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ENABLING BALANCE TRAINING IN ROBOT-ASSISTED GAIT REHABILITATION

A thesis submitted to attain the degree of

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presented by

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"Experience never errs; it is only your judgments that err by promising themselves effects such as are not caused by your experiments.
- Leonardo da Vinci.

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Abstract

Stroke is the leading cause of disability in the developed world. The ability to walk is often affected and a main focus for rehabilitation to regain mobility and independence. Several methods have shown effective, including robot-assisted treadmill training. Clinical trials with individuals post-stroke suggest that especially those individuals with severe impairments may benefit from robot-assisted training, even though only certain elements of gait can be trained with current devices. As soon as patients overcome the sub-acute stage of stroke, however, they need less restrictive forms of therapy. At this point, motor learning needs to be promoted by greater challenge.

It is hypothesized that dynamic balance, like standing balance, plays an important role for gait rehabilitation. Dynamic balance especially concerns balance in the frontal plane during walking, namely lateral movements of the center of mass, which are controlled by foot placement. This training requires a lateral degree of freedom (DoF), which is not present in most current robotic devices.

The goal of this work is to develop a new design for a robotic gait rehabilitation environment that enables balance training. To define the requirements for such a system, knowledge concerning human gait and the influence of robotic interaction is needed, in terms of kinematics and kinetics.

To first assess the role of kinematics, a small study with individuals post-stroke is conducted using the treadmill-based Lokomat FreeD, which features DoF that enable weight shifting, and which guides the subjects' legs and pelvis rigidly. Comparing this device to a conventional Lokomat, which restricts guided locomotion to the sagittal plane, no significant changes are found in the subjects' muscle activa-

tion patterns. This suggests that the incorporation of balance-related DoF alone does not suffice, but that the type of human-robot interaction is critical to enable effective balance training.

To further guide design, the effects of forces acting on the pelvis and upper body on balance control during walking are assessed in a study with healthy subjects. The combination of additional inertia and partial body weight support (BWS) leads to reduced step width. Analysis of these results suggests that BWS induces a "pendulum effect" that assists balance, thereby reducing the challenge.

In light of these findings, the mechanical design paradigm of "constrain-as-needed" is proposed: Instead of design for rigid guidance, the system should leave freedom for kinematic variability wherever possible. This leads to a second main requirement: a high degree of transparency. Transparency refers to the ability of the robot to "get out of the way" and not interfere with the user's movements. The design paradigm is in line with clinical studies showing that more freedom increases active patient participation during gait training, which is believed to be a key factor for therapy success.

To achieve high BWS transparency and overcome the pendulum effect, an extension to two-dimensional BWS is designed. The design mechanically decouples the additional actuated low-force lateral DoF from the high-force vertical DoF. It is demonstrated that the system accurately tracks lateral user movements at different levels of vertical support load, which leads to very low lateral interaction forces.

To assist weight shifting, a multidimensional compliant decoupled actuator (MUCDA) for pelvic support is proposed. The mechanism comprises one motor and a spring assembly in which each spring serves as a series actuation component for lateral forces, and simultaneously provides passive compliance in the five unactuated DoF. Six-DoF force and torque sensing are realized via a model of spring deformation characteristics in combination with low-cost inertial and optical sensors. Experimental evaluation demonstrates that the system follows physiological weight shifting with low interaction forces and also has little undesired impact on other pelvis motions.

The BWS and MUCDA modules can be combined with each other and also with a low-inertia exoskeleton that supports foot placement by means of compliant parallel actuators. In the future, the system can be used to realize challenging balance exercises, and to conduct clinical studies assessing the role of balance for rehabilitation.

Zusammenfassung

Schlaganfall ist die Hauptursache für Behinderungen in den Industrieländern. Häufig ist nach einem Schlaganfall die Gehfähigkeit beeinträchtigt und steht im Mittelpunkt des Rehabilitationsprozesses, um Mobilität und Selbstständigkeit im Alltag wieder zu erlangen. Mehrere Therapieansätz haben sich bewährt, darunter auch roboterunterstütztes Laufbandtraining. Klinische Studien mit Schlaganfallpatienten deuten darauf hin, dass besonders die stark beeinträchtigten Patienten von einem robotergestützten Training profitieren. Sobald Patienten jedoch das subakute Stadium des Schlaganfalls überwunden haben, brauchen sie eine Therapie mit mehr Freiraum. Dann benötigt das motorische Lernen mehr Herausforderung.

Dem dynamischen Gleichgewicht, wie auch dem Gleichgewicht im Stand, wird eine wichtige Rolle für Gangrehabilitation zugeschrieben. Das dynamische Gleichgewicht beschreibt das Gleichgewicht in der Frontalebene beim Gehen, insbesondere die Verschiebung des Körperschwerpunkts, die durch die Platzierung der Füsse geregelt wird. Dieses Training erfordert vor allem einen seitlichen Freiheitsgrad des Robotersystems, der in vielen gängigen Geräten nicht vorhanden ist.

Das Ziel dieser Arbeit ist es, ein neues Design eines Robotergang-Rehabilitationssystems vorzustellen, welches Gleichgewichtstrainings beim Gehen ermöglicht. Um die Anforderungen an ein solches System zu definieren, muss der menschliche Gang und der Einfluss der Interaction mit einem Roboter verstanden werden, im Sinne von Kinematik und Kinetik.

Um zunächst die Rolle der Kinematik zu bestimmen, wird eine

Studie mit Schlaganfallpatienten mit dem Lokomat FreeD durchgeführt, dessen Freiheitsgrade auch die seitliche Gewichtsverlagerung ermöglichen, und der die Beine und den Pelvis steif führt. Im Vergleich zum konventionellen Lokomat, welcher den geführten Gang auf die Saggitalebene beschränkt, konnten keine signifikanten Veränderungen im Muskelaktivierungsmuster festgestellt werden. Dies deutet darauf hin, dass zusätzliche gleichgewichtsbezogene Freiheitsgrade nicht ausreichen, sondern dass die Art der Interaktion zwischen Mensch und Roboter entscheidend zu effektivem Gleichgewichtstraining beiträgt.

Um den Entwurf weiter zu lenken, werden die Auswirkungen von Kräften am Pelvis und am Oberkörper auf Gleichgewicht während des Gehens in einer Studie mit gesunden Probanden untersucht. Die Kombination zusätzlicher Trägheit und teilweiser Körpergewichtsentlastung (BWS) führt zu reduzierten Schrittweiten. Analyse dieser Resultate deutet auf einen stabilisierenden "Pendeleffekt" von BWS hin, was die Herausforderung der lateralen Gleichgewichtsaufgabe reduzieren kann.

Daraus leitet sich hier das Prinzip der "Einschränkung-wie-erforderlich" ab: Statt einer Konstruktion für steife Führung sollte das System wo immer möglich Freiheit für kinematische Variabilität lassen. Dies führt zu einer zweiten Hauptanforderung: Einem hohen Grad an Transparenz. Transparenz bezieht sich auf die Fähigkeit des Roboters, "aus dem Weg zu gehen" und die Bewegungen des Benutzers nicht zu stören. Das Konstruktionsleitprinzip passt zu klinischen Studien, die gezeigt haben dass Freiraum die aktive Teilnahme des Patienten am Training fördert, was als Schlüsselfaktor für den Rehabilitationserfolg angesehen wird.

Um hohe Transparenz für BWS zu erreichen und den Pendeleffekt zu vermeiden, wird eine Erweiterung auf ein zweidimensionales System entworfen. Der Entwurf entkoppelt den zusätzlichen, angetriebenen lateralen Freiheitsgrad, in dem niedrige Kräfte nötig sind, vom vertikalen Freiheitsgrad, in dem hohe Kräfte auftreten. Es wird gezeigt, dass das System den seitlichen Bewegungen eines Nutzers bei unterschiedlicher vertikaler Unterstützungskraft exakt folgt und dadurch sehr niedrige Seitenkräfte erzielt.

Zur Unterstützung der Gewichtsverlagerung beim Gehen wird ein

mehrdimensionaler, nachgiebiger, entkoppelter Aktuator (MUCDA) für das Becken vorgestellt. Der Mechanismus besteht aus einem Motor und einem Federkissen, in dem jede Feder die Rolle eines seriellen Aktuatorelements übernimmt und gleichzeitig passive Nachgiebigkeit in den nicht-aktuierten Freiheitsgraden gewährleistet. Kraft- und Positionsmessung in sechs Freiheitsgraden werden über ein Modell der Federverformungseigenschaften in Kombination mit einem kostengünstigen IMU und optischen Sensoren realisiert. Die experimentelle Auswertung zeigt, dass das System der physiologischen Gewichtsverlagerung mit geringen Interaktionskräften effektiv folgen kann und auch die anderen Beckenbewegungen kaum beeinflusst.

Die BWS- und MUCDA-Module können miteinander und auch mit einem Exoskelett geringer Trägheit kombiniert werden, das Fusspositionierung auf dem Laufband mithilfe nachgiebiger, paralleler Aktuatoren unterstützt. Zukünftig kann das System genutzt werden, um herausforderndes Gleichgewichtstraining zu realisieren und um in klinischen Studien die Rolle des Gleichgewichts für Rehabilitation zu untersuchen.

Preface

The experimental work, evaluation and writing of this thesis was performed at the Sensory-Motor Systems (SMS) Lab, Institute of Robotics and Intelligent Systems (IRIS), Department of Health Science and Technology (D-HEST), ETH Zurich, Switzerland.

The following chapters of this cumulative dissertation are based on publication manuscripts:

Chapter 3, based on the conference publication [74]:
Andrew Pennycott, Dario Wyss, Heike Vallery and Robert Riener. Effects of Added Inertia and Body Weight Support on Lateral Balance Control During Walking. In *IEEE International Conference on Rehabilitation Robotics (ICORR)*, pages 1-5, 2011
 Part of the introduction of the original article has been moved to the introduction of Chapter 2, for consistency of the narration.

Chapter 4, based on the conference publication [113]:
Dario Wyss, Volker Bartenbach, Andrew Pennycott, Robert Riener and Heike Vallery. A Body Weight Support System Extension to Control Lateral Forces: Realization and Validation. In *IEEE International Conference on Robotics and Automation (ICRA)*, pages 328-332, 2014

Chapter 5, based on the journal publication [114]:

Dario Wyss, Andrew Pennycott, Volker Bartenbach, Robert Riener and Heike Vallery. A MUltidimensional Compliant Decoupled Actuator (MUCDA) for Pelvic Support during Gait. In *IEEE Transactions on Mechatronics*, 2018

The chapters are introduced by a short section for the reader's understanding and are edited to avoid redundancy throughout the thesis, and to achieve homogeneous structure.

Declaration of Originality

I hereby declare that the written work I have submitted entitled

Enabling balance training in robot-assisted gait rehabilitation

is original work which I alone have authored and which is written in my own words.[1]

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Contents

Chapter 1

General Introduction

1.1 Background

Stroke is a main contributor to acquired disability in the world [24]. Enhanced medical care after a stroke leads to increasing survival rate and consequently to an increasing population with stroke-related disability [25]. Gait impairment is prominent within this population and contributes to long-term disability [58]. Improved walking was found to be one of the most important goals stated by people undergoing rehabilitation [10].

Promising results have been shown with rehabilitation programs involving body weight supported training on a treadmill [31]. Two or three therapists supervise and guide an individual during walking on the treadmill. This training is task-specific and allows a high repetition of the movements. A major drawback is that the training duration can be limited by the fitness of the therapists, as the correction of a patient's gait pattern can be physically strenuous. To overcome this limitation, robot-assisted therapy was developed to reduce the physical effort of the therapists and thereby to increase the duration and intensity of the training. In addition, a robot offers the possibility to quantitatively assess an individual's performance and progress.

1.2 Robotic Gait Rehabilitation

Robot-assisted training has been of great interest in research over the past years and is now widely used in rehabilitation, as several devices are available as commercial products. These robotic systems can be classified broadly into two groups: overground and stationary walking devices.

Stationary systems consist of a fixed structure, a body weight support system (BWS) to partially unload the user's body weight, and a movable ground platform. A division into subclasses can be made on the basis of this platform into end-effector-based systems and exoskeleton-based systems.

End-effector-based systems use movable footplates connected to the patient's feet to reproduce physiological gait trajectories and simulating stance and swing phases. Examples are the G-EO-system (Reha Technology AG, Olten, CH) and the Gait Trainer (Reha-Stim Medtec, Berlin, DE). The Haptic Walker [88] extends this principle even to simulate walking on different surfaces.

Exoskeleton-based systems use a treadmill with a robotic exoskeleton connected to the patient's legs to guide and control lower-limb joints directly. Representatives of this group, the Lokomat (Hocoma AG, Volketswil, CH) and the ReoAmbulator (Motorika USA Inc., New York, USA), are both commercially available devices and share a common principle of operation. The user is guided by a powered exoskeleton through the gait cycle, where the motions are restricted to the sagittal plane. The Lower Extremity Powered ExoSkeleton (LOPES) [61] was designed by the University of Twente to combine the advantages of end-effector and exoskeletal systems. It comprises a so-called shadow leg as an intermediate element between the motors and the user and has eight actuated DoF.

Other commercially available systems, like overhead support systems, provide more freedom for the user. Therefore, these are mainly suitable for individuals who have already regained some of their walking ability and rather need a safe environment and assistance in weight bearing. Examples are the Zero-G [34], The Float [95], or the RYSEN [78], which provide body weight support during free walking overground in a large workspace. Within these systems, the patient is fixed in a harness in order to avoid falling. There is no actuation

of any joint that could facilitate leg movements.

Overground walking can also be enabled by mobile devices that support the user's movements by either partially unloading the body weight or by inducing forces on the human limbs. KineaDesign developed the KineAssist [75], which is a mobile robot attached to the user at the pelvis, providing assistance at the pelvis and torso while having no influence on the lower limb's movements. The Rewalk device from Rewalk Robotics is an actuated exoskeleton fixed to the patient's legs. As the system is autonomously powered by a battery, the patient can walk freely overground. The device is built for individuals with enough residual function to initiate movements as it does not provide body weight support and does not offer specialized training. Another representative of this group featuring a powered exoskeleton is the Hybrid Assistive Leg (HAL) [48].

1.3 Evidence of Robotic Gait Rehabilitation

Most studies in this field have been conducted as pilot studies, as the majority are research devices. These studies thus cannot make a large contribution to the general body of clinical evidence. Nonetheless, comparative studies have been carried out with the Lokomat and the Gait Trainer; these devices will be in focus in this review.

Individuals in the acute phase after stroke were assessed in a pilot study comparing the progress of a group treated with conventional training and a group treated with a combination of robotic (Lokomat) and conventional training. Both groups demonstrated significant improvements in terms of functional walking scores but no significant difference was found between the groups [44]. However, a study with a similar design found greater improvements for the group trained with robotic assistance with the Gait Trainer [76].

Subacute patients were enrolled in tests with one group trained with conventional training and another group undergoing a combination of conventional and Lokomat training. The walking scores and neuro-assessments were in favour for the robotic group, whereas no difference was found for the timed walking tests [89]. Similar re-

sults were found in a multi-center study with subacute patients in Germany. Individuals treated with a combination of conventional training and with the Gait Trainer showed more improvements in independent walking and secondary variables such as speed and endurance [79]. More improvements in timed walking tests and motor scores after Lokomat training compared to conventional training were also found in a further study [59]. Conversely, another study comparing conventional training with Lokomat training found that participants in the conventional training group achieved better results in timed walking tests. No differences were found concerning functional ambulation [36]. Also, no significant differences between a group training with the Gait Trainer and a group undergoing manual treadmill therapy were found at six months follow-up [108]. A study investigating the differences between Lokomat therapy and manual assisted over-ground therapy with similar therapy volume in subacute patients found no differences in motor scores and walking speeds [101].

Individuals in the chronic phase after stroke were investigated in a study comparing Lokomat training with manual treadmill therapy, where no significant differences were found [109]. Similar results were demonstrated in a study comparing training with the Lokomat and overground training. The participants from both groups improved in motor score tests but no differences were found between the groups [49]. Another study found contrary results, with greater improvements in stance time and walking speed after therapist-aided training compared to robot-assisted training with the Lokomat being found [42]. A study with five participants was performed using the LOPES, and the authors observed significant gains in walking speed, distance and joint range of motion [91].

In summary, the studies show conflicting results, and the efficacy of robot-assisted training seems to be comparable to manual training. Especially in the early stage of the rehabilitation process of stroke survivors, the potential of robotic devices seems to be greatest. Acute patients are believed to benefit most from the high degree of assistance and guidance the robotic devices offer in contrast to chronic patients with partially regained walking ability.

1.4 Relevance of Balance Training

An underlying cause of reduced efficacy could include a generally lower level of volitional input due to over-reliance on robotic support for producing gait, leading to subject passivity. Furthermore, the pelvis is often locked in the horizontal plane, which is the case, for example in the Lokomat. There is evidence that a horizontally locked pelvis changes the gait pattern significantly [104].

Some further assumptions are given by Hornby and Hidler to explain the results of their studies: the conventional Lokomat system restricts pelvis and trunk movement, thus rendering volitional postural control superfluous [41] and preventing weight shifting from right to left leg [36]. It has been demonstrated that even when healthy subjects walk in the Lokomat with different levels of guidance, there is a reduction in the activity of muscles needed for maintaining stability [100].

Weight shifting and maintaining stability feature as the main focus of the labor-intensive manual therapy; indeed, these factors may be the main reasons for the better effects of manual therapy compared to robot-assisted training in chronic patients [36]. Patients in the sub-acute and chronic phases may require a form of therapy that is specifically focused on training balance, coordination, postural or distal control. Such training may lead to a more physiological gait pattern that is more energy-efficient, less strenuous, faster and more stable. This is also supported by the study of [28], which underlines that the training of balance is a key goal of rehabilitation, and is also important for lowering the risk of falling for individuals post-stroke [17].

Whilst these restrictions are not necessarily a limitation - and may in fact be advantageous - during the early stages of rehabilitation where training should be simplified and not overwhelm patients [35], they impose limitations on the efficacy of training with less impaired individuals. It is, therefore, desirable to remove these overly restrictive constraints in order that the Lokomat be more effective in later training phases and for people who have lower levels of impairment.

After a stroke, the complex interplay of balance and locomotion control mechanisms needs to be relearned to regain a stable gait pattern. It is difficult to perform the training for such re-learning in a

safe environment during locomotion. Conventional balance training devices such as wobble boards train balance during stance but are detached from locomotion and support.

During walking, the main challenge of balance control is posed in the frontal plane, where foot placement is used to keep the centre of gravity in a step-to-step-wise stable trajectory, and is mainly realized via hip abduction. Critical factors include step width; variability in step width indicates the degree of difficulty experienced for balance and vulnerability to falls. Further variables such as the orientation of the pelvis, particularly in the frontal plane, may play a secondary role in balance maintenance. The lack of stabilization of these variables in neurologically impaired people is also often the cause of secondary gait deviations, suggesting that these correspond to an important aspect for training during gait rehabilitation.

1.5 Goal and Outline of this Work

In this work, it is hypothesized that enabling balance training within a robotic gait training device is crucial for rehabilitation efficacy. The basic prerequisite to tackle this hypothesis is to provide a robotic gait trainer allowing for balance training. How balance training can be made technically possible and which principles to follow in the design process, is investigated in this work. An overview of the thesis is shown in Fig. 1.1.

A clinical trial with stroke survivors is conducted to assess the efficacy of a commercially available extension to the Lokomat (Lokomat FreeD), which allows for weight shifting through lateral translation of the pelvis and the leg cuffs. The subjective comfort as well as muscle activation patterns and the kinematics of the upper body are studied. The results show that replicating the kinematics of balance training is not sufficient and attention must be paid rather to impedance: that is, on the interaction forces between the human and the robot.

The interaction forces are mainly exchanged over the pelvis attachment and the body weight support system. To better understand these interaction forces and how they affect the gait pattern of the human, a study with healthy subjects is conducted. The effects of added inertia on the pelvis and partial body weight support on lat-

eral balance control during walking are assessed. The results suggest a stabilizing effect on the subject, rendering the task of dynamic balancing less challenging.

Based on these findings, a two-dimensional body weight support system extension is designed which reduces lateral forces induced on the subject by means of linearly translating the cable pulley according to lateral movements of the subject. To this end, a pelvis module is designed and evaluated to precisely track and render interaction forces between the human and the robot and provide a safe but transparent environment. These modules form the basis of a new type of a robotic gait trainer.

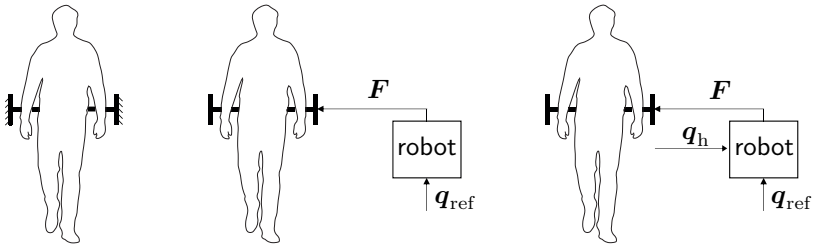


Figure 1.1: Overview of the thesis. Firstly it is investigated how blocking DoF influence the human gait (a). Secondly, the implications of a robot imposing kinematics $\boldsymbol{q}_{\mathrm{ref}}$ on the human gait is assessed (b) and lastly, modules to make the robot interact in a compliant manner and exert forces $\boldsymbol{F}$ based on human movements $\boldsymbol{q}_{\mathrm{h}}$.

Chapter 2

Effects of Physiological Kinematics during Robot-Assisted Training

Foreword

To further gain knowledge on the mechanical requirements of a robot offering balance training, individuals post-stroke participated in a study training with two different robotic devices with different kinematics. As benchmark device, the Lokomat Pro was used and compared to the Lokomat Pro with the extension module FreeD, allowing for a combined lateral shift and rotation about the vertical axis.

Abstract

This study investigates the efficacy of the commercial robotic rehabilitation device Lokomat PRO with the FreeD module, which allows more physiological kinematics of the pelvis. Patients were assigned to training with or without the position-controlled module, and the difference in muscular activation and balance scores was investigated. No significant changes between the groups were found.

2.1 Introduction

Robot-assisted gait training is applied to rehabilitation of neurologically impaired persons who have had a stroke or a spinal cord injury [44, 59, 111]. The robot be used to provide extra support, particularly for gait phases where impairments could prevent unaided walking. The support of driven-gait devices can be advantageous, given the physical demands of therapist-assisted training; in fact, these can limit the duration of therapy. Furthermore, walking may even be impossible due to spasticity [15]. Robot-driven gait orthoses can enable longer duration training and produce a more consistent locomotive pattern than could be realized with manual therapist assistance [85, 84]. Moreover, the robotic instrumentation facilitates accurate assessments of key indices such as muscle strength and spasticity [84].

Various therapeutic benefits of robot-assisted gait training have been observed. These include effects from training in the robotic gait orthosis Lokomat [46]: increases in gait speed [59] and muscle tone [59] have been shown for stroke patients, while increases in gait speed [111], endurance [111] and joint range of motion [65] have been demonstrated for spinal cord injured patients. The gait trainer LOPES [105] has shown joint range of motion improvements as well as walking speed for individuals post-stroke [91]; walking in the Gait Trainer GT I [33] has led to improvements in the independent walking abilities of post-stroke participants [31].

Nevertheless, in spite of the potential advantages of robot-driven systems, various studies have demonstrated a greater degree of improvement by means of conventional, therapist-assisted training as compared to robotic-based training for moderately to severely impaired patients [36]. This could be due to the constraints on the pelvis from the device [36], which could bring about kinematics changes during gait [37, 104]. Passivity, where there is a lack of active patient participation in the movement and too much reliance on assistance from the machine, may be detrimental to motor learning, for which active participation is very much beneficial [55]. Furthermore, passivity could reduce the effectiveness of cardiovascular training aspects [45].

This chapter investigates the influence of introducing additional pelvis kinematics on balance and gait performance of individuals

post-stroke.

2.2 Methods

2.2.1 Robotic Device

The commercial FreeD-module is an extension to the Lokomat Pro system from Hocoma. In addition to the exoskeleton employed in the Lokomat, which features actuated knee and hip joint flexion/extension in the sagittal plane, this module extends the Lokomat by one degree of freedom at the pelvis, a combined lateral shift and rotation along the vertical axis of the pelvis. This movement pattern conforms to the natural gait kinematics, as rotation and translation are coupled during gait. The combined movement is actuated by an additional electric motor.

The patient is fixed with the body weight support harness to this pelvis guidance, to comply with the lateral shift of the human pelvis, the attachments of the cuffs are mounted on linear guides. The most distal cuff located just above the ankle is fixed, ensuring a constant foot placement of the patient. The main features of the device are shown in Fig. 2.1.

A study comparing the Lokomat with and without this extension module is conducted to assess the influence of the additional DoF on the user. Except for the mechanical differences of the devices, the various settings such as guidance force and gait trajectory are maintained to the same values with and without the extension for each subject.

2.2.2 Outcome Measures

The goal of this study is to assess the comfort perceived by the user and to find surrogate markers indicating that the user is actively performing the dynamic balancing task. The comfort is evaluated by questionnaires and measuring the relative movements between the robot and the patient. High relative movements lead to high undesired interaction forces induced to the patient's body, which is perceived as discomfort.

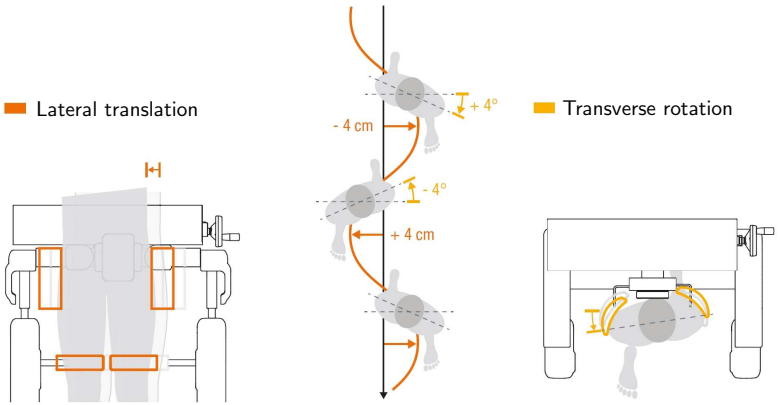


Figure 2.1: Adjustable lateral translation of up to 4 cm (left) and transverse rotation of the pelvis of up to 4° (right) to each side during walking (middle). Image with courtesy of Hocoma AG.

Quantification of the active participation in lateral weight shift is evaluated with EMG measurements at the patient's legs (Gluteus medius, Gastrocnemius, vastus lateralis, biceps femoris) and the kinematic data provided by the robotic system. The muscle activation during the assessment session (session 3) for both devices was recorded at both legs for the following muscles: Gastrocnemius, Biceps Femoris, Vastus Medialis, Gluteus Medius. The electrodes were attached to the patients prior to training with device A and not removed before the end of the training with device B. The recorded signal is rectified and pass band filtered (3Hz-20Hz). The data is then split into single gait cycles to calculate the mean and 95% confidence interval.

The pelvis and upper-body movements of the patients were recorded by an optical tracking system recording marker clusters attached to the patient's body. With additional data from the robotic devices, the relative movement between the patient and the robot was calculated. This relative motion represents the compliant behavior of both the pelvic attachment system and the human tissue. Large relative motions are considered as undesired and lead to unwanted interaction

forces, which may interfere with the natural gait pattern.

Outcome measures will be a significantly higher rated patient comfort and or patient satisfaction using the new version, compared with the established Lokomat. Furthermore, muscle activity and kinematic movement patterns are expected to show a greater number of participating muscle groups and a more natural walking pattern, involving more degrees of freedom during application of the new version compared with the established Lokomat. Exploratory endpoints will be a significantly stronger involvement of muscles needed for balance, more physiological walking patterns and more change on the lateral positioning of the feet due to training with the new version, compared with the established Lokomat.

2.2.3 Participants

The clinical study was conducted in the rehabilitation and research center Cereneo in Vitznau. Ten stroke survivors (one drop out) and nine healthy subjects participated (Table 2.1). All patients were in the sub-acute or chronic phase of the stroke recovery process.

Table 2.1: Basic Patient Characteristics

Number	9
sex (m/f)	6/3
age (SD)	57.9 (11.4)
years since stroke (SD)	2.8 (3.0)
MMST (SD)	28.8 (2.5)
NIH (SD)	3.3 (1.9)

2.2.4 Study Design

The participants were asked to take part in three sessions, a training session with device A, a training session with device B and an assessment session with both devices on different days (2.2)). The order of training with device A and device B was randomized. Questionnaires were filled out in every session to assess the comfort of the patient

and the general condition. Before and after sessions 1 and 2, a 6 minute walking test and a balance assessment was conducted. The balance assessment was performed with the Balance Master (Natus Medical Inc. USA) recording the step width of the patient. The case report form can be found in Appendix A.

Table 2.2: Study Design Overview

Session 1	Session 2	Session 3
6min WT	6min WT	Training Device A
Balance Test	Balance Test	Questionnaire
Training Device A	Training Device B	Training Device B
Questionnaire	Questionnaire	Questionnaire
Balance Test	Balance Test	
6min WT	6min WT	

2.3 Results

2.3.1 Questionnaire

The patients were asked to answer the questionnaire twice during the recording therapy session: after training with device A and after training with device B. The questions were partitioned into four groups: Effort to hold balance, interaction with the therapist, comfort and motivation during the training. Overall, no differences between the two devices were found, which shows that the new device was accepted by the patients as well as the already established device. The results are shown in Fig. 2.2.

2.3.2 Balance performance

The balance ability of the patients before and after the training sessions was quantified by measuring the step width on a force measurement plate. The step width acts as a strong indicator for the balance

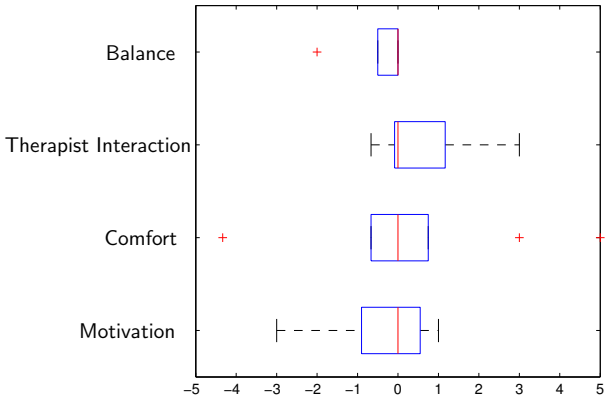


Figure 2.2: Results of the questionnaire. Patients could choose from 1-10 and results are shown as differences between the two devices. Positive values rate the new module as "better".

ability [16, 22]. No significant difference was found in balance performance before and after the training sessions with the different devices (Fig. 2.3)

2.3.3 Pelvis and upper body movements

An increased variability over a large number of gait cycles in the pelvic and upper body movements was observed (Fig. 2.4). For the upper body, there is a trend towards lower variability with the module (Fig. 2.5). The variability of pelvic motions while training with the module was significantly reduced (Fig. 2.6).

2.3.4 6 min Walking Test

The 6 minute Walking Test was conducted prior and after the Sessions 1 and 2. The distance covered was assessed and the results are normalized with the performance data prior to the training. The results are shown in 2.7 and no significant differences were found.

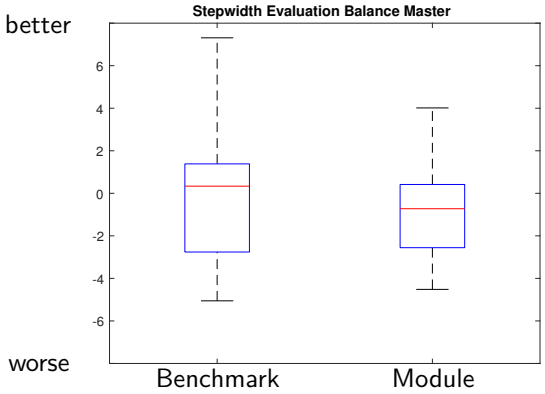


Figure 2.3: Results of the balance assessment with the Balance Master.

2.3.5 Electromyography

The activation patterns of the patients do not differ in the different devices, for both the devices the patterns resemble what we expect for treadmill walking. In Fig. 2.8 the results for one patient is shown exemplary.

2.4 Discussion

Finding surrogate markers for an improved balance training in this study with only three sessions was very ambitious. This hypotheses could not be verified, but the results can be used for further improvements in designing lower-limb rehabilitation robots.

Evaluation of the questionnaires shows that patients feel comfortable and safe with a guided lateral shift of the pelvis, suggesting the feasibility of including this degree of freedom in a robotic device. The reduced variance in the pelvis and upper body movements indicates that the unwanted interaction between the user and the robot is reduced with the FreeD module.

The functional assessments (balance test and 6 min WT) showed

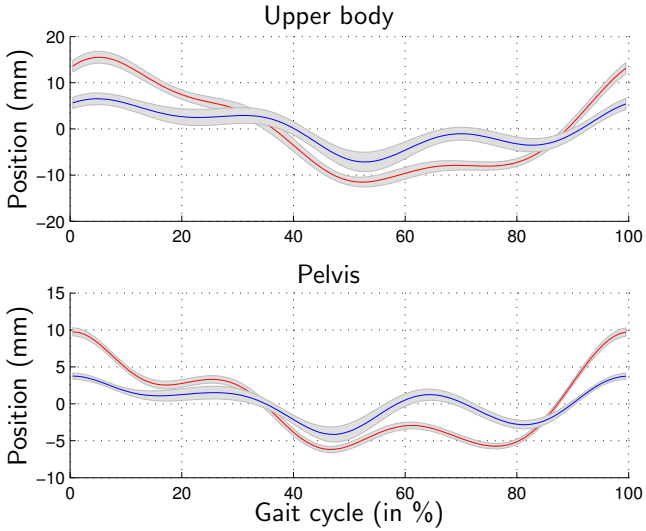


Figure 2.4: Lateral movements of the upper body and pelvis over the gait cycle from one representative individual post-stroke. The blue line represents the motion recorded from the benchmark device, the red line from the training with the module. Areas in grey show one standard deviation.

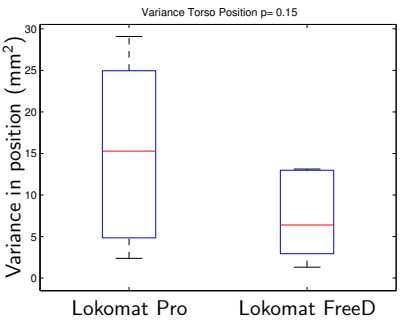


Figure 2.5: Variance in torso movements

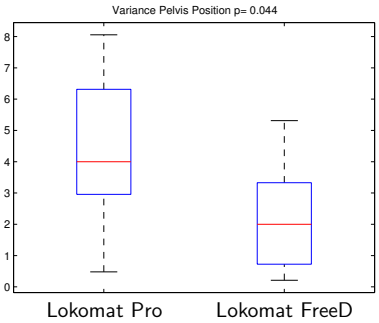


Figure 2.6: Variance in pelvic movements

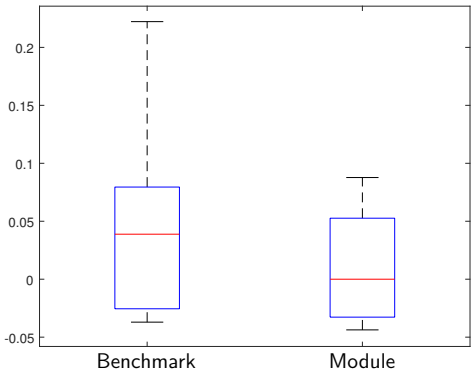


Figure 2.7: Normalized distance covered during 6 minutes as difference prior to after the training sessions.

no difference, which could be due to only three training sessions being carried out. Muscle activation patterns during walking differed largely between the patients and showed a large variability with the module. The results of this study concerning muscle activation are also supported by a current publication, in which the authors compared three different conditions of training with the Lokomat [4]. One condition featured the training with the presented FreeD module; the study found that walking with the module led to unphysiological muscle activation in the tested sample of 15 adolescents. It is striking that even while the kinematics resemble physiological gait more closely, muscle activation patterns do not seem to become more physiological, implying further developments of robotic gait trainers should include even more than one additional DoF at the pelvis.

2.5 Conclusion

The comfort and efficacy of the Lokomat FreeD module has been evaluated in a clinical study in the rehabilitation and research center Cereneo in Vitznau with patients in a comparison to the Lokomat

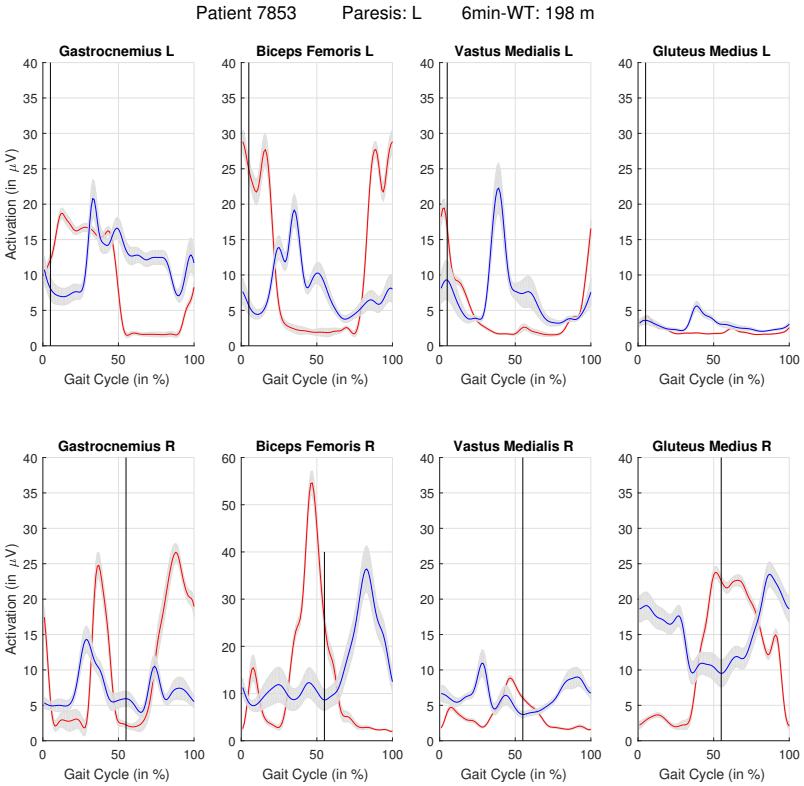


Figure 2.8: Exemplary EMG activation patterns over 300 gait cycles in one patient. Red lines represent muscle activation while walking with the FreeD module, blue lines without the module.

Pro V6 by Hocoma as a benchmark. The results suggest that the Lokomat Pro with the module providing more DoF was well accepted by the individuals post-stroke, but is not necessarily better than the benchmark device.

The recorded muscle activity (surface EMG) shows an activation pattern very similar to unassisted treadmill training, but no significant differences could be shown so far for the different devices. The variability of the pelvic movements is significantly reduced, indicat-

ing a more physiological movement of the pelvis and upper body. No influence was found on the balance task and on the performance of the 6 min walking test, both assessed before and after training with both devices.

The results imply that simply providing more physiological kinematics is not sufficient to enable balance training. The interaction forces between the human and the robot remain undetermined and require further investigation.

Chapter 3

Effects of Added Inertia and Body Weight Support on Lateral Balance Control During Walking

Foreword

This chapter documents a study on the effects on added inertia on gait in healthy subjects to create a basis for further development of modules to enable balance training within a robotic device. The findings guide the design process on the question which degrees of freedom are essential so as not to interfere with physiological movement of the human pelvis.

Abstract

A robot-driven gait orthosis which allows balance training during gait would further enhance the capabilities of robotic treadmill training in gait rehabilitation. In this paper, additional mass is attached to walking able-bodied subjects to simulate the effects of additional inertia and body weight support on the lateral balance task. The combination of additional inertia and body weight support led to reduced step widths, suggesting a stabilizing effect which may reduce the challenge of the lateral balance task.

3.1 Introduction

In the previous chapter, the influence of pelvis kinematics during position-controlled Lokomat training has been investigated, and it was concluded that more physiological kinematics alone are likely not sufficient to improve therapeutic outcomes, particularly regarding balance. Therefore, this chapter investigates a different factor, namely the influence of supporting forces in the frontal plane.

Maintaining balance in the frontal plane represents, together with providing propulsion and support against gravity, one of the major tasks involved in walking. During gait, people must actively control balance in the frontal plane [68]. Conversely, balance in the sagittal plane is thought to be passively stable and, therefore, has less of an active control role as compared to the frontal plane [8]. Lateral balance control is mostly achieved by prediction of the future position of the center of mass and then adjustment of subsequent foot placement [93, 110]. Movements of the center of mass during walking can be interpreted as possessing step-to-step stability; this is in contrast to the scenario of quiet standing, which is continuously stable [68].

Some robotic devices such as the Lokomat restrict movement to the sagittal plane only [35]. This can make weight shifting from one leg to the other difficult [37], and as a result, prevent active training of balance in the frontal plane. Hence, it has recently been proposed that robotic rehabilitation technology be enhanced by incorporating balance training into gait therapy with the rehabilitation robot Lokomat by means of adding of further degrees of freedom for the pelvis

and legs. This would have the additional effect of permitting a more natural gait pattern.

In investigating the control of balance in the frontal plane, it is useful to measure step width since this is recognized as a key index of lateral balance. Very narrow or wide steps are metabolically costly, and there is hence a preferred step width for human walking at which the metabolic cost is minimized [20]. Furthermore, results have suggested that external factors which artificially stabilize the body lead to a corresponding reduction in step width [16, 22].

Body weight support systems and robotic actuators are common features of robotic gait technology. They are required not only for helping the user with weight bearing but also for propulsion throughout the gait cycle. However, these features could also lead to a human-robot system with quite different dynamics as would be observed in the walking human alone. The hardware is likely to impose additional inertia, friction and weight, and these factors - inertia in particular - can only be partially compensated for by control algorithms [14, 94]. As well as providing vertical support against gravity, the body weight support system could provide additional support in other directions, thereby reducing the challenge and effort needed on the part of the subject for balance and postural control during walking.

Using measurements of step width, the work presented here investigates whether the lateral balance task is affected by additional inertia and also by horizontal forces from the body weight support system.

3.2 Methods

3.2.1 Mathematical Modeling

The overall system can be idealized as an inverted pendulum (Fig. 3.1). This concept has formed the basis of conditions determining necessary step widths for stability [39]. It has also provided the foundation for more complex modeling and simulation of lateral balance during walking [53].

Let I denote the inertia of the subject alone, and ΔI be the added inertia, which can be approximated as mL^2, where m is the addi-

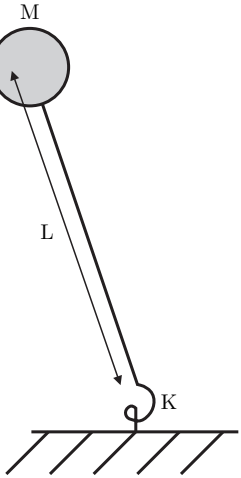


Figure 3.1: Inverted pendulum idealization of lateral balance during walking with lateral body weight support effects modeled as a torsional spring of stiffness K.

tional mass and L the pendulum length. Linearizing about the vertical equilibrium position and assuming small angles, allows equation (3.1) to be written, which is the differential equation of the system's dynamics.

$$(I + \Delta I)\ddot{\theta} - MgL\sin\theta + K\theta = 0 \tag{3.1}$$

Furthermore, an additional moment is to be provided by the subject in order to make the system marginally stable (such that it behaves as an undamped system), and this moment is equal to $K_s\theta$. The equation representing the dynamics becomes

$$(I + \Delta I)\ddot{\theta} - MgL\sin\theta + K\theta + K_s\theta = 0 \tag{3.2}$$

and therefore the poles of the corresponding Laplace transform are

$$s = \pm\sqrt{\frac{MgL - K - K_s}{(I + \Delta I)}}. \tag{3.3}$$

The effects of the additional inertia and stiffness can be seen from the system's poles. By definition, the poles must be on the imaginary axis for the system to be marginally stable and therefore we have

$$\frac{MgL - K - K_s}{(I + \Delta I)} \leq 0 \tag{3.4}$$

giving

$$MgL - K - K_s \leq 0. \tag{3.5}$$

Equation (3.5) implies that the greater the value of K, the less the minimum value of K_s that is required to achieve marginal stability. Therefore, greater stiffness via the body weight support system leads to the subject having to exert a smaller moment to maintain stability for a given perturbation. Conversely, the added inertia ΔI has no effect on the minimum value of K_s needed for marginal stability and consequently on the effort needed by the subject to maintain stability. This simple analysis suggests that the body weight support system has a stabilizing effect on the lateral balance task, while the added inertia is neutral in that it does not influence the degree of active stabilization required from the subject.

$$I\ddot{\theta} - K\theta - Mgl\sin\theta = 0 \tag{3.6}$$

Assuming that the tension is equal to the weight of the additional mass and is thus simply mg, the moment - the cross product between moment arm and force - acting on the pendulum is $mgd(\theta + \beta)$, and thus the stiffness (the ratio between force and angle) of the equivalent spring is

$$K = \frac{mgd(\theta + \beta)}{\theta} = 5mgd. \tag{3.7}$$

The nominal values of $M = 80\text{kg}$, $I = 0\text{Nms}^2$, $L = 1\text{m}$, and $d = 2\text{m}$ are used along with equation (3.3) to assess the relative effects of inertia and stiffness on the poles of the system.

3.2.2 Experimental Procedure

Five able-bodied subjects whose characteristics are shown in Table 3.1 walked on a Mercury treadmill (h/p cosmos, Nussdorf-Traunstein,

Germany). This treadmill is equipped with eight force sensors embedded in two individual plates (at the front and back of the treadmill), allowing computation of the centre of pressure (CoP) of each step.

Table 3.1: Subject characteristics.

Subject	Age	Gender	Mass (kg)	Height (cm)	Leg (cm)
A	26	M	76	190	98
B	26	F	54	164	88
C	30	M	72	186	97
D	26	F	47	164	86
E	29	M	64	181	94

Subjects walked with additional mass attached to their waists using diving weights. Walking was performed at three different mass conditions of 0 kg, 14kg and 28kg. 28kg was found to be the maximum mass that could be securely fastened to all subjects using the diving weight equipment.

Furthermore, two different treadmill speeds of 3km/h and 5km/h were used. The lateral position of the centre of mass could be approximated as a sinusoidal function of the form $a\sin(\omega t + p)$ where a is the amplitude and p the phase, and so the corresponding acceleration would be equal to $-a\omega^2\sin(\omega t + p)$. Therefore, peak inertial forces would be expected to vary with the squared frequency ω^2 and consequently, if added inertia were influential in the lateral stability task, an effect of step cadence on the step width should be observed. Therefore, different walking speeds were incorporated into the tests to permit such an effect.

The six walking conditions for each subject are summarized in Table 3.2. The order of these conditions was randomized for each person.

The static weight of the additional mass was compensated for using the body weight support system Levi (Hocoma AG, Volketswil, Switzerland), so that the overall effect was an increase in inertia, but not static weight, of the walking subject. The experimental setup is illustrated in Fig. 3.2.

The body weight support system will tend to produce a horizon-

Table 3.2: Different conditions of additional mass and walking speeds used in the tests.

Condition	Added Mass (kg)	Walking Speed (km/h)
C1	0	3
C2	14	3
C3	28	3
C4	0	5
C5	14	5
C6	28	5

tal force, since deviations from the vertical position of the cable will produce a lateral force component towards the centre line, as shown in Fig. 3.3. With the force acting in opposite direction to the subject's lateral movement, the body weight support system behaves as a spring which, due to its stabilizing effect, provides assistance in the lateral balance task.

3.2.3 Data Processing and Statistical Analysis

Data from the front plate of the treadmill were used to calculate the CoP for the first portion of each step. The average lateral position of the CoP was calculated, and differences in the lateral positions between left and right steps were then used to calculate a vector of step widths. This subsequently allowed the overall mean step width, and also the mean step cadence to be computed for that condition. The process of calculating the step widths is illustrated in Fig. 3.4.

The effects of added mass and walking velocity on step width were tested for using a 2-way ANOVA. The data was checked for Gaussianity by visual inspection. It should be noted that the mean step widths of each subject were normalized using the leg length of that subject. Moreover, the effects of speed and loading on step cadence were also tested using a 2-way ANOVA. The significance level was set at 5%.

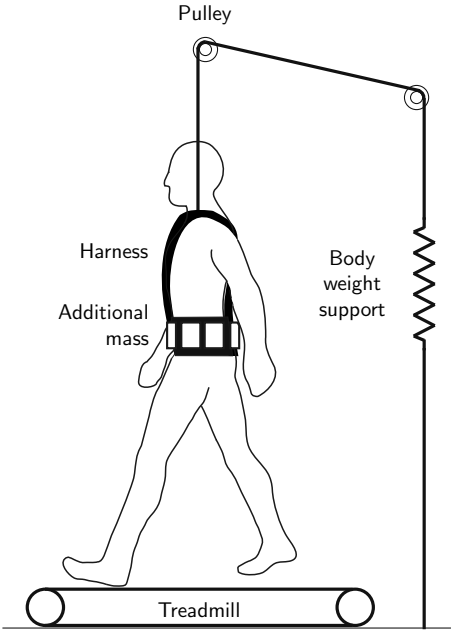


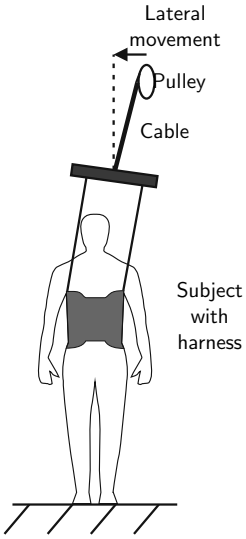
Figure 3.2: Experimental setup showing treadmill, body weight support system and additional mass attached to the subject.

3.3 Results

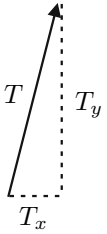
Mean step width (across subjects and walking speeds) was 20.5% lower with a load of 28 kg as compared with no load. Fig. 3.5 shows an example of the mean trajectory of the centre of pressure for two loading conditions. It can be seen that the centre of pressure for the heavier loading case lies medial to that of the unloaded scenario, giving a correspondingly smaller mean step width.

The ANOVA on cadence showed that there was a significant effect of treadmill speed ($p \ll 0.01$) but not of the loading condition ($p = 0.87$) on the step cadence. Fig. 3.6 shows the step cadences at different loading and treadmill speed conditions in box plot form.

Box plots for the step width data are given in Fig. 3.7.



(a) Subject's lateral movement induces an angle in the cable.



(b) Lateral component of tension.

Figure 3.3: Lateral subject movement induces an angle in the body weight support cable, causing a stabilizing force to be applied to the subject in the lateral direction.

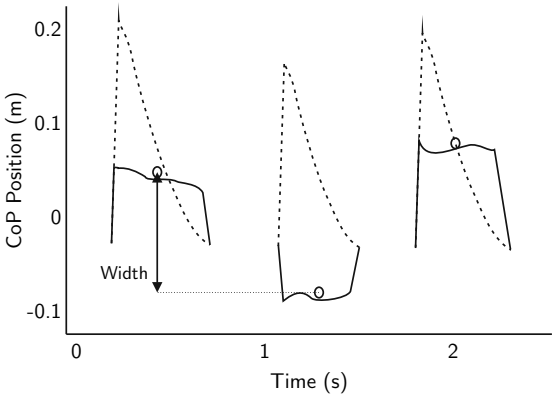


Figure 3.4: Peaks in the longitudinal position of the CoP (dashed) indicate the beginning of each step, allowing averaged lateral positions (open circles) to be calculated from the lateral CoP position (solid lines), giving the step widths.

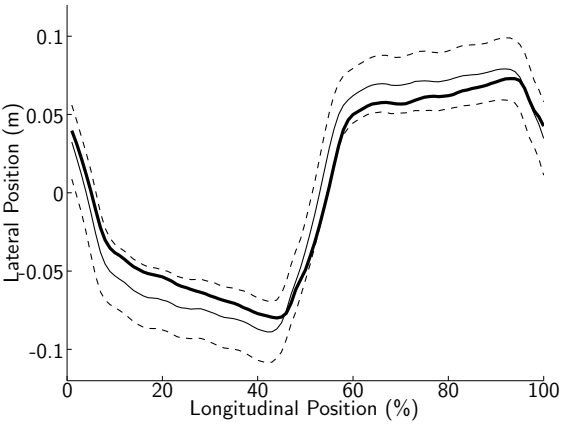
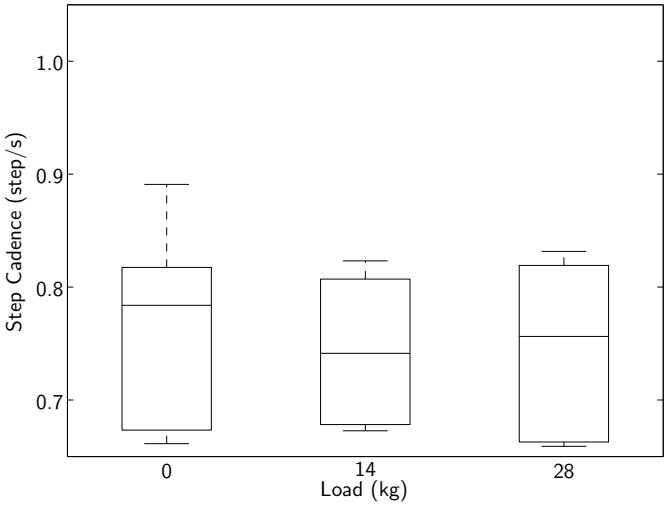
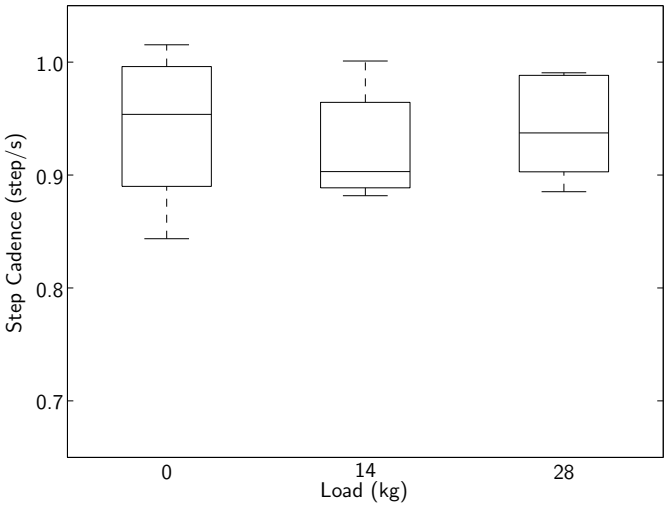


Figure 3.5: The mean positions of the centres of pressure are depicted by the solid and lines for subject E at 3 km/h. The thin solid line is used for the 0 kg case and the thick solid line for the 28 kg case. The dashed lines represent standard deviations in the lateral position in the 0kg case.

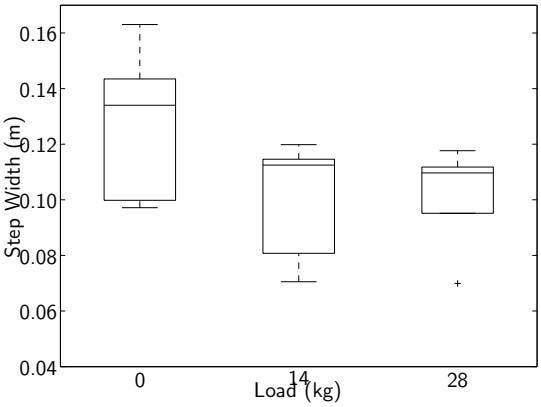


(a) 3 km/h

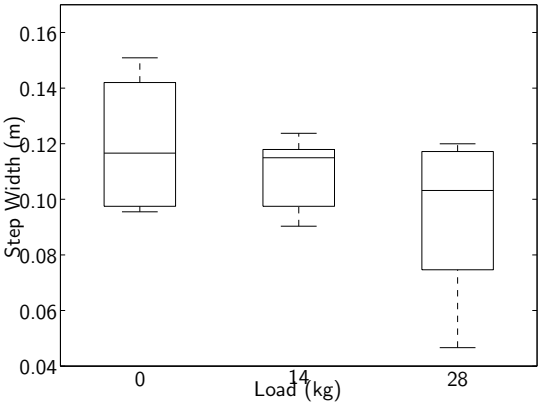


(b) 5 km/h

Figure 3.6: Step cadences at different loading and speed conditions.



(a) 3 km/h



(b) 5 km/h

Figure 3.7: Step widths at different loading and speed conditions.

The ANOVA on step width indicated a significant effect of loading condition on the (normalized) step widths ($p = 0.015$) but not of walking speed ($p = 0.781$), and no significant interaction between the two factors ($p = 0.493$). In general, subjects took narrower steps when walking with higher loads (and thus at higher levels of body weight support).

3.4 Discussion

The decrease in step width with increasing load suggests that the combination of additional inertia and body weight support has a stabilizing effect on the body, reducing the required level of active lateral stabilization via foot placement. The lack of influence of walking speed on step width is consistent with other work that found no correlation between speed and lateral measures of stability including step width [86].

The results show that the additional mass, compensated for using the body weight support system, has two effects: one of additional inertia and a spring-like (stiffness) effect from the lateral forces induced in the cable of the body weight support system. Results from other investigations that employed external lateral stabilization via a stiffness approach have also demonstrated a reduction in step width [16, 22], albeit using much greater levels of stiffness than in the current study.

Inertial forces are expected to change with the square of the frequency of lateral movement, yet different cadences of walking did not influence the resulting step widths taken by the subjects. This implies that the effect was mostly due to the lateral stabilization from the body weight support system.

Models of balance using an inverted pendulum have been proposed and have formed the basis of conditions determining necessary step widths for stability, using a prescribed initial lateral velocity as an input [39]. Models using a constant initial velocity would predict a greater required step width for a larger inertia due to the greater initial angular momentum. However, this pre-step velocity is likely to itself depend on the system's inertia. Therefore, more complex models are required to study the influence of inertia on step width.

It may be necessary to include the kinematics of the pelvis and coupling between sagittal plane propulsion and frontal plane balance. Furthermore, energy expenditure is an important factor in the control of step width during walking [21], and is likely to be required in mathematical modelling of lateral balance.

Although not available in the current study, measurements of the lateral support force could be used to accurately determine the relative contributions of body weight support and additional inertia on balance control. Alternatively, the effect of inertia alone could be investigated using a system able to maintain the body weight support cable at a consistently vertical orientation, preventing any lateral forces from being developed. Such an investigation necessitates a control system of relatively high bandwidth.

Robotic gait therapy will typically have both additional inertia and stabilization from the body weight support system. The stabilizing effect could be much greater in actual clinical application than in this study since the magnitude of the moment from the person's weight is also reduced. Furthermore, a substantial portion of a person's body weight can be needed to allow patients with a high level of impairment to perform stepping in the devices. Stabilization effects are also seen in the sagittal plane, where it has been observed that patients use the body weight support system and harness for support in anterior tilt by leaning forwards. Although such additional stabilization may be beneficial in the early stages of rehabilitation where a large degree of assistance is required, in later stages the additional support is likely to reduce the amount of effort required of a subject in postural and balance tasks in both frontal and sagittal planes.

3.5 Conclusion

The combination of additional inertia and the body weight support system significantly decreased the step widths used by the subjects to maintain balance in the frontal plane. The reduction in step width arose mainly due to the stabilizing lateral forces developed in the body weight support cable. Since this could significantly reduce the challenge of maintaining lateral balance during walking, body weight support mechanisms in which horizontal force components are prevented may be useful in robot-driven gait training incorporating bal-

ance and postural control.

Chapter 4

Body Weight Support System Extension

Foreword

The results of the study on added inertia (chapter 3) suggest that a body weight support system inducing purely vertical forces on the user is a requirement for a robot allowing balance training. This finding motivates the development of a lateral extension for body weight support systems, which is documented in this chapter.

Abstract

Body weight support systems are frequently used as part of robotic gait training to provide unloading in order to help subjects perform walking, but can also induce stabilizing forces and render the task of maintaining balance less challenging. In this paper, a two-dimensional body weight support system extension is presented which reduces lateral forces induced on the subject by means of linearly translating the cable pulley according to lateral movements of the subject. It is demonstrated that the system accurately tracks lateral movements of the pelvis at different levels of vertical support load

and thereby lowers the induced lateral forces. The system will be used in advanced robotic body weight supported treadmill walking incorporating a balance training element.

4.1 Introduction

Regaining the ability to walk is a major focus of the rehabilitation of stroke survivors and patients suffering from a spinal cord injury. Being ambulatory is crucial for accomplishing activities of daily living and, therefore, strongly contributes to quality of life [27]. Body weight supported treadmill training is frequently used as part of rehabilitation for spinal cord injury patients or people with neurological disorders such as stroke. The body weight of the subject is partially unloaded according to their walking ability and strength. Positive results have been shown regarding the effectiveness of body weight supported treadmill training for stroke patients [32] and for spinal cord injured patients [18].

The training is often combined with a form of supporting forces to move the neurologically impaired leg, which can be provided manually by therapists or by powered orthoses. Robotic assisted therapy can enhance the intensity and frequency of training as it reduces the physical workload of the therapists. On the other hand, a robotic device with limited degrees of freedom restricts pelvis and trunk movements and can thus alter gait kinematics [37]. These constraints hinder lateral movement - an important element underlying the control of balance during walking [8] - and, in this way, can limit the scope of and reduce the challenge of the training [73].

Most body weight support systems (BWS) are realized with a cable system which is connected to a harness worn by the subject. The cable is guided by a passive or active weight unloading system over an overhead-mounted deflection pulley to the subject. An example of a commercially available body weight support system is the Lokolift [26]. The scope of body weight supported gait training has recently been expanded to large workspace overground walking with the development of the ZeroG (Bioness, Inc., Valencia, CA, U.S.), which follows the subject in the walking direction by means of a trolley that runs on a rail and contains a pulley mechanism with a series

elastic actuator to unload the subject [34], and with the development of the FLOAT (Lutz Medical Engineering, Switzerland), which allows transparent three-dimensional support during overground gait, by means of four actuators and moving deflection units on two rails [95].

Preliminary experiments [74] indicate that most of the currently available body weight support systems with a fixed pulley system impose lateral forces on the subject. These forces tend to pull the subject back towards the center line as shown in Fig. 4.1 and thus cause a stabilizing effect, which may reduce the challenge of the dynamic balance control task.

This paper presents an extension to body weight support systems incorporating an additional degree of freedom (DoF) to minimize these lateral forces. The device hardware and validation of its performance is shown. Finally, a comparison between a conventional, fixed BWS system and one equipped with the new extension is made.

4.2 Methods

4.2.1 Concept and Model

The main application of the new BWS will be in the context of the new gait rehabilitation device. To clarify the interaction between the two components, the pelvis module (see Chapter 5) is simplified as a linear actuator to support the lateral movement of the pelvis, as depicted in Fig. 4.2.

The main function of the body weight support extension is to laterally translate the main pulley according to pelvic movements using a linear actuator[1]. However, a lateral deviation of the pulley would alter the cable length of the main BWS, which would induce large forces on the pulley.

Therefore, a design is proposed that cancels out the static forces and compensates for changes in cable length. The actuator force needed is small as it only compensates for friction effects and inertia. A system

[1]Two linear actuators are used: the first mimics the pelvis module for the experiments of this study (Fig. 4.2, 4.6 and 4.7) while the second is a component of the presented 2D BWS (Fig. 4.3 and 4.4).

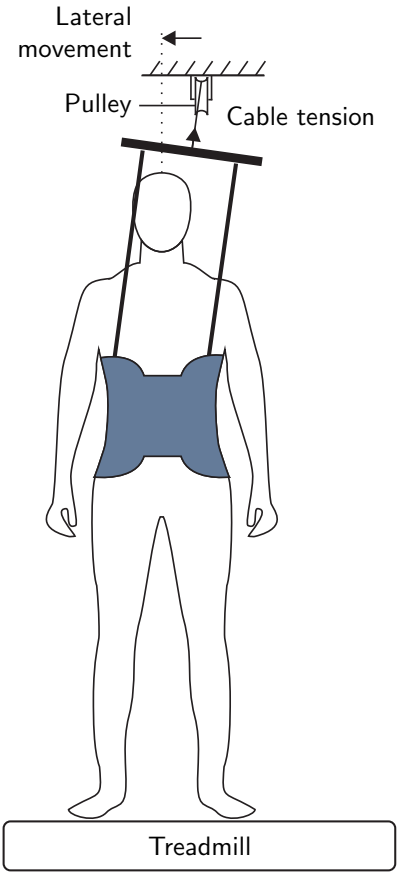


Figure 4.1: Principle problem with missing lateral DoF.

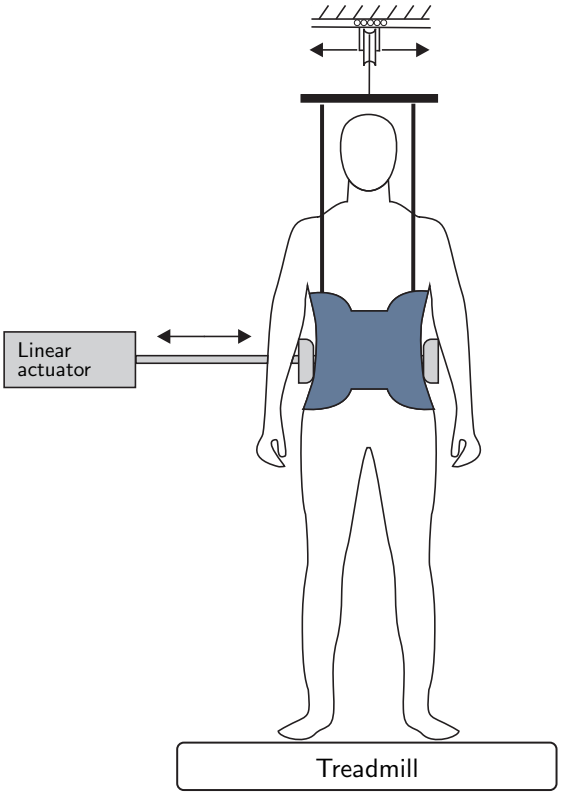


Figure 4.2: The linear actuator supports the natural horizontal weight-shifting movement of the pelvis and emulates the pelvis module of the complete platform.

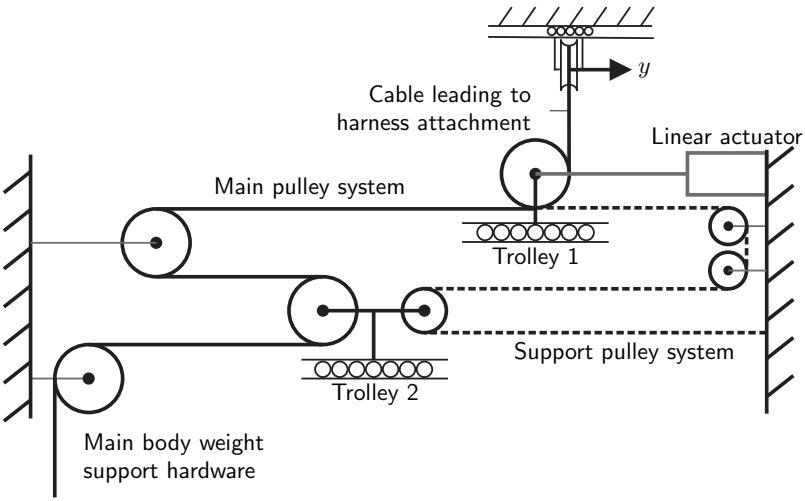


Figure 4.3: Schematic of BWS extension.

of pulleys (Fig. 4.3) is used to ensure that changes in the overall cable length in response to lateral movements of the main pulley are kept to a minimum. Further details on the concept, including a mathematical model and a simulation study of the system, are presented in [72].

In comparable applications such as cranes with trolleys, another approach is often used, namely a free hanging pulley. The concept presented here was chosen because uncontrollable degrees of freedom should be avoided. Another advantage is the compatibility with existing body weight support systems, as the extension does not alter the forces of the main support system. In addition, it will be possible to use the system to command nonzero lateral forces. This could be useful to render subject-specific balance training environments (e.g. stabilizing or destabilizing).

4.2.2 Hardware and Sensor Implementation

The components of the new BWS extension - illustrated in Fig. 4.4 - are mounted on a plate that is rigidly connected to the frame of the BWS system. The pulleys are manufactured from polyamid and are

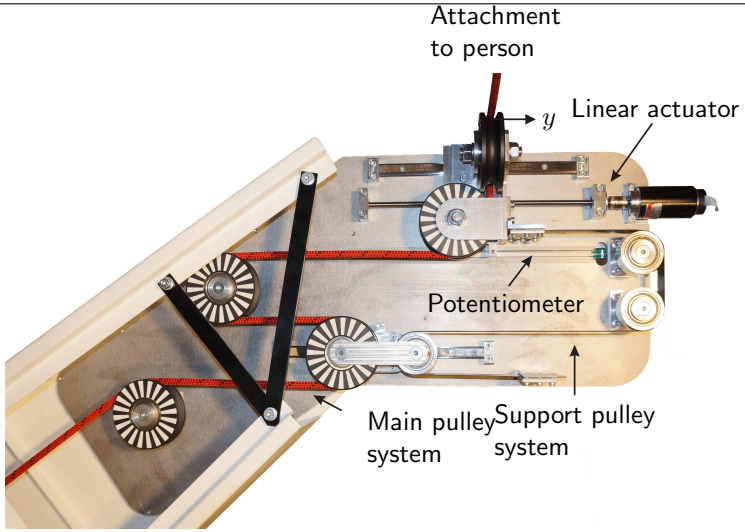


Figure 4.4: The mechanical implementation of the pulley system.

each equipped with two low friction ball bearings. The movable pulley units are mounted on the trolley of a linear guideway (Hiwin®). The trolley of the first pulley system is actuated by a ball screw (with a lead of 2 mm) and electric motor (Maxon®RE40). The small lead was chosen in order to make an additional transmission superfluous.

Sensing of the lateral pulley position is realized by an encoder on the motor shaft and a linear potentiometer for redundancy and ease of initialization. The support pulley system is driven by a length stable belt and guided over crowned pulleys for self-centering. The maximal achievable force and speed of the linear unit are 565 N and 0.35 m/s, respectively, while the maximal lateral amplitude of the lateral pulley position is 0.1 m.

4.2.3 Control System

The lateral position of the new BWS extension, y, is position controlled using a simple proportional derivative (PD) controller, shown in Fig. 4.5; the parameters of the PD controller were tuned manually. The controller produces, via a motor torque, an actuator force, F_a,

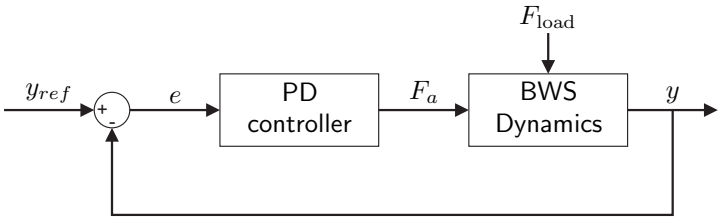


Figure 4.5: The position control system of the lateral degree of freedom of the BWS system.

based on the error, e, between the reference (y_{ref}) and actual (y) lateral displacements.

This arrangement was chosen since the pelvis module has the task of supporting and controlling the lateral movement of the human and the idea behind the control system of the BWS in standard training mode is to minimize the relative lateral position between the pelvis and BWS cable attachment point, and therefore, the unintended lateral forces which are induced by this relative displacement.

Currently, two control modes are implemented. In the first control mode, the desired position of the BWS is given by the actual measured lateral position of the pelvis module. In this way, the BWS tracks the lateral position of the pelvis module (and thus also the lateral position of the subject). This mode will be applied when the impedance of the lateral control of the pelvis module is low, giving the human subject a high degree of influence on their lateral position.

In the second control mode, the desired position for the control system of the BWS is directly given by the desired trajectory of the pelvis module. The movement of the two modules is thus better synchronized and the tracking error and thus the lateral disturbing forces should be minimized. This mode is intended for when the impedance of the lateral pelvis guidance is high; in this case, the actual trajectory of the human pelvis should be tightly controlled to a reference trajectory.

Concerning other applications, the linear correlation of the displacement/angle and the disturbing lateral force provides the possibility of actually using the BWS to induce defined disturbing or stabilizing forces on the human subject. This could be used in future

applications such as balance training in which perturbations are applied to the subject during walking to increase the challenge of maintaining balance. As a safety feature, virtual 'walls' are implemented at the end points of the lateral range of motion. These provide an additional assisting component and secure against excessive lateral movements (e.g. when a subject is stumbling and in danger of falling to one side). Other safety features are current limits for the motors, emergency stop buttons for both the subject and operator, and the mechanical end-stops of the device (Fig. 4.4) which limit the extent of lateral movement.

4.2.4 Experimental Protocol

In order to evaluate the concept and implementation of the system, two experiments were conducted to answer the following questions:

1. Does the pulley mechanism function in the intended way such that the supportive load for the subject does not affect the lateral system behavior (e.g. due to friction effects)?

2. Does the additional degree of freedom of the BWS system succeed in reducing the induced lateral forces?

Influence of Attached Mass on System Behavior

As discussed in section 4.2.1, the force from the attached mass of the person should be canceled out within the pulley mechanism and should thus not unduly influence the system's lateral behavior. To evaluate whether the friction and other effects caused by the attached load alter the mechanical behavior, the BWS was set to a sinusoidal trajectory (6 cm amplitude) and three different masses (5 kg, 25 kg and 45 kg) were attached to the cable, as shown in Fig. 4.6.

This range was chosen as it represents realistic loads for body weight supported treadmill training: 45 kg would represent 50% support for a 90 kg person. To ensure that the additional load is only applied in a vertical (negative y-direction) and that no (or very low) dynamic lateral forces are applied to the BWS, the added masses were additionally guided by a linear actuator that moved them in a lateral direction. The fixation utilized ensured that no vertical forces

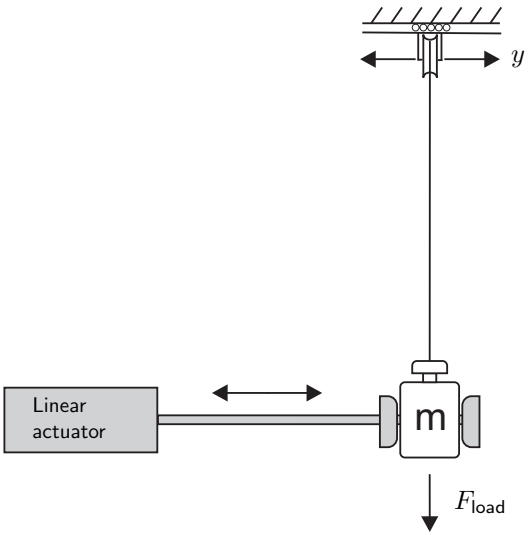


Figure 4.6: The experimental setup to evaluate the system behavior under different loading conditions.

were transferred to this pelvic lateral guidance system. The linear actuator concept is applied here in order to simulate the intended application in which all the lateral forces between the subject and the robot should be transferred to the pelvis module of the gait rehabilitation system.

As an additional indicator of whether the system behavior is altered by the different levels of supported mass, the control outputs of the position control were recorded for the three different load cases and then compared.

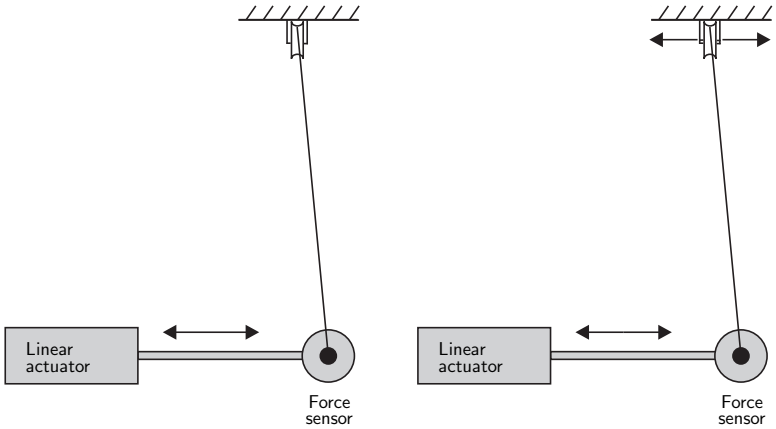
Influence of the Lateral DoF on the Disturbing Forces

In order to answer the second question - that is, whether the new lateral degree of freedom of the BWS system effectively helps to reduce the disturbing lateral reaction forces - an experiment was conducted to measure this force. Firstly, the experiment was conducted with a locked DoF, while in the latter phases, the lateral DoF was unlocked

and the control system was activated in two different modes. There were thus three different settings:

1. Lateral DoF locked;
2. Lateral DoF permitted, first control mode;
3. Lateral DoF permitted, second control mode.

The setup of the experiment is illustrated in Fig. 4.7. The cable of the BWS was attached to a 6 DoF force sensor that in turn was attached to an additional linear actuator that could move laterally (with all other DoF fixed). The vertical distance between the force sensor and the attachment point was approximately 0.75 m, a height that represents a realistic value in the setup. During the tests, the linear actuator moved the added mass in a sinusoidal pattern with an amplitude of 6 cm.



(a) 2D BWS locked (b) 2D BWS active

Figure 4.7: Experimental setup for the experiments on the disturbing forces, with the lateral DoF of the BWS blocked (left) and active (right).

The resulting forces were recorded (no torques were transmitted due to the mechanical interface to the sensor) and low pass filtered with a 20 Hz cutoff frequency.

Table 4.1: Lateral forces (RMS and amplitude) for the three different conditions.

Condition	RMS (N)	Amplitude (N)
1	5.8	8.5
2	1.25	2
3	0.71	1.5

4.3 Results

Influence of Attached Mass on System Behavior

Fig. 4.8 compares the desired and actual sinusoidal trajectories for the different loads.

Time histories of the control outputs are shown in Fig. 4.9. The root mean square force output is increased by around 16% when the attached load is increased from 5 kg to 45 kg.

Influence of the Lateral DoF on the Disturbing Forces

Fig. 4.10 shows the lateral forces that were measured under the three different settings.

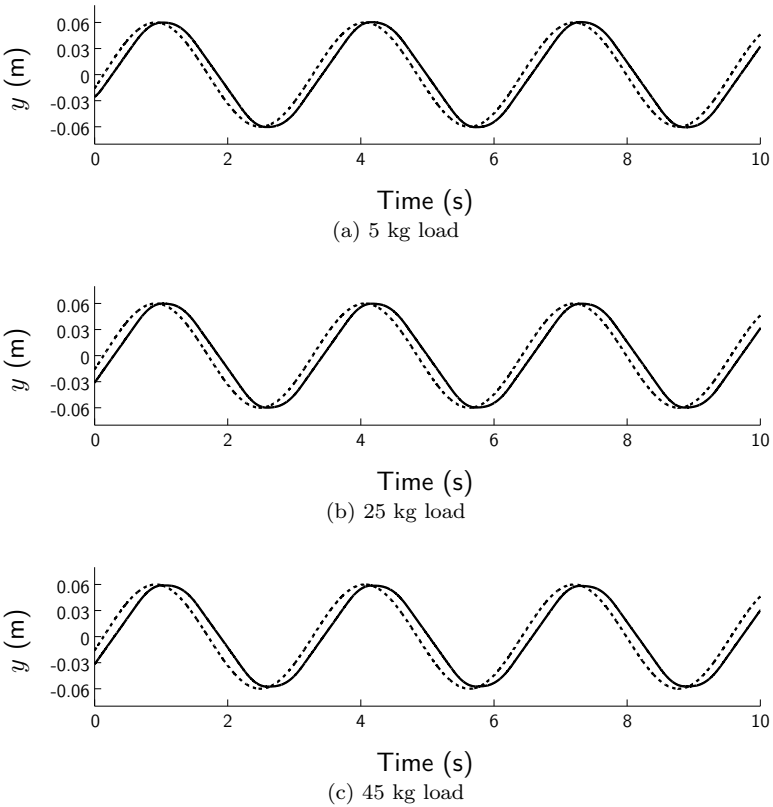
The root mean square (RMS) results are summarized in Table 4.1. The RMS values are reduced by around 80% from condition 1 to condition 2 and by another 40% from condition 2 to condition 3.

The correlation between the lateral force and the lateral displacement between the cable attachment point at the BWS and the lateral position of the force sensor is shown in Fig. 4.11.

4.4 Discussion

It can be seen that there is almost no difference in the tracking performance of the position control system for the three different loads attached to the BWS, so the high-force vertical DoF is indeed effectively decoupled from the lateral DoF.

The results demonstrate that the lateral degree of freedom with the feedback controller greatly reduces the induced lateral forces.



(a) 5 kg load

(b) 25 kg load

(c) 45 kg load

Figure 4.8: The desired sinusoidal trajectory compared to the actual recorded trajectories with three different loads (5 kg, 25 kg and 45 kg) attached to the BWS. Dotted and solid lines represent the reference and actual positions, respectively.

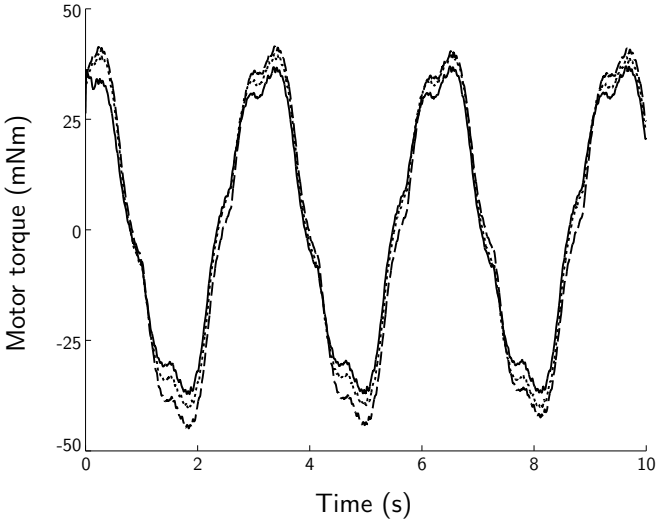


Figure 4.9: Motor torque of the control system for the three different load cases. Solid, dotted and dashed lines represent loading at 5 kg, 25 kg and 45 kg, respectively.

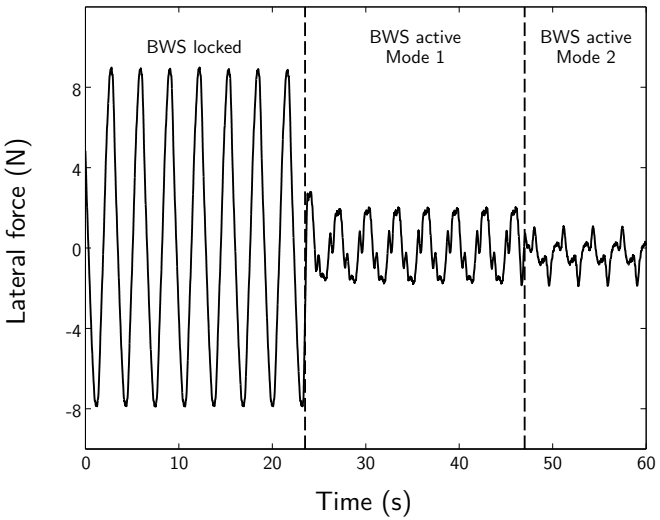


Figure 4.10: Measured lateral forces under different control settings, with the lateral DoF was blocked (left-hand section), unlocked (middle section) and position control set to the tracking mode, and the trajectory of the BWS synchronized with that of the linear actuator (right-hand section).

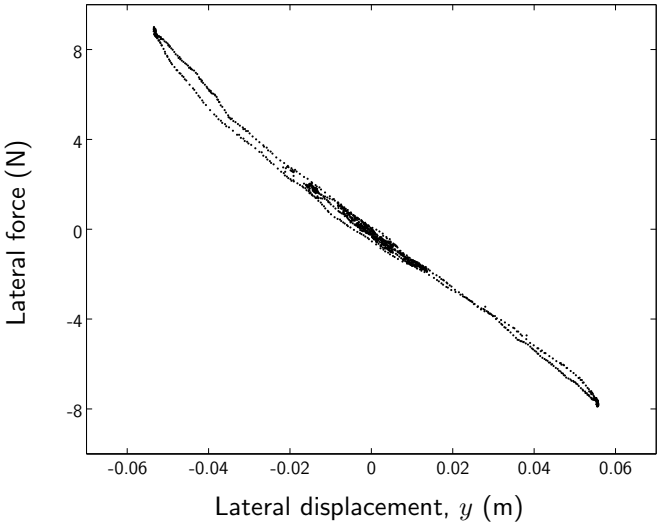


Figure 4.11: The lateral forces over the lateral displacement between the attachment point of the rope at the BWS and the force sensor.

It can also be seen that the relationship between lateral forces and lateral displacement is fairly linear over the analyzed range of motion and that the degree of hysteresis is low. This shows that in addition to the reduction in the lateral disturbance, the BWS extension presented in this paper could be used to apply stabilizing or disturbing forces, for example, in a stand-alone application without the additional modules (i.e. the robotic orthosis device) of the complete gait rehabilitation system.

Tests need to be conducted to evaluate the performance when human subjects are supported by the device and walking at different speeds. This will introduce additional challenges such as a variation in vertical load and a more complex pattern of lateral movement.

4.5 Conclusion

An extension for body weight support systems which reduces lateral stabilizing forces through tracking lateral movements of the subject's

pelvis has been designed, constructed and tested. The system can accurately follow a reference sinusoidal lateral displacement under various levels of vertical supporting force. The body weight support system extension substantially reduces the induced lateral forces, while effects on the vertical support forces are kept to a minimum. The system will be applied to advanced body weight supported treadmill training in which subjects are required to actively maintain their balance.

Chapter 5

Compliant Pelvis Module

Foreword

This chapter describes the development and evaluation of a novel pelvis module to realize the requirements of additional DoF for balance, as stated in chapter 1.

Abstract

Series Elastic Actuation decouples actuator inertia from the interaction ports and is thus advantageous for force-controlled devices. Parallel or even passive compliance can fulfill a complementary role by compensating for gravitational or periodic inertial forces or by providing passive guidance. Here, these concepts are combined in an underactuated six degree of freedom (DoF) compliant manipulator with one actuated DoF. The mechanism comprises a spring assembly in which each spring serves as an actuation element and simultaneously provides passive compliance in the unactuated DoF. The device is designed to assist weight shifting via controlled lateral forces on a human pelvis during treadmill walking and its eigenfrequencies are

tuned to align with normal gait. Six-DoF force and torque sensing are realized via a model of the spring deformation characteristics in combination with low-cost inertial and optical sensors. Experimental evaluation demonstrates that the system can effectively follow physiological lateral pelvis movement with low interaction forces and also has little impact on remaining pelvis motions.

5.1 Introduction

The traditional approach to robotics has tended to emphasize serial kinematic structures and stiff actuation which, although suitable for many position control tasks found in manufacturing, is not ideal for applications that require transparent force control, due to undesired inertial effects. Particularly in haptics, in which robots operate in direct contact with humans, control over robot-human interaction forces in six degrees of freedom (DoF) is critical for safe operation.

An important area of application for highly transparent robotic interfaces is the field of rehabilitation robotics. A robotic interface able to render low impedance can enhance active patient participation and achieve low robotic interference whilst retaining the option of more support when needed [73]. Gait training represents a particularly important area of rehabilitation for stroke survivors [54] and individuals with spinal cord injury [60]. The human pelvis moves in a six DoF pattern during normal gait [92], the kinematics of which can be represented as periodic oscillations in the range of 1-2 Hz in able-bodied individuals [7]. The pelvis translates in longitudinal, lateral and vertical directions in an approximately sinusoidal manner. The lateral direction has the greatest amplitude of around 20 mm. The orientation of the pelvis is governed by three angles [5] that influence the vertical position of the center of mass during gait [50]. In addition to control of angular momentum in the sagittal plane [67], balancing during walking is a crucial element of human gait involving weight shifting from one leg to the other in the lateral direction (i.e. the frontal plane) [56]. During walking, the center of mass constantly moves outside the base of support, and must be actively stabilized through foot placement [8]. As such, the lateral degree of freedom has an important role to play during walking and it could be

argued that it should be afforded particular attention during design of robotic devices for rehabilitation.

Restraining the pelvis adversely affects gait dynamics [104] and thereby may impede the rehabilitation process in stroke survivors [36]. Therefore, designers of modern gait rehabilitation robots generally incorporate multiple DoF for pelvis motion. For example, the LOPES II treadmill-based exoskeleton [64] features two actuated (lateral and forward/backward translation) and four passive (one vertical and three rotational) DoF for the pelvis. Another pelvis manipulator mounted on a moving cart, the KineAssist [71], supports users at the pelvis and enables unrestricted and highly transparent pelvis rotations about all three axes as well as vertical translation. Horizontal translations are enabled via the device's moving frame. Both aforementioned devices use stiff design and closed-loop force control, which can partially mitigate unwanted effects by reducing the robot's reflected inertia [40, 14]. In LOPES II, the stiff design in combination with force sensors allows rendering masses at the pelvis which are lower than the actual structural masses of the device components. For the KineAssist, given the high mass and inertia of the frame, transparency is more difficult to achieve for the translational DoF. The extent of inertia reduction for stiff robots remains limited [14], and the required force sensors tend to substantially increase cost.

For reducing mass and inertia, most dedicated pelvis actuation devices rely on parallel actuation to enable transparent interaction during the six-dimensional pelvic movement: a number of rigid or elastic members such as cables, rods, or springs are each connected to the pelvis on one end, and to a fixed frame on the other. Varying numbers of these members are actuated to change their length, for example six in the A-TPAD [47] or in PAM [2], three in the BAR [69], and one in a dedicated 1-DoF pelvis perturbator [30].

In underactuated parallel manipulators, the design-inherent coupling of DoF can be an issue. For example, to assist only in weight shifting i.e. to apply lateral forces, the standard paradigm of parallel actuation would involve one single member as in [30], or an antagonistically arranged pair of members connected to a fixed base. While this enables free movement in five of the six DoF, such a mechanism does not allow the applied forces to be perfectly lateral. Instead, force

components can occur in other directions as well as moments about all axes. This can be alleviated by connecting the members not to a fixed base, but rather to moving points that passively follow major pelvis movement, particularly anterior-posterior movement [11]. However, this further complicates the design and still does not ensure full decoupling of DoF.

Both in series and parallel manipulators, the incorporation of physical compliance in different forms is now an established design strategy to achieve high-fidelity force and impedance control [102]. The Series Elastic Actuator (SEA) is a prominent example of the use of physical compliance to improve force control. Its key feature is an elastic element in series with a stiff actuator [81]. The concept transforms force control into a position control task because the deflection of the elastic element acts as an indirect force measurement. Although the elastic element lowers the overall bandwidth and inhibits stiff actuation [98], its effect is favorable for many applications [51, 103] as it allows back-drivability and limits mechanical impedance, and thereby contributes to the safety and comfort for operation with humans [115]. While most SEAs have hitherto had a single DoF, the present authors have demonstrated two extension concepts to realize SEA manipulators with multiple DoF, both applied to gait training. The first extension was fully actuated and involved placing a stiffly actuated n-DoF robot in series with a single end-effector module that is elastic in the same n DoF and rigid in the $6 - n$ others. It was applied to a gait training robot for rats that actuated four DoF of the animal's trunk (translations and yaw) and constrained the two others (pitch and roll) to enforce upright gait [19]. Interacting with humans, it is less beneficial to constrain rotational DoF as these may interfere with natural motion. Therefore, a second extension was an underactuated SEA-based cable robot that transmits a single force vector to a human trunk by means of a harness, while not imposing kinematic rotational constraints [95].

Passive compliant elements such as springs are routinely applied to compensate for constant (e.g. gravitational) forces. Furthermore, passive compliance can allow some freedom around an equilibrium configuration without enabling excessive deviation from this point, for example, to provide guidance. With appropriate tuning, passive

compliance can also compensate for periodic inertial forces [97].

In this article, a six-DoF end-effector mechanism is proposed that achieves decoupling of actuated and unactuated DoF by unifying the concepts of passive compliance and serial elasticity: the MUltidimensional Compliant Decoupled Actuator (MUCDA). The MUCDA exhibits series elastic actuation in one constant direction only, and fully passive elasticity in the remaining five DoF. This decoupling is enabled by means of a linear actuator in combination with an assembly of multiple individual coil springs. Each spring in the assembly provides serial and passive elasticity, thus minimizing overall mass and complexity.

The MUCDA is realized and experimentally evaluated in a gait rehabilitation platform that interacts with the human pelvis in six DoF. A single actuated DoF allows the crucial task of lateral weight shifting to be supported to differing degrees according to the walking ability of the user - for more able walkers, the device can be transparent, but more guidance can be provided by the machine when required. Compliance in the remaining five DoF keeps the user in a safe range around the origin position without restriction to a fixed point in space. This combination of passive compliance and the SEA concept leads to high transparency in all six DoF, which is expected to encourage a higher degree of active participation and thereby more positive gait training outcomes.

5.2 Methods

5.2.1 Hardware Concept

The compliant interaction of a robotic device with the human pelvis represents a challenging task for gait rehabilitation robots.

The system, shown in Fig. 5.2, is designed such that it actively guides the human pelvis to support lateral weight shift during treadmill walking, while providing only minimal passive support to the remaining DoF of the pelvis. The aim is thus to support and guide the human pelvis without completely restraining it. The support should hold the user approximately in position without interfering with physiological movements, including arm swing.

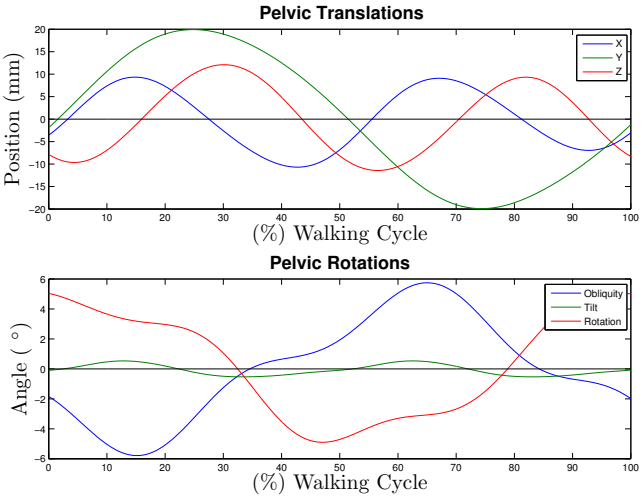


Figure 5.1: Pelvic motions during gait. Obliquity is the rotation about the sagittal axis, tilt the rotation about transverse axis and transverse rotation comprises rotation about the vertical axis. All rotations are shown as projections on the planes of movement.

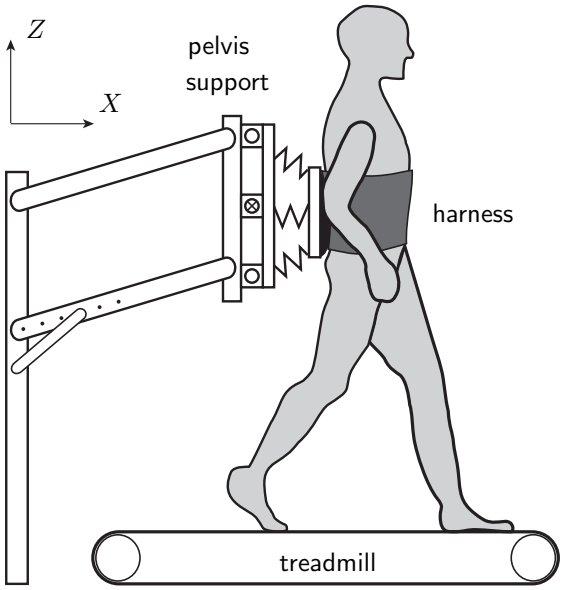


Figure 5.2: System overview

The MUCDA concept is applied to fulfill these requirements. A single motor is used to support lateral weight shifting and produces a configuration with five passive and one actuated DoF. The MUCDA pelvis module comprises two functional groups - an actuated element and a passive component - as summarized in Fig. 5.3. An actuator and linear guides are placed between a fixed and a middle plate in order to translate the latter component. This lateral translation is used to support the lateral movements of the user. The user is mechanically connected to the pelvis plate, which in turn is attached to the middle plate via a multi-DoF compliant module, which is an assembly of linear springs.

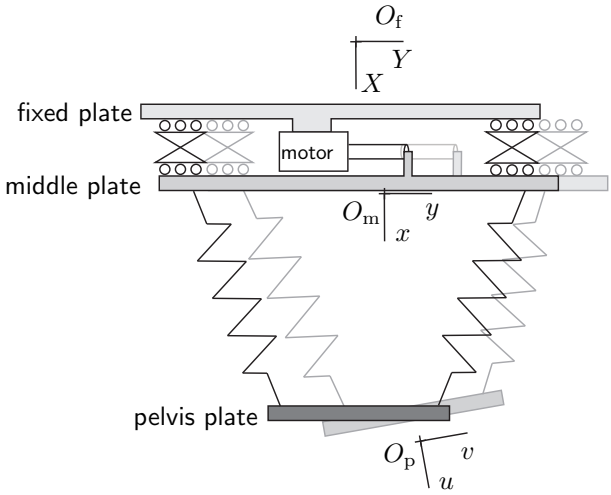


Figure 5.3: Pelvis support module comprising: 1. the actuated module, where the middle plate translates in Y-direction relative to the fixed plate by linear guides and an actuator; 2. a passive module in which the pelvis plate follows the user's movements in six DoF and is connected by compliant components to the middle plate. Coordinate systems corresponding to the fixed, middle and pelvis plates are shown as O_f, O_m and O_p, respectively

This spring assembly enables relative movements of the two plates

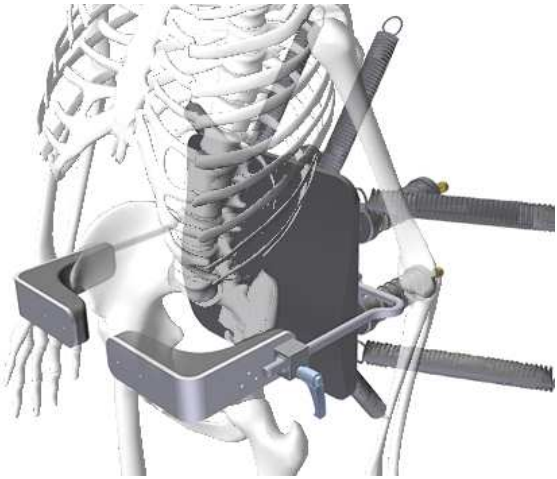


Figure 5.4: User fixation on the pelvis module

with respect to each other. The elastic behavior of this connection - governed by the stiffness matrix - defines how users are influenced when moving their pelvis in all five unactuated DoF. In the lateral direction, the linear motor and the elasticity in this DoF together form a series elastic actuator.

The system comprises a frame connected to the environment via linear guides movable in the y-direction. The pelvis plate is connected to the linear actuator using the spring assembly depicted in figure 5.5. The lateral DoF is actuated by a direct-drive electromagnetic linear motor (P01-48x240 from NTI AG LinMot, Spreitenbach, Switzerland). This actuator can produce forces and velocities of up to 585 N and 1.7 $^{\mathrm{m}}/_{\mathrm{s}}$, respectively. This installed peak force is almost six times the amount expected for normal operation to enable stiff guidance near the end of the allowed workspace. An image of the pelvis module - showing the positions of the motor and linear guides as well as the fixed, middle and pelvis plates - is provided in Fig. 5.5.

Since the compliant behavior is realized through the pelvis module, the mechanical connection to the human user should be as stiff as

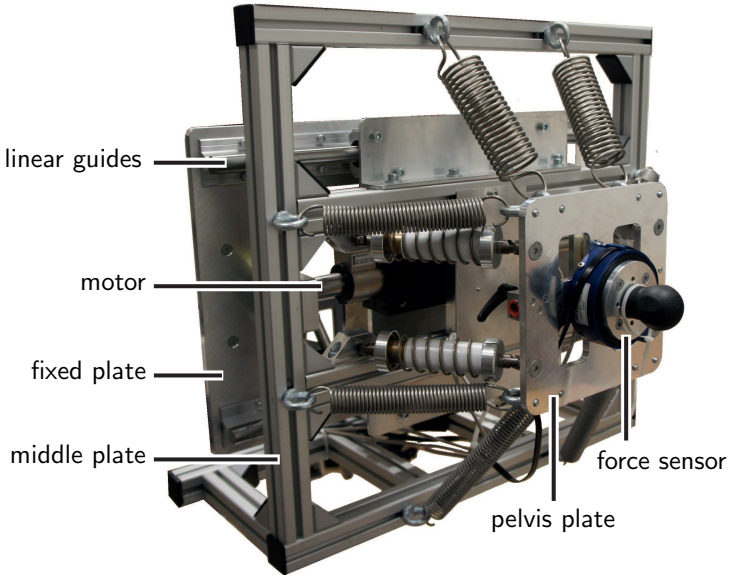


Figure 5.5: Compliant pelvis module (with force-torque sensor for evaluation)

possible. For comfort, the user wears a cushioned harness around the waist connected to leg loops that interface to the pelvis module as well as to an external body weight support system by adjustable straps. The moving plate of the pelvis module is padded with an anatomically formed foam. The user is secured to this pad over an adjustable bracket as seen in Fig. 5.4. For the supporting frame and treadmill, slightly adjusted components of the Lokomat environment (Hocoma AG, Volketswil, CH) were used.

5.2.2 Design of the Passive Compliant Module

Desired Stiffness Matrix

The compliant module must fulfill stiffness requirements in different DoF derived from the physiological movement pattern of the human pelvis during walking. A second objective is to minimize the coupling

between the various DoF by ensuring that the off-diagonal terms of the stiffness matrix are as close to zero as possible.

Pelvic motion is described by a set of six generalized coordinates $\boldsymbol{q}_\mathrm{p}$ with the three translational coordinates x_p, y_p, z_p of the pelvis center of mass and the zyx Euler angles θ_z, θ_y, θ_x. When the angles are zero, the x-axis points in the subject's anterior direction, y in the lateral direction (left with respect to the subject's perspective), and the vertical axis z in the subject's longitudinal axis direction. The angle θ_z denotes the transverse rotation (yaw) about the vertical z-axis, θ_y the tilt (pitch) angle about an intermediate y'-axis, and θ_x is the obliquity (roll) angle about the anterior-posterior u-axis. The forces and moments generated by the device on the user in the inertial coordinate directions are denoted F_x, F_y, F_z and M_x, M_y, M_z, respectively.

The main direction-specific objectives are:

- x-translation: the forward and backward motion of the user are to be constrained such that no low-frequency drift will occur, while still permitting some oscillation.
- y-translation: the resulting stiffness is chosen according to the desired performance of the SEA since the stiffness influences the bandwidth of the force controller and the achievable stiffness in impedance control.
- z-translation: any unwanted interaction forces caused by the device are to be avoided.
- θ_z-rotation: allow some freedom for users to deviate from a nominal configuration but keep them within a safe range of motion.

In the x-direction, the fundamental harmonic of human pelvis motion when walking on a treadmill must be enabled. This oscillation has less than 2 cm amplitude [116]. More oscillation is not desired when used in combination with a body weight support system and exoskeletal legs mounted to the treadmill frame. The user should stay directly under the deflection pulley of the BWS. Therefore, we chose to associate an excursion of 1 cm with a force of 100 N, which is very difficult to resist as previously observed with the FLOAT robot [95].

A study by [62] even found lower values of forces that could not be resisted. This leads to a desired stiffness of 10000 $^{\mathrm{N}}/_{\mathrm{m}}$.

The y-direction is the only actuated DoF. To enable low interaction forces in zero-impedance control, the stiffness of the spring should be made as low as possible. However, the stiffness must be at least as high as the maximum expected virtual stiffness that will be rendered by an assistive controller [98]. From experience with balance assistance on a similar robot [29], we know that making use of feedforward assistance, a relatively low stiffness of 3000 $^{\mathrm{N}}/_{\mathrm{m}}$ is sufficient.

Reducing interaction forces in the vertical direction is particularly challenging when the module is used together with an active leg orthosis, which typically has a high inertia. As no active elements are present to compensate for robot dynamics, the passive structure itself must minimize the forces. To achieve this goal, the eigenfrequency of the oscillating mass-spring assembly must be in the same range as an average human gait frequency [97]. The medium walking cadence is expected to be $f_0 = 0.75\,\mathrm{Hz}$ and the estimated mass of the oscillating parts of the device $m = 33.6\,\mathrm{kg}$, which is mainly caused by the mass of an additional exoskeletal structure for the legs, giving a resulting stiffness k_z equal to $m\omega_0^2$.

Rotational stiffness about the z-axis was chosen such that yaw rotation would be constrained in order to avoid crossing legs on the treadmill, using a relatively high stiffness of 100 $^{\mathrm{Nm}}/_{\mathrm{rad}}$. We chose the same value also for the two other rotational DoF.

The stiffness matrix $\mathbf{K}_{\mathrm{c}}$ has the matrix form:

$$\mathbf{K}_{\mathrm{c}} = -\begin{bmatrix} \frac{\partial F_x}{\partial x_{\mathrm{p}}} & \frac{\partial F_x}{\partial y_{\mathrm{p}}} & \cdots & \frac{\partial F_x}{\partial \theta_z} \\ \frac{\partial F_y}{\partial x_{\mathrm{p}}} & \frac{\partial F_y}{\partial y_{\mathrm{p}}} & \cdots & \frac{\partial F_y}{\partial \theta_z} \\ \vdots & \vdots & \ddots & \vdots \\ \frac{\partial M_z}{\partial x_{\mathrm{p}}} & \frac{\partial M_z}{\partial y_{\mathrm{p}}} & \cdots & \frac{\partial M_z}{\partial \theta_z} \end{bmatrix} \tag{5.1}$$

Summarizing the requirements for the individual DoF, the desired numerical values for the stiffness matrix $\mathbf{K}_{\mathrm{c,des}}$ are:

$$\mathbf{K}_{\mathrm{c,des}} = \mathrm{diag}\,(10000, 3000, 723, 100, 100, 100) \tag{5.2}$$

Units are $^{\mathrm{N}}/_{\mathrm{m}}$ and $^{\mathrm{Nm}}/_{\mathrm{rad}}$ for the respective entries.

Spatial Stiffness Synthesis

The spring assembly needs at least six individual springs for its six DoF, and the stiffnesses and configuration of the springs need to be determined from the desired stiffness matrix.

Huang and Schimmels introduced a screw theory-based numerical method to synthesize a desired stiffness matrix [43]. A prerequisite is that the off-diagonal terms of the stiffness matrix be zero; a further drawback of the method is that the stiffness matrix is valid only for one chosen point and can greatly change due to small deviations from this point. The stiffness matrix would ideally remain constant over the workspace of the end-effector but the deviations from the origin are not negligible. In addition, further constraints arise from practical considerations: for instance, the device must not interfere with the user's movements, especially concerning arm swing [12].

Therefore, an alternative approach was selected to determine the attachment points of the springs and their corresponding spring constants. This intuitive approach decomposes the three-dimensional problem into three two-dimensional problems as seen in Fig. 5.6. For each of these two-dimensional problems, the attachment points for four simple springs were found through geometric considerations and the spring stiffnesses and resting lengths were calculated via parameter optimization. The number of springs used and their attachment points on both plates were chosen to fulfill the space limitations and to achieve symmetry. Consequently, twelve springs were used in total, comprising four compression springs and eight tension springs. As few as possible compression springs are used since peripheral mechanics (i.e. two universal joints and a linear guide) are needed for these springs, which introduces additional weight into the system.

Springs 1 - 4, shown as projections in the z-y plane as in Fig. 5.6 serve to create the elasticity in lateral direction, while springs 5 - 8 perform the same function in the vertical direction. These springs also influence the rotational elasticity around the x-axis. The compression springs 9 - 12 in Fig. 5.6 act mainly in the x-direction but cover also the rotations about the y and z-axes. This arrangement of the springs minimally interferes with other motions of the user.

To perform the optimization procedure to determine the individual spring characteristics, the stiffness matrix must be written as a func-

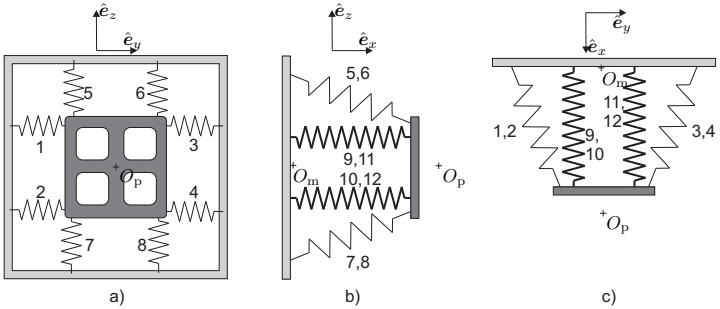


Figure 5.6: Geometric decomposition of the 3D problem into three 2D problems. Thin lines represent tension springs, bold lines

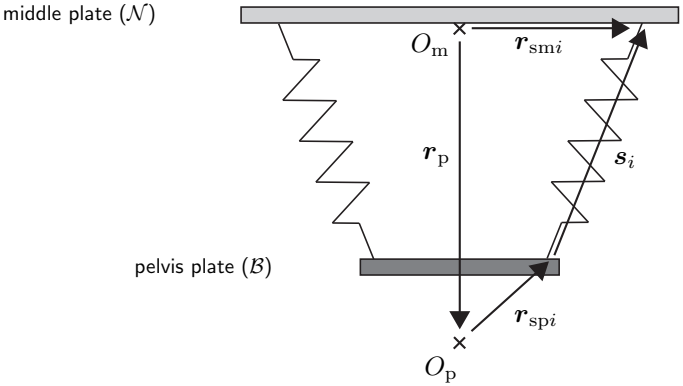


Figure 5.7: Definition of vectors describing spring geometry.

tion of the spring stiffnesses and resting lengths. As shown in Fig. 5.7, each spring i, $i = 1...n$ has an attachment point on the pelvis plate of position vector $\boldsymbol{r}_{\mathrm{p}} + \boldsymbol{r}_{\mathrm{sp}i}$ with respect to O_{m}. The attachment point of the same spring on the middle plate has position vector $\boldsymbol{r}_{\mathrm{sm}i}$ with respect to O_{m}.

An inertial frame $\mathcal{N}$ uses a set of direction vectors $\hat{\boldsymbol{e}}_x$, $\hat{\boldsymbol{e}}_y$, and $\hat{\boldsymbol{e}}_z$ connected to the middle plate. Because the middle plate does not rotate, these and the relative position vectors $\boldsymbol{r}_{\mathrm{sm}i}$ are constant. A frame $\mathcal{B}$ is connected to the pelvis plate and rotates with the human pelvis. The rotation matrix ${}^{\mathcal{N}}\mathbf{C}_{\mathcal{B}}$ describes the orientation of the human pelvis and thereby the pelvis plate with respect to the middle plate by mapping vectors with components expressed in the pelvis-fixed $\mathcal{B}$-frame to the inertial $\mathcal{N}$-frame.

Vector addition can be used to express the vector pointing along the i-th spring as:

$$ {}^{\mathcal{N}}\boldsymbol{s}_i = {}^{\mathcal{N}}\boldsymbol{r}_{\mathrm{sm}i} - {}^{\mathcal{N}}\boldsymbol{r}_{\mathrm{p}} - {}^{\mathcal{N}}\mathbf{C}_{\mathcal{B}} \cdot {}^{\mathcal{B}}\boldsymbol{r}_{\mathrm{sp}i}. \tag{5.3} $$

The associated spring force $\boldsymbol{F}_i$ that the i-th spring exerts on the moving plate is given by

$$ \boldsymbol{F_i} = k_i(|\boldsymbol{s}_i| - l_{0i})\frac{\boldsymbol{s}_i}{|\boldsymbol{s}_i|} \tag{5.4} $$

and the spring's moment vector $\boldsymbol{M}_i$ with respect to the pelvis center O_{p} as

$$ {}^{\mathcal{N}}\boldsymbol{M_i} = ({}^{\mathcal{N}}\mathbf{C}_{\mathcal{B}} \cdot {}^{\mathcal{B}}\boldsymbol{r}_{\mathrm{sp}i}) \times {}^{\mathcal{N}}\boldsymbol{F_i}, \tag{5.5} $$

with l_{0i} as resting length of the i-th spring.

Based on these equations, forces and moments acting on the pelvis can be compactly written as

$$ \boldsymbol{\tau} = \mathbf{A}\boldsymbol{p}, \tag{5.6} $$

whereby the parameter vector $\boldsymbol{p}$ is a function of stiffnesses k_i and resting lengths l_{0i}:

$$ \boldsymbol{p} := \begin{bmatrix} k_1 & k_2 & ... & k_n & k_1 l_{01} & k_2 l_{02} & ... & k_n l_{0n} \end{bmatrix}^{\mathrm{T}}, \tag{5.7} $$

the vector $\boldsymbol{\tau}$ subsumes the generalized forces of all $n = 12$ springs

$$\boldsymbol{\tau} := \begin{bmatrix} \sum_{i=1}^{n} \boldsymbol{F}_i \\ \sum_{i=1}^{n} \boldsymbol{M}_i \end{bmatrix}^{\mathrm{T}}, \tag{5.8}$$

and the matrix $\mathbf{A}$ encodes the geometric configuration:

$$\mathbf{A} := \begin{bmatrix} \boldsymbol{s}_1 & \dots & \boldsymbol{s}_n & \frac{\boldsymbol{s}_1}{|\boldsymbol{s}_1|} & \dots & \frac{\boldsymbol{s}_n}{|\boldsymbol{s}_n|} \\ \boldsymbol{a}_1 & \dots & \boldsymbol{a}_n & \frac{\boldsymbol{a}_1}{|\boldsymbol{s}_1|} & \dots & \frac{\boldsymbol{a}_n}{|\boldsymbol{s}_n|} \end{bmatrix}, \tag{5.9}$$

with entries

$$\boldsymbol{a}_i := \boldsymbol{r}_{\mathrm{sp}i} \times \boldsymbol{s}_i. \tag{5.10}$$

This optimization problem is linear in the parameters, and therefore, linear least-squares optimization is applied to find the optimal parameter vector $\boldsymbol{p}$ such that the forces and moments subsumed in $\boldsymbol{\tau}$ match a desired spatial profile $\boldsymbol{\tau}_{\mathrm{des}}(\boldsymbol{q}_{\mathrm{p}})$, which is a function of the pelvis generalized coordinate vector $\boldsymbol{q}_{\mathrm{p}}$. The training data $\boldsymbol{\tau}_{\mathrm{des}}(\boldsymbol{q}_{\mathrm{p}})$ for this optimization are generated in form of a grid with boundaries and resolution summarized in Table 5.1 using the desired diagonal stiffness matrix.

Table 5.1: Boundaries and resolution for the optimization grid

Coordinate	Boundaries	Resolution	Unit
p_x	$\pm$ 20	1	mm
p_y	$\pm$ 30	1	mm
p_z	$\pm$ 50	1	mm
θ_x	$\pm$ 10	0.1	deg
θ_y	$\pm$ 2	0.1	deg
θ_z	$\pm$ 10	0.1	deg

To map the individual spring stiffnesses to the stiffness matrix $\mathbf{K}_{\mathrm{c}}$ in end-effector space, the Jacobian matrix is used. Given a vector of individual spring lengths $\boldsymbol{L} = \begin{bmatrix} l_1 & l_2 & \dots & l_i \end{bmatrix}^{\mathrm{T}}$, which in turn

depends on the position vector $\boldsymbol{r}_\mathrm{p}$, the Jacobian matrix $\mathbf{J}$ is defined as:

$$\mathbf{J} = \frac{\partial \boldsymbol{L}}{\partial \boldsymbol{r}_\mathrm{p}} = \begin{bmatrix} \frac{\partial l_1}{\partial x_\mathrm{p}} & \frac{\partial l_1}{\partial y_\mathrm{p}} & \frac{\partial l_1}{\partial z_\mathrm{p}} & \frac{\partial l_1}{\partial \theta_x} & \frac{\partial l_1}{\partial \theta_y} & \frac{\partial l_1}{\partial \theta_z} \\ \frac{\partial l_2}{\partial x_\mathrm{p}} & \frac{\partial l_2}{\partial y_\mathrm{p}} & \frac{\partial l_2}{\partial z_\mathrm{p}} & \frac{\partial l_2}{\partial \theta_x} & \frac{\partial l_2}{\partial \theta_y} & \frac{\partial l_2}{\partial \theta_z} \\ \vdots & & & & & \\ \frac{\partial l_i}{\partial x_\mathrm{p}} & \frac{\partial l_i}{\partial y_\mathrm{p}} & \frac{\partial l_i}{\partial z_\mathrm{p}} & \frac{\partial l_i}{\partial \theta_x} & \frac{\partial l_i}{\partial \theta_y} & \frac{\partial l_i}{\partial \theta_z} \end{bmatrix} \tag{5.11}$$

The individual spring stiffnesses k_i are contained in matrix $\mathbf{K}_\mathrm{s}$:

$$\mathbf{K}_\mathrm{s} := \mathrm{diag}\left(k_1, k_2, \ldots, k_n\right), \tag{5.12}$$

and the stiffness matrix $\mathbf{K}_\mathrm{c}$ is

$$\mathbf{K}_\mathrm{c} = \mathbf{J}^\mathbf{T}\mathbf{K}_\mathrm{s}\mathbf{J}. \tag{5.13}$$

Table 5.2 shows the characteristics for the actually chosen physical springs that are used in the hardware realization.

Table 5.2: spring characteristics

Spring	Initial lengths	Stiffness
$1-4$	0.196 m	1577 N/m
$5-8$	0.161 m	628 N/m
$9-12$	0.19 m	1680 N/m

The resulting stiffness matrix $\mathbf{K}_\mathrm{c,opt}$ with the actual springs (which has the same units as above) is:

$$\mathbf{K}_\mathrm{c,opt} = \begin{bmatrix} 11134 & 0 & 0 & 0 & 0 & 0 \\ 0 & 3379 & 0 & 0 & 0 & 91 \\ 0 & 0 & 1017 & 0 & 28 & 0 \\ 0 & 0 & 0 & 79 & 0 & 0 \\ 0 & 0 & 28 & 0 & 85 & 0 \\ 0 & 91 & 0 & 0 & 0 & 126 \end{bmatrix} \tag{5.14}$$

The maximum workspace is determined by the excursion limits of the springs and by the physical endstops. It is verified that the device can also accommodate users who exhibit pathological (excessive)

pelvis motions when walking freely. The workspace boundaries are shown in Table 5.3, together with the forces occurring at these boundaries, for the case of single-DoF excursions from the neutral position. The limits of the workspace cannot be reached during walking due to the high forces occurring there. In the lateral direction, forces as low as 40-60 N can no longer be resisted [62].

Note that the device is not meant to support the weight of a person or to catch a person when falling. During gait rehabilitation training, such tasks would be performed by via an external body weight support system.

Table 5.3: Workspace of the pelvis module

Direction	Workspace	Max Force / Torque
x	-40 mm, +156 mm	-445 N, +1742 N
y	± 214 mm	723 N
z	± 288 mm	523 N
rot x	± 27 °	± 71 Nm
rot y	± 39 °	±86 Nm
rot z	± 39 °	±86 Nm

5.2.3 Sensing and Control

The absolute position and orientation of the MUCDA's end-effector are determined via a miniature camera module and a 6-DoF inertial measurement unit (IMU) as depicted in Fig. 5.8. The low-cost camera is equipped with a filter permeable only to infrared light and tracks an array of four active infrared markers (LEDs) mounted on the pelvis plate. The maximal spatial and temporal resolutions are 100 Hz and 0.14 mm, respectively. The image sensor (PixArt Imaging Inc.) is equipped with custom-made peripherals in order to read the information via serial bus. The obtained image of 128 x 96 pixel resolution is oversampled onboard to generate a 1024 x 768 resolution view. The IMU (MPU9250, Invensense, San Jose, USA) uses a three-axis accelerometer along with a three-axis gyroscope and is

sampled at a frequency of 1 kHz.

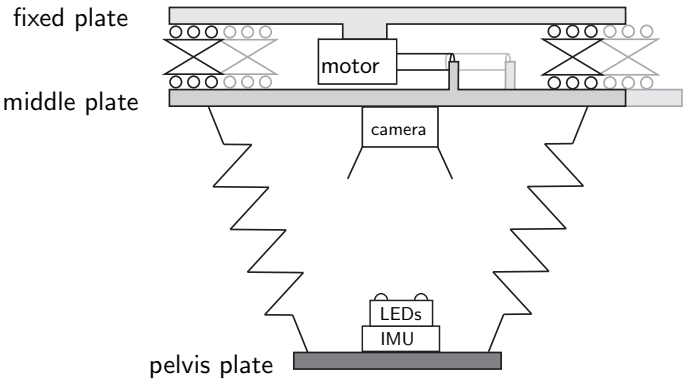


Figure 5.8: Sensing system with IMU, infrared camera and infrared LED-array

A multiplicative quaternion extended Kalman filter, which provides fast and reliable sensing of the end-effector position and orientation, is used to combine the IMU and camera data to estimate the position and orientation of the pelvis plate. The linear actuator has an incremental encoder for its displacement with a resolution of 0.05 mm and has an additional linear potentiometer for calibration.

A cascaded approach with an inner proportional velocity loop and an outer PI force control loop is applied, as often employed in unidimensional SEAs [112], [98]. The inner velocity loop is integrated into the motor drive, enabling a high sampling rate of 20 kHz; the external force loop is executed on a Matlab xPC target PC with a sampling rate of 1 kHz.

5.2.4 Evaluation Protocol

To validate the sensing concept and to evaluate the performance of the system, additional external sensor systems were used in the set-up as described below. Additionally, to evaluate the system concerning its interaction with the pelvis during gait, an experiment with a human subject was conducted.

Mechanical Stiffness Verification

To investigate the stiffness matrix and the achievable lateral transparency, tests were conducted using an evaluation setup consisting in a pelvis support system, an optical tracking system and a six-DoF force-torque sensor mounted on the pelvis plate. The external multi-axis force-torque sensor was the model 45E15A4 from manufacturer JR3 (JR3 Inc., Woodland, USA). The measurement range is $\pm$ 200 N for F_y and F_z with a resolution of 0.025 N, and $\pm$ 200 N for F_x with a resolution of 0.05 N.

The optical tracking system (Qualisys AB, Gothenburg, Sweden, Oqus camera series) comprising four cameras provided a spatial resolution of 0.5 mm and covered the entire range of motion of the system. The force data was captured at a frequency of 1 kHz and filtered with a lowpass second-order Butterworth filter with a cutoff frequency of 50 Hz, applied firstly in the forward and then in the backward direction for zero phase shifting. The position data was captured at a frequency of 1 kHz, while the calibration of the Qualisys system indicated a spatial accuracy of 0.5 mm.

The calculated stiffness matrix was validated by blocking the lateral DoF of the pelvis module (so that the springs have to deflect to let the pelvis plate move in the Y-direction) and by recording the forces and torques using the external six-DoF sensor and the position and orientation in 3D space.

External forces were induced into the system by manually moving the handle in 3D space, generating up to approximately 100 N and 10 Nm in all directions, respectively, corresponding to the maximum expected range of motion during normal operation. The precise range of the evaluation data is shown in Table 5.4.

All movements were conducted slowly so as to minimize the effects of dynamic forces on the results. It was verified that the pelvis plate acceleration plate did not exceed 0.1 g.

The stiffness matrix was estimated from the external force and position sensor data by computing mean values over the range of motion.

Table 5.4: Range of motion for validation

Direction	Range of Motion
x	$\pm$ 9 mm
y	$\pm$ 24 mm
z	$\pm$ 43 mm
rot x	$\pm$ 4 °
rot y	$\pm$ 6 °
rot z	$\pm$ 6 °

Sensor System Performance Evaluation

The performance of the position sensor was evaluated by comparing the internally calculated position data with the reference data from the optical tracking system. Similarly, the forces estimated from these positions in combination with the modeled spring characteristics were compared to the measurements of the external reference force sensor (shown in Fig. 5.5). Root mean square errors were calculated both for forces and positions. The same data set used to calculate the stiffness matrix was also employed for a quasi-static performance evaluation. In addition, dynamic measurements were conducted by manually moving the force sensor attached to the pelvis plate in all DoF with up to 3 Hz for 25 seconds. The range of motion is denoted in Table 5.4.

Control Evaluation

Firstly, to quantify the transparency of the module in the (lateral) Y-direction, the pelvis module was unblocked and the control system was activated in zero-impedance mode. The handle was then manually moved around in a periodic manner from left to right at different speeds and the lateral (Y-direction) force and position data were recorded.

The resulting data was used to estimate the parameters of a dynamic equation that describes the reflected end effector mass m_v and

damping d_v perceived at the pelvic plate according to

$$F_{u,Y} = m_v \cdot \ddot{Y}_p + d_v \cdot \dot{Y}_p. \tag{5.15}$$

in which $F_{u,Y}$ is the force applied by the user in Y-direction. The coefficients of the equation were estimated using linear regression. The position data were not filtered, while the velocity and acceleration were derived from low-pass-filtered position data using a second-order Butterworth filter with cutoff frequency 5 Hz, again applied in backward and forward directions to avoid phase shifting.

Secondly, a force-tracking experiment was conducted to evaluate force tracking performance, with the end effector being manually restrained and the force controller commanded to a reference sinusoidal force varying both in amplitude and frequency. Frequency slowly increased from 0.1 Hz to 10 Hz, while amplitude decreased with frequency from 40 N to 8 N. The recorded force estimate was then compared to the reference force in terms of phase lag and amplification in steady-state conditions for each frequency.

Usability Evaluation

A usability test with one male healthy subject (age: 31, height: 1.88 m, weight: 84 kg) was performed. The subject walked on a treadmill at 3 $^{\mathrm{km}}/_{\mathrm{h}}$ without the device while the pelvic movements were assessed by the external optical tracking system. The subject then performed the task with the pelvis module attached to his pelvis, without body weight support. The marker clusters were placed on the spina iliaca anterior/superior, just above the user fixation of the pelvis module. The position data was firstly partitioned into 30 gait cycles by finding the maximum values of the first derivative of the Y-translation. Subsequently, the data was averaged across these gait cycles and the standard deviation was calculated.

5.3 Results

5.3.1 Mechanical Stiffness Evaluation

The stiffness matrix $\mathbf{K}_{\mathrm{c,m}}$ obtained from the measured data comprising (mean values over the entire workspace) is

$$\mathbf{K}_{\mathrm{c,m}} = \begin{bmatrix} 10979 & 332 & 74 & 20 & 99 & 4 \\ 194 & 3160 & 25 & 8 & 18 & 71 \\ 376 & 155 & 1802 & 39 & 93 & 41 \\ 25 & 136 & 28 & 164 & 8 & 14 \\ 68 & 30 & 7 & 14 & 133 & 26 \\ 17 & 52 & 29 & 2 & 10 & 146 \end{bmatrix}. \tag{5.16}$$

The x- and y-direction levels are close to the design values, while in the z-direction and for the rotations, the stiffness values are somewhat higher than desired.

5.3.2 Sensor System Performance Evaluation

An excerpt of forces and moments as measured by the system's sensors and by the external reference sensors in dynamic conditions is shown in Fig. 5.9 [1].

The root mean square values corresponding to the difference between the target and actual positions and forces across the entire static and dynamic data sets are shown in table 5.5.

5.3.3 Closed-Loop Performance Evaluation

The identified reflected virtual mass in the Y-direction is $m_v = 3.6\,\mathrm{kg}$ and damping is $d_v = 14\,\mathrm{Ns/m}$. The corresponding R^2 value of 0.84 indicates a close fit of the model to the measured data.

Force tracking of the system is demonstrated in Fig. 5.10, in which measured and reference forces are shown for different frequencies. The experimental frequency response for force tracking is shown in Fig. 5.11, from which it is apparent that the bandwidth of the device is approximately 5 Hz. However, the phase at that frequency is

[1] Data is available at https://doi.org/20.500.11850/297732

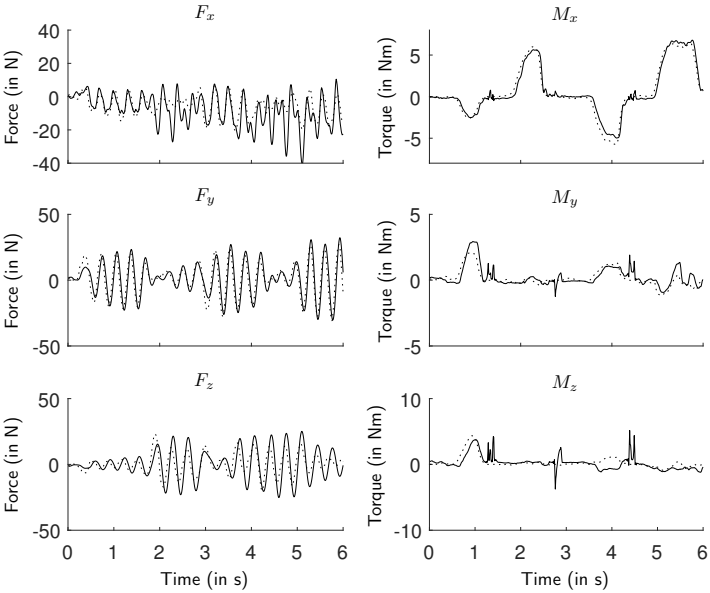


Figure 5.9: Forces and torques calculated from the system's sensors (solid line) compared to an external reference force sensor (dotted line) during dynamic movements.

Direction	RMSE position		RMSE force	
	static	dynamic	static	dynamic
x	4.24 mm	1.10 mm	19.30 N	8.70 N
y	1.31 mm	2.60 mm	3.03 N	6.85 N
z	1.41 mm	2.87 mm	3.07 N	5.94 N
rot x	2.13 deg	3.43 deg	352 Nmm	473 Nmm
rot y	3.00 deg	4.20 deg	731 Nmm	868 Nmm
rot z	3.87 deg	4.98 deg	818 Nmm	912 Nmm

Table 5.5: Root Mean Square Error (RMSE) of the position and force sensing system

already well beyond $-90°$, such that tracking of high-frequent references may not be possible with simple feedback controllers.

5.3.4 Usability Test

The results of the two conditions of the usability test with and without the device are shown in Fig. 5.12, while Fig. 5.13 shows the forces in the compliant module with their respective RMS errors. These errors were the deviation from the originally desired forces, as calculated with the desired stiffness matrix (5.2) for all DoF except for lateral translation, where the reference was zero force.

Movements in the three translational DoF as well as the rotation about the vertical axis appear to be little influenced by the human-robot interaction during walking. However, the pelvic tilt and obliquity rotations are less consistent, with both showing a decrease in amplitude with respect to free walking.

5.4 Discussion

Through the application of a new type of compliant actuator with multidirectional series and passive elasticity ,the MUCDA, a gait training platform has been realized. The proposed training device

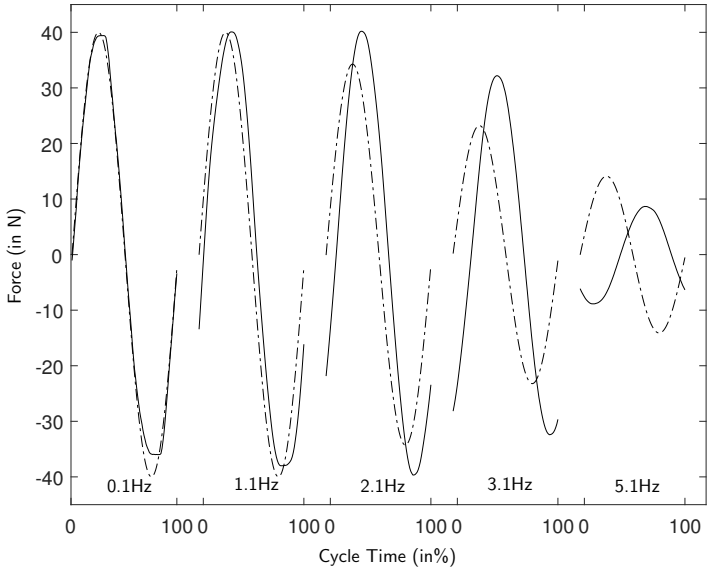


Figure 5.10: Experimental tracking of the reference force. Reference values are dotted, measured forces in solid lines.

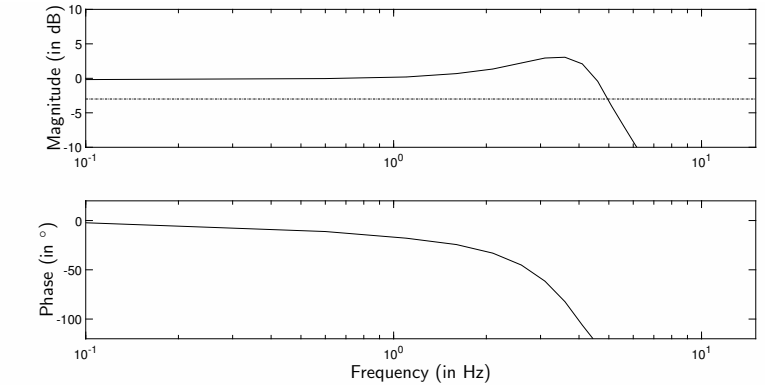


Figure 5.11: Experimental tracking frequency response of the pelvis module.

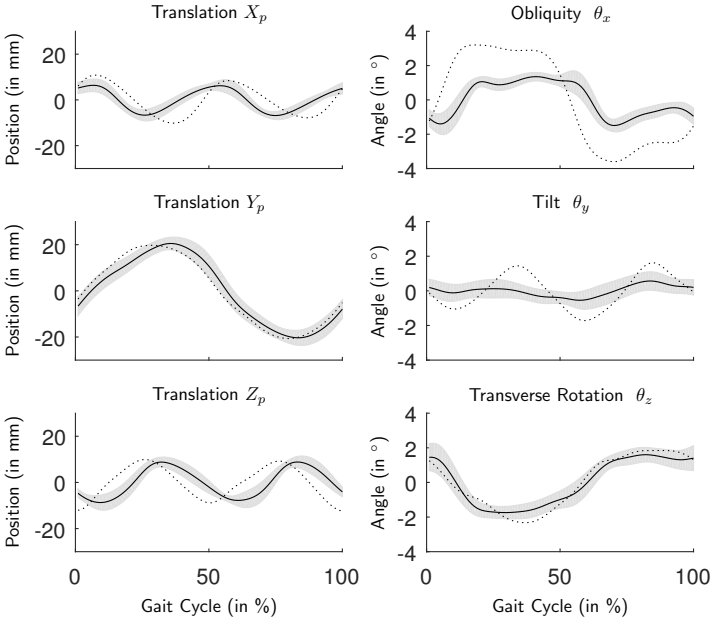


Figure 5.12: Pelvic motions during usability test, with dotted lines representing the physiological motions when walking freely on the treadmill, solid lines lines representing the mean of the motions for 30 gait cycles of the same healthy subject with attached pelvis module. The gray area represents the standard deviation. Data was acquired with an external optical tracking system.

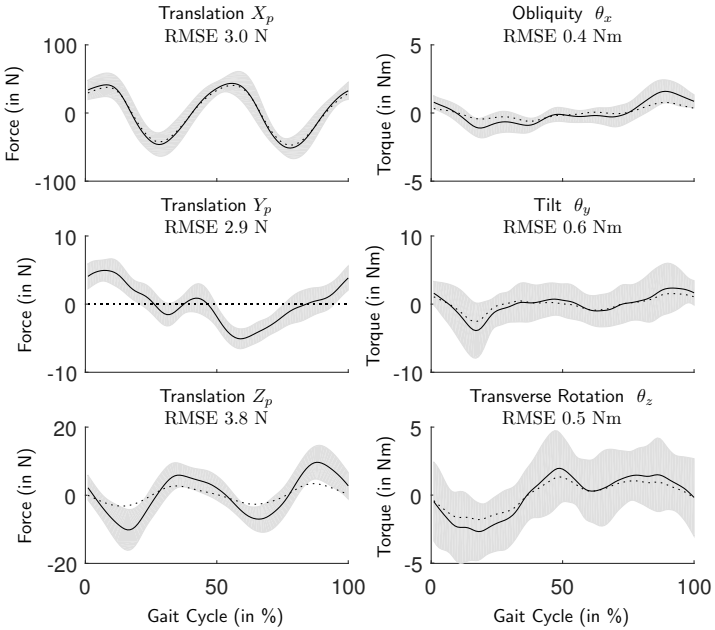


Figure 5.13: Forces and torques in the pelvic interface. Dotted lines represent the desired forces according to desired stiffness matrix (5.2). Solid lines represent the forces and torques estimated by the observer, the gray area the standard deviation. RMSE forces indicate the deviation from the desired forces. Note the different scales.

supports the posture and movement of a human user at the pelvis in all six DoF and actuates only weight shifting in the lateral direction.

The force and position sensing systems appear sufficiently reliable for the application. The force sensing errors in the $y-$ and z-directions were in the range of static friction, which was identified as 3.5 N in the y-direction and arises from the spring suspension system. The sensing error in the x-direction is due to the reduced accuracy of the tracking camera in this direction and the high stiffness in this direction, which tend to map small position errors to larger force errors.

Additionally, the results indicate that the device has a very low reflected inertia. Although the reflected inertia in Y-direction is still higher than the reported in [30], which is 1 kg, it is well below the 5.3 kg that can be added to the pelvis without significantly affecting gait [63]. The small forces occurring during the test can mostly be attributed to reflected damping. Consequently, the lateral interaction forces did not substantially influence pelvis kinematics. Despite the elastic interaction forces, the amplitudes of the other translations as well as transverse rotation are not perceivably affected either. Only obliquity and tilt seem to be reduced in amplitude, due to the chosen stiffnesses and resulting moments. Particularly obliquity is a DoF in which strong pathological movements can occur, including the so-called "hip-hiking" in stroke survivors. Depending on the training paradigm, reducing this movement to some extent via the spring forces may be desirable.

It can be seen that the relative phase of the three pelvis translations changes slightly when using the compliant module. This might be due to the compliance favoring a certain limit cycle that is not identical to the human pelvis motion. This effect would need to be further analyzed, and it could even be exploited in future designs in order to favor certain desired (physiological) motions over others.

The training platform may require further modification in order to accommodate the differences from normal gait frequently shown by individuals post-stroke. This group frequently shows increased kinematic variability e.g. in swing and stride times [6], as well as different gait patterns arising from compensatory mechanisms such as a prolonged swing time [70], [82]. Nevertheless, it has been

demonstrated that on average, individuals post-stroke have a walking cadence of as much as 80% that of able-bodied individuals [107]. Furthermore, the use of a spring to compensate for inertial effects of an exoskeleton is effective across a range of frequencies [97]: a spring structure demonstrably reduces vertical interaction forces even if the stiffness is not optimal. Therefore, despite being based on walking and step cadence data for able-bodied subjects, the proposed device should also be suitable for gait training for many individuals post-stroke.

The choice of the stiffness matrix entries for the passive DoF also depends on the adopted training paradigm. For example, some clinicians advocate constraining DoF in which pathological movements occur, such as hip hiking in obliquity. Others advocate leaving all DoF free that are not critical to the locomotion task and pose no safety risk during training. Because it is known that abnormal muscle activity does not necessarily cease even if the associated DoF is constrained during walking [66], the parameters in this paper were chosen closer to the latter paradigm. However, the proposed mechanical design principle can be equally used to adopt the former: For example, to fully suppress hip hiking passively, a higher stiffness would be desired for the fourth element on the diagonal of the stiffness matrix.

There remains some dispute about which DoF are most relevant to assist in pelvic movement of stroke survivors. For example, while [64] lists lateral translation as the only pelvis DoF that needs actuation, namely to assist balance, [77] advocates actuating obliquity instead to intervene with hip hiking. The actuation concept presented here can be adjusted to any other DoF by exchanging the motor and redesigning the stiffness matrix to desired values.

Though a single actuated DoF is considered in this article, the actuation concept could be extended to additional actuated DoF by keeping the same passive components while introducing more actuation on the input side. It would also be possible to place linear actuators in parallel to the springs. In the latter case, the individual spring elements serve as a series *and* parallel element to support the actuators, and possibly still act additionally as passive elastic components. Moreover, a particularly interesting property of elastic

elements is their capacity to store potential energy. This is possible both in series and parallel configuration; it depends on the specific task which configuration is more energy-efficient [9]. It remains to be investigated how the dual use of the springs in the MUCDA influences energetic aspects.

The sensing principle used in this device, consisting of a camera and an IMU, could also be replaced by alternative measurement methods. For example, it was recently shown that it is possible to precisely measure the length of coil springs using their inductance [99]. In that case, the springs themselves could be used as sensors.

In this work, we tested only the force tracking abilities of the pelvis manipulator with a simple force feedback controller and focus on zero-force tracking. Future research should address how to actuate weight shifting on a higher level when needed during therapy. One option would be to provide assistance (or challenge) in a low frequency fashion, for example to tackle asymmetry by constant offset forces, or to guide periodic weight shifting from step to step. Given the low fundamental frequency of human lateral pelvis motion, the limited bandwidth of the compliant actuator should not pose a limitation. However, when attempting to apply impulsive forces on the pelvis, the simple force feedback controller may not suffice, and the reference force could be added to the motor as a feed-forward term to improve force tracking. Impulsive forces could for example assist balance by inducing quick recovery movements. Previously, we had proposed a model-based control scheme that assisted lateral balance [29] using a modified Lokomat robot with lateral actuation. That controller was based on the concept of the extrapolated center of mass [38] or capture point [80]. Given that such a high-level controller is predictive, the low-level force controller can also be non-causal, further improving force tracking. Alternatively, one could choose stiffer springs or switch from impedance control to open-loop position control of the motor, exploiting the peak stiffness capabilities of the springs for more impulsive interaction.

In addition to guidance of the pelvis and possibly the legs, most individuals of the target group require partial support of their body weight during the single-support phase, which can be provided by an external body-weight-support system (BWS). Therefore, the sys-

tem's harness was designed such that it can be used directly for this purpose. Vertical unloading with a conventional BWS can interfere with the dynamic balancing task [74], and so to avoid any such undesired stabilizing effects, a system able to translate laterally with the user is preferred [113].

5.5 Conclusion

The feasibility of assigning dual roles to elastic elements in order to unify series elastic actuation in selected DoF and passive compliance in other DoF has been demonstrated. The MUltidimensional Compliant Decoupled Actuator concept opens up new perspectives for actuators with tuned compliance for high-performance haptic devices. A prototype that exhibits highly compliant interactions with the dynamic movement of human gait has been conceived and realized, allowing controlled lateral forces on the pelvis in combination with passive compliance in the other five DoF. Nevertheless, functional benefits of lateral weight shifting assistance, as well as the effectiveness for rehabilitation remain to be investigated.

Chapter 6

Discussion and Conclusion

6.1 Supporting Kinematics Reduces the Challenge

Replicating physiological kinematics alone is not sufficient to render challenging balance training for individuals post-stroke. In chapter 2, it was found that training using a device with fixed kinematics leads to unphysiological muscle activation patterns in individuals post-stroke, suggesting a limited rehabilitation effect on dynamic balance. The study with healthy subjects in chapter 3 indicated that gait training that helps maintain upright posture, via a BWS-induced pendulum effect, reduces the challenge of the balancing task. The findings of this pilot study are also supported by another study with similar a setup [23]. These findings show that guiding a patient on predefined trajectories or supporting kinematics that are believed to be physiological do not render an optimal rehabilitation environment. Conversely, the impedance should be in focus when designing a robotic gait trainer.

Enhancing active subject participation by a robot design focusing on transparency and sufficient degrees of freedom (DoF) is also sug-

gested by the review publication [73].

Other research groups have developed robotic devices following similar principles. An example is the RGR Trainer presented by [77] featuring a single actuator to alter the pelvis movement of individuals post-stroke in the frontal plane. They focused on correcting excessive pelvic obliquity ("hip hiking"), a secondary gait deviation, while leaving other DoF free. A study with four healthy subjects imitating hip hiking was conducted with a training session and an assessment session past the training. The participants did not seem to profit from the training, two of them even showed higher gait deviations after the training. The approach enforces physiological kinematics by guiding forces, thereby masking an individual's movement errors, which may hamper learning in individuals with neurological impairments.

Impedance control and control strategies as "assist-as-needed" suggest to be more effective than position control strategies [52]. This approach is now transferred from the domain of control to the domain of hardware development.

In this thesis, a corresponding paradigm for hardware design is followed: constrain-as-needed. Two modules are presented following this paradigm. In contrast to the traditional approach of actuating the desired DoF and blocking the residual ones, the passive DoF are not constrained to a certain position but to within a range. In the application presented in this thesis, one pelvis DoF is identified as necessary to be actuated for supporting a human user maintain lateral balance, namely lateral translation. The DoF believed to be of less importance allow movement variability while ensuring a certain safe range of motion and ensuring low interaction forces (transparency).

A very similar conclusion was reached by Jos Meuleman as design requirements for LOPES II, as presented in [61]. This end-effector based gait rehabilitation robot incorporates two actuated DoF for the pelvis, namely forward/aft and pelvis mediolateral, while all other DoF are free. It also includes exoskeletons for the leg, actuating hip flexion/extension, hip abduction/adduction, knee flexion/extension, which are the same DoF as those of the exoskeleton that can be paired with the MUCDA (see appendix C).

6.2 Multi-DoF Compliance Enables Challenging Training

In order to follow the paradigm of constrain-as-needed, an engineering solution needs to be found to allow near-free movement in some DoF, while still actuating others, in a decoupled fashion.

Achieving decoupling is challenging, and undesired interaction forces due to uncompensated robot dynamics in the "free" DoF are unavoidable for many design solutions. For example, the LOPES II requires shifting a massive structure in mediolateral direction, such that peak interaction forces of 50 N are reported at the pelvis, even in the most transparent setting [61]. For the RGR, neither interaction forces nor subject movements in other degrees of freedom than the target (obliquity) have been reported. However, given that also in that device the actuation is arranged in a serial fashion, the entire actuator structure for the actuated DoF has to be moved by the subject in lateral direction. This still affects the free DoF by the robot's inertia, such that the transparency is limited.

In the devices presented in this thesis, transparency in the passive DoF is achieved by decoupling the DoF, allowing very low inertia. More specifically, challenging balance training is enabled by the development of a two-dimensional BWS system and a six-dimensional pelvis support module. These modules provide a safe gait training environment for individuals with neurological impairments, and may offer support without restraining human movement.

An extension for body weight support systems which reduces lateral stabilizing forces through tracking lateral movements of the subject's pelvis has been designed, constructed and tested. The system is based on fully decoupling the high-force vertical support from an additional, highly transparent actuated lateral DoF. It can follow an individual's lateral displacement under various levels of vertical supporting force, but also apply lateral forces to support or perturb, when desired. The body weight support system extension substantially reduces the induced lateral forces while effects on the vertical support forces are kept to a minimum. This way, the pendulum effect is avoided and the challenge is not diminished compared to unassisted training. The system can be applied to advanced body weight supported treadmill

training, in which subjects are required to actively maintain their balance.

The concept of the MUltidimensional Compliant Decoupled Actuator opens up new perspectives for elastic actuators with tuned compliance for high-performance haptic devices for humans. The feasibility of assigning dual roles to elastic elements in order to unify Series Elastic Actuation in selected DoF and passive compliance in other DoF has been demonstrated. A prototype that exhibits highly compliant interactions with the dynamic movement of human gait has been conceived and realized, allowing controlled lateral weight-shifting assistance in combination with passive compliance that supports near-physiological oscillations in the other five DoF. The system offers the possibility of more challenging training sessions, although it remains unclear how well this will be accepted by patients and what adjustments to the different ability levels of the patients will have to be realized. Compared to the above mentioned LOPES II, the undesired interaction forces in mediolateral direction at the pelvis are reduced by a factor of 10 measured at a similar walking speed. The MUCDA was not combined with a leg exoskeleton in the presented study, while the LOPES II was. A robot featuring MUCDA combined with an exoskeleton can be found in Appendix C. The exoskeleton is suspended at a fixed frame and actuated independently from the pelvis module, to ensure the mediolateral forces at the pelvis are not increased.

6.3 Outlook

A possible realization of a robotic gait trainer featuring all the presented modules is shown in Appendix C. Such a setup can be used to investigate how the presence of the robot influences the gait parameters of healthy subjects. The modules presented in this thesis can be used to investigate the benefit of balance training on the rehabilitation process and thereby guide future developments in the field of robotic gait rehabilitation. No clinical tests have been conducted so far with theses modules.

Control strategies to support human posture and the dynamic balancing task in the early stages of rehabilitation, as well as strategies to

challenge individuals in later stages, remain to be developed. These could follow up on controllers such as the feed-forward model-based balance assistance scheme [96] or rendering a spring [57]. Residual movements of the pelvis and muscle activation are of special interest. If promising results are achieved, a study with individuals post-stroke should be conducted and functional ambulation should be compared to a training with a conventional robotic gait trainer. The outcome measures should focus on functional tests like the Six-Minute Walk Test.

The 'constrain-as-needed' design paradigm can also enhance affordability of devices, as production and maintenance costs decreases. The employed hardware for the pelvis module, including sensor technology, is low-cost and does not require high manufacturing accuracy. Reduced cost may lead to higher density of such tools in rehabilitation facilities, relieving therapists from some of the physical labor and allowing them to focus more on coaching and tailoring training sessions to their clients.

In connection with current international initiatives on semi-automated collection of data, as in the International Stroke Trial Database [87], National Stroke Registries [13], the Res-Q database [3], the Integrated Stroke Outcomes Database (ISOD) [83], or the EMSCI database for spinal cord injury [1], a higher density of robots in rehabilitation clinics may allow inexpensive data collection on a larger number of patients. Later analysis of this data may help further improve the efficiency and efficacy of gait rehabilitation programs, and it could drastically reduce the need for expensive dedicated clinical trials.

Training of dynamic balance during gait could also be beneficial for other target groups such as individuals receiving a lower-limb prosthesis who need to relearn weight shifting. The stationary nature of the device however, imposes limitations on its application, as tasks like obstacle avoidance cannot be trained on a treadmill. Therefore, other groups man benefit less from treadmill-based robot-assisted training, like individuals with Parkinson's disease, who often have difficulty with gait initiation rather than steady-state walking, or elderly people, who may just need a BWS system to prevent falling.

Besides training, a secondary purpose of the developed technology could be assessment of an individual's balance capabilities, aiding

therapists determine objective measures of the current status and progress of their clients. For this to be a viable solution, a standardized procedure for testing needs to be developed, ideally independent of the particular robotic hardware. To make sure such tests are reliable, it is important to reduce any artifacts induced by the robot [90]. In this aspect, the developed modules may offer particularly suitable solutions thanks to their very low inertia. Needed steps would include setting up specific standardized assessment protocols that uncover balance strategies, e.g. based on CoM and CoP [106], collecting normative reference data of a range of physiological gaits and reactions to perturbations, to enable identifying deviations [90], and linking outcomes to existing clinical measures.

Future possible hardware developments include a detachable and interchangeable pelvis module spring assembly with different spring characteristics. The therapist could choose a version with higher spring stiffness for severely affected individuals and a version with lower spring stiffness for patients in later rehabilitation stages. The inexpensive parts as springs and sensors enable a set of different modules.

Appendix A

Clinical Report Form to Study with Lokomat FreeD module

CRF

Case Report Form
Version 1.4, 22.04.2014

Lokomat Free Walking

Advanced robotic gait training with additional degrees of freedom: towards a free-walking training system

Name und Anschrift des verantwortlichen Prüfarztes (in Schriftform) Stempel

Prof. Dr. med. Andreas Luft, Leitender Arzt, Universitätsspital Zürich, Neurologische Klinik, Frauenklinikstrasse 26, 8091 Zürich

Ablaufdiagramm

Screening:

	Screening Session
Informed consent	*X*
Demografische Daten	*X*
Krankengeschichte	*X*
Ein- Ausschlusskriterien	*X*
Medizinische Untersuchung	*X*

Grp A:

	Mixed session	***Training Device A***	***Training Device Std***
6min Walking Test		*X*	*X*
Movement Tracking	*X*		
EMG	*X*		
Lokomat Training	*A & Std*	*A*	*Std*
Frageogen	*X*	*X*	*X*
6min Walking Test		*X*	*X*
Adverse Events	*X*	*X*	*X*

Inhaltsverzeichnis Visiten

Screening Datum: I__I__I / I__I__I / I__I__I__I__I

Aufklärung und Einverständniserklärung
Der Patient wurde vom Prüfarzt vollständig über die Studie EKNZ…….. aufgeklärt. Eine schriftliche Patienteninformation wurde ausgehändigt und allfällige Fragen des Patienten wurden beantwortet.

Datum Einverständniserklärung: I__I__I / I__I__I / I__I__I__I__I
Wenn die Einverständniserklärung nicht unterschrieben wurde, hat dies eine Nichtaufnahme in die Studie zur Folge!

Prüfarzt: Ich bestätige, dass der Patient/die Patientin (oder sein/ ihr gesetzlicher Vertreter) entsprechend den ICH/ GCP-Richtlinien ordnungsgemäss über die Prüfung informiert wurde und sein/ ihr schriftliches Einverständnis zur Teilnahme an dieser Studie, zu oben genannten Daten, gegeben hat.

Datum & Unterschrift: I__I__I / I__I__I / I__I__I__I__I _____

1. Demographische Daten

1.1 Geburtsjahr: I__I__I__I__I

1.2 Geschlecht: ☐ männlich ☐ weiblich

2. Einschluss-/ Ausschlusskriterien

2.1 Einschlusskriterien:	**Ja**	**Nein**
1. ≥ 18 Jahre alt	☐	☐
2. Schlaganfall in der Vorgeschichte	☐	☐
3. Schwäche der unteren Extremitäten (MRC <5)	☐	☐
4. Fähigkeit und Bereitschaft an der Studie mitzumachen	☐	☐
5. Freiwillig unterschriebene Einverständniserklärung	☐	☐

Wenn eines der Einschlusskriterien mit “Nein” angekreuzt wurde, ist der Patient von der Studie auszuschliessen

2.2 Ausschlusskriterien:	**Ja**	**Nein**
1. Schwere Spastik der unteren Extremitäten (Ashworth 4 od. 5)	☐	☐
2. Körpergewicht > 135 kg?	☐	☐
3. Orthose des Roboters ungeeignet für die unteren Extremitäten der TeilnehmerIn?	☐	☐
4. Instabile Knochenkrankheit (z.B. Osteoporose, nicht-verheilte Brüche, instabile Wirbelsäule)	☐	☐
5. Schwere steife Kontrakturen	☐	☐
6. Arthrodese der Hüfte, des Knies oder Knöchels	☐	☐
7. Schwere Komorbidität oder medizinische Kontraindikation (z.B. Herz-Kreislauf Erkrankung)	☐	☐
8. Schwere Gefässerkrankung in den unteren Extremitäten	☐	☐
9. Immobilität oder Immobilisierung (z.B. durch Osteomyelitis oder andere Infektionskrankheiten)	☐	☐
10. Erosive oder wuchernde Hautkrankheit der unteren Extremitäten oder des Rumpfes	☐	☐
11. Schwere kognitive Defizite (z.B. Demenz, Aphasie)	☐	☐
12. Schwere Motivationsstörung (z.B. Depression, CFS)	☐	☐

13. Unkooperatives oder aggressives Verhalten (z.B. bei Psychosen) oder bekannte oder erwartete non-compliance ☐ ☐
14. Bekannter oder vermuteter Drogen oder Alkoholmissbrauch ☐ ☐
15. Einschluss in eine klinische Studie in den letzten 4 Wochen ☐ ☐
16. Schwangerschaft (ggf. mittels Urintest ausgeschlossen) ☐ ☐

Wenn eines der Ausschlusskriterien mit “Ja” angekreuzt wurde, ist der Patient von der Studie auszuschliessen.

Eignung des Patienten: Erfüllt der Patient alle Selektionskriterien für die Teilnahme an der Studie? ☐ Ja ☐Nein. Falls nein, muss das Studienabschluss-Formular auf Seite 20 ausgefüllt werden.

Screening

Ich habe die Angaben zu diesem Assessment überprüft. Die Angaben sind vollständig und korrekt.

Datum und Unterschrift des Prüfarztes: _____

Persönliche Daten

1. Studienindikation Anamnese

☐ Schlaganfall

2. Datum des Schlaganfalles |__|__|__|__|__|__|__|

3. Parese **Rechtes Bein** ☐
Linkes Bein ☐

4. Aktuelle Rehabilitationstherapie und Häufigkeit?

☐ Keine
☐ Ergotherapie __xlWoche
☐ Physiotherapie __xlWoche
☐ Medizinische Trainingstherapie __xlWoche
☐ Logopädie __xlWoche

5. Händigkeit

Rechtshänder ☐
Linkshänder ☐

Relevante medizinische Vorgeschichte und Begleiterkrankungen			
Diagnose/ Erkrankung	Beginn TT/MM/JJJJ	Medikamentöse Therapie	Bemerkungen:
		☐Ja ☐ Nein	
		☐Ja ☐ Nein	
		☐Ja ☐ Nein	
		☐Ja ☐ Nein	

6. BMI - Akut (1-14d):	
Grösse [cm]	
Gewicht [kg]	

Klinische Scores

NIH Stroke Scale - Screening

1a. Bewusstseinsgrad 0 = wach 1 = somnolent (durch geringe Stimulation weckbar) 2 = soporös (benötigt wiederholte Stimulationen reagiert nur auf starke schmerzhafte Reize) 3 = Koma (antwortet nicht oder nur mit motorischen Reflexen oder automatischen Antworten)	__/3
1b. Bewusstseinsgrad-Fragen Fragen nach dem aktuellen Monat und dem Alter des Patienten (keine Hilfestellung, nur erste Antwort zählt) 0 = beide Antworten richtig 1 = eine Antwort richtig oder Patient kann nicht sprechen wegen Dysarthrie oder Intubation 2 = keine Antwort richtig oder aphasischer Patient oder stuporöser/komatöser Patient	__/2
1c. Bewusstseinsgrad-Befehle Augen öffnen und schliessen lassen, dann öffnen und schliessen der nicht betroffenen Hand (falls Hand nicht möglich, anderen Befehl verwenden; falls Aphasie, Pantomime benutzen) 0 = beide Befehle richtig ausgeführt; 1 = einen Befehl richtig ausgeführt 2 = keinen Befehl richtig ausgeführt	__/2
2. Augenbewegungen Nur horizontale Bewegungen testen. Nur willkürlicher od. reflektorischer, kein kalorischer Test 0 = keine Blicklähmung 1 = partielle Blickparese (abnormal bei beiden Augen aber Besserung bei occulocephalem Manöver oder abnormal bei einem Auge) 2 = starke Abweichung oder komplette Blickparese beider Augen	__/2
3. Gesichtsfeld Alle Quadranten testen 0 = normal oder monookkulare Blindheit ohne Gesichtsfelddefizit des anderen Auges 1 = Quadrantenanopsie 2 = komplette Hemianopsie 3 = Blindheit (auch kortikale Blindheit)	__/3
4. Motorik des Gesichts (Facialisparese) Lachen, danach Augen schliessen (bei Aphasie, Pantomime benutzen oder auf die Symmetrie der Grimasse bei Schmerzreizen achten) 0 = normale, symmetrische Bewegung 1 = geringe Parese (glatte Nasolabialfalte, Asymmetrie beim Lachen); 2 = komplette oder fast komplette Parese der unteren Gesichtshälfte 3 = komplette Parese im unteren und oberen Gesichtsbereich	__/3
5. a Motorik des rechten Arms Rechten Arm für 10 sec. bei 90° im Sitzen oder 45° im Liegen halten. 0 = kein Absinken in 10 sec 1 = Absinken nach weniger als 10 sec aber ohne die Unterlage zu berühren 2 = Patient kann den Arm halten aber nicht vollständig extendieren oder Arm sinkt nieder und berührt die Unterlage. 3 = keine Anstrengung gegen die Schwerkraft möglich 4 = keine Bewegung möglich (Plegie) x = nicht beurteilbar / Amputation	__/4
5. b Motorik des linken Arms Linken Arm für 10 sec. bei 90° im Sitzen oder 45° im Liegen halten. 0 = kein Absinken in 10 sec 1 = Absinken nach weniger als 10 sec aber ohne die Unterlage zu berühren 2 = Patient kann den Arm halten aber nicht vollständig extendieren oder Arm sinkt nieder und berührt die Unterlage. 3 = keine Anstrengung gegen die Schwerkraft möglich 4 = keine Bewegung möglich (Plegie) x = nicht beurteilbar / Amputation	__/4
6. a Motorik des rechten Beins Rechtes Bein für 5 sec. bei 30° im Liegen halten.	__/4

0 = kein Absinken in 5 sec 1 = Absinken nach weniger als 5 sec aber ohne die Unterlage zu berühren 2 = partielle Überwindung der Schwerkraft (Patient kann das Bein halten aber nicht vollständig extendieren oder das Bein sinkt nieder und berührt die Unterlage). 3 = keine Anstrengung gegen die Schwerkraft möglich 4 = keine Bewegung möglich (Plegie) x = nicht beurteilbar / Amputation	
6. b Motorik des linken Beins Linkes Bein für 5 sec. bei 30° im Liegen halten. 0 = kein Absinken in 5 sec 1 = Absinken nach weniger als 5 sec aber ohne die Unterlage zu berühren 2 = partielle Überwindung der Schwerkraft (Patient kann das Bein halten aber nicht vollständig extendieren oder das Bein sinkt nieder und berührt die Unterlage). 3 = keine Anstrengung gegen die Schwerkraft möglich 4 = keine Bewegung möglich (Plegie) x = nicht beurteilbar / Amputation	**__/4**
7. Ataxie Beidseits Finger-Nasen- und Ferse-Schienbeinversuch bei geöffneten Augen testen. Nicht testen bei unvollständiger Wachheit, Verständnisproblemen oder Schwäche 0 = keine Ataxie oder aufgrund von Schwäche 1 = vorhanden in einer Extremität 2 = vorhanden in 2 oder mehr Extremitäten X = nicht beurteilbar / Amputation	**__/2**
8. Sensibilität Prüfung mit spitzigem Holzstäbchen. Falls Patient aphasisch oder soporös, Verwendung von schmerzhaften Stimuli. Prüfung an Gesicht, Stamm, Armen und Beinen. 0 = normal; 1 = partieller Verlust (Patient bemerkt Berührung auf der betroffenen Seite weniger als auf der gesunden Seite oder Patient bemerkt eine Berührung aber nicht die Spitze auf der betroffenen Seite oder Patient reagiert auf schmerzhaften Stimulus) 2 = schwerer oder völliger Verlust (Pat. bemerkt die Berührung nicht)	**__/2**
9. Sprache Nachsprechen lassen, Gegenstände benennen, Situation beschreiben, Mehr-Folgen-Befehl 0 = normal 1 = milde bis mässige Aphasie (Paraphasien, Wortverwechslungen), Kommunikation möglich 2 = schwere Aphasie, Kommunikation weitgehend unmöglich 3 = stumm, globale Aphasie	**__/3**
10. Dysarthrie 0= normale Artikulation 1= milde bis mässige Dysarthrie (einzelne Wörter verwaschen); 2 = nahezu unverständlich oder schlecht oder mutistisch X = nicht beurteilbar; nur bei Intubation oder mechanischer Barriere	**__/2**
11. Neglekt 0 = kein Neglekt (Wahrnehmung zu beiden Seiten normal) 1 = Neglekt in einer Modalität (zB. visuell oder taktil) bzw. Hemineglekt 2 = kompletter Neglekt oder Hemineglekt in mehr als einer Modalität (nimmt die eingene Hand nicht wahr oder orientiert sich nur zu einer Seite)	**__/2**
N**IH Stroke Scale Gesamtscore:**	**__/42**

MMST - (Patienten mit Schlaganfall) - Pretraining

(Bereit halten: Uhr, Bleistift, leeres Blatt) Blatt mit Beschriftung / Figur → teilen

1. Orientierung	
Welches Jahr haben wir?	☐ OK
Welche Jahreszeit ist jetzt?	☐ OK
Welchen Monat haben wir?	☐ OK
Welches Datum haben wir heute?	☐ OK
Welcher Wochentag ist heute?	☐ OK
In welchem Kanton sind wir hier?	☐ OK
In welcher Stadt?	☐ OK
In welcher Strasse oder Stadtteil (oder der eigene Wohnort)?	☐ OK
In welchem Gebäude / welcher Klinik sind wir hier?	☐ OK
Auf welchem Stockwerk?	☐ OK ___ Pkt.
2. Merkfähigkeit	
Sofortige Wiederholung nach Vorsprechen (1 Sek pro Wort); erste Wiederholung bestimmt die Punktzahl, max. 5 Wiederholungen (wenn nicht alle 3 Wörter gelernt, kann das Erinnern in Punkt 4 nicht geprüft werden	
„Zitrone“	☐ OK
„Schlüssel“	☐ OK
„Ball“	☐ OK ___ Pkt.
3. Aufmerksamkeit und Rechnen	
Rückwärtszählen von 100 in 7er Schritten. Stopp nach 5 Subtraktionen und Rückwärtsbuchstabieren „PREIS“ (SIERP=5; SIREP=3). Höhere der beiden Wertungen wird gezählt.	
„93“ „S“	☐ OK ☐ OK
„86“ „I“	☐ OK ☐ OK
„79“ „E“	☐ OK ☐ OK
„72“ „R“	☐ OK ☐ OK
„65“ „P“	☐ OK ☐ OK
4. Erinnern	
„Zitrone“	☐ OK
„Schlüssel“	☐ OK
„Ball“	☐ OK ___ Pkt.
5. Benennen	
Armbanduhr	☐ OK
Bleistift	☐ OK ___ Pkt.
6. Wiederholen	
Nachsprechen: „Kein Wenn und Aber“ (nur 1 Versuch erlaubt)	☐ OK ___ Pkt.
7. Dreiteiliger Befehl	
„Nehmen Sie ein Blatt in die Hand	☐ OK
falten es in der Mitte	☐ OK
und legen es auf den Boden“.	☐ OK ___ Pkt.
8. Reagieren	
Dem Patienten ein Blatt zeigen mit der Beschriftung „Schliessen Sie die Augen“. Den Patient soll den Text lesen und ausführen. 1 Punkt wenn der Patient die Augen schliesst.	☐ OK ___ Pkt.
9. Schreiben	
Der Patient soll einen beliebigen Satz aufschreiben. Nicht diktieren. Der Satz muss einen Sinn ergeben, ein Subjekt und ein Verb enthalten. Grammatik und Punktion nicht relevant. 1 Punkt	

	☐ OK ___ Pkt.
10. Abzeichnen	
Vorzeichnen zweier sich überschneidender Fünfecke, die der Patient abzeichnen soll. 1 Punkt für insgesamt 10 Ecken und 1 Punkt für 2 Überschneidungen.	☐ OK ☐ OK ___ Pkt.
Gesamtscore:	**___/30 Pkt.**

CRF Lokomat	**Patienten Nr.** I__I__I **Prüfer ID** I__I__I
Training Mixed	**Datum:** I__I__I / I__I__I / I__I__I__I__I

Training 1: Beide Geräte

Ich habe die Angaben zu dieser Visite überprüft. Die Angaben sind vollständig und korrekt.

Datum und Unterschrift des Prüfarztes: _____

Lokomat Seriennummer: I__I__I

Beginn Training Lokomat Pro:	Uhrzeit: I__I__I : I__I__I
Ende Training Lokomat Pro:	Uhrzeit: I__I__I : I__I__I
Beginn Training Lokomat FREEW:	Uhrzeit: I__I__I : I__I__I
Ende Training Lokomat FREEW:	Uhrzeit: I__I__I : I__I__I

Fragebogen: Patientenbefragung vor dem Training mit Lokomat Pro:

	Trifft voll zu	Trifft überhaupt nicht zu
Mein Allgemeinbefinden ist gut	☐☐☐☐☐☐☐☐☐☐	
Beschwerden	_____	

Unerwünschte Wirkungen:
Der Patient wurde über unerwünschte Wirkungen nach der Anwendung des Medical Devices gefragt.
Unerwünschte Wirkungen ☐Ja ☐Nein

Datum: I__I__I / I__I__I / I__I__I__I__I Zeit: I__I__I : I__I__I

Falls unerwünschten Wirkungen aufgetreten sind werden diese auf Seite 20 beschrieben.

Fragebogen: Patientenbefragung nach dem Training mit Lokomat Pro:

	Trifft voll zu	Trifft überhaupt nicht zu
Ich hätte gerne länger trainiert	☐☐☐☐☐☐☐☐☐☐	
Das Training war zu anstrengend	☐☐☐☐☐☐☐☐☐☐	
Das Training wurde sehr langweilig	☐☐☐☐☐☐☐☐☐☐	
Das Training hat Spass gemacht	☐☐☐☐☐☐☐☐☐☐	
Das Training sollte abwechslungsreicher sein	☐☐☐☐☐☐☐☐☐☐	
Das Training hat zu lange gedauert	☐☐☐☐☐☐☐☐☐☐	
Ich hatte das Gefühl, die Trainingsstunde ging sehr schnell vorbei	☐☐☐☐☐☐☐☐☐☐	

Ich würde das Trainingssystem gerne häufiger nutzen	☐☐☐☐☐☐☐☐☐☐
Ich habe das Gefühl, das heutige Training war erfolgreich	☐☐☐☐☐☐☐☐☐☐
Das Training war zu wenig anstrengend	☐☐☐☐☐☐☐☐☐☐
Das Eingespannt-werden in den Lokomat empfand ich als angenehm.	☐☐☐☐☐☐☐☐☐☐
Das Eingespannt-sein im Lokomat empfand ich als angenehm.	☐☐☐☐☐☐☐☐☐☐
Ich fühlte mich wohl während des Eingespanntseins im Lokomat.	☐☐☐☐☐☐☐☐☐☐
Ich war in gutem Kontakt zum Therapeuten.	☐☐☐☐☐☐☐☐☐☐
Ich fühlte mich vom Therapeuten wahrgenommen.	☐☐☐☐☐☐☐☐☐☐
Es war anstrengend das Gleichgewicht zu halten.	☐☐☐☐☐☐☐☐☐☐
Ich fühlte mich im Gerät sicher.	☐☐☐☐☐☐☐☐☐☐

Beobachtungen des Prüfers, spezielle Ereignisse:

Fragebogen: Patientenbefragung vor dem Training mit Lokomat FREEW:

	Trifft voll zu									Trifft überhaupt nicht zu
Mein Allgemeinbefinden ist gut	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐
Beschwerden	_____									

Fragebogen: Patientenbefragung nach dem Training mit Lokomat FREEW:

	Trifft voll zu									Trifft überhaupt nicht zu
Ich hätte gerne länger trainiert	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐
Das Training war zu anstrengend	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐
Das Training wurde sehr langweilig	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐
Das Training hat Spass gemacht	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐
Das Training sollte abwechslungsreicher sein	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐
Das Training hat zu lange gedauert	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐
Ich hatte das Gefühl, die Trainingsstunde ging sehr schnell vorbei	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐
Ich würde das Trainingssystem gerne häufiger nutzen	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐
Ich habe das Gefühl, das heutige Training war erfolgreich	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐
Das Training war zu wenig anstrengend	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐
Das Anlegen der Orthosen empfand ich als angenehm.	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐
Den Sitz der Orthosen im Lokomat empfand ich als bequem.	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐
Ich hatte das Gefühl, die Kontrolle zu verlieren	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐
Ich fühlte mich wohl während des Eingespanntseins im Lokomat.	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐
Ich war in gutem Kontakt zum Therapeuten.	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐
Die Orthosen haben mich bei den Bewegungen gestört.	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐
Ich fühlte mich vom Therapeuten wahrgenommen.	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐
Ich wäre am liebsten schnell wieder ausgestiegen	☐	☐	☐	☐	☐	☐	☐	☐	☐	☐

Es war anstrengend das Gleichgewicht zu halten.	☐☐☐☐☐☐☐☐☐☐
Die Orthosen haben mich geschmerzt.	☐☐☐☐☐☐☐☐☐☐
Ich fühlte mich ausgeliefert.	☐☐☐☐☐☐☐☐☐☐
Ich fühlte mich im Gerät sicher.	☐☐☐☐☐☐☐☐☐☐

Beobachtungen des Prüfers, spezielle Ereignisse:

Training 2: Lokomat PRO

Ich habe die Angaben zu dieser Visite überprüft. Die Angaben sind vollständig und korrekt.

Datum und Unterschrift des Prüfarztes: _____

Lokomat Seriennummer: I__I__I

Ergebnis 6min Gehtest pre-Training:	Strecke: I__I__I__I__I m
Ergebnis Balance-Board pre-Training:	Score: I__I__I__I
Beginn Training Lokomat Pro:	Uhrzeit: I__I__I : I__I__I
Ende Training Lokomat Pro:	Uhrzeit: I__I__I : I__I__I
Ergebnis 6min Gehtest post-Training:	Strecke: I__I__I__I__I m
Ergebnis Balance-Board post-Training:	Score: I__I__I__I

Fragebogen: Patientenbefragung vor dem Training mit Lokomat Pro:

	Trifft voll zu	Trifft überhaupt nicht zu
Mein Allgemeinbefinden ist gut	☐☐☐☐☐☐☐☐☐☐	
Beschwerden	_____	

Unerwünschte Wirkungen:
Der Patient wurde über unerwünschte Wirkungen nach der Anwendung des Medical Devices gefragt.
Unerwünschte Wirkungen ☐Ja ☐Nein

Datum: I__I__I / I__I__I / I__I__I__I__I Zeit: I__I__I : I__I__I

Falls unerwünschten Wirkungen aufgetreten sind werden diese auf Seite 20 beschrieben.

Fragebogen: Patientenbefragung nach dem Training mit Lokomat Pro:

	Trifft voll zu	Trifft überhaupt nicht zu
Ich hätte gerne länger trainiert	☐☐☐☐☐☐☐☐☐☐	
Das Training war zu anstrengend	☐☐☐☐☐☐☐☐☐☐	
Das Training wurde sehr langweilig	☐☐☐☐☐☐☐☐☐☐	
Das Training hat Spass gemacht	☐☐☐☐☐☐☐☐☐☐	
Das Training sollte abwechslungsreicher sein	☐☐☐☐☐☐☐☐☐☐	
Das Training hat zu lange gedauert	☐☐☐☐☐☐☐☐☐☐	

Ich hatte das Gefühl, die Trainingsstunde ging sehr schnell vorbei	☐☐☐☐☐☐☐☐☐☐
Ich würde das Trainingssystem gerne häufiger nutzen	☐☐☐☐☐☐☐☐☐☐
Ich habe das Gefühl, das heutige Training war erfolgreich	☐☐☐☐☐☐☐☐☐☐
Das Training war zu wenig anstrengend	☐☐☐☐☐☐☐☐☐☐

Das Anlegen der Orthosen empfand ich als angenehm.	☐☐☐☐☐☐☐☐☐☐
Den Sitz der Orthosen im Lokomat empfand ich als bequem.	☐☐☐☐☐☐☐☐☐☐
Ich hatte das Gefühl, die Kontrolle zu verlieren	☐☐☐☐☐☐☐☐☐☐
Ich fühlte mich wohl während des Eingespanntseins im Lokomat.	☐☐☐☐☐☐☐☐☐☐
Ich war in gutem Kontakt zum Therapeuten.	☐☐☐☐☐☐☐☐☐☐
Die Orthosen haben mich bei den Bewegungen gestört.	☐☐☐☐☐☐☐☐☐☐
Ich fühlte mich vom Therapeuten wahrgenommen.	☐☐☐☐☐☐☐☐☐☐
Ich wäre am liebsten schnell wieder ausgestiegen	☐☐☐☐☐☐☐☐☐☐
Es war anstrengend das Gleichgewicht zu halten.	☐☐☐☐☐☐☐☐☐☐
Die Orthosen haben mich geschmerzt.	☐☐☐☐☐☐☐☐☐☐
Ich fühlte mich ausgeliefert.	☐☐☐☐☐☐☐☐☐☐
Ich fühlte mich im Gerät sicher.	☐☐☐☐☐☐☐☐☐☐

Beobachtungen des Prüfers, spezielle Ereignisse:

Training 3: Lokomat FREEW

Ich habe die Angaben zu dieser Visite überprüft. Die Angaben sind vollständig und korrekt.

Datum und Unterschrift des Prüfarztes: _____

Lokomat Seriennummer: I__I__I

Ergebnis 6min Gehtest pre-Training:	Strecke: I__I__I__I__I m
Ergebnis Balance-Board pre-Training:	Score: I__I__I__I
Beginn Training Lokomat Pro:	Uhrzeit: I__I__I : I__I__I
Ende Training Lokomat Pro:	Uhrzeit: I__I__I : I__I__I
Ergebnis 6min Gehtest post-Training:	Strecke: I__I__I__I__I m
Ergebnis Balance-Board post-Training:	Score: I__I__I__I

Fragebogen: Patientenbefragung vor dem Training mit Lokomat FREEW:

	Trifft voll zu	Trifft überhaupt nicht zu
Mein Allgemeinbefinden ist gut	☐☐☐☐☐☐☐☐☐☐	
Beschwerden	_____	

Unerwünschte Wirkungen:
Der Patient wurde über unerwünschte Wirkungen nach der Anwendung des Medical Devices gefragt.
Unerwünschte Wirkungen ☐Ja ☐Nein

Datum: I__I__I / I__I__I / I__I__I__I__I Zeit: I__I__I : I__I__I

Falls unerwünschten Wirkungen aufgetreten sind werden diese auf Seite 20 beschrieben.

Fragebogen: Patientenbefragung nach dem Training mit Lokomat FREEW:

	Trifft voll zu	Trifft überhaupt nicht zu
Ich hätte gerne länger trainiert	☐☐☐☐☐☐☐☐☐☐	
Das Training war zu anstrengend	☐☐☐☐☐☐☐☐☐☐	
Das Training wurde sehr langweilig	☐☐☐☐☐☐☐☐☐☐	
Das Training hat Spass gemacht	☐☐☐☐☐☐☐☐☐☐	
Das Training sollte abwechslungsreicher sein	☐☐☐☐☐☐☐☐☐☐	
Das Training hat zu lange gedauert	☐☐☐☐☐☐☐☐☐☐	

Ich hatte das Gefühl, die Trainingsstunde ging sehr schnell vorbei	☐☐☐☐☐☐☐☐☐☐
Ich würde das Trainingssystem gerne häufiger nutzen	☐☐☐☐☐☐☐☐☐☐
Ich habe das Gefühl, das heutige Training war erfolgreich	☐☐☐☐☐☐☐☐☐☐
Das Training war zu wenig anstrengend	☐☐☐☐☐☐☐☐☐☐
Das Anlegen der Orthosen empfand ich als angenehm.	☐☐☐☐☐☐☐☐☐☐
Den Sitz der Orthosen im Lokomat empfand ich als bequem.	☐☐☐☐☐☐☐☐☐☐
Ich hatte das Gefühl, die Kontrolle zu verlieren	☐☐☐☐☐☐☐☐☐☐
Ich fühlte mich wohl während des Eingespanntseins im Lokomat.	☐☐☐☐☐☐☐☐☐☐
Ich war in gutem Kontakt zum Therapeuten.	☐☐☐☐☐☐☐☐☐☐
Die Orthosen haben mich bei den Bewegungen gestört.	☐☐☐☐☐☐☐☐☐☐
Ich fühlte mich vom Therapeuten wahrgenommen.	☐☐☐☐☐☐☐☐☐☐
Ich wäre am liebsten schnell wieder ausgestiegen	☐☐☐☐☐☐☐☐☐☐
Es war anstrengend das Gleichgewicht zu halten.	☐☐☐☐☐☐☐☐☐☐
Die Orthosen haben mich geschmerzt.	☐☐☐☐☐☐☐☐☐☐
Ich fühlte mich ausgeliefert.	☐☐☐☐☐☐☐☐☐☐
Ich fühlte mich im Gerät sicher.	☐☐☐☐☐☐☐☐☐☐

Beobachtungen des Prüfers, spezielle Ereignisse:

Posttraining

Ich habe die Angaben zu dieser Visite überprüft. Die Angaben sind vollständig und korrekt.

Datum und Unterschrift des Prüfarztes: _____

Denken Sie bitte an das Folgende:

-Dokumentation unerwünschter Ereignisse auf Seite 51

Studienabschluss

Datum 1. Anwendung des Medical Device: | | | | | 2 | 0 | | |

Datum letze Anwendung des Medical Device: | | | | | 2 | 0 | | |

Hat der Patient die Studie vollständig durchlaufen? ☐ Ja
☐ Nein (unterer Abschnitt ausfüllen)

Ausfüllen, falls der Patient die Studie **nicht** vollständig durchlaufen hat!

Datum des vorzeitigen Studienabbruchs: | | | | | 2 | 0 | | |

Primärer Grund für den Studienabbruch:

☐ Der Patient hat die Ein- und Ausschlusskriterien nicht erfüllt.

☐ Der Patient zieht seine Einwilligung zur Teilnahme an der Studie zurück (nicht im Zusammenhang mit einem UE oder einer mangelnden Wirksamkeit der Behandlung).

☐ Der Patient ist nicht konform zum Studienprotokoll.

☐ Aufgrund eines unerwünschten Ereignisses hält es der Prüfarzt im Interesse des Patienten für notwendig, diesen aus der Studie auszuschliessen.

☐ Aufgrund eines unerwünschten Ereignisses hält es der Patient für notwendig, die Studie abzubrechen.

☐ Anderer, bitte erläutern:_____

_____

Ich habe alle Angaben überprüft. Die Angaben sind vollständig und korrekt.

Datum und Unterschrift des Prüfarztes:

	Trifft voll zu				Trifft überhaupt nicht zu
Ich hätte gerne länger trainiert	☐	☐	☐	☐	☐

Unerwünschtes Ereignis/ Adverse Event (AE) Diagnose (wenn bekannt) oder Zeichen/Symptom	**Beginn und Ende** Datum und Zeit	**Ausgang** 1= Geheilt ohne Folgeerkrankung 2= Geheilt mit Folgeerkrankung 3= Gebessert, aber noch nicht geheilt 4 = Pat. nicht mehr erreichbar	**Schweregrad** 1= Mild 2= Moderat 3= Schwer	**Kausalität zum Studiengerät** 1= kein Zusammenhang 2= Unwahrscheinlich 3= Möglich 4= Wahrscheinlich 5= Sicher	**Getroffene Massnahmen** 1= Keine 2= Dosis Therapie geändert 3= Therapie abgesetzt 4= Medikation angewendet 5= (Verlängerung der) Hospitalisation 6= Andere (Erläuterung unten)	**Serious Adverse Event (SAE)*** 0= Nein 1= Ja
AE: _____ _____	**Beginn:** ☐☐/☐☐/☐☐ Tag /Monat / Jahr ☐☐:☐☐ 24 Stunden **Ende:** ☐☐/☐☐/☐☐ Tag /Monat / Jahr ☐☐:☐☐ 24 Stunden	☐ Falls 2, 3 oder 4 Erläuterung: _____ _____ _____	☐	☐	☐ Erläuterung: _____ _____ _____	☐
AE: _____ _____	**Beginn:** ☐☐/☐☐/☐☐ Tag /Monat / Jahr ☐☐:☐☐ 24 Stunden **Ende:** ☐☐/☐☐/☐☐ Tag /Monat / Jahr ☐☐:☐☐ 24 Stunden	☐ Falls 2, 3 oder 4 Erläuterung: _____ _____ _____	☐	☐	☐ Erläuterung: _____ _____ _____	☐

Unterschrift Prüfarzt:	**Datum** ☐☐/☐☐/☐☐ Tag /Monat / Jahr	***Serious Adverse Event** (Zutreffendes bitte ankreuzen und SAE-Formblatt ausfüllen): ☐ Tödlich ☐ Lebensbedrohlich ☐ Hospitalisation oder deren Verlängerung ☐ Persistierende Behinderung ☐ Angeborene Anomalie/Geburtsdefekt ☐ Anderes relevantes medizinischen Ereignis

Appendix B

EMG Data to Study with Lokomat FreeD

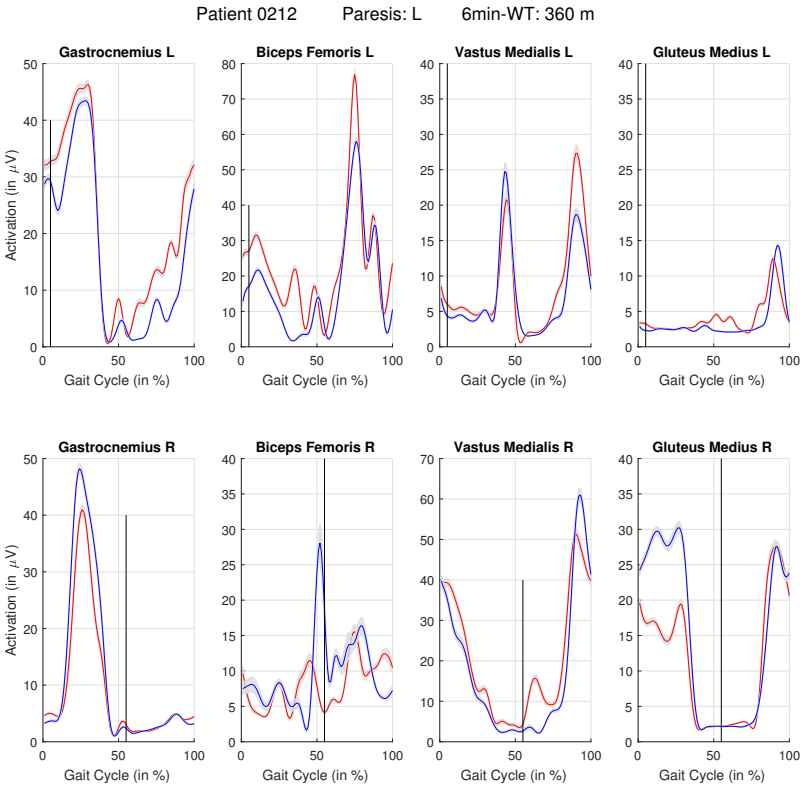


Figure B.1: EMG activation patterns from one patient over 300 gait cycles in one patient. Red lines represent muscle activation while walking with the FreeD module, blue lines without the module.

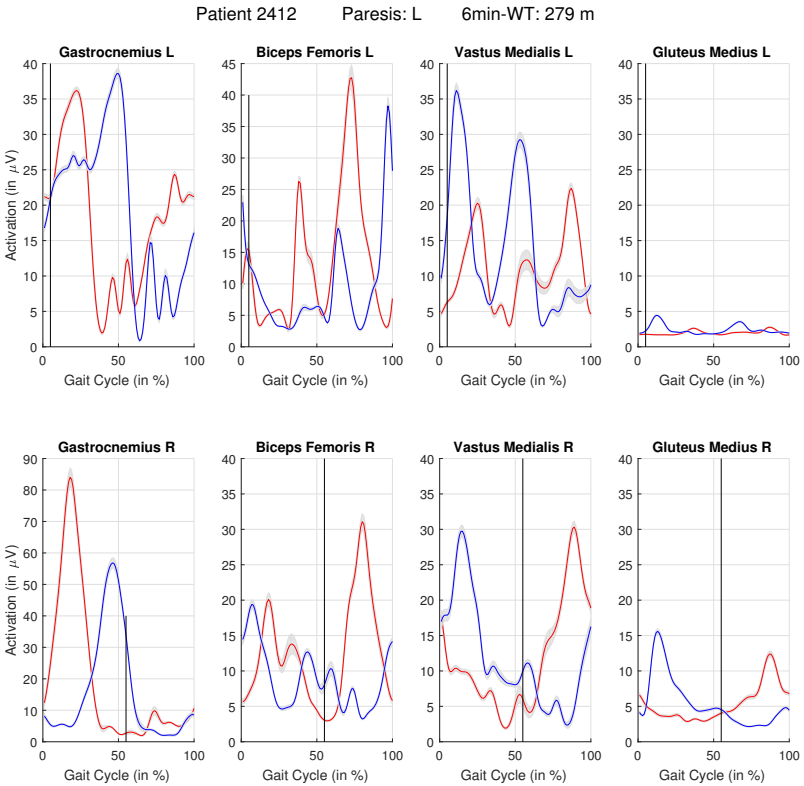


Figure B.2: EMG activation patterns from one patient over 300 gait cycles in one patient. Red lines represent muscle activation while walking with the FreeD module, blue lines without the module.

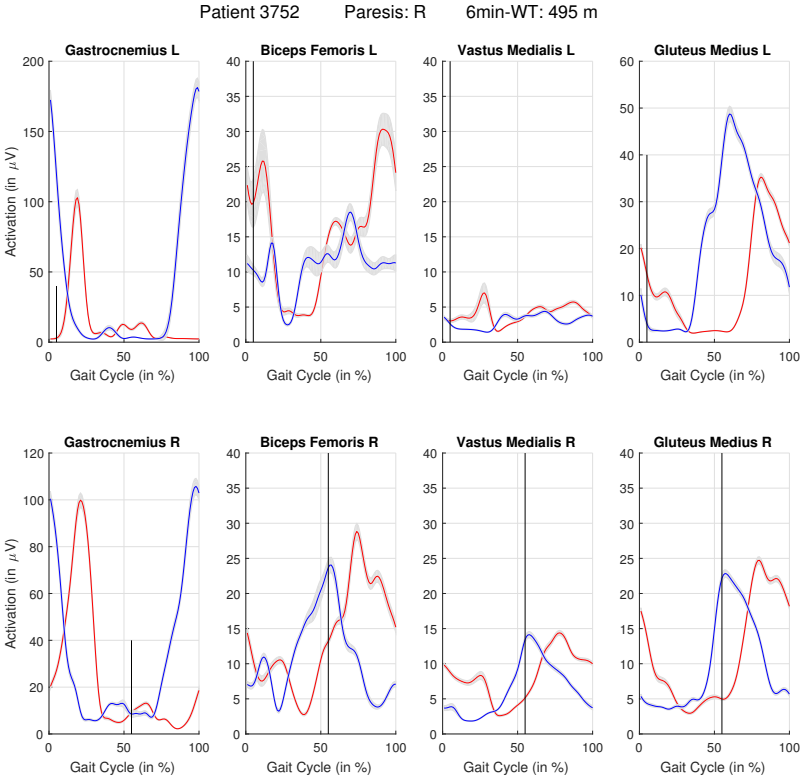


Figure B.3: EMG activation patterns from one patient over 300 gait cycles in one patient. Red lines represent muscle activation while walking with the FreeD module, blue lines without the module.

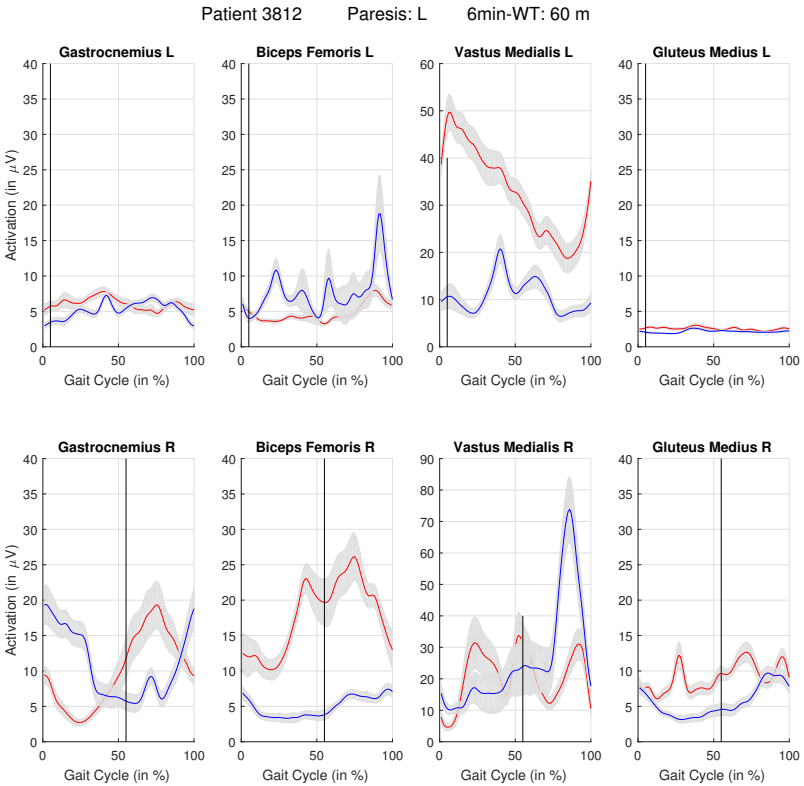


Figure B.4: EMG activation patterns from one patient over 300 gait cycles in one patient. Red lines represent muscle activation while walking with the FreeD module, blue lines without the module.

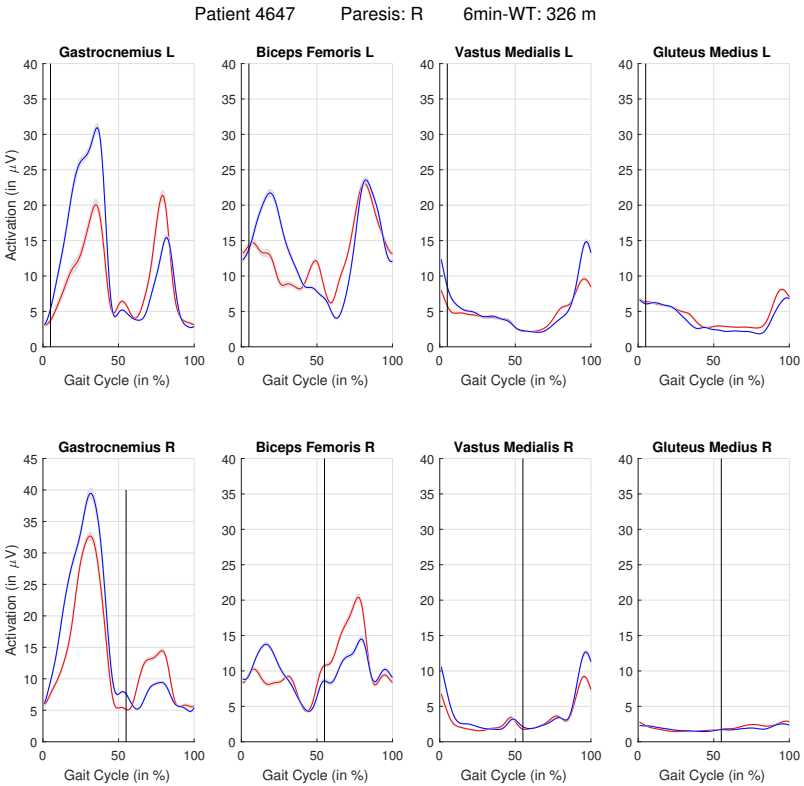


Figure B.5: EMG activation patterns from one patient over 300 gait cycles in one patient. Red lines represent muscle activation while walking with the FreeD module, blue lines without the module.

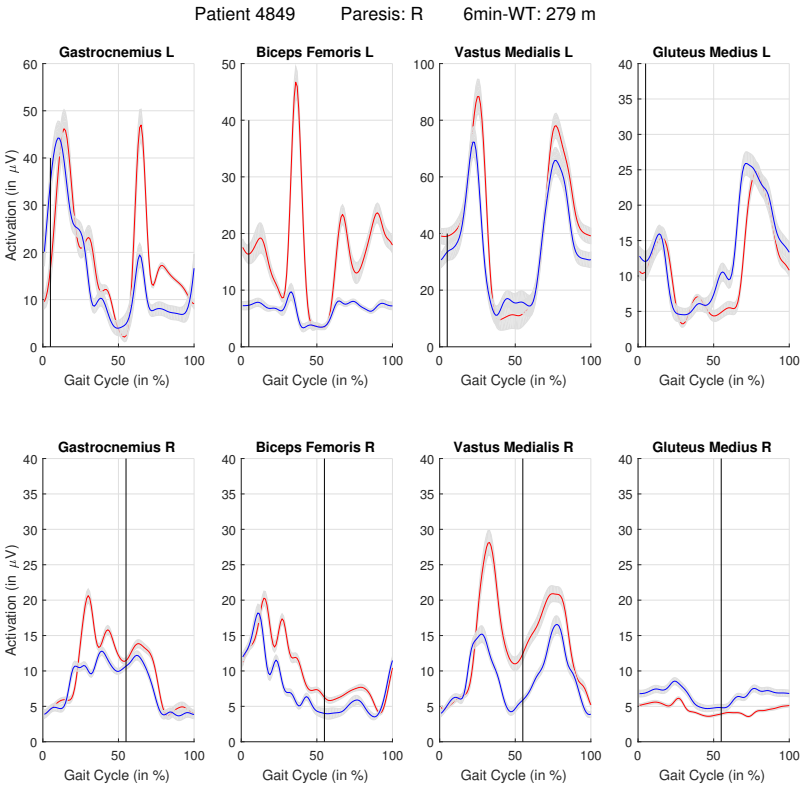


Figure B.6: EMG activation patterns from one patient over 300 gait cycles in one patient. Red lines represent muscle activation while walking with the FreeD module, blue lines without the module.

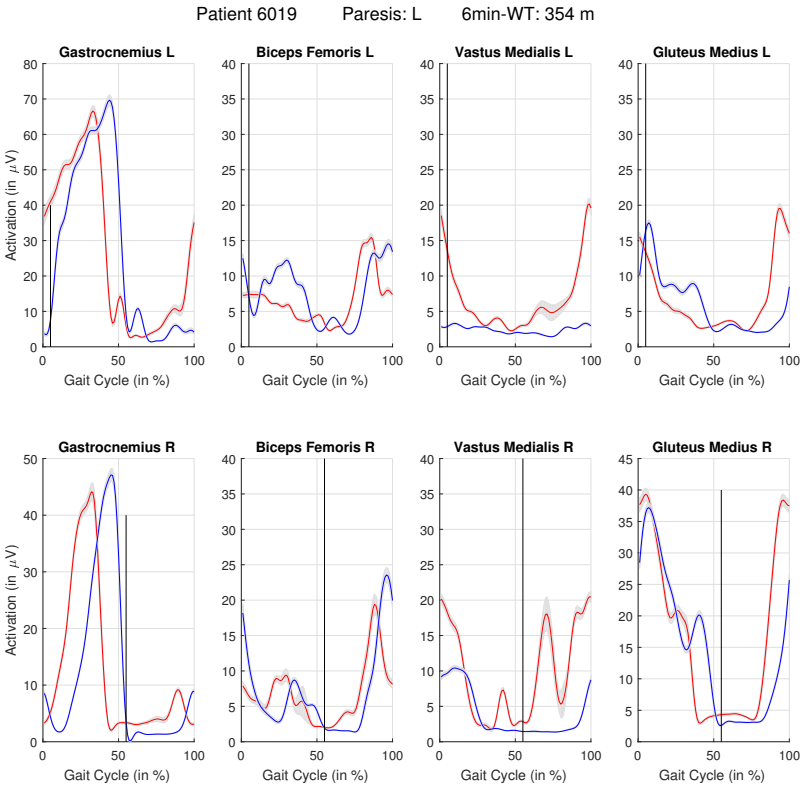


Figure B.7: EMG activation patterns from one patient over 300 gait cycles in one patient. Red lines represent muscle activation while walking with the FreeD module, blue lines without the module.

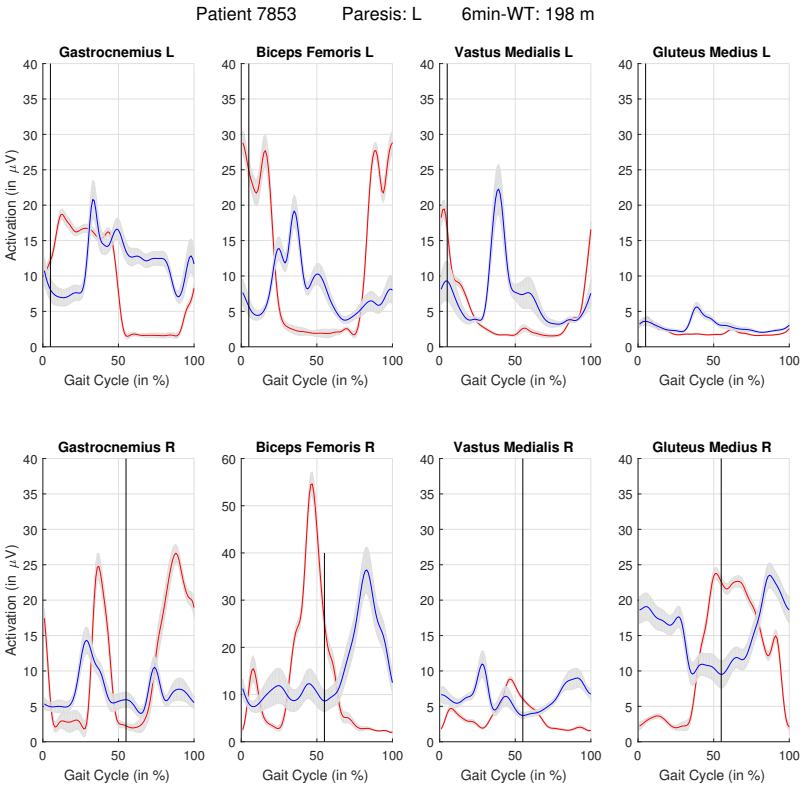


Figure B.8: EMG activation patterns from one patient over 300 gait cycles in one patient. Red lines represent muscle activation while walking with the FreeD module, blue lines without the module.

Appendix C

Overall System Design

For a proof-of-concept, the modules developed within this work (2D-BWS and MUCDA) were integrated into a base frame with a treadmill and robotic orthoses, and combined with a new actuation concept to assist foot placement (Fig. C.1). The base frame as well as the orthoses originate from a Lokomat V6 (Hocoma AG, Volketswil, CH). Knee flexion/extension remain actuated via the DC motor spindle drive of the standard Lokomat.

Hip flexion/extension and abduction/adduction is incorporated via a series-elastic actuation approach with parallel kinematics. The design approach offers the possibility to let the patient walk on a broad treadmill with self-adjusted foot placement. Two linear motors (each with 500 N peak force, NTI AG, Spreitenach, CH) are used per leg and are connected to the human leg via compression springs and a parallel structure consisting of push-pull rods and hinge joints (Fig. C.2). Synchronous movement of the linear actuators results in hip flexion/extension (Fig. C.2 b), asynchronous movements create a hip abduction/adduction movement (Fig. C.2 c). The actuators are situated posterior to the leg to allow arm swing close to the body. Fig. C.3 shows a photographic impression of the parallel actuation mechanism, and the location of all its hinge joint axes.

Motor position is measured both via internal position sensors and secondary potentiometers for calibration and redundancy. Additional potentiometers measure spring length and thereby interaction force,

Figure C.1: System overview including frame, treadmill, 2D BWS, MUCDA pelvis module, and actuated orthoses.

which is used for closed-loop control of the interaction forces.

Given that the weight of the actuators is supported by the fixed frame, and the actuators are direct-drive and also series-elastic, gravitational and inertial forces transmitted to the user due to the actuation are low.

The orthoses are connected to the pelvis module MUCDA by a spherical joint, and the weight of the orthoses is suspended by the BWS system. Although the spherical joint is not aligned with the human hip joint, we found that in practice, the linear displacement of the cuffs along the leg's longitudinal axis as a function of ab/adduction was negligible, such that there was no need for an additional

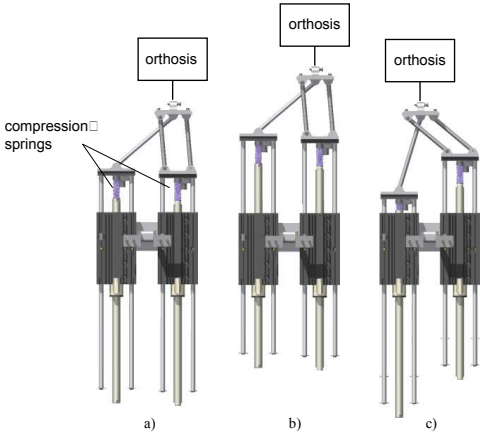


Figure C.2: Top view of one actuator unit for hip flexion/extension and abduction/adduction, in three different configurations. a) initial position of the linear actuators, b) motor position corresponding to hip flexion, c) motor position corresponding to hip abduction.

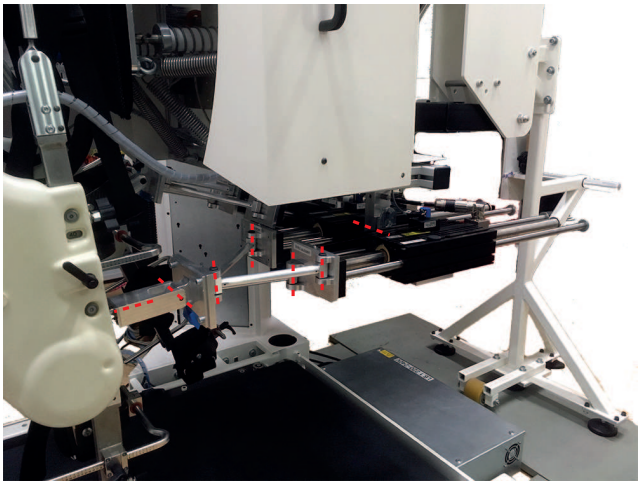


Figure C.3: The actuator unit for the hip flexion/extension and abduction/adduction. Red lines indicate hinge joint axes of the mechanism.

passive sliding joint along this axis.

The combined system is fully functional and is currently being used as a demonstrator setup.

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