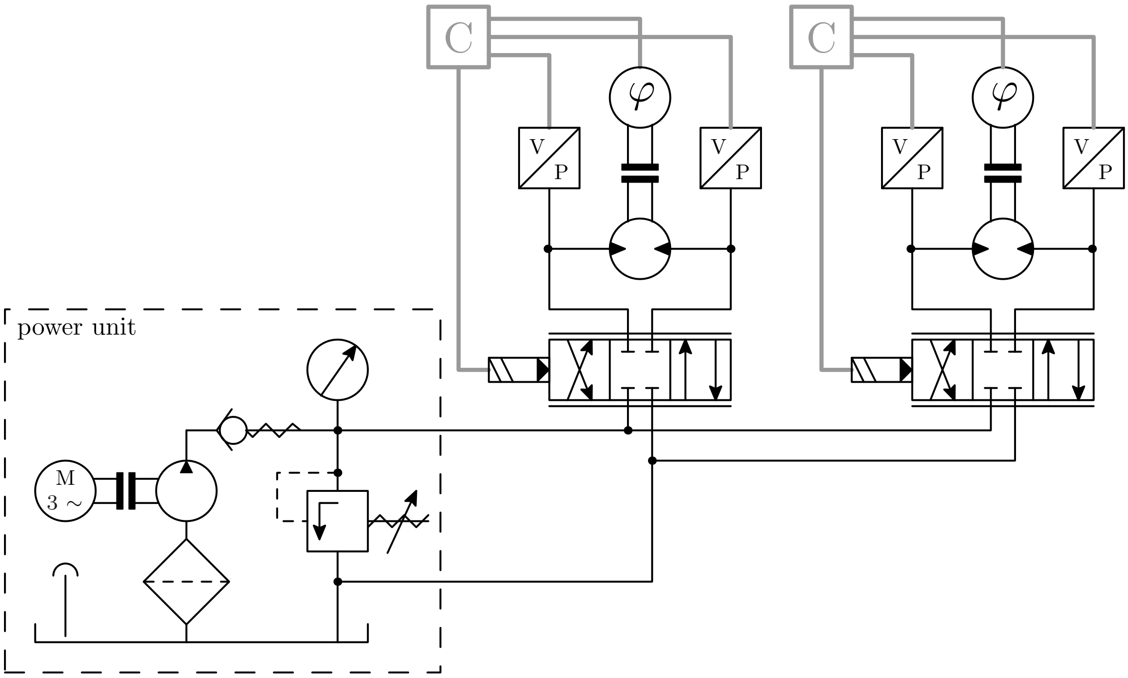


Figure 3.4.: Electro-hydraulic servo circuit of the MRF. The movements in the sagittal and frontal planes are controlled by separate electro-hydraulic servo circuits. These two circuits, arranged in parallel, are driven by a common source of power (dashed box). The electrical parts of the system are shown in grey. The inclination angle of the MRF is measured by shaft encoders, indicated by the symbol "φ". The control of the angle and moment is performed by the blocks labelled "C" (see Figure 3.5 for more details). For hydraulic symbols refer to Zoebl [112]. For the specification of the hydraulic components refer to Appendix A. *Adapted from [38].*

The two circuits are connected in parallel and control the movement of the frame in the sagittal and frontal planes independently. The hydraulic power is provided by a common source which consists of a three-phase motor driving a hydraulic pump and providing a constant operating pressure. With a 4-way servo valve, driven by an electric DC torque motor, the flow through the rotary hydraulic actuators is controlled. Under no-load conditions the flow through the valve is proportional to its driving current. The torque applied at the frame is provided by the hydraulic actuator and is proportional to the pressure difference between the inlet and the outlet of the actuator. This difference is measured by two pressure transducers. For the control of the angle the measurements from an absolute shaft encoder are used.

3.2.3. Hydraulics and Sensors

For the supporting moment a hydraulic pump with a nominal power of 1.1 kW at 3000 rev/min is used. The pump and the two hydraulic actuators applying the supporting stiffness to the frame in the sagittal and frontal planes are supplied by KNAPP Micro Fluid GmbH, Barbing, Germany (*www.knapp-microfluid.de*). The applied pressure is measured with pressure transducers attached to the hydraulic actuators (MP Filtri, Italy; *www.mpfiltri.it*). The flow of hydraulic fluid is regulated by servo valves manufactured by MOOG Inc. (MOOG Inc., East Aurora, NY, USA; *www.moog.com*). Further specifications of the equipment used can be found in appendix A.

The forces and moments applied by the subject onto the ground during standing are measured by two force plates, one under each of the subject's feet. This equipment is manufactured by Advanced Mechanical Technology, Inc., Massachusetts, USA (*www.amtiweb.com/*).

For the measurements of the inclination angles of the lower body in the sagittal and frontal planes two absolute shaft encoders are used, supplied by Hengstler GmbH, Aldingen, Germany (*www.hengstler.com*). To measure the movement of the upper body an ultrasonic motion analysis system is used, provided by zebris Medical GmbH, Isny, Germany (*http://web.zebris-medical.de/*). The zebris system measures the travelling time of ultrasound pulses emitted from a sender and received by markers attached to the body of the subject. The computing unit of the zebris system computes from these measurements the distances between the sender and the respective marker in the three dimensions. With simple triangulation the inclination angles of the upper body can then be computed.

3.2.4. Control of the MRF

The structure of the control blocks labelled "C" in Figure 3.4 is shown in Figure 3.5.

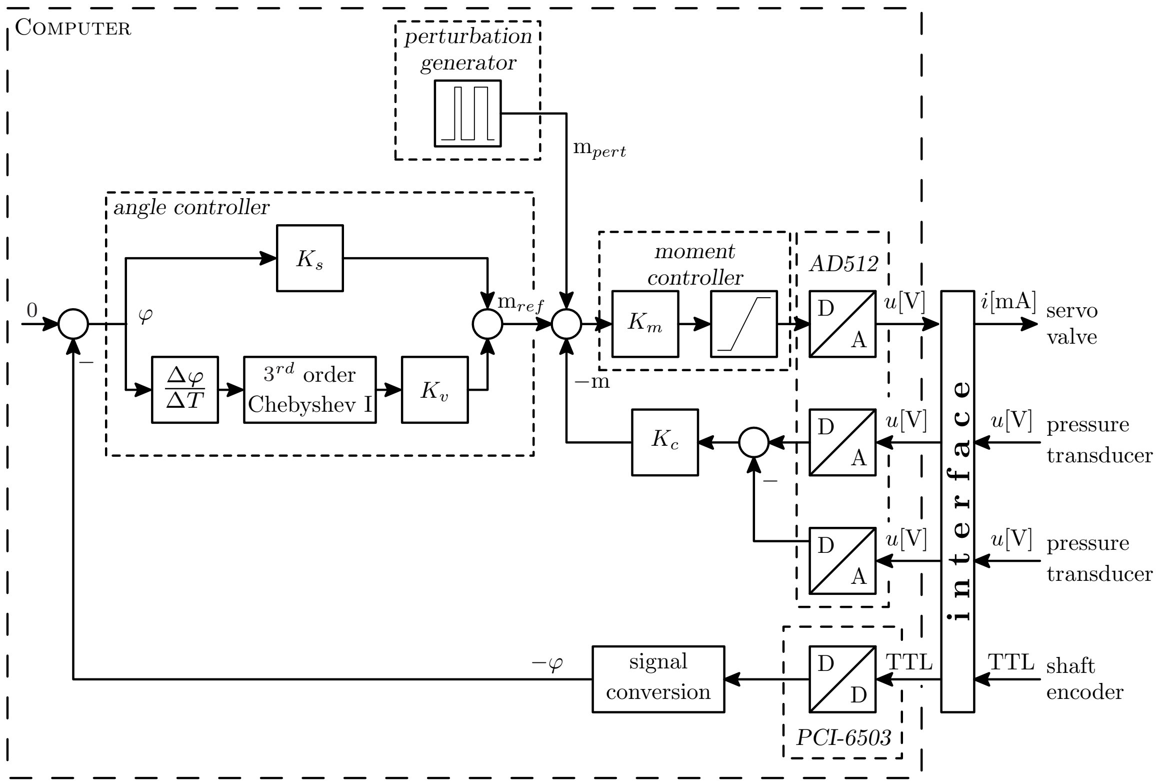


Figure 3.5.: Nested control structure of the MRF. Each of the blocks labelled "C" in Figure 3.4 is of this structure. The input/output signals are connected to the acquisition boards via an external interface. For the data acquisition boards refer to Appendix A. *Adapted from [38].*

The desired initial position of the MRF is its vertical position defined as the inclination angle of 0°. The actual angle φ, measured by the shaft encoder, is compared with the reference value. The angle controller employed is a proportional-derivative controller (PD-controller) with an adjustable stiffness (K_s) as the proportional part of the controller and an adjustable viscosity (K_v) as the derivative part. Due to the quantisation noise the derivative of the angle signal is filtered using a third-order Chebyshev (Type I) filter with a cut-off frequency of 5.47 Hz. As a result, a desired reference for the moment is calculated and used for the inner loop of this nested control loop which controls the pressure with a proportional controller. The pressure difference measured between the in- and outlet of the hydraulic actuator is converted to a corresponding moment value using the conversion factor $K_c = 1$ according to the transducer data sheet. The measured moment is controlled by a proportional controller with the controller gain K_m. The outcome of the control is an electrical current which causes the servo valve to adjust the hydraulic flow.

Real time control of the MRF

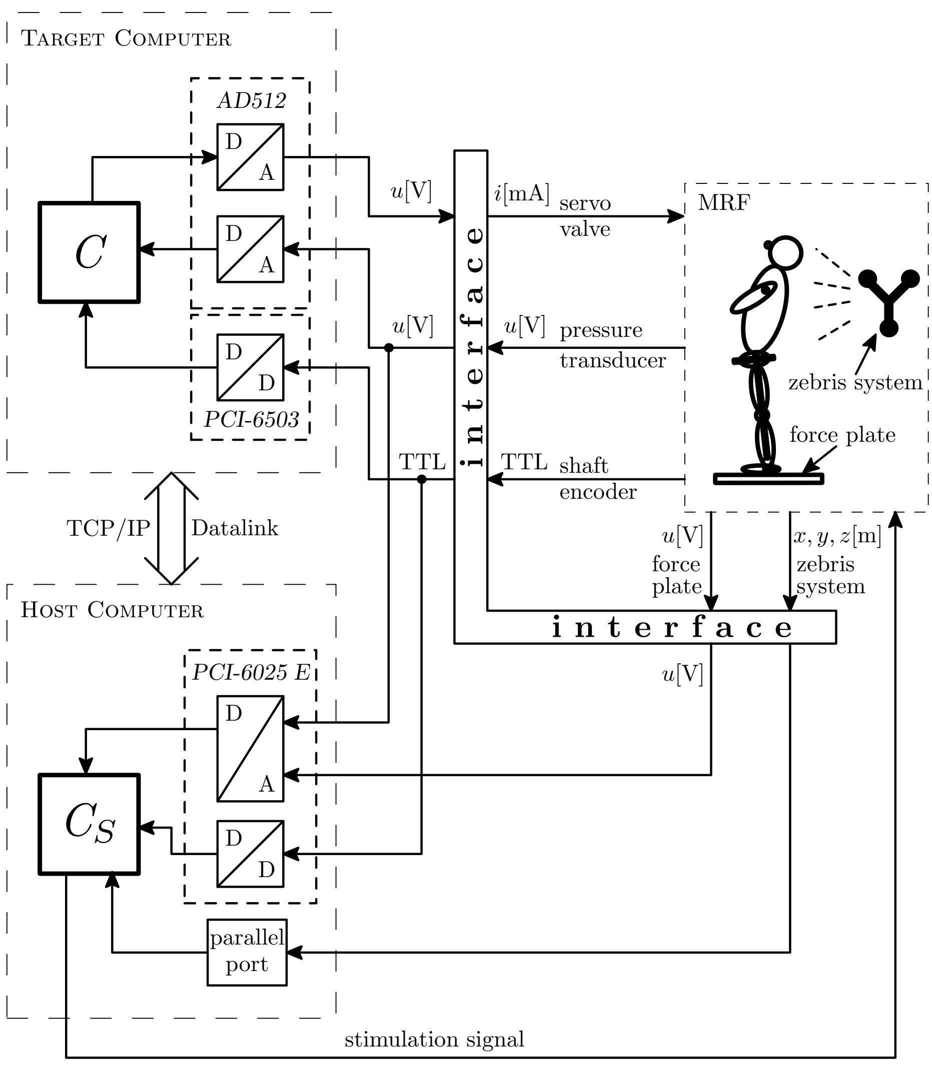


Figure 3.6.: Real time control structure of the MRF. The block labelled "C" has the structure as shown in Figure 3.5. The structure of the block labelled "C$_S$" is shown in chapters 4 and 5. The host and the target computers are connected via a TCP/IP connection.

For the control of the MRF with a person standing in the frame two computers are used. The main computer, the so called host computer (Intel Celeron, 1.4GHz), controls the stimulation signal needed to stimulate the shank muscles of the subject and communicates with the pressure transducers, the shaft encoder, the force plates and the ultrasonic motion system (denoted as "zebris system" in figure 3.6) via an interface.

On the second computer, the target computer (CPU: Intel Pentium III, 650MHz), a real-time kernel allows control of the servo valves of the MRF in real-time. Both computers are connected via a TCP/IP link (up to 100 Mbit/s) (cf. Figure 3.6).

For real-time control the *xPC Target* environment (The Mathworks Inc.) is used which allows to prototype, to test, and to deploy real-time systems using standard PC hardware [113]. The model controlling the physical system is built in the Matlab/Simulink working environment, which is installed on the host PC. After the model has been created, source C-code is generated using the Real-Time Workshop which is part of the Matlab/Simulink working environment. After this step a C/C++ compiler creates executable code which is then downloaded onto the target PC via the TCP/IP data link connection. Now it is possible to run and test the target application in real-time. With certain I/O interface blocks it is possible to observe changes of the measurements in real-time on the target PC. These interface blocks also allow interaction with the actual physical system to send control signals which are changed during operation. The measured data is stored in a ring-buffer in RAM on the target PC. After the execution of the physical system has been stopped the data stored can be uploaded to the host computer, again using the TCP/IP data link.

With the structure shown in figure 3.6 the target PC is used to run the control of the MRF at small sample times which allows the safe performance of the MRF and ensures therefore the safety of the subject standing in the MRF. Although the change of parameters and the visualisation of results is possible, these tasks remain difficult to achieve. Additionally, the communication with the zebris system and the stimulator is not possible using the real-time kernel. Therefore, the host PC is used to compensate for the disadvantages of the target PC allowing an easy communication with the zebris system and the stimulator. Furthermore, the modification of parameters and the visualisation of results can be achieved more easily using comparably large sample times.

3.3. The BalanceTrainer

The advantage of the use of hydraulic actuators with the MRF is the precise adjustment of the supporting moment and the possibility to apply exact perturbations. This is important for research purposes but is not really necessary for everyday rehabilitation in a clinical environment.

Clinicians showed interest in the potential of the MRF in rehabilitation not only for paraplegic patients, but also for stroke patients. However, the hydraulic components make the machine difficult to handle for nontechnical staff, it is bound to one fixed location close to a hydraulic pump, the noise level of the hydraulic pump makes it difficult to use it for long periods of time, and it is expensive to buy.

In order to make the standing frame mobile and easier to use, the MRF was modified and tailored for the purposes of clinical use. A hight adjustable table is placed at pelvis height on two vertical bars which are connected to the base of the frame with two-degrees-of-freedom mechanical joints. These joints consist of helical springs placed in steel cylinders with one end mounted firmly to the base of the frame and the other connected to the vertical bar. These springs give the subject using the frame a certain support which can be adjusted by varying the active length of the springs and replace the hydraulic system used with the MRF. In order to allow the subjects to get into the frame securely, a locking mechanism prevents the frame from tilting. A belt, wrapped around the subject's pelvis, is attached to the table and secures the subject during exercise. The outcome of the mechanical changes resulted in a device called the BalanceTrainer which is now marketed by Medica Medizintechnik GmbH, Hochdorf, Germany (*http://www.medica-medizin.de/*). A more detailed description can be found in [56].

For the case study described in chapter 6 a modified version of the BalanceTrainer was used which is shown in figure 3.7 and described in detail in [114]. It is very similar to an ordinary standing frame. A table at pelvis height is placed on two vertical bars which are connected to the base of the frame with two-degrees-of-freedom mechanical joints. These joints consist of helical springs placed in steel cylinders with one end mounted firmly to the base of the frame and the other connected to the vertical bar. These springs give the subject using the frame a certain support which can be adjusted by varying the active length of the springs. In order to allow the subjects to get into the frame securely, a locking mechanism prevents the frame from tilting. A belt, wrapped around the subject's pelvis, is attached to the table and secures the subject during exercise. In order to apply perturbations, the BalanceTrainer was fitted with four electric motors (two at each side) which are connected via ropes to the frame (see figure 3.7). To perturb the frame in a certain direction the appropriate electric motor winds up the rope and pulls the frame out of its upright position. The motors are powered by two car batteries connected in series.

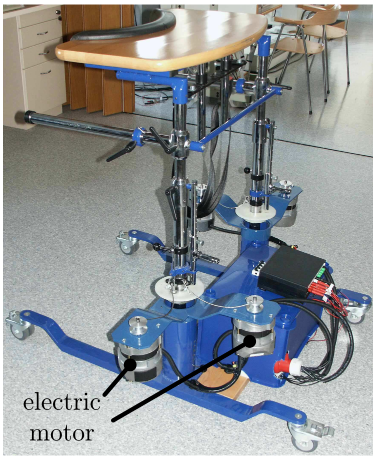


Figure 3.7.: The modified BalanceTrainer with four electric motors used to apply perturbations.

When the frame is perturbed from its upright position the subject standing in the frame is pulled from his/her neutral upright position. The perturbation is a pulse with a constant amplitude, which is the torque magnitude of the electric motor, and a variable duration. This value is adjusted depending on the size and weight of the subject. After choosing a duration this time stays fix for all experiments. Simultaneous activation of two motors allows a total of eight directions of perturbation (see also Figure 3.8).

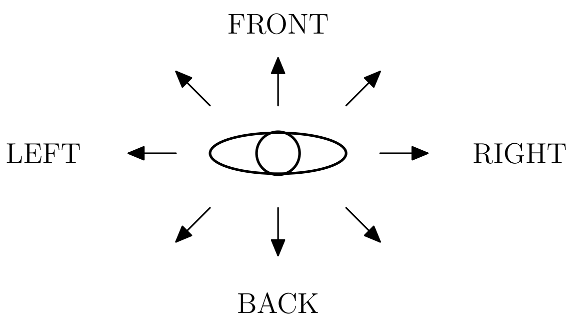


Figure 3.8.: Perturbation directions.

Experimental results using the BalanceTrainer with a stroke patient performing balance training are presented in chapter 6.

4. Evaluation of Standing

4.1. Summary

In this chapter an alternative way of evaluating the performances of able-bodied and spinal cord injured subjects during standing is introduced.

For validating this alternative evaluation value experiments were carried out using the MRF (see section 3.2). Furthermore, the performances of subjects during standing presented in this chapter are used for comparison later on with the results achieved using the new integrated voluntary control approach of chapter 5 (see for results chapter 5.4).

Able-bodied subjects stood on a rotating platform attached to the MRF which immobilised their ankles and prevented the subjects this way from stabilising themselves with the muscle strength of their own shank muscles. The stabilising moment was provided by artificially controlled hydraulic actuators instead. The spinal cord injured subject, however, stood on force plates which measured the moment applied onto the ground. These measurements were used for the control of the stimulation signal which was applied to the subject's gastrocnemius muscles in order to stabilise him during standing. All the subjects were kept in a double-link inverted pendulum configuration allowing voluntary movement of the upper body and were perturbed in a pseudo random binary way. The controllers used where the ones suggested by Matjačić *et al.* [59] and Jaime *et al.* [105].

With the alternative evaluation, for which the variance of a time signal around its mean is computed, it is now possible to evaluate the behaviour of the lower and upper body separately.

As a consequence the evaluation value allows to judge whether a performance was good or bad and makes statistical analysis possible. The performance of a paraplegic subject during a balancing task is qualified as "good", when the lower body does not sway too much whereas the upper body would do the majority of the balancing work allowing a prolonged period of time of standing.

4.2. Background

The overall aim within the framework presented here is to prolong the duration of standing while controlling the posture of the paralysed subject. Simply measuring the time of standing is not suitable for evaluating the quality of control as this is very time demanding and the subjects would need to rest a period of time long enough for the muscles to recover. Therefore, an alternative way for evaluating the performance during standing is used indicating the quality of standing via sway size.

In systems designed to reject disturbances it is assumed that these disturbances alter the output of the controller. Depending on the characteristics of the controller used the disturbance is rejected differently.

In order to evaluate the quality of disturbance rejection, the system is exited with a certain disturbance signal. Commonly, either a step or an impulse function are used for the disturbance as indicated in figure 4.1.

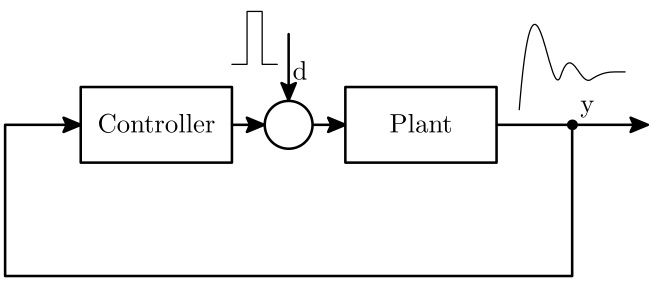


Figure 4.1.: General control loop with indicated impulse function as the disturbance d and a respective response y as output.

Common values to evaluate the response towards an impulse or step like disturbance in the time-domain are

- the rise time, which is typically defined as the time the controlled signal needs to rise from 10% to 90% of the steady state value,
- the steady state value,
- the maximal overshoot value, and

- the maximal undershoot value.

These different characteristic values of the response signal allow the direct comparison of different control performances even if different controllers were used.

Frequency domain techniques, such as Bode plots, Root Locus diagrams as well as Nyquist and Nichols plots can be used to evaluate the behaviour of the closed loop in certain frequency ranges. Furthermore, these methods can not only be used to compare control performances but to tune control parameters so that the closed control loop behaves in a desired manner.

Due to the construction of the MRF the subject is perturbed by a moment applied around the ankle. A pulse like disturbance, with an amplitude constrained due to safety reasons, is applied. As a single pulse would not reveal very much about the subject's balance performance, several pulses are applied over a fixed period of time. In order to reduce the effect of learning by the subject to a minimum the pulses are applied in a pseudo random way, i.e. the duration of a pulse as well as the time between the pulses varies pseudo randomly. For using the evaluation values mentioned on page 41, first an ARX- or ARMAX model would need to be identified in order to model the relationship between the PRBS-input and the measured output. Then a step response or an impulse response could be computed and evaluated for this model. In order to simplify this evaluation procedure a simpler way of evaluating the behaviour of a subject during standing is needed, which would allow the comparison of different performances.

4.2.1. Evaluation Methods in Literature

The main aim of the different exercise approaches which incorporate FES for the functional movement of paralysed muscles is to allow the paralysed person to perform the exercise for as long as possible. In literature different measures are proposed to quantify the quality of standing which comprise mainly the measurements of elapsed standing times, ground reaction forces, and changes in the centre of pressure during repeated stereotypical disturbances.

As the paralysed muscle tends to fatigue relatively quickly, researchers developed various strategies to optimise the amount of stimulation in order to prolong the duration of exercise.

Kralj *et al.* [108] introduced a fatiguing index describing the rate of muscle fatigue over a certain time. Upright standing in paraplegia with the application of FES to the quadriceps was chosen as the functional task. The aim was to predict the functioning endurance of the subject and to prolong exercising time using this index for adjusting the applied stimulation.

Önell [115] found that the heart beat of a person standing on a force plate during quiet standing can be seen as small force peaks in the ground force measurements. Results comparing the performance of balance impaired and healthy adults revealed that the heart beats in the balance impaired population were seen more clearly than those in healthy people. From these observations the author proposed to use the vertical ground reaction force to assess the balance performance during quiet standing.

Popović *et al.* [109] propose a stability criterion to assess the current standing situation of a balance impaired subject which can be used for the control of supporting means (e.g. a standing prosthesis) which the subject uses for standing. The authors defined four different stability zones relating to certain positions of the centre of pressure (CoP). A general model was developed to compute a priori the stability zones for any subject and consequently to assess the actual stability using the actual CoP measurements.

4.3. Variance based Evaluation

For assessing the performances achieved with the control approaches described in chapter 5 the evaluation methods suggested in literature (see section 4.2.1) would be suitable to only a limited extent.

Able-bodied subjects are standing during the experiments on a rotating platform attached to the MRF (see for more details section 3.2) and therefore neither forces nor moments are applied onto the ground. Consequently, the evaluation of the forces and moments applied onto the ground as suggested by Önell and Popović is not possible to be used with this group of subjects.

For evaluating the performance of SCI subjects, however, the CoP could be used as a possible evaluation value, as these subjects are standing on force plates during the experiments. As the reaction forces show changes induced by both, the upper and the lower body, the CoP would, however, only allow the evaluation of the overall stability of the subject.

The fatiguing index proposed by Kralj *et al.* is used for adjusting the stimulation in SCI subjects in order to prolong the duration of standing. This fatiguing index, defined by the rate of muscle fatigue over a certain time, is, however, not suitable as it is a very time demanding way for comparing performances achieved with different control approaches.

To solve the disadvantages of the known evaluation methods mentioned above, a new evaluation method is necessary which would not require time consuming experiments and allow to assess the performance of the upper and of the lower body separately.

It is suggested to use the variance of a signal around its mean value as a measure for the effort during perturbed standing. For this any measurement (inclination angle, velocities, accelerations, etc.) associated with the movement of the upper and lower body, respectively, can be assessed. The variance allows also the comparison of results achieved with different control approaches easily.

The variance of a certain signal around its mean value is calculated using following formula [116]:

$$S^2(x) = \frac{1}{N-1}\sum_{i=1}^{N}(x_i - \overline{x})^2 \qquad (4.1)$$

Here, N is the number of sample points, x_i is the sample point at the time point i, and $\overline{x}$ is the mean value of the signal x, with x denoting the signal to be evaluated. S^2 has the squared unit of variable x.

The measurements, which allow to draw a conclusion concerning the effort generated by the voluntary movement of the subject as well as by the artificial control, are:

- the measured torque applied by the hydraulic actuators at ankle hight (τ_{meas})
- changes in the inclination angles of the lower body (φ_1), and
- changes in the inclination angles of the upper body (φ_2).

4.4. Methods for Experiments with Able-bodied Subjects

To verify the validity and usefulness of the variance measure for evaluation standing experiments with able bodied subjects were carried out using the MRF. All experimental procedures were approved by the Ethics Committee of the Faculty of Biological and

Life Sciences (FBLS) at the University of Glasgow and the subjects provided written, informed consent prior to participation.

4.4.1. Setup

For the experiments with able-bodied subjects the ankles are disabled mechanically by letting them stand on a rotating platform attached to the MRF (see also section 3.2 and figure 3.3(a)). The subject is kept in a double-link inverted pendulum configuration with the knee and hip joints being mechanically locked and allowing voluntary movement of the upper body around the lumbar joint. The necessary stabilising supporting torque at the ankle is provided by hydraulic actuators and is controlled artificially.

4.4.2. Subjects

For the experiments three male and one female able-bodied subject participated. Following table contains the data of the participating subjects regarding age, body hight, and body weight.

Subject	weight [kg]	hight [cm]	age	sex
1	62	175	32	male
2	60	168	26	female
3	75	188	30	male
4	68	182	23	male

Table 4.1.: Characteristic data of the participating subjects.

4.4.3. Control Structures

In order to verify the evaluation method of standing introduced in section 4.3, the control structure used by Jaime in his work [38] is employed.

The control structure consists of a cascaded control loop to control the upright standing posture of a subject in the MRF (for a more detailed description see section 3.2). The inner loop of this nested structure controls the moment produced at the ankle, whereas in the outer loop the inclination angle of the lower body is controlled. For the moment controller in the inner loop a proportional controller is used (see also figure 3.5).

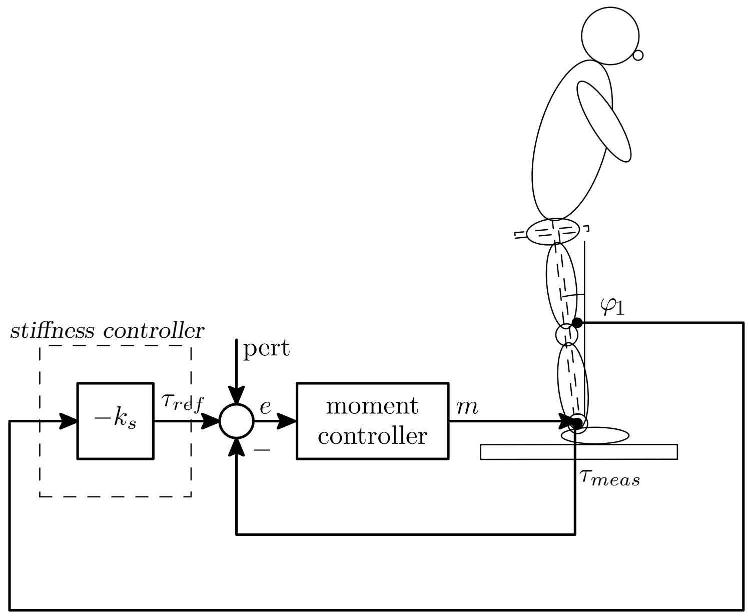


Figure 4.2.: Control loop using a stiffness controller for the angle control.

For the outer angle control loop, two approaches are compared: (i) a simple P-controller ("stiffness controller") with a gain factor k_s, see figure 4.2, and (ii) a PD-controller ("stiffness-viscosity controller") with a proportional gain k_s and a differentiator with a gain k_v and a time constant T_D, see figure 4.3. The signal "pert" indicates the perturbation applied in form of a pseudo random signal.

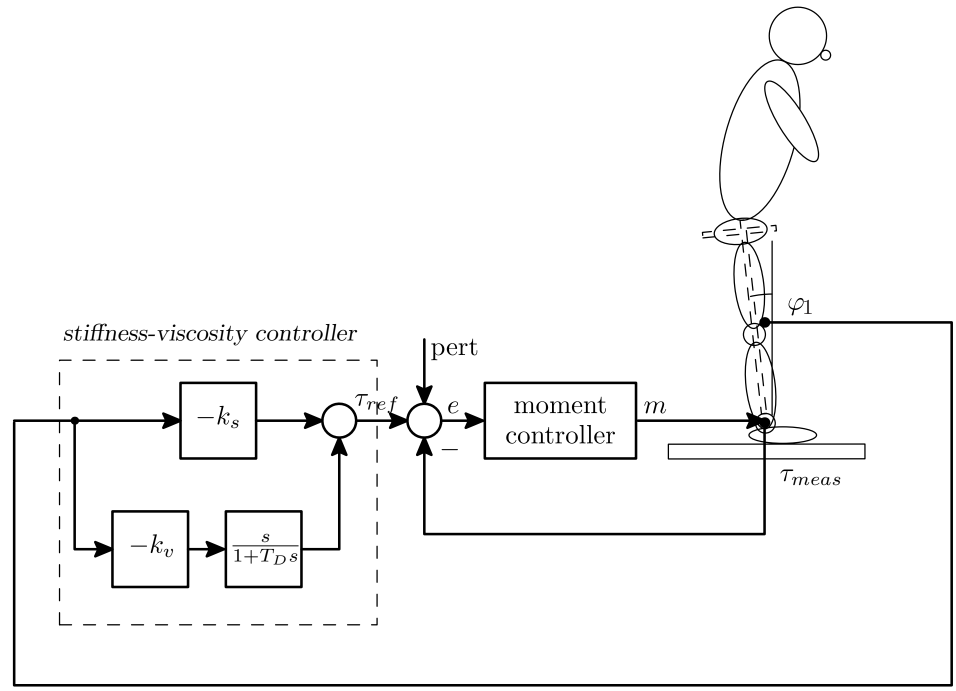


Figure 4.3.: Control loop using a stiffness-viscosity controller for the angle control.

4.4.4. Stiffness and Viscosity Values

In order to use the stiffness and stiffness-viscosity controllers stiffness and viscosity values which are typical for natural perturbed standing, are needed. For this purpose the procedure suggested by Black *et al.* [117] was carried out using the natural balance response to pseudo random perturbations applied to neurologically healthy people during perturbed standing in a double-link inverted pendulum configuration. For this the knee joints were locked mechanically, allowing the subject to move only around the ankle and lumbar joints. For the experiments, the MRF, as described in chapter 3.2, was used. The subjects were standing on two force plates which measured the moment delivered onto the ground. The inclination angle of the MRF was measured by the shaft encoder attached to it. The subjects were perturbed by a pseudo random binary signal conveyed by the MRF applying two different types of perturbation signals using different amplitudes which were chosen in such a way that the subjects would not lose balance during the experiments. No instructions to the subjects were given in how they should react to the perturbations.

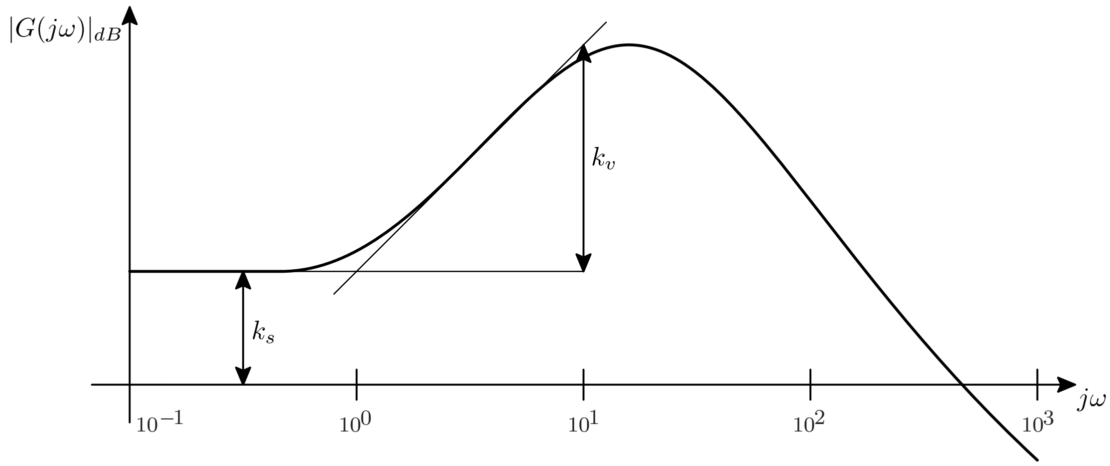


Figure 4.4.: A frequency response plot with denoted stiffness k_s and viscosity k_v values.

For the estimation of model parameters describing the relation between the measured inclination angle as input and the resulting moments, Black *et al.* [117] showed that a third order ARX (autoregressive with exogenous inputs) model structure fitted the measurements best. The frequency responses of the resulting ARX models were used to establish the respective stiffness and viscosity value of the response for every subject. As indicated in figure 4.4, the static gain of the frequency response denotes the muscle stiffness k_s whereas the first slope is an indication of the muscle viscosity (denoted k_v). The stiffness and viscosity values which were found for each subject are shown in table 4.2.

Subject	k_s [Nm/deg]	k_v [Nm/deg s]
1	14.29	1.66
2	12.16	2.36
3	24.27	1.48
4	16.03	1.39

Table 4.2.: Nominal stiffness and viscosity values determined for the four able-bodied subjects.

4.4.5. Experimental Protocol

A rotating foot platform was attached to the frame (see figure 3.3(a)) providing an artificially actuated ankle joint. Furthermore, the subjects were kept in a double-link inverted pendulum configuration allowing the voluntary movement of the upper body. In order to keep the subjects in an upright position the stiffness and stiffness-viscosity controller from section 4.4.3 were used to apply a stabilising moment at the ankle. As controller parameters the nominal stiffness and viscosity values from table 4.2 were used.

The nominal stiffness and viscosity values reflect the stable situation of a subject during applied perturbations. To see the robustness of the control approach, the angle control parameters are altered by using a scaling factor K_{scale}. The scaling values were chosen as 1, 0.7 and 0.4. The resulting stiffness and viscosity values are shown in table 4.3.

Subject	$1 \cdot k_s$	$0.7 \cdot k_s$	$0.4 \cdot k_s$	$1 \cdot k_v$	$0.7 \cdot k_v$	$0.4 \cdot k_v$
1	14.29	10.0	5.72	1.66	1.16	0.66
2	12.16	8.51	4.86	2.36	1.65	0.94
3	24.27	16.99	9.71	1.48	1.04	0.59
4	16.03	11.22	6.41	1.39	0.97	0.56

Table 4.3.: Stiffness and viscosity values determined for the four able-bodied subjects.

The subject was perturbed only to the front and to the back by applying a pseudo random binary sequence signal (PRBS signal) alternating between the positive and negative value of the amplitude in a pseudo random way as shown in figure 4.5.

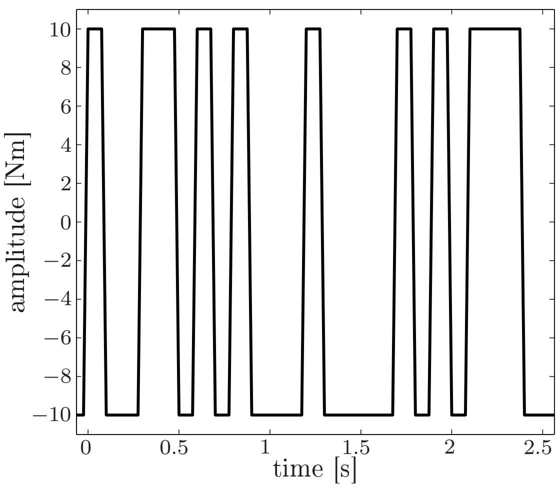


Figure 4.5.: Excerpt from a Pseudo Random Binary Sequence signal alternating between fixed amplitude values with a varying pulse width.

For the experiments two different perturbation levels were applied. The signal with the low level (abbreviated as "LP") applied a moment of ±10 Nm, whereas for the experiments applying the high perturbation level (abbreviated as "HP") a moment of ±15 Nm was used. These perturbation levels were the same as during the experiments for finding the subjects' individual stiffness and viscosity values (see also [117]). The different scenarios depending on the scaling factor used are denoted as e.g. 40%LP and 40%HP when the scaling factor 0.4 with a low (LP) or high perturbation level (HP) is applied. For the scaling factor 0.7 the scenarios are denoted as 70%LP and 70%HP whereas the results with the nominal value are denoted as 100%LP and 100%HP. The different stiffness and stiffness-viscosity value combinations for the angle controller are shown in table 4.4.

	stiffness control	stiffness-viscosity control
scenario 40% LP, HP	$0.4 \cdot k_s$	$0.4 \cdot k_s$ & $0.4 \cdot k_v$
scenario 70% LP, HP	$0.7 \cdot k_s$	$0.7 \cdot k_s$ & $0.7 \cdot k_v$
scenario 100% LP, HP	$1.0 \cdot k_s$	$1.0 \cdot k_s$ & $1.0 \cdot k_v$

Table 4.4.: Combination of stiffness and viscosity values used for the angle control.

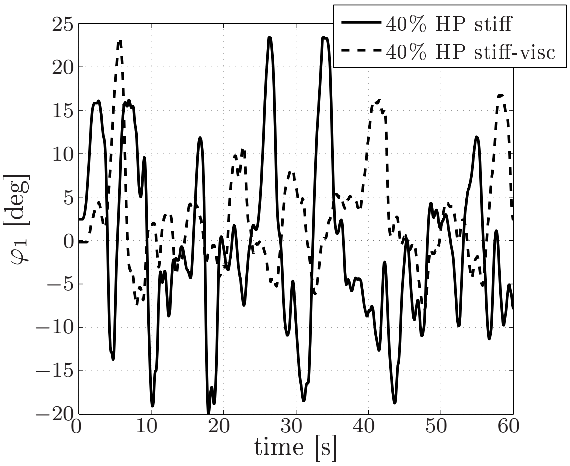
The combination of the three stiffness and three stiffness-viscosity controllers with these two different perturbation levels resulted in 12 trials every subject had to perform during every session with each trial lasting 60 seconds. The order of the trials during one session was random and changed from day to day. This minimised the effect of learning. The resting time between trials was just long enough to save measured data of a performed trial and set the necessary control values for the next trial. Overall one experimental session took around 20 minutes. The experiments were repeated over five consecutive days.

4.5. Results (Able-bodied Subjects)

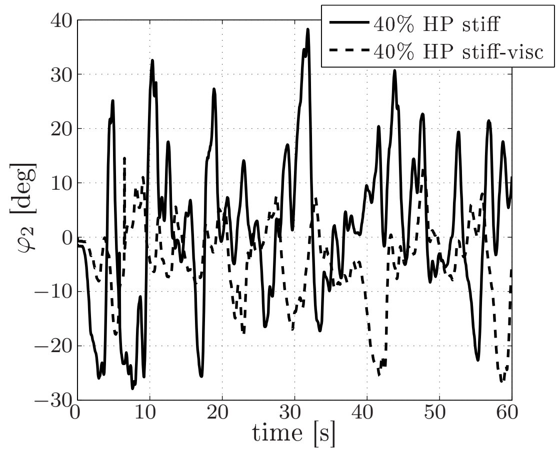
In this section the variance measure introduced in section 4.3 is applied on the performance of able-bodied subjects and the responses of different subjects with the same controllers and with the same perturbation levels are compared.

4.5.1. Different controllers used with one subject

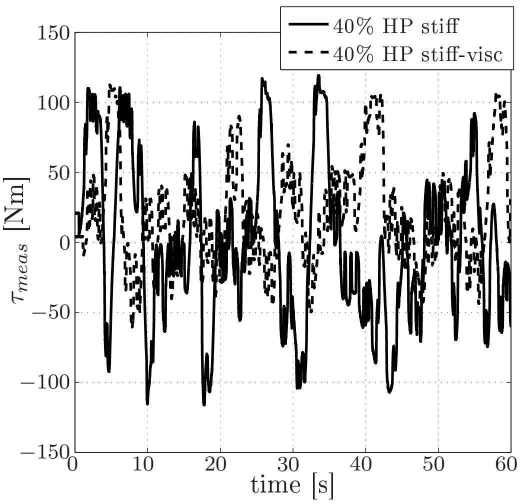
To demonstrate the applicability of the variance as evaluation value the angular change of the lower body (φ_1), of the upper body (φ_2), and the torque applied at the ankle (τ_{meas}) over 60 seconds on one testing day is shown for subject 4 as an example (cf. figure 4.6). The perturbation level was set to ±15Nm and a stiffness value of $k_s = 6.41$ Nm/deg and a viscosity value of $k_v = 0.56$ Nm/deg s (see table 4.3) were used.



(a) Time plot of inclination angle φ_1



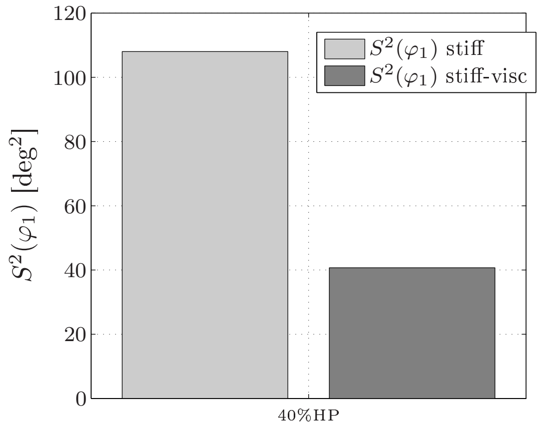
(b) Time plot of inclination angle φ_2.



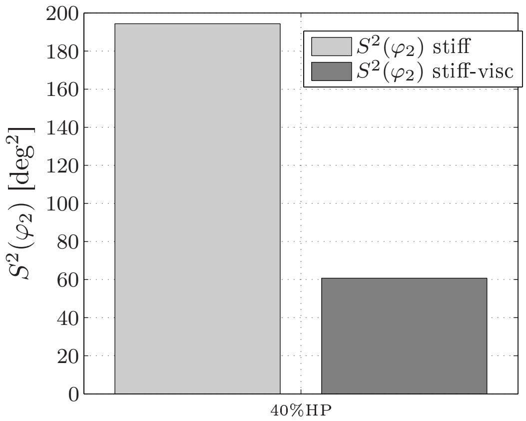
(c) Time plot of torque τ_{meas}.

Figure 4.6.: Results of the inclination angles φ_1, φ_2, and the torque τ_{meas} of subject 4 as he is perturbed with a high perturbation level ("HP"). This plot shows the results of a stiffness and a stiffness-viscosity (denoted as "stiff-visc") controller. The controller values are 40% of the nominal values.

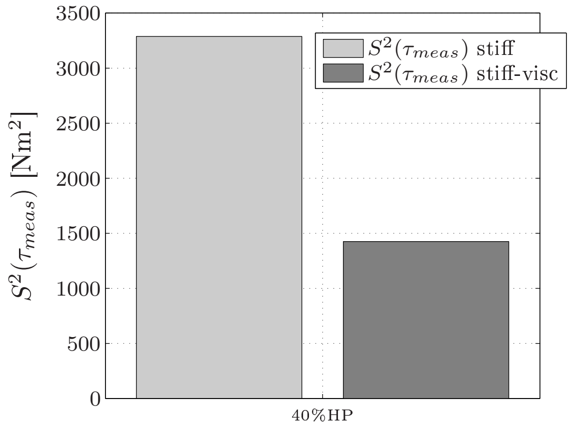
The comparison of performance using on the one hand a stiffness controller (see bold lines in figure 4.6) and on the other hand a stiffness-viscosity controller (dashed lines) shows that the subject was more active with the applied stiffness controller than with the stiffness-viscosity controller. This can especially be seen in the angular values achieved as the subject was leaning to the front and to the back. Figure 4.6(a), for example, shows that the subject leaned with his lower body eight times further than $\pm 15°$ with the stiffness controller applied and only three times when using a stiffness-viscosity controller.



(a) Variance of inclination angle φ_1.



(b) Variance of inclination angle φ_2.



(c) Variance of torque τ_{meas}.

Figure 4.7.: Bar plot of the computed variances $S^2(\varphi_1)$, $S^2(\varphi_2)$, and $S^2(\tau_{meas})$, respectively of the time plot shown in figure 4.6.

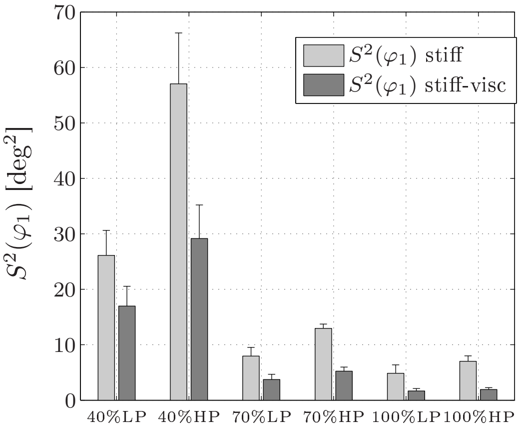
The evaluation of the time plots (see figure 4.6) with the help of the variance according to formula 4.1 confirms the observation that the subject was more active using the stiffness controller compared to the performance achieved with a stiffness-viscosity controller. Figure 4.7 depicts the variance values of the respective time signals as bars and shows that the variance is suitable to be used as evaluation value of the performance during perturbed standing.

4.5.2. Comparison of performance during standing

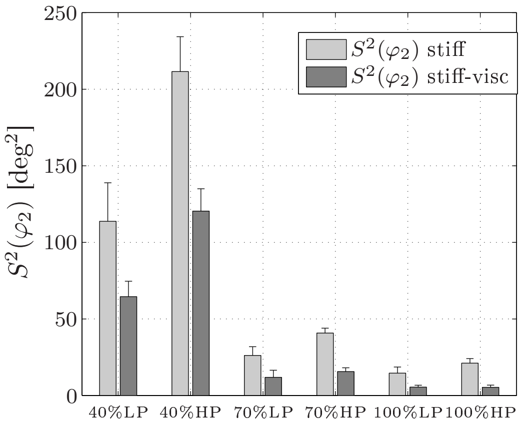
The following figures show bar plots of the mean variances of the responses towards different test scenarios for every subject. Every scenario was performed once per day for five consecutive testing days and the mean values shown are the mean values of these five measurements. For the experiments three different stiffness and three different stiffness-viscosity controllers were used (see also table 4.4). For the bar plots, responses achieved with the same scaling factor and same perturbation strength are grouped. The whiskers in the bar plots indicate the standard deviation of the variances of the inclination angles φ_1 and φ_2 as well as of the torque τ_{meas}.

The performances of all the able-bodied subjects did not vary considerably with time and therefore the changes over time could be ignored. Nonetheless, the experiments were carried out several times in order to test the equipment and procedure before performing these experiments with the SCI-subject.

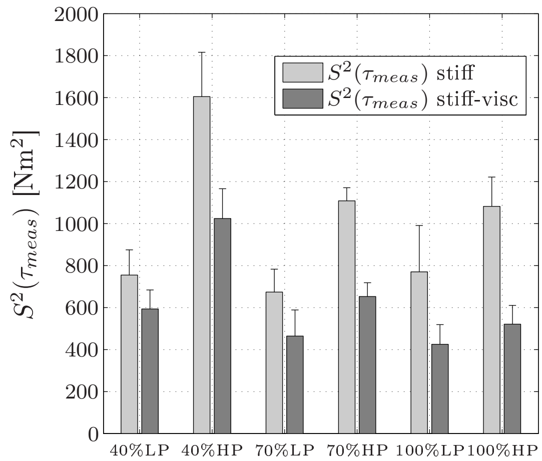
Responses of subject 1



(a) Variance of inclination angle φ_1.



(b) Variance of inclination angle φ_2.



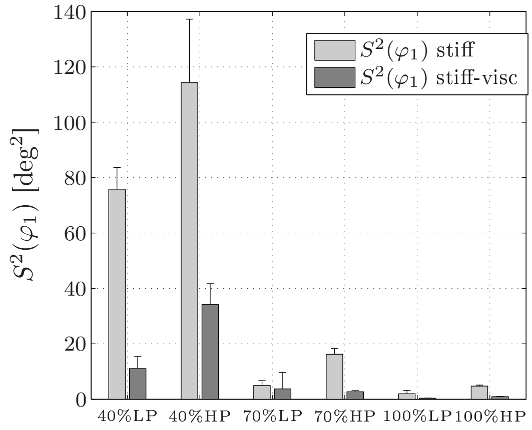
(c) Variance of torque τ_{meas}.

Figure 4.8.: Comparison of variances $S^2(\varphi_1)$, $S^2(\varphi_2)$, and $S^2(\tau_{meas})$ for subject 1 with the respective standard deviations (whiskers) regarding the results achieved over five consecutive testing days.

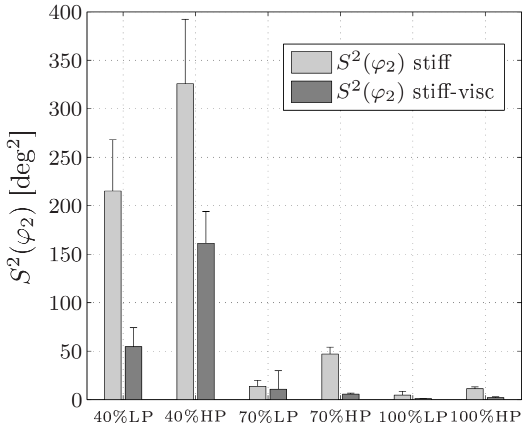
Subject 1 showed for all measured values (φ_1, φ_2, τ_{meas}) during all scenarios smaller variance values while using a stiffness-viscosity controller compared to the respective results acquired with a stiffness controller. With increasing support by the stiffness and stiffness-viscosity controllers (K_{scale}: $0.4 \Rightarrow 0.7 \Rightarrow 1.0$) the variances $S^2(\varphi_1)$ and $S^2(\varphi_2)$ get smaller. This behaviour can also be seen in $S^2(\tau_{meas})$ only with the stiffness-viscosity controller. With the stiffness controller, however, $S^2(\tau_{meas})$ shows approximately the same values for the "70%HP"- and "100%HP"-scenarios while the variance values of the results measured during the"40%LP"- and "100%LP"-scenarios have approximate values.

Responses of subject 2

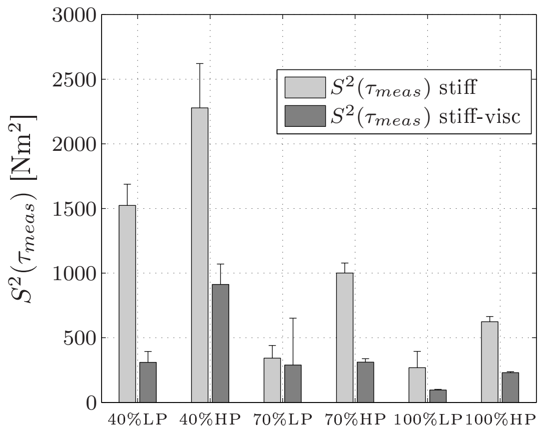
The behaviour shown by all the other subjects is very similar to the performance described for subject 1.



(a) Results for φ_1.



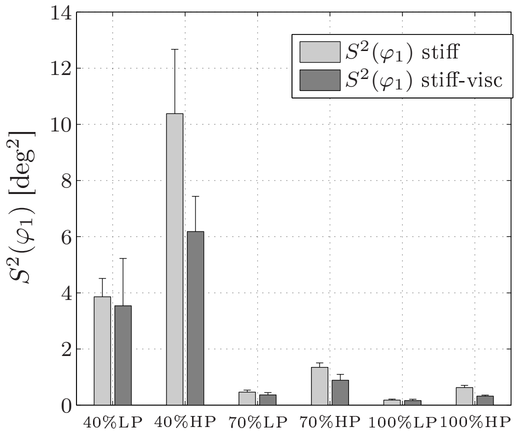
(b) Results for φ_2.



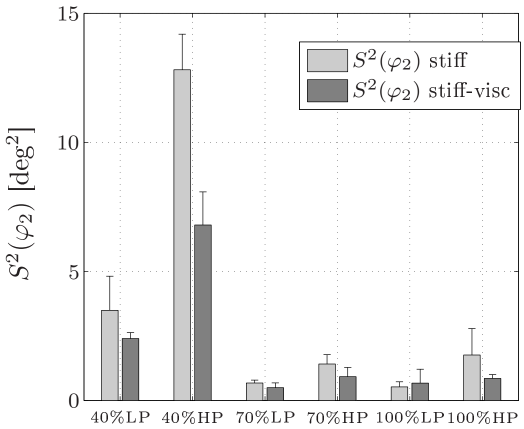
(c) Results for τ_{meas}.

Figure 4.9.: Comparison of variances $S^2(\varphi_1)$, $S^2(\varphi_2)$, and $S^2(\tau_{meas})$ for subject 2 with the respective standard deviations (whiskers) regarding the results achieved over five consecutive testing days.

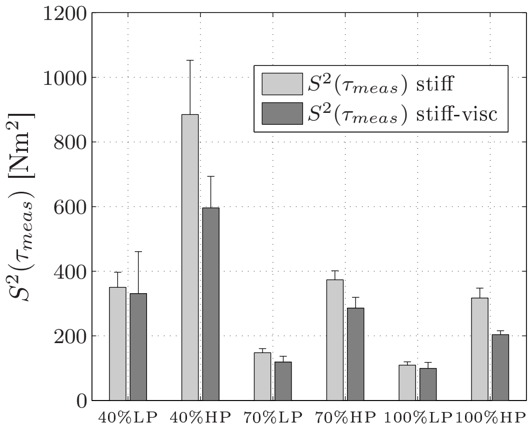
Responses of subject 3



(a) Results for φ_1.



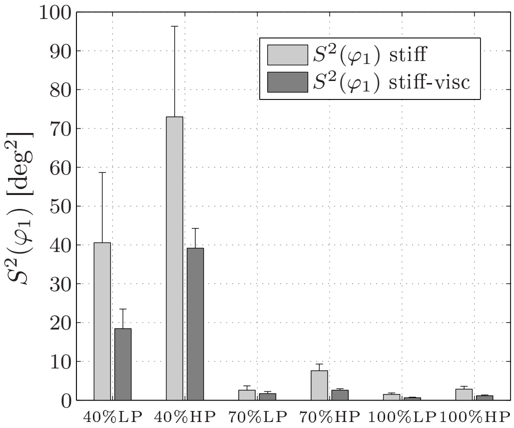
(b) Results for φ_2.



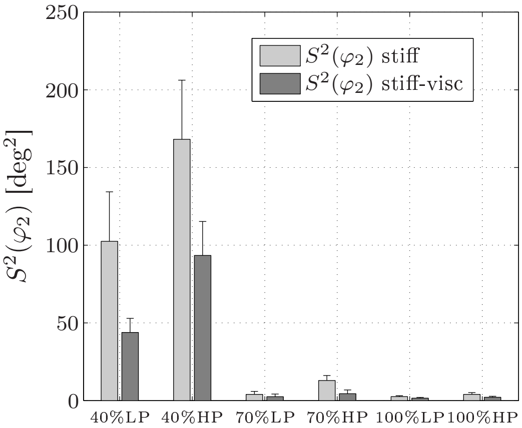
(c) Results for τ_{meas}.

Figure 4.10.: Comparison of variances $S^2(\varphi_1)$, $S^2(\varphi_2)$, and $S^2(\tau_{meas})$ for subject 3 with the respective standard deviations (whiskers) regarding the results achieved over five consecutive testing days.

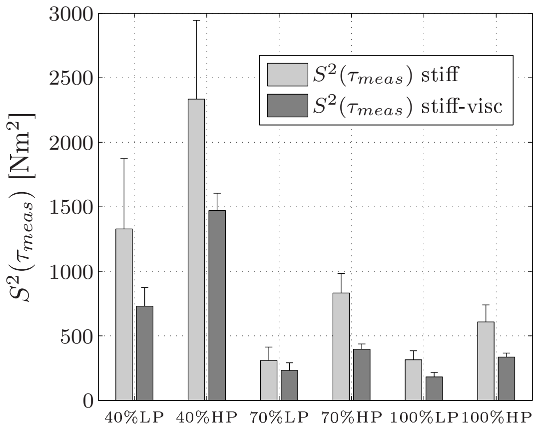
Responses of subject 4



(a) Results for φ_1.



(b) Results for φ_2.

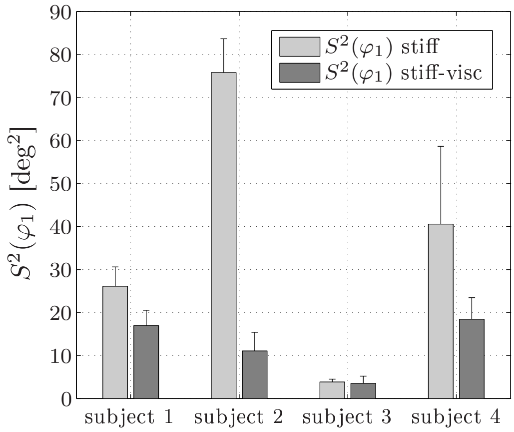


(c) Results for τ_{meas}.

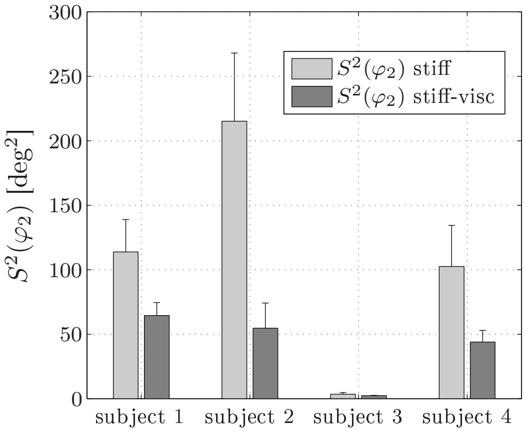
Figure 4.11.: Comparison of variances $S^2(\varphi_1)$, $S^2(\varphi_2)$, and $S^2(\tau_{meas})$ for subject 4 with the respective standard deviations (whiskers) regarding the results achieved over five consecutive testing days.

During all scenarios the subjects were less active with the stiffness-viscosity controller than with the stiffness controller. The comparison of the results achieved during high and low perturbation show also that all subjects were more active during the "HP"-scenarios. Furthermore, the variance of all measured signals decreases as the support of the controllers increase (K_{scale}: $0.4 \Rightarrow 0.7 \Rightarrow 1.0$).

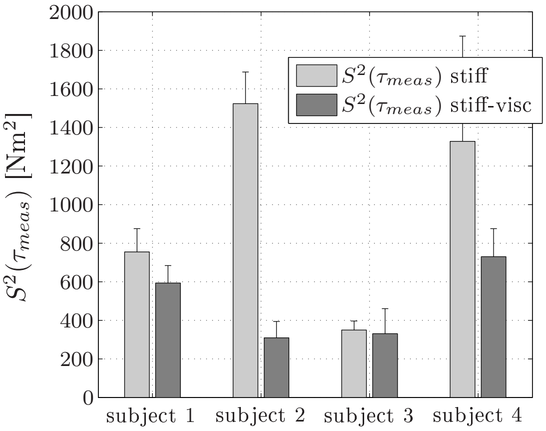
4.5.3. Comparison of performance between all subjects



(a) Results for φ_1.



(b) Results for φ_2.



(c) Results for τ_{meas}.

Figure 4.12.: Comparison of variances of all subjects for the scenario "40%LP".

Figure 4.12 shows a comparison of results of different subjects for the same test scenario. Exemplary the responses for the scenario "40%LP" have been chosen. Results for the other scenarios are similar For all compared variables (φ_1, φ_2, and τ_{meas}) subject 2 showed the highest average variances using a stiffness controller whereas subject 3 showed the smallest. The same applies for the behaviour using a stiffness-viscosity controller except for the measured torque, which is nearly equal to the variance seen in the performance of subject 2 (cf. figure 4.12(c)).

4.6. Discussion and Conclusions (Able-bodied Subjects)

In this chapter an alternative way of evaluating the performance of subjects during standing was introduced. The known strategies for evaluating the behaviour of able-bodied subjects during standing in combination with the experimental setup were not suitable and therefore it was suggested to use the variance of certain measured variables around their means as an evaluation value.

For the experiments with able-bodied subjects results were reproduced using the control approach suggested by Matjačić *et al.* [59] and Jaime *et al.* [105]. Before the performance during standing was evaluated the stiffness and viscosity of a feedback loop using the subjects' gastrocnemius muscles as actuators was established. For the evaluation of standing the subject stood on a rotating platform attached to the standing frame used. The stabilising moment was applied by a hydraulic actuator at the base of the standing frame who was controlled with the help of either a stiffness or a combined stiffness-viscosity controller using the earlier established natural stiffness and viscosity of the subject as controller parameters. For the experiments evaluating the performance during standing these controller parameters were altered using a scaling value with which the support of the artificial control was reduced. The subjects had the task to counterbalance applied perturbations by moving the upper body voluntarily.

The variances of the measured inclination angles of the upper and lower body as well as the ones of the measured torque showed higher values when only reduced support by the hydraulic actuators was given. These results concurred with the observation that the subjects tried to compensate the reduced support by stronger movement of the upper body. The variance as evaluation value reflected also correctly the differences in response when different levels of perturbations were applied. Using a higher perturbation level caused the subjects to be more active with the upper body which resulted in higher variance values.

Furthermore, it was observed that the subjects moved less with their upper body when a stiffness-viscosity controller was used compared to performances using a stiffness controller only. The computed variance values verified this observation with smaller values for results achieved with the stiffness-viscosity controller compared to the results acquired with a stiffness controller (see figures 4.8-4.11).

The comparison of the variances between subjects for the responses for the scenario

"40%LP" shows the influence of the stiffness and stiffness-viscosity controller on the performance during perturbed standing. Subject 3 has the largest natural muscle stiffness and second smallest viscosity (see table 4.3) and shows the smallest value of all for $S^2(\varphi_1)$ and $S^2(\varphi_2)$ achieved with both types of controller (see figure 4.12. Only for $S^2(\tau_{meas})$ achieved with a stiffness-viscosity controller subject 2 and subject 3 showed nearly equal results. Subject 2, however, has the smallest muscle stiffness and the highest muscle viscosity and showed the highest variances of all participating subjects in the inclination angles and measured torque.

These findings again reflect the fact correctly that it is easier to balance a double-link inverted pendulum with two stiff links than a pendulum with two less stiff links and show that the viscosity value has not such big impact on the ability to counterbalance perturbations as the stiffness value has.

To conclude, the results in the previous section show that the variance is suitable:

i) for the evaluation of the performance of a subject during standing in general,

ii) to compare the performance achieved with different control approaches of one subject, and

ii) to compare performances between different subjects.

The use of the variance of a measurement around its mean as evaluation value is a simple approach for comparison and the results presented in this chapter verify the suitability of the variance as evaluation value for perturbed standing.

4.7. Methods for Experiments with a Paraplegic Subject

This section describes the experimental setup for evaluating the performance of a spinal cord injured person during standing and compare the performance achieved with different control approaches. The methods for carrying out the experiments with the paraplegic subject are very similar to the ones conducted with the able-bodied subjects described earlier in this chapter and differ mainly in the moment controller used. The applied control approach is explained in detail in [38] and is described only briefly below.

All experimental procedures were approved by the Ethics Committee of the Faculty of Biological and Life Sciences (FBLS) at the University of Glasgow and the subject provided written, informed consent prior to participation.

4.7.1. Setup

For the experiments with the paraplegic subject the MRF is used (see section 3.2 and figure 3.3(b)). The subject is kept in a double-link inverted pendulum configuration with the knee and hip joints being mechanically locked and allowing voluntary movement of the upper body around the lumbar joint. The necessary stabilising supporting torque at the ankle is provided by functional electrical stimulation of the paralysed shank muscles and is controlled artificially. Although the paralysed muscles are able to provide a stabilising moment with the help of stimulation, they fatigue quickly and can not stabilise the subject for the length of an experimental session. Therefore, an additional supportive stiffness of 5 Nm/deg is applied by the hydraulic actuators of the MRF which was identified by trial and error. With this supportive stiffness alone, the subject is only stable during quiet standing but not during perturbed standing. This supporting stiffness stays constant and is applied throughout the whole experimental session.

4.7.2. Subject

The experiments are performed with one male subject, aged 33, with a motor-complete spinal cord injury at level T9 and 8 years post injury.

4.7.3. Control structure

In the same way as for the experiments with able-bodied subjects a cascaded control structure is used (see section 4.4.3). In the inner loop again the moment at the ankle and in the outer loop the inclination angle of the lower body is controlled. The difference to the experiments described in section 4.4.3 lies therein that the moment produced at the ankles is a result of an applied stimulation current with a varying pulse width to the paralysed calf muscles. The measured moments applied at the force plates the subject is standing on are used for the comparison with the reference torque τ_{ref} (see figures 4.13 and 4.14).

4.7.4. Outer Control Loop: Controlling of Inclination Angle

For the outer angle control loop, again two different controllers are used: (i) a simple P-controller ("stiffness controller") with a gain factor k_s, see figure 4.13, and (ii) a PD-controller ("stiffness-viscosity controller") with a proportional gain k_s, a differentiator with a gain k_v, and a time constant T_D, see figure 4.14.

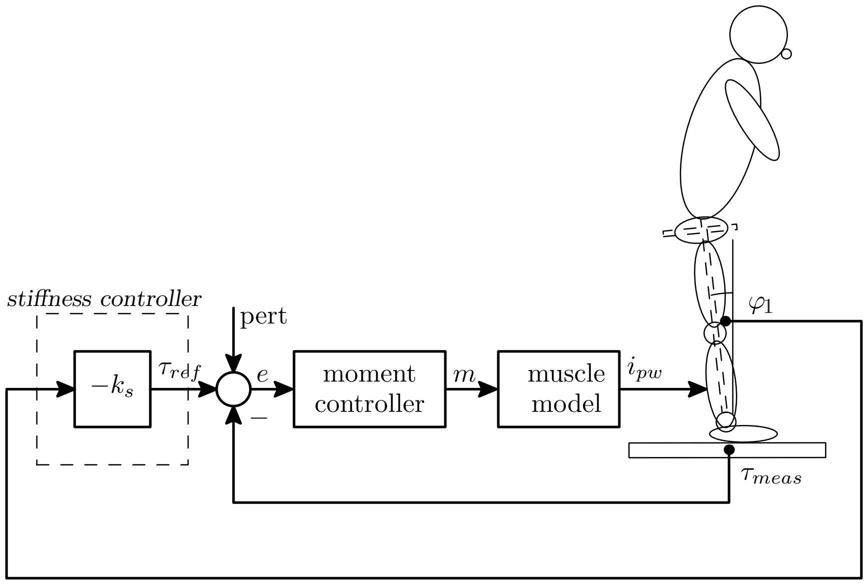


Figure 4.13.: Control loop using a stiffness controller for the moment control.

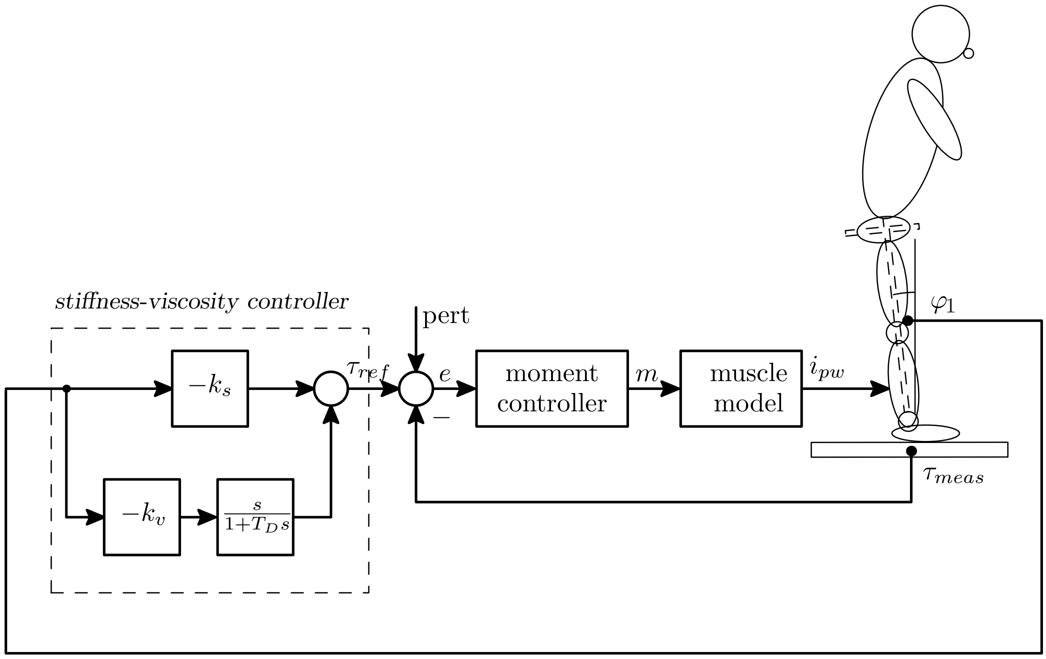


Figure 4.14.: Control loop using a stiffness-viscosity controller for the angle control.

Stiffness and Viscosity Values

Due to the injury of the spinal cord it is not possible to perform similar preliminary tests for establishing the natural stiffness and viscosity value of the shank muscles of the SCI patient as it was done with the able-bodied subjects (see section 4.4.4). Therefore, values for stiffness and viscosity are estimated. In order to get an indication for stiffness and viscosity values a few preliminary tests are carried out. For these preliminary tests the same control structures shown in figures 4.13 and 4.14 are used.

After several preliminary experiments with varying stiffness and viscosity values the muscle stiffness is set to $k_s = 5$ Nm/deg and the viscosity to $k_v = 0.5$ Nm/(deg s). These values are quite low compared to the ones used with able-bodied subjects but this way the subject was able to stand for some reasonable time as otherwise the muscles of the subject would fatigue quick with higher stiffness values.

4.7.5. Inner Control Loop: Controlling of Moment

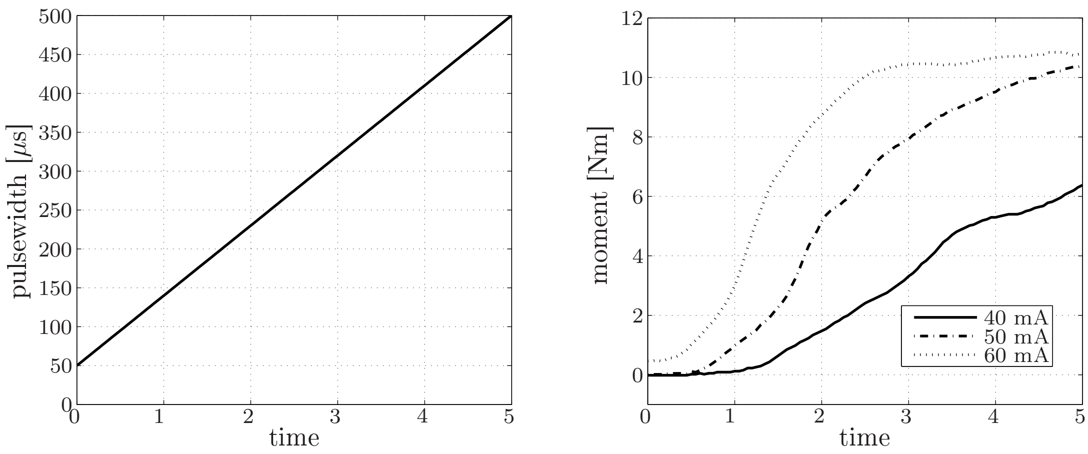
For the control of the moment at the ankle an identified model of the shank muscles and a controller with two poles and one zero is used. This again is based on the suggestions made by Jaime in [38].

Muscle Model Identification

The necessary model identification is carried out before the first experimental session. For the model identification, two different tests are carried out while the paraplegic subject is kept in isometric conditions. For this the subject stands in the MRF on the force plates with the ankles as well as the knees being locked mechanically preventing movement around the ankles. Signals are sampled with a sample time of 0.05 s. Although the describing model of the muscles during movement could be different, the identified model achieved during isometric conditions is used in all experiments and stays always the same.

(i) C-Test: Before the actual model identification the optimal amplitude of the stimulation current has to be determined. Based on a starting **c**urrent value the pulsewidth of the stimulation signal ramps up from 50 μs to 500 μs within 5 seconds. Then the current is increased by 10mA and the same stimulation signal is applied as before. The current is increased until the measured moment reaches a steady value at high pulsewidths without making the subject uncomfortable (see also figure 4.15). The chosen stimulation current is the value with which the muscles neither saturate too early (stimulation current too high) nor too late (stimulation current value too low). For this test the same stimulation signal is applied to both legs at the same time. The recorded moment, as shown in figure 4.15(b), is the moment produced by both legs together.

(ii) PRBS-Test: For the acquisition of the data used for the model identification a stimulation signal, whose pulsewidth varies in **PRBS** form, is applied. The amplitude of the PRBS signal varies by ± 35 μs around the mean value with a period of 155 samples and is constant for at least 5 samples after each transition. The current of the stimulation signal is the one determined during the C-Test earlier and is kept constant throughout the trial. The behaviour of the muscles is measured while PRBS signals with different mean values are applied. Contrary to the C-Test, each PRBS-Test trial lasts for 20 seconds. After one trial is performed the mean value of the pulsewidth is increased by 50 μs. This is continued until the muscles saturate and show no reaction towards changes in pulsewidth.

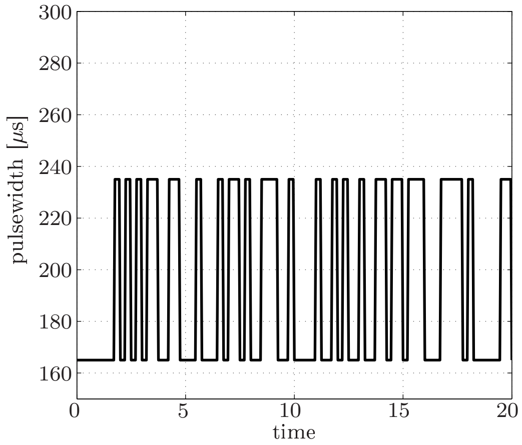


(a) Ramping pulsewidth of stimulation signal.

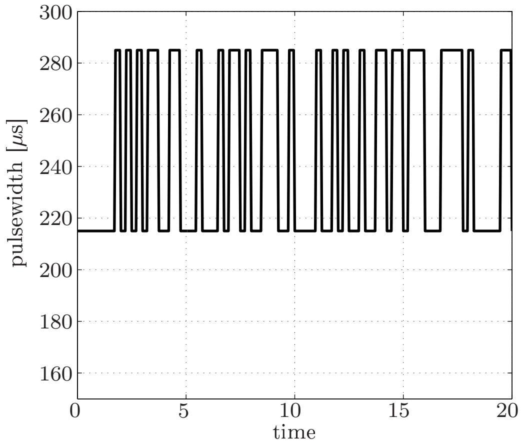
(b) Resulting moment using different stimulation currents.

Figure 4.15.: Resulting total moments after the subject muscles were stimulated at different current levels.

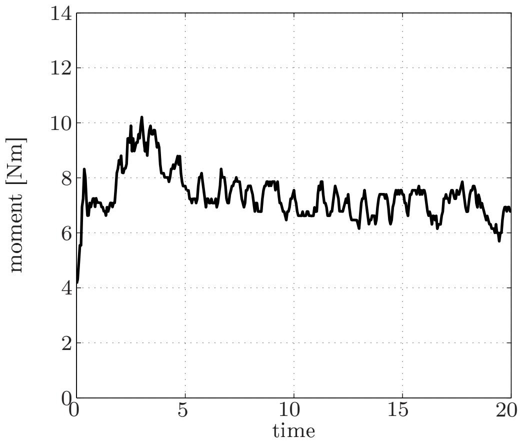
With a few preliminary C-Tests it is established that the muscles of the paraplegic subject respond best with an applied current value of 60 mA on the left and right side. With this current value, several PRBS-Tests are carried out to identify the respective models of the calf muscles. The responses to applied PRBS signals are shown in figure 4.16. For more details regarding the C- and PRBS-Tests see [38].



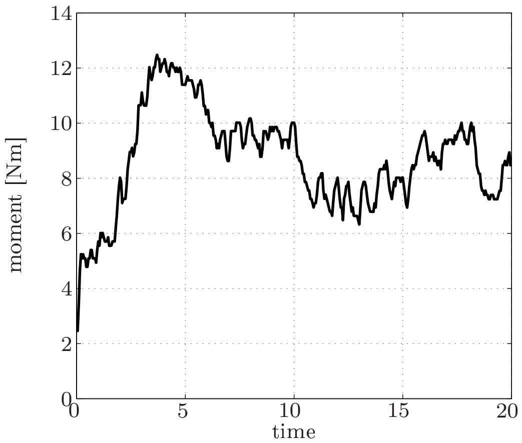
(a) PRBS signal with mean value at 200μs.



(b) PRBS signal with mean value at 250μs.



(c) Resulting total moment using excitation signal of figure 4.16(a).



(d) Resulting total moment using excitation signal of figure 4.16(b).

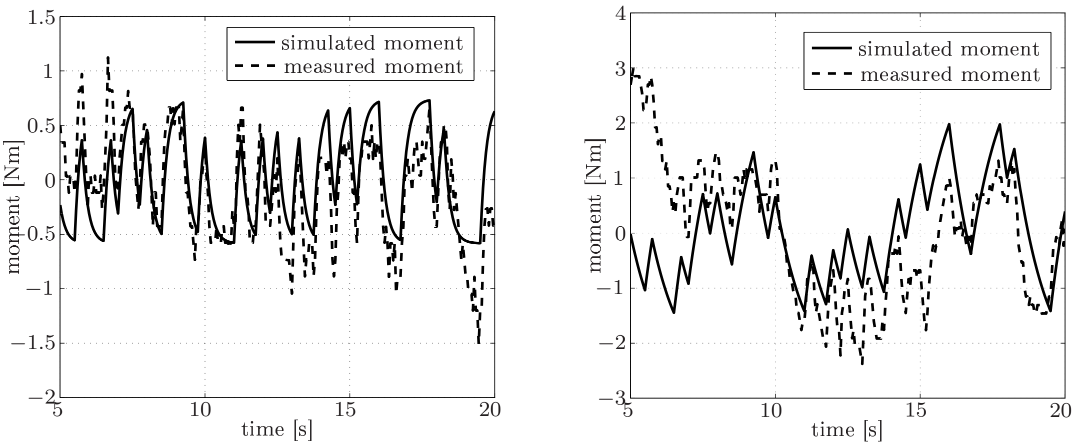
Figure 4.16.: Resulting total moments after the subject muscles were stimulated with different PRBS signals.

For the model identification the moment measurements recorded during the first 5 seconds are ignored. This way the transient behaviour of the muscels is omitted for identification. For the estimation of the models the least squares estimation method is applied. Table 4.5 shows the identification results for two operating points using two different PRBS signals with the mean pulsewidths of 200 and 250 μs, respectively.

mean pulsewidth [μs]	transfer function $G_z(z)$	rise time [s]	static gain [Nm/μs]
op. 1: 200	$\frac{0.003979z}{z^2-0.8306z+0.04126}$	0.4380	0.0189
op. 2: 250	$\frac{0.006756z}{z^2-1.554z+0.6472}$	0.2834	0.0724

Table 4.5.: Models identified at two different operating points (abbreviated by ”op“). The gray shaded row shows the results with the highest static gain (model 2).

Figure 4.17 shows a comparison of results of the simulated and the actually measured moments using the models shown in table 4.5 with the mean values being removed. As input signals the PRBS signals shown in figures 4.16(a) and 4.16(b) for models 1 and 2, respectively, are used.



(a) Simulated and measured moments for operating point 1 using the input signal shown in figure 4.16(a).

(b) Simulated and measured moments for operating point 2 using the input signal shown in figure 4.16(b).

Figure 4.17.: Simulated and measured moments for two operating points with the mean value being removed.

Although the output of the model 2 (see figure 4.17(b)) does not match with the real measurement as well as the model identified at operating point 1 does, the model identified at operating point 2 is chosen for all further experiments. The reason for this choice is to ensure robust stability of the moment loop for varying stimulation levels due to its higher static gain compared to the static gain of model 1.

Controller Parameters

For the design of the moment controller the pole placement technique is used. The starting point for this approach is the control structure shown in figure 4.18 (adapted from [118]).

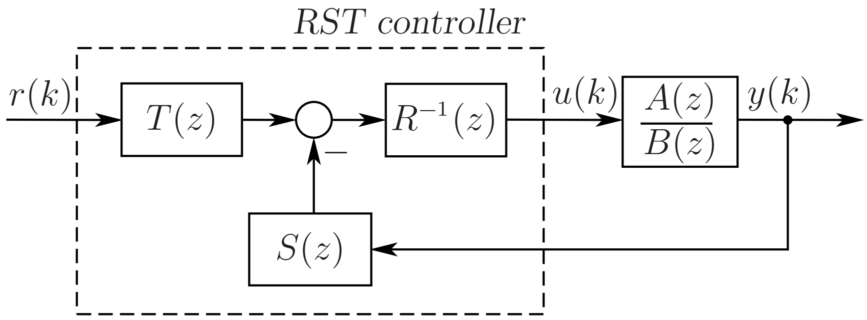


Figure 4.18.: An RST-control structure with the discrete reference $r(k)$, the discrete input $u(k)$, and the output $y(k)$ and their respective weighting functions $R(z)$, $S(z)$, and $T(z)$ used for controlling the system. The model to be controlled is described by the polynomials $A(z)$ and $B(z)$.

The idea behind the pole placement technique lies in defining the close-loop behaviour by corresponding poles and zeros. As a result control parameters have to be computed which satisfy the conditions set out for the closed-loop system. The plant is defined by polynomials $A(z)$ and $B(z)$ which describe the input/output behaviour of the system to be controlled.

$$A(z) \cdot y(k) = B(z) \cdot u(k) \tag{4.2}$$

The plant input $u(k)$ as shown in figure 4.18 can be computed in the following way:

$$u(k) = R^{-1}(z) \cdot (T(z) \cdot r(k) - S(z) \cdot y(k)) \tag{4.3}$$

with the discrete reference $r(k)$, the discrete input $u(k)$, the output $y(k)$ as well as their respective weighting functions $R(z)$, $S(z)$, and $T(z)$ used to control the system. By replacing $u(k)$ in equation 4.2 with equation 4.3 it follows that:

$$(A(z) \cdot R(z) + B(z) \cdot S(z)) \cdot y(k) = B(z) \cdot T(z) \cdot r(k) \tag{4.4}$$

The characteristic polynomial A_{cl} of the closed-loop system is:

$$A_{cl}(z) = (A(z) \cdot R(z) + B(z) \cdot S(z)). \tag{4.5}$$

Equation 4.5 is also called the *Diophantine equation*. Let $A(z)$, $B(z)$ and $A_{cl}(z)$ be polynomials with real coefficients, which are known. Then the Diophantine equation has a solution if and only if the greatest common factor of $A(z)$ and $B(z)$ divides $A_{cl}(z)$ [118]. Furthermore, there exist unique solutions to equation 4.5 such that

$$\begin{aligned} degR &< degB \text{ or} \\ degS &< degA. \end{aligned} \tag{4.6}$$

Consequently, there are infinitely many solutions to the Diophantine equation. In order to reduce the amount of possible solutions some constraints are introduced which have to be fulfilled by the solution as well.

One constraint is the satisfaction of causality. It is assumed that the model is causal which is expressed by the fact that $deg\ B \leq deg\ A$. As the controller is required to be causal as well, it follows that $deg\ S \leq deg\ R$. If $deg\ A = n$ then the maximal degree of S has to be $n - 1$ in order to fulfil the condition set out in 4.6 [118]. The identified muscle model is described by a polynomial A, which has the degree $n = 2$, and a polynomial B, which has the degree 1 (see polynomials in table 4.5). Consequently, the maximal degree of S has to be 1. Since the muscle contains no inherent integrating behaviour, an integrator must be incorporated in the controller otherwise a control error would stay. This means, the degree of the controller polynomials is increased by 1 and has therefore a degree of $n = 2$.

In a standard pole placement design using an RST-controller (see figure 4.18) the desired closed loop characteristic polynomial $A_{cl}(z)$ is split into a controller polynomial A_c and an observer polynomial A_o [118].

According to the suggestions of Jaime [38] for the placement of the poles of the polynomial A_c a rise time of $t_c = 0.5$ s and a damping value of $\zeta_c = 1$ was chosen, while for the placement of the poles of A_o a rise time of $t_o = 0.3$ s and a damping value of $\zeta_o = 1$ was used. Together with the muscle model from table 4.5 following polynomials were computed, which incorporate integral behaviour and are denoted as $\tilde{R}(z)$, $\tilde{S}(z)$, and $\tilde{T}(z)$:

$$\begin{aligned}\tilde{R}(z) &= z^2 - 1.2809 \cdot z + 0.28087 \\ \tilde{S}(z) &= z^2 - 1.6374 \cdot z + 0.6989 \\ \tilde{T}(z) &= z^2 - 1.1736 \cdot z + 0.3445\end{aligned}$$

These polynomials stayed always the same throughout all the experiments which were carried out.

For further details concerning the method of pole placement and the solution of the Diophantine equation see also [118].

4.7.6. Experimental Protocol

Prior to the experiments the subject was asked to stimulate his shank muscles at home over several weeks in order to build up his muscle strength. The data necessary for modelling the muscles (see section 4.7.5) were acquired after the subject had built up his muscle strength. In order to reduce the rate of muscle fatigue, an experimental trial is carried out for only 20 seconds (instead of 60 seconds as with the able bodied subjects).

Contrary to the experiments with able-bodied subjects the controller parameters are scaled using only the scaling values 1 and 0.4. Together with the additional constant supporting stiffness of the MRF (5 Nm/deg) the subject is performing the experiments with the values set out in table 4.6.

$1 \cdot k_s$ [Nm/deg]	$0.4 \cdot k_s$ [Nm/deg]	$1 \cdot k_v$ [Nm/deg s]	$0.4 \cdot k_v$ [Nm/deg s]
5 + 5 (MRF)	2 + 5 (MRF)	0.5	0.2

Table 4.6.: Stiffness and viscosity values used for the paraplegic subject.

For the perturbation, again a PRBS signal with the two amplitudes of 10 Nm ("low perturbation") and 15 Nm ("high perturbation") is used. The experiments could only be carried out once a week due to the availability of the subject and were conducted over 6 consecutive weeks performing 6 test sessions in total. Throughout the week between the test sessions the subject was asked to stimulate his shank muscles at home in order to maintain muscle strength. Only during the week before session 5 the subject was not able to stimulate his muscles on a regular basis.

During a session the subject performs trials using two different stiffness and two different stiffness-viscosity controllers. With each of these four controllers two different perturbation signals are applied resulting in a total of 8 trials. The order of the different controllers is random and varies from session to session in order to minimise the effect of learning. The resting time between trials is just long enough to save measured data of a performed trial and to set the necessary control values for the next trial. Overall one experimental session takes around six minutes.

Prior to an experimental session the subject stands for about 20 minutes in an ordinary passive standing frame to stretch the leg muscles. As part of the "warm-up" and in order to acquaint the subject to the balancing he stands for a few minutes in the MRF without stimulation using only the support of the hydraulic actuators of the MRF.

From time to time during a trial the subject lost his balance. As soon as this happened and he could not return to the upright position he was pulled back thereto by the experimenter and the subject continued with the ongoing trial.

4.8. Results (SCI Subject)

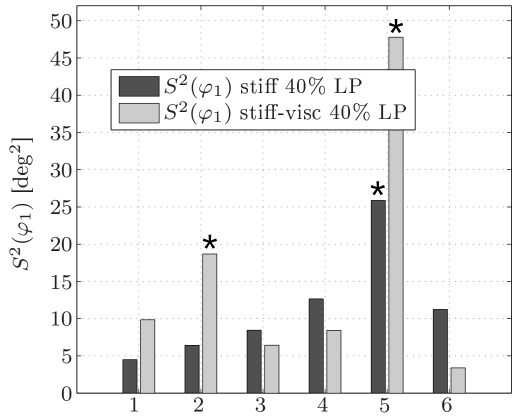
4.8.1. Performance evaluation

For the evaluation of performance of the spinal cord injured subject during standing the time signals of the inclination angles of the lower body (φ_1) and upper body (φ_2), the measured torque applied onto the ground (τ_{meas}), and the applied stimulation pulsewidth (pw) are used applying the approach for evaluation introduced in section 4.3. The results of the trials, during which a loss of balance occurred, are marked by a star (see i.e. figure 4.19).

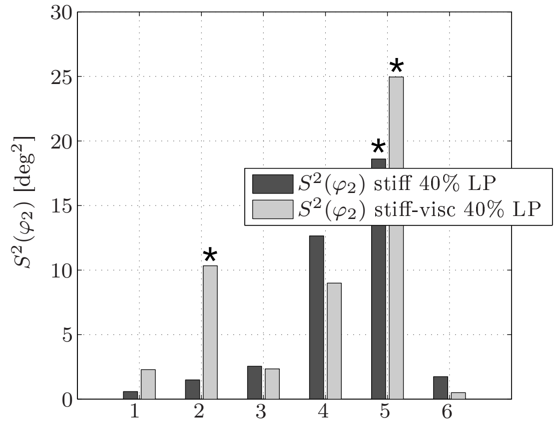
A performance is seen as "good" when the evaluation value for the pulsewidth ($S^2(pw)$) and inclination angle of the lower body ($S^2(\varphi_1)$) is comparatively small and, at the same time, high for the inclination angle of the upper body ($S^2(\varphi_2)$). This indicates small movements of the lower body (with only small variations in stimulation intensity), together with involvement of the upper body in the balancing task.

Scenarios with scaling factor 0.4 and low perturbation

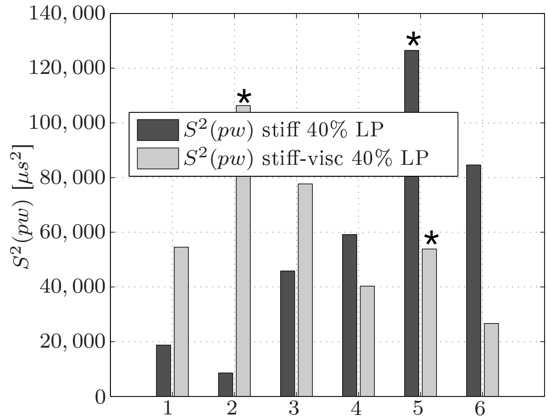
Figure 4.19 shows the results for the variance S^2 while the scaling factor 0.4 and low perturbation are applied ("40%LP" scenario).



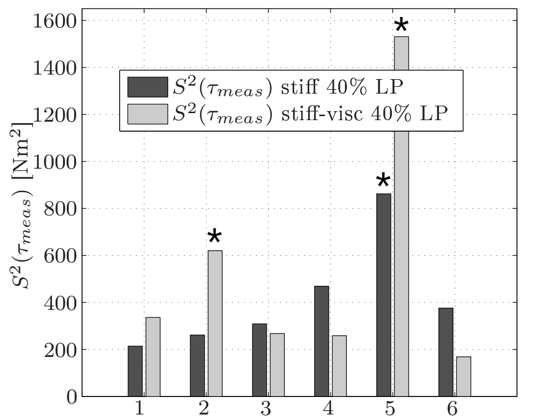
(a) Results for φ_1 at 40%LP.



(b) Results for φ_2 at 40%LP.



(c) Results for the pulsewidth at 40%LP.



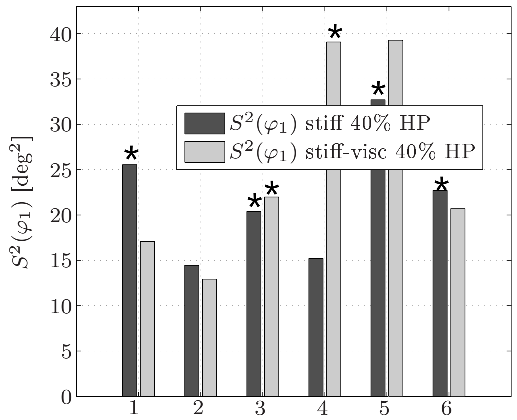
(d) Results for τ_{meas} at 40%LP.

Figure 4.19.: Comparison of evaluation values of all signals during the 40%LP scenario for the spinal cord injured subject. The stars mark a loss of balance.

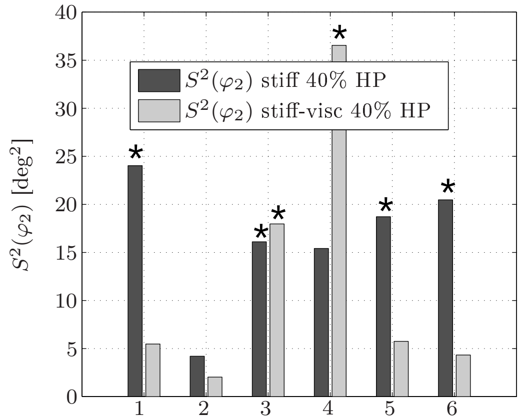
In figure 4.19 $S^2(\varphi_1)$, $S^2(\varphi_2)$, and $S^2(\tau_{meas})$ indicate the loss of balance during the second and fifth training session using a stiffness-viscosity controller with comparably high values. The evaluation value of the stimulation signal reflects this fact only for the second but not for the fifth session.

Scenarios with scaling factor 0.4 and high perturbation

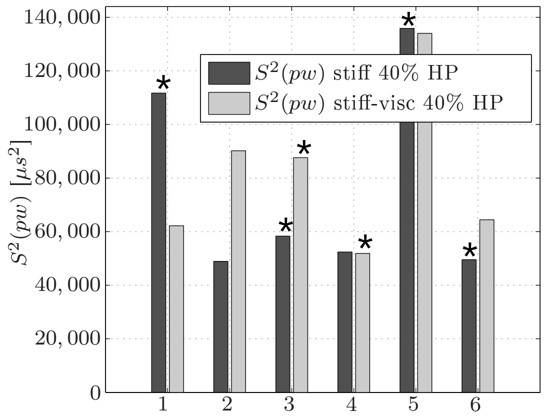
Figure 4.20 shows the results for the variance S^2 while the scaling factor 0.4 and high perturbation are applied ("40%HP" scenario).



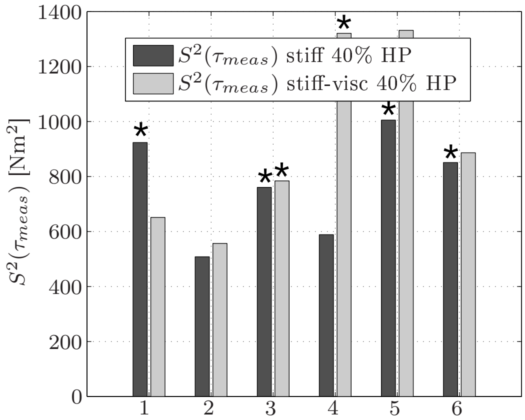
(a) Results for φ_1 at 40%HP.



(b) Results for φ_2 at 40%HP.



(c) Results for the pulsewidth at 40%HP.



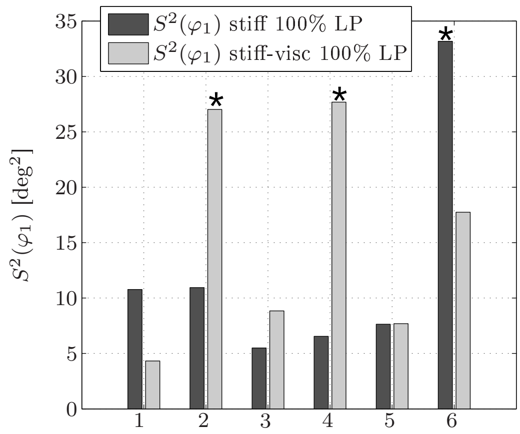
(d) Results for τ_{meas} at 40%HP.

Figure 4.20.: Comparison of evaluation values of all signals during the 40%HP scenario for the spinal cord injured subject. The stars mark a loss of balance.

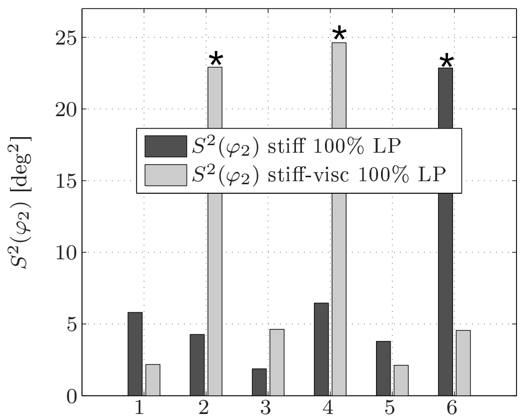
With the high perturbation the subject lost his balance more often. When using the stiffness controller a loss of balance appeared during the first, third, fifth, and sixth session, whereas using the stiffness-viscosity controller the subject lost his balance during the third and fourth session.

Scenarios with scaling factor 1 and low perturbation

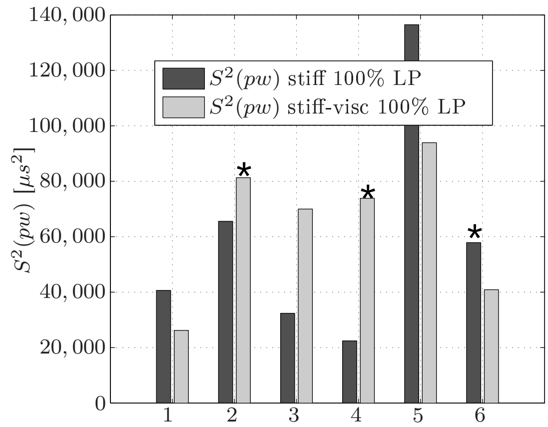
Figure 4.21 shows the results for the variance S^2 while the scaling factor 1 and low perturbation are applied ("100%LP" scenario).



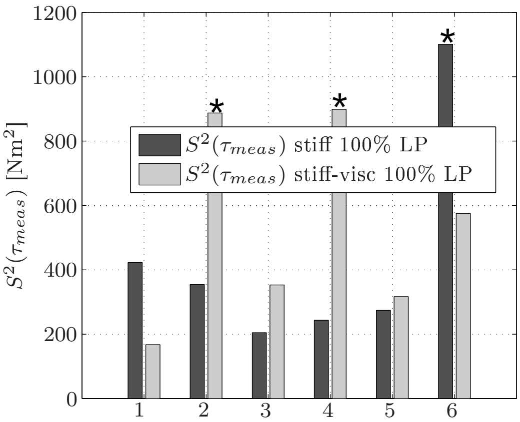
(a) Results for φ_1 at 100%LP.



(b) Results for φ_2 at 100%LP.



(c) Results for the pulsewidth at 100%LP.



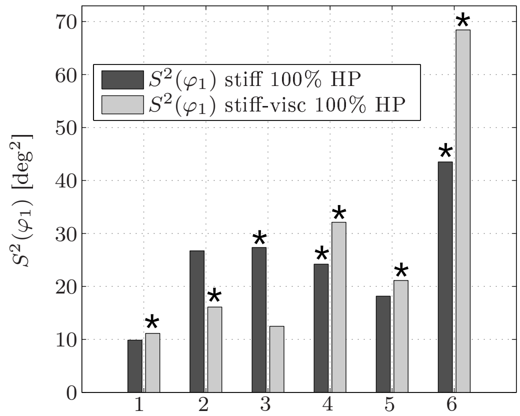
(d) Results for τ_{meas} at 100%LP.

Figure 4.21.: Comparison of evaluation values of all signals during the 100%LP scenario for the spinal cord injured subject. The stars mark a loss of balance.

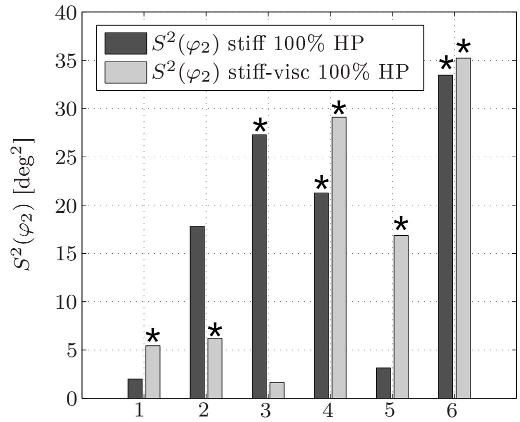
During the 100%LP scenario the subject lost his balance with the stiffness controller only during the last experimental session whereas with the stiffness-viscosity controller this happened during the second and fourth session.

Scenarios with scaling factor 1 and high perturbation

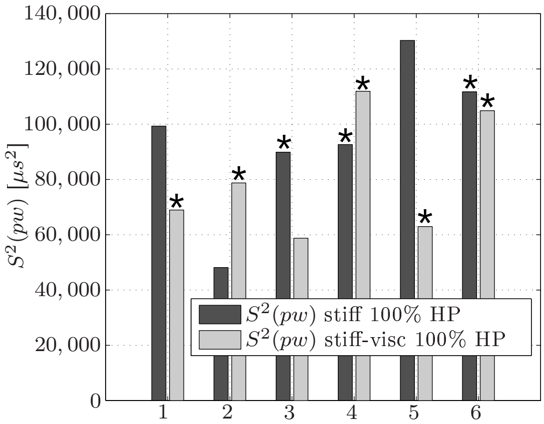
Figure 4.22 shows the results for the variance S^2 while the scaling factor 1 and high perturbation are applied ("100%LP" scenario).



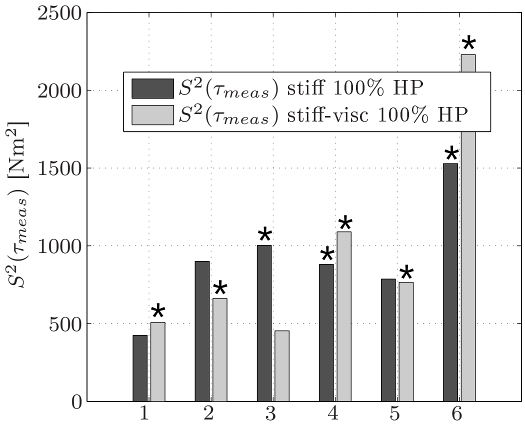
(a) Results for φ_1 at 100%HP.



(b) Results for φ_2 at 100%HP.



(c) Results for the pulsewidth at 100%HP.



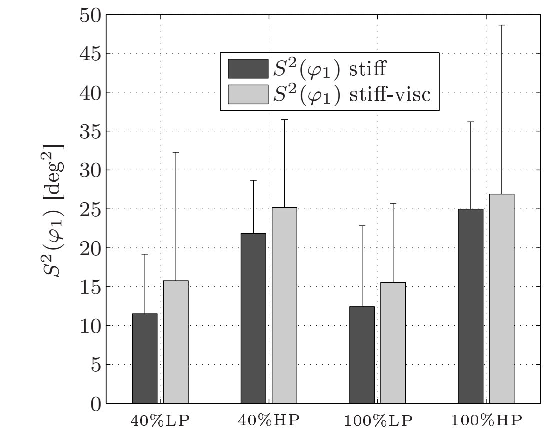
(d) Results for τ_{meas} at 100%HP.

Figure 4.22.: Comparison of evaluation values of all signals during the 100%HP scenario for the spinal cord injured subject. The stars mark a loss of balance.

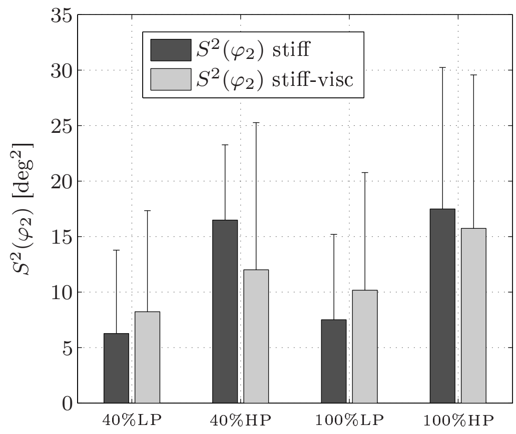
With the stiffness controller the subject lost his balance during the third, fourth, and sixth session, whereas with the stiffness-viscosity controller the subject lost his balance during every session except during the third.

4.8.2. Comparison of different stiffness and stiffness-viscosity controllers

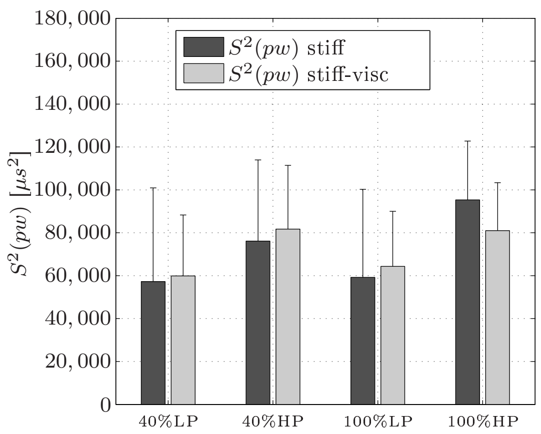
Figure 4.23 shows the average values of $S^2(\varphi_1)$, $S^2(\varphi_2)$, $S^2(\tau_{meas})$, and $S^2(pw)$ for each scenario with their standard deviations. These averaged values are calculated using the results presented in section 4.8.1.



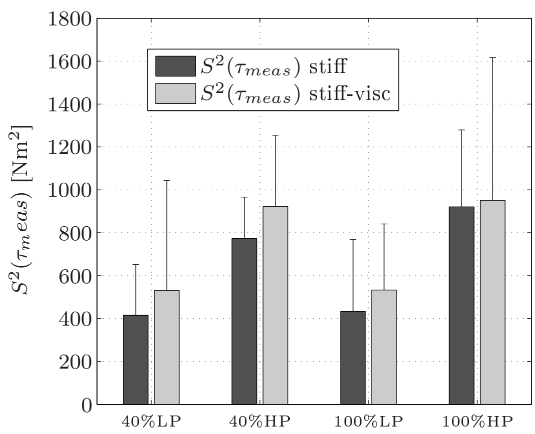
(a) Averaged results for φ_1.



(b) Averaged results for φ_2.



(c) Averaged results for the pulsewidth.



(d) Averaged results for τ_{meas}.

Figure 4.23.: Averaged variances and standard deviations for the spinal cord injured subject.

The comparison of the variances shows for the scenarios with low perturbation

(40%LP and 100%LP) a lower value than for the respective scenarios with high perturbation. Further, the variances for the trials using a stiffness-viscosity controller are in general higher compared to the results achieved with a simple stiffness controller. Additionally, the variances for the scenarios 40%LP and 100%LP differ not very much from each other. This is also the case for the scenarios 40%HP compared to 100%HP.

4.9. Discussion and Conclusions (SCI Subject)

In this chapter an alternative way of evaluating the performance of a person during standing was introduced. Experiments were performed with able-bodied and paraplegic subjects. For the evaluation of the performance of a person during standing the variance of a measured signal was used. In section 4.6 the results of the performance of able-bodied subjects were discussed. The results showed that the variance is suitable for evaluating the performance of an able-bodied subject during standing in general. It is also possible to compare the performance achieved with different control approaches of one subject as well as compare performances between different subjects.

Similar experiments have been carried out with a SCI subject as with the able-bodied subjects. The SCI subject was standing in the MRF and his posture was controlled with the help of a cascaded control structure using either a stiffness or a stiffness-viscosity controller in the outer loop and a moment controller in the inner loop. For the stabilisation of the subject functional electrical stimulation was applied to the shank muscles of the subject. The subject was perturbed by a perturbation in PRBS form. The resulting inclination angles of the lower and upper body as well as the moment applied onto the ground and the pulse width of the stimulation signal were then evaluated.

Contrary to the able-bodied subjects the paraplegic subject lost sometimes his balance during an experimental trial. As the variance of a signal is used for evaluation, a comparatively high value is expected when the subject is loosing his balance. In general the results reflect correctly this fact. Only sometimes the evaluation value shows higher values when no loss of balance occurred compared to trials where the subject lost his balance. This can be seen, for example, in figure 4.19(b) where the results with the stiffness controller show a higher value for the fourth trial without losing the balance, compared to the results of the second trial, where he lost the balance using a stiffness-viscosity controller. This also can be seen in figure 4.19(c) where the results with the stiffness controller show a higher value for the sixth trial without losing the balance, compared to the results of the fifth trial achieved with a stiffness-viscosity controller, where he lost the balance. For the 40%HP (see figure 4.20) and 100%LP-scenarios (see figure 4.21) only the results of the pulsewidth do not always reflect the loss of a balance with a high value. For the 100%HP-scenario (see figure 4.22) $S^2(\varphi_1)$, $S^2(\varphi_2)$, and $S^2(\tau_{meas})$ show for the second trial a higher value when using a stiffness controller where no loss of balance occurred compared to the results achieved during the same trial with a stiffness-viscosity controller where the subject lost his balance.

The reason for such outcome is that during the experimental trials with no loss of balance but comparatively high evaluation values the subject swayed with a high amplitude around the upright position managing to keep his balance whereas during the trials where the subject lost his balance and the evaluation value shows a comparatively small value the subject leaned during the course of balancing already in one direction when the perturbation pushed him even further in this same direction letting him lose his balance fairly quickly without much swaying.

The comparison of results achieved for different perturbation amplitudes (see figure 4.23) shows higher values for scenarios with a high perturbation applied, independently whether a stiffness or a stiffness-viscosity controller was used. This observation was also made with the able-bodied subjects.

Contrary to the results of able-bodied subjects, the performance of the SCI subject using a stiffness controller was basically the same as with a stiffness-viscosity controller for the same perturbation strength. As the experiments were carried out with only one SCI subject, there is not enough data for deriving a trend. Another role play the chosen stiffness and viscosity values. In the course of finding the appropriate supportive torques provided by the MRF a relatively high value was used as the muscles were not able to provide the sufficient stabilising moment for the whole duration of the experimental session due to fatigue. With these higher value, however, the subject made the impression of being not really challenged. Therefore the supportive moment was reduced in order to give the subject some enjoyment and also allow him to train his muscles more effectively.

In conclusion it can be said that the variance as an alternative evaluation value is also suitable to be used for evaluating the performance of a SCI subject during standing. Furthermore, it is possible to qualify a performance as good or bad by the analysis of the stimulation pulsewidth and the inclination angles of the upper and lower body. A subject has performed well, when the evaluation values for the inclination angle of the lower body and the pulsewidth of the stimulation signal are comparatively small and, at the same time, the evaluation value for the inclination angle of the upper body is high.

For the experiments shown in this chapter the voluntary movement of the upper body is assumed to be an unknown disturbance which has to be counterbalanced. If the movement of the upper body, however, is estimated and used for the calculation of the amount of stimulation actually needed, the subject might perform better by working more with the upper body and less with the lower body. This approach might reduce the amount of stimulation as the subject is able to compensate most of the perturbations applied by actively moving more with his upper body. The results of this approach are presented in chapter 5.

5. Integrated Voluntary Control

5.1. Summary

A new control approach is suggested which allows a paraplegic subject to stand in a more natural way by combining artificial with natural control.

For this approach the subject is standing in the MRF and kept in a double-link inverted pendulum configuration which allows the subject to move his upper body voluntarily. With the measurements of the inclination angles of the upper and lower body and the help of a double-link inverted pendulum model the torque applied onto the ground is estimated. This estimated torque is then used in an artificial controller controlling a stimulation signal applied to the gastrocnemius muscles.

The performance using this **I**ntegrated **V**oluntary **C**ontrol (IVC) approach is then compared with the performance achieved with a control approach which does not integrate the voluntary movement of the subject in the control. The comparison of these two different approaches shows no significant difference in performance.

5.2. Background

The improvement of control systems, which use additional control input from a human operator, started to draw interest during World War II, when engineers and psychologists attempted to improve the performance of pilots, gunners, and bombardiers. To design satisfactory manually controlled systems these researchers began analysing the neuromuscular characteristics of the human operator. Their approach was to consider the human as an intermittent correction servomechanism with a well-defined input and output interpreting the human as an error-activated compensation element. To describe the behaviour of human operators the same linear, constant-coefficient differential equations as for linear servomechanisms were applied [119, 120].

Many authors suggest to use neural networks for modelling the human behaviour performing certain controlling tasks in systems with human-machine interactions. Nechyba *et al.* [121], for example, describe methods for modelling human controlling behaviour with the help of learning cascade neural networks which model thereby the behaviour of the human brain carrying out certain control tasks. With this approach it is possible to analyse human behaviour more easily, to improve human-robot coordination, to train other humans in simulators, and to develop human-like intelligent machines. These insights are intended to be applied in areas as human-machine interfaces, space telerobotics, agile manufacturing, and others.

Robots and machines are nowadays also used in rehabilitation of patients. These devices support the patients in the process of relearning certain movement patterns which are impaired due to neural injury. Examples include the MIT-Manus [122–124] which assists the rehabilitation of elbow and shoulder movement in stroke patients, the gait trainer which allows chronic stroke and paraplegic patients to train gait-like movement [125, 126], and the Lokomat [127], a robotic orthosis supporting spinal cord injured and chronic stroke patients during treadmill training rehabilitation. Devices specific for balance retraining include balance platforms (such as the BalanceMaster and the Biodex Balance System) which are based on a moving standing platform combined with biofeedback, and the BalanceTrainer [56] which is a dynamic standing frame allowing balance training and step-like movements. These devices allow the patients to train in a repetitive manner the movement of healthy people.

5.3. Methods

This section describes a control approach which integrates the voluntary movement of a SCI subject during standing into the control of his posture.

All experimental procedures were approved by the Ethics Committee of the Faculty of Biological and Life Sciences (FBLS) at the University of Glasgow and the subject provided written, informed consent prior to participation.

5.3.1. Setup

For the experiments the MRF is used (see section 3.2 and figure 3.3(b)). The subject is kept in a double-link inverted pendulum configuration with the knee and hip joints being mechanically locked and allowing voluntary movement of the upper body around the lumbar joint just above the hip. The necessary stabilising supporting torque at the ankle is provided by functional electrical stimulation of the paralysed shank muscles and is controlled artificially. Although the paralysed muscles are able to provide a stabilising moment with the help of stimulation, they fatigue quick and can not stabilise the subject for the length of an experimental session. Therefore, an additional supportive stiffness of 5 Nm/deg is applied by the hydraulic actuators of the MRF. With this supportive stiffness alone, the subject is only stable during quite standing but not during perturbed standing. This supporting stiffness stays constant and is applied throughout the whole experimental session. The inclination angles of the upper and lower body as well as the torque at the ankle are measured.

5.3.2. Subject

The experiments are performed with one male subject, aged 33, with a motor-complete spinal cord injury at level T9 and 8 years post injury.

5.3.3. Control Structure

For the integrated voluntary control structure a different way of estimating the additional control input of the subject is proposed. As the subject itself is the plant to be controlled a controller is used which is based on the mechanical model of the subject .

Matjačić *et al.* [59] and Jaime *et al.* [105] suggested control strategies which would allow paraplegic subjects to stand in the MRF and being able to move the upper body freely. These authors, however, did not incorporate the voluntary movement of the subject in the control action and considered it merely as an unknown input disturbing the control system. For the stabilisation of the person in the upright standing position functional electrical stimulation was applied to the shank muscles.

Originating from the control structure suggested by Matjačić *et al.* and Jaime *et al.* a controller is proposed which takes into account the inclination angles φ_1 and φ_2 for the estimation of the torque τ_{ref} applied onto the ground. This modified control structure is shown in figure 5.1.

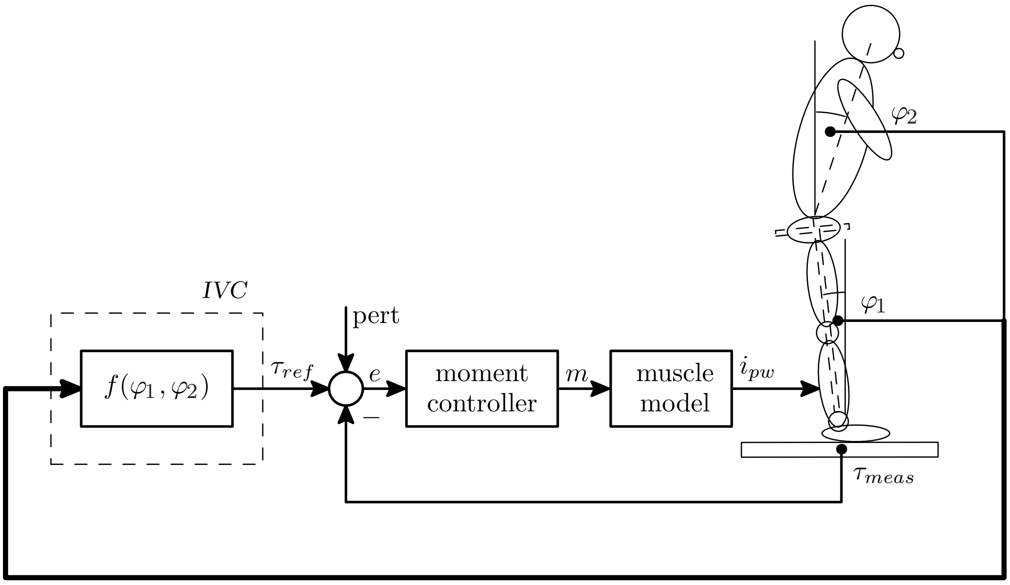


Figure 5.1.: IVC structure using a cascaded control loop with a moment controller in the inner loop.

Double Link Inverted Pendulum Model

For the prediction of the torque τ_{ref} (see figure 5.1) a mechanical model of the subject standing in the frame is used. The most suitable model would be a double inverted pendulum model representing the legs as the lower and the upper body as the upper link. The active parts of this model are the ankle and the lumbar joint where counter-balancing torques are produced due to the contraction of the shank muscles and the voluntary movement of the upper body.

In figure 5.2 a double inverted pendulum model is depicted:

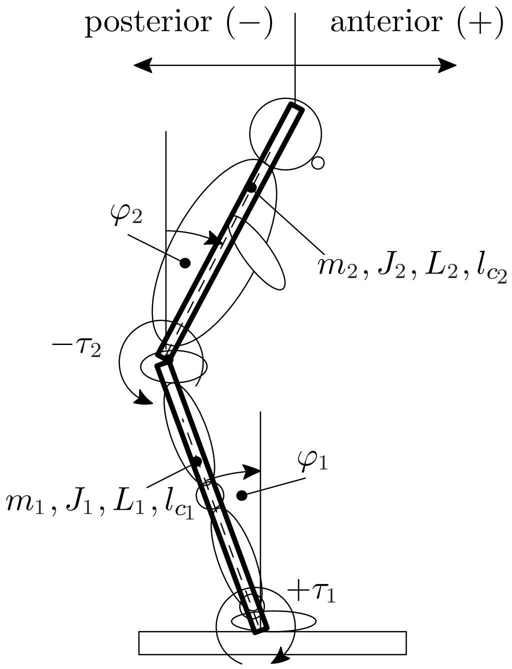


Figure 5.2.: Double inverted pendulum model of a human body with the legs as the lower link and the upper body as the upper link.

φ_1 and φ_2, as denoted in figure 5.2, describe the angles of the lower and upper body relative to the upright position. Torque τ_1 is applied at the ankle either by muscle force or by hydraulic actuators and τ_2 is the torque voluntarily applied by the subject. m_1 and m_2 are the masses, J_1 and J_2 the moments of inertia, l_{c_1} and l_{c_2} the distances of the centre of mass from the joints, and L_1 and L_2 are the lengths of the lower and upper body, respectively. The states of the system are the two angles φ_1 and φ_2 and their first derivative $\dot{\varphi}_1$ and $\dot{\varphi}_2$,respectively. The nonlinear equations of motion were derived using the Newton-Euler method (see also [59]).
Equation 5.1 describes the effect of the torque applied at the ankle

$$\begin{aligned}\tau_1 = &- m_2 g l_{c_2} \sin\varphi_2 - (m_1 g l_{c_1} + m_2 g l_1) \sin\varphi_1 \\ &- m_2 l_1 l_{c_2} \sin(\varphi_2 - \varphi_1) \left(\dot{\varphi}_2^2 - \dot{\varphi}_1^2\right) \\ &+ \left[J_2 + m_2 l_{c_2}^2 + m_2 l_1 l_{c_2} \cos(\varphi_2 - \varphi_1)\right] \ddot{\varphi}_2 \\ &+ \left[J_1 + m_1 l_{c_1}^2 + m_2 l_1^2 + m_2 l_1 l_{c_2} \cos(\varphi_2 - \varphi_1)\right] \ddot{\varphi}_1\end{aligned} \tag{5.1}$$

and equation 5.2 describes the effect of the torque produced voluntarily by the subject

$$\begin{aligned} \tau_2 = &- m_2 g l_{c_2} \sin\varphi_2 + m_2 l_1 l_{c_2} \sin(\varphi_2 - \varphi_1)\,\dot{\varphi}_1^2 \\ &+ \left[J_2 + m_2 l_{c_2}^2\right]\ddot{\varphi}_2 + m_2 l_1 l_{c_2} \cos(\varphi_2 - \varphi_1)\,\ddot{\varphi}_1. \end{aligned} \tag{5.2}$$

The most convenient and energy efficient posture is the upright position, i.e. $\varphi_1 = 0$ and $\varphi_2 = 0$. For linearisation around these angles it is assumed that $\dot{\varphi}_1^2 \approx 0$ and $\dot{\varphi}_2^2 \approx 0$. The sin- and cos-functions can be approximated by following series:

$$\begin{aligned} &\sin\alpha = \alpha - \frac{\alpha^3}{3!} + \frac{\alpha^5}{5!} - \frac{\alpha^7}{7!} + \ldots + (-1)^n \frac{\alpha^{2n+1}}{(2n+1)!} + \ldots \\ &\cos\alpha = 1 - \frac{\alpha^2}{2!} + \frac{\alpha^4}{4!} - \frac{\alpha^6}{6!} + \ldots + (-1)^n \frac{\alpha^{2n}}{(2n)!} + \ldots; \text{ with } \alpha = \varphi_1,\, \varphi_2 \end{aligned}$$

Depending on how accurate the linear model has to be the sin- and cos-functions can be approximated by one or more members of the series. In this application it is sufficient to approximate the sin- and cos-functions by the first member of a series. For an error of 1% of α the approximation $\sin\alpha \approx \alpha$ can be used for a range of $\pm 14^\circ$ of α. The approximation $\cos\alpha \approx 1$ shows a 1%-error for the range of $\pm 8.1^\circ$ of α (see [116]).

With these modifications the linearised model can be written in following state space notation:

$$\begin{aligned} \dot{x} &= Ax + Bu \\ y &= Cx + Du \end{aligned} \tag{5.3}$$

with

$$
A = \frac{1}{\Delta}\begin{bmatrix} 0 & \Delta & 0 & 0 \\ c \cdot f & 0 & d \cdot (f-b) & 0 \\ 0 & 0 & 0 & \Delta \\ -c \cdot e & 0 & d \cdot (a-e) & 0 \end{bmatrix},
$$

$$
B = \frac{1}{\Delta}\begin{bmatrix} 0 & 0 \\ f & -b \\ 0 & 0 \\ -e & a \end{bmatrix}, \quad C = \begin{bmatrix} 1 & 0 & 0 & 0 \\ 0 & 1 & 0 & 0 \\ 0 & 0 & 1 & 0 \\ 0 & 0 & 0 & 1 \end{bmatrix},
$$

$$
D = \begin{bmatrix} 0 \\ 0 \\ 0 \\ 0 \end{bmatrix}, \quad \Delta = a \cdot f - b \cdot e
$$

and

$$
\begin{aligned}
a &= J_1 + m_1 l_{c_1}^2 + m_2 l_1^2 + m_2 l_1 l_{c_2} \\
b &= J_2 + m_2 l_{c_2}^2 + m_2 l_1 l_{c_2} \\
c &= m_1 g l_{c_1} + m_2 g l_1 \\
d &= m_2 g l_{c_2} \\
e &= m_2 l_1 l_{c_2} \\
f &= J_2 + m_2 l_{c_2}^2 \\
x &= \begin{bmatrix} \varphi_1 - \varphi_{1_0} \\ \dot{\varphi}_1 \\ \varphi_2 - \varphi_{2_0} \\ \dot{\varphi}_2 \end{bmatrix}, \quad u = \begin{bmatrix} \tau_1 - \tau_{1_0} \\ \tau_2 - \tau_{2_0} \end{bmatrix}
\end{aligned} \tag{5.4}
$$

with the initial states defined as: $\varphi_{1_0} = 0$, $\varphi_{2_0} = 0$, $\tau_{1_0} = 0$, and $\tau_{2_0} = 0$. The transfer function $G(s)$ of the linearised model can be derived as

$$
\begin{aligned}
G(s) &= \underbrace{C}_{I} (sI - A)^{-1} B + \underbrace{D}_{0} \\
&= (sI - A)^{-1} B
\end{aligned} \tag{5.5}
$$

The characteristic polynomial $CP(s)$ is the determinant of the matrix $(sI - A)$.

$$\det(sI - A) = \frac{1}{\Delta}\left(\Delta s^4 + \left(d\left(e-a\right) - cf\right)s^2 + dc\right) := \frac{1}{\Delta}CP(s) \tag{5.6}$$

with Δ from equation 5.4. Consequently, equation 5.5 has following solution:

$$G = \frac{1}{CP(s)}\begin{bmatrix} fs^2 - d & -bs^2 + d \\ -es^2 & as^2 - c \end{bmatrix} \tag{5.7}$$

τ_1 and τ_2 are the torques at the ankle and at the lumbar joint, respectively, with the resulting output values being the inclination angles of the lower (φ_1) and upper body (φ_2) of the subject. The outputs $\dot{\varphi}_1$ and $\dot{\varphi}_2$ (see also equation 5.3) are not considered due to cancellations in the course of computing the transfer function G. Therefore the transfer function is only a 2-by-2 matrix.

The two links are interconnected with each other in such a way that input one (torque at the ankle) has an impact on the upper limb as input two (voluntarily applied torque by the subject) has on the lower limb. This is shown in figure 5.3.

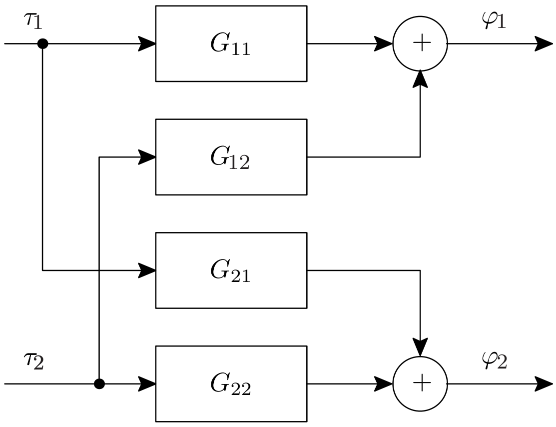


Figure 5.3.: Relationship between the two inputs and two outputs of the system described by the transfer functions G_{11} to G_{22}

From figure 5.3 the following relationship between in- and output can be derived:

$$\begin{bmatrix} \varphi_1 \\ \varphi_2 \end{bmatrix} = \underbrace{\begin{bmatrix} G_{11} & G_{12} \\ G_{21} & G_{22} \end{bmatrix}}_{G}\begin{bmatrix} \tau_1 \\ \tau_2 \end{bmatrix} \tag{5.8}$$

The equations 5.1–5.8 describe the resulting angular changes of the lower and upper body of the subject caused by the torques applied at the ankle and the lumbar joint. By comparing equation 5.7 with equation 5.8 it follows that:

$$G_{11} = \frac{fs^2 - d}{CP}; \qquad G_{12} = \frac{-bs^2 + d}{CP};$$

$$G_{21} = \frac{-es^2}{CP}; \qquad G_{22} = \frac{as^2 - c}{CP}.$$

In order to use this mathematical description for the integrated voluntary control approach the torques applied at the lumbar joint and at the ankle have to be known. The direct measurement of the applied torque at the lumbar joint is not possible. The only torque measurements possible are acquired from the force plates the subject is standing on measuring the moment applied by the subject onto the ground. On the other hand the outputs of the linearised model for the human being, i.e. the inclination angles of the upper and lower body, can be measured.

To calculate the possible input values which result in the measured output values the following equation can be evaluated:

$$\begin{bmatrix} \tau_1 \\ \tau_2 \end{bmatrix} = G^{-1} \cdot \begin{bmatrix} \varphi_1 \\ \varphi_2 \end{bmatrix} \tag{5.9}$$

With the use of equations 5.6 and 5.8 the inverse of the matrix G with the transfer functions G_{11} to G_{22} results in following new matrix:

$$\begin{aligned} G^{-1} &= \frac{1}{G_{11} \cdot G_{22} - G_{21} \cdot G_{12}} \begin{bmatrix} G_{22} & -G_{12} \\ -G_{21} & G_{11} \end{bmatrix} \\ &= \frac{CP(s)}{\underbrace{\left(fs^2 - d\right)\left(as^2 - c\right) - \left(bs^2 - d\right)\left(-es^2\right)}_{CP(s)}} \begin{bmatrix} as^2 - c & bs^2 - d \\ es^2 & fs^2 - d \end{bmatrix} \end{aligned} \tag{5.10}$$

Consequently equation 5.9 results in

$$\begin{bmatrix} \tau_1 \\ \tau_2 \end{bmatrix} = \begin{bmatrix} as^2 - c & bs^2 - d \\ es^2 & fs^2 - d \end{bmatrix} \begin{bmatrix} \varphi_1 \\ \varphi_2 \end{bmatrix} \tag{5.11}$$

Due to the inversion of the transfer function the system described by equation 5.11 is now not proper. To make the system proper the ideal differentiator s in $G^{-1}(i,j)$ is replaced by a realisable differentiator $s/(T_d s+1)$ where T_d is small. With these considerations the system of equation 5.11 becomes

$$\begin{bmatrix} \tau_1 \\ \tau_2 \end{bmatrix} = \begin{bmatrix} a\left(\frac{s}{T_d s+1}\right)^2 - c & b\left(\frac{s}{T_d s+1}\right)^2 - d \\ e\left(\frac{s}{T_d s+1}\right)^2 & f\left(\frac{s}{T_d s+1}\right)^2 - d \end{bmatrix} \begin{bmatrix} \varphi_1 \\ \varphi_2 \end{bmatrix} \tag{5.12}$$

As the system is run with a sample time of 0.05 seconds, T_d is set to 0.1 seconds in order to satisfy the Nyquist criteria.

Figure 5.4 shows the control structure of the integrated voluntary control approach with the matrix G^{-1} from equation 5.12 as the angle controller.

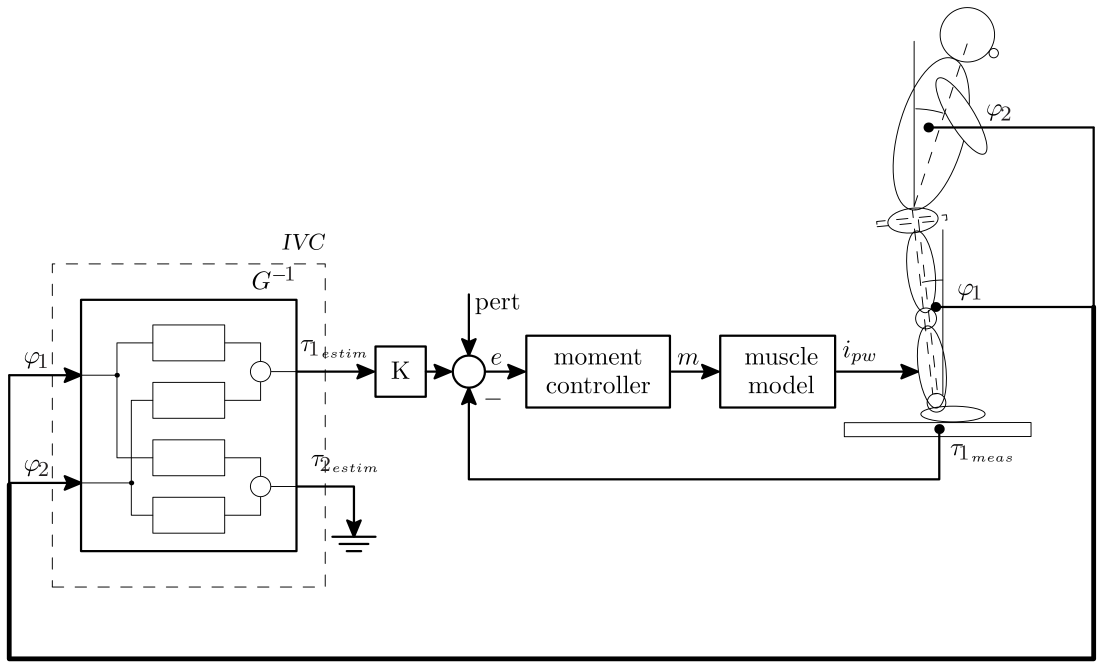


Figure 5.4.: Control structure with integrated voluntary control with the measured inclination angles ($\varphi_{1,2}$) and the torque measured by the force plates ($\tau_{1_{meas}}$). Scaling factor K can be 1, 0.7, or 0.4.

The scaling factor K, with which the signal $\tau_{1_{meas}}$ of the IVC controller is scaled, can be 1, 0.7, or 0.4. This scaling factor is used in order to prove the robustness of this approach.

Anthropometric Data

Zatsiorsky [128] defines formulas based on averaged anthropometric measurements performed with healthy people to calculate the centre of mass, lengths, and moments of inertia of different parts of the human body in relation to their dimensions and weight.

The mass of the subject was $m = 63$ kg and the body height $h = 1.68$ m. The length of the lower (l_1) and upper body (l_2) was measured as 0.84 m each. The weight of the upper body (w_2) and lower body (w_1) were calculated with following formulae:

$$w_1 = 2 \cdot 0.161 \cdot w$$
$$w_2 = 0.678 \cdot w$$

For the calculation of the moment of inertia of the upper and lower body following formulae were used:

$$I_1 = w_1 \cdot (0.56 \cdot l_1)^2$$
$$I_2 = w_2 \cdot (0.798 \cdot l_2)^2$$

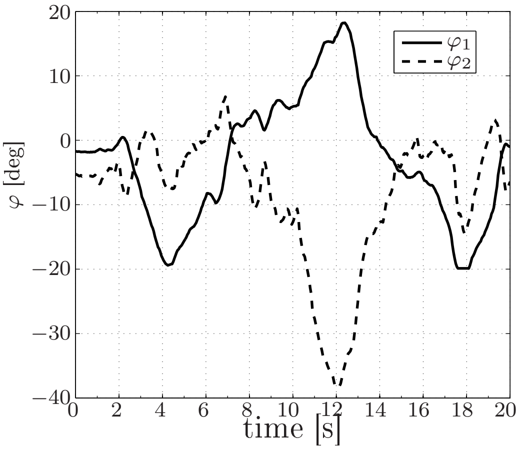
resulting in the anthropometric values summarised in table 5.1

	moment of inertia [kg$\cdot$ m^2]	segment length [m]	segment mass [kg]	centre of mass [m]
lower body	4.3	0.84	20	0.38
upper body	19.32	0.84	43	0.53

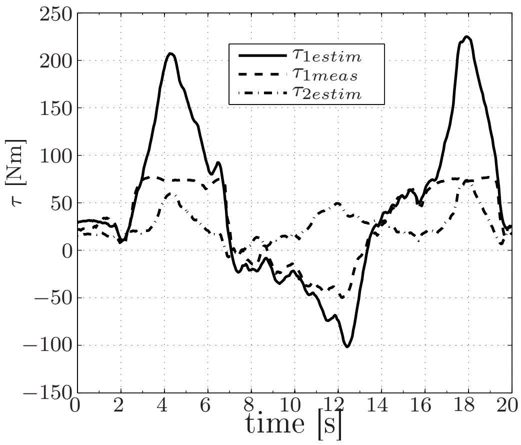
Table 5.1.: Anthropometric data according to calculation tables by Zatsiorsky [128].

5.3.4. Adjustment of Anthropometric Data

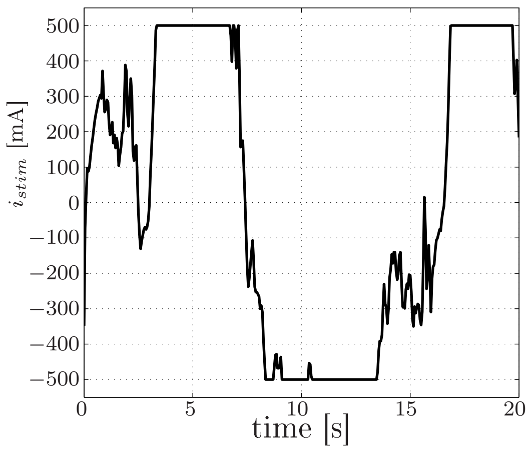
The time plots shown in figure 5.5 are the results of initial tests carried out using the control structure shown in figure 5.4. For the control the anthropometric data of a paraplegic subject as set out in table 5.1 is used. For this scenario a PRBS as the perturbation signal with an amplitude of ± 15 Nm is applied using a gain of $K = 1$.



(a) Time plot of angles.



(b) Time plot of estimated torques τ_{1estim}, τ_{2estim} and measured torque τ_{1meas}.



(c) Time plot of stimulation signal.

Figure 5.5.: Time plots of the measured angles, the ankle torque, and the stimulation signal acquired with the model using the anthropometric data from table 5.1.

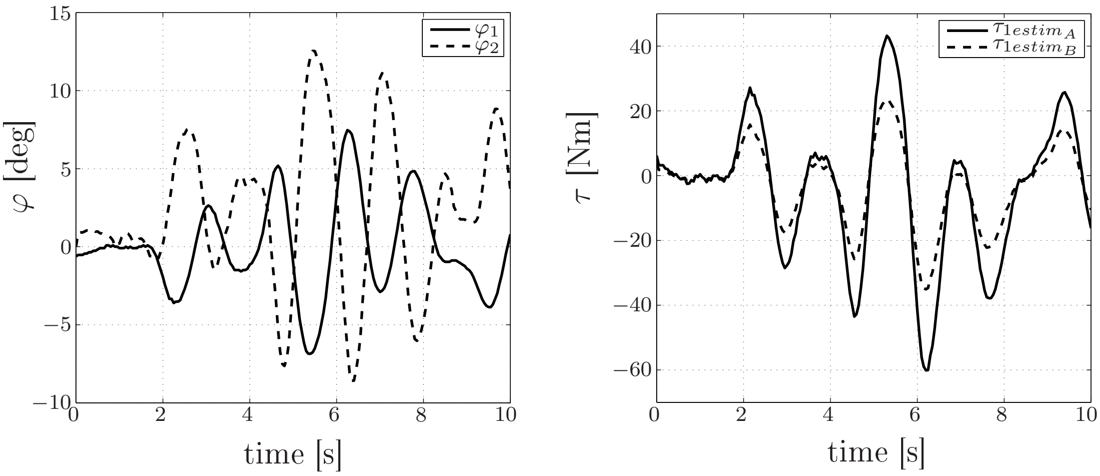
At the time of 4s and 18s peaks in the torque τ_{1estim} are estimated (see figure 5.5(b)) which are much higher than the actual measurements. The inclination angles at these times have the same sign indicating that the upper and lower body lean in the same direction showing the loss of balance of the subject who had to be pulled back to the upright position. The angle controller detects correctly the instability and tries to stabilise the system with a high stimulation current which is applied at the shank muscles. At the time of 12s the subject tries to stabilise himself by leaning to the front but was not able to get back to the upright position and had to be pulled back again to the upright position by the experimenter. Again the controller estimated correctly the instability and commanded a high stimulation current to stabilise the subject. As can be seen in figure 5.5(a), small deviations from the upright position lead very quick to instability (see e.g. the behaviour between $t = 2s$ and $t = 4s$) giving the subject no chance to recover. Within this trial the subject lost his balance three times. Several other trials applying different perturbation levels using the anthropometric data from table 5.1 showed similar results to the ones of figure 5.5. As the subject never managed to keep his balance throughout other trials as well, it can be assumed that the anthropometric values of table 5.1 are not suitable to be used for the control as the torque reference estimated by the IVC (see also figure 5.4) is higher than the subject can achieve (see figure 5.5(b)) in order to allow satisfactory stability.

In order to describe the subject's performance more precisely the anthropometric values were adjusted. For this the values of the upper link were set to the same ones as of the lower link. The resulting anthropometric values are shown in table 5.2:

	moment of inertia [kg· m^2]	segment length [m]	segment mass [kg]	centre of mass [m]
lower body	4.3	0.84	20	0.38
upper body	4.3	0.84	20	0.38

Table 5.2.: Modified anthropometric data for the paraplegic subject.

In order to show the effect of the adjustment to the IVC a simulation was carried out using angle measurements as inputs of a trial where the subject did not lose his balance. Figure 5.6 shows the angle measurements and the resulting torques the IVC from figure 5.4 provides as reference values.



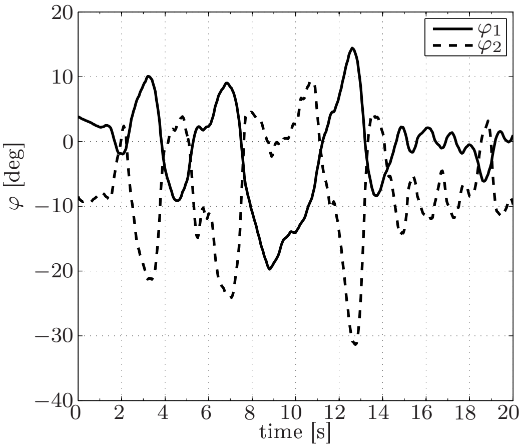
(a) Time plot of measured angle measurements.

(b) Estimated torque τ_{1estim_A} using data from table 5.1 and torque τ_{1estim_B} using data from table 5.2

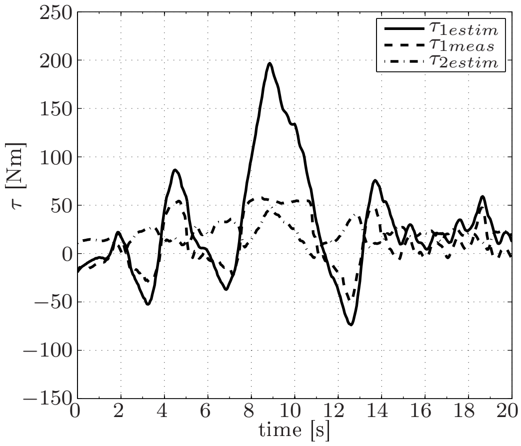
Figure 5.6.: Time plots of measured angle measurements and resulting simulated torques achieved with a model using the anthropometric data from table 5.1 (τ_{1estim_A}) and from table 5.2 (τ_{1estim_B}), respectively.

Figure 5.6 shows e.g. at $t = 4.7s$ an inclination angle of $\varphi_1 = +5°$. The torque reference τ_{1estim_A} achieved with the anthropometric values from table 5.1 is nearly double as high as achieved with the amended anthropometric data from table 5.2. This can be observed for all occasions where the inclination angle φ_1 of the lower body is grater than $\pm 5°$.

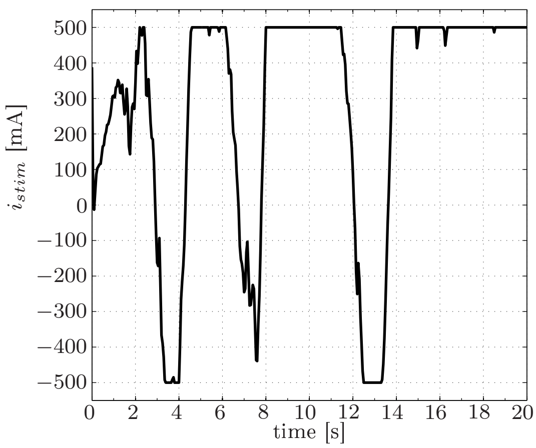
Preliminary tests showed that the amendments of the anthropometric data have a stabilising effect on the performance of the subject during perturbed standing. Figure 5.7 shows the behaviour of the paraplegic subject while applying a PRBS as the perturbation signal with an amplitude of ± 15 Nm and using a gain of $K = 1$ (same conditions as for figure 5.5).



(a) Time plot of angles.



(b) Time plot of τ_1.



(c) Time plot of stimulation signal.

Figure 5.7.: Time plots of the measured angles, the ankle torque, and the stimulation signal acquired with the model using the anthropometric data from table 5.2.

Due to the change of anthropometric data the subject was better able to react to perturbations. Although between the times 8s and 14s the inclination angles of the upper and lower body were quite high, they were always inclined in opposite directions which means the subject managed to keep his balance throughout the trial. Similar behaviour could be seen during other trials using different perturbation levels. For the further evaluation of the IVC-approach the anthropometric data from table 5.2 are used.

5.3.5. Experimental Protocol

The experiments using the integrated voluntary control approach were conducted just after the experiments described in section 4.7.6 and following. The muscle model necessary for the moment control of the inner loop of the IVC structure (see figure 5.4) was identified as

$$G_{muscle} = \frac{0.006756z}{z^2 - 1.554z + 0.6472} \tag{5.13}$$

and was the same as used before (see identification results in table 4.5).

The duration of an experimental trial was carried out for 20 seconds. Furthermore, the subject was supported by an additional constant supporting stiffness of the MRF (5 Nm/deg) in order to prevent the muscles of the subject to fatigue too quickly. In order to show the robustness of this new control approach the signal $\tau_{1_{estim}}$ was multiplied with the gains $K = 1$, $K = 0.7$ and $K = 0.4$, respectively (see figure 5.4). Two different perturbations in PRBS form were applied. The amplitudes of the signals were ± 10 Nm ("low perturbation level") and ± 15 Nm ("high perturbation level").

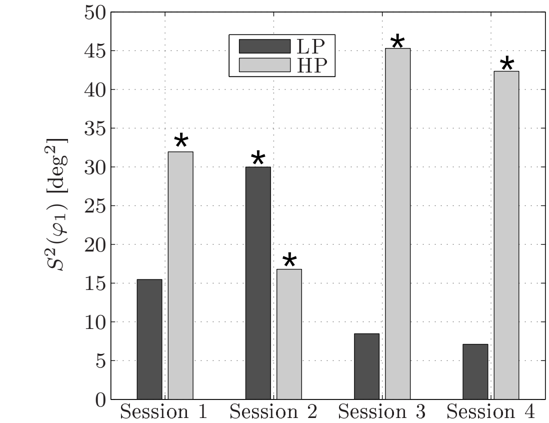
The subject performed six trials during one session. The trials varied in the strength of perturbation and scaling gain K. Due to time constraints the spinal cord injured subject was able to participate in four experimental sessions which were carried out once per week with a two week gap between the second and third session. The subject was asked to continue with the stimulation of the shank muscles at home between experimental sessions in order to maintain muscle strength.

5.4. Results

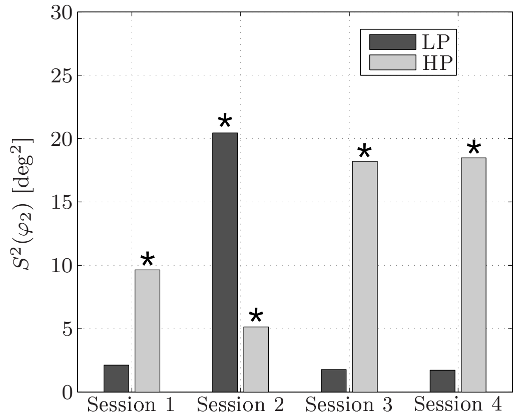
For the evaluation of the results the approach introduced in section 4.3. The evaluated signals are the time signals of the inclination angles of the lower body (φ_1) and upper body (φ_2), the measured torque applied onto the ground ($\tau_{1_{meas}}$), and the applied stimulation pulsewidth (pw). The results of the trials, during which a loss of balance occurred, are marked by a star (see i.e. figure 5.8).

5.4.1. Scenarios with scaling factor 0.4

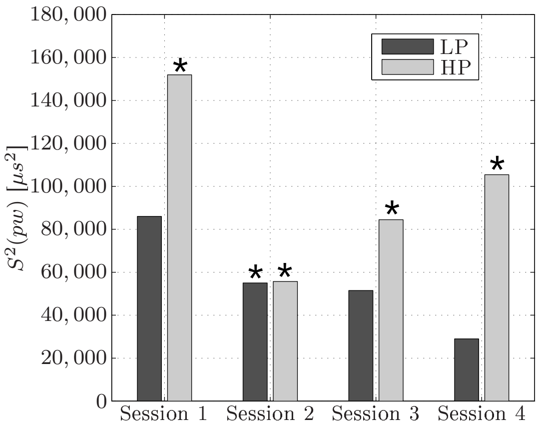
Figure 5.8 shows the results for the variance S^2 using the scaling factor $K = 0.4$ with low ("40%LP" scenario) and high perturbation ("40%HP" scenario) applied.



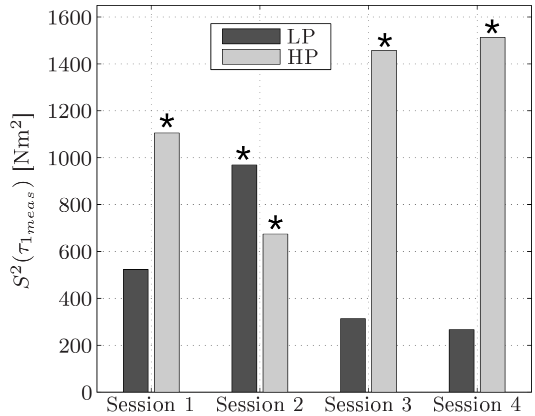
(a) Variance for φ_1.



(b) Variance for φ_2.



(c) Variance for the stimulation signal.



(d) Variance for the measured torque.

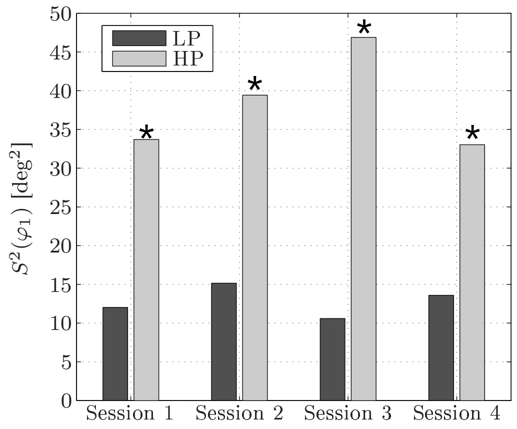
Figure 5.8.: Comparison of the variance for all signals for the scenarios 40%LP and HP achieved with the spinal cord injured subject.

The variance values for the 40%HP scenario, as depicted in figure 5.8, show a comparatively high variance for all measured signals in the first, third, and fourth session and reflect this way the loss of balance. The measurements of the second session, however, do not show a comparatively high evaluation value despite the occurrence of the loss of balance.

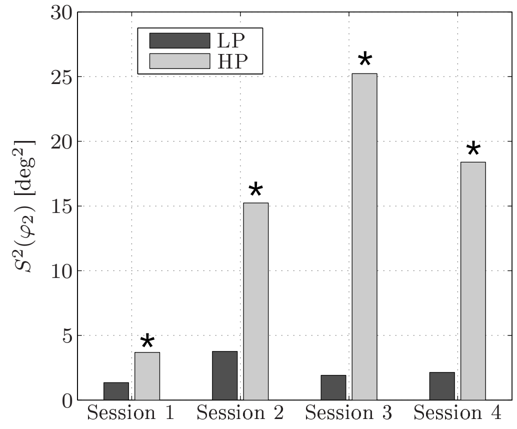
For the 40%LP scenario all measured values show a comparatively high variance when a loss of balance occurs and show small values when this does not happen. The only exception can be seen in $S^2(pw)$ for the second session. There the variance is smaller, despite the loss of balance, compared to the variance in the first session, where no loss of balance occurred.

5.4.2. Scenarios with scaling factor 0.7

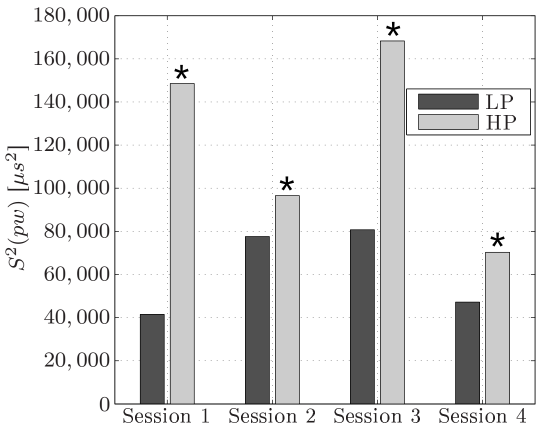
Figure 5.9 shows the results for the variance S^2 using the scaling factor $K = 0.7$ with low ("70%LP" scenario) and high perturbation ("70%HP" scenario) applied.



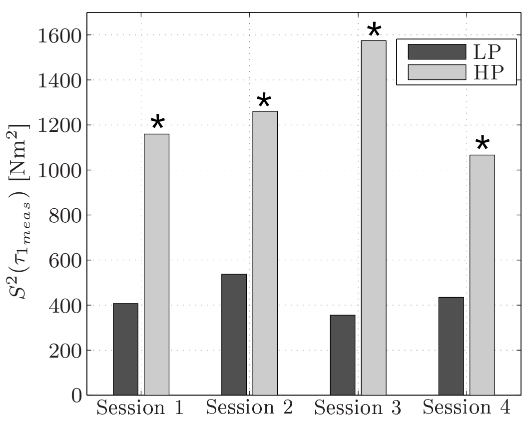
(a) Variances of φ_1.



(b) Variances of φ_2.



(c) Variances of the stimulation signal.



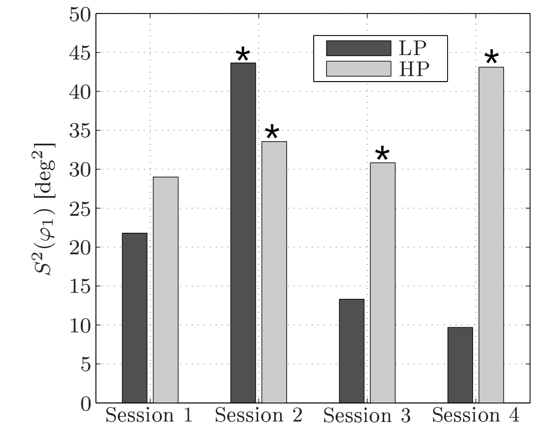
(d) Variances of the measured torque.

Figure 5.9.: Comparison of variances of all signals for the scenarios 70%LP and HP achieved with the spinal cord injured subject.

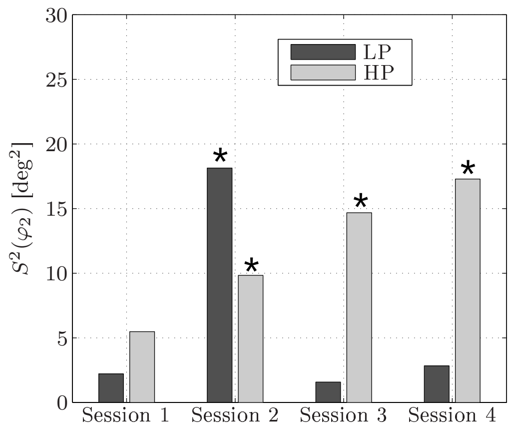
In figure 5.9 the variances $S^2(\varphi_1)$ and $S^2(\tau_{1_{meas}})$ show in all sessions for the 70%HP scenario a higher variance compared to the results achieved during scenario 70%LP. $S^2(\varphi_2)$ shows only during sessions 2-4 high values for the performance during the 70%HP scenario while $S^2(pw)$ shows high values during the first three sessions.

5.4.3. Scenarios with scaling factor 1.0

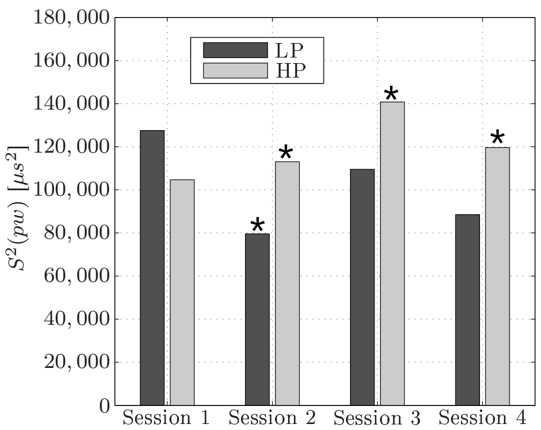
Figure 5.10 shows the results for the variance S^2 using the scaling factor $K = 1.0$ with low ("100%LP" scenario) and high perturbation ("100%HP" scenario) applied.



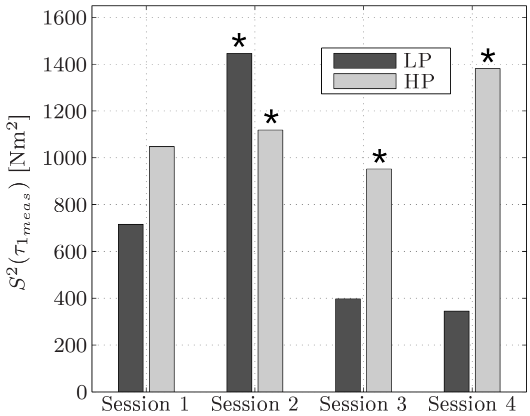
(a) Variances of φ_1.



(b) Variances of φ_2.



(c) Variances of the stimulation signal.



(d) Variances of the measured torque.

Figure 5.10.: Comparison of variances of all signals for the 100%LP and 100%HP scenarios achieved with the spinal cord injured subject.

During the second session when a low perturbation was applied (100%LP scenario) a loss of balance occurred and so $S^2(\varphi_1)$, $S^2(\varphi_2)$, and $S^2(\tau_{1_{meas}})$ show a high value compared to the variances achieved during the other sessions for this same scenario. Only the variance in the stimulation signal shows a smaller value during the second session if compared to the results of the other sessions.

The variances $S^2(\varphi_1)$, $S^2(\varphi_2)$, and $S^2(\tau_{1_{meas}})$ achieved during a loss of balance when the high perturbation was applied (100%HP scenario) show only during session 4 a high value if compared to the results of session 1 where no loss of balance occurred while the variance $S^2(pw)$ shows the highest value during session 3. The values $S^2(\varphi_1)$ and $S^2(\tau_{1_{meas}})$ of sessions 2 and 3, where a loss of balance happened, show nearly the same or even smaller values if compared to the results of session 1 where no loss of balance occurred.

5.4.4. Comparison of results with different controllers

The following figures show the comparison of the results achieved with the integrated voluntary control approach and with the stiffness and stiffness-viscosity controllers as presented in chapter 4.7.

Comparison of Results for $S^2(\varphi_1)$

Figure 5.11 shows the variances for φ_1 resulting from the experiments with the stiffness-stiffness-viscosity controller and the integrated voluntary control approach.

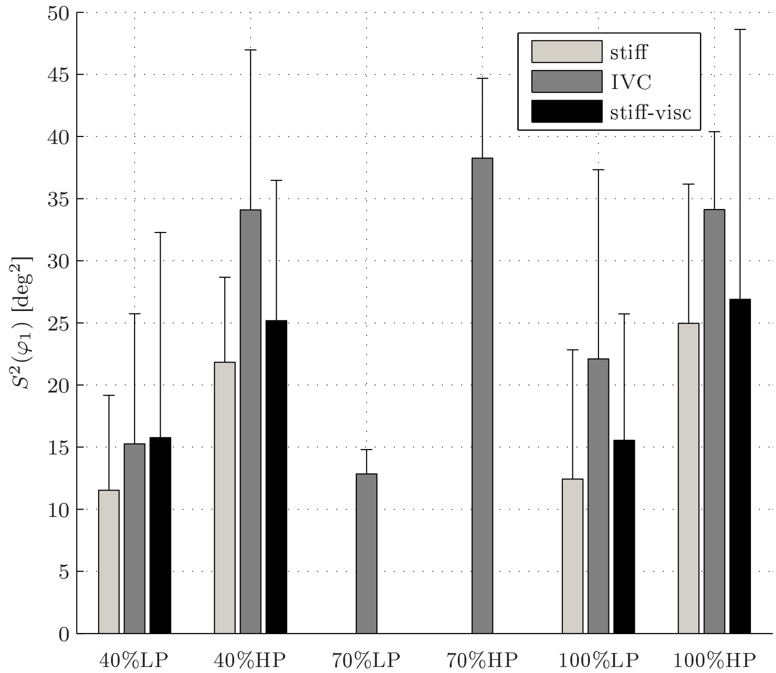


Figure 5.11.: Evaluation values for the inclination angle of the lower body (φ_1) resulting from the experiments with the stiffness, the stiffness-viscosity, and the IVC controller. Maximal measured values are indicated by the whiskers.

40%LP scenario: The use of a stiffness controller results in smaller variances while the performance with a stiffness-viscosity controller shows nearly the same variance as with the IVC result for this scenario. The measurements achieved with the stiffness-viscosity controller show, however, a higher standard deviation than it is seen using the IVC controller.

40%HP scenario: The results for this scenario indicate clearly that with the IVC controller the subject moved more with the lower body than during the trials using the stiffness and stiffness-viscosity controller.

70%LP scenario: Only the IVC approach was used during this scenario. The results have basically the same results as achieved with the stiffness controller during the 40%LP and 100%LP scenarios.

70%HP scenario: The variance achieved with the IVC approach during this scenario is higher than the ones achieved during the other scenarios.

100%LP scenario: During this scenario the subject moved more using the IVC controller than with the stiffness and stiffness-viscosity control approach.

100%HP scenario: The subject moved during this scenario less with the stiffness and stiffness-viscosity controller than with the IVC controller. However, the standard deviation of the results achieved with the stiffness-viscosity controller are higher than the ones seen in the results measured with the IVC approach.

Comparison of Results for $S^2(\varphi_2)$

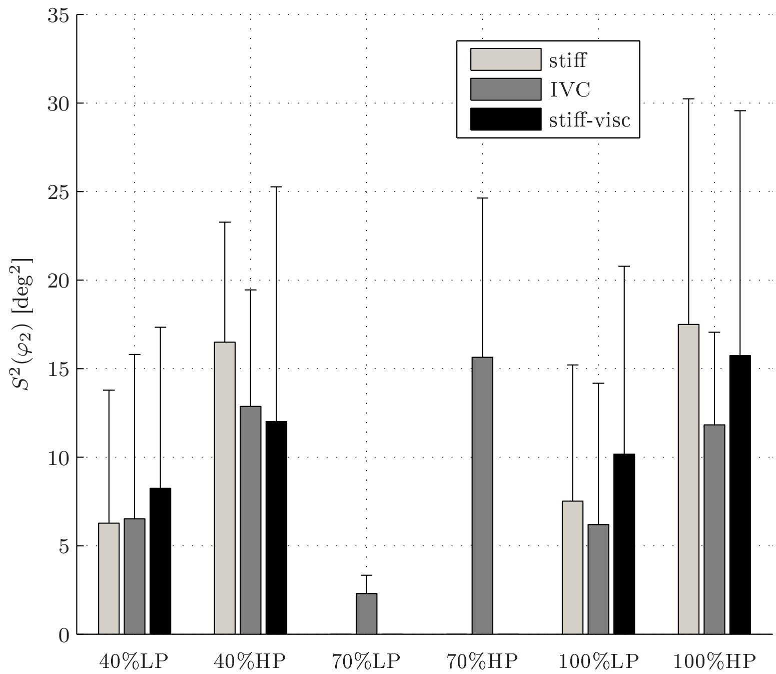


Figure 5.12.: Evaluation values for the inclination angle of the upper body (φ_2) resulting from the experiments with the stiffness, the stiffness-viscosity, and the IVC controller. Maximal measured values are indicated by the whiskers.

40%LP scenario: Results show basically the same variance with the stiffness and IVC controller whereas the stiffness-viscosity controller caused the subject to be slightly less stable.

40%HP scenario: Variances with the stiffness-viscosity and the IVC controller are nearly the same whereas the results with the stiffness controller are slightly higher.

70%LP scenario: The variance shows the smallest value if compared with all other scenarios

70%HP scenario: Shows basically the same variance value as the one achieved during the 40%HP scenario with the stiffness controller and during the 100%HP scenario achieved with the stiffness-viscosity controller.

100%LP scenario: The performance with the IVC controller and with the stiffness controller are nearly equally stable whereas the performance with the stiffness-viscosity controllers is less stable.

100%HP scenario: Results with the stiffness and stiffness-viscosity controller show higher variances than achieved with the IVC controller.

Comparison of Results for $S^2(pw)$

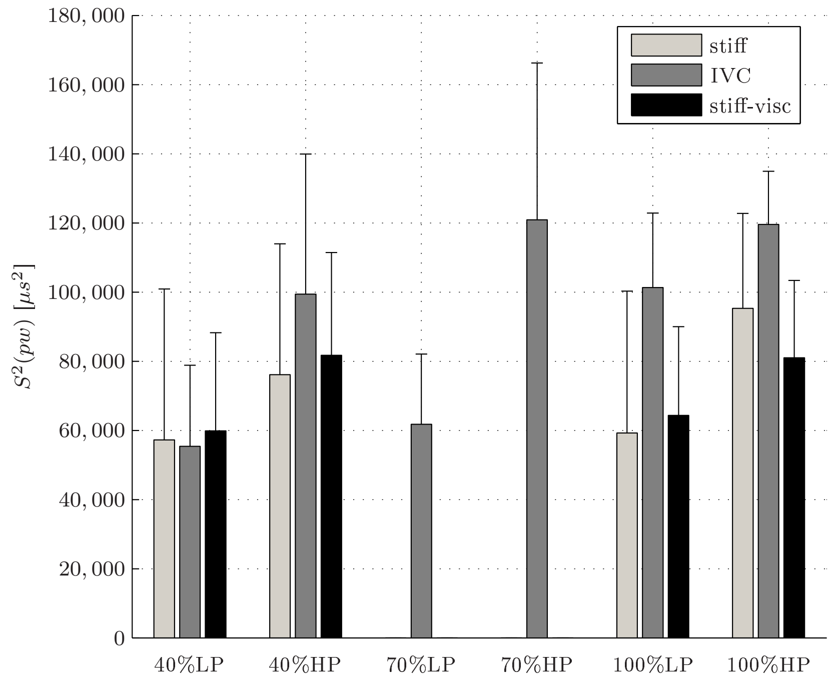


Figure 5.13.: Variances for the the stimulation signal resulting from the experiments with the stiffness, the stiffness-viscosity, and the IVC controller. Maximal measured values are indicated by the whiskers.

40%LP scenario: The variances show that the amount of stimulation was approximately the same during this scenario for all three different control approaches.

70%LP scenario: Shows approximately the same value as achieved during the 40%LP scenario with all the different controllers.

70%HP scenario: The variance has basically the same value as achieved during the 100%HP with the IVC approach.

40%HP, 100%LP and 100%HP scenarios: During these scenarios the stimulation was always higher using the IVC controller than with the stiffness and stiffness-viscosity controller.

Comparison of Results for $S^2(\tau_{1_{meas}})$

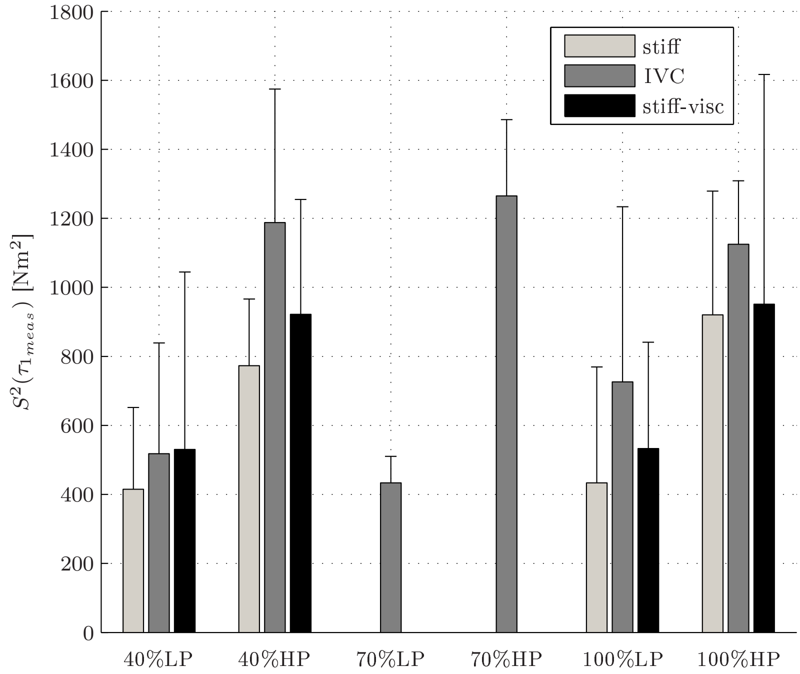


Figure 5.14.: Evaluation values for the torque at the ankle ($\tau_{1_{meas}}$) resulting from the experiments with the stiffness, the stiffness-viscosity, and the IVC controller. Maximal measured values are indicated by the whiskers.

40%LP scenario: The variance of the torque signal measured with the stiffness-viscosity and the IVC controller are basically the same whereas the use of the stiffness controller caused a smaller torque output.

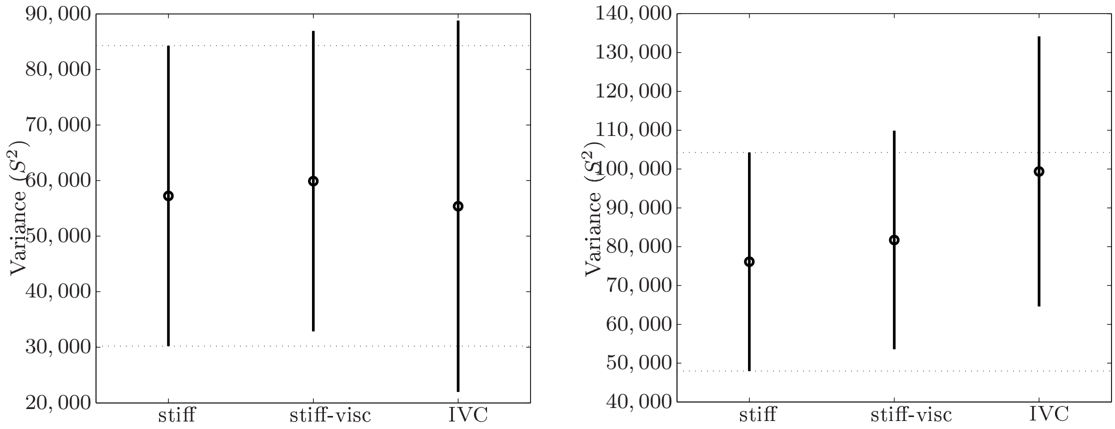
70%LP scenario: The variance shows approximately the same value as achieved during the experiments with the stiffness controller in the 40%LP scenario.

70%HP scenario: The variance shows a slightly higher value if compared with the values achieved with the IVC approach in the 40%HP and 100%HP scenarios.

40%HP, 100%LP and 100%HP scenarios: The torques achieved with the IVC controller during these scenarios are higher than with the stiffness and stiffness-viscosity controller.

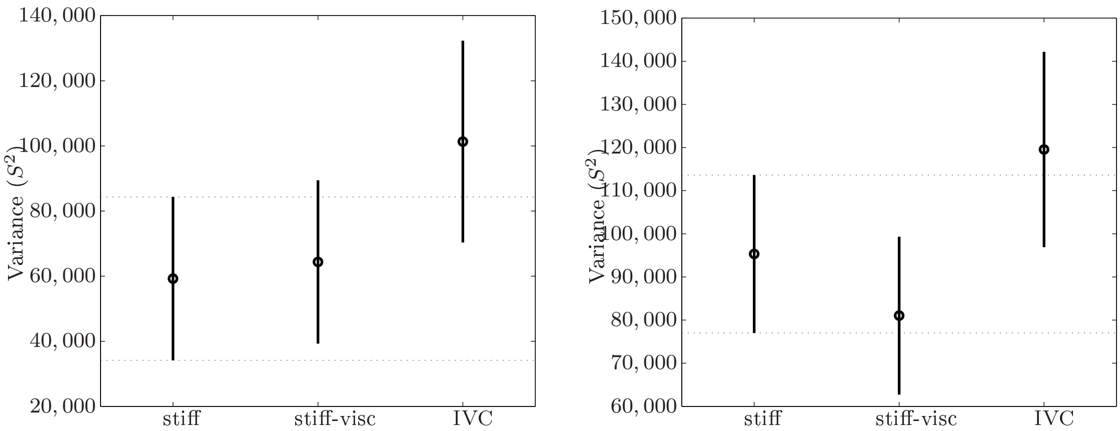
5.4.5. Statistical Analysis

For the statistical analysis of the measured results a one-way analysis of variance was computed by comparing the means of the variances of each measured signal resulting from the use of the three different controllers (stiffness, stiffness-viscosity, and IVC-controller).



(a) Statistical results for the 40%LP scenario; $p = 0.9782$.

(b) Statistical results for the 40%HP scenario; $p = 0.5998$.



(c) Statistical results for the 100%LP scenario; $p = 0.1330$.

(d) Statistical results for the 100%HP scenario; $p = 0.06838$.

Figure 5.15.: Statistical comparison of the mean variances achieved with the three different controllers (stiffness, stiffness-viscosity controller, and IVC controller) for the stimulation signal.

This analysis was performed in order to test the hypothesis that the compared mean values are all the same, against the general alternative that they are not all the same. To concretise which pairs of means are significantly different, additionally a multiple comparison procedure was performed using the Dunn-Šidák procedure [129]. For the comparison only the results of the 40%LP/HP and 100%LP/HP scenarios were used. The significance value was set to $p = 0.05$.

The results of the statistical analysis regarding all scenarios and the respective evaluated signals were very similar. Therefore, representative for the other values only the results using the measurements of the stimulation signal will be presented. The displayed graphs in figure 5.15 show the mean value, indicated by a circle, and a comparison interval around the circle. Two means are significantly different if their comparison intervals are disjoint, and are not significantly different if their intervals overlap.

The subplots of figure 5.15 show the 40%LP/HP and 100%LP/HP scenarios which were performed with the three controllers used. As all the comparison intervals overlap for all scenarios the mean variances of the stimulation signal are not significantly different.

5.5. Discussion and Conclusions

In this chapter a new control approach controlling the posture of a paraplegic subject during standing was introduced. For this approach a combined effort of the voluntary movement of the subject's upper body together with the controlled functional electrical stimulation of the paralysed shank muscles was used while the subject was standing in the MRF. The aim of this concept was to reduce the movement of the lower body and let the upper body do the main work in the balancing task. With this approach the amount of stimulation might be reduced and consequently cause the stimulated muscles to fatigue less.

In order to being able to combine the natural with artificial control, the contribution of the upper body movement to the overall torque applied onto the ground had to be estimated. This estimation was achieved by using a double-inverted pendulum model describing the mechanical properties of the subject standing in the MRF. As control structure, a nested control loop was used. The estimation of the subject's contribution due to voluntary movement was placed in the outer loop whereas the artificial controller controlling the stimulation signal was put in the inner loop using the estimated torque as reference value.

The results achieved with this new approach were compared with the results measured using already known controllers taking the voluntary movement of the subject not into account. For the comparison the evaluation value defined in section 4.3 was used.

The results show that the evaluation values, in general, indicated as expected a loss of balance with a high value. The reason why the evaluation value showed in some cases of a loss of balance a similar value as for sessions where the subject did not lose his balance (e.g. figure 5.8(a)), lies therein that the subject got surprised by the perturbation and therefore could not react in time, which resulted in a loss of balance fairly quickly. Furthermore, in some cases the subject did not lose his balance but swayed with high inclination angles around the upright position which resulted in a high evaluation value. In order to get a more precise evaluation of the performance during standing the measurement of velocities and accelerations of the upper and lower body might give more detailed insight what happened during a trial.

The comparison of the evaluation values achieved with the different control concepts (see figures 5.11–5.14) show that the performance achieved with the IVC approach did not differ very much from the performances achieved with the stiffness and stiffness-viscosity controllers. Although differences for the respective mean variances were observed, the statistical analysis showed that these differences are not significant (see e.g. figure 5.15).

For future work a more extensive study with several spinal cord injured subjects should be carried out to see what impact this IVC approach has on the balancing task and on the amount of stimulation used.

Furthermore, experiments could be carried out where the hydraulic system supplies the dynamic moment of the lower body in the same way as the subject did via stimulation of the gastrocnemius muscles, and let then the stimulated muscles contribute the necessary static moments.

Additionally, tests could be carried out to see whether a model of the double inverted pendulum with the inertia $J = 0$ would be sufficient for allowing standing with reduced stimulation as the inertia of the lower and upper body of a SCI-subject is difficult to predict.

6. Sensory electrical nerve stimulation for training dynamic balance responses in a chronic stroke patient

6.1. Summary

Similar to the approaches presented in chapters 4 and 5 the combination of voluntary control of a subject during perturbed standing and electrical stimulation was adapted for a training regime carried out with a stroke patient.

A dynamic standing frame was modified with electrical actuators which allow the application of unexpected perturbations to neurologically impaired people during standing, while protecting the subject from falling. The subject underwent two different periods of perturbation training, each lasting ten days. During the first period the subject was only perturbed in eight different directions. During the second period the subject was also perturbed, but was assisted by sensory electrical stimulation of the soleus (SOL), tibialis anterior (TA), tensor fascia latae (TFL), and vastus muscles (VAS) in the impaired leg. After each period of training, an assessment was carried out to measure the forces the subject applied on the ground via two force plates and the EMG responses of the SOL, TA, TFL, and VAS muscles. The subject improved his ability to balance throughout the training, with the largest improvements occurring during the final period when sensory electrical stimulation was used. These observations suggest to carry out a testing series with more subjects to evaluate the value of this rehabilitation method in a clinical environment.

The content of this chapter has been published with the Journal of Medical and Biological Engineering (JMBE) [1].

6.2. Background

In Scotland with a population of 5 Mio people, annually approximately 15,000 people suffer strokes for the first time with approximately 80% surviving beyond 30 days. Of all surviving stroke patients who start with a rehabilitation programme, around 50% will remain impaired on their affected side [44].

For the rehabilitation of stroke patients, a therapist can usually work with only one patient at a time and therefore the rehabilitation is very labour intensive. Additionally, the physical effort required by the therapist can be very high in assisting the patient during rehabilitation [125]. Therefore, assistive devices were developed in order to reduce the physical effort of the therapist as well as the need for human attendants [130]. Examples include the MIT-Manus [122–124] which assists the rehabilitation of elbow and shoulder movement in stroke patients, the gait trainer which allows chronic stroke and paraplegic patients to train gait-like movement [125, 126], and the Lokomat [127], a robotic orthosis supporting spinal cord injured and chronic stroke patients during treadmill training rehabilitation. Devices specific for balance retraining include balance platforms (such as the BalanceMaster and the Biodex Balance System) which are based on a moving standing platform combined with biofeedback, and the BalanceTrainer [56] which is a dynamic standing frame allowing balance training and step-like movements. Initial results with these devices showed an improvement in rehabilitation outcome [55, 123, 131–133].

According to the findings of Field-Fote [134], the spinal and cortical neural circuitry are modified by applied electrical stimulation as the neural circuitry underlying motor performance on a short- and long-term basis is modulated. Studies which combined robotic rehabilitation approaches with functional electrical stimulation (FES) also showed an improvement in rehabilitation outcome [135, 136]. However, Tong *et al.* [137] stated that there was no significant difference in performance achieved after using a combination of rehabilitation robot and FES compared to the performance achieved after using a rehabilitation robot only.

Other studies have shown that stroke patients can regain independence in activities of daily life using **T**ranscutaneous **E**lectrical **N**erve **S**timulation (TENS) [138–142]. This type of stimulation uses only a small electrical current applied to the skin which can usually be felt and will, at normal strength, only stimulate sensory nerves [143] and not motor nerves as it happens with FES.

Based on the reported potential benefits of employing assistive devices, the feasibility of combining such therapy with sensory transcutaneous electrical nerve stimulation in stroke rehabilitation is investigated, using a modified BalanceTrainer. Based on an alternating training protocol it is investigated whether this combined approach would have the potential to lead to a significant change in performance compared to rehabilitation using the modified BalanceTrainer alone. In this chapter a control method and apparatus for applying sensory electrical stimulation during perturbed stance in a modified BalanceTrainer is presented. The outcome of a case study is shown where the change in balance performance in a chronic stroke patient during perturbed standing is investigated while applying transcutaneous electrical nerve stimulation. Force plate measurements and EMG data were used to evaluate the balance performance at assessment points throughout the training. The results are discussed and the feasibility of this approach evaluated.

6.3. Methods

6.3.1. The Standing Frame

The standing frame which was used for the experiments is an adapted version of the commercially available BalanceTrainer and is described in chapter 3.3.

6.3.2. Subject

The experiments were performed with one chronic stroke patient (male, 45 years old, with a height of 1.85 m and a weight of 85 kg which remained unchanged throughout participation in the study). He was 19 months post stroke, had completed the stroke rehabilitation programme and no longer received physiotherapy treatment at the time of the study. The impairment affected his right side. The subject needed no support during quiet standing, but was using an orthosis to prevent foot-drop during gait, due to inactive tibialis anterior muscle on the affected side. The orthosis was removed for the training and assessment sessions. All experimental procedures were approved by the Slovenian National Ethics Committee and the subject provided written, informed consent prior to participation.

6.3.3. Measurements

In order to assess changes in ground reaction forces the subject stands on two force plates (AMTI, Massachusetts, USA).

The force distribution between the two legs, as well as changes in the centre of pressure (CoP) were assessed. The CoP components in x- and y-direction were calculated as,

$$CoP_x = -M_y/F_z$$
$$CoP_y = M_x/F_z$$

with M_x and M_y denoting the moments in x- and y-direction and F_z being the vertical force. The sample time of the force measurements is 1 kHz. Before every session the force plates were reset.

For the acquisition of the EMG data, repositionable surface electrodes (3M^{TM}Red Dot^{TM}Monitoring Electrodes with Foam Tape, 3M^{TM}, USA) were used. The signals were amplified (MyoSystem 2000 Amplifier, Noraxon Inc., USA), and recorded with a sample rate of 1kHz, after appropriate anti-aliasing filtering. The raw EMG signals were inspected to ensure that the electric fields from the motors or other environmental disturbances did not interfere with the recordings. EMG data were rectified and the linear envelop extracted by applying a 4th-order low pass Butterworth filter with a cut-off frequency of 7Hz [144].

6.3.4. Sensory Electrical Stimulation

Sensory electrical stimulation was applied to the skin areas over the soleus (SOL), tibialis anterior (TA), tensor fascia latae (TFL), and vastus (VAS) muscle groups in the impaired leg as the subject was perturbed. These stimulations sites were selected as the corresponding muscle groups are important for ankle stabilisation (SOL, TA), knee extension (VAS) and medial-lateral movement (TFL), and therefore affect balance control [57]. Depending on the direction of perturbation, the muscles which are mainly involved in the recovery of the perturbation were stimulated (see table 6.1). As the subject's right side is impaired the stimulation was applied only for perturbations in the sagittal plane (front, back) and towards the right (right, front/right, back/right).

direction	TFL	VAS	TA	SOL
front				✓
back		✓	✓	
right	✓			
front/right	✓			✓
back/right	✓	✓	✓	

Table 6.1.: Directions of perturbation and stimulated muscle groups for impairment on the right side. The ticks indicate which muscle groups were stimulated. Note that stimulation was only applied to the affected right leg.

The stimulation was current controlled, monophasic, and charge balanced using the Stanmore Stimulator [110] and delivered via self-adhesive surface electrodes (PALS, 50mm round, Axelgaard Mfg. Co., Ltd., Denmark). The aim was to stimulate during the time when the subject was trying to return to the starting position after he had been perturbed. The intensity of stimulation was regulated by the current level of the stimulation pulses.

The start of stimulation was triggered by a signal which initiates the perturbation of the frame. The timing of the stimulation as well as the triggering of the perturbation were controlled by PCs running Matlab/Simulink. A preliminary test with an able-bodied person was carried out to determine an appropriate pulse width and the duration of the stimulation of the different muscle groups. The aim was to determine the precise on-set for the stimulation and to make sure that the stimulation was active only during the time the subject was reacting to the perturbation. Note that these stimulation times were not obtained from measurements of muscle activity, but are based on observations of the recovery pattern following perturbation. The same starting and finishing times of the stimulation were used for all directions of perturbations and are summarised in table 6.2.

muscle group	start [s]	finish [s]
TFL	0.25	1.5
VAS	0.5	1.5
TA	0.25	1
SOL	0.25	1

Table 6.2.: Start and finishing time of stimulation for each muscle group after the initiation of perturbation.

A stimulation frequency of 20Hz and a constant pulse width of 250μs were chosen, while the stimulation currents were adjusted individually at the start of each session to compensate for variations in the placement of the electrodes. The current levels were selected for each muscle group separately in such a way that the subject had to feel the stimulation clearly without having the stimulated muscles contracting due to the stimulation. For TFL, VAS and SOL, a current range of 20-40mA was used, while for TA the current level was 40-50mA.

6.3.5. Experimental Protocol

The experimental protocol is summarised in figure 6.1.

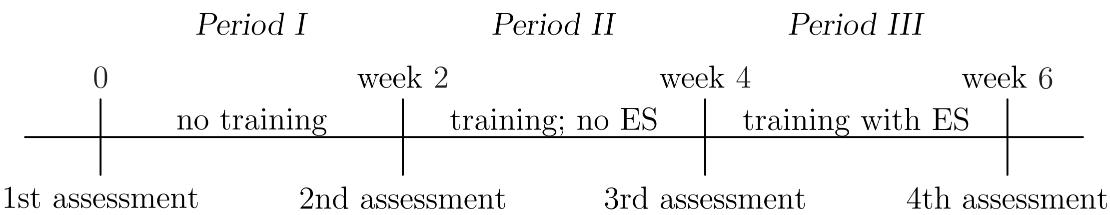


Figure 6.1.: The time scale of training using electrical stimulation (ES) in the last training period.

At the beginning of period I, a baseline assessment of the subject's balancing performance was carried out (1st assessment). After two weeks without training (period I) the performance of the subject was reassessed (2nd assessment). A two-week session with training in the BalanceTrainer (period II) followed. After a 3rd assessment the subject underwent a final period (period III) of training which was identical to training period II, except that this time sensory electrical stimulation was applied. At the end of this training period the performance was assessed again (4th assessment).

During periods II and III the subject trained five days a week. While training, the subject stood in the standing frame with close contact at the pelvis and the feet in a normal parallel quiet standing position. He was perturbed in eight different directions (see figure 3.8) and was asked to react to the perturbations in the way he thought most appropriate without moving his feet. To recover from the perturbation, a combination of ankle and hip strategy was typically required. A round of perturbations was completed when the subject had been perturbed once in all eight directions. The order of perturbation direction changed randomly from round to round. The time between perturbations also varied randomly, but was chosen large enough to allow the subject to return to the initial upright position before the next perturbation was applied. It took the subject less than five seconds to react to the perturbation and to return to the initial position. At each training session, the subject performed 16 rounds, resulting in a total duration of approximately 20 minutes per session. During period III, sensory electrical stimulation was applied using the procedure outlined in section 6.3.4. For the assessments the subject carried out the same exercises as during normal training days, but surface EMG data of the SOL, TA, TFL and VAS muscle groups in the impaired leg as well as force plate measurements were recorded. No stimulation was applied during the assessment sessions.

6.3.6. Data analysis

Statistical analysis of force data characteristics

To allow a more detailed statistical analysis of the changes in the force data between assessments, a number of key characteristic values were extracted from the vertical force response, F_z, of the impaired leg to the perturbations during the assessment sessions. These values, which are summarised in figure 6.2, include: the starting value of F_z, the peak value (i.e. the difference between maximum and starting value of F_z), the peak time (i.e. the time at which the peak has been reached), the undershoot value (i.e. the difference between the minimum and the starting value of F_z), the undershoot time (i.e. the time at which the minimum has been reached), and the final value of F_z.

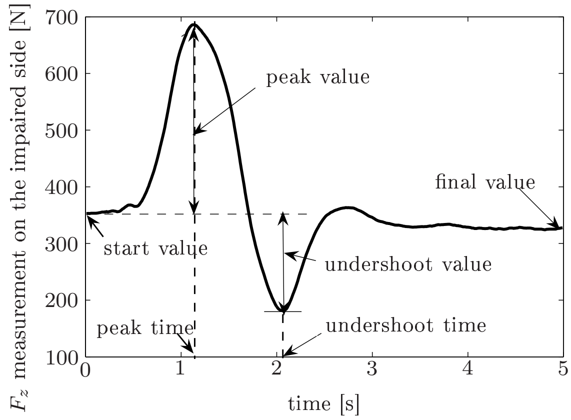


Figure 6.2.: Characteristic values used for the evaluation of the ground force measured under the impaired foot during the perturbation to the right.

The characteristic values, are based on the 16 sets of data for each direction obtained during one assessment. A one-way analysis of variance (ANOVA) was used to analyse these, giving a statistical test of whether the means of the characteristic values obtained at the four assessments are equal. To obtain further details about which pairs of means are significantly different, a multiple comparison algorithm was applied using the Dunn-Łidák procedure [129]. The confidence interval was set to 95% ($p<0.05$). The Matlab Statistics Toolbox (The Mathworks, USA) was used for the statistical analysis.

Analysis of EMG data

To enable the analysis of the relative changes in EMG following a disturbance, the EMG data were normalised and their offsets removed. Since the maximal contraction the subject was able to produce with the impaired limb could not be established directly, the maximal value of the existing EMG measurements over the 4 assessments for each muscle group for normalisation was used. The EMG data were averaged over the 16 rounds of perturbation which comprise each assessment.

6.4. Results

Measurement results reported here were obtained during the four assessment sessions. Corresponding data were averaged over the 16 rounds which constituted one assessment. The results show the voluntary response of the subject to the perturbations since no sensory electrical stimulation was applied in the assessment sessions.

6.4.1. Force Data

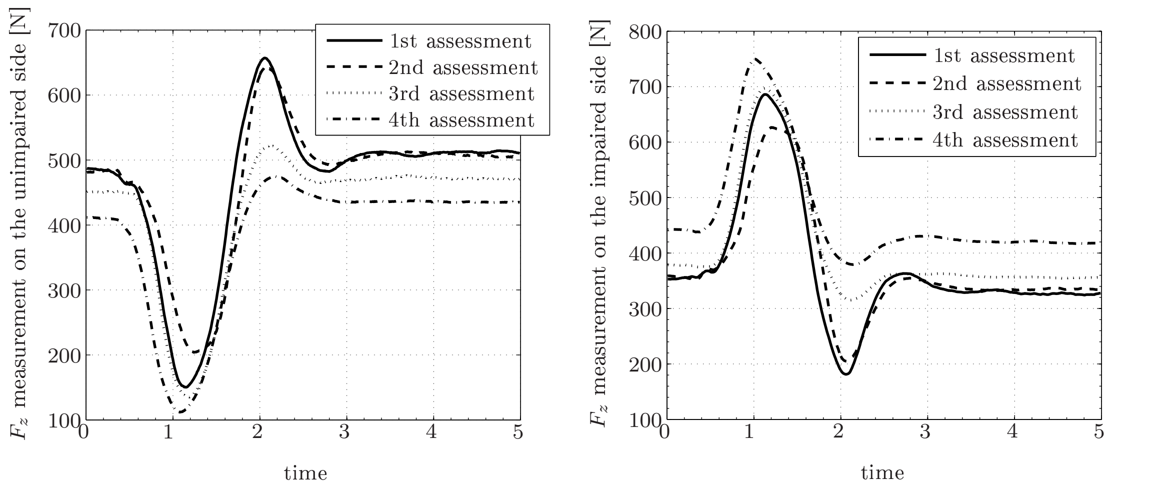
Although the subject was perturbed in eight directions in each assessment (as shown in figure 3.8), changes in the force data were most apparent for perturbations in the direction of the subject's impaired side, i.e. to the right. For this reason the presentation of the force data focuses on the reactions to perturbations to the right. The trajectory of vertical force data is presented, followed by the weight distribution between the two legs and the displacement of the CoP.

6.4.2. Force measurements

Vertical force

Vertical force data of the unimpaired and impaired side for all four assessments are shown in figure 6.3. The performance during the first two assessments (solid and dashed lines) shows a very similar pattern of behaviour, except that the undershoot value in figure 6.3(a) and the respective peak value in figure 6.3(b) reached during the second assessment are smaller than the ones achieved during the initial assessment.

The trajectories of the vertical force data following the initiation of the perturbation at time 0 are shown in figure 6.3 for the unimpaired and impaired side for all four assessments. The corresponding upper limits for the standard deviation values are summarised in table 6.3 (the trajectories of the standard deviations were omitted from figure 6.3 for clarity.). The generic shape of the response is similar for all assessments: On the unimpaired side, the initial period of constant force is followed by a reduction in as the subject is pushed away from this side. As he regains balance, an overshoot in the force on this side can be observed which is followed by a period of relatively constant force as he has recovered from the perturbation. On the impaired side, the initial period of constant force is followed by an increase in as the subject is pushed towards this side. As he regains balance, an undershoot in the force on this side can be observed which is followed by a period of relatively constant force. The results show that it took the subject approximately 3.5 sec to fully recover from the perturbation.



(a) Vertical force F_z on the unimpaired (left) side. (b) Vertical force F_z on the impaired (right) side.

Figure 6.3.: The change in the vertical force F_z on the impaired and unimpaired side after the subject was perturbed to the right; measured during all four assessments. Perturbation was initiated at 0s.

assess-ment	standard deviation for F_z left [N]	standard deviation for F_z right [N]
1	< 77.3	< 77.6
2	< 72.3	< 68.7
3	< 59.1	< 57.7
4	< 43.6	< 45.5

Table 6.3.: Upper limits of standard deviation values of the vertical force F_z on the unimpaired (left) and impaired (right) side for each assessment, as shown in figure 6.3.

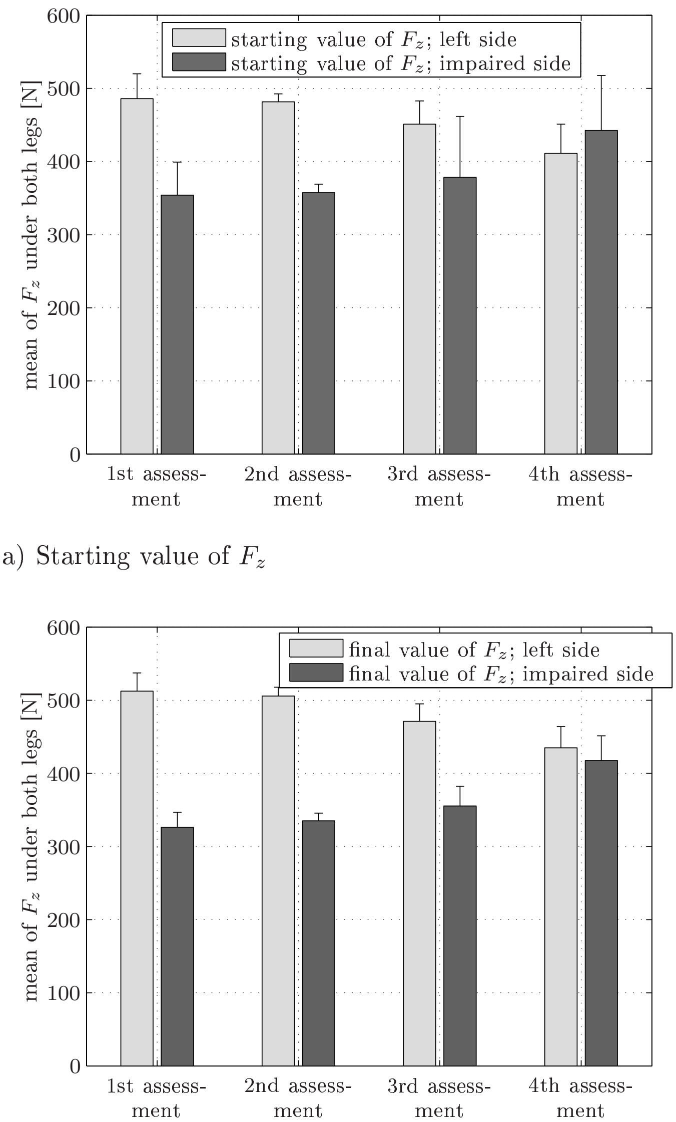
The performance during the 1st and 2nd assessments (solid and dashed lines in figure 6.3) shows a very similar pattern of behaviour.

Following two weeks of training without sensory electrical stimulation (period II) the most obvious change in performance during assessment 3 (dotted lines) can be observed during the recovery from the perturbation: On the unimpaired side (see figure 6.3) the overshoot is reduced, while on the unimpaired side, the corresponding undershoot is smaller. After another two weeks of balance training (assessment 4, dash-dotted lines in figure 6.3), this time with sensory electrical stimulation (period III), a marked increase in starting and final values on the impaired side can be observed when compared to the third assessment while the corresponding values are reduced under the unimpaired leg. In addition, a further reduction in overshoot on the unimpaired side and undershoot on the impaired side can be noted. The peak standard deviations reported in table 3 show that their values decrease throughout the programme, with the largest decrease when the subject is participating in the training programme. This indicates that the balance performance is becoming more consistent throughout participation in the intervention.

Weight distribution between the two legs

The results shown in figure 6.4 give an indication of the weight distribution between the unimpaired and the impaired legs by comparing the vertical forces at the start (figure 6.4(a)) and at the end (figure 6.4(b)) of the perturbation trial. Values were averaged for each assessment and are shown together with the respective standard deviations.

Figures 6.4(a) and 6.4(b) show that before and after the perturbation is applied, the subject puts more weight on his unimpaired (left) side during assessments 1, 2 and 3. Only during the final assessment is the weight distribution more balanced, with a slightly larger force under the impaired leg.



(a) Starting value of F_z

(b) Final value of F_z

Figure 6.4.: Bar plots of the starting, the final and overall mean values of the vertical force F_z with the respective standard deviations (whiskers) regarding the 16 repetitions of the vertical forces measured under both feet during the four assessments. The subject was perturbed to the right.

Centre of Pressure

Figure 6.5 shows the position of the centre of pressure (CoP) obtained from averaged measurements during each of the four assessments. The corresponding upper limits for the standard deviation values are summarised in table 6.4.

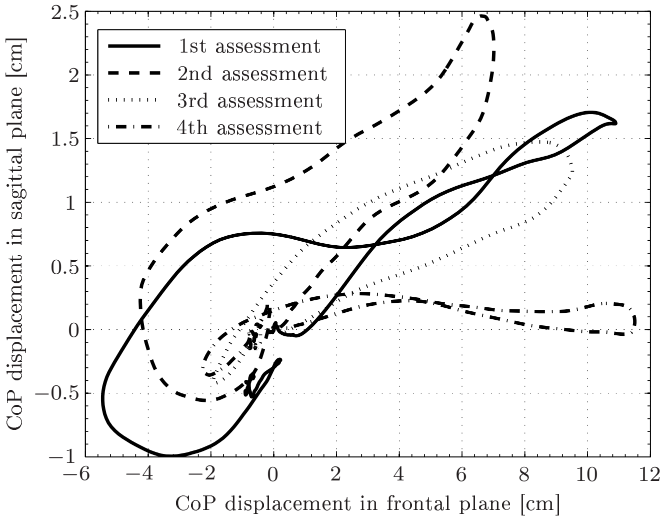


Figure 6.5.: Change in the centre of pressure (CoP) after the subject had been perturbed to the right; measured during all four assessments.

assess-ment	standard deviation CoP_x [cm]	standard deviation CoP_y [cm]
1	< 0.91	< 1.6
2	< 0.86	< 2.1
3	< 0.57	< 1.9
4	< 0.52	< 1.9

Table 6.4.: Upper limits of standard deviation values of the centre of pressure CoP in frontal (CoP_x) and sagittal (CoP_y) plane for every assessment as shown in figure 6.5.

The shape of the CoP distribution during the 1st and 2nd assessments (solid and dashed lines) is similar, with a relatively large forward movement and a significant overshoot in the direction opposite to the perturbation.

After two weeks of training (3rd assessment, dotted line in figure 6.5) the subject was still moving slightly to the front as he is perturbed to the right. During the return to the starting position, however, the movement backwards and to the left is reduced. The final assessment (dash-dotted line in figure 6.5) shows a straight movement to the right with only a small movement to the front and back as the subject reacts to the perturbation.

The results reported in table 6.4 show that the peak standard deviations of the CoP in the frontal plane decrease after assessment 2. This confirms that the balancing performance becomes more consistent throughout the training programme which starts following the 2nd assessment. The standard deviations of the CoP in the sagittal plane remain unaffected throughout the programme.

The characteristic values defined in section 6.3.6 were analysed using the methods described previously. Figure 6.6 shows the means (marked with a circle) of the characteristic values, together with their standard deviations (marked by the whiskers). The groups of measurements which are not significantly different from other groups are presented as thin lines whereas those which are statistically significantly different are marked bold.

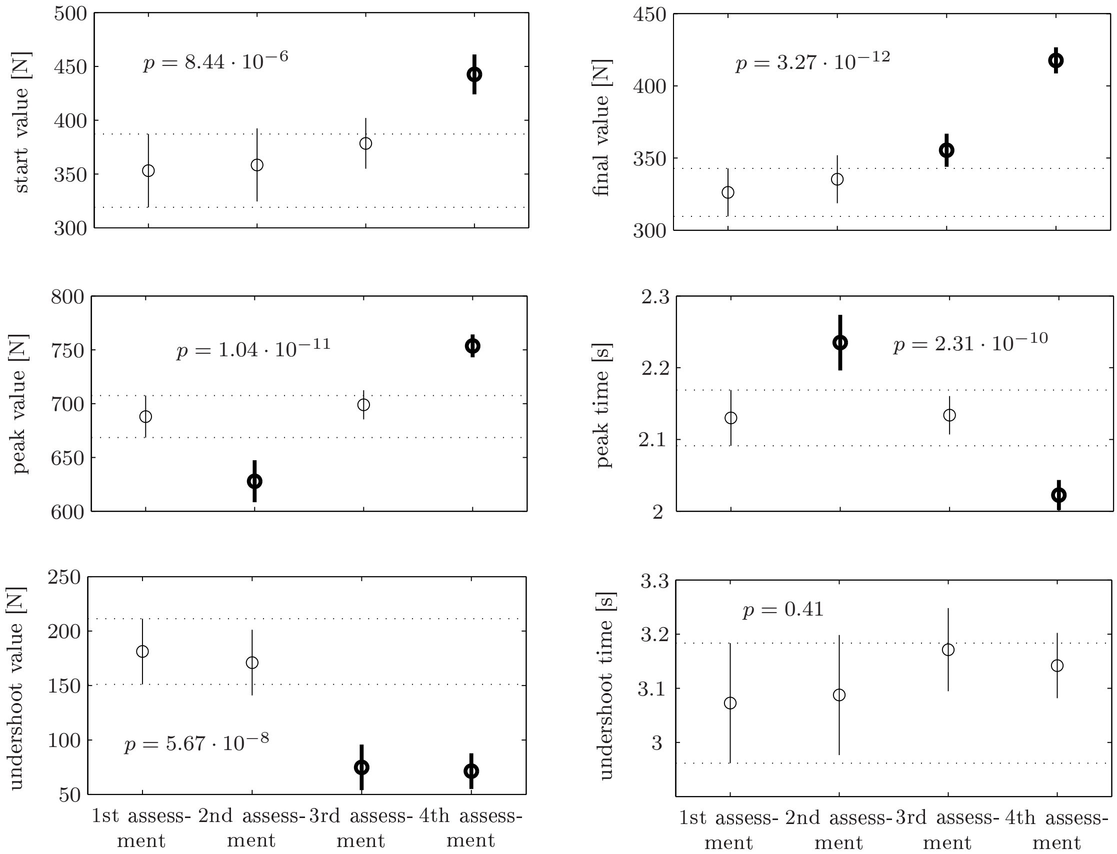


Figure 6.6.: Statistical evaluation of the characteristic values measured on the impaired side. Variables which are statistically significant different are marked bold. Significance indicated when $p < 0.05$

6.4.3. EMG measurements

In order to evaluate to which extent training influences the recruitment of the muscle groups of the impaired leg, EMG data recorded from the right leg during perturbations to the front, back and right were analysed. During perturbations to the front and back it is expected that the shank muscles will be used, while during perturbations to the right mainly the hip muscles will be involved in helping the subject to recover from a perturbation. The knee joint was slightly flexed (ie. not hyper-extended) throughout the experiments. No signs of interference from environmental disturbances or the electric motors could be observed in the recorded EMG signals. Data shown are averaged over the 16 rounds of perturbations which constituted one assessment.

Perturbation to the right

Figure 6.7 shows the EMG data recorded during the four assessments as the subject was perturbed to the right.

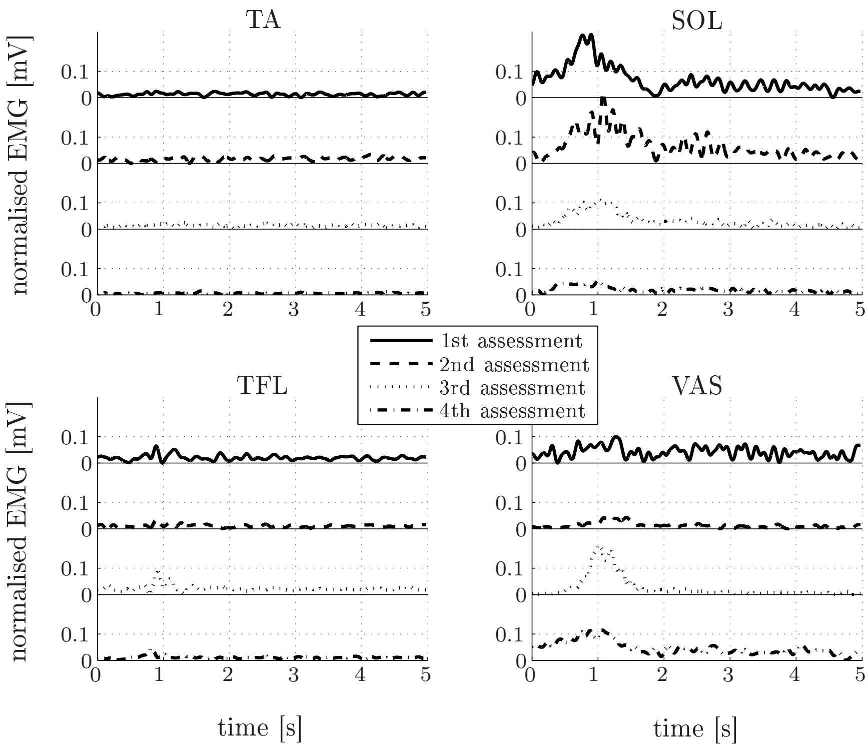


Figure 6.7.: Average normalised measurements of the EMG signal measuring the activities of TA, SOL, TFL, and VAS muscle groups of the right leg during all the assessments. The subject was perturbed to the right.

The main muscle group involved in counterbalancing the perturbation to the right is the TFL muscle group. Figure 6.7 shows a slight activation during the first, third and fourth assessments in this muscle group.

As shown in figure 6.5 the subject moved slightly to the front during the first three assessments as he was perturbed to the right. This can be clearly seen in figure 6.7, as the EMG signals of the soleus muscle group (SOL), which stabilises the movement to the front, indicate activity. As the subject is successfully able to avoid movement to the front during the final assessment, no activation of SOL can be observed. This also coincides with a more even distribution of the subject's weight between left and right leg during this assessment (see figure 6.4). The activation peak in the EMG data of the VAS muscle during the third assessment indicates that the subject tended to extend his knee in response to the perturbation. With a perturbation to the right the tibialis anterior (TA) was, as expected, not active.

Perturbation to the front

Figure 6.8 shows the plots of the EMG data measured during all four assessments as the subject was perturbed to the front.

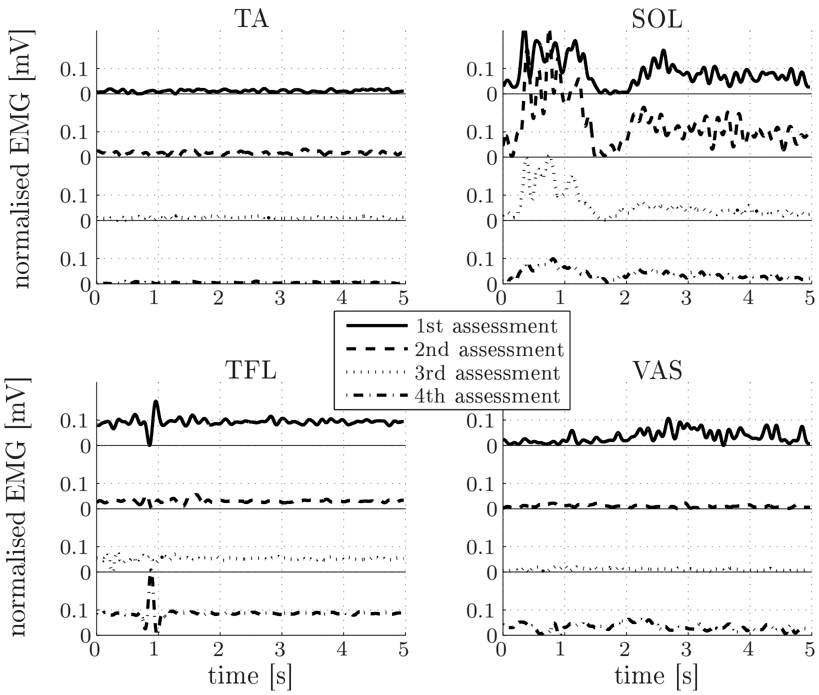


Figure 6.8.: Average normalised measurements of the EMG signal measuring the activities of TA, SOL, TFL, and VAS muscle groups of the right leg of all the assessments. The subject was perturbed to the front.

Stabilisation of the body during perturbations to the front mainly involves the SOL muscle group and therefore the TA muscle group is not activated. Activation of the VAS muscle group during the 1st assessment indicates that the subject extended his knee. During the subsequent assessments the VAS muscle group remained inactive. In figure 6.8 a distinct activation of the SOL muscle group during the first three assessments can be seen which might correspond to knee flexion in response to the perturbation. This is reduced during the last assessment. The activation of the TFL muscle group indicates that the subject moved not only to the front but to the right as well.

Perturbation to the back

Figure 6.9 shows the EMG data obtained during the assessments as the subject was perturbed to the back.

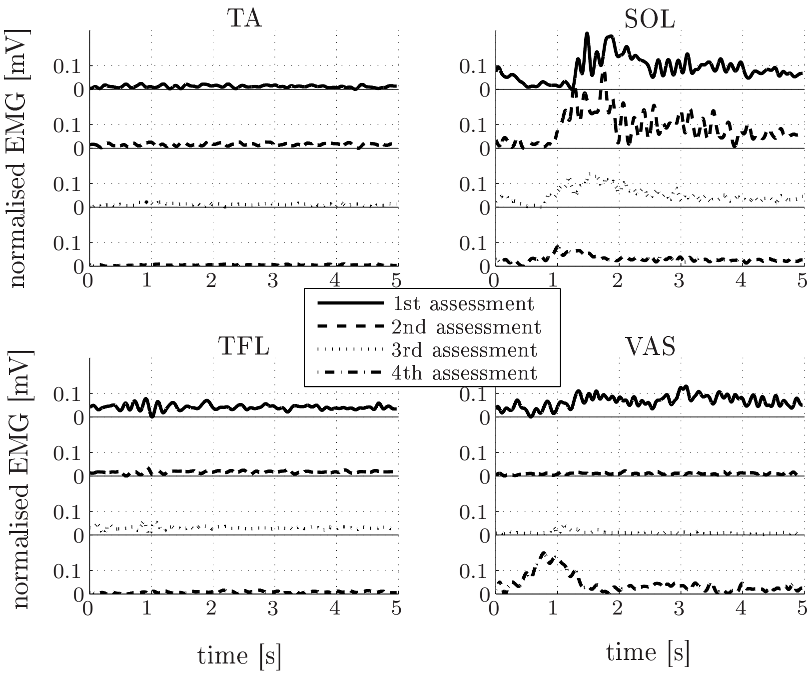


Figure 6.9.: Average normalised measurements of the EMG signal measuring the activities of TA, SOL, TFL, and VAS muscle groups of the right leg of all the assessments. The subject was perturbed to the back.

During perturbations to the back the TA and VAS muscle groups should be activated. Figure 6.9 shows a clear activation of the VAS only during the last assessment, while the subject did not activate the TA muscle group during any of the four assessments.

The results recorded during the first two assessments give a clear sign of SOL activity which shows that the subject moved not only to the back due to the perturbation but later to the front as he returned to the upright position. The TFL muscle group was active during the first assessment but was no longer in use during later assessments. In the 4th assessment, the VAS is active before the SOL which indicates that the subject extended his knee in response to the perturbation before recovering.

6.5. Discussion

The results show that throughout the training programme, changes in ground reaction forces and in the muscle activation did occur, with effects on the subject's ability to balance.

6.5.1. Statistical analysis of force data

The force trajectory data shown in figure 6.3 indicate that the vertical forces during the recovery from the perturbation were reduced throughout the training period. Analysis of the characteristic values (cf. figure 6.6) shows that all characteristic values, except for the undershoot time, had changed significantly during the final assessment.

The start value (top left plot in figure 6.6) indicates how much weight is initially placed on the impaired leg. As confirmed in figure 6.4, the subject shifted more weight onto the unimpaired side during the first 3 assessments. The significant increase in this value during the final assessment shows that the subject was confident to distribute his weight more evenly between the two legs. Note, that although the data shown here focus on perturbations to the right, the subject was perturbed randomly and could therefore not anticipate the direction of a perturbation.

Similarly to the start value, the final values of the vertical force (F_z) measured under the right foot (top right plot in figure 6.6) show a significant change during the last two assessments which show that the subject used his impaired leg more than at the beginning of the experiments.

Changes in peak value and time (middle plots in figure 6.6) are inconclusive and cannot be attributed to the training progress, as they are mainly a direct result of the perturbation application. However, a significant reduction in peak time during the final assessment indicates that the subject was able to counteract the perturbation faster, probably as a result of the more favourable initial weight distribution.

While the undershoot time (bottom right plot in figure 6.6) did not change significantly throughout the assessments, the undershoot value (bottom left plot) decreased significantly during the last two assessments. This shows that the subject put less weight onto the left (unimpaired) leg as he was returning to the initial position which indicates that the perturbation was counteracted more accurately and that his balance improved.

This analysis indicates that the subject's ability to balance improved significantly over the course of training, together with the confidence to put more weight onto the impaired leg. The improved balance ability is a result of a combination of improved ability in the affected leg, together with better coordination and integration with upper body movement. Observation of the subject during the assessments gave the impression that he was more at ease at the very end of the experimental period than during the first sessions.

6.5.2. Analysis of EMG Data

The EMG data presented in figures 6.7-6.9 show a tendency of reduction in muscle activity in response to the perturbation following the intervention period. It appears that during the course of training the subject develops a strategy of muscle activation which allows him to react to the perturbation in a more efficient way. This might be due to the fact that better coordination between leg muscle groups and with the upper body may allow the subject to reduce the contribution from the leg muscles. The change in CoP trajectory shown in figure 6.5 also indicates that the improved balancing skills lead to a decreased movement of the CoP and, consequently, decreased EMG activity.

The responses in the SOL muscle group indicate that over the intervention period a strategy is developed in response to perturbations to the front which reduces the effort of this muscle group (cf. figure 6.8).

As figure 6.9 shows, the subject is not able to activate his TA muscles as there is no sign of contraction in the EMG data. This also can be seen in the fact that the subject's foot still dropped after the experiments were concluded.

Although the EMG data presented here give some indications of neuromuscular adaptations following the training period, the results remain overall inconclusive. It may be necessary to include activity at the hip in the analysis to obtain a more complete picture of the activity following perturbations.

6.5.3. Balance Training and Sensory Electrical Stimulation

After the first two weeks of training the subject showed more confidence in shifting his body weight onto the impaired leg as the vertical force measured under the right foot increased significantly compared to the values measured during the first two assessments (see figures 6.3(b) and 6.6). This suggests that using balance training for rehabilitation in chronic stroke could improve confidence during standing and walking and reduce the risk of falling.

Our findings show the largest improvements in balance ability during the final training period, when sensory electrical stimulation combined with the use of a rehabilitation assisting device, the modified BalanceTrainer, was applied. While the value of the undershoot remained unchanged during the final training period, the start and final values of the vertical force continued to improve and were significantly different from the results using the rehabilitation assisting device only (see figure 6.6). In addition, the CoP displacement (cf. figure 6.5) illustrates that the subject was able to counteract the perturbation after the final training period in a more confident and precise way, without significant movement to the front which was still present at the 3rd assessment.

This study shows that the active balance training is a rehabilitation technique which may be combined with sensory stimulation. While it illustrates the feasibility of combining active balance training with sensory electrical stimulation, the limitation to a single subject case does not allow to attribute the improvements during the final training period to the added electrical stimulation. It indicates, however, that adding electrical stimulation may benefit the outcome of the rehabilitation programme.

The primary aim of the electrical stimulation used in this study was to provide sensory input to aid neurological rehabilitation. Transcutaneous electrical stimulation will, however, affect both sensory and motor neurons. While the stimulation level was chosen in such a way that no superficial muscle contraction could be observed, the stimulation may still have activated motor units and therefore acted not solely as sensory stimulation. More detailed neurophysiological assessments would need to be conducted to ascertain the precise effect of the stimulation on the different sensory and motor pathways.

While our stimulation procedure as described in section 6.3.4 requires the subject to be able to feel the sensation in order to set the stimulation intensity, sensory electrical stimulation may also be applicable in subjects without sensation, but in whom lower sensory pathways are intact. In these subjects our approach may still lead to peripheral or central neural adaptations as a result of afferent inputs elicited by stimulation [134].

6.6. Conclusions

In this case study a new training approach for chronic stroke patients was introduced using the modified BalanceTrainer. Contrary to the experiments carried out with the SCI subject (see chapters 4 and 5) where functional electrical stimulation was used as part of the controlling system supporting the subject in his task of balancing, the training performed with a stroke subject used the voluntary response combined with a stimulation only on a sensory level in order to help the subject to "remember" which muscle groups to use and when to activate them for respective directions of perturbation.

Before the training started the balance performance of the subject was assessed. For the evaluation of performance the known characteristic values for an impulse response were used.

Measurements of vertical forces under the subject's feet show that the subject improves balance over the course of training, with the biggest change seen during the final assessment following a training period with applied sensory stimulation. This may suggest that this type of stimulation can enhance the outcome of dynamic balance training. Further investigations with a larger subject group together with a training regime which randomises the order of training with or without electrical stimulation are needed to verify this hypothesis. A further suggestion for future studies would be to record the kinematics of the subject's lower and upper limbs as well as of the upper body and the movement of the pelvis, in addition to the force data. While making the experimental setup more complex, these measurements would give a more detailed picture of the changes in performance, allowing to analyse the changes in the upper body movement.

The EMG responses did not show whether the stimulation had an effect on neural adaptations leading to a reactivation of the paralysed muscles.

While conclusions drawn from the results in this case study are limited and at this stage cannot be generalised, evidence in the literature suggests that sensory electrical stimulation can modulate the neural motor circuitry after neurological impairment [134] and appears to be a valuable addition to training programs [145]. Future experiments performed in a clinical setting with a larger subject group, using the methods and techniques introduced here, might give an answer to the question of whether balance training with sensory stimulation consistently improves balance performance in this population, while additional neurophysiological assessments would be necessary to verify the source of any adaptation observed.

6.7. Acknowledgements

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7. Conclusions and Outlook

This thesis focused on a new approach of supporting paraplegic patients during standing by employing a control approach which takes the voluntary movement of the paralysed person into account. Furthermore, a new training approach for chronic stroke patients was introduced combining sensory electrical stimulation with active balance training using a specially adapted standing frame.

7.1. Conclusions

For supporting paraplegic subjects during standing Jaime *et al.* [105] proposed a nested control loop with a moment controller in the inner loop controlling the ankle moment and an angle controller in the outer loop controlling the inclination angle of the lower body. The subject was kept in a double-link inverted pendulum configuration and was able to move with the upper body freely. The paralysed shank muscles produced due to functional electrical stimulation enough stiffness to stabilise the paraplegic subject during standing. With the control approach suggested by Jaime *et al.* the voluntary movement of the subject's upper body was assumed to be an unknown disturbance.

The aim of this thesis was to develop a new control approach which would allow paraplegic subjects to stand in a more natural way. For this new control approach, called Integrated Voluntary Control (IVC), a modified nested control loop was used. In the outer loop the ankle moment, produced by the paraplegic subject due to his voluntary movement of the upper body, was estimated with the help of a mathematical model using the inclination angles of upper and lower body. This estimated ankle moment was then compared with the actual moment applied onto force plates the subject was standing on. With this control error a moment controller in the inner loop induced an appropriate stimulation signal which was then applied to the paralysed shank muscles.

For comparison of performances achieved with different control approaches an evaluation value is needed. Due to the nature of the applied pseudo random binary signal as perturbation the commonly used evaluation values, like steady state values or rise times, were not suitable. Therefore, a new evaluation method was defined and verified, which is based on the variance of a certain measured time signal around its mean value.

To see whether the variance is suitable for evaluating the performance of a subject during standing, experiments were carried out with four able-bodied subjects using the approach presented by Jaime *et al.* [105]. For these experiments the subjects were standing in a specially adopted standing frame that prevented the able-bodied subjects to stabilise themselves by their own muscle force. This was achieved by letting them stand on a rotating platform attached to the standing frame. Hydraulic actuators attached to the base of the standing frame provided the stabilising torque which depended on a subject specific muscle stiffness and viscosity. Before the experiments were carried out these specific stiffness and viscosity values were established using the approach suggested by Black *et al.* [117]. These nominal values were then reduced in order to see whether or not the evaluation value is reflecting correctly the increased effort with the decreased support by the hydraulic actuators.

The evaluation value reflected correctly the increased effort of the subject when decreasing the support of the hydraulic actuators by higher variance values. Furthermore, this evaluation value allowed the comparison of performances of different subjects achieved with the same control approach. One outcome of such comparison suggests that the muscle stiffness has a bigger impact on the balance performance than the muscle viscosity. Thirdly, it was also possible to compare performances of one subject carrying out experiments with different control approaches. Concluding from these results the variance as evaluation value proved to be suitable for evaluating the performance of subjects during standing.

Similar experiments were carried out with one spinal cord injured subject. The experimental setup was the same as it was used with the able-bodied subjects. Instead of the rotating footplate, FES at the shank muscles was now used to generate an appropriate stabilising moment at the ankle. As the stiffness and the viscosity of the SCI subject could not be established in the same way as it was done with the able-bodied subjects these values were chosen by trial and error using a value which would not cause the muscles to fatigue too quick but on the other hand stabilise the subject over a certain period of time.

These established results were then compared with the results using the new IVC approach. For comparison, the variances of the inclination angles of the lower and upper body, the pulsewidth of the applied stimulation signal, and the torque applied onto the ground were used. Although the results achieved with the new IVC approach showed e.g. for the stimulation signal slightly higher variance values, a statistical analysis showed no significant differences between these two approaches.

These concepts of balance control were then developed for applications in stroke patients, where a new training approach was introduced combining the voluntary abilities of a patient during balancing together with stimulation applied to the subject's paralysed leg. For the experiments a special standing frame was used which allowed perturbations in eight different directions. Depending on the direction of perturbation a different group of muscles was stimulated. The stimulation strength was chosen only to stimulate the sensory nerves and not to provide any functional support. The subject was brought out of balance using an impulse like perturbation. After the subject had been perturbed the sensory stimulation of the paralysed leg was switched on at a distinct point of time and was kept on for a certain period of time. With this approach it was intended to help the subject to "remember" which muscles to use during the respective direction of perturbation. The results achieved without stimulation were then compared with the results acquired during the experiments with stimulation. For the evaluation of the subject's performance known evaluation values like the steady state value, the rise time and others were used. The statistical evaluation showed a significant improvement with the use of the stimulation compared to the experiments performed without stimulation.

7.2. Outlook

The experimental studies with the paraplegic and stroke subject were carried out as single case studies to show the feasibility of the new approaches. To give a more precise statement about the significant improvement of the approaches compared to the known ones it would be worth to investigate both approaches with a bigger subject group.

Specifically, extended experiments with several paraplegic subjects should clarify whether the IVC approach could help to improve their performance significantly in terms of reduced applied stimulation and consequently in prolonged duration of standing. The use of the recommended values for inertias of the lower and upper body from the literature resulted in a very sensitive model of a paraplegic subject towards disturbances. In order to increase the robustness of the model and improve the safety of the subject the inertia of the upper body was adjusted heuristically. A deeper investigation for determining the inertias more precisely might lead to an improved model and therefore to improved performance of the paraplegic subject. A further way of improving performance could be using a more detailed model as suggested e.g. by Kim *et al.* [146].

As for the second part, experiments with the stroke patient were performed first without stimulation and then in a second series sensory stimulation was added. Experiments with several stroke patients might reveal whether the observed significant improvement in performance will be achieved too, if the experiments will be carried out with stimulation first and then without. These findings suggest to carry out this new training method in a clinical trial with more stroke patients and compare it to the results achieved with traditional rehabilitation methods.

Despite all the research and suggestions of advanced control approaches controlling the posture of SCI patients during standing the use of static standing frames is still very common. Beside the high costs of such devices, the systems suggested are far too complicated to be used by medical personnel in a hospital environment or by the patients themselves at home and are therefore not attractive to be used. Dynamic standing frames, like the commercially available BalanceTrainer (see also section 3.3) have been used successfully in rehabilitation with stroke patients already. Therefore, it is suggested to tailor this existing BalanceTrainer towards the needs of spinal cord injured subjects. The BalanceTrainer could be adapted with sensors measuring the inclination angles. The necessary control could be performed by a smart phone connected to the standing frame via its existing interface using an application running on the smart phone. This way, the equipment would be ready to be used for many different control approaches which the user could simply download onto the smart phone and would motivate the patient more to exercise.

Appendix

A. Specification of the MRF

Hydraulics

Hydraulic Pump

model	Knapp Microfluid AKA 5K T2A
motor power	1.1 kW
maximum work pressure	90 bar
theoretical flow rate	5.8 l/min
electrical supply	380 V AC 3-phase

Servo Valve

model	MOOG E760–100
rated no–load flow @ 70 bar	3.85 l/min
maximum work pressure	210 bar
rated current	$\pm$ 15 mA parallel
rise time	ca. 6 ms

Rotary Actuator

model	Knapp Microfluid DA12 270 W
maximum torque	120 Nm
maximum work pressure	100 bar
angle of rotation	270°
absorption volume	68 cm^3/a
friction breakaway pressure	$<$ 10 bar

Sensors

Pressure Transducer

model	MP Filtri TR4002
pressure range	0 - 100 bar
output	0 - 10V DC
rise time	1ms
electrical supply	13 - 30V DC

Shaft Encoder

model	Hengstler absolute rotary encoder RA58
resolution	12 bit
output	TTL
interface	parallel, Gray code
electrical supply	5V DC

Data Acquisition Cards

model	Humusoft AD512	
bus system	ISA	
analog input	number	8 SE
	resolution	12 bit
	input range	$0-5$, $0-10$, ± 5, ± 10V
	sampling rate	100 kHz
analog output	number	2
	resolution	12 bit
	output range	$0-5$, $0-10$, ± 5, ± 10V
	maximum output current	10mA
digital input/output	number	8
	level	TTL

model	National Instruments PCI-6503

bus system	PCI	
digital input/output	number	24, programable
	level	TTL

Force Plates

model	Advanced Mechanical Technology (AMTI) OR6–7–2000	
capacity	F_x, F_y	4450 N
	F_z	8900 N
	M_x, M_y	2300 Nm
	M_z	1100 Nm
natural frequency	F_x, F_y	370 Hz
	F_z	530 Hz

sensitivity	F_x	0.3358 μV/V/N
(force plate left)	F_y	0.3346 μV/V/N
	F_z	0.0853 μV/V/N
	M_x	0.7789 μV/V/Nm
	M_y	0.7800 μV/V/Nm
	M_z	1.6829 μV/V/Nm

sensitivity	F_x	0.3372 μV/V/N
(force plate right)	F_y	0.3370 μV/V/N
	F_z	0.0862 μV/V/N
	M_x	0.7957 μV/V/Nm
	M_y	0.7939 μV/V/Nm
	M_z	1.6888 μV/V/Nm

Zebris System

system	CMS-HS
number of marker channels; basic version	10 + 2 pointer

Measurement distance (for one measuring unit)	max. 2.5 m
	@ 80-100Hz: 1.8m - 2.5 m
Measurement	max. 100Hz per marker

Ultrasound Markers

dimensions of marker with attachment plate	7 x 6 mm (diameter x hight)
emission angle	min. 120°

Electric Motors

BLPM - Brushless Permanent Magnet Motors Type AMG 24V, 0.2kW http://www.iskra-ae.com/eng/blpm_motors.php

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