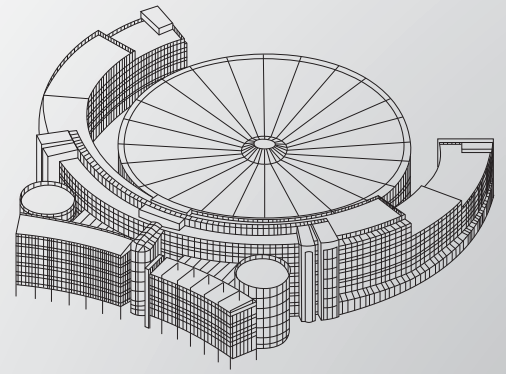


BERICHTE AUS DEM PRODUKTIONSTECHNISCHEN ZENTRUM BERLIN



Sami Hussein

**A haptic control framework for end-effector
based gait simulation interfaces and its
application in patient-adaptive rehabilitation
training**

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A haptic control framework for end-effector based gait simulation interfaces and its application in patient-adaptive rehabilitation training

vorgelegt von
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von
der Fakultät V - Verkehrs- und Maschinensysteme
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Vorwort des Autors

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Abbreviations and Symbols

Uncommon abbreviations and symbols and differing denotations are defined in the respective sections separately.

*Vectors are denoted by bold lower case symbols, e. g. **a**.*

*Matrices are denoted by bold upper case symbols, e. g. **A**.*

Abbreviations

<i>AAN</i>	<i>assist as needed</i>
<i>BWS</i>	body weight support
<i>CNS</i>	central nervous system
<i>CoM</i>	center of mass
<i>DoF</i>	degrees of freedom
<i>EMG</i>	electromyography
<i>FAC</i>	functional ambulation category
<i>FT</i>	force-torque
<i>GRF</i>	ground reaction forces
<i>GUI</i>	graphical user interface
<i>ILC</i>	iterative learning control
<i>RT</i>	realtime
<i>SCI</i>	spinal cord injured
<i>TCP</i>	tool center point

Symbols

Latin notation

<i>D</i>	damping
<i>M</i>	inertia matrix
<i>B</i>	viscose friction matrix
<i>V</i>	velocity coupling vector
<i>G</i>	gravitational force vector

C	stiffness matrix
J	Jacobian
I	unity matrix
q	generalized joint coordinates
$\dot{q}$	generalized joint velocities
$\ddot{q}$	generalized joint accelerations
x	cartesian end-effector coordinates
$\dot{x}$	cartesian end-effector velocities
$\ddot{x}$	cartesian end-effector accelerations
f	vector of forces in cartesian space
t	vector of torques in cartesian space
Z_h	passive, linear human impedance
Z_d	linear mechanical device impedance
T_{dp}	linear closed loop position controlled system dynamics
$\hat{T}_{dp}$	nominal, linear closed loop position controlled system dynamics
Z_t	target impedance – coupling controller
Z_e	virtual environment impedance

Greek notation

Δ_p	unnormalized physical uncertainty
Δ	normalized uncertainty
δ	penetration of the virtual environment by the probe
μ	static friction coefficient
τ	vector of actuator forces and torques

Kurzfassung

In Industrieländern sind neurologische Erkrankungen, insbesondere der Schlaganfall, die Hauptursache für erworbene motorische Behinderungen bei Erwachsenen. Die demographische Entwicklung lässt eine steigende Zahl von Vorfällen für die Zukunft erwarten. Eine dieser motorischen Funktionen, deren Verlust die Eigenständigkeit im Alltag stark beeinträchtigt, ist das Gehen. Ein viel versprechender Ansatz zur Verbesserung der Behandlungsergebnisse, ist roboterunterstütztes Rehabilitationstraining. Dies umfasst neuartige Geräte zur Unterstützung des Patienten sowie anspruchsvolle Regelungsstrategien für die physische Mensch-Maschine-Interaktion und die Trainingsunterstützung.

Diese Arbeit stellt die erste systematische Anwendung eines Haptik-Frameworks und so genannter *assist as needed* Trainingsparadigmen für Endeffektor-basierte Laufsimulatoren dar, die über einen natürlichen Arbeitsraum verfügen und das Gewicht des Patienten vollständig auf der unterstützenden Kinematik tragen.

Das entwickelte Haptik-Framework liefert eine einheitliche Systemarchitektur und einen Ansatz für den Reglerentwurf basierend auf robuster Regelung und identifizierten linearen Modellen. Dabei werden bekannte Impedanzgrenzen der virtuellen Umgebung im Entwurf berücksichtigt. Der resultierende Regler stabilisiert die Interaktion mit den simulierten haptischen Umgebungen. Um die Performance der haptischen Simulation zu verbessern, wird eine Adaption der Reglerparameter in Abhängigkeit des Interaktionszustands vorgenommen. Somit ist das Framework gut auf verschiedene Endeffektorgeräte portierbar und erlaubt eine flexible Synthese komplexer Trainingsszenarien aus einfachen Objektprimitiven. Die Umgebung wird durch ein Netzwerk aus parallel angeordneten Impedanzen beschrieben und ist dadurch gut in den Reglerentwurf integrierbar. Das Framework stellt die Basis zur Entwicklung und Umsetzung von nachgiebigen und patientenadaptiven Trainingsstrategien dar. Die Unterstützung wird in Form eines elliptischen Fensters in das Trainingsszenario integriert, das sich entlang einer nominellen Trainingstrajektorie bewegt. Das dadurch entstehende Unterstützungsniveau wird maßgeblich von der Steifigkeit des Fensters bestimmt. Aufbauend auf diesem Fensteransatz wird ein ILC basierter Ansatz als Beispiel einer patientenadaptiven Unterstützung vorgeschlagen. Er orientiert sich an Ansätzen, die aus der Anwendung in Exoskelettgeräten bekannt sind.

Das Framework wurde auf dem Endeffektor-basierten Laufsimulator *HapticWalker* implementiert. Zur Validierung wurden Experimente mit einem gesunden Probanden durchgeführt. Die Experimente zeigen, dass die stabile Simulation von statischen und dynamischen virtuellen Umgebungen möglich ist. Der Proband konnte im Experiment ohne Unterstützung willkürlich und aktiv Bewegungen in statischen und dynamischen Umgebungen durchführen. Im Trainingsmodus wurde der Proband erfolgreich bei der Bewegung entlang einer physiologischen Fußtrajektorie unterstützt.

Abstract

Neurological injuries, especially stroke, are the main cause for acquired motor function disabilities in adults in industrial countries and the current demographic development implies an increasing number of incidents in the future. One of those motor functions is gait, which is essential for independence in daily life. A promising approach to improve treatment outcome in rehabilitation is the application of robot-assisted training, which involves innovative devices for assisting the patients as well as sophisticated control methods for physical human machine interaction and training assistance.

The presented work is the first systematic application of a haptic control framework and *assist as needed* training paradigms to end-effector based gait simulation interfaces with a natural walking workspace, which fully carry the operator's bodyweight on the assisting kinematic structure.

The developed haptic control framework provides a unified controller architecture and a design method based on robust control and identified linear models, which takes account of known environment impedance bounds. The resulting controller ensures stable interaction with the simulated virtual environment. To improve the performance of the haptic simulation an adaptation of the controller parameters between free motion and contact state is performed. Thus the framework is easily portable to various end-effector based devices. Moreover it allows the synthesis of complex environments from simple polygonal primitives into a parallel impedance network, which is well incorporated in the control design. This framework is the basis to develop and implement compliant and patient-adaptive training strategies. The assistance is integrated into the simulated training scenario via an elliptic window object, which is moving along a nominal training trajectory. The assistance level is mainly determined by the window's stiffness. Based on the assistance window an ILC based algorithm, similar to present approaches in exoskeleton devices, is proposed as a first example for patient-adaptive gait rehabilitation training.

The proposed framework was implemented on the end-effector based, robotic walking simulator *HapticWalker*. The validation includes various experiments with one healthy subject, addressing basic stability issues, voluntary motion through static and dynamic virtual scenarios and assisted training modes. The haptic simulation of static and dynamic scenarios was shown to be stable and in the experiments the subject was able to perform the desired voluntary tasks on the haptic interface. The assisted training modes successfully guided the subject near nominal training trajectories and provided support to fulfill the training task.

1 Introduction

Neurological injuries are the main cause for acquired motor function disabilities in adults in industrial countries. From those neurological injuries stroke accounts for the most incidents by far. The annual stroke incidence is approximately 180 patients per 100.000 inhabitants in the European Community (EC). In Germany alone between 200.000 and 250.000 people suffer from stroke every year. Twenty-five percent of the surviving patients remain wheelchair-bound, and in 60% of the incidences gait velocity and endurance are considerably reduced three months after stroke [5].

Gait rehabilitation primarily aims at the restoration of an independent gait and secondarily at the improvement of walking functions to meet daily requirements. Complete or partial dependence of stroke survivors on care due to a reduced control of their motor functions impose a great challenge to the health system. Especially the impairment of gait functions leads to an increased risk of accidents and injuries. Moreover stationary therapy periods become shorter in order to reduce treatment costs, which leaves even less time to restore cognitive and motor functions and thereby often leaves patients with less ability for an independent life. Therefore the research on technical assistance systems for rehabilitation, which improve the effectiveness of rehabilitation treatment, i. e. the therapy outcome for the patients, and increases time efficiency is a crucial topic.

1.1 Daily routine in conventional clinical gait rehabilitation

In rehabilitation hospitals most commonly manual rehabilitation therapy is applied. This starts with functional exercises, e. g. physiotherapy according to Bobath, and is extended in later rehabilitation phases with task specific exercises. In gait rehabilitation mostly only one task is trained, which is floor walking. Fig. 1.1 shows examples of gait rehabilitation exercises commonly practiced in rehabilitation hospitals. During the last decade a few supported training methods were established, the most common of which is the use of a treadmill combined with bodyweight support (BWS). The patient is secured and supported by the BWS system and the treadmill forces leg motion during stance. The swing motion is still assisted by the therapists.



(a) Manual non-task-specific therapy (property of Department for neurological rehabilitation, Charité Berlin)



(b) Manual task-specific therapy (property of Department for neurological rehabilitation, Charité Berlin)



(c) Locomotion studio in the Medical Park Clinic, Berlin Humboldtmühle (property of Department for neurological rehabilitation, Charité Berlin)

Figure 1.1: Examples for manual therapy, which is most commonly provided in rehabilitation hospitals

Even using a treadmill with BWS manual therapy is a very physically straining and exhausting work for the therapist. Especially in task specific gait training exercises the patients must be supported and secured while their limb motion is assisted. This often requires more than one therapist to handle the task. Consequently the patient often receives less training intensity than he could tolerate, because the therapists tire or too few therapists are available to safely practice task specific exercises.

1.2 Motor learning and relearning – implications for robot assisted rehabilitation

Throughout the whole lifetime the human central nervous system (CNS) is able to form new neuronal connections and thereby learn and relearn skills of every sort. This fundamental property of the CNS is called plasticity. This ability of neuronal reorganisation of the CNS is the reason that any recovery from neurological injuries, such as stroke, is possible at all. Neurological rehabilitation training aims at the facilitation of that reorganization process by providing stimuli to the CNS.

According to the current state of knowledge in neurological research, the relearning process is strongly facilitated by the mechanism of afferent stimulation of the CNS. The CNS is stimulated by external proprioceptive cues via the afferent nerves. These cues involve muscle activations, external force and kinematic cues, i.e. joint angles.

The success of the rehabilitation process is based on the effect of afferent stimulation of the CNS by the activation of muscles and the rendering of forces during training. This leads

to the conclusion that high training intensity and a task specific training of movement patterns in daily life are major factors that affect the quality of rehabilitation treatment [6]. These perceptions also influenced guidelines for the rehabilitation process after stroke published in the Netherlands [7].

1.3 Potential of robot assisted rehabilitation training to improve therapy outcome

Concerning the question of the optimal rehabilitation treatment, according to known mechanisms of motor learning (see Sec. 1.2) modern concepts of motor rehabilitation favor a task-specific repetitive training, i. e. for the field of gait training: who wants to relearn walking, has to walk [8]. Besides the therapy goal to achieve more independence after neurological injuries and the tendency to reduce therapy periods, in recent times the so called evidence based paradigm becomes more popular in the medical field. It implies the systematic quantitative assessment of therapy progress and results.

Regarding both aforementioned issues the use of robots to assist rehabilitation training presents great potential for the improvement of rehabilitation treatment. The potentials for the facilitation of evidence based rehabilitation therapy by providing continuous gait analysis data during the therapy process are discussed in [9]. The motivation for the present work, however, lies in the potential improvement of therapy efficiency by robot-assisted rehabilitation treatment. Known mechanisms in motor learning described in Sec. 1.2 imply that training intensity, accuracy and task specificity of the trained movements and the patient's active contribution facilitate the outcome of therapy treatment. Compared to conventional therapy robot-assisted rehabilitation treatment is considered advantageous regarding these aforementioned issues.

Accordingly a repetitive locomotor therapy on an electromechanical gait trainer in a non-compliant training mode proved effective in restoring the ability of independent gait in non-ambulatory stroke patients [4].

In rehabilitation robot design two major concepts can be distinguished:

1. end-effector type machines, e.g. Gait Trainer GT I [10], HapticWalker [11]
2. exoskeleton type machines with a treadmill, e.g. Lokomat [12], AutoAmbulator [13], LOPES [14].

On end-effector type machines the patients' feet are fixed on footplates and the patient is secured and supported in a trunk suspension harness. While the foot motion is guided by the device, the remaining degrees of freedom (DoF) of the leg and hip are unrestricted. In case of a patient with an unstable knee joint, the physiotherapist has to manually stabilize the knee. A major advantage of this design approach is that leg and hip motion, e.g. in lateral directions, are not restricted.

In contrast on exoskeleton type devices the patients' legs and hips are fixed along a robot arm. Due to the complexity of the human lower limb kinematics in practice only few of the DoF are actively moved and controlled. The motion of the remaining joints is usually restricted to the sagittal plane. Due to their lightweight kinematic design and small drives, exoskeleton type robots with treadmill are inherently limited to the simulation of floor walking condition.

To increase the patient's active contribution and motivation during robot assisted rehabilitation training compliant and cooperative training modes are considered the key issues to work on. Such cooperative control strategies are supposed to enhance the efficiency of rehabilitation treatment on both exoskeleton and end-effector devices. Yet to date in gait rehabilitation they have been only applied to exoskeleton devices.

1.4 Scope of this work

The previous section showed the potential gains of using adjustable, robotic assistance in rehabilitation therapy. Hence this work provides a contribution towards the goal of patient cooperative and performance adaptive robot-assisted gait training paradigms on end-effector type rehabilitation devices.

The goal is to develop a haptic control framework, which allows for the synthesis and simulation of arbitrary training tasks and at the same time integrates adjustable and adaptive assistance to perform the desired target movement. This provides the foundation to develop sophisticated *assist as needed* training strategies on end-effector based gait interfaces. One important issue is to find a trade off between the need for robust interaction stability and interaction performance. Moreover a modular approach provides flexibility and portability of the framework. The developed framework is applied to the robotic walking simulator *HapticWalker* and exemplary training scenarios are implemented to evaluate its feasibility.

Though this work was mainly inspired by the potential benefits of the developed system in rehabilitation, its transfer to various other applications is possible, e. g.:

- robotic assistance systems in surgery
- partially automated assembly systems in small production series
- high-end gaming and training applications
- VR-enhanced cultural applications

The aforementioned potential applications are envisioned in more detail in Chap. 7.

2 State of the art in interaction control in robot assisted gait rehabilitation

2.1 Interaction control for haptic interfaces

This section provides an overview of the nature of haptic displays and their interaction with human operators and common approaches to control this interaction in a stable manner. A lot of work has been done on the control of haptic displays in the last two and a half decades.

Fig. 2.1 shows the components of a haptic display and their interconnection with each other. What is called the haptic display is defined as follows and incorporates several

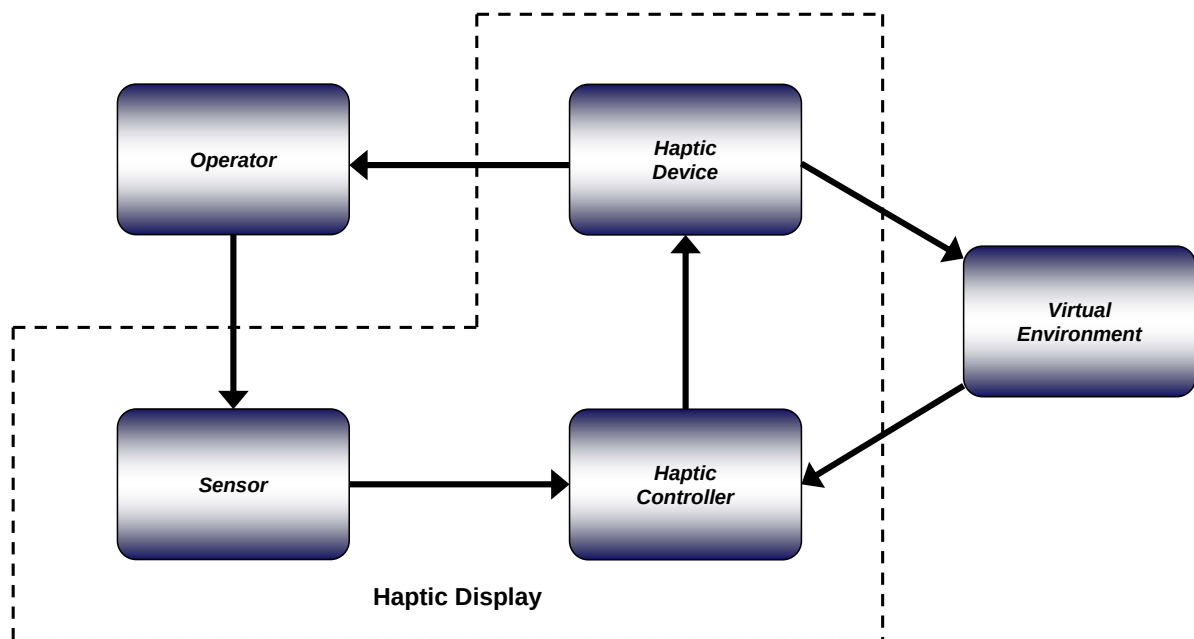


Figure 2.1: Haptic display components and their interaction

components including both software and hardware. Throughout this book following terms and definitions will be used:

Haptic display is anything lying between the operator and the virtual environment, in our case the combination of the *haptic device*, the *sensor* and the *haptic controller*.

Operator is the human user connecting to and interacting with the haptic display.

Virtual environment is a representation of the physical characteristics of virtual objects, to be rendered to the human user.

Haptic device is any kind of actuated, mostly robotic, device used to display a relation of force and interconnected motion at an interaction port to a human user.

Haptic controller or coupling controller is in general an algorithm used to generate forces and motions for the haptic device based on the mechanical characteristics of the virtual environment and the operator input

2.1.1 Haptic interaction – description and analysis

Conventional methods widely used in control engineering to analyze the behavior of dynamic feedback controlled systems provide a good basis for the analysis of haptic interaction but omit certain crucial issues emerging from the physical bilateral coupling of several systems each possibly containing its own in general unknown control loop. Several approaches, extending conventional analysis tools, were proposed for the analysis of haptic human robot interaction.

The main issues to be encountered are:

- bilateral influence between the several components of the haptic display among each other as well as the human operator,
- unpredictable behavior of the human operator and the wide range of virtual environments.

It proved to be useful to describe the chain of haptic interaction in terms of the components' interaction ports coupled to each other. These interaction ports are modeled by their impedance or admittance respectively, relating an input velocity to an output force and vice versa.

Generally the term impedance defines a transfer system relating an input flow $\Phi(t)$ to an output effort $E(t)$. The product of flow and effort is always the power $P(t) = E(t)\Phi(t)$. In haptic displays the flow is the velocity $v(t)$ and the effort the force $f(t)$ at the interaction port. Admittance is the inverse of impedance. In the LAPLACE domain the impedance and admittance are defined as

$$Z(s) = \mathcal{L} \left\{ \frac{f(t)}{v(t)} \right\} \quad and \quad (2.1)$$

$$Y(s) = Z(s)^{-1} = \mathcal{L} \left\{ \frac{v(t)}{f(t)} \right\} \quad , \quad (2.2)$$

respectively.

One way to look at such a coupled system consisting of haptic display components and human operator is to use transfer blocks, widely used for conventional control engineering problems [15, 16, 17, 18]. Fig. 2.2 shows such a block diagram of an operator coupled

to a haptic display. Here Z_H is the human operators impedance which is assumed to

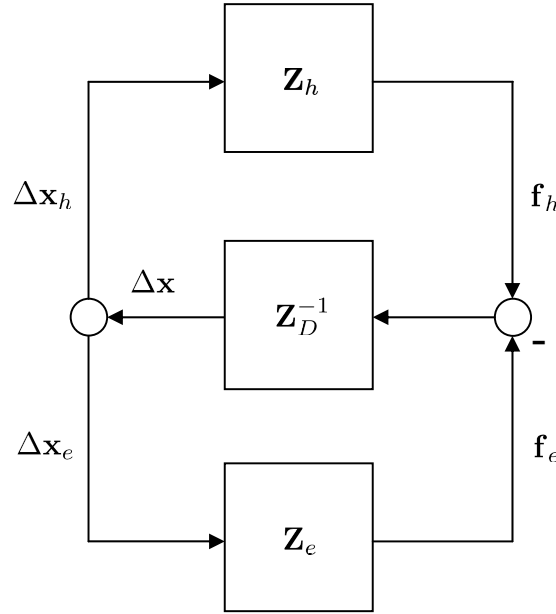


Figure 2.2: Block diagram of an operator (Z_h) interacting with a virtual environment (Z_e) through a haptic display (Z_D^{-1})

be unknown and time-varying. It is coupled in a feedback loop with the haptic device impedance Z_D which in general can be determined quite accurately and therefore is assumed as known. This loop is then coupling with a virtual environment Z_E . The virtual environment impedance is supposed to display a wide range of environmental properties.

Another approach uses a network representation consisting of one and two ports to model the haptic interaction [19]. This network representation is shown in fig. 2.3. Again Z_H represents the human operator, Z_D the device impedance and Z_E the virtual environment. The most common and promising approaches for stability analysis and controller design for haptic interfaces are described in the following.

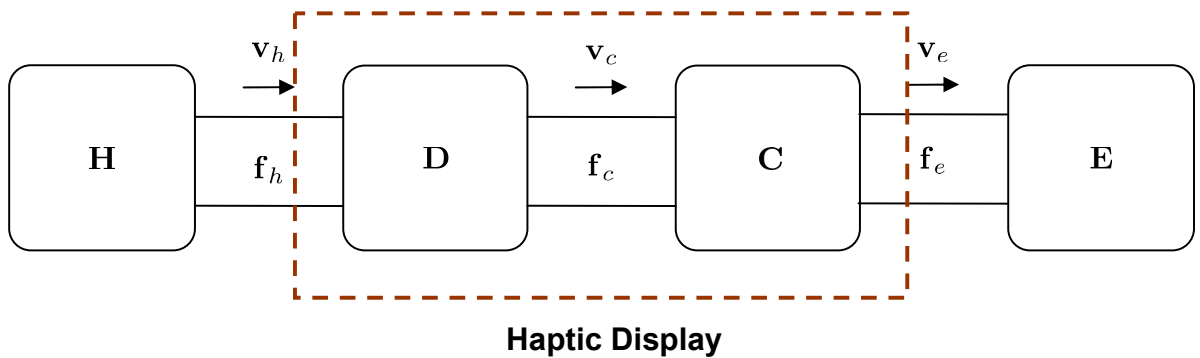


Figure 2.3: Coupled system two port network representation – human (H), device (D), coupling controller (C) and virtual environment (E)

Passivity

Based on Hogan's impedance control concept Colgate developed a toolbox for impedance control [15]. The approach uses passivity theory for the control design in interaction systems. The basic assumption is that any combination of coupled passive systems will result in a stable overall system. To achieve stable haptic interaction the interface is designed to remain passive and therefore interacts stably with any passive user and virtual environment. However this approach often results in conservative control parameters. Miller extended the approach by allowing a lack of passivity in the (virtual) environment, that is compensated by a strictly passive haptic display [20, 21]. To influence the degree of strict passivity, a coupling controller is introduced. This virtual block connects the haptic device with the virtual environment and is part of the haptic display, see Fig. 2.3. The classes of permitted virtual environments were systematically extended in the passivity approach. From linear, passive environments in the beginning, the set of environments, for which a passive coupled system could be guaranteed, extended to non-passive, delayed and non-linear environments [22, 20].

Passivity is a basic concept for stability analysis of interacting systems, that can be defined as follows:

Definition 1 *A system with input u and output y is passive if a non-negative storage function $W(x)$ exists, such that*

$$W(x(t)) \leq W(x(0)) + \int_0^t y(\tau)u(\tau)d\tau, \quad (2.3)$$

$$\forall x \in \mathbb{R}^n, \quad x, t \geq 0, \quad ,$$

with the state x .

Passivity is a sufficient property for coupled stability. If each of the coupled systems is passive, the interconnected resulting system will also be passive and therefore stable.

Robust stability

Another approach is the application of robust control design to the coupled system. A set of models, e. g. of environments [16] or human dynamics [23], is described as a combination of a nominal model and a description of uncertainty. The robust control design can also handle device model uncertainties, e. g. non-linearities. The designed controller remains stable if connecting a passive operator to an environment within the defined range. Furthermore a systematic, iterative control design algorithm can be formulated to determine the controller parameter set, using e. g. $\mathcal{H}_\infty$ synthesis.

In [17] a complementary stability criterion is provided. The robot is assumed to interact with an environment with a known nominal model and variation of parameters. The closed loop consisting of the impedance controlled robot's end-point impedance and the perturbed environment is said to be complementary stable if it satisfies robust stability. The control design method was applied to a one DoF screw driven robot for upper ex-

tremity rehabilitation using a range of known parameters for the human arm impedance in [23].

From [24, 18] two contact transition stability conditions, which also ensure coupled stability, were derived for industrial robots interacting with real and virtual bounded environments 1) from robust control theory and 2) from passivity theory as follows:

Definition 2 *Sufficient conditions for a stable contact transition of a haptic device control system under impedance control from free space to unilateral contact with any passive user, are that 2-norm/2-norm system gain of the linearized feedback system with the input-output pair $\mathbf{f}_h, \mathbf{f}_e$, i.e. ∞ -norm of the corresponding transfer function matrix is less than 1. Were $G_e(s)$ is the upper environment impedance bound and $\mathbf{Z}_t(s)^{-1}$ is the target admittance used as the virtual coupling.*

$$1) \quad [I + \mathbf{G}_e^{-1}(s)\mathbf{Z}_t(s)^{-1}]^{-1} \leq 1 \quad (2.4)$$

$$2) \quad [I + 2\mathbf{G}_e^{-1}(s)\mathbf{Z}_t(s)^{-1}]^{-1} \leq 1 \quad (2.5)$$

The applied passivity based approach also incorporates the upper bound on the environment impedance and hence is not as restrictive as the methods, which design a controlled device to be passive. Hence the robust stability based criterion obviously is the more restrictive one, as it allows twice the upper environment impedance bound compared to the passivity based condition. The above stability conditions were derived for and used on

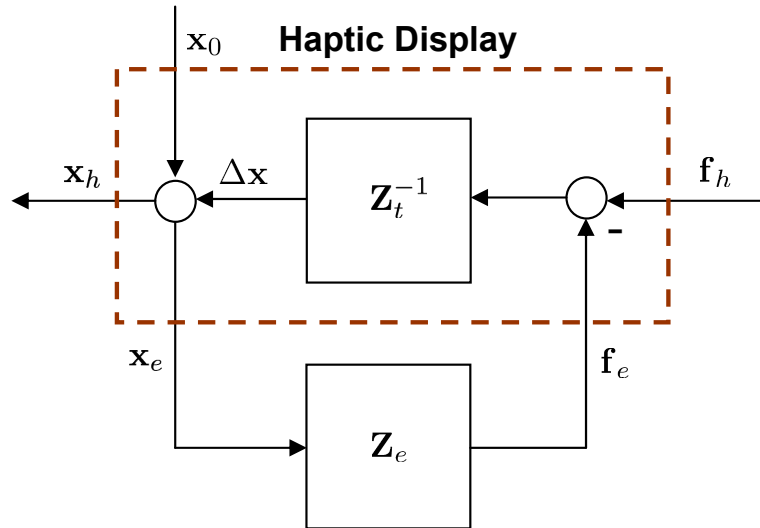


Figure 2.4: Simplified model of coupled system including haptic display ($\mathbf{Z}_t^{-1}$) and virtual environment ($\mathbf{Z}_e$)

industrial robots with an improved impedance control compensator $\mathbf{T}_p(s)^{-1}\mathbf{G}_t(s)$, where $\mathbf{T}_p(s)$ is the linear, diagonal approximation of the robot position control loop's complementary sensitivity. In this case the sensitivity function of the impedance controlled industrial robot can be substituted by the target impedance (see [18] for details). Stability conditions and control design are then based on the simplified model in Fig. 2.4.

Equations 2.4 and 2.5 result from the relation between the operator input force and the environment reaction force generated within the haptic display. The motion for operator and environment are equal in the simplified interaction model in Fig. 2.4. This clearly illustrates that the contact transition stability is defined by means of the energy transferred during contact. Intuitively the contact remains stable if the energy transferred via the haptic display into the environment is less than the energy contained in the environment reaction, which means that energy is dissipated within the haptic display.

Network and amplifier analysis techniques

Adams and Hannaford developed a Framework derived from electrical network analysis techniques for coupled haptic interaction [19]. This Framework allows for the use of stability analysis methods developed for electrical two-port amplifiers, e. g. [25]. These analyze the stability of a haptic display terminated with arbitrary passive user and environment impedances. The haptic display is thereby not restricted to passive behavior. A theoretical framework based on the so called absolute stability condition was derived for impedance and admittance displays in [26] and later applied to a three degree of freedom manipulator [27]. Absolute stability is under certain circumstances a less restrictive criterion than passivity. In [25] Llewellyn derived a necessary and sufficient criterion for the absolute stability of electrical two-port networks.

Definition 3 *A linear two-port is absolutely stable if and only if there exists no set of passive terminating one-port immittances for which the system is unstable.*

Consequently a two-port with immittance matrix P is absolutely stable if and only if

$$\begin{aligned} \operatorname{Re}(P_{11}) &\geq 0, \\ 2 \operatorname{Re}(P_{11}) \operatorname{Re}(P_{22}) &\geq |P_{12}P_{21}| + \operatorname{Re}(P_{12}P_{21}), \forall \omega \geq 0. \end{aligned} \tag{2.6}$$

Based on the same two-port representation, Ryu and Hannaford used a continuous passivity observer [28] that was combined with a passivity controller. This method presents a time domain approach for stability analysis and control design of haptic interaction systems.

Likewise adopted from the analysis of electrical transmission amplifiers and describing the user, haptic display and environment in terms of their power exchanging ports, is the approach in [29]. Therein Haddadi and Hashtrudi-Zaad adopted a graphical method from Edwards and Sinsky [30] for the analysis of teleoperation. Again the absolute stability condition is adopted and transformed into scattering domain. This representation provides means to comfortably include operator and environment impedance bounds in the stability analysis and leads to the proposed *bounded impedance absolute stability* (BIAS).

2.1.2 Virtual environments for haptic simulation

The analysis and control design techniques, described in Sec. 2.1.1, lead to a low level haptic controller providing stable interaction for certain environments and user characteristics. To complete the haptic simulation a virtual environment, also called scenario, is needed which the user is interacting with. In the following the components for such a scenario are defined and the terms will be valid throughout this thesis.

Virtual objects are the smallest units from which a scenario is composed. These objects can vary in complexity from simple cubes or spheres to complex shapes as later on shown by the tea pot example.

Probes are the interface of the operator to the virtual world. They are the projection of haptic devices pose and movement into the virtual scenario and enable interaction with the virtual objects therein. The probe can represent e. g. the operator's fingertip, whole hand, foot or a tool used by him.

Model finally terms the computational basis to mathematically describe the objects and probes. The most common methods of modeling the virtual scenario are given later in this section.

Depending on the application of the haptic display the complexity of the virtual environment is varying in wide ranges. This complexity, i. e. the degrees of freedom, of the rendering task is defined by the dimension of the virtual objects on one hand and the probe exploring the environment on the other hand. Combinations of probes and objects of different complexities and the resulting degrees of freedom of the respective rendering task are shown in Tab. 2.1. In the previously described research work focusing on the anal-

	1D (e. g.. virtual wall)	3D (e. g.. polygonal objects)
1D (point probe)	1	3
3D (polygonal probe)	3	6

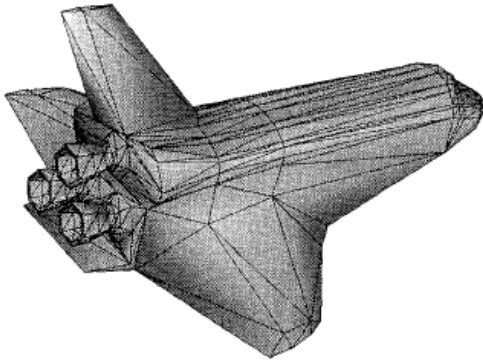
Table 2.1: Combinations of probe and virtual object complexities and the resulting degrees of freedom of the rendering task

ysis of stability and low level haptic control design mostly simple, i. e. one dimensional, rendering tasks were used to experimentally verify theoretical results. On the contrary other research activity focused on the task of rendering detailed complex environments with a wide range of tactile characteristics, e. g. in surgical training simulations. Therefore mostly very lightweight and thereby inherently transparent haptic devices are used and the range of object stiffness to be displayed is rather low. The challenge is a computationally efficient representation of complex object shapes and characteristics and the collision detection between probes and virtual objects. With the complexity of the virtual environment the computational effort of the rendering task rises. This is why restrictions

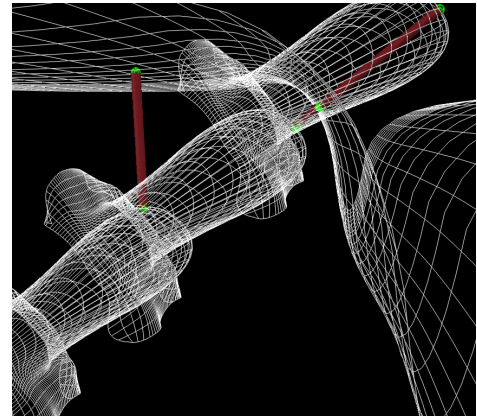
and simplifications of some sort have to be accepted. E. g. the complex virtual environments in surgery training simulation are commonly explored with point (1D) probes. A good survey on the field of haptic rendering of complex environments is provided by [31].

Depending on the requirements of the rendering task following methods for the modeling of virtual 3D-objects (also known in the field of computer graphics) can be used. The most common approaches are:

1. polygonal contour description (convex and concave)
2. spline contour description (e. g.. NURBS)
3. voxel contour description



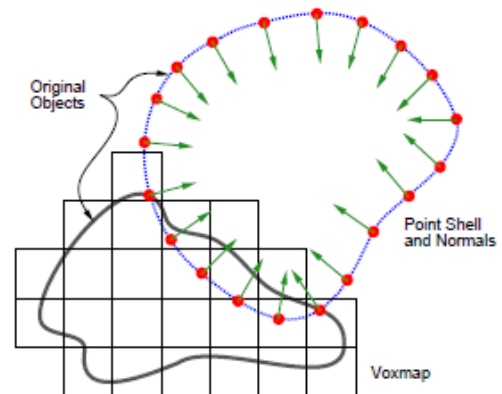
(a) Model of a complex object composed of polygons (triangle) (from [32])



(b) Interaction of objects composed of splines (from [33])



(c) Voxel map virtual object representation (from [34])



(d) Point shell of probe interacting with voxel map (from [34])

Figure 2.5: Examples for three selected types of models used for virtual environments in haptic rendering: polygon models (a), spline models (b) and voxel maps/point shells (c, d)

Polygon models are a common method to model objects in various applications, such as computer graphics and computer aided design (CAD). Here surfaces are synthesized of polygons, i. e. nodes and edges. In most applications convex poly-

topes can be used to model the respective objects. As they are widely used a set of well tested and performing algorithms for storage and collision detection exists for this class of objects. Common methods are namely linear programming, Lin-Canny-Algorithm and GJK algorithm. The GJK-Algorithm provides the best global performance ($\mathcal{O}(\log^2 n)$). For arbitrary sets of non-closed polygonal objects closed polygonal hulls can be placed around the object, or the objects are spatially partitioned. Polygonal object surfaces present a computational efficient model for haptic rendering of virtual objects [32], if the number of polygons is not too high. Furthermore polygonal object descriptions are common in computer graphics and tools for the synthesis of objects can be obtained therefrom. However polygonal contours comprise limitations regarding the constructions of smooth, round contours. Either the roundness is lacking fidelity or the number of vertices for the contour becomes extremely high neglecting the computational advantage of simple polygon models. A rather complex polygon model is shown in Fig. 2.5 a.

Spline models present an alternative to polygon models, improving the ability to design smooth, rounded object contours. Here the linear edges between nodes used in polygon models are replaced by polygonal functions satisfying a continuous transition at the nodes. This can be desirable in certain applications where continuous transition between surface segments are needed, e. g. in CAD and CAM for finite elements methods (FEM). A special, very flexible case are non-uniform rationale B-splines (NURBS). Algorithms for contact detection for spline objects are related to those used for generalized polygon models. An application of NURBS models for haptic rendering is presented in [33, 35].

Voxel models A third approach to model three dimensional objects for haptic rendering is the use of voxels, i. e. the 3D cubic equivalent to image pixels. Voxel models are applied when the simulation requires the rendering not only of surface properties but also of the distribution of physical proprioceptive properties throughout the object's volume. An example for such an application could be surgical simulations. This type of model was first used in [34] for haptic rendering of virtual objects. The complete virtual environment is therefore transformed into a so called voxel map (see Fig. 2.5 c) and the probe is described by a set of points on its surface, called point shell (see Fig. 2.5 d). To determine the interaction between probe and environment, on each point of the probe's point shell lying inside an object voxel an interaction force is acting. The sum of all these forces determines the resulting probe environment interaction. The resolution of surface structures, that can be rendered to the user depends on voxel size and the number of points on the point shell. However as voxel size is reduced and point number increased, the computational effort and memory usage for the simulation increases considerably.

Available software for models and contact detection in haptics

Below a selection of software libraries available for the modeling of virtual objects and contact detection between them is listed. Among other applications these libraries are adequate for the application in haptic rendering. Additional references for more detailed information on these tools can be found in the respective short description.

SWIFT++ is a library, written in C++, which uses polygonal objects for the construction of a virtual scene. It provides binary collision detection as well as approximate and exact computation of distances between arbitrary fixed or moving objects within the virtual scene. The distance computation and collision detection is based on a modified Lin-Canny-Algorithm [36]. To use SWIFT++ for the determination of object penetrations the nominal objects need to be reduced in size and the distance to the reduced object is determined. Together with the object shrinkage ratio this leads to a penetration distance into the nominal object. The determination of object penetrations is described in detail in the concept section Sec. 5.

(<http://gamma.cs.unc.edu/SWIFT++/>)

DEEP is an extension of the SWIFT++ library and additionally provides direct computation of object penetration depths for overlapping convex polygons. The computed penetration is the smallest distance needed to separate the respective objects.

(<http://gamma.cs.unc.edu/DEEP/>)

Extended GJK algorithm This software extends the GJK algorithm [37], which is supposed to be the fastest algorithm for 3D distance computation.

(<http://www.com.ox.ac.uk/oucl/users/stephen.cameron/distances.html>)

PIVOT is a library that features collision detection, computation of distances and penetrations and computation of contact regions of deformable objects. The library uses an image based approach to make use of the graphics acceleration hardware.

(<http://gamma.cs.unc.edu/PIVOT/>)

2.2 Devices to assist gait rehabilitation

Throughout the past 20 years several devices to assist in the rehabilitation of the human motor functions have been developed. The first of those devices addressed training of the upper extremity. After almost one decade the first devices for training support for the lower extremity, i. e. robot assisted gait training, were developed. The devices for both upper and lower extremity can be basically divided into two categories:

1. exoskeleton type devices and
2. end-effector based devices.

The presented work is dedicated to lower extremity rehabilitation. Hence the following overview is limited to gait rehabilitation devices of the aforementioned two types.

2.2.1 End-effector based devices

In end-effector based gait interfaces the patient's feet are fixed on actuated footplates while the patient is secured and supported in a trunk suspension harness. While the foot motion is guided by the device, the remaining DoF of the leg and hip are unrestricted. In case of a patient with an unstable knee joint, the physiotherapist needs to manually stabilize the knee. Further in early stages of rehabilitation treatment an additional lateral stabilization or guidance of the hip and pelvic motion may be required to prevent the patient from losing balance during training. A major advantage of this design approach is that leg and hip motion are not restricted beyond preventing an overly large lateral hip motion. The general idea of end-effector based rehabilitation is depicted in Fig. 2.6. Below

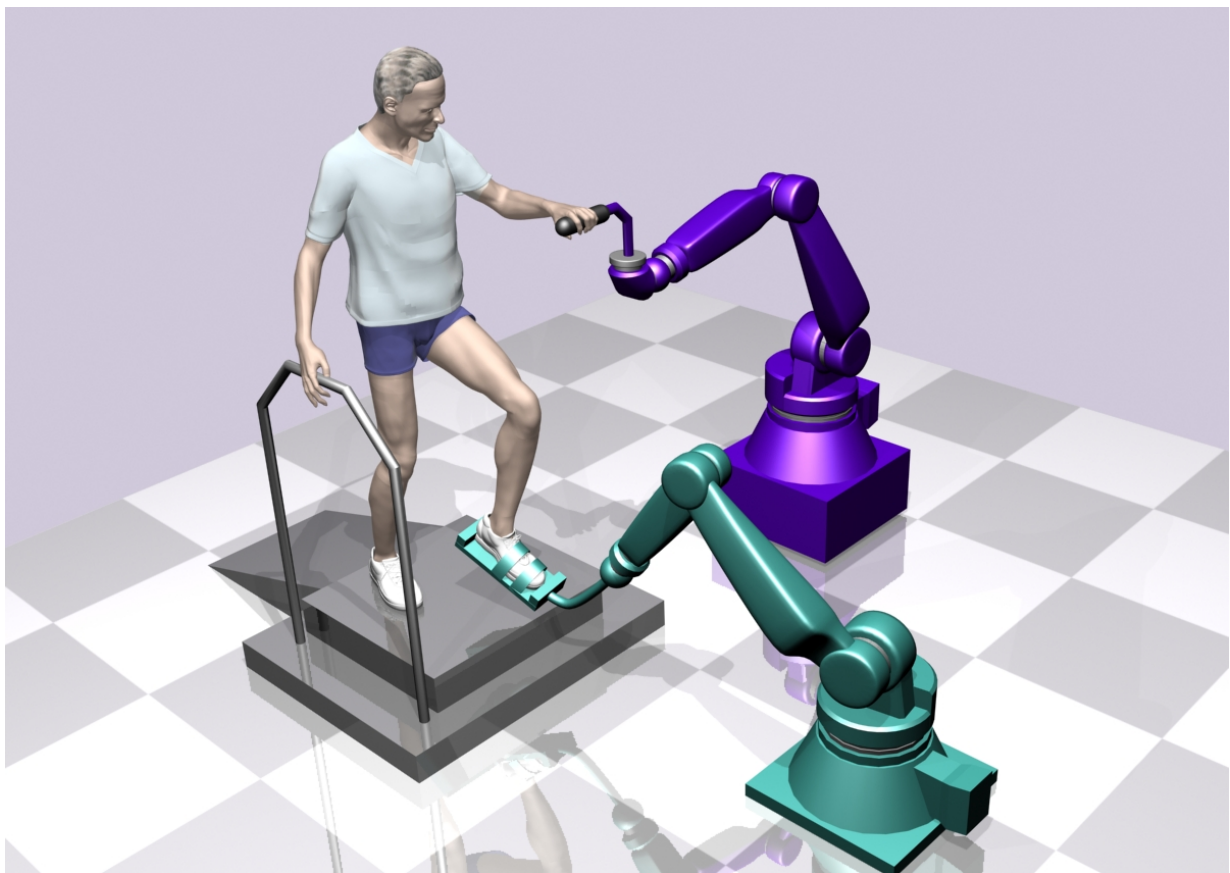


Figure 2.6: Basic idea of end-effector based robot assisted rehabilitation training

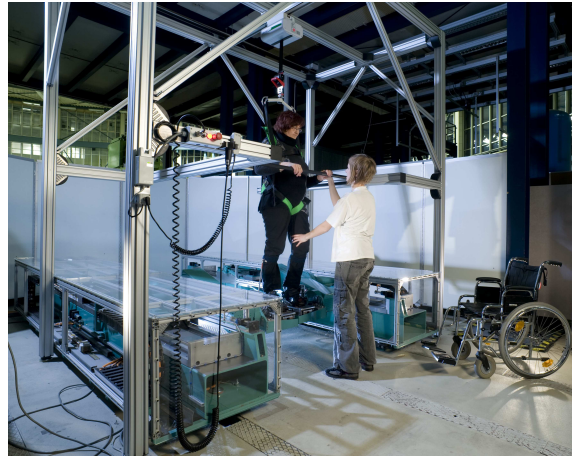
four selected end-effector based devices designed for robot assisted gait rehabilitation are described.

Gait Trainer GT I

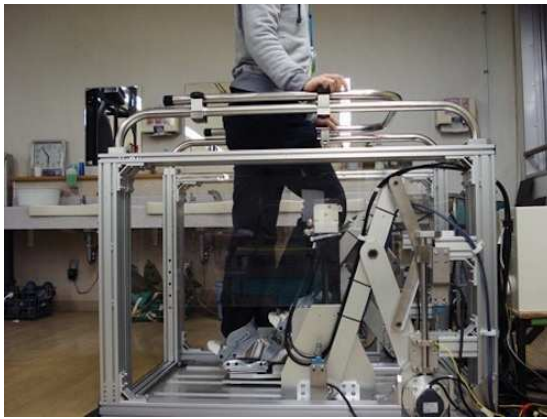
In 1999 the first version of the *Gait Trainer GT I* (Fig. 2.7 a) was developed at the department for Neurorehabilitation of Charité University Hospital in Berlin. It is an electromechanical device which implements a coupled one dimensional motion using one velocity controlled electrical drive. Thereby the patient's feet are moved at a constant speed on a mechanically predefined trajectory, which can be adjusted in length. At the same time the



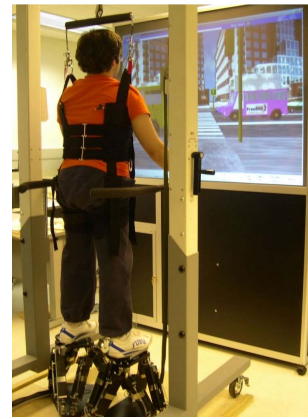
(a) *Gait Trainer GT I* (RehaStim, Germany)



(b) *HapticWalker* (Fraunhofer IPK, Germany)



(c) *Gaitmaster V* (VR Lab, University of Tsukuba, Japan)



(d) *Rutgers Mobility Simulator* (Rutgers State University of New Jersey)

Figure 2.7: End-effector based devices for robot assisted gait rehabilitation

controlled drive is used for a synchronized hip guidance and vertical movement of the patient suspension. The training movement provided by the GT-I is non-compliant, i. e. the device performs the desired movement, and thereby guides the patient, regardless of the patient's contribution. Partial substitution of manual training sessions with sessions on this device proved very beneficial for the therapy outcome in subacute stroke patients [4].

***HapticWalker*– robotic walking simulator**

The *HapticWalker* (Fig. 2.7 b) is a robotic walking simulator featuring two three DoF hybrid parallel-serial robots to guide the motion of a users feet on freely programmable trajectories within the bounds of its workspace. It was developed at the Fraunhofer Institute for production Systems and Design Technology (department of Automation Technology) in close cooperation with the department for Neurorehabilitation of Charité University Hospital in Berlin [11]. The device is the platform for the implementation and evaluation

of the presented haptic control framework and a more detailed description will be given in Sec. 5.1.

GaitMaster I-V

The *GaitMaster* series are end-effector based locomotion interfaces developed at the VR-Lab of the University of Tsukuba. In the meantime five revisions of the device were built with different degrees of freedom. The first version featured two omni-directional motion platforms whereas later devices were reduced to the simulation of motion in the sagittal plane. Fig. 2.7 c shows the latest version, the *GaitMaster V*. The motion interface of the *GaitMaster* devices is able to simulate floor walking as well as stair climbing [38]. The *GaitMaster* provides no permanent contact between foot and machine.

Rutgers Mega-Ankle Mobility Simulator

The Mobility Simulator comprises two six DoF Stewart platforms, called *Rutgers Mega-Ankle*, a bodyweight support system (Unweighting System by Biodex Medical Systems) and a screen to display virtual reality scenarios [39, 40]. Fig. 2.7 d shows the Mobility Simulator setup. The workspace of the Stewart platforms provides only a short step length of max. 0.3 m and restricts the training movements to level walking tasks. On the mobility simulator various ground characteristics can be simulated and an animated training scenario is presented to the patient. The device has passed first validation tests with healthy subjects. No clinical trials with patients of any sort were published for the *Rutgers Mega-Ankle* Mobility Simulator.

2.2.2 Exoskeleton devices

In contrast to end-effector based devices exoskeletons intend to control the complete lower limb motion of a patient. The patient's legs and hips are fixed along a robot arm. Due to the complexity of the human lower limb kinematics in practice only a small set of the DoF is actively controlled and supported. The motion of the remaining joints is usually either completely restricted or supported via passive joints and springs. Hence the motion of whole lower limb kinematics is often reduced to the sagittal plane. Due to their lightweight kinematic design and small drives, exoskeleton type robots with treadmill are inherently bound to the simulation of floor walking condition. The lightweight kinematic structure on the other hand promotes the compliant control of the devices and therefore the implementation of cooperative control and training strategies. All presented devices use two actuated degrees of freedom for each leg and a varying number of passive DoF held by springs.

Lokomat

The first exoskeleton type gait rehabilitation robot was the *Lokomat* (Fig. 2.8 a), which was originally developed at the ETH Zürich and is now commercially available from Hocoma



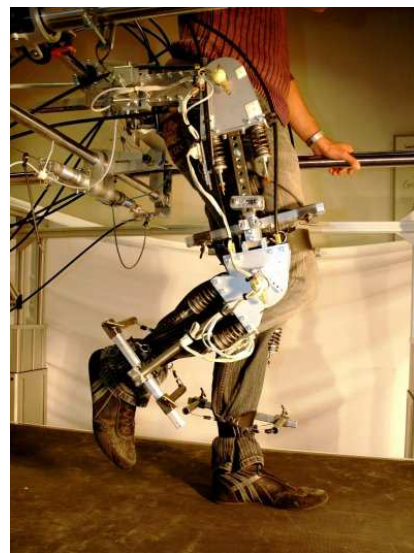
(a) Lokomat (Hocoma, Switzerland)



(b) ALEX (University of Delaware, USA)



(c) AutoAmbulator (Healthsouth, USA)



(d) LOPES (University Twente, Netherlands)

Figure 2.8: Exoskeleton type devices for robot assisted gait rehabilitation

[12]. It features two active DoF each leg acting in the sagittal plane, one at the hip joint and one knee joint. To prevent a drop foot during swing phase, a passive foot lifter is used. Like all Exoskeleton devices the *Lokomat* is based on a treadmill on which the patient is walking while guided by the robot. The patient is secured in a harness fixed on a patient lifter.

ALEX

The Active Leg EXoskeleton (Fig. 2.8 b) is a powered orthosis for the support of human gait [41, 42]. Its trunk suspension is connected to a walker supporting the device. The trunk suspension features three passive spring held DoF. The leg orthosis uses actuated DoF for the hip and knee joint in the sagittal plane and a passive spring held joint for thigh adduction and abduction. To walk in the orthosis it is combined with a treadmill.

AutoAmbulator/ReoAmbulator

The AutoAmbulator was developed by HealthSouth health care service provider (Fig. 2.8 c). The patient is secured in a bodyweight support system and walking on a treadmill. Similar to the *Lokomat* an exoskeleton is used to guide the patient's motion. It comprises two active DoF for each leg guiding the hip and knee angle in the sagittal plane. The device is commercially available as ReoAmbulator by the Motorika company.

LOPES

At the university of Twente the exoskeleton type gait rehabilitation robot *LOPES* (*LOW*er-extremity *POWER*ed *EXO*skeleton) (Fig. 2.8 d) was developed [43]. It comprises a compliant actuator concept based on Bowden cables, called Series Elastic Actuators (SEA). These compliant actuators act as sources of force. This actuation concept suggests to use impedance control in this device for the interaction between user and robot. The low inertia of *LOPES* allows for a very good transparency of the impedance display. On the other hand this property limits the ability to display stiff environments. Interaction control was designed using virtual dynamic elements (e. g. virtual springs).

2.3 Control strategies in robot assisted gait rehabilitation

In [44] a review of a wide range of control strategies developed for robot assisted therapy is presented, not all of which are relevant for the presented work.

As mentioned in section 1.2 repetitive training of physiological movements needed in daily life is considered one of the most promising approaches for an efficient, successful rehabilitation therapy. Promising factors to increase the impact of the aforementioned repetitive training are intensity and motivation. Increasing training intensity can be achieved by a higher number of repetitions on the one hand and the facilitation of active contribution of the patient during training on the other hand. Both are considered useful to improve the outcome of the rehabilitation treatment [6, 45, 4]. To achieve both of these goals, a high training intensity and facilitation of the patient's motivation, robot aided training is a method with high potential. The focus of the present thesis is to enable the development of interactive training paradigms on end-effector based devices, which is more dedicated to facilitate training intensity. Therefore this section likewise focuses on existing approaches in other gait rehabilitation robots for cooperative training. Approaches as visual feedback and influencing the patient's state of awareness and arousal, which aim at improving the patient's motivation are left out. Having this in mind several control paradigms were developed for robot assisted rehabilitation, trying to fulfill the aforementioned aspects of the training. Fig. 2.9 shows a schematic overview of these paradigms and ranks them in terms of the development order. For upper extremity rehabilitation, especially with the MIT-Manus device, many compliant and cooperative control strategies have been developed which also served as inspiration for the development



Figure 2.9: Basic control paradigms used in robot assisted gait rehabilitation

of lower extremity rehabilitation robot control methods, e. g. [46, 47, 48]. The following sections however concentrate on control methods used for the latter devices, because this thesis is dedicated to lower extremity rehabilitation.

2.3.1 Fixed trajectory training

The simplest paradigm for device assisted training is a stiff guidance of the patient by the machine on a fixed trajectory. The trajectory is adjusted to the patient's physical requirements, e. g. step length and height. To implement this paradigm the device operates in position or velocity controlled mode. For freely programmable robotic devices the trajectories are usually derived from the motion patterns of healthy human subjects. No variation e. g. of the trajectory geometry or assistance level depending on the patient's performance is done online, yet. The training speed can be adjusted by the therapist. Despite its simplicity clinical studies partially indicate the benefit on the therapy outcome using position controlled robot-assisted therapy compared to conventional therapy. Probably the largest study performed on an end-effector based device is published in [4]. A meta study published in the Cochrane database addressing both therapy with end-effector and exoskeleton devices is found in [49].

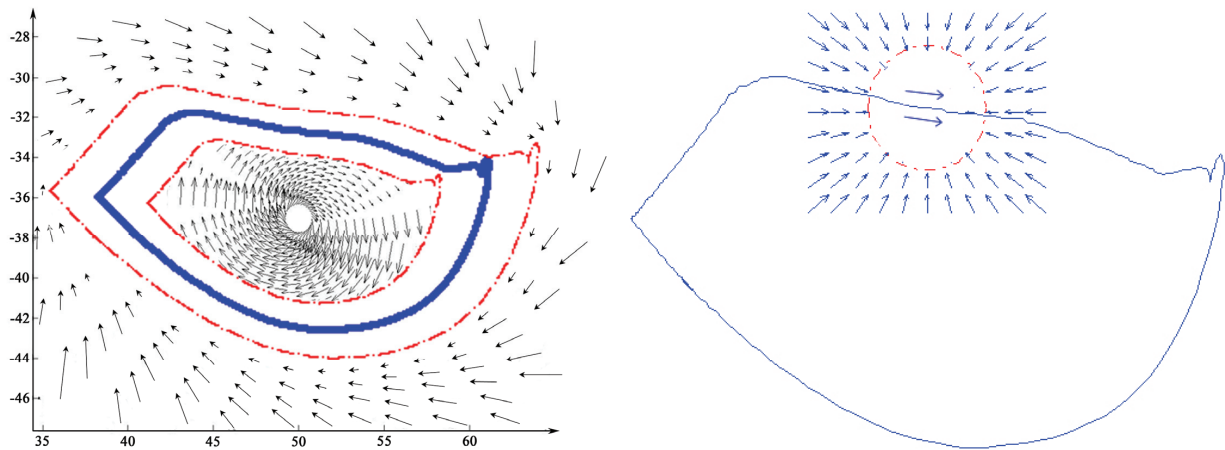
Although the stiff guidance has proven to enhance rehabilitation outcome in subacute stroke patients in several studies, several recent research results indicate that taking into account the natural inherent variability of human motor control and adapting the support provided to the patient is supposed to further improve rehabilitation outcome [50, 51, 1].

The approaches and algorithms providing compliant guidance in gait training and even more sophisticated adaptation of the level of assistance with respect to the patients training performance is commonly referred to as *assist as needed* (AAN) training.

2.3.2 Compliant training paradigms

Currently known characteristics of motor learning mechanisms show that the variability of movements and the subject's active contribution to the training are important factors in learning new movements. Even though relearning motor control is not proven to correspond exactly to learning new motor tasks, the facilitation of the patient's own contribution to the training movement related to a performance based adaptation of assistance can be considered a beneficial goal in rehabilitation training.

In [1, 52] two approaches of robot assisted AAN training are presented and tested in animal trials with mice after spinal transection. One approach uses a tolerance band about the desired end-effector trajectory of the robot, within which a constant velocity field provides gentle assistance without enforcing a constraint on inter-limb timing. Within the band the leg motion is mostly unhindered and dictated by the patient. The other control method uses a circular tolerance window, within which the leg motion is unhindered. Outside of the window a velocity field draws the leg back inside. This method allows for stepping variation but implies an inter-limb timing constraint. Fig. 2.10 shows the implementation of the aforementioned control approaches. The results from the trials



(a) AAN training with tolerance band (no interlimb timing constraint) (b) AAN training with tolerance window (with interlimb timing constraint)

Figure 2.10: Velocity fields used for non-adaptive AAN control methods in [1]

with spinally transected mice from [1, 52] indicate that any form of the proposed assist as needed strategies leads to better results than the training of a fixed trajectory. Further the window control leads to better results in step rhythmicity and shape consistency than the band approach. Therefrom a positive effect of imposing an inter-limb coordination during training is assumed.

For the *Gait Trainer GT I* a moving window approach similar to the one depicted in Fig. 2.10 b was proposed in [53, 54]. At that time sensors for the integration into the device's footplates and a PC based control system were still under development, hence the control paradigm was only tested in computer simulations.

At the ETH Zurich several approaches for compliant training on the *Lokomat* were developed. In [55] three algorithms for adaptation of the gait pattern during training were developed and tested. The first one estimates the human's voluntary force component and minimizes the interaction torques of this voluntary component based on inverse dynamics. The measured interaction forces are divided into passive and active forces. The joint angle trajectory adaptation is performed indirectly by minimizing a torque functional. This approach however is highly model dependent. The second algorithm uses a direct dynamic model and an estimation of the desired acceleration variation in the gait pattern. The last algorithm uses impedance control to directly adapt the angular trajectories.

A path control approach, comparable to the *AAN* training algorithm used in [1], was developed for the *Lokomat*, defining the virtual tunnel in the two dimensional hip-knee joint space [56]. Within the tunnel the user's motion is unrestricted. To support the motion a force field is used, which is tangential to the nominal trajectory Fig. 2.11.

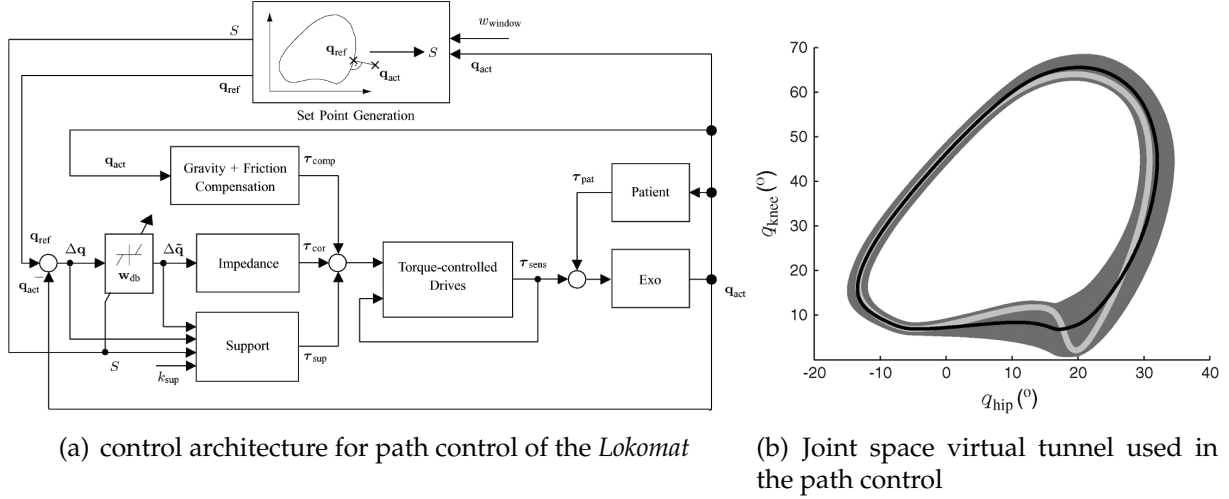


Figure 2.11: Control architecture and virtual tunnel in joint space used in the *Lokomat* [2]

The compliant control of the lightweight exoskeleton *LOPES* (University Twente) with its serial elastic actuators (SEA) is implemented as passivity based impedance control in joint space [57]. The various virtual model control structures rely on the impedance control's functionality. Virtual model control uses virtual dynamic elements attached to the device to influence the robot's behavior during the training interaction with the user [58, 59].

2.3.3 Patient cooperative training paradigms

The previous section contains control strategies, which already allow the patient a certain level of control over the training process by the compliance of the training motion. A variable assistance, which is adapted to the patient's performance, is supposed to additionally facilitate and maximize his active contribution to the practiced movement compared to the aforementioned control approaches.

Based on the gait pattern adaptation developed for the *Lokomat* in [55], an adaptation method for impedance control parameters was developed and tested [60]. The goal is here to minimize the interaction torques between the patient and the robot. This optimization problem results in the control law. No direct measure of assistance is incorporated.

During the past five years several approaches using ILC algorithms to adapt robot parameters to practicing individuals during assisted motor learning tasks were developed and tested at the University of California, Irvine. For the training support the robotic device Arthur [61] was used, which is mounted on a treadmill. The cooperative training was applied to motor adaptation tasks, i.e. learning to compensate a disturbance force during floor walking. The ILC algorithm adapts the control parameter P_{i+1} (e.g.

supporting force [50, 51] or robot impedance [62]), used to additionally compensate the disturbance, depending on performance error e caused by the disturbance and the last iteration's controller output P_i . Solving an optimization problem minimizing a cost function incorporating the error and the amount of robot support the yields following ILC algorithm

$$P_{i+1} = f \cdot P_i + g \cdot e_i \quad (2.7)$$

where f is a forgetting factor decreasing the influence of past controller actions over time and g is the error gain.

Similar performance-based adaptation approaches were also applied on several other devices. In the patient-cooperative framework, an adaptive impedance controller for the *Lokomat* was developed in which the machine impedance is increased when little subject contribution is detected, and decreased when participant effort is detected [63, 2].

Another approach of patient cooperative training is proposed in [64, 65]. Here the interaction control loop is closed via vital functions of the subject. The goal is a systematic control of the subject's mental and physical strain and state of awareness during training by changing the physical and mental stimuli imposed by the robot assisted training task.

For the active leg exoskeleton (*ALEX*) [41] a force field control scheme was developed to provide assistance as needed in gait rehabilitation. Therefore the parameters of the controller were changed with the progress of the rehabilitation training to reduce the assistance for the patient as the training proceeded. This Training mode was combined with a visual feedback of the current foot trajectory compared to the desired trajectory [42].

Control strategies that incorporate the online estimation of the user's behavioral and/or dynamic model parameters were up to now only used for upper extremity rehabilitation [66].

2.4 Clinical experience using robot assisted gait rehabilitation

The state of the art in clinical gait rehabilitation is most commonly manual therapy combined with bodyweight supported treadmill training, as stated in Sec. 1.1. The use of rehabilitation robots is still restricted to selected hospitals and the application in research and clinical studies.

From Sec. 1.2 the following factors are supposed to facilitate the positive effect of an intervention on rehabilitation outcome (e. g. [6, 67, 68]):

- training intensity
- task oriented training

- early mobilization after incidence

This section is meant to provide an overview of clinical experience using robots in clinical gait rehabilitation. First results from studies with healthy subjects are shown, which are always the first step to investigate the implication of certain interventions to support motor learning. Further results from clinical studies with patients are presented.

2.4.1 Healthy subject trials

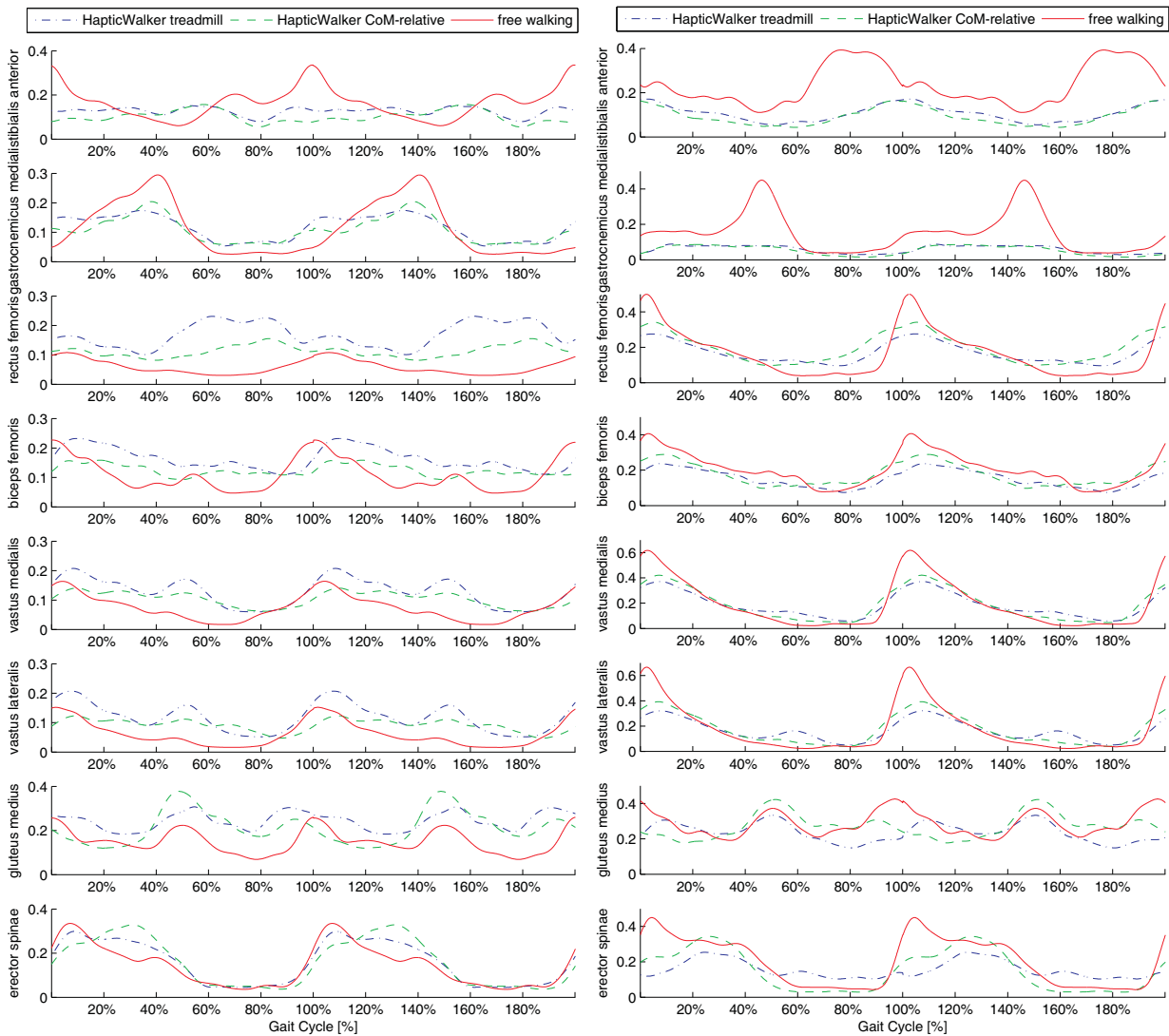


Figure 2.12: Mean EMG of 8 muscles (top to bottom: tibialis anterior, gastrocnemius medialis, rectus femoris, biceps femoris, vastus medialis, vastus lateralis, gluteus medius, erector spinae) during free and robot assisted floor walking and stair climbing [3]

First studies applying robot aided gait training on healthy subjects investigated muscle activities and ground reaction forces, which can be assumed as important afferent stimuli,

during assisted gait training. In [69] muscle activation in healthy subjects were recorded during *Lokomat* training in position controlled mode, i.e. the machine performs a fixed trajectory movement. The study showed significant differences to the control measurements during treadmill walking. The authors related the differences in muscle activation patterns mainly to the restriction of gait motion to the sagittal plane due to the *Lokomat* orthosis.

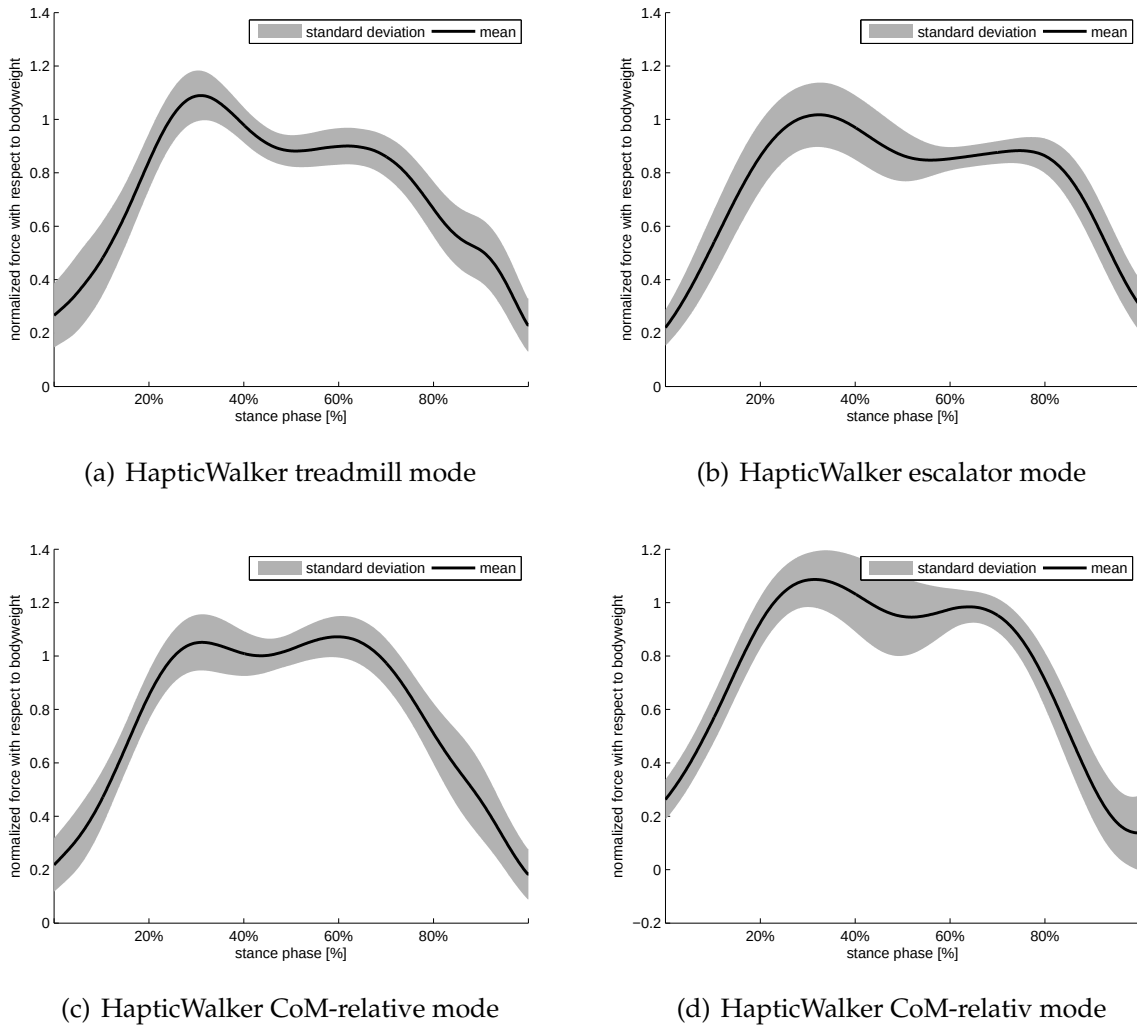


Figure 2.13: Vertical ground reaction forces during floor walking (a and c) and stair climbing (b and d) at 90 steps/min

On the end-effector type device *HapticWalker* a similar study was conducted to investigate the influence of the position controlled training mode on muscle activation [70, 3] and ground reaction forces [71, 72] in healthy subjects during floor walking and stair climbing. Measurements of the study subjects during free floor walking and stair climbing were used as reference data. As already seen in [69] for exoskeleton type devices, position controlled training on the end-effector type rehabilitation robot *HapticWalker* led to differences in muscle activations and ground reaction forces.

Fig. 2.12 shows the mean muscle activations obtained from the aforementioned study. Muscle activation timing was only slightly affected in most muscles but due to the robotic guidance an increased muscle tone could clearly be seen. The activation peaks in the thigh muscles observed during robot assisted walking were at least equal, in stair climbing even higher, compared to free walking. The shank muscles were considerably less actuated with robotic assistance than in free walking as a result of the completely guided foot motion.

For the analysis of ground reaction force (GRF) data, characteristics obtained from literature were used since measurement data was not available. As the suspension belt was used for safety reasons only and no body weight support was provided the amplitude of the force peaks can be compared to data from literature which was recorded during free floor walking and stair climbing without any assistive devices or body weight support. The observed characteristics in the presented data were comparable to those presented in [73]. Even in the evaluated position controlled operating mode the vertical GRF of healthy subjects show the characteristic maxima at heel strike and push off observed in free walking, which are referred to as haptic cues. This is an important insight as the input to the central nervous system induced by rehabilitation training is considered an essential requirement for the therapy success and the position controlled operating mode is mandatory for severely affected stroke patients in early rehabilitation. The stiff guidance of the subjects' feet resulting from the position controlled operation of the device allows the subjects an earlier loading and hence slower load transfer to the swing leg and unloading of the stance leg respectively. This leads to flattened slopes of the force graphs compared to free floor walking and stair climbing. The flattened slopes result in a delay of the force peak during heel strike and a premature force peak during push off. Fig. 2.13 depicts the GRF during floor walking and stair climbing in healthy subjects on the *HapticWalker* during fixed trajectory training mode.

2.4.2 Patient trials

Pilot trials with chronic ambulatory stroke patients were conducted for the *HapticWalker*. Single measurements were performed on two stroke patients. In [74] electromyography (EMG) data of eight relevant muscles was measured in position controlled training mode and compared to the EMGs of healthy subjects walking under the same conditions.

[75] presents a single case therapy trial with a chronic, ambulatory stroke patient one and a half years post stroke. The trial assesses the short-term impact of task specific training, i.e. floor walking and stair climbing condition, on the patient's ability to perform the respective tasks without assistance under real-life conditions. The patient was provided 17 sessions of assisted, repetitive locomotor training with each session lasting between 30 – 45 min. The sessions comprised equal contingents of floor walking, climbing upstairs and climbing downstairs. The patient's maximum walking speed improved in all three situations, however the change was not always significant. A significant improvement

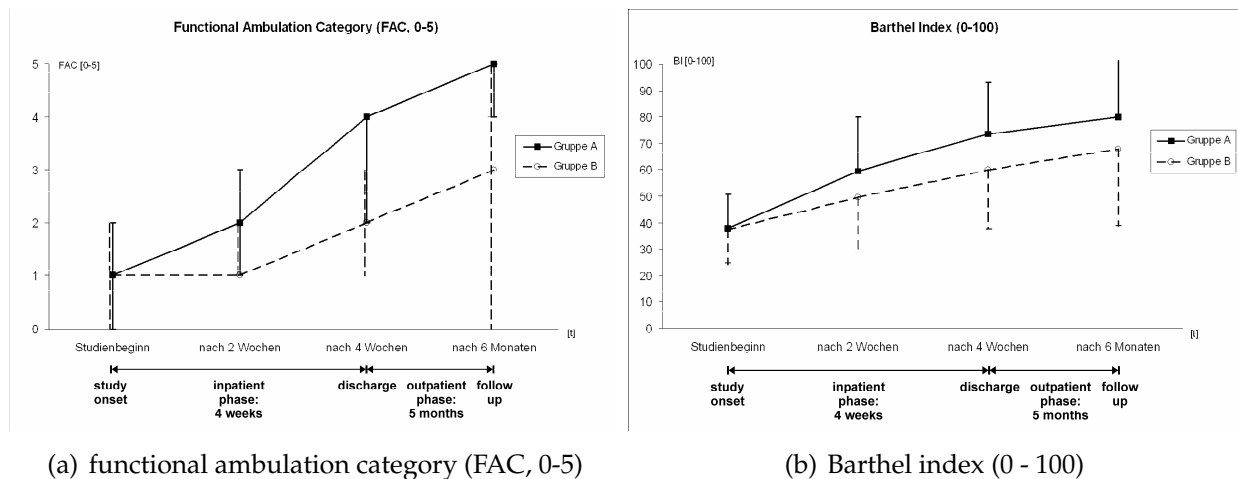


Figure 2.14: FAC and Barthel index of subacute stroke patients after repetitive locomotion training combined with physiotherapy (group A) vs. physiotherapy only (group B) (DEGAS multicenter study [4]).

was observed in the six min. walking test. One of the major improvements was the excessive reduction of the patient's circumduction movement with the affected leg.

The *HapticWalker's* ancestor the *Gait Trainer GT I* as well as the *Lokomat* are already commercially available and clinical studies were conducted comparing conventional rehabilitation therapy to robot assisted therapy in stroke patients. For the *Gait Trainer GT I* a controlled randomized multi center study (DEGAS [4]) showed a considerably higher number of ambulatory patients after *Gait Trainer GT I* therapy combined with conventional physiotherapy compared to the control group exclusively receiving physiotherapy. This indicates that for end-effector type devices even using only the position control paradigm yields a positive effect on therapy outcome. The therapy progress was assessed after two and four weeks of training and at a follow-up after six months. Fig. 2.14 shows the functional ambulation category (FAC) and Barthel index for the *Gait Trainer GT I* group and the control group from the DEGAS study. Recently a randomized controlled multi-center study concerning the effect of *Lokomat* training in ambulatory sub-acute stroke patients was conducted and the results are described in [76]. At the beginning of the treatment the *Lokomat* was operated in position controlled mode forcing the patients to follow the trajectories of the device. For patients who improved so far, that they were able to walk at maximum training speed without bodyweight support (BWS), the assistance force driving the patient on the predefined trajectory was reduced. The results however provided that the *Lokomat* group achieved even less improvement in walking speed and endurance than the control group receiving conventional physiotherapy only. The result is contradictory to the result of other studies using the same device as well as the one in [4]. A potential explanation for this is the restriction of the pelvis movement due to the devices design.

In a Cochraine review the effect of electromechanically assisted gait rehabilitation therapy after stroke was investigated [49]. The result of this meta study showed that electrome-

chanically assisted gait training can improve the outcome of rehabilitation treatment. The study did not address differences of the benefit of different devices used in the included studies.

3 Thesis objective and outline

The current situation in neurological rehabilitation and the description of ongoing research in the field of robot assisted gait rehabilitation clearly demonstrated its potential. Especially the implications for so called *assist as needed* control approaches became clear.

The presented work is the first systematic application of a haptic control framework and *assist as needed* training paradigms to end-effector based gait simulation interfaces with a natural walking workspace and fully carrying the operator's bodyweight on the device.

3.1 Objective and issues

This thesis addresses three main issues related to haptic and *assist as needed* control of end-effector based gait simulation interfaces:

- A. haptic control framework
 - A.1 low level haptic control system design
 - A.2 task level design procedure for training scenarios
- B. end-effector based *assist as needed* training paradigms
 - B.1 patient-adaptive adjustment of assistance

The main challenges arising from the aforementioned issues are detailed in the following sections.

A.1 Low level haptic control system design

Impedance and admittance control schemes are considered which have been widely used in conventional low force haptic interface implementations. The aim is to find a trade off between the following two competing design goals:

$$\text{robustness and stability} \iff \text{controller performance}$$

Robustness and stability aim at the issue of safe interaction between the device and the human user. Different from conventional haptic interfaces for the upper extremities the user cannot break contact with the haptic foot device in case of instability, because the feet need to be firmly attached to the device to provide proper assistance during training. Therefore in the presented application the preservation of stable interaction is an essential

design goal. The low level controller has to stabilize the interaction with all task level training scenarios. It further takes account of model inaccuracies and simplifications to provide robust stabilization. For control design the dynamics of the haptic display are usually assumed linear. However in general this is a strong simplification of reality.

On the other hand there is performance. In haptic interfaces performance is usually related to the accuracy and range which the task level scenarios are rendered with by the haptic display. This is also called the fidelity of the haptic simulation. As stated above, in freely programmable end-effector type devices like e.g. the *HapticWalker* no physical training environment is present. This results in the need to render the basic training environment, such as a floor or a stair case, in the haptic simulation. The haptic display is therefore required to carry the users full bodyweight and render high object impedances (stiffnesses) when it is in contact with the virtual ground. This leads to forces at the interaction port, which may range up to 1.5 kN . These forces are much higher, than in common haptic displays. To provide the necessary forces a stiff kinematic structure and powerful drives are needed which causes very high moving masses and a considerable friction in the mechanical guiding elements. Above the virtual ground a free motion with minimum impedance is desired to provide a feeling of unhindered walking, as much as possible. In this case the controller must compensate for the device's inherent impedance (i. e. inertia and friction) as far as the actuator dynamics allow. Otherwise the proprioceptive sensation in the walking simulation will be of poor fidelity. This requirement competes with the desire for stable rendering of highly stiff environments.

The challenge is to find a control scheme and parameter set suitable to realize the aforementioned trade off. Commonly used control design approaches, e.g. based on passivity theory, usually result in conservative controller parameters not suitable for the presented system. Consequently within this thesis known constraints on virtual environment parameters are incorporated into the control design procedure, which leads to less conservative controller parameters. A robust control design approach allows to take account of simplifications on device models, necessary to easily apply the control framework to various devices. Furthermore the chosen controller uses parameter adaptation to optimize fidelity both in contact with stiff environments and free motion.

A.2 Task level design procedure for training scenarios

In literature a variety of methods to model virtual environments for haptic simulations is proposed. For the target application the virtual environment model needs to allow the simulation of concave and convex contours with basic surface textures. For haptic simulations the virtual environment must be computed and rendered in realtime. In this application the task level simulation is intended to run at the same update rate as the low level controller, which means 500 Hz or higher. Hence the computational cost, and consequently the complexity of the environment model, needs to be kept as low as possible. Additionally the environment must be modular to enable a flexible synthesis of arbitrary training tasks. This is achieved by using object primitives from which the training envi-

ronment is synthesized by superposition. The constructed training scenario can consist of static and moving objects. The adjustable training support is one of these separate entities and hence integrated into the virtual environment. As part of this thesis a set of primitives is defined, which allows for the synthesis of appropriate gait training scenarios. Furthermore the range of necessary mechanical characteristics is determined, which is needed as input to the control design procedure.

The intended modular nature of the developed framework as described in the previous section raises the question of interaction stability with an arbitrary combination of given primitives. The interaction stability with superimposed primitives must be retained under all circumstances.

B.1 Patient-adaptive adjustment of assistance

The developed haptic control framework is the foundation for the development of patient-adaptive training assistance. The goal is to provide integration concepts for variable adaptive assistance of movement. An adaptation algorithm based on ILC is implemented and evaluated. The adaptation scheme resembles those also applied in exoskeleton devices. As mentioned before, training assistance is integrated into the framework as dynamic objects, called windows. Compliant task-specific assistance is achieved by the windows moving along a predefined training trajectory. The windows allow free motion close to their center and present a springlike behavior outside. To adapt the assistance level according to the user's performance, i.e. in the case of gait rehabilitation the patient's ability to actively contribute to the training, the window parameters are adjusted. The performance is determined from the kinematic errors during the assisted stepping movements.

Evaluation of the developed haptic control framework and training modes

The evaluation of the developed haptic control framework comprises two stages: simulations and experimental investigation. The proposed framework is applied to the robotic walking simulator *HapticWalker*. The simulations are carried out in MATLAB® and use a detailed dynamic model of the *HapticWalker*. Experimental results are obtained from trials with healthy subjects on the machine. Gait characteristics under various training modes are investigated and compared.

3.2 Outline

This thesis provides a systematic haptic control framework for end-effector based gait simulation interfaces, which allows the development of patient-adaptive *assist as needed* algorithms for this class of devices for the first time.

The goal is to establish foundations for the development of the aforementioned sophisticated *assist as needed* algorithms. Within the scope of the thesis further a patient-adaptive example training scenario is developed. It is derived from *assist as needed* approaches previously used in other devices [42, 50, 51, 63]. The parameters used in the evaluation are determined empirically. The development of new adaptation algorithms or the model-based selection and optimization of parameters would presume extensive collection of data from healthy subjects and patients as well as more detailed exploration of the mechanisms of motor learning and relearning. This however is beyond the scope of this thesis. Likewise beyond this thesis is the derivation of novel, specialized and clinically relevant training strategies based on the developed control framework.

4 Concept: Haptic control framework for patient-adaptive gait training

The design and implementation of patient-adaptive training modes on end-effector based gait interfaces is based on the haptic simulation of virtual training environments and assistance of the motion along a task-specific training trajectory. The goal is to derive a unified haptic control framework, which allows for the flexible synthesis of virtual training scenarios, integrates task-specific assistance and can be applied for the whole class of end-effector devices. The concept includes:

1. selection of the basic force feedback control scheme,
2. interaction control design procedure,
3. synthesis and simulation of virtual training environments,
4. integration of adjustable training assistance into the framework and
5. a basic approach for performance-based assistance adaptation.

Linear models and control design methods are used, which avoids the need for complex, non-linear analytical device models. Working with simplified models, identified from measurement data, makes the framework more flexible for the use with varying devices.

For the definition of the nomenclature used for the description of haptic interfaces and interaction refer to Sec. 2.1.

4.1 Target application properties and requirements

This section summarizes the system requirements, which the concept needs to take account of, arising from the target application. The first part in Sec. 4.1.1 is about the class of devices being addressed. Sec. 4.1.2 considers the low level control system design and the influence of device characteristics on it. In Sec. 4.1.3 the requirements for the task level design and rendering of training scenarios are summarized.

4.1.1 Target devices

The presented concept aims at delivering a framework applicable for the whole class of end-effector based gait rehabilitation devices. The basic idea of these devices is outlined in Sec. 2.2.1. Lower extremity walking simulation devices based on an end-effector design can vary e. g. in their workspace, the number of DoF or their range of dynamics. They share however common properties, crucial for the presented haptic control framework.

To implement rehabilitation training the patient is secured via a bodyweight support system and his or her feet are fixed on and guided by the end-effectors, i. e. the footplates, of the device. Consequently for the presented concept it is assumed, that the patient's bodyweight is carried during stance by the same kinematic structure delivering the guidance of the unloaded leg during swing. As a result the kinematic structure and device's actuators must be strong enough to support the patient's dynamic loads caused by impact and weight shifting during walking movements. These loads can be estimated from literature, e. g. [77, 73], and depend on the maximum bodyweight allowed for the user. Therefore end-effector based gait rehabilitation devices generally feature powerful high force actuators and a stiff kinematic structure resulting in high device inertia and friction.

Further the presence of an underlying position or velocity control loop is assumed for the target devices. This assumption is based on the configuration known from the most prominent end-effector based devices *Gait Trainer GT I* and *HapticWalker*.

4.1.2 Low level haptic controller design and rendering

The low level control scheme for the haptics framework is intended for various haptic foot devices varying in kinematics, complexity (number of DoF) and dynamics. This leads to the desire for a unified control system architecture, including the low level controller as well as the task level haptic simulation, for all of these devices. For practical reasons the design of device specific control parameters is desired to rely on linearized dynamics and measured device behavior, which makes the derivation and online computation of complex non-linear dynamic models unnecessary. The controller is rather desired to be robustly stable to cope with unmodeled system dynamics. The selection of the control scheme will be elaborated on in Sec. 4.2.1.

For the haptic control framework, which is developed in this thesis, the target device characteristics imply that the haptic simulation needs to be able to render stiff objects bearing the patient's bodyweight, i. e. also high interaction forces, as well as compliant guidance objects with low stiffness and even free motion. Despite the variety of desired haptic properties the fidelity of the simulation should be sufficient to render the feeling of free voluntary motion and stiff impact of the foot on the ground.

4.1.3 Task level environment synthesis and training support

The task level rendering system holds the training scenario and elements for training assistance according to the training task. It provides data regarding the interaction state between the probes and the virtual environment, which is used by the low level control system to compute interaction forces and generate the haptic device output to the user. Sec. 1.2 reveals factors, which from the current state of knowledge have positive impact on therapy outcome. Task specific training is one of them. As a result it is desired to provide various individualized training tasks to the patient. Consequently the training scenarios provided by the task level system should be modular and easily reconfigurable. For continuous training movements static as well as dynamic, i. e. moving, objects are needed within the scenario.

To facilitate the cooperation with clinicians in developing new training concepts and strategies, the framework should provide the basic interfaces to later add an easy-to-use user interface for environment design.

4.2 Control design concept for end-effector based gait interfaces

The following section deals with the selection, development and design of a haptic control system, which ensures stable and robust operation of a haptic interface for walking simulation coping with a wide range of user characteristics and training environments. Therefore the constraints imposed by the considered class of devices as well as the range of desirable training environment characteristics as described in Sec. 4.1 are defined and taken into account.

4.2.1 Selection of control scheme

The first step in control design is always the selection of a suitable control scheme. In the presented case the intention is to select a uniform control structure fit for a whole class of devices, i. e. electromechanical and robotic end-effector based gait rehabilitation devices. A good overview of control methods for robot contact tasks and interaction is given in [18, 78]. In the field of haptics two basic types of control schemes are commonly used (as stated e. g. in [79]):

- A. impedance display
- B. admittance display

The definition of impedance and admittance causalities is provided in Sec. 2.1.1 eq. 2.2. Additional to this definition commonly found in literature, the terms impedance and

admittance will be also used for systems relating positions and forces in this work, to achieve a unified description for velocity and position controlled devices.

In the following the advantages and disadvantages of aforementioned implementation schemes will be discussed in light of the targeted application, to derive a uniform control concept fit for the display of haptic training environments and providing variable support to a patient during assisted gait training on end-effector based devices.

A Impedance display

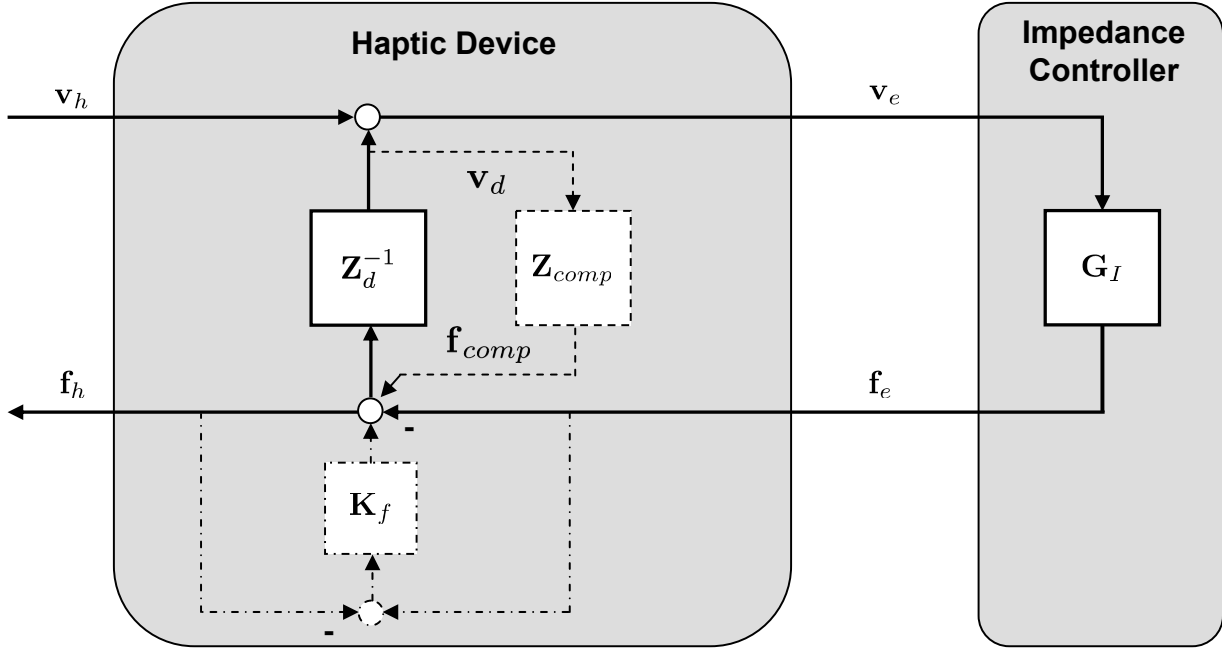


Figure 4.1: Impedance display implementation of a haptic display (solid components). Possible extensions are model feedforward (dashed components) and closed loop force feedback (dash-dotted components)

In an impedance controlled haptic device, i. e. an impedance display, position or velocity deviations are measured as inputs and an output force is displayed to the operator (Fig. 4.1).

According to the generic impedance display scheme in Fig. 4.1, the user has to cause a position or velocity deviation, in order for the impedance display to output a force. It can be easily shown (as e. g. in [26, 27]), that the minimum realizable impedance is the device's mechanical impedance stated in its non-linear form in eq. 4.1. This can be linearized for control design and analysis according to eq. 4.4. Therefore this control scheme is only useful for devices with low inertia and friction. Otherwise the haptic display provides poor transparency for free motion simulation.

An enhancement to the simple impedance control is the compensation of the haptic device's dynamics, i. e. inertia and friction, via feed-forward, also referred to as computed torque control. This augmentation of an impedance display is depicted by the dashed compensation block Z_{comp} in Fig. 4.1. To apply dynamic compensation to the impedance

display, an accurate dynamic model of the haptic device is necessary. Unavoidable modeling errors and delay between determination of the device position, velocity and acceleration and the application of the feed-forward torques, result in erroneous feed-forward computed torque values. Therefore it is never possible to completely compensate for the device dynamics. In the worst case erroneous model based feedforward compensation can lead to instabilities. Another possibility to enhance fidelity and compensate for gravitational effects is to close an underlying force-feedback loop ensuring the realization of the desired setpoint force at the interaction port. This is shown by the dash dotted force control block K_f in Fig. 4.1. Increasing the force feedback gain reduces the accuracy of the realized impedance and therefore also the lowest achievable impedance. High feedback gains however can lead to stability issues. The maximum gain is limited by the drive torque control characteristics, e. g. sampling rate and mechanical parameters. A major advantage of the previous two impedance control implementations is, that no force measurement at the interaction port is needed. On the contrary the implementation with force feedback requires operator force measurement.

The impedance display requires direct access to the force/torque control loops of the haptic device's actuators. Yet most commercially available robots as well as servo actuators and controllers used to build custom devices, ship with integrated velocity or position controllers, which provide superior position tracking performance and robust disturbance rejection. The impedance control strategy introduced above is not able to make use of an integrated position controller.

B Admittance display

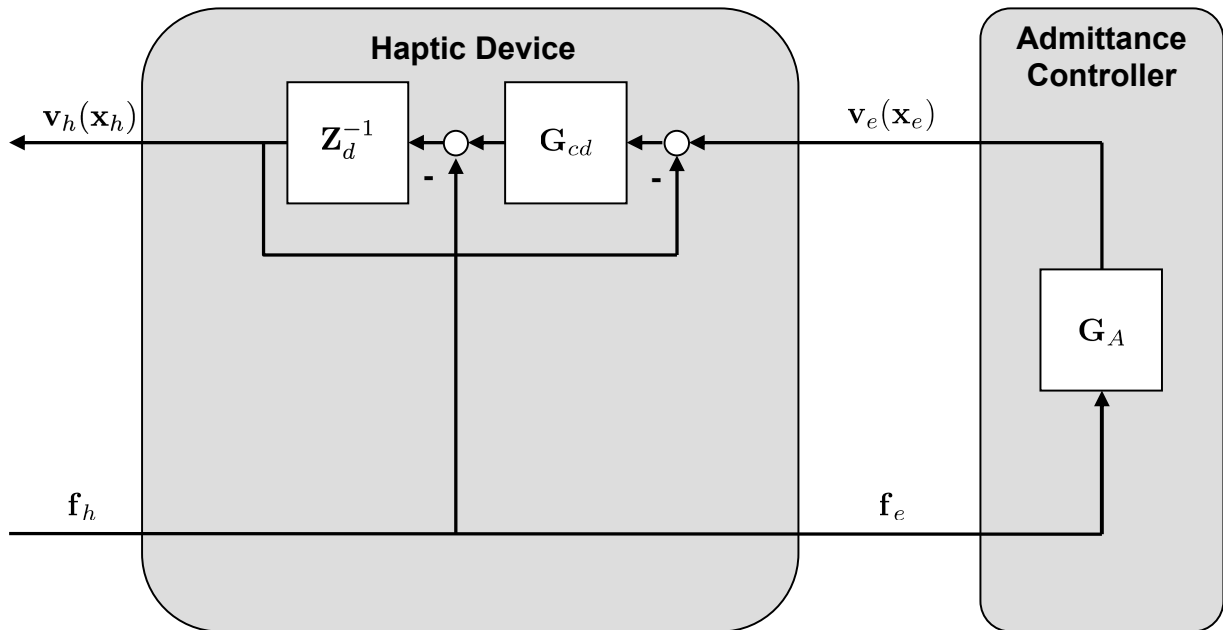


Figure 4.2: Admittance display implementation of a haptic display

In a haptic device under admittance control, i. e. an admittance display, the user's input forces at the interaction port are measured and a position deviation of the tool center

point (TCP) corresponding to the displayed admittance model G_A is generated. Therefore in kinesthetic devices the measurement of forces or force proportional data, such as motor currents, is mandatory. For a diagram of a general admittance display refer to Fig. 4.2. The device block diagram represents a closed loop velocity or position controlled device and G_{cd} the respective controller. In a specific application, e. g. an industrial robot with cascaded position controller, the position control loop might look different and more complex.

In contrast to impedance displays an admittance display control structure inherently makes use of a device's position or velocity control loop. Using a well tuned position control loop the control structure is thereby able to effectively compensate for the device inherent inertia and friction up to the bandwidth of the position control loop. Further due to the high gains of the underlying position control a high stiffness in the simulated environment can be achieved.

Selection of a uniform control structure

As pointed out in Sec. 4.1 the dominant characteristics of the present application are very high interaction forces and consequently a high mechanical device impedance. Further the application requires very high environment stiffnesses to be rendered to the operator. Having in mind the results of the previous section's comparison between impedance and admittance display haptic interface structure, either an impedance display with either model based or force feedback inertia and friction compensation (closed loop impedance display) or admittance display structure could be used. A simple open loop impedance display solution is not feasible due to the device's high mechanical impedance, which can not be compensated with that approach. However for the class of end-effector devices it is assumed, that in general a well tuned, underlying velocity or position control loop is present. To benefit from this underlying control loops, and present, well proven robot controller software, which already utilizes joint level position or velocity control, an admittance type control structure is favored for the haptic control design for end-effector based gait rehabilitation devices.

The presented application does not require overly precise impedance and nominal position tracking compared to industrial robot contact tasks considered e. g. in [18]. Neither needs subtle tactile information to be transferred via the interaction port to the human user during walking simulation. The crucial challenge is rather to achieve good performance for contact transition between free motion and medium stiff assistance elements as well as extremely stiff environment objects, such as a virtual floor, stairs or undeformable obstacles.

To achieve a framework which allows for a practical, straight forward design of basic control parameters for end-effector gait rehabilitation devices, the uniform control system structure shown in Fig. 4.3 suitable for a whole class of end-effector devices, which comprise position or velocity control loops, is chosen. Using a coupling controller theoretically allows for an arbitrary combination of impedance/admittance display and

impedance/admittance environments. For the presented application the chosen control architecture as depicted in Fig. 4.3 consists of an admittance display connected to an impedance type environment. This combination is useful, as the environments to be displayed comprise position dependent mechanical characteristics and in the selected approach position information is implicitly transferred to the virtual environment. Fig. 4.3 already shows a linearized representation of the real system. The goal is to use linear control design techniques to determine the parameters of the coupling controller, which is used to ensure interaction stability between operator, haptic display and virtual environment. Therefore later the non-linear closed loop device dynamics are replaced by a linear approximation and associated model uncertainty. The presented concept assumes

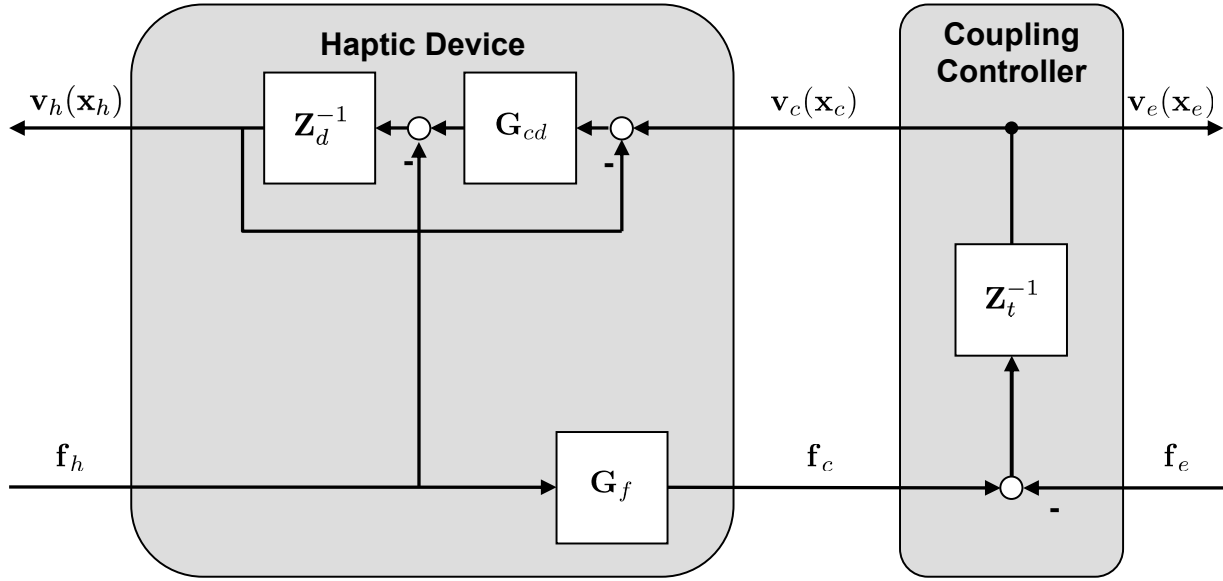


Figure 4.3: Chosen control structure: admittance display with impedance environment causality. It consists of the mechanical device dynamics Z_d^{-1} , device's inherent velocity or position control loop G_{cd} , measured force filter G_f and haptic controller Z_t^{-1}

that in general the drive components powering the device feature integrated velocity or position control loops represented by G_{cd} . This is valid for most industrial electric drive components and robot systems, which are often convenient to use for the design of new rehabilitation devices, as the control loops coming with the aforementioned drive components commonly provide robust stability and very good performance.

The coupling controller Z_t^{-1} is used to stabilize the interaction of the operator with a whole class of virtual environments via the haptic display. It can be a linear or a non-linear coupling control function. The parameters need to be determined such that the controller ensures stability of a user with passive limb impedance providing a voluntary force input with a class of environment characteristics defined depending on the target application. As mentioned before, for the current work the focus is rather to achieve a straight forward practical controller design procedure, providing robustly stable interaction with the lower extremity haptic display. The control structure and design procedure

is based on an experimentally identified linear device model, to avoid the need for elaborate and time consuming formulation of non-linear device dynamics. To enhance the fidelity of the haptic simulation of free motion and contact transition to various types of virtual environments, the controller parameters are adapted when approaching interaction with the virtual environment according to the environmental characteristics.

The methods derived in the following sections always refer to the basic control system structure selected previously within this section.

4.2.2 Robust interaction stability

In the previous section a unified control structure for the class of end-effector gait interfaces was selected for the haptic control framework. In this section a control design procedure is derived, which takes into account device specific dynamics and requirements, i. e. bounds, for the training environment parameters. The goal is to provide a control system structure and design procedure, which is valid for the whole class of end-effector based gait interfaces.

Linearized interaction model

For the design of a coupling controller, which ensures stable interaction between the operator and arbitrary virtual environments with known bounds on their mechanical impedance, a robust control design approach is used. Thereby the use of linear approximations of the non-linear parts of the considered system is possible, because the robust control design accounts for the resulting model inaccuracies.

The presented control design framework can be applied to admittance displays with either underlying velocity or position joint controllers. As the device, which the approach will later be applied to, is position controlled, the following derivations refer to admittance displays with underlying position control loop.

In general the dynamics of a haptic device must be assumed to be non-linear. According to [80] the non-linear rigid body dynamics of a robot can be expressed in joint space as

$$\mathbf{M}(\mathbf{q})\ddot{\mathbf{q}} + \mathbf{V}(\mathbf{q}, \dot{\mathbf{q}}) + \mathbf{G}(\mathbf{q}) + \mathbf{B}(\dot{\mathbf{q}}) = \boldsymbol{\tau} + \mathbf{J}(\mathbf{q})^T \mathbf{f}_{ext} \quad (4.1)$$

with the inertia matrix $\mathbf{M}$, velocity coupling vector $\mathbf{V}$, the vector of gravitational forces $\mathbf{G}$ and the vector of friction $\mathbf{B}$. The vector $\boldsymbol{\tau}$ contains the drive output forces and torques acting on the robot rigid body structure, $\mathbf{J}(\mathbf{q})^T \mathbf{f}$ are external forces acting on the robot end-effector transformed to joint space and $\mathbf{q}$ are the joint coordinates.

Using the configuration dependent Jacobian $\mathbf{J}(\mathbf{q})$ the dynamic equations can be equally expressed in Cartesian space as follows

$$\overline{\mathbf{M}}(\mathbf{q})\ddot{\mathbf{x}} + \overline{\mathbf{V}}(\mathbf{q}, \dot{\mathbf{q}}) + \overline{\mathbf{G}}(\mathbf{q}) + \overline{\mathbf{B}}(\dot{\mathbf{q}}) = \mathbf{J}(\mathbf{q})^{-T} \boldsymbol{\tau} + \mathbf{f}_{ext} \quad (4.2)$$

where the cartesian model parameters denoted as $\bar{\cdot}$ are

$$\begin{aligned}\bar{\mathbf{M}}(\mathbf{q}) &= \mathbf{J}(\mathbf{q})^{-T} \mathbf{M}(\mathbf{q}) \mathbf{J}(\mathbf{q})^{-1} \\ \bar{\mathbf{B}}(\dot{\mathbf{q}}) &= \mathbf{J}(\mathbf{q})^{-T} \mathbf{B}(\dot{\mathbf{q}}) \mathbf{J}(\mathbf{q})^{-1} \\ \bar{\mathbf{G}}(\mathbf{q}) &= \mathbf{J}(\mathbf{q})^{-T} \mathbf{G}(\mathbf{q}) \\ \bar{\mathbf{V}}(\mathbf{q}, \dot{\mathbf{q}}) &= \mathbf{J}(\mathbf{q})^{-T} \mathbf{V}(\mathbf{q}, \dot{\mathbf{q}}) - \bar{\mathbf{M}}(\mathbf{q}) \dot{\mathbf{J}}(\mathbf{q}, \dot{\mathbf{q}})^{-1} \dot{\mathbf{q}}\end{aligned}\quad (4.3)$$

Fig. 4.4 shows the block diagram of a position controlled admittance display with respect to Cartesian workspace coordinates, where $\bar{\mathbf{f}}_{dyn}(\mathbf{q}, \dot{\mathbf{q}})$ are the non-linear robot dynamics, $\bar{\mathbf{G}}_{pos} = \mathbf{J}(\mathbf{q})^{-T} \mathbf{G}_{pos} \mathbf{J}(\mathbf{q})^{-1}$ the position controller and $\mathbf{Z}_t^{-1}$ the transfer function of the coupling controller in Cartesian space.

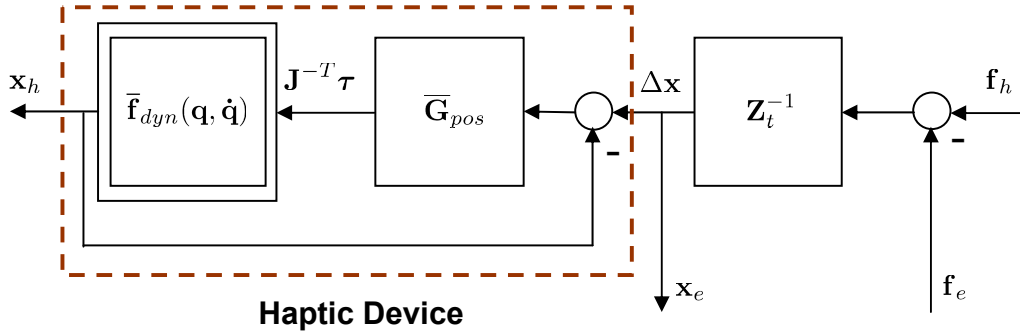


Figure 4.4: Admittance display with non-linear robot dynamics and underlying position control loop

The first step is to derive a linear approximation in frequency domain from the above non-linear differential equation of the controlled system, i. e. the haptic device according to Fig. 4.4, which can be used as the basis for the following design of the robustly stable coupling controller.

An analytical approach for this linearization is to replace the device's non-linear dynamic behavior by a linear device impedance. As stated before impedance can in the present case be the transfer function relating velocity to force or position to force. The model incorporates the velocity proportional friction term $\mathbf{B}$ and the device inertia. The configuration dependent inertia matrix is averaged with respect to the workspace to obtain the nominal model. The velocity coupling vector $\mathbf{V}$ and the gravitational term $\mathbf{G}$ are assumed as unmodeled disturbances to the position controller. From the linearization results the nominal linear dynamic model

$$\boldsymbol{\tau} = \mathbf{Z}_D \mathbf{q} = s(\mathbf{M} \cdot s + \mathbf{B}) \mathbf{q} \quad . \quad (4.4)$$

Together with the known position controller transfer matrix, the linear approximation of the position controlled device's closed loop behavior $\mathbf{T}_{dp}$ can be derived. The individual system order and form of the closed loop transfer matrix depends on the implementation of the position control loop.

For this analytical approach exact knowledge of the device's dynamic parameters is essential. However the identification of dynamic parameters can be complicated and yields uncertainties in parameters in any case. Moreover the structure and parameters of underlying velocity and position control loops are often not accessible and therefore unknown. Accordingly in practice similar to the approach in [18] a further simplification can be done by identifying the linear approximation of the closed loop device characteristics using fixed system order from step response measurement data. Using the step response has the advantage of having an excitation with broad bandwidth and usually leads to good results. A diagonal transfer matrix of 2nd order models according to eq. 4.5 will be used here to approximate the closed loop dynamic Cartesian behavior of the position controlled robot end-effector.

$$\hat{\mathbf{T}}_{dp} = \mathbf{diag} \left(\frac{1}{a_2 s^2 + a_1 s + a_0} \right) \quad (4.5)$$

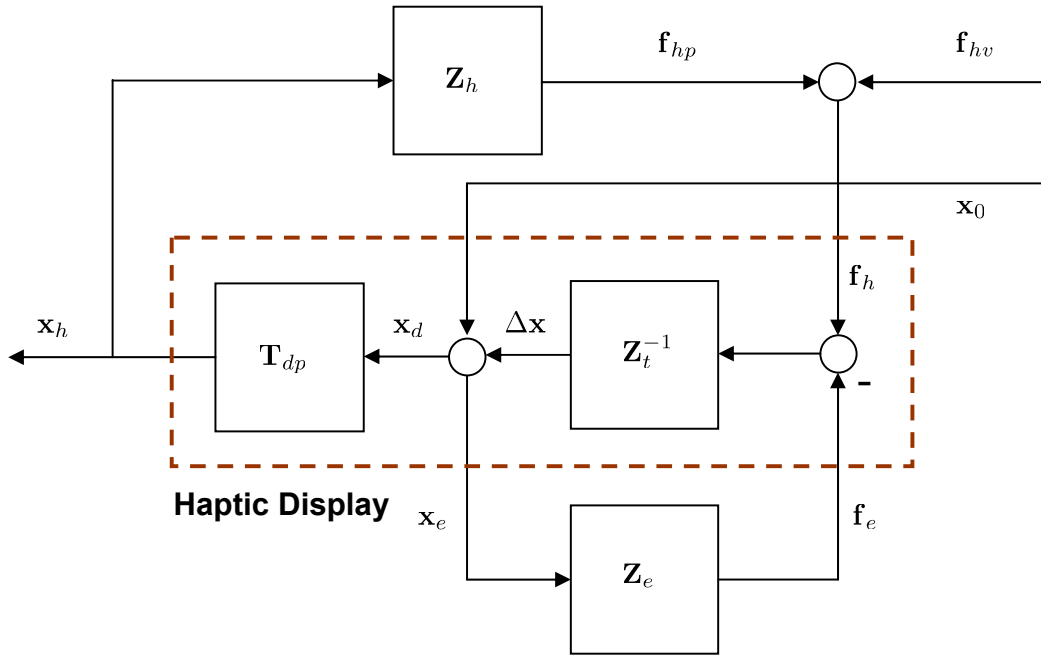


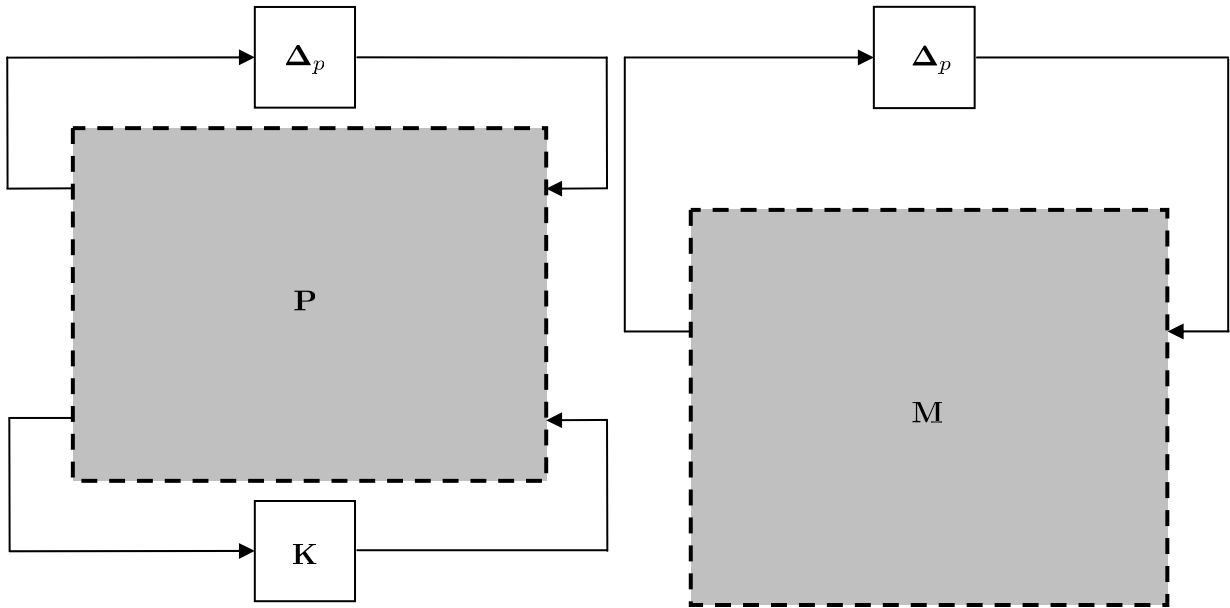
Figure 4.5: Complete coupled system including human operator (Z_H), haptic display (T_{DP} and Z_t^{-1}) and virtual environment (Z_E)

With the derived nominal, linear device model the complete block diagram for the interaction of a human operator and a virtual environment between the haptic display depicted in Fig. 4.5 is obtained. The coupled system consists of the operator Z_h , the haptic display T_{dp} and Z_t^{-1} and the virtual environment Z_e . The system structure matches the two port representation from [26, 27]. The admittance coupling controller Z_t^{-1} and position controlled device T_{dp} link the virtual environment to the operator. The environment uses the position input $\mathbf{x}_e = \mathbf{x}_d$ as input whereas the operator receives the actual position $\mathbf{x}_h = T_{dp}\mathbf{x}_d$ position output from the haptic device.

Coupled and contact transition stability of the haptic gait interface in the presence of bounded environments

As mentioned in the beginning of this section, the simplifications of using linear nominal models need to be accounted for in the design of the controller, which must provide stability for all possible variations from the nominal controlled system. Therefore a robust control design is applied for coupled and contact transition stability in the following. Extensive basics on robust and multivariable control theory is provided in [81, 82].

For the design of robustly stable controllers either a structured or unstructured model of the uncertainties is required. The simplifications made on the system model make the exact determination of non-linear device dynamics unnecessary. However for structured model uncertainties at least part of these dynamics would have to be explicitly determined. Therefore unstructured uncertainties are used in the present approach, which only use an estimate for the upper bound of model errors. A generalized representation of a nominal closed loop control system perturbed by unstructured uncertainties is shown in Fig. 4.6. Δ_p is the physical model uncertainty. In general this uncertainty is described



(a) Nominal controlled system P with uncertainty Δ_p and controller K (b) Nominal closed loop controlled system M with uncertainty Δ_p

Figure 4.6: Generalized nominal closed loop controlled system with model uncertainty

by a normalized uncertainty Δ and weighting matrices $W_1(s)$ and $W_2(s)$ providing an upper bound on the model uncertainties (see eq. 4.6).

$$\Delta_p = W_2(s)\Delta(s)W_1(s) \quad , \|\Delta(j\omega)\|_\infty \leq 1, \quad \forall \omega \quad (4.6)$$

Here Δ is a stable transfer function with $\|\Delta\|_\infty \leq 1$. The weights $W_i(s)$ are likewise assumed to be stable and proper. Fig. 4.6 a comprises the nominal open loop system P controlled by a controller K and perturbed by Δ_p . This system can be transformed into the $M\Delta$ structure where M describes the complete nominal closed loop system (Fig. 4.6 b).

To isolate the normalized uncertainty Δ we can move the weighting matrices from eq. 4.6 to the nominal system M which yields

$$\overline{M} = W_1(s)M(s)W_2(s) \quad . \quad (4.7)$$

The weights define the set of possible models which is subject to the robust control design. In this section the general formulation with two weighting matrices is used. Later in the implementation a simplified form using a single diagonal weighting matrix $W = \text{diag}(w_i)$ is used. For the determination of the weights the linearized models throughout the workspace are identified. From these tests the nominal (average) transfer function $\hat{T}_{dp}$ is computed and the weights are chosen such that $T_{dp} = (I + \Delta W)\hat{T}_{dp}$ contains all measured behaviors.

According to [82] from the $M\Delta$ representation a criterion for robust stability can be derived using a generalized form of the Nyquist theorem for multivariable systems.

$$\max_{\Delta} \rho(\overline{M}\Delta(j\omega)) < 1 \quad , \forall \omega \quad (4.8)$$

Here ρ is the spectral radius of the matrix, which equals the largest singular value. Another property of the spectral radius is that it provides a lower bound on any matrix norm. This allows for the use of norms for stability analysis in the following. With the aforementioned property of the spectral radius and $\|\Delta\|_{\infty} \leq 1$ from eq. 4.8 the robust stability criterion in eq. 4.9 is obtained.

$$\|\overline{M}\|_{\infty} < 1 \quad , \forall \omega \quad (4.9)$$

This multivariable robust stability criterion will be applied to the interaction with virtual environments via the haptic gait interface in the following.

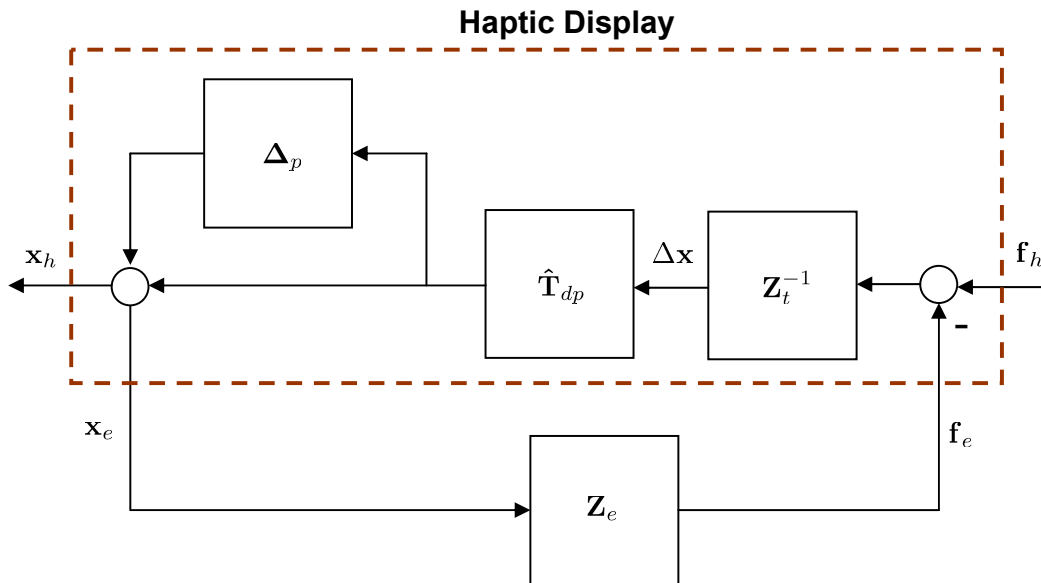


Figure 4.7: Linearized admittance control scheme with uncertain closed loop robot model $(I + \Delta_p)\hat{T}_{dp}$, coupling controller Z_t^{-1} and virtual environment Z_e

Several perturbation models can be used, to account for model uncertainties. To handle the uncertainties and non-linearities in the current case unstructured model perturbations are used. The perturbed linear closed loop model of the position controlled haptic device is shown in Fig. 4.7. $\hat{T}_{dp}$ is the nominal closed loop transfer function of the position controlled device. Using eq. 4.6 the set containing the perturbed closed loop device transfer functions for output multiplicative perturbations can be written as

$$T_{dp} = (I + W_2(s)\Delta(s)W_1(s))\hat{T}_{dp} \quad (4.10)$$

Within the perturbed system, the position input to the virtual environment x_e corresponds now to the device's position output to the operator x_h . The applied approach is based on the assumptions derived in [18] for the analysis of coupled and contact transition situations. The coupled behavior and contact transition remains stable, when the power transferred during contact transition into the haptic simulation by the operator, is larger than the power returned by the virtual environment to the operator. Both amounts of power correspond to the product of the respective forces and position changes. In the system depicted in Fig. 4.7 the respective position changes for operator and environment $\dot{x}_h$ and $\dot{x}_e$ are equal. Therefore the energy transfer corresponds to the relation of the forces f_h and f_e , which is given by the transfer function in eq. 4.11.

$$\begin{aligned} \hat{G}(s) &= Z_e(s)\hat{T}_{dp}(s)Z_t^{-1}(s)[I + Z_e(s)\hat{T}_{dp}(s)Z_t^{-1}(s)]^{-1} \\ &= [I + Z_t(s)\hat{T}_{dp}^{-1}(s)Z_e^{-1}(s)]^{-1} \end{aligned} \quad (4.11)$$

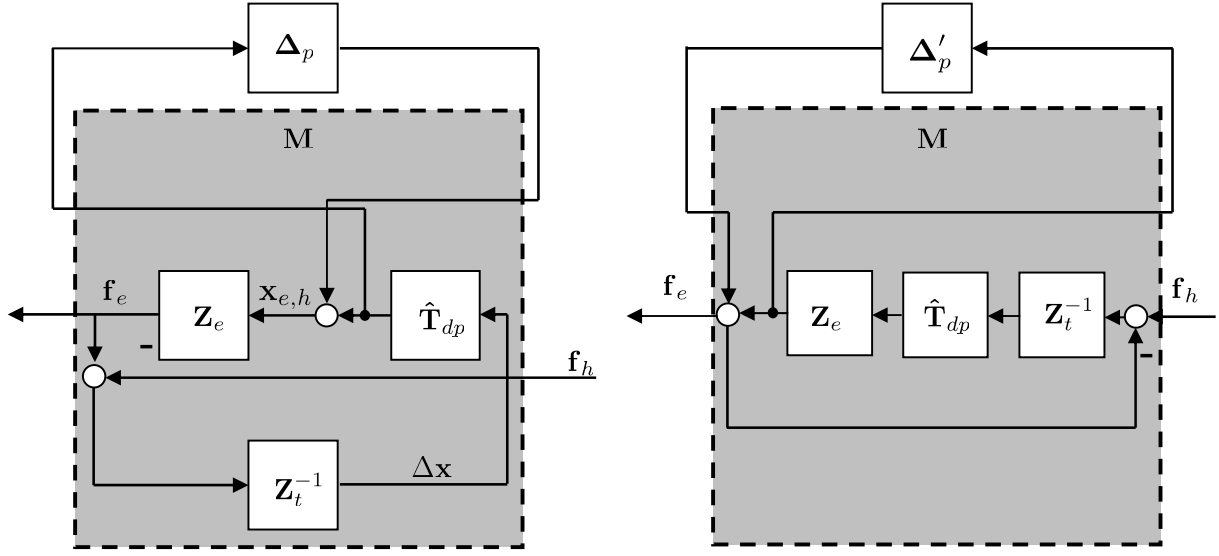
The demand formulated above for contact transition is met by applying the small gain theorem to the nominal haptic system according to Fig. 4.7 with $\Delta_p \equiv 0$.

Nominal coupled and contact transition stability for virtual environments with a maximum stiffness according to Z_e is ensured for the closed loop haptic control system with the nominal device dynamics $\hat{T}_{dp}$ and the coupling controller Z_t^{-1} if for the closed loop transfer function in eq. 4.11 holds

$$\left\| [I + Z_t(s)\hat{T}_{dp}^{-1}(s)Z_e^{-1}(s)]^{-1} \right\|_{\infty} < 1 \quad . \quad (4.12)$$

The *nominal coupled and contact transition stability* condition in eq. 4.12 yields asymptotic stability of the nominal system. It can also be interpreted as a bounded impedance passivity condition, which provides coupled and contact transition stability and is less conservative than conventional passivity design (e. g. [15, 22, 83]).

Fig. 4.8 shows detailed diagrams of the closed loop haptic system with multiplicative output perturbation in the $M\Delta$ structure. Fig. 4.8 a is the perturbed closed loop system with the multiplicative perturbation Δ_p according to eq. 4.12. Fig. 4.8 b shows the influence of the perturbation on the nominal transfer function yielding coupled and contact transition



(a) Nominal closed loop haptic control system M (b) Nominal closed loop system with uncertainty influence on f_h to f_e transfer behaviour

Figure 4.8: Detailed structure of the generalized nominal closed loop controlled system with model uncertainty

stability from eq. 4.12. The transformed perturbation is

$$\begin{aligned}\Delta'_p &= W'_2 \Delta W'_1, \\ W'_1 &= W_1 Z_e^{-1}, \\ W'_2 &= Z_e W_2.\end{aligned}\quad (4.13)$$

From these block diagrams the closed loop system's transfer matrix M for the analysis of the robustness of contact transition stability can be explicitly derived (eq. 4.14).

$$\begin{aligned}M &= Z_e(s) \hat{T}_{dp}(s) Z_t^{-1}(s) [I + Z_e(s) \hat{T}_{dp}(s) Z_t^{-1}(s)]^{-1} \\ &= [I + Z_t(s) \hat{T}_{dp}^{-1}(s) Z_e^{-1}(s)]^{-1}\end{aligned}\quad (4.14)$$

This leads to a robust stability condition for coupled and contact transition of end-effector based gait interfaces with the proposed haptic control system from the general robust stability criterion in eq. 4.9.

Robust coupled and contact transition stability for virtual environments with a maximum stiffness according to Z_e is ensured in the perturbed haptic control system with $\Delta_p = W_2 \Delta W_1$ if *nominal coupled and contact transition stability* according to eq. 4.12 is given and the system transfer matrix $\bar{M}$ meets eq. 4.9 and hence

$$\begin{aligned}\|W'_1(s) [I + Z_e^{-1}(s) Z_t(s) \hat{T}_{dp}^{-1}(s)]^{-1} W'_2(s)\|_\infty &< 1 \\ \Rightarrow \| [I + Z_e^{-1}(s) Z_t(s) \hat{T}_{dp}^{-1}(s)]^{-1} \|_\infty &< \frac{1}{\|W'_1(s)\|_\infty \|W'_2(s)\|_\infty}.\end{aligned}\quad (4.15)$$

Eq. 4.12 and eq. 4.15 present a general criterion for stability tests. The robustness criterion differs from the nominal stability only in the right side of the inequality. If the maximum gain of the weighting transfer functions W'_1 and W'_2 is smaller or equal than 1 ($\|W_i\|_\infty \leq 1$), the bounded impedance passivity condition is the more restrictive and therefore relevant condition. In the case of large perturbations, i. e. $\|W_i\|_\infty > 1$, the robustness condition is more restrictive and must be used.

The evaluation of the perturbation bounds for the gait interface, which the implementation is based on, will be presented in Sec. 5.1.2 before providing the controller parameters resulting from the proposed design criteria.

In many systems further simplifications can be introduced before designing the controller.

1. Device dynamics: if the closed loop position (velocity) control bandwidth b_d is above the one of voluntary motor control, the nominal closed loop device transfer function is selected as unity $\hat{T}_{dp} = I$. In literature the human voluntary motor control bandwidth is given between 7 Hz - 15 Hz [84]. For the lower extremities the necessary device bandwidth is assumed to be $b_d > 10$ Hz.
2. Weighting functions: in most devices the weights for model uncertainties can be reduced to a single diagonal weighting matrix $W = \mathbf{diag}(w_i)$ and hence $\Delta_p = \Delta W$ to improve practical usability of the control design procedure.

Sampled data system

In practical application the control system will be a sampled data system. Therefore the stability conditions are transferred to discrete time domain. The bilinear (Tustin's) transform preserves the $\mathcal{H}_\infty$ norm of a transformed system (see [24]). Hence the *nominal coupled and contact transition stability* and *robust coupled and contact transition stability* conditions can directly be written in discrete time domain, when using the bilinear transform. Further delay present in the force feedback and position control loops can be accounted for. The *nominal coupled and contact transition stability* and *robust coupled and contact transition stability* conditions in discrete time domain are given in eq. 4.16 and eq. 4.17, respectively.

$$\left\| [I + z^n Z_t(z) \hat{T}_{dp}^{-1}(z) Z_e^{-1}(z)]^{-1} \right\|_\infty < 1 \quad (4.16)$$

$$\left\| [I + z^n Z_t(z) \hat{T}_{dp}^{-1}(z) Z_e^{-1}(z)]^{-1} \right\|_\infty < \frac{1}{\|W_1(z)\|_\infty \|W_2(z)\|_\infty} \quad (4.17)$$

Coupling controller structure and parameters – target admittance

Optimization based methods, e. g. $\mathcal{H}_\infty$ synthesis, could be used for the control design. They yield the controller structure (system order of Z_t^{-1}) and parameters by numerical optimization utilizing the stability conditions derived above. However in haptic interfaces mechanical properties are simulated and rendered to the operator. Hence it is convenient to derive the controller structure, i. e. system order of Z_t^{-1} , from considerations on equivalent mechanical models.

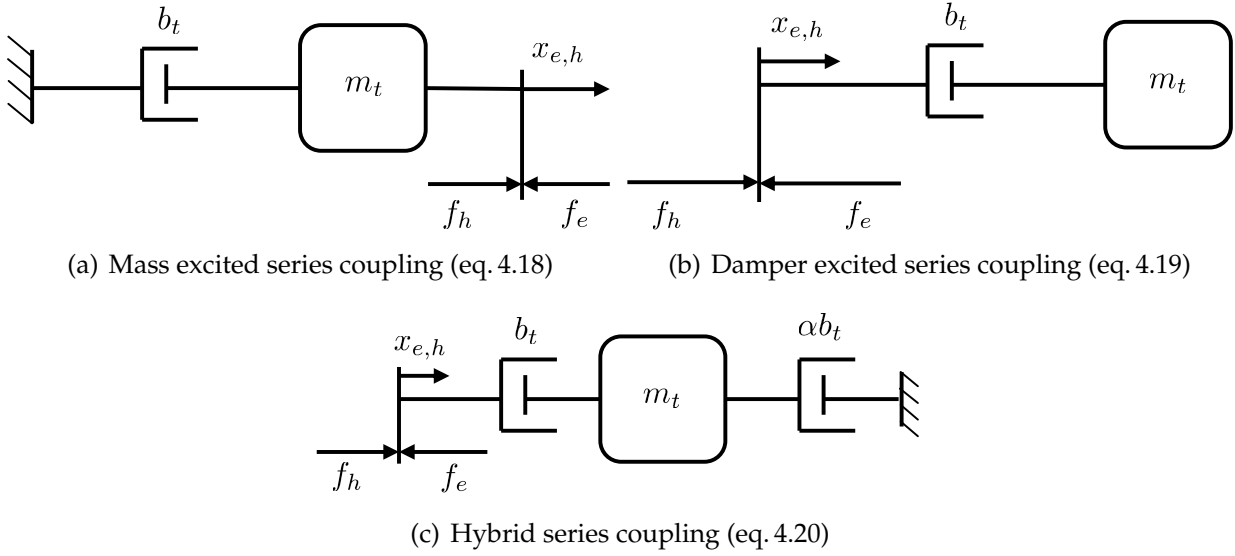


Figure 4.9: Virtual coupling physical representation using mass and damping elements

In [16, 18] a series mass-damper target impedance with mass excitement was used as coupling controller function (eq. 4.18) which resembles a 1st order low-pass. In [26, 27] a series mass-damper structure with damper excitement (eq. 4.19) was proposed to provide maximum transparency for the haptic display. It behaves as a frequency dependent damper with no damping in steady state.

$$Z_{t,mex}^{-1} = \frac{1}{m_t s^2 + b_t s} \quad (4.18)$$

$$Z_{t,dex}^{-1} = \frac{1}{b_t s} + \frac{1}{m_t s^2} \quad (4.19)$$

For the present application a target impedance combining the properties of the two aforementioned controllers $Z_{t,mex}^{-1}$ and $Z_{t,dex}^{-1}$. The corresponding mechanical representations of all three coupling schemes are shown in Fig. 4.9. The hybrid target impedance is described by eq. 4.20, which adds a steady state damping to the damper excited series mass-damper by providing the damper $\alpha b_t s$ connected to a fixed point.

$$\begin{aligned} Z_{thyb}^{-1} &= \frac{1}{b_t s} + \frac{1}{m_t s^2 + \alpha b_t s} \\ &= \frac{1}{s} \cdot \frac{m_t s + b_t(1 + \alpha)}{b_t(m_t s + \alpha b_t)} \end{aligned} \quad (4.20)$$

This leads to a steady state damping of $b_0 = \frac{\alpha}{1+\alpha} b_t$. For very small values $\alpha \ll 1$ and large values $\alpha \gg 1$ the target admittance tends to $Z_{t,dex}^{-1}$ and $Z_{t,mex}^{-1}$ respectively. Using moderate values $0.1 \leq \alpha \leq 1$ the hybrid controller provides slightly less damping for low frequencies than $Z_{t,mex}^{-1}$ but considerably higher damping for high frequencies.

Similar to the design procedure proposed in [18] the determination of the coupling parameters is divided into the following steps:

1. empirically choose α , which is the ratio between damping for high and low frequencies,
2. select initial target mass m_t ,
3. check if for selected m_t a target damping b_t exists, such that eq. 4.16 and eq. 4.17 are fulfilled for the maximum environment stiffness $k_{e,max}$,
4. if b_t exists, reduce m_t and repeat step 3.

Very low values for α reduce the damping for low frequencies and hence the remaining uncompensated end-effector mass must be actively decelerated by the operator, in free motion at constant velocity. This behavior feels uncomfortable. α is chosen empirically depending on the application.

When for the first time for a selected mass m_t no valid b_t can be found to fulfill the stability conditions a backward iteration is started from that value. The target mass is increased again in smaller increments until the stability condition can be met.

4.2.3 Continuous adaptation of target admittance in the vicinity of contact

To improve transparency in free motion and still retain stable contact transition and interaction with stiff virtual objects a gain scheduling approach is used [85].

The virtual object parameters and locations are fully known at every time. Further the distance to all objects is permanently monitored for collision detection. This raises the possibility to adjust the controller parameters to the characteristics of the nearest objects to provide always the minimum damping, which is needed to stabilize the upcoming interactions. Using the controller design procedure from the last section parameter set, i. e. damping values, for a number of expected environment characteristics are computed. In the following scheduling rule the parameter $b_{t,min}$ is the one currently active. $b_{t,max}$ is the parameter designed for smallest stiffness k_e , which is equal or larger than the one for the expected contact.

The controller gain is adapted by continuous change of the target admittance in the vicinity of contact. Therefore the target damping is scheduled with respect to the distance d between probe and nearest object as well as the direction of probe motion as follows:

$$b_t(d) = \begin{cases} b_{t,min}, & d \geq d_{crit}, \\ b_{t,min}, & \frac{d}{dt}d > 0 \wedge d \geq 0, \\ b_{t,min} + \Delta b_t(1 - e^{-8\frac{d_{crit}-d}{d_{crit}}}), & d_{crit} > d > 0, \\ b_{t,max}, & d \leq 0 \end{cases} \quad (4.21)$$

When the probe approaches the an object's boundary with higher impedance than the current environment, below the critical distance d_{crit} the controller's damping is increased

according to the 3rd case in eq.4.21. The difference between minimum and maximum damping is Δb_t . To improve the fidelity when breaking contact with stiff objects the condition $\frac{d}{dt}d > 0$ of the second case adds sensitivity to movement direction to the damping adaptation. If the probe is not approaching the object, i. e. the distance is decreasing, the minimum damping for free motion is used. Fig. 4.10 shows the various states described

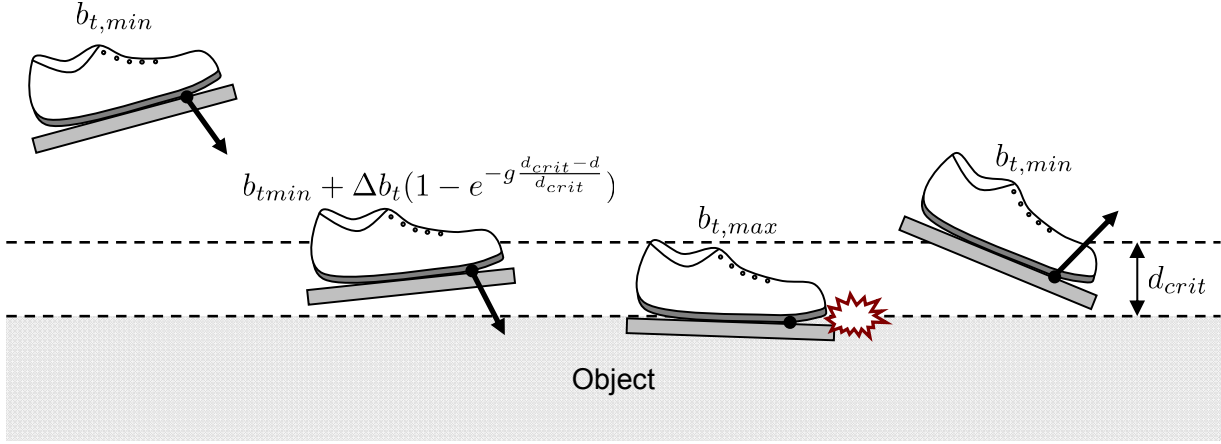


Figure 4.10: Adaptation of target damping in the vicinity of contact with virtual environment

by the cases in eq. 4.21 and the corresponding target damping.

4.3 Concept for synthesis and rendering for virtual walking environments

The first part of the task level control system is the simulation of virtual training environments, which render various scenarios encountered in daily life. The task level control is designed to make use of and extend the low level control architecture presented in the previous section.

4.3.1 Virtual environment and probe model

From Sec. 2.1.2 various types of mathematical models are known for virtual environment representation in haptic simulations. The task is now to select an appropriate model type and implementation for the application of haptic gait simulation. The goal is to keep the computational effort low and therefore the model as simple as possible. Modularity is another important design goal to enable flexible synthesis of new scenarios from reusable primitives. For the decision on the representation of virtual walking scenarios, two major entities are considered:

- A. the *virtual objects* are the smallest components to form the haptic environment and
- B. the *probes* as the operator's interface to explore the environment.

A Virtual objects

The virtual objects are the components for the design of the previously mentioned every day's life walking scenarios. Reusing and recombining these primitives enables fast and flexible synthesis of new scenarios. Therefore single objects as well as their superposition must complement the presented low level control architecture and the characteristics of the complete scenario present a fitting input to the control design procedure. Situations of daily life, which could be desirable to practice, are e. g.:

- floor (treadmill) walking,
- stair climbing (upstairs and downstairs),
- passing obstacles like e. g. a threshold or
- walking on uneven ground with slopes and holes.

The objects needed for these scenarios are rather simple polygonal shapes, such as steps, floor tiles or a door's threshold. To each object individual physical properties are assigned as an impedance Z_o consisting of mass, damping and stiffness.

B Probes

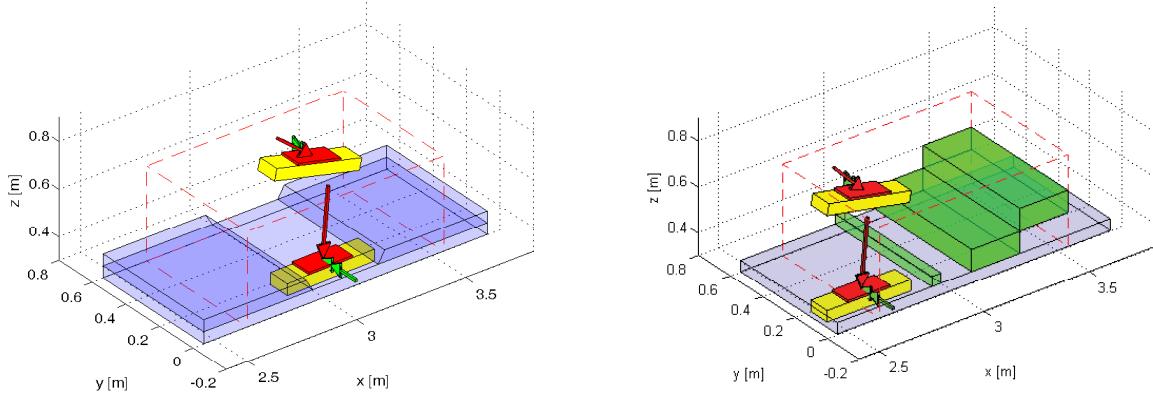
The probes of end-effector type haptic gait interfaces are the virtual feet of the operator, which match the position of the device's footplates mapped into the virtual environment coordinate frame. In the current devices no separate DoF exists for the metatarsal joint therefore the foot tip is not considered part of the probe. The metatarsal joint is the joint at the ball of the foot. The interaction of the foot with geometric objects is addressed, i.e. any contact of the foot sole between heel and metatarsal joint is rendered to the operator. Considering these requirements, the probe is chosen to be a cubic rigid body object representing the foot sole. The size of the probe depends on the operator's shoe size. The probe objects are depicted in red in Fig. 4.11 a and b.

Model for virtual walking scenarios

From the considerations on virtual objects and probes for haptic gait simulation a suitable model type can be selected. Here we also remember, that the model should be kept as simple as possible to allow sample rates also for the task level control system. Aforementioned examples of walking scenarios imply the rendering of proprioceptive information to the operator, whereas subtle tactile information is not intended to be transported. This means that the structure of the required surfaces remains rather simple without very small and delicate structures. Neither is the simulation of inner mechanical characteristics of the object's volumes needed.

Therefore the use of polygonal models is a good choice, as the number of nodes stays manageable. For further simplification of contact detection and computation of object penetration and interaction only convex objects are used. Concave structures, like e. g.

a stair case or a floor with a hole, are composed of a number of convex, possibly overlapping objects. Two examples for virtual gait training scenarios composed of convex polygonal shapes are given in Fig. 4.11. Under the above assumptions, polygonal models provide rather low computational cost. Furthermore they afford numerous interfaces for object representation and import from other tools. Further polygonal models comply with common visualization systems, e. g. using OpenGL. These well-tested interfaces and integration mechanisms are advantageous for future extensions. Intended extensions are the integration of virtual reality into gait training and the implementation of easy and systematic design tools for new training scenarios, which are usable for clinical staff.



(a) Floor tile with a hole made of three convex objects

(b) Floor with threshold and staircase

Figure 4.11: Two examples of virtual gait training scenarios composed of convex polygonal shapes

For consistency with the control design concept developed in the previous section the virtual haptic environment consisting of a set of primitives, is represented as a network of parallel spatially confined, linear, discrete time impedance transfer function blocks (eq. 4.22).

$$\mathbf{Z}_{obj}(z, \mathbf{x}) = \begin{cases} \hat{\mathbf{Z}}_{obj}, & \mathbf{x} \in \mathbf{x}_{obj} \\ \mathbf{0}, & \text{otherwise} \end{cases} \quad (4.22)$$

4.3.2 Rigid body dynamics and texture simulation for probe-object interaction

Rigid body dynamics

According to the definition in Sec. 2.1.2 the probe is the operator's interface to explore the virtual environment. In the haptic gait interface the probes are three dimensional rigid body objects interacting with the three dimensional environment. Hence the equations to describe the probe-environment interaction consider all six dimensions, three translatory and three rotatory, of the Cartesian space. More specific the interaction is described by

contact penalty forces and torques. Mostly throughout this section only the term forces, referring to both forces and torques, will be used.

Fig. 4.12 shows the user input and environment penalty forces acting on a probe. For simplicity Fig. 4.12 shows a simplified planar case. However, the equations below are valid for both planar and the three dimensional case. The probe's motion is determined by

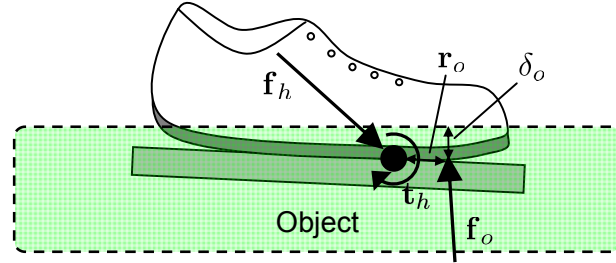


Figure 4.12: Planar example for forces acting on probe during interaction with virtual environment

the operator's external input forces and the virtual environment penalty forces. For the derived admittance control scheme the equations of motion, which relate external forces to probe motion, are given by the coupling controller's differential equations and transfer functions, respectively (see Fig. 4.7). The controller equation is stated in eq. 4.20. In contrary to many other haptics applications, in the presented concept the motion equations differ from the classical mass-spring-damper configuration. In the following the computation of environment penalty forces in the task level control system is addressed.

Both operator forces and penalty forces are measured and computed in the Cartesian workspace coordinate frame. Accordingly in the general six DoF case, for each object, which is penetrated by the probe a 6 element force-torque vector is computed.

$$\begin{bmatrix} f_{oi} \\ t_{oi} \end{bmatrix} = \begin{cases} 0 & \forall \delta_i \leq 0 \\ \begin{bmatrix} f_x & f_y & f_z & t_x & t_y & t_z \end{bmatrix}^T & \forall \delta_i > 0 \end{cases} \quad (4.23)$$

For continuous training movements a dynamic scenario with moving objects, such as a treadmill or an escalator, is necessary. Therefore the simulation has to handle probe and object motion. However, if we consider these examples, it is convenient to neglect any influence of the operator force on the object motion or dynamic interaction between virtual objects.

The penalty forces for each detected contact case are computed separately according to the simplified Newton-Euler equations eq. 4.24 and eq. 4.25.

$$f_i(\delta_i) = \mathbf{M}_i \cdot \frac{d^2}{dt^2} \delta_i + \mathbf{B}_i \cdot \frac{d}{dt} \delta_i + \mathbf{C}_i \cdot \delta_i \quad (4.24)$$

$$t_i(\delta_i) = \hat{f}_i(\delta_i) \times \mathbf{r}_i \quad (4.25)$$

The i th object is characterized by its diagonal matrices of inertia $\mathbf{M}_i$, damping $\mathbf{B}_i$ and stiffness $\mathbf{C}_i$. The aforementioned matrices represent decoupled (diagonal), direction dependent mechanical object characteristics. Further $\mathbf{r}_i$ is the vector from the probe's contact

point on the object to its center of rotation, which is equal to the probe's center of mass. The vector $\hat{f}_i(\delta_i)$ is the effective force used to compute the penalty torques. Under certain circumstances this effective force equals the penalty force resulting from the rigid body dynamics $\hat{f}_i(\delta_i) = f_i(\delta_i)$. Details on the effective force are given later in this section. To simplify the computation of penalty forces and probe motion, the probe's center of rotation is placed in the robot's TCP. In this case no further transformations of external and contact forces are needed.

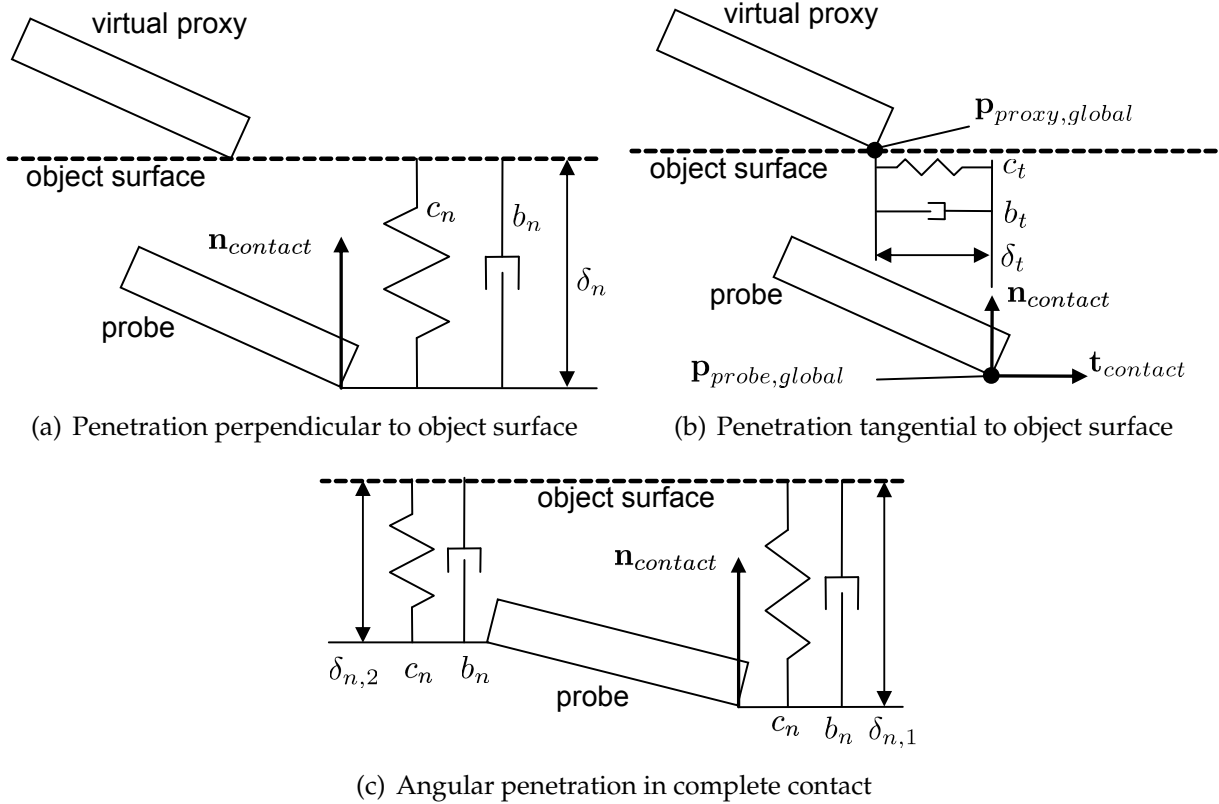


Figure 4.13: Planar interaction of probe with virtual object – penetration determination

Following these assumptions and eq. 4.24 and eq. 4.25, the penalty forces during contact depend only on the mechanical object characteristics and the penetration δ of the probe into the object contour as depicted in Fig. 4.13. The penetration is computed as

$$\delta = \delta_n \mathbf{n}_{contact} + \delta_t \mathbf{t}_{contact} \quad , \mathbf{n}_{contact} \perp \mathbf{t}_{contact} \quad (4.26)$$

where δ_n and δ_t are the penetration depths perpendicular and tangential to the object surface and $\mathbf{n}_{contact}$ and $\mathbf{t}_{contact}$ the corresponding unit vectors in the global coordinate frame.

The remaining task is now to determine the penetrations for all Cartesian coordinates. To be more comprehensible the figures depict planar examples. The vector equations, however, are valid for the three dimensional interaction. Normally contact detection algorithms provide the point of maximum penetration $p_{probe,global}$, the perpendicular pen-

etration depth δ_n and the corresponding vector perpendicular to the penetrated surface $\mathbf{n}_{contact}$.

For the determination of the penetration tangential to the object surface the so called proxy position, i. e. the position of the first penetration on the object $\mathbf{p}_{proxy,object}$ and probe $\mathbf{p}_{proxy,probe}$ in local coordinates is used. Both points are remembered for future computations. When the probe and/or the object moves those points change their relative position in environment coordinates. Let the transformation from probe to environment coordinate frame be $\{\mathbf{t}_{probe}^{global}, \mathbf{R}_{probe}^{global}\}$ and from object to environment frame $\{\mathbf{t}_{object}^{global}, \mathbf{R}_{object}^{global}\}$. Hence the tangential penetration is

$$\delta_t = [(\mathbf{t}_{probe}^{global} + \mathbf{R}_{probe}^{global} \mathbf{p}_{proxy,probe}) - (\mathbf{t}_{object}^{global} + \mathbf{R}_{object}^{global} \mathbf{p}_{proxy,object})] \cdot \mathbf{t}_{contact} \quad . \quad (4.27)$$

When the probe only partly penetrates the object, the force vector used to compute the penalty torques is $\hat{\mathbf{f}}_i = \mathbf{f}_i$. In case the probe completely penetrates the object (see Fig. 4.13 c), e. g. when standing on a virtual floor, this can lead to oscillations. Small changes in the operator input force, e. g. slight shift of weight, can change the probe orientation and the point of largest penetration switches from heel to metatarsal joint or vice versa. Therefore if the probe completely penetrates an object, the force is computed as a function from an effective perpendicular penetration

$$\hat{\mathbf{f}} = f(\underbrace{\delta_{n,2} - \delta_{n,1}}_{\delta_{eff}}) \mathbf{n}_{contact} \quad . \quad (4.28)$$

Here the penetration $\delta_{eff} = (\delta_{n,2} - \delta_{n,1})$ is called effective penetration and $f(\cdot)$ is a scalar function of it. The influence of the tangential force component is neglected as it is almost in line with the vector $\mathbf{r}_i$.

With the straight forward mechanical model described by eq. 4.24 – eq. 4.28 a variety of object types can be modeled. For stiff objects like floor tiles, steps etc. a pure stiffness C is used. Viscose obstacles e. g. are modeled by using objects with mass and damping characteristics.

Surface texture simulation

The above model provides the means to simulate most mechanical characteristics of virtual objects for a walking scenario. However, it's means to simulate surface characteristics are limited. When attempting to move along the surface the operator either sticks to the object or experiences the feeling of pure viscose, velocity dependent friction. To simulate stick-slip effects on the object surface, the model from the last section is extended by a static friction component. Therefore the basic stick-slip model from [86] is applied to the present application (see Fig. 4.14).

If the norm of the tangential penetration δ_t is below $\delta_{t,static}$ the virtual proxy position on the object $\mathbf{p}_{proxy,object}$ is not changed and hence the operator sticks to the object. When $|\delta_t| > \delta_{t,static}$, i. e. the operator exerts a force larger than the maximum static friction force

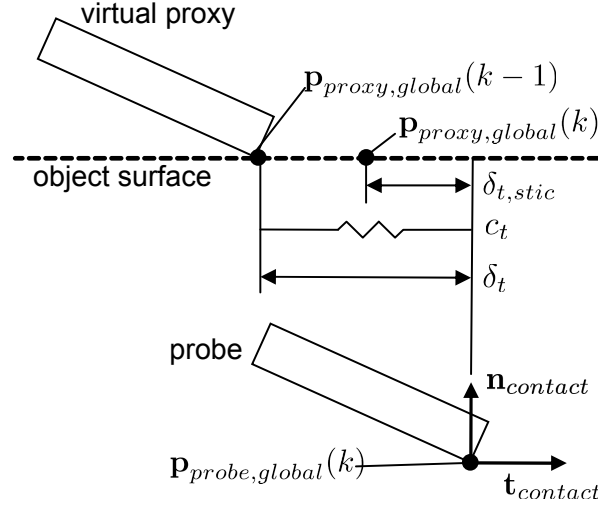


Figure 4.14: Static friction in the haptic simulation

$f_{t,static,max}$, the point $p_{proxy,object}$ is moved along the object's surface and the operator feels a viscose friction and slides along the object surface.

$$p_{proxy,object}(k) = \begin{cases} p_{proxy,object}(k-1) + \text{sign}(\delta_t) |\delta_t - \delta_{t,static}| \cdot t_{contact} & , |\delta_t| > \delta_{t,static} \\ p_{proxy,object}(k-1) & , \text{else} \end{cases} \quad (4.29)$$

The maximum static friction force $f_{t,static}$ is depending on the normal force

$$f_{t,static} \leq \mu |f_n| \quad \text{and} \quad (4.30)$$

$$f_{t,static,max} = \mu |f_n| \quad . \quad (4.31)$$

To apply the static friction to the haptic simulation eq. 4.31 must be formulated in terms of penetration depths. Therefore only the stationary component of the environment force is used, which is the part resulting from environment stiffness. The stiffness tangential to the surface is c_t the perpendicular stiffness is c_n . Therewith results

$$\delta_{t,static} = \mu |\delta_n| \frac{c_n}{c_t} \quad . \quad (4.32)$$

4.3.3 Synthesis and haptic rendering of complex, multi-object training environments

The previous section provided a mathematical model to compute the interaction forces between a probe and an object. For complex training environments multiple of these objects, or primitives, are superposed. The use of primitives promotes an easy synthesis of scenarios and combination of static and dynamic, i. e. moving, components. The scenarios consist of unmoving and moving primitives alike, which do not interact with each other and can even be partly or totally collocated. As stated previously, from the low level control point of view this results in an environment impedance network [87, 88].

The network of transfer function blocks replaces the single transfer function $Z_e(z)$ in Fig. 4.7. For the environment network a single equivalent worst case impedance $\hat{Z}_e(z, \mathbf{x})$ can be determined. Thereby the low level control design procedure in Sec. 4.2 can be applied. The bounded impedance robust coupled and contact transition stability allows for the arbitrary combination of environment components as long as the bounds used in the haptic controller design are maintained for the environment network's equivalent shunt impedance $\hat{Z}_e(z, \mathbf{x})$.

Basic network structure for the virtual environment

To enable an easy and flexible, yet systematic synthesis of arbitrary training environments, the simple parallel arrangement of virtual object blocks depicted in Fig. 4.15 seems the most suitable choice. Fig. 4.15 a shows the transfer function network used in the theoretical analysis, Fig. 4.15 b provides a physical interpretation of what is happening in a contact situation with two overlapping objects.

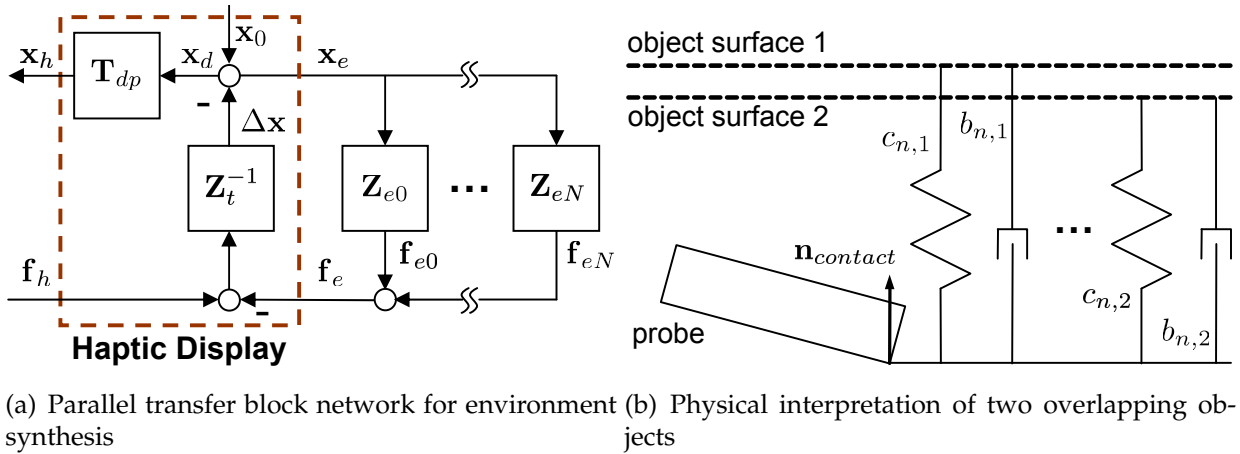


Figure 4.15: Parallel virtual objects network arrangement forming the virtual environment

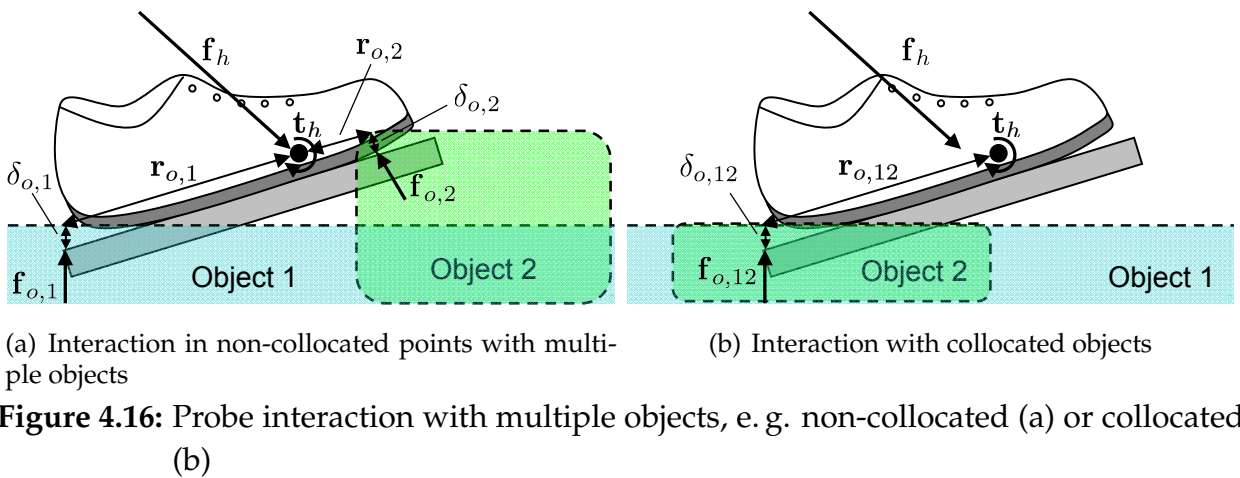


Figure 4.16: Probe interaction with multiple objects, e. g. non-collocated (a) or collocated (b)

The object transfer functions are linear spatially confined impedances, i. e. they present a linear behavior in a defined location. The superposition of a number of N objects yields

for a position x_h

$$\mathbf{Z}_e(\mathbf{x}_h) = \sum_{i=0}^N \mathbf{Z}_i(\mathbf{x}_h) \quad (4.33)$$

$$f_e(\mathbf{x}_h) = \sum_{i=0}^N f_{o,i}(\mathbf{x}_h) \quad , \quad (4.34)$$

with $\mathbf{Z}_i(\mathbf{x}_h)$ according to eq. 4.22. If two or more objects overlap and the probe is penetrating more than one object, eq. 4.33 yields the effective environment impedance which is the sum of all penetrated object impedances.

Maintaining necessary environment bounds

The control design approach (see Sec. 4.2) makes use of the fact, that the worst case environment regarding the stability of the operator environment interaction is a pure stiffness. Further the objects rendered to the operator possess a known upper bound for their stiffness, which is defined by the application's demands. In the present case e. g. maximum users bodyweight defines the demands for floor stiffness.

Since the designer is free in the choice and placement of objects, several objects might overlap locally. Since the network of transfer blocks resulting from the environment synthesis is a parallel structure, the objects impedances of overlapping objects would add up. Generally this is not a problem as long as the actual upper bound, i. e. the sum of the stiffness of all potentially overlapping objects is used for control design. However, this could lead to quite conservative controller parameters, though the upper stiffness bound may not be demanded by the application but results from the local overlapping of objects.

Thus for the case of contact with more than one overlapping object, where the single stiffnesses would add up, only the maximum object stiffness is effectively rendered (eq. 4.35).

$$k_e(\mathbf{x}) = \max_i k_i(\mathbf{x}) \quad (4.35)$$

Synthesis process

For the synthesis of training scenarios including the environment, guidance and support elements as well as low level control parameters the systematic synthesis chain depicted in Fig. 4.17 is applied.

First the training task is defined by the therapist, e. g. floor walking, stair climbing or walking up or down a slope. Additionally obstacles can be added to the scene. Definition of the training task involves the choice of a nominal movement trajectory, supposed by the therapist to be most adequate to solve the task. In this thesis the nominal training trajectory is assumed to be available. It can e. g. be derived from motion tracking data recorded in healthy subjects and is then scaled to the individual subject [89].

Based on the defined training task a basic environment consisting of stiff objects, e. g. floor tiles or steps, is designed. Additionally the training environment can contain stiff

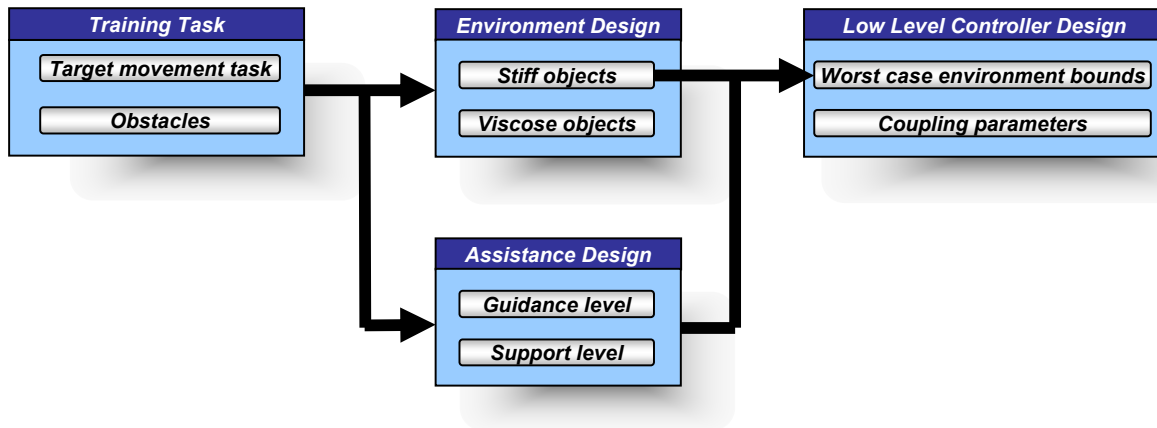


Figure 4.17: Design process for the synthesis of training scenarios

or viscose obstacles. The concept for assistance will be elaborated on in detail in the following section. The maximum stiffness of the assisting objects is used for the design of the coupling controller.

The haptic display low level controller parameters are computed based on the environment impedance bounds. As stated in Sec. 4.2.2 the worst case environment, in terms of haptic display stability, is a pure spring characteristic. Hence a set of controller parameters is computed for all values of object stiffness occurring in the scene, starting with the highest value. The coupling mass determined for this parameter set remains constant in all sets and only different damping values are determined for all controller sets. Typically the lowest of the stiffnesses is that of the assisting objects and the associated controller parameters are also used for free motion.

4.4 Concept for patient-adaptive training assistance on end-effector based haptic foot devices

The training task, corresponding to a certain virtual haptic environment, is defined via a nominal trajectory. To ensure the fulfillment of the task by the patient, the performed movement needs to be:

- guided and
- supported.

The previous sections referred to challenges and solution concepts, which apply to the operator and haptic gait interface interaction in general. This section specifically addresses the integration of assistance for rehabilitation training. Therefore the operator in the following section is the patient receiving the gait training.

Guidance means to prevent deviations from a nominal trajectory, defined to accomplish the training task, above a given tolerance value.

Support defines elements in the haptics framework, which help a patient to move along the desired path to accomplish the training task.

These two subtasks are summarized under the term assistance. Sec.2.3 provides an overview of control strategies for gait training. Sophisticated compliant and adaptive training algorithms were to date only developed for the use with exoskeletons.

Based on the simulation of virtual training scenarios from the previous sections and the concept of path control applied in exoskeleton devices [2, 42] an approach to provide adjustable task specific assistance on end-effector devices is developed. Further a basic method for the adaptation of the assistance level to the patient's performance is introduced.

4.4.1 Adjustable, task specific assistance in the haptic control framework

Assistance object representation

Besides the polygonal shapes forming the basic training environment a special class of objects is used to implement assistance modes during robot assisted gait training. Like the polygonal objects forming the walking scene, the objects implementing the assistance are integrated into the haptic control framework as dynamic objects interacting with the haptic display's probes.

The goal is to implement assistance similar to the path control strategy using dynamic objects to compliantly guide and support the patient on the training path. Therefore a window moving along a predefined training trajectory is chosen. A special case used here is an ellipsoid which contrary to all other objects is not approximated by a polygon but described by its quadratic form in the window's local coordinate frame in the ellipsoid center

$$\frac{x^2}{l_x^2} + \frac{y^2}{l_y^2} + \frac{z^2}{l_z^2} = 1 \quad . \quad (4.36)$$

The current ellipsoid position in the global coordinate frame is defined by the translational offset to its center $p_{act} = \begin{bmatrix} x_{p_{act}} & y_{p_{act}} & z_{p_{act}} \end{bmatrix}^T$ and its orientation given by the rotation matrix R_{window}^{global} between the global and the window coordinate frames. The current position p_{act} corresponds to a sample point on the nominal trajectory and the orientation of the window frame x-axis is tangential to the trajectory.

Within the ellipsoid no assistance is provided, i. e. the patient is able to move freely with minimum displayable device impedance. If the probe leaves the ellipsoid, an impedance Z_w correcting the motion is displayed to the patient. To adjust the assistance level, the parameters, i. e. size and impedance, of the window and its velocity along the trajectory can be modified.

Assistance computation model

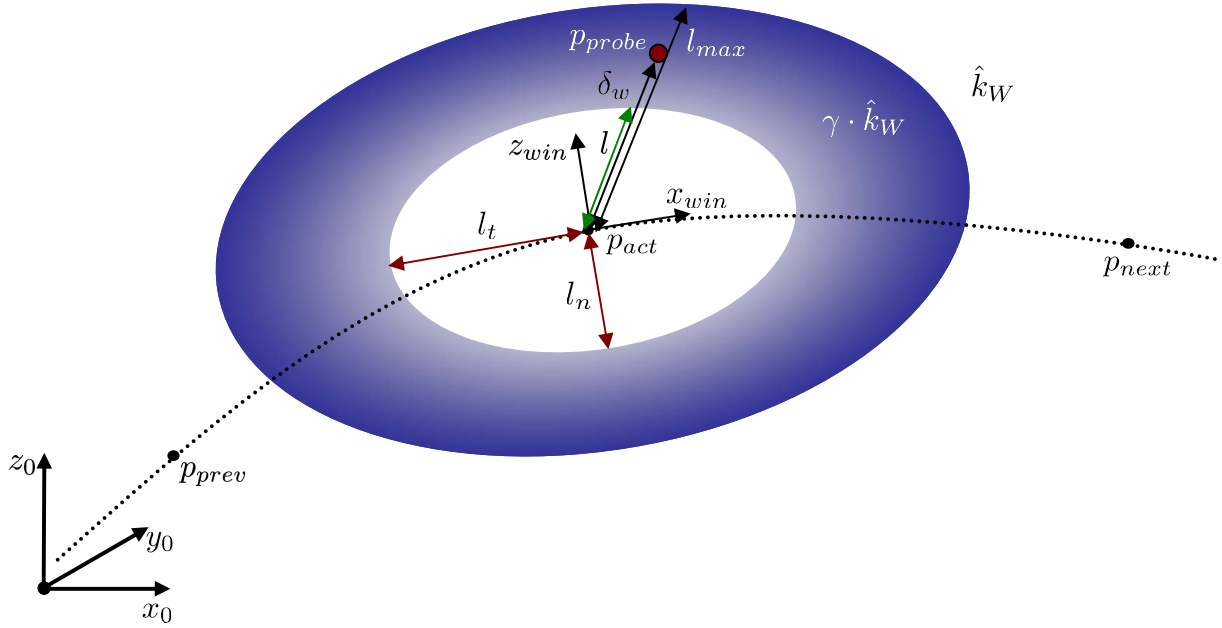


Figure 4.18: Stiffness characteristic of the assistance window. Inner white region comprises no stiffness, outer region with color gradient from white to blue represents progressive stiffness $\gamma \cdot \hat{k}_W$. Outside of the outer elliptic region maximum window stiffness $\hat{k}_W$ is present.

The parameters of the general ellipsoid window from the previous section are modified for the assistance concept, which is applied within the framework. Fig. 4.18 shows the window in the x-z plane.

The dimensions of the elliptic window are the window length tangential to the trajectory $l_t = l_x$ and the window size in the plain perpendicular to the trajectory $l_n = l_y = l_z$. Within this ellipsoid region the window impedance is $Z_w = 0$. Outside of this inner tolerance region, the window impedance is selected as a pure stiffness. The stiffness increases progressively from 0 to $\hat{C}_w$ in the region between the inner ellipsoid defined by l_t and l_n and an outer ellipsoid defined by $l_{t,max}$ and $l_{n,max}$.

Distances and penetrations between probe and window are always computed with respect to the window center. The forces produced by the window impedance on the probe are likewise always directed towards the center point. Therefore the distance vector

$$\mathbf{d}_w = \mathbf{p}_{act,global} - \mathbf{p}_{probe,global} \quad (4.37)$$

is always perpendicular to the window surface. The penetration vector δ_w has the same direction as $\mathbf{d}_w$. l is the radius of the inner ellipse in the direction of $\mathbf{d}_w$. The penetration vector results to

$$\delta_w = \underbrace{(\|\mathbf{d}_w\| - l)}_{\delta_w} \frac{\mathbf{d}_w}{\|\mathbf{d}_w\|} \quad (4.38)$$

The euclidean length of the window penetration vector is δ_w . The effective stiffness of the assistance window for the $i - th$ coordinate direction results to

$$c_{w,i}(\delta_w, \lambda) = \gamma(\delta_w)(\lambda \hat{c}_{w,i}) \quad . \quad (4.39)$$

Here λ is a factor with $0 \leq \lambda \leq 1$ to adjust the assistance by scaling the maximum window stiffness. The progressive behavior is implemented with the weighting function $\gamma(\delta_w)$.

$$\gamma(\delta_w) = \begin{cases} 0 & \delta_w < 0, \\ 1 & \delta_w > |l_{max} - l| \\ \left(\frac{\delta_w}{l_{max} - l}\right)^2 & \text{else} \end{cases} \quad . \quad (4.40)$$

The assistance force provided by the window results to

$$f_{assist} = \text{diag}(c_{w,i}(\delta_w, \lambda)) * \delta_w \quad . \quad (4.41)$$

From moving window to virtual tunnel

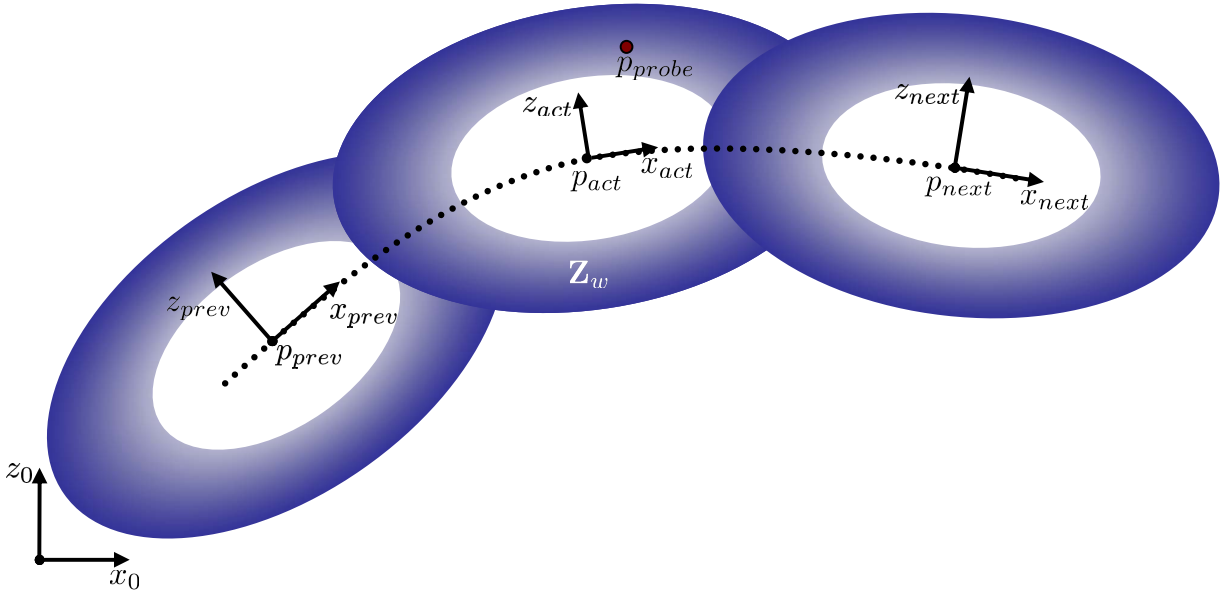


Figure 4.19: Tunnel resulting from elliptic window moving along the nominal task trajectory

When the window is moving along the nominal trajectory, a tunnel similar to those defined in the path control approaches is obtained. The example in Fig. 4.19 shows three window positions at sequential instants in the x-z plane. The current center of the window, which lies on the nominal trajectory, is p_{act} . The probe position is denoted as p_{probe} . If the window is moved along the trajectory using a high sampling resolution, the free motion inner area of the moving window results in a tunnel within which unhindered motion is possible. Fig. 4.20 shows the resulting tunnel for three different resolutions for a treadmill trajectory.

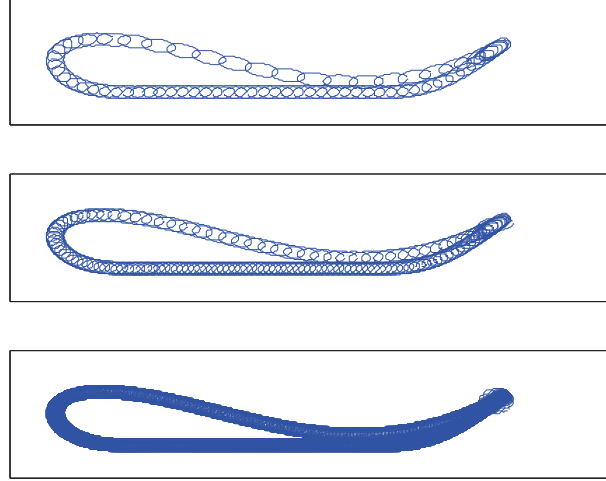


Figure 4.20: Tunnel resulting from window moving with three resolutions – treadmill example

4.4.2 Assistance modes

The previous section describes the basic window object, used to render assistance during gait training. As mentioned above the assistance is divided into the subtasks guidance and support. Two approaches (modes) to provide assistance are developed in this section, which are intended to

- A. adjust guidance and support with one unified scaling parameter or
- B. allow to address guidance and support via separate scaling parameters.

In both approaches one window object for each foot is used. The windows are displaced by half the gait cycle with respect to each other. Thereby the coordination between both legs is taken into account, when providing training assistance.

Mode A Window with user selected training velocity

In this assistance mode the windows move at a user defined velocity along the specific trajectory for the apparent training task. The trajectory is sampled at equidistant time intervals $\frac{\Delta T}{N}$. ΔT is the sample time of the RT system running the low level admittance control and N a positive integer representing the oversampling. The whole number of points sampled for one gait cycle are $\hat{N} = \frac{T_{s,min}}{\Delta T} N$. $T_{s,min}$ is the minimum time needed for one gait cycle, i.e. for the maximum training velocity. The index of the i -th window position on the trajectory n_i is

$$n_i = n_{i-1} + v_w \cdot N \quad 0 \leq n \leq \hat{N}. \quad (4.42)$$

In eq. 4.42 v_w scales the training velocity between zero and the maximum value. It ranges between -1 and 1 . Negative values represent a backward movement.

In this assistance mode the window is dragging the patient along, if he moves slower than the target training velocity or holds him back if he moves faster. At the same time

the window stiffness restricts deviations perpendicular to the training trajectory. As a result both guidance and support are provided by the moving window and depend on the windows stiffness (see eq. 4.39). Consequently only one factor λ is used to scale both guidance and support level in equal measure and is in this case called assistance level (see eq. 4.43).

$$\lambda_{guid} = \lambda_{sup} = \lambda \quad (4.43)$$

The nominal trajectory comprises the geometry of the foot motion. Both windows moving at a defined velocity along the nominal training trajectory are shifted by half a gait cycle and provide a correct inter-limb coordination. The stiffness is usually high enough to prevent the loss of this inter-limb coordination.

Mode B Window following the probe motion and additional support force

The second assistance mode synchronizes the window position to the probe movement. The center of the window is still always on the trajectory and the trajectory is sampled as described before. The actual trajectory point, which the window is positioned on, is the nearest sampled point on the trajectory to the current probe position

$$n_i = \{n \mid \|p_{global}(n) - p_{probe,global}\| = \min\} \quad n_{i-1} - N < n_i < n_{i-1} + N \quad . \quad (4.44)$$

The interval, which the minimum distance search is performed in, is delimited by the number of oversampling points N eq. 4.44. N is the maximum number of points the window should move along the trajectory as it corresponds to the highest desired training velocity, at which the trajectory was sampled offline.

By moving the window along with the probe, the result is a virtual tunnel around the nominal training trajectory, which size is determined by the window dimensions. The patient can move unhindered along the tangential direction on the trajectory. Perpendicular to the nominal trajectory the window restricts the deviation of the probe.

Therefore the window provides only guidance depending on its stiffness (eq. 4.39). The guidance level, which scales the window stiffness, is λ_{guid} . For training support a force field tangential to the trajectory is applied. The amount of support depends on the maximum strength of the force field $\hat{f}_{sup}$ and the support level λ_{sup} .

$$f_{sup} = \lambda_{sup} \hat{f}_{sup} \quad (4.45)$$

In this approach the subtasks guidance and support are addressed and scaled separately with λ_{guid} and λ_{sup} .

If the motion of each side's windows is left decoupled, no inter-limb coordination is forced on the user. However, this can lead to a patient leg motion which deviates from the natural phase shift between both legs and is therefore uncoordinated. This is not desirable in robot assisted gait training and studies about robot-assisted gait training in spinalized mice showed that imposing inter-limb coordination by the robot leads to better training results [1]. To impose inter-limb coordination two different approaches are intended.

In **master-slave** mode one foot is master. The common window position is the nearest point to the master (eq. 4.46).

$$n_{common} = n_{master} \quad (4.46)$$

In multi-master mode both feet are equal and the common point is computed according to eq. 4.47.

$$n_{common} = n_{right} + \frac{n_{left} - n_{right}}{2} \quad (4.47)$$

In this assistance mode the current window velocity is computed as

$$v_{w,i} = \frac{n_{common,i} - n_{common,i-1}}{N} . \quad (4.48)$$

4.4.3 Performance based adaptation of training assistance

Both approaches from the previous section to render training assistance within the haptic control framework integrate one or more scaling factors λ to adjust the assistance level between 0 % and 100 %. One way to do this is a manual adjustment by the therapist. A more sophisticated method is to use an adaptation algorithm to modify the assistance level according to the patient's training performance. This work provides basic ideas, which intend to lead toward an adaptive, therapist-like behavior of end-effector based rehabilitation robots. The human therapist adapts inherently to the patient's needs and possibilities. Likewise in the future the robotic assistance should be adapted automatically similar to the human therapist.

The proposed performance based adaptation of training assistance is directly derived from known methods established in other devices [51, 90, 42], which use iterative learning control (ILC) methods. It consists of two major parts:

- A. performance measure and
- B. iterative learning algorithm.

The iterative learning algorithm uses a performance measure, which is used to modify the assistance level from the previous iteration to compute the current assistance level. The iterative learning is based on the model for human motor learning and *assist as needed* control approach from [51]. Firstly an appropriate performance measure, or in terms of ILC tracking error, is derived. This is used afterwards in the iterative learning algorithm.

A Performance measure for error based assistance scaling

For the application of training assistance adaptation the term usually called tracking error in the ILC, is the performance measure. It describes how well the patient performs the training task. The optimal training performance would be to track the nominal training trajectory the window is moving on in terms of geometry and timing. Consistent to

the other approaches cited above kinematic deviations are used to classify the training performance.

The two assistance modes proposed in the previous section provide different means to render and adjust guidance and support. Consequently, for each mode a different performance measure is used in the adaptation of the assistance level.

Assisting the training movement in **Mode A** leaves only one factor to scale both guidance and support. The window strictly follows its nominal trajectory and the patient is dragged along by the window. Hence he can be assumed to always stay close to the nominal window position and will never drift away from the nominal movement timing. In this case it is feasible and convenient to directly use the absolute value of geometric distance from the nominal training path.

$$e_A(n) = \|\delta_w(n)\| = \delta_w(n) \quad (4.49)$$

Eq. 4.49 provides an error measure e_A for both the deviations perpendicular to the trajectory and the tendency to lag or advance the nominal motion.

In **Mode B** guidance and support are addressed separately. The window moves along the nominal trajectory according to the probe's position and restricts the perpendicular deviation. The support is provided by a tangential force field. However the window does not necessarily follow the motion along the training trajectory regarding the timing. In this mode the patient should be allowed to move constantly slower or faster than the nominal training velocity. Compared to a nominal position computed from the nominal training velocity, the actual position lag can be substantial and even continuously increasing if the patient moves constantly slower than the nominal velocity. Hence geometric deviations can only be used as a measure to adjust the guidance level. For the support level the difference between the nominal velocity $v_{w,nom}$ and the actual velocity v_w is used.

$$e_{B,guid}(n) = \delta_w(n) \cdot \mathbf{n}_w(n) \quad (4.50)$$

$$e_{B,sup}(n) = v_{w,nom}(n) - v_w(n) \quad (4.51)$$

B Iterative learning algorithm

The iterative learning algorithm comes from the field of ILC. Theoretical background on ILC can be found in [91]. ILC is particularly useful to control repetitive tasks. The principle of ILC is to use knowledge about the performance from previous iterations, e. g. tracking error, and adjust the control action of the current iteration using this knowledge. The simplest form of an ILC uses the sampled series of control actions from the previous iteration $\mathbf{u}^{(k)}(n)$ and adds for each sampling point a tracking error correction term.

$$\mathbf{u}^{(k+1)}(n) = \mathbf{u}^{(k)}(n) + \mathbf{G}_e \mathbf{e}^{(k)}(n) \quad (4.52)$$

In eq. 4.52 $\mathbf{G}_e$ is the error gain matrix and $\mathbf{e}^{(k)}(n)$ is the sampled tracking error from each sample point of the previous iteration.

For the application in the performance based assistance adaptation the vector equation eq. 4.52 becomes a scalar equation. The control action $u(n)$ equals the assistance level λn and the tracking error $e(n)$ is selected according to the previous section. An effect, which occurs for this simple ILC, is that the control action converges to a certain state. However this is not desired for the present application as the patient should be continuously challenged during the training, which is not the case if the control action converges to a static state. A similar effect is known from online system identification algorithms which may get insensitive to parameter variations. To avoid this problem [51] suggests a forgetting factor scaling down the contribution of the previous iteration's control action. The ILC control law for the performance based assistance adaptation can be written as

$$\lambda^{(k+1)}(n) = f\lambda^{(k)}(n) + g_e e^{(k)}(n) \quad f < 1. \quad (4.53)$$

In the ILC law f is the scalar forgetting factor, λ the guidance or support level, g_e the scalar error gain and e the respective performance measure.

4.5 Evaluation concept for device assisted rehabilitation training

To evaluate the developed haptic control framework for device assisted rehabilitation it is implemented on the robotic walking simulator *HapticWalker* [89]. Using the implementation on this device, experiments with healthy subjects are conducted.

The evaluation covers the following three stages, starting from basic functionalities and proceeding to the complete extent of features proposed in the framework's concept.

- A. Basic rendering of haptic walking environments
- B. Voluntary free motion on the haptic gait interface
- C. Assisted training modes

A Basic rendering of haptic walking environments

The stability of contact transition and interaction with virtual training objects is evaluated in three setups on the *HapticWalker*.

1. Interaction with a single stiff object

The contact transition from free motion to a single static floor tile is simulated with the same controller parameters and varying object stiffness. Thereby stable and especially unstable interaction cases can be investigated without a danger to damage the experimental apparatus.

2. Interaction with multiple objects

The second case to investigate is the interaction with multiple virtual objects. Therefore the interaction with two objects in a collocated and non-collocated setup is sim-

ulated. The stiffness of both objects will be selected as the bound used for the controller design.

3. Rendering of surface friction and stiction

The proposed concept to render surface friction and stiction is investigated by sliding along the surface of a single object.

B Voluntary free motion on the haptic gait interface

The second evaluation step is the voluntary free motion of a subject in various complex walking environments. From these experiments the haptic display behavior in free motion, i. e. fully guided by the operator, and the possibility to perform real walking movements in a complex virtual walking environment can be investigated. At last the feasibility of dynamic training environments with moving weight bearing objects is addressed.

1. Free motion simulation

Free motion above the floor is performed while standing on the virtual ground on the second leg.

2. Static stair case

The free motion through a complex walking environment is shown by a subject walking up and down two static steps.

3. Virtual treadmill

Walking on a virtual treadmill consisting of two floor tiles shows the interaction with a dynamic walking environment.

C Assisted training modes

The final part of the evaluation considers the proposed assisted training modes to perform different training tasks in dynamic walking environments. Both assistance modes described in Sec. 4.4.2 are investigated in healthy subjects with a set of different training scenarios. Firstly various, constant assistance levels are applied in each mode. At last the experiments are repeated with the ILC based automatic adaptation of the assistance level.

1. Moving window mode

The healthy subject performs different training tasks assisted by the moving window with varying constant assistance levels.

2. Patient-adaptive training. The subject performs different training tasks in the moving window assistance mode with automatic adaptation of the assistance level.

5 Implementation of six DoF haptic gait simulation and training assistance

5.1 Robotic six DoF haptic foot device *HapticWalker*



Figure 5.1: Robotic gait trainer *HapticWalker* with patient and therapist

The developed haptic control framework for cooperative gait training was implemented and evaluated on the *HapticWalker*, an end-effector based walking simulator for neurological gait rehabilitation. Fig. 5.1 shows the rehabilitation robot with patient and therapist.

5.1.1 *HapticWalker*– hybrid parallel-serial robot for gait training

The *HapticWalker* comprises two hybrid parallel-serial robot kinematics, each of which moves one footplate, located at the end-effector. Corresponding to the general principle of end-effector based devices given in Sec.2.2.1 The patient's feet are fixed on the footplates and can be guided on arbitrary trajectories within the workspace. The *HapticWalker* kinematic structure for one side (leg) with its joint coordinates as well as the Cartesian coordinate frames are depicted in Fig. 5.2.

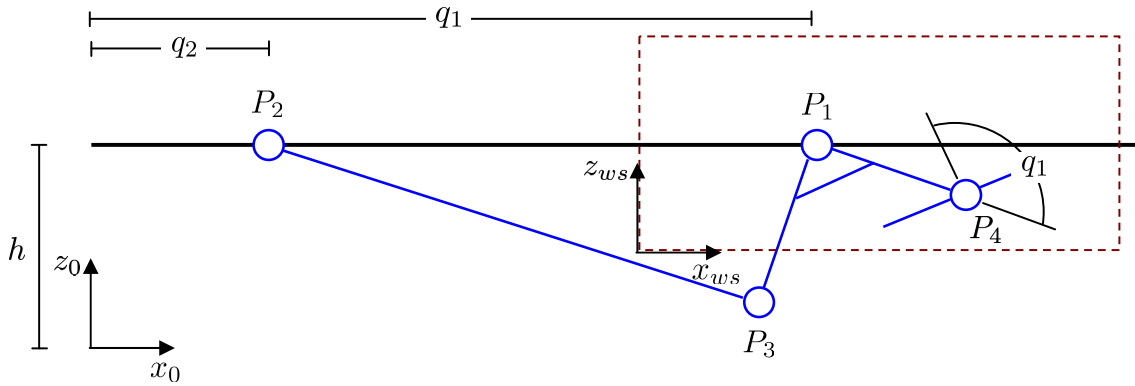


Figure 5.2: *HapticWalker* kinematics with joint, world and workspace coordinates

The device is designed to perform arbitrary planar movements in the sagittal plane for both feet. It therefore consists of two separate robots with three DoF, respectively. Each of the first two joints q_1 and q_2 feature a linear direct drive forming a parallel mechanism. The third joint comprises a geared rotary actuator the footplate angular motion. All drives are brushless DC servo actuators, i.e. their motor controllers feature an integrated cascaded position control loop. The actuators are connected to the PC based control via a SERCOS ring, where the motor controllers are the SERCOS slaves. Corresponding to industrial robots, these position control loops are utilized. They allow for highly dynamic robot motion and good disturbance rejection.

The *HapticWalker* comprises output position encoders to measure the position of the robot joints. For the force measurement a six DoF force-torque (FT) sensor is placed beneath each footplate. The FT sensors contain an additional six DoF accelerometer. The sensor data is amplified and digitalized within the sensor and transferred using a proprietary protocol. All twelve channels from each sensor are captured via a special DSP PCI board.

The robot control system runs on a RTAI Linux industrial PC (IPC) system. Already available was a kinematics module, which computes the robot's forward and backward kinematics, Jacobian matrices and more. Further point-to-point and trajectory movement functions are included. The set of cyclic walking trajectories already implemented in the control system, are described in the next section. More details on the general software architecture and the most important modules of the control system are given in Sec. 5.2.

Definition of training task

The training task is defined by the therapist operating the *HapticWalker*. It can be an arbitrary desired movement within the device's work space. According to the assumption, that the tasks needed by the patient later on in daily life should also be the ones practiced during the rehabilitation process, currently three different training scenarios, i. e. nominal trajectories, from everyday's life are implemented in the *HapticWalker*:

- even floor walking
- stair climbing: upstairs
- stair climbing: downstairs

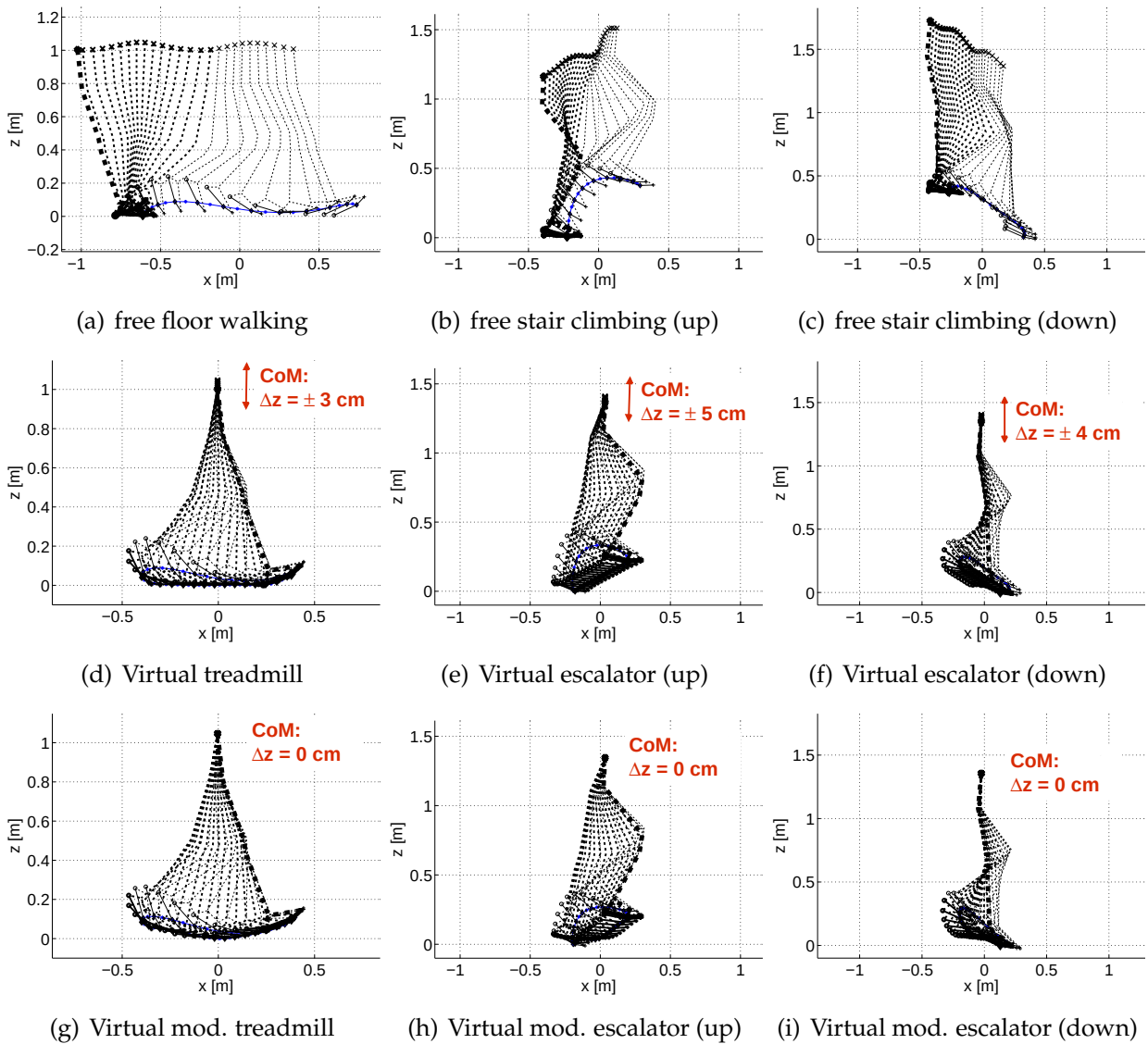


Figure 5.3: Captured Motion tracking data of floor walking, climbing upstairs and climbing downstairs (a - c) and the derived *HapticWalker* nominal training trajectories (d - f). CoM-relative trajectories incorporated the vertical CoM motion in the foot trajectory (g - i)

All foot trajectory types were recorded from healthy subjects using a marker based motion tracking system and transformed into cyclic robot motion trajectories [92]. The interpolation uses a Fourier decomposition of the periodic trajectories. The trajectories can be adjusted offline using the three variable parameters length, height and foot angle range. During the training the velocity can be changed online by the therapist operating the *HapticWalker*.

For each trajectory type two training modes are available. These are the normal mode, which simulates walking on a treadmill or on an escalator respectively, and the CoM-relative mode. Compared to the normal training mode the CoM-relative mode takes advantage of the freely programmable footplates to support the vertical trunk movement by offline incorporating it into the footplate trajectory, see [72]. Figure 5.3 shows the foot trajectories for the available training scenarios and both *HapticWalker* training modes. Moreover Fig. 5.3 a - c show the captured free walking movements, which the nominal training trajectories were derived from.

5.1.2 Low level control design – stable rendering of training environments

Closed loop dynamics of position controlled device

For the control design the complementary sensitivity T_D is needed to determine the bandwidth of the position controlled device. Instead of analytically computing T_D , the complementary sensitivity is approximated by a 2nd order low pass eq. 5.1.

$$\hat{\mathbf{T}}_D = \text{diag} \left(\frac{1}{s^2 + a_1 s + a_0} \right) \quad (5.1)$$

Therefore the device's step response is measured and the parameters of the 2nd order system are computed using a least squares estimation. The motion commands of the control system always rely on a movement interpolator, which takes care of the limitations of the drive dynamics. Hence the steps are not real unit step responses, but smoothed step functions. Consequently the magnitudes of excitation for higher frequencies are smaller compared to an ideal step function.

	$\hat{T}_{dp,x} = \frac{x_h}{x_d}$	$\hat{T}_{dp,z} = \frac{z_h}{z_d}$	$\hat{T}_{dp,\phi} = \frac{\phi_h}{\phi_d}$
TF	$\frac{1}{1+2.072 \cdot 10^{-2} s + 1.832 \cdot 10^{-4} s^2}$	$\frac{1}{1+2.074 \cdot 10^{-2} s + 1.82 \cdot 10^{-4} s^2}$	$\frac{1}{1+2.076 \cdot 10^{-2} s + 1.72 \cdot 10^{-4} s^2}$
bandwidth	11.7627 Hz	11.7627 Hz	11.7627 Hz

Table 5.1: Nominal complementary sensitivity transfer functions

To determine the nominal 2nd order closed loop transfer function and the uncertainty the step responses were measured for end-effector positions throughout the workspace. Due to the device's kinematic structure, the dynamic characteristics vary only with respect

to the vertical end-effector position. To move the end-effector along the x -axis means equally moving the joints q_1 and q_2 . In this case the configuration of the kinematic chain remains the same. 14 measurements were performed to cover the complete height of the workspace. Fig. 5.4 shows the movements performed for identification. The figure is a schematic diagram and the dimensions are not to scale. The starting position on the x -axis was always in the middle of the workspace ($x = 0.45\text{ m}$) and on the z -axis varied from $z = 0\text{ m}$ to $z = 0.4\text{ m}$ in 0.05 m steps. The movement amplitude in both x and z direction were $\pm 0.1\text{ m}$ and the angle was $+45^\circ / -30^\circ$. The black dots are the starting positions, the arrows represent the movement direction of a single trial and the red lines depict the corresponding end-effector pose at the start and target position respectively.

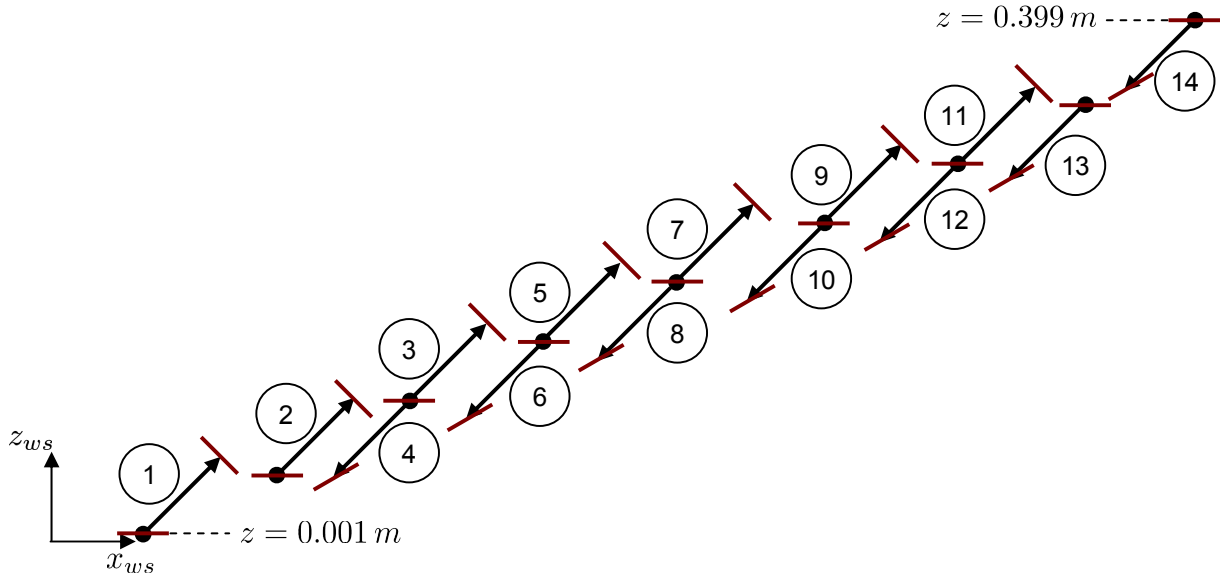


Figure 5.4: Identification measurements: black dots are the starting positions, arrows symbolize the movement direction and red lines the footplate pose at start and target positions

In Fig. 5.5 the magnitudes of the 2^{nd} order models identified from the 14 measurements were drawn for the three Cartesian coordinates. It is seen, that the variation in transfer behavior throughout the workspace is almost negligible. Further the bandwidth in all directions is very similar.

This behavior of the position controlled device is advantageous for the implementation. A nominal set of parameters can be determined by simply averaging the transfer function coefficients. There is no danger of violating the constraint of $\|\Delta\|_\infty < 1$. The nominal closed loop dynamics for each Cartesian direction as well as the corresponding bandwidth of the position controlled system is given in Tab. 5.1.

Environment stiffness bound

The maximum allowable weight for patients on the *HapticWalker* is 135 kg . Together with the object stiffness in the simulation this defines the maximum static penetration of the

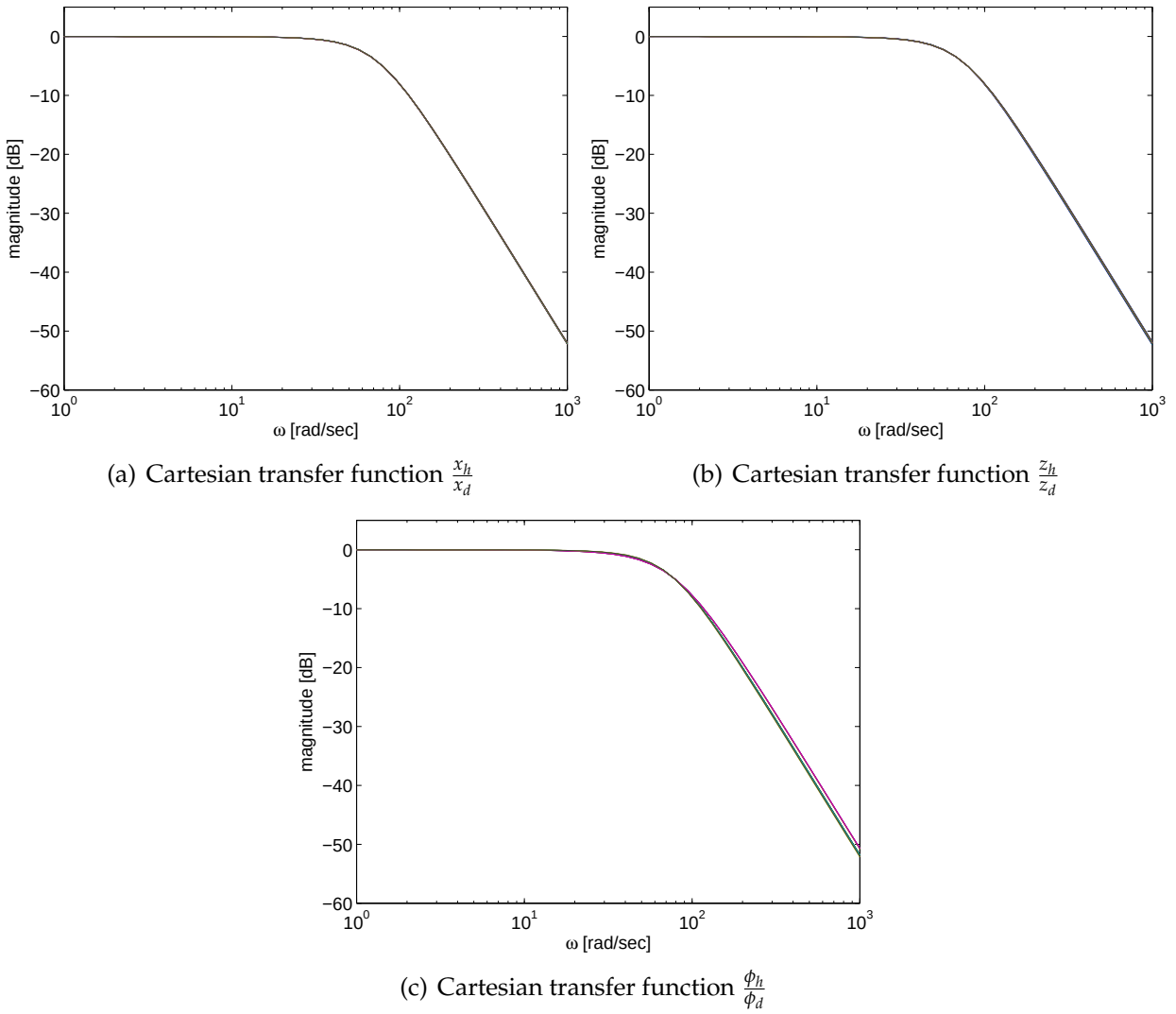


Figure 5.5: Magnitudes of the closed loop transfer functions estimated from measured step responses throughout the workspace

probes into objects. If the stiffness of weight bearing objects is too low, they feel soft. A large penetration into the object could further lead to a break through for smaller objects as well as a violation of workspace boundaries.

object type	patient weight [kg]	stiffness [$\frac{N}{m}$]	static penetration [m]
weight bearing objects	135	10^5	0.0135
assistance window	135	5000	0.27

Table 5.2: Stiffness parameters for different object types

The stiffness of the weight bearing objects, e. g. the floor or stairs, and the stiffness of the assistance window are needed for the design of the contact and free motion coupling parameters, respectively. The parameters used for this implementation on the *HapticWalker* together with the allowed patient's weight and resulting static object penetrations are listed in Tab. 5.2.

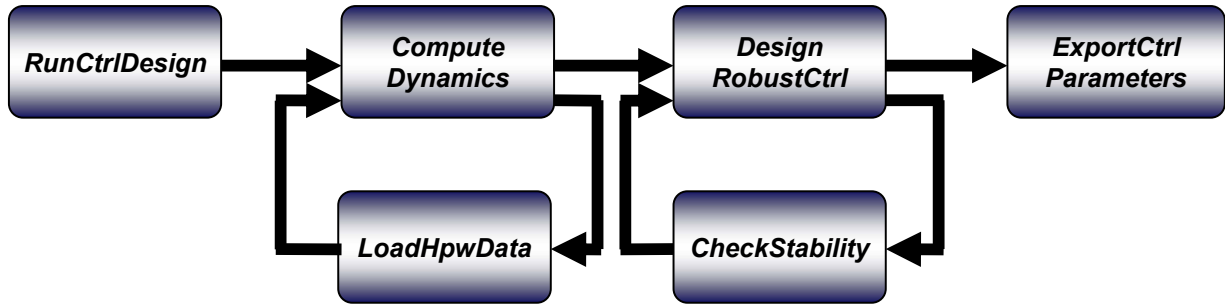


Figure 5.6: MATLAB® tool chain for the coupling controller design

Coupling parameters for robust interaction stability

The concept of continuous adaptation of the coupling gain according to the virtual environment allows for an arbitrary number of predefined parameter sets for different environments. In the implementation on the *HapticWalker* currently only two parameter sets are needed:

1. one set for free motion, which is designed for the assistance window stiffness and
2. a second for contact with the weight bearing environment objects.

Parameter set	Direction	Mass m_t	Damping b_t	α []
1. free motion	translatory	5.0 kg	1245.0 $\frac{Ns}{m}$	0.2
	rotatory	0.3 kgm ²	126 $\frac{Ns}{rad}$	0.2
2. contact	translatory	5.0 kg	7951.0 $\frac{Ns}{m}$	0.2
	rotatory	0.3 kgm ²	956 $\frac{Ns}{rad}$	0.2

Table 5.3: Coupling controller parameter sets for free motion vs. contact transition and virtual environment interaction

Tab. 5.3 shows the controller parameters for both situations occurring during the haptic simulation. α was chosen empirically. For small values of α the damping for low frequencies is too small and the footplate overshoots when the user force jumps to zero. $\alpha = 0.2$ proved to be a good choice. The mass and damping parameters were computed in MATLAB® as described in Sec. 4.2.2 and using the condition in eq. 4.17 assuming $\|W_1\|_\infty = \|W_2\|_\infty \leq 1$.

Fig. 5.6 shows the MATLAB® tool chain for the design of the coupling controller parameters. The script *RunControllerDesign* initializes basic constants, e.g. sampling time and target environment impedance as well as filenames for data sources. The second function implements the computation of the nominal device dynamics and closed loop bandwidth. In the third stage the controller parameters are iteratively determined as described in Sec. 4.2.2, where the robust stability condition is evaluated in the function *CheckRobustStability*. The last step is to export the computed controller parameters into an ASCII file, which can be read by the *HapticWalker* control system.

5.1.3 Task level haptic control of the *HapticWalker*

Available training tasks in haptic control mode

The assistance modes, described in the concept, rely on the elliptic window as their major component. This window is always associated with a task specific nominal training trajectory. In the current implementation this can be one of the cyclic trajectories with vertical CoM motion, depicted in Fig. 5.3 d - f.

	non-compliant	haptic (moving window)	haptic (virtual tunnel)
treadmill (normal)	x	x	x
treadmill (CoM-rel.)	x		
upstairs (normal)	x	x	x
upstairs (CoM-rel.)	x		
downstairs (normal)	x	x	x
downstairs (CoM-rel.)	x		

Table 5.4: Available training tasks (i. e. trajectories) and assistance modes

Implementation of *assist as needed* training modes

Both assistance modes from the concept are available in the current implementation.

For the experiments conducted for evaluation of the system the semi-axes of the windows are selected equal, which makes the window a circle. The selected basic parameters of the window are shown in Tab. 5.5

Assistance Mode	$\hat{C}_w \left[\frac{N}{m} \right]$	inner radius $r_{min} [m]$	outer radius $r_{max} [m]$
window / tunnel	5000	0.02	0.03

Table 5.5: Assistance window parameter set

5.1.4 Force data conditioning

For the proper functionality of the haptic control system a clean force signal is required. The raw sensor signal is disturbed by a constant bias, measurement noise and the inertial forces of the tool, which is mounted on the sensor. On the *HapticWalker* the tool is the footplate with the active safety release binding.

The force sensor signal is conditioned within the following three successive preprocessing stages:

- A. bias correction
- B. low pass filtering
- C. compensation of tool inertia

Throughout this section the force and acceleration sensor outputs are divided into their translational and rotatory components according to eq. 5.2.

$$\begin{aligned} \mathbf{f} &= \begin{bmatrix} f_x & f_y & f_z \end{bmatrix}^T, & \mathbf{t} &= \begin{bmatrix} t_x & t_y & t_z \end{bmatrix}^T \\ \mathbf{a} &= \begin{bmatrix} a_x & a_y & a_z \end{bmatrix}^T, & \boldsymbol{\omega} &= \begin{bmatrix} \omega_x & \omega_y & \omega_z \end{bmatrix}^T \end{aligned} \quad (5.2)$$

A Bias correction

Force sensor signals are always biased by a constant offset, which differs from usage to usage. This constant bias is computed once during the machine startup procedure when the footplates have reached the entry position. The bias values are stored in the DSP (Digital Signal Processor) on the PCI sensor board and internally subtracted from the raw signal. However the force measured in the entry position contains additionally the tool gravitational force, which has to be subtracted. The constant sensor bias results to

$$\begin{aligned} \mathbf{f}_{bias} &= \mathbf{f}_{raw,bias} + m_{tool} \mathbf{g} \begin{bmatrix} \sin(\phi) \\ 0 \\ \cos(\phi) \end{bmatrix}, \\ \mathbf{t}_{bias} &= \mathbf{t}_{raw,bias} + \mathbf{r} \times m_{tool} \mathbf{g}. \end{aligned} \quad (5.3)$$

where g is the gravitational acceleration and $\mathbf{g} = g\mathbf{e}_z$ the corresponding vector. The unbiased raw force sensor output is

$$\begin{aligned} \mathbf{f}_{raw} &= \mathbf{f}_{raw,bias} - \mathbf{f}_{bias}, \\ \mathbf{t}_{raw} &= \mathbf{t}_{raw,bias} - \mathbf{t}_{bias}. \end{aligned} \quad (5.4)$$

B Low pass filtering

To reduce measurement noise the bias corrected raw force signal is low pass filtered. The DataAcquisitionModule provides filters for all input data. The filters are 2nd order Butterworth low pass filters with selectable cutoff frequencies up to half the control system frequency, which is currently 500 Hz.

From literature the bandwidth of human voluntary motor functions is known to be in the range of 7 Hz - 15 Hz, see e. g. [84]. This fact suggested a force filter cutoff frequency around 10 Hz. However the data from literature usually refers to upper extremity human machine interaction. In the application on the *HapticWalker* a cutoff frequency of 10 Hz led to a shaky interaction behavior in free motion. Possible reasons could be:

- different voluntary motor control characteristics and fine motor skills in lower compared to upper extremities,
- additional unmodeled effects from the permanent foot machine contact enforced by the binding.

The straight forward solution for this problem, which was used for this implementation, was to reduce the cutoff frequency of the force filter. The best results were obtained with a value of 3 Hz. For a filter frequency below 3 Hz the fidelity of the free motion simulation reduced noticeably, which made the robot motion overshoot when free motion direction was changed.

C Compensation of tool inertia

The bias corrected, filtered force signal results from two influences. It is a combination of the inertial and gravitational forces of the tool mounted behind the sensor (in this case the safety release binding) and the human-machine interaction forces. To determine the effective input forces, caused by the operator, the tool inertia must be compensated for.

The acceleration of the tool's CoM, which is needed for the inertia compensation, is computed from the end-effector point acceleration. This can be acquired from two possible sources on the *HapticWalker*. Firstly the acceleration can be computed by numerical differentiation from the end-effector position. However the delay introduced by numerical differentiation is a disadvantage. Secondly the end-effector acceleration can directly be derived from the force sensor acceleration, which is measured by the six DoF accelerometer.

To avoid the aforementioned delay from numerical differentiation, the accelerometer data is used in this implementation. The gravitational acceleration is not eliminated in the measurement and therefore is still contained in the acquired data. Hence gravitational and dynamic inertial forces need not be computed separately. The compensation forces and torques are

$$\begin{aligned} \mathbf{f}_{tool} &= m_{tool} \mathbf{a}_{tcp}^{(C)} \quad \text{and} \\ \mathbf{t}_{tool} &= \mathbf{r} \times (m_{tool} \mathbf{a}_{tcp}) + I_{tool}^{(C)} \boldsymbol{\omega} \quad , \end{aligned} \tag{5.5}$$

respectively.

The parameters, which need to be identified for the compensation, are the tool mass m_{tool} , the vector from the tool center point and the tool's CoM $\mathbf{r}_{tcp,C}$ and the moment of inertia I_{tool} . It turned out that using a single parameter set led to a model, which could only be tuned to compensate the tool forces well in some spatial directions. At least in one direction the compensation remained poor. The reason for this remains unclear. Yet instead of modeling non-uniform, non-linear sensor characteristics and influences from sensor mountings, the remedy was to use separate parameter sets for each Cartesian direction x , z and ϕ . Thereby a good dynamic compensation of tool inertia could be achieved.

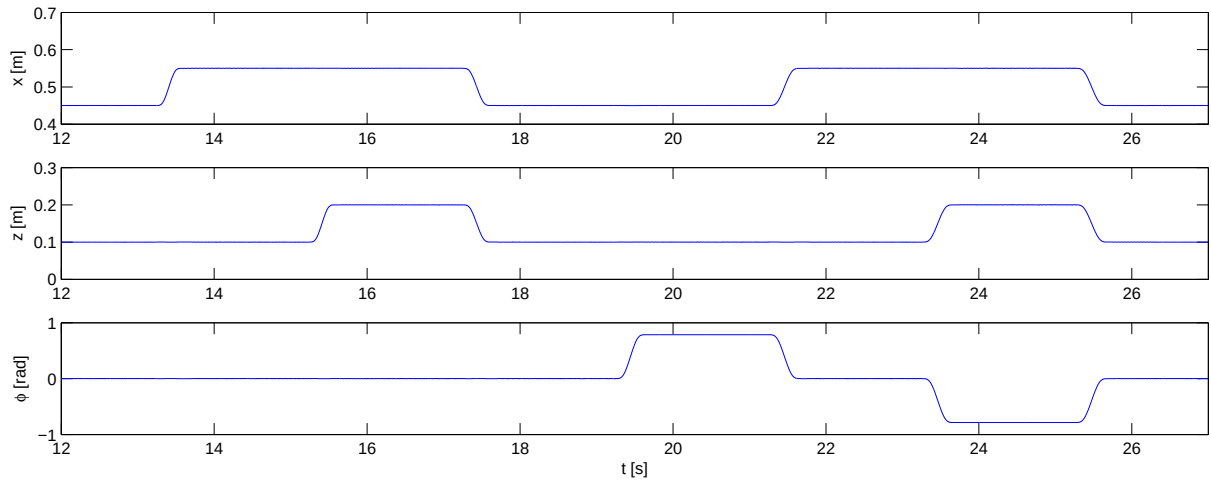


Figure 5.7: Identification trajectory for compensation model parameters

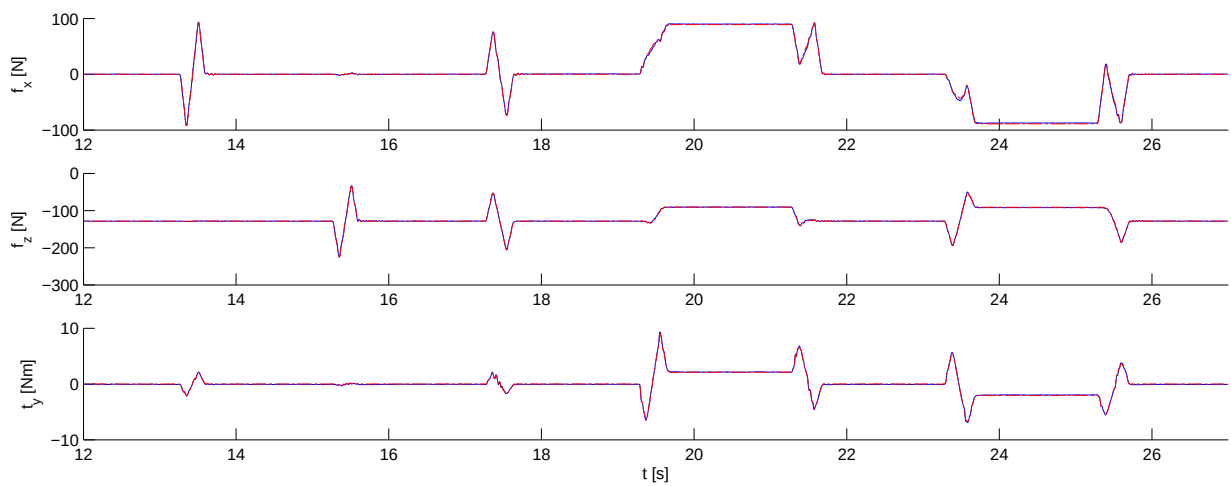


Figure 5.8: Measured vs. estimated forces and torques in the sagittal plane during identification measurements

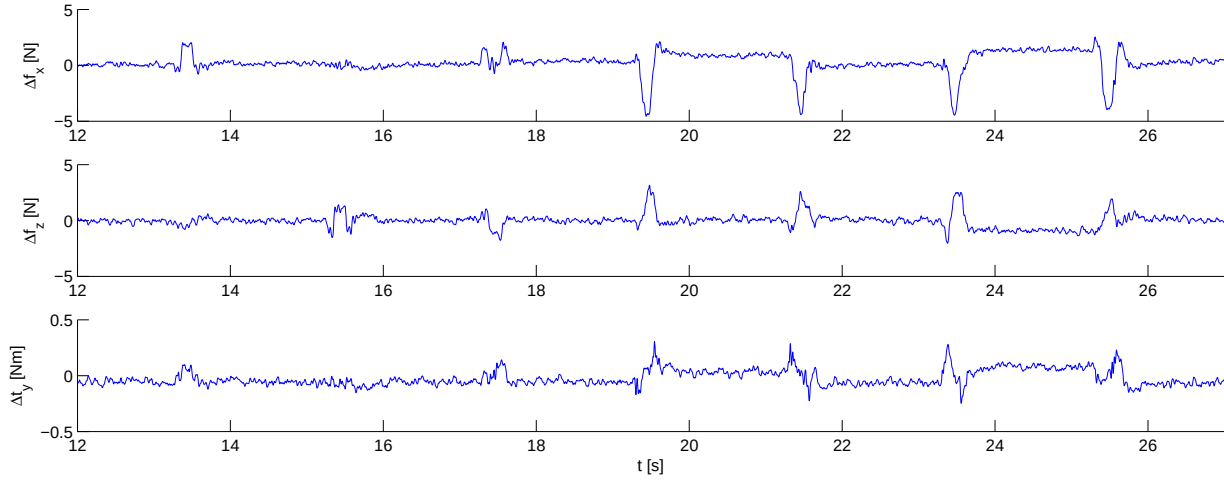


Figure 5.9: Remaining compensation error – forces and torques in the sagittal plane

For parameter identification, the trajectory depicted in Fig. 5.7 was performed. It comprises step-like movements exciting single as well as multiple force channels at a time. The parameter sets were then determined by a least squares algorithm minimizing the error between measured and estimated inertial forces. Fig. 5.8 shows the measured versus the estimated forces.

(a) right side			(b) left side		
Parameter	Unit	Value	Parameter	Unit	Value
m_x	[kg]	12.83	m_x	[kg]	12.50
$\mathbf{r}_x$	[m]	$\begin{bmatrix} -0.032 & 0.018 \end{bmatrix}$	$\mathbf{r}_x$	[m]	$\begin{bmatrix} 0.013 & 0.019 \end{bmatrix}$
m_z	[kg]	13.01	m_z	[kg]	13.18
$\mathbf{r}_z$	[m]	$\begin{bmatrix} -0.001 & -0.002 \end{bmatrix}$	$\mathbf{r}_z$	[m]	$\begin{bmatrix} -0.001 & -0.027 \end{bmatrix}$
m_{phi}	[kg]	16.60	m_{phi}	[kg]	13.00
$\mathbf{r}_{phi}$	[m]	$\begin{bmatrix} 0.0 & 0.018 \end{bmatrix}$	$\mathbf{r}_{phi}$	[m]	$\begin{bmatrix} 0.002 & 0.020 \end{bmatrix}$
I	[kgm ²]	0.175	I	[kgm ²]	0.152

Table 5.6: Parameters for inertial compensation of tool mass for right (a) and left (b) side

Tab. 5.6 shows the parameters determined for the right and left side, respectively.

The remaining error between the measured and the estimated forces from Fig. 5.8 for the performed identification measurements is plotted in Fig. 5.9. Due to sensor noise, non-linearities and parameter uncertainties the compensation will never be perfect. With respect to the maximum measurement range of the sensor the remaining error lies between 0.1 % and 0.2 %. To avoid erroneous detection of interaction forces, values below

the remaining error level are set to zero according to eq. 5.6.

$$f_{out} = \begin{cases} f_{mess} - f_{min} & f_{mess} \geq f_{min} \\ 0 & \text{else} \end{cases} \quad (5.6)$$

The absolute minimum values for interaction forces and torques in the sagittal plane, which can be detected are listed in Tab. 5.7.

$f_{min,x} [N]$	$f_{min,z} [N]$	$t_{min,y} [Nm]$
5.0	5.0	0.5

Table 5.7: Minimum detectable interaction forces and torques in the sagittal plane

5.2 Control software architecture

5.2.1 RTAI-Linux based control system framework

The control software used for the implementation of the developed control schemes for patient-adaptive training scenarios is based on the realtime operating system RTAI-Linux. The control framework and *HapticWalker* control system consists of two main parts:

- A. RT Linux kernel thread and
- B. Non-RT user space threads.

The non-RT and RT parts of the control software communicate via FIFO (first in first out) shift register pairs in a shared memory region. This FIFO communication mechanism synchronizes asynchronous events from the non-RT system with the RT thread cycle. Fig. 5.10 provides an overview of the control software structure and the communication mechanisms throughout all levels from the user input to the robot hardware. Relevant parts of the system will be looked upon in the following sections.

A RT Linux kernel thread

The cyclic realtime thread is started when the core modules, mainly *RTServer* and *RTServo* are loaded. After that any device specific kernel modules can be loaded and register periodic handler functions, which are then called in the thread context in every cycle. Additionally asynchronous RT-command handlers can be registered, which handle commands that are issued from the non-realtime context.

Within the realtime kernel thread data acquisition, setpoint generation and system image logging are handled by device specific modules. It directly accesses device drivers to perform I/O operations.

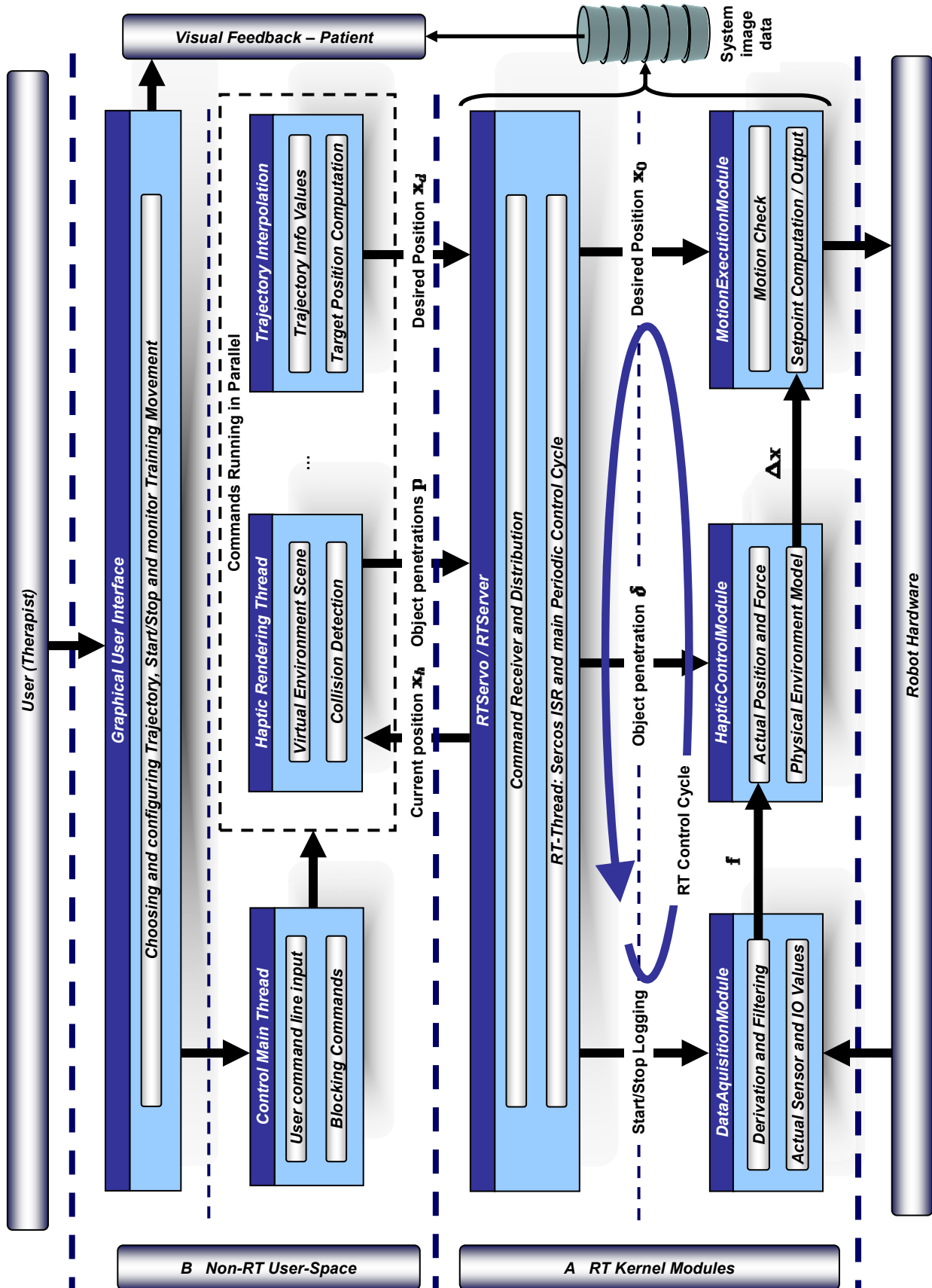


Figure 5.10: General control software structure used for implementation of the haptic control framework on the *HapticWalker*

The thread is triggered either by a system timer interrupt or a hardware I/O interrupt. For *HapticWalker* control system the cycle time of the SERCOS bus, which performs the communication to and from the servo drives, defines the realtime thread timing. Therefore the thread is triggered on the interrupt generated by the SERCOS PCI board.

The device specific implementation developed within this thesis uses mainly three additional kernel modules on top of the core framework:

DataAcquisitionModule is called first in the realtime thread. All data from the robot's sensors is collected in this module. Numerical computation of axes and Cartesian position derivatives is performed and the acquired data is filtered. Further the tool inertia is compensated from the force sensor data.

HapticControlModule is called after data acquisition and preprocessing performed in the DataAcquisitionModule. It comprises a state machine to perform the transition from position controlled to haptics mode. When the haptics mode is enabled the interaction forces produced in the virtual environment are computed in two separate models for each robot side, i.e. for each foot (see Sec. 5.2.2). From the measured operator forces and the virtual environment's interaction forces the resulting end-effector motion is computed via the coupling controller.

MotionExecutionModule is the last module called in the realtime control cycle. It provides another state machine providing the startup procedure and switching between motion states. In this module desired position values and motion offsets computed by the HapticControlModule are combined to the setpoint. Before the position setpoint is transferred to the drives, its validity regarding the robot workspace and allowed velocities, accelerations and jerks is checked.

B Non-RT user space threads

The user space application starts the *control main thread*. Within this thread runs a command line interface to operate the device. During startup the program loads the framework and device specific RT kernel modules and registers a set of commands, that can be accessed via the command line prompt by entering their command alias and additional arguments.

In addition to directly entering commands into the command prompt, the console interface can be mirrored via TCP/IP. The mirrored console can be accessed by sending a string, which contains the command alias and arguments. The command issued via the mirrored console is executed as if entered in the command prompt. The mirrored console interface is used for the graphical user interface (GUI).

Commands running in the user space can either run within the main thread context or in a separate one. The first type is called *blocking command* the latter *parallel command*. Commands executing robot motions and the haptic control command are examples for parallel commands running in their own user space thread. They can either only affect

the non-RT application or issue a command to a RT kernel module. The communication between RT and non-RT components is implemented through FIFO pairs, which lie in a shared memory area. One of these FIFO pairs is permanently reserved for the *control main thread*. Commands started as parallel thread reserve one of the free communication resources for the execution period. To pass commands to a RT kernel module, a command block, containing the kernel space command type and optional data, is written to the command FIFO. The commands are dispatched and distributed to the respective destination module by the RTServo/RTServer kernel modules at the beginning of each RT cycle. An answer can be written by the respective receiving kernel module into the answer FIFO. After receiving the appropriate answer and finishing the command execution the user space application returns to command prompt (blocking command) or frees the communication resource (parallel command).

5.2.2 Multi-threaded haptic rendering system

Using the existing software framework and the *HapticWalker* motion control based thereon, additional software modules were developed to integrate the haptic control features into the system. A GUI allows for comfortable operation of the device and a 3D visualization of the training scenario provides visual feedback to the operator using the haptic walking simulation.

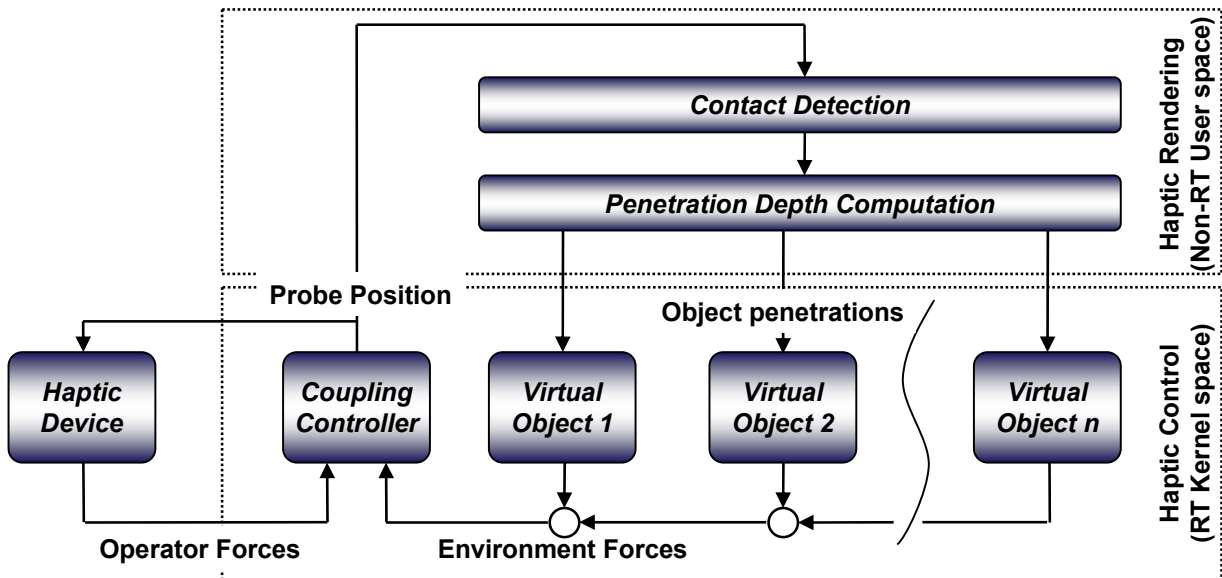


Figure 5.11: Software scheme for multi-threaded haptic simulation of virtual walking environments

The software structure implementing the haptic rendering and control system consists of two interacting threads:

- A. *Haptic rendering thread* – Contact detection and penetration determination within the virtual environment running in user space as parallel command and

- B. *Haptic control module* within the RT kernel thread – RT computation of virtual object interaction forces based on mechanical object models and calculated penetrations.

Fig. 5.11 shows the multi-threaded haptic rendering system with the *haptic rendering thread* (upper dashed box) running at a high priority in non-RT user space and the low level *haptic control thread* (lower dashed box) running as a kernel module within the RT control context. In Fig. 5.10 the *haptic control thread* corresponds to the RT kernel thread executing all periodic and command handlers of the kernel modules loaded. The main part however is located in the *HapticControlModule*, which will be addressed in this section. The single block on the left side represents the robot hardware rendering velocities or positions to the operator and measuring his forces as input to the haptic simulation.

A Haptic rendering thread – Contact detection and penetration determination

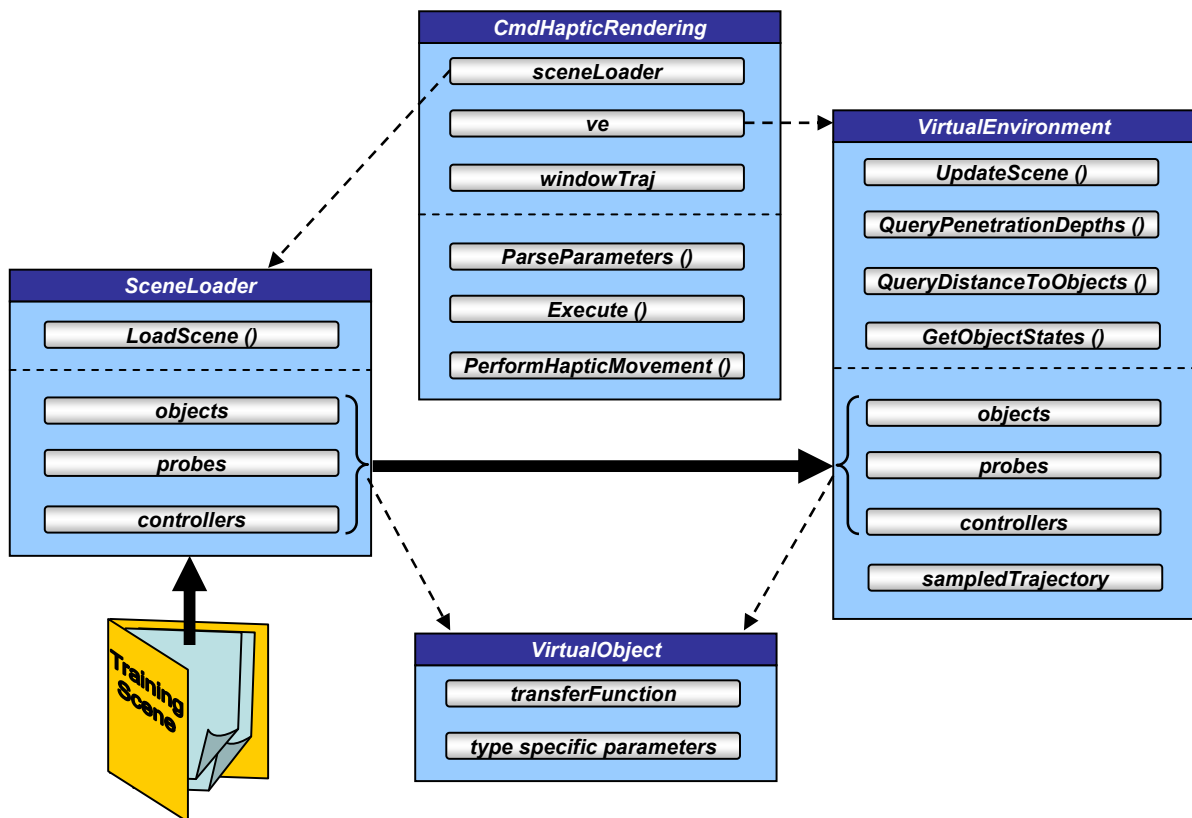


Figure 5.12: Class diagram of the user space haptic rendering application (solid bold lines indicate data transfer, thin dashed lines indicate that an attribute is of the appointed class)

Fig. 5.12 shows a simplified class diagram of the most important classes of the haptic rendering user space application and their major methods and attributes. The single classes, their functionality and their interaction will be detailed in the following paragraphs.

CmdHapticRendering

The computation of contact states between virtual probes and fixed and moving objects runs in a separate user space thread, which is invoked via the parallel command interface

provided by the core framework. The main class is *CmdHapticRendering*, which implements the command and is derived from the framework class *BasicMovementCommand*. It contains a *SceneLoader* (sceneLoader) to parse the training scenario directory, a *VirtualEnvironment* object (ve) for collision detection and penetration computation and a trajectory interpolator (windowTraj) to generate the sampled trajectory. The sequence of actions

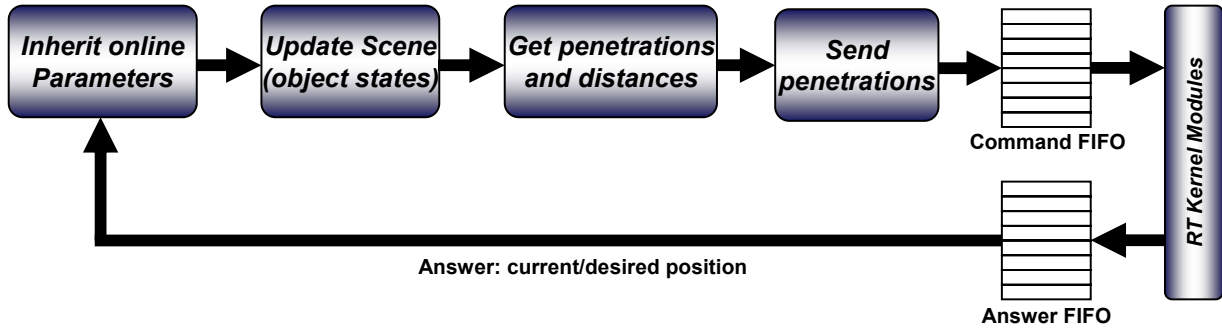


Figure 5.13: *CmdHapticRendering* execution loop

during the rendering loop is shown in Fig. 5.13. Parameters, which are adjustable online, are the desired training cadence and the guidance and support levels. Update of the scene and penetration and distance computation is called via the *VirtualEnvironment* object *ve*. The communication with the Linux kernel RT system runs through the shared memory FIFO message system, which is provided by the framework. The FIFOs for the exchange of penetrations and current end-effector positions have a length of one single block. This is important to force the realtime thread to read transferred messages in every RT cycle and hence always use the newest penetration data. In contrary to classic motion control applications a buffer underrun is tolerated. The execution of the loop is further triggered on the response from the RT kernel thread containing the current and desired position. Ideally this means that the rendering loop runs at the same rate as the RT control cycle.

	scene	trajectory switch	trajectory type	format	trajectory	controller
1	dir	''	- treadmill	-	-	name
2	dir	-t	upstairs downstairs	f p	filename coeffs	name

Table 5.8: Command line arguments for the command alias *hapticRendering*

The command is started by the alias *hapticRendering* and accepts several command line arguments listed in Tab. 5.8.

VirtualEnvironment and VirtualObject

VirtualObject is the class, which is used to store all object, probe and controller parameters loaded from the scenario's directory. It comprises transfer function parameters, valid for all types of objects, and specific parameters depending on the object type.

The class *VirtualEnvironment* includes the loaded objects, probes and controllers and performs contact detection and penetration computation. Therefore it uses the external C++ library *SWIFT++*, developed at the University of North Carolina at Chapel Hill [36]. This library allows for the definition of polygonal objects and probes in 3D and allows for the computation of the shortest distance vector between any two objects/probes. When objects collide and overlap no valid value is returned. To compute object penetrations by the probes the actual object is a shrunk version of the desired one. The probe then penetrates the virtually expanded object when it approaches nearer than the shrinkage distance. The difference between shrinkage and actual distance to the shrunk object is the penetration. By shrinking the object the corners of polygonal objects are rounded, which is not an issue in this application. This is shown in Fig. 5.14 The *SWIFT++* li-

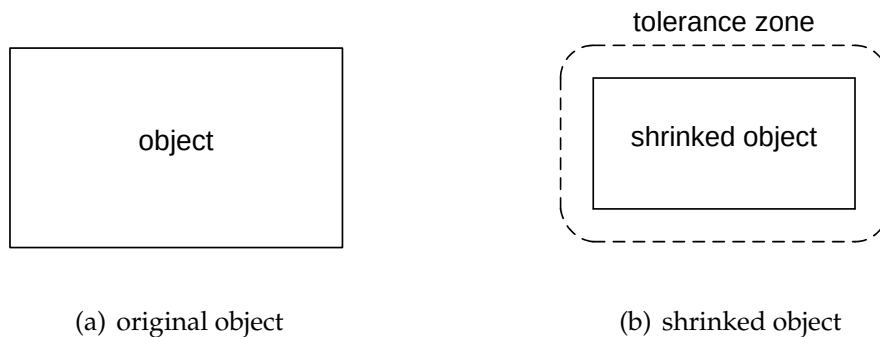


Figure 5.14: Object shrinking to compute virtual penetrations from distances of polygons

brary is used to handle contacts between probes and weight bearing environment objects, e. g. floor tiles and steps. For the definition of assistance window objects and workspace boundaries, as well as the corresponding collision detection and penetration computation, new functions were implemented. In addition to the contact states and penetrations the shortest distances between objects and probes need to be computed and transferred to the *HapticControlModule*. They are used to schedule the coupling controller gain in the vicinity of contact.

SceneLoader and training scenarios

The *SceneLoader* class is used to parse a training scenario, load all objects, probes and controllers and pass them to the *VirtualEnvironment* object. Each training scenario is represented by a folder containing ASCII configuration files for all objects, probes and controller parameter sets. The configuration files are generated via MATLAB[®] scripts. The various parameter sets are distinguished via the file endings. Following endings are valid:

- *.obj for virtual objects forming the scenario and the assistance,
- *.pro for the probes, i. e. the virtual feet and
- *.ctl for controller parameter sets, designed for stable interaction with certain environment bounds.

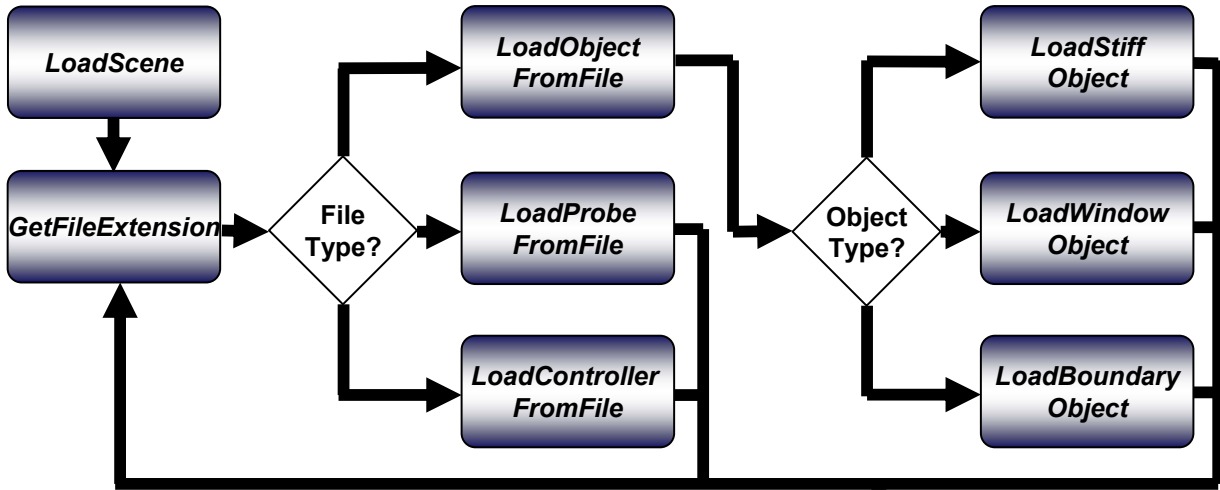


Figure 5.15: Dynamic loading of training scenario during runtime

Fig. 5.15 presents a flow chart of the loading procedure for training scenarios. This procedure is run before actually enabling the haptic control realtime module and contact detection in user space. The selected scenario path is passed as parameter to the command starting the haptic control mode. It is therefore loaded dynamically during control system runtime, i. e. when the *hapticRendering* command is restarted the complete scenario is reloaded and all changes in the ASCII files become valid.

B Haptic control module – Realtime virtual object mechanical models

Within the RT control cycle the *HapticControlModule* handles the computation of the virtual environment forces. The module contains a state machine to handle startup and shutdown of the haptic rendering mode. The state machine is shown in Fig. 5.16. When

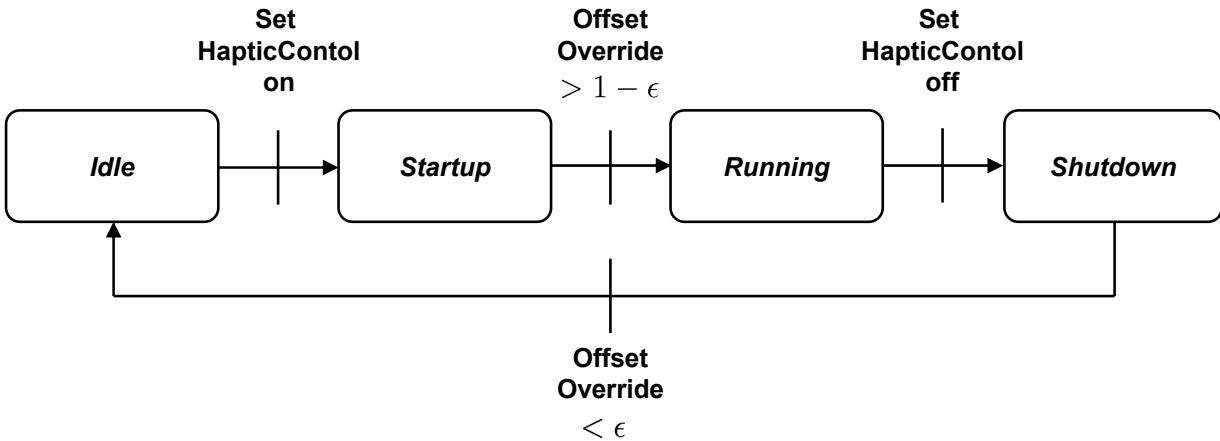


Figure 5.16: Haptic control state machine in RT kernel module

the module is loaded the state machine starts in the *Idle* state. It switches to *Startup* when the RT-command *SetHapticControl on* is issued by the user space command *hapticRendering*. Within the *Startup* state the setpoint offset computed by the haptic control is continuously increased using a 2nd order low pass step response. The corresponding scaling factor is called override offset. When the offset override is larger than $1 - \epsilon$ with $\epsilon = 10^{-4}$

the state machine switches to *Running*. On the RT-command *SetHapticControl off* the state machine switches to *Shutdown* and decreases the offset override until it is smaller than ϵ and returns to the *Idle* state.

To compute the virtual interaction forces the module integrates functions to define and solve block based models. According to the concept described in section 4.3 the virtual environment block is synthesized from a set of cascaded blocks, each of which corresponds to an object in the scenario. At least three blocks are present in every haptic simulation: 1) Input, 2) Output and 3) coupling controller. During the loading procedure the controller parameters are configured and all other objects in the training scenario are added as transfer function blocks in the configuration shown in Fig. 4.15. For the present implementation the controller and the object blocks are all discrete linear transfer functions. The controller additionally implements the algorithm for gain scheduling.

5.2.3 Graphical User Interface

For a user friendly operation of the control system the GUI was developed on top of the command line console interface. The GUI is implemented using the graphical interface library QT by Trolltech. The GUI consists of following tabs:

- database interface to select a patient for the next training session (Fig. 5.17 a),
- splash screen through SERCOS boot (Fig. 5.17 b) and reference tab (Fig. 5.17 c) to control machine start up,
- training mode selection – training scene and haptic or non-haptic mode (Fig. 5.18 a),
- training parameterization – offline adjustment of scene geometry, online adjustment of cadence and assistance (Fig. 5.18 b).



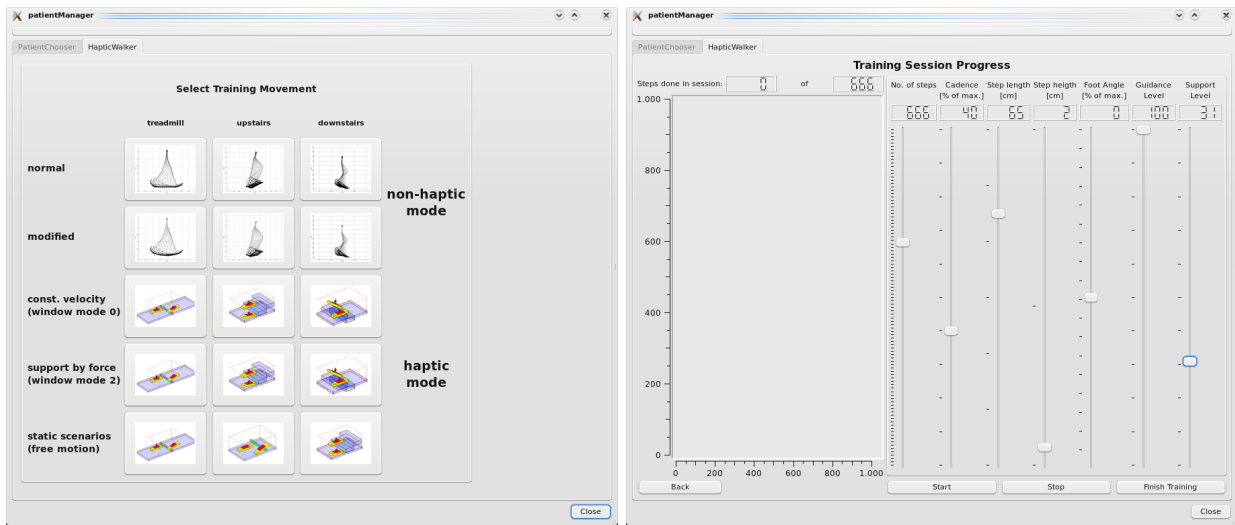
(a) Database interface to load patient data

(b) Splash screen during SERCOS boot

(c) Reference screen with checklist for actions before start of drive reference procedure

Figure 5.17: Startup windows of the control system's GUI

In the future the database interface will be used to access and manage patient data within clinical studies. It is the first screen displayed when starting the control system and can afterwards be accessed to load new patient data and settings and initiate a new training



(a) Menu to select training mode

(b) Offline and online modification of training parameters

Figure 5.18: Main interface of the control system's GUI for selection and adjustment of training modes

session. The splash screen indicates the progress during SERCOS boot up and switches to the reference screen after SERCOS is completely booted. The reference screen requires the user to check a list of necessary steps and finally start the reference procedure of the machine.

When the machine is completely started and ready for motion, the training mode selection screen shows up. It provides a matrix of buttons to select position controlled (row 1 - 2) and haptic training modes (row 3 - 5) as well as three static demonstration scenarios (row 6). After selecting a training mode the parameterization screen displays. It provides sliders to adjust the number of steps, training velocity, step length, step height, foot angle range, guidance and support (assistance) levels. From these parameters the velocity and assistance levels can be adjusted online.

Further a 3D visualization of the training scene is displayed in 3D for the operator, in the present case the subject or patient, during haptic training mode to facilitate the understanding of the presented training task.

3D Training Scenario Visualization

The visual feedback displays a 3D visualization of the virtual environment and the current position of the footplates therein. Additionally the nominal trajectory, defined for the training task and the position of the window object is shown. The interaction forces at the footplates are envisioned by arrows, which are scaled in length and diameter according to the force value. The arrows point in the direction of the measured force and the tips end at the footplate. Fig. 5.19 shows screen-shots of the visualization of the three training tasks, which are currently implemented on the *HapticWalker*: a) treadmill walking, b) climbing upstairs and c) climbing downstairs.

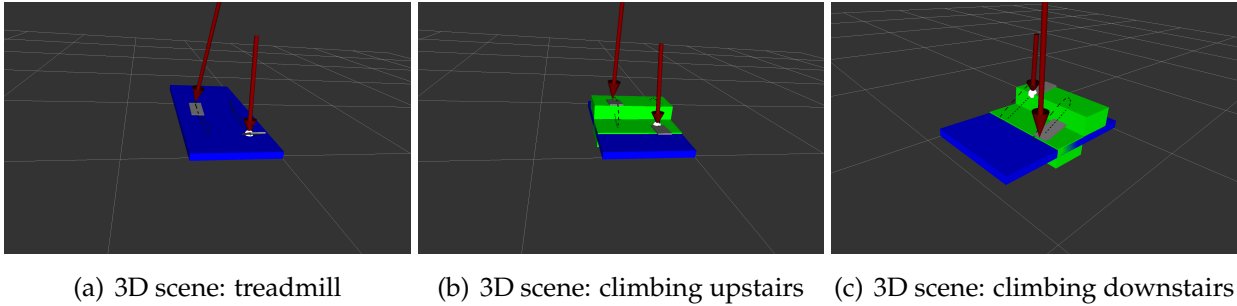


Figure 5.19: 3D visual feedback of training scenarios, current foot position within the virtual environment and foot interaction forces

It uses *libqglviewer*, an open source extension library for QT, which provides an interface to invoke OPENGL windows in a QT application and implements basic functionality. Among other features *libqglviewer* allows to adjust the view on the scene (direction and scale) with the mouse, to render a grid or arrows and to define the frame rate, out of the box.

Visual rendering		Haptic rendering
SceneViewer : public QGLViewer	$\hat{=}$	CmdHapticRendering
WalkingScene	$\hat{=}$	Virtualenvironment
WalkingObject	}	$\hat{=}$ VirtualObject
WalkingProbe		
SceneLoader	$\hat{=}$	SceneLoader

Table 5.9: Structural similarity of haptic and visual rendering implementation: corresponding classes

The implementation of haptic environment rendering and it's corresponding visual feedback are closely related in their design. Hence the class diagram in Fig. 5.12 can be applied to the visual feedback using the correspondents of the respective classes, given in Tab. 5.9. The main class *SceneViewer* inherits the library class *QGLViewer* and handles the initialization of the OPENGL scene. The major attributes are a *WalkingScene* and a *SceneLoader* object.

The *SceneLoader* loads the same scenario folder, which is used by the *hapticRendering* command, and stores the objects and probes in the *WalkingScene*. The nominal training trajectory is obtained from the GUI application. The *WalkingScene* object handles the periodic redraw of the OPENGL window, calling the redraw handler of all *WalkingObject* and *WalkingProbe* objects.

6 Evaluation

According to the evaluation concept proposed in Sec. 4.5 the evaluation of the developed framework is done in three stages addressing different levels of functional complexity.

For the experiments one healthy subject performed test movements and exercises on the *HapticWalker*. The subject was a 31 years old male with no known neurological or motor impairments. His bodyweight was approx. 80 kg. To keep balance the subject was allowed to hold on to the hand rails on both sides. Other than the grip on the rail no bodyweight support was applied.

For the haptic control system's components the parameters given in the previous chapter were used, unless explicitly denoted otherwise.

6.1 Basic rendering of haptic walking environments

This first section addresses the validation of the proposed control design, environment synthesis and rendering methods for the application in haptic gait simulation. Therefore these experiments assess the interaction with basic virtual scenes, which is stabilized by the proposed coupling controller.

6.1.1 Interaction with a single stiff object

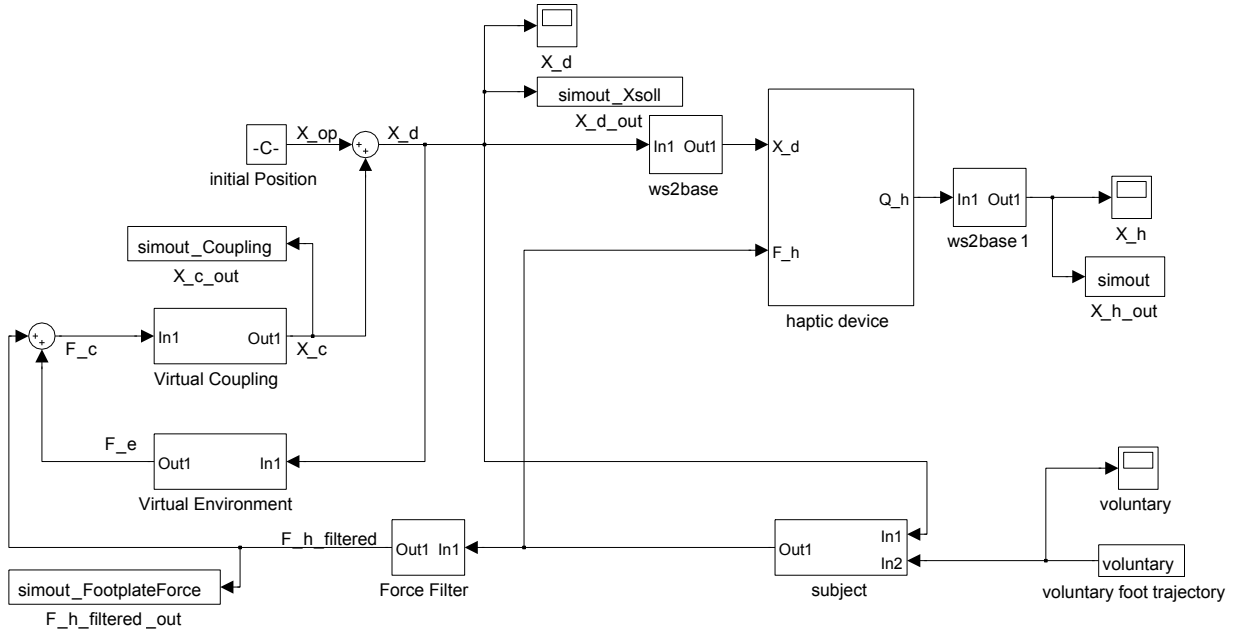
As pointed out in Sec. 3.1 and 4.1.2 one of the main issues is the interaction between operator and stiff objects with high force via the haptic display. Therefore the very first evaluation step is to ensure the stability of such interactions.

Simulation

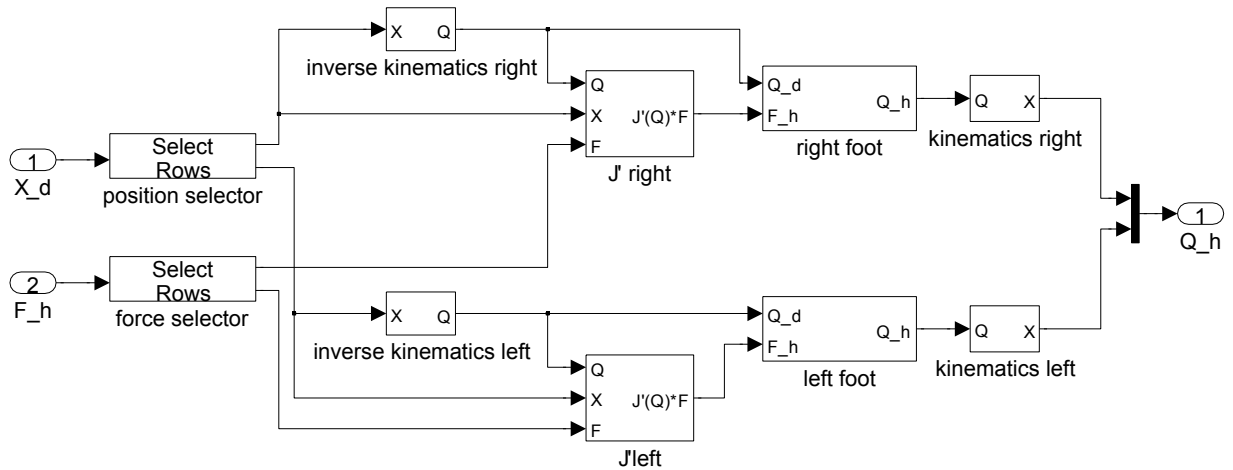
The simulation model used for the first simulation based stage of the evaluation is depicted in Fig. 6.1 a. It comprises the detailed dynamic model of the *HapticWalker*, the virtual environment and the coupling controller.

The machine's model includes the dynamics of both robots, which are moving the footplates and is shown in Fig. 6.1 b. The operator input can be simulated by a constant force signal or an arbitrary force function obtained from a MATLAB[®] workspace variable.

A unidirectional transition from free motion to contact with a virtual floor is simulated using the model described above. The floor is located 0.05 m above the lower end of the workspace and comprises a stiffness $k_{floor} = 10^5 \frac{N}{m}$. The probe (footplate) starts 0.1 m above the ground. The operator input force at the probe is a step in vertical direction with an amplitude of $f_z = -200\text{ N}$. In each simulation the coupling controller parameters are left constant and all controllers use the same mass $m_t = 5\text{ kg}$. The first trial uses damping parameters designed for the actual object stiffness. To demonstrate unstable interaction the coupling controller parameters are changed. In the second and third trial two parameter sets, designed for the contact with lower environment stiffness than the actual object stiffness, are used.



(a) Top level model



(b) HapticWalker (haptic device) model

Figure 6.1: Simulation model including haptic device, virtual environment, coupling controller and operator input

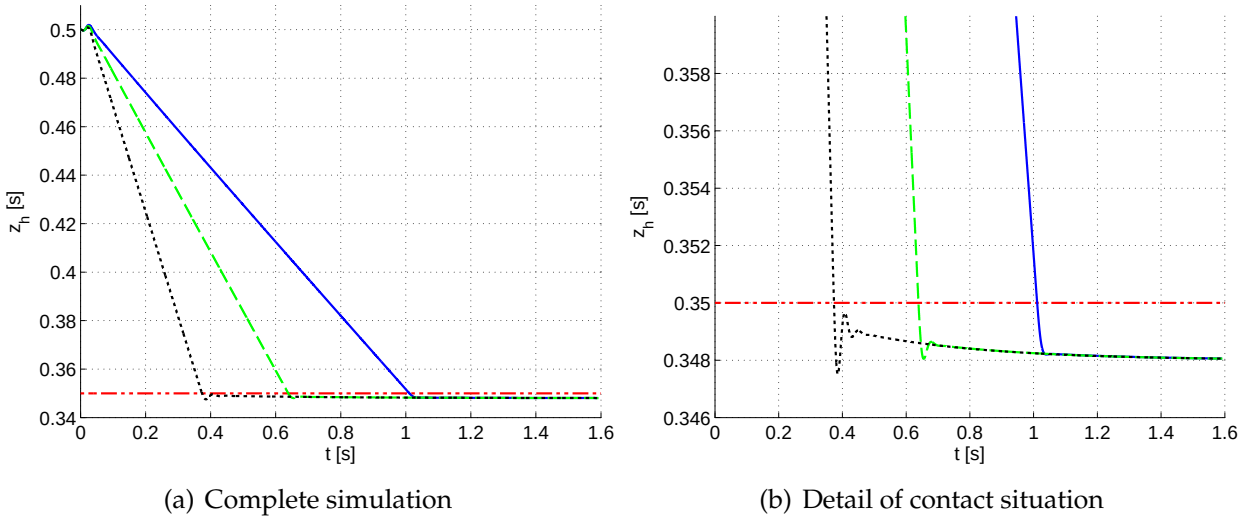


Figure 6.2: Unidirectional transition from free motion to stiff floor object ($k_e = 10^5 \frac{N}{m}$) – TCP position vs. time for the three controller parameter sets: A —, B --, C ...

The controller parameter sets used for the simulation as well as the experimental trials are designed for following maximum environment characteristics:

- A. $k_e = 10^5 \frac{N}{m}$
- B. $k_e = 5 \cdot 10^4 \frac{N}{m}$
- C. $k_e = 2 \cdot 10^4 \frac{N}{m}$

Fig. 6.2 a shows the vertical footplate position z_h over time for all three parameter sets and Fig. 6.2 b the contact transition in detail. The position of the trial using the appropriately designed parameter set A is plotted as solid blue line, for the parameter set B as dashed green line and for set C as dotted black line. Other than expected none of the two inadequate parameter sets leads to real contact instability, i. e. temporary loss of contact or continuous vibrations. However the chatter during contact transition increases notably for lower controller damping and with the parameter set C the probe is close to losing contact again.

Experiment

Experiments, corresponding to the above simulations, conclude the validation of simple interaction stability with weight bearing objects. They are performed on a single, static virtual floor tile. The operator is standing on the *HapticWalker* with his feet strapped to the footplates. He lifts and stamps down the right foot to the virtual floor. In the single stance phases the experiment shows the ability of the simulation to carry the operator's full bodyweight. The experimental design resembles the conditions used in the simulation, i. e. floor stiffness and three controller parameter sets are the same. Accordingly the colors and line styles used in Fig. 6.3 equal those in Fig. 6.2. Only for the parameter set C, designed for the lowest environment stiffness, the operator stands on secure ground with

the left foot, instead of the machine's footplate, and moves one footplate with the right one. Moreover only in this trial the bodyweight is not fully shifted to the virtual floor.

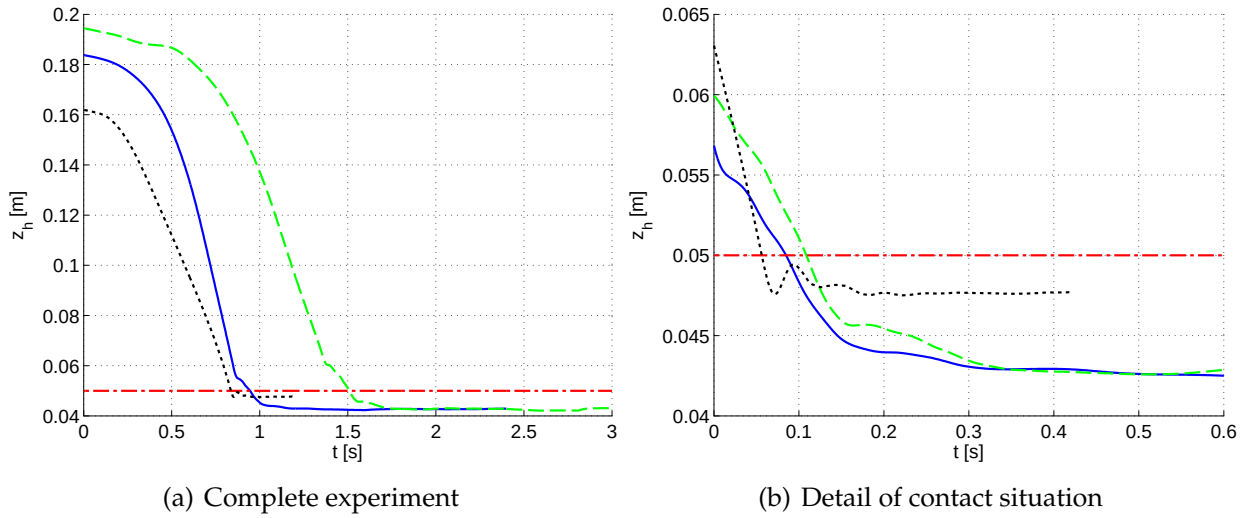


Figure 6.3: Contact transition experiment – vertical TCP position z_h during unidirectional contact transition to a virtual floor ($k_e = 10^5 \frac{N}{m}$) with the three different controller parameter sets: A —, B ---, C ···

The vertical position of the right foot plate for all three trials is depicted in Fig. 6.3a and the detailed view of the contact in Fig. 6.3b. In addition the operator input force compared to the simulated environment reaction force for the three trials is plotted in Fig. 6.4a - c.

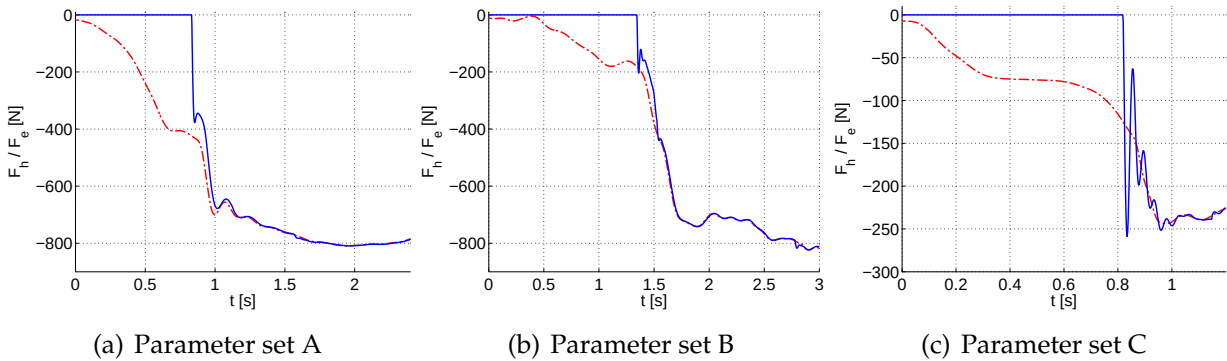


Figure 6.4: Contact transition experiment – vertical operator force z_h -- and simulated vertical environment reaction force z_e —

Corresponding to the simulation results, the experiment shows no unstable contact situation for the applied parameters. Parameter set B. is designed for an environment stiffness $k_e = 5 \cdot 10^4 \frac{N}{m}$, half the actual object stiffness used. Suggesting very small model errors in the linearized model, a parameter variation of 100 % can be tolerated for the environment. Accordingly the position during contact shows little chatter and the simulated force tracks the operator input almost as good as in the trial with parameter set A (Fig. 6.4a and b). Using parameter set C the position and the simulated reaction force show strong oscillations. These are distinctly felt in the experiment. The subjective im-

pression of the impact and the following oscillations with parameter set C is very unsafe. Moreover these impacts might well activate the safety foot binding during a real heel strike and bodyweight shift.

In [16] a robustly designed coupling controller with the structure from Fig. 4.9 a (eq. 4.18) was tested. In this case one controller parameter set was used for unidirectional contact with environments comprising the design target stiffness, $\frac{5}{3}$ and $\frac{5}{2}$ of this value, respectively. The coupling achieved stabilization of the contact with the first two environments, but not with the last one.

The increased robustness regarding environment parameter variations compared to the controller used in [16], which is seen from the third trial with parameter set C, probably results from the frequency dependent damping. It provides comparable or even lower damping for low frequencies and considerably higher damping for high frequencies. This higher damping for high frequencies reduces or eliminates the oscillations occurring for inadequate parameters. As a result the contact transition remains stable, but feels very unsafe due to the notably harder contact remaining oscillations.

6.1.2 Interaction with multiple objects

The scenario used to investigate the interaction with multiple superposed objects consists of a floor tile and a threshold object atop the floor. A healthy subject moved through the example scenario. The experiment involved interaction with only one as well as both objects at a time. Contrary to the unidirectional contact transition experiment, the parameter adaptation of the coupling controller is already used in this experiment. However, the importance of reducing the controller damping in free motion, which is the same as to increase the gain, will become evident for the voluntary motion within dynamic scenarios later.

Fig. 6.5 a - c show three 3D snapshots of the virtual experimental scenario, the footplates and the operator trajectory created in MATLAB[®]. Fig. 6.5 d depicts the virtual scene and the operator motion during the first part of the experiment, in which the operator steps onto the threshold with his right foot.

The measurement data shown in Fig. 6.6 includes the first part of the experiment up to the point where the subject is standing with his full bodyweight on the threshold. The subject lifted the right leg and placed the foot atop the threshold (approx. 4 s – 9 s). Then he shifted his weight to the right leg pushing down on the heel until standing with the heel on the floor and the forefoot on the threshold, see Fig. 6.5 b (approx. 10 s – 15 s). At this time he interacted simultaneously with both floor and threshold (approx. 15 s – 17 s). Finally he shifted his weight forward again, fully standing on the threshold with the right leg (approx. 18 s – 20 s), and afterwards stepped over the obstacle.

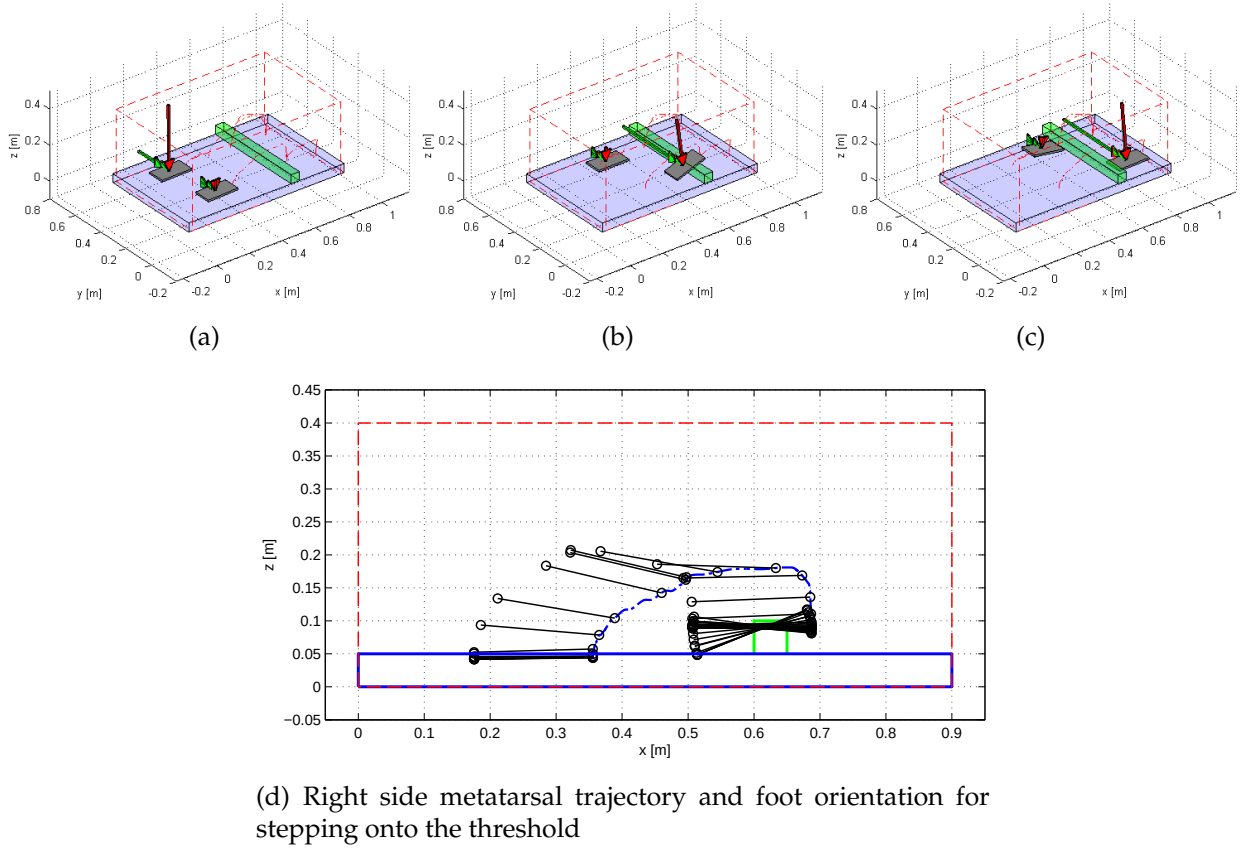
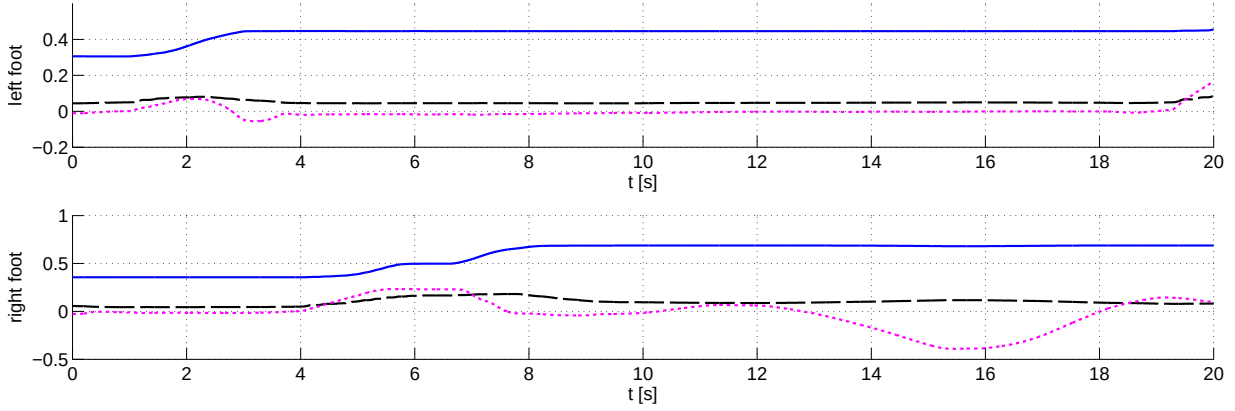


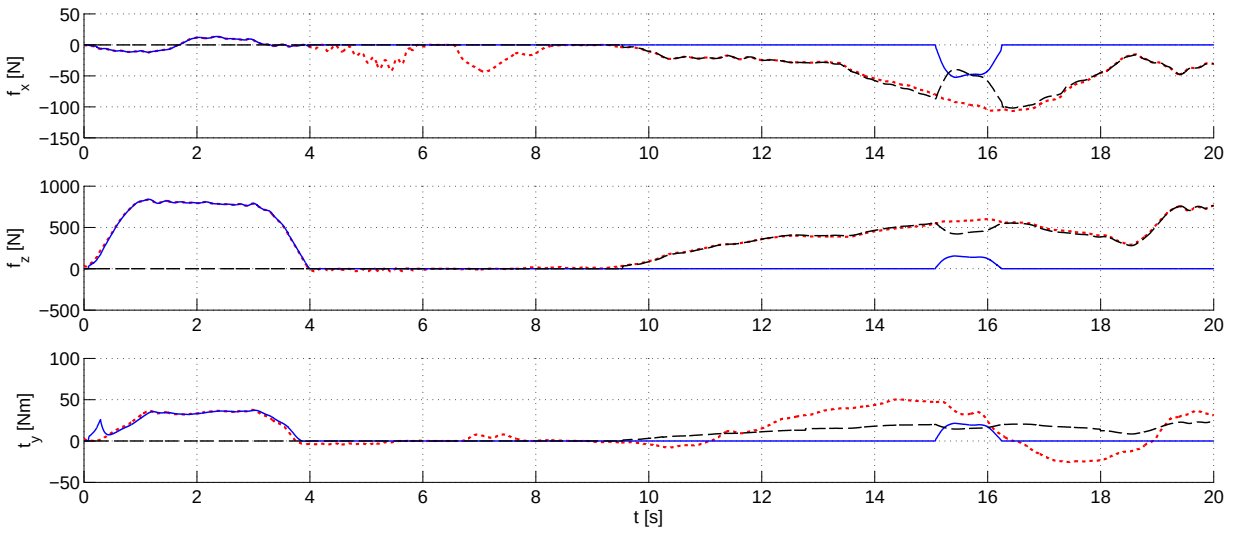
Figure 6.5: MATLAB[®] visualization of simulated scenario and measured operator trajectory and force for interaction with multiple objects (floor, threshold) – exemplary 3D snapshots at three instants (a – c) and 2D metatarsal trajectory (d)

The position of the user's left and right foot are presented in Fig. 6.6 a. Anterior-posterior positions (x_h) are plotted as solid blue line, vertical positions (z_h) as dashed black line and foot orientation (Φ_h) as dotted magenta line. The active foot in this experiment is the right one.

The force and torques presented in Fig. 6.6 b are the measured operator input as dotted red line, the floor reaction force as solid blue line and the threshold reaction force as dashed black line. The environment reaction forces correspond well to the operator contact forces during contact, as they are supposed to. When the operator interacts with both floor and threshold the forces adequately distributes to both objects and the interaction remains perfectly stable. While the subject interacts only with the threshold object, the measured operator torque differs from the environment interaction torque. This is because the rotatory DoF is not in a static equilibrium but the footplate rotates backwards until the heel hits the floor object and afterwards the footplate rotates forward when the weight is shifted to the right leg. During the static period when the subject rests on floor and threshold (approx. 15 s – 17 s) the operator torque corresponds to the environment reaction.



(a) Cartesian coordinates x_h (—), z_h (--) and Φ_h (· · ·) of left and right footplate for stepping on a threshold



(b) Measured operator and object reaction forces / torques – operator (· · ·), floor (—), threshold (--)

Figure 6.6: Operator positions, operator forces and environment reaction forces during stepping on top of a threshold

6.1.3 Rendering of surface characteristics

The simulation of training scenarios of daily life involves rendering of different kinds of surface characteristics, especially stick-slip effects and friction. The approach to achieve these basic surface characteristics in the simulation is given in Sec. 4.3.2. Specifically the stiction effect is evaluated in this section.

For the experiment the probe is pushed onto a virtual floor object with a perpendicular force of 200 N. This is implemented by placing a 20 kg weight on the footplate. Then a tangential force is produced manually in the positive direction of the x -axis. The force is slowly increased until the stiction is overcome and the footplate starts to move.

Fig. 6.7 shows the tangential probe position x_h with respect to the surface vs. the corresponding tangential operator force for four different friction coefficients $\mu = 1.0, 0.7, 0.5$ and 0.0 . According to Sec. 4.3.2 the breakaway force is supposed to be $\mu \cdot f_n$ and $f_n \approx 200$ N is the contact force perpendicular to the surface.

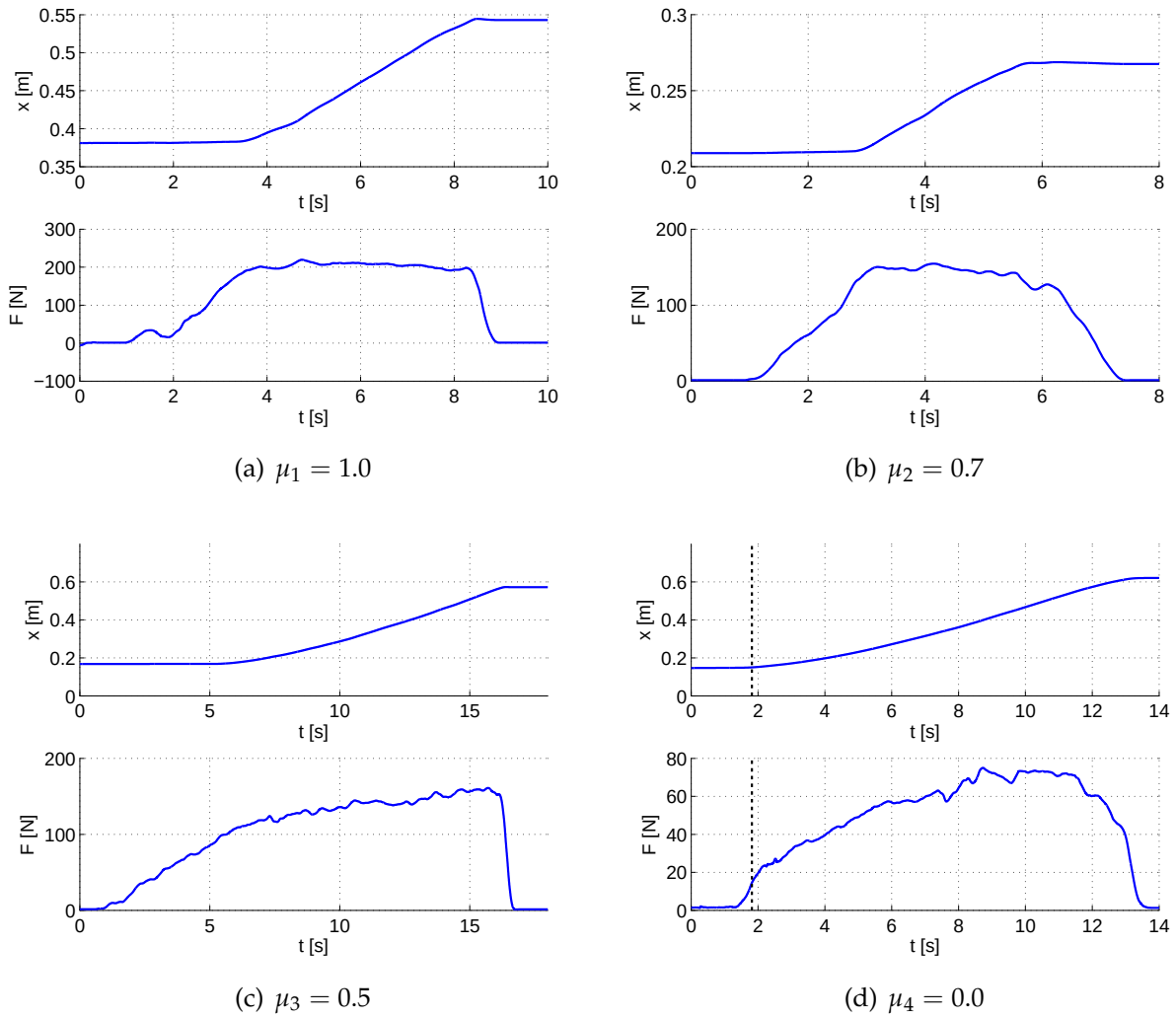


Figure 6.7: Experimental evaluation of simulated object surface stiction with four different friction coefficients μ_i – tangential force and probe position during contact with a virtual floor

The tangential probe position vs. the corresponding force, seen in Fig. 6.7, validate this intended behavior. Up to the breakaway force the probe moves only slightly due to the finite object stiffness. Only when the breakaway force is reached the probe starts to slide. This force depends on the friction coefficient as described in the concept. In Fig. 6.7 a - c the breakaway point, which is marked by the vertical dotted line, corresponds well with the expected force. Due to the minimum coupling impedance for $\mu = 0$ of course still a certain force $f_t \approx 10 \text{ N}$ is needed to cause a notable motion of the probe.

6.2 Voluntary motion on the haptic gait interface

Based on the validity of proposed control system and control design, shown in the previous section, this section assesses voluntary, unassisted movement through virtual envi-

ronments, which are simulated on the robotic gait interface *HapticWalker* using the proposed haptic control framework.

6.2.1 Free motion above a virtual floor

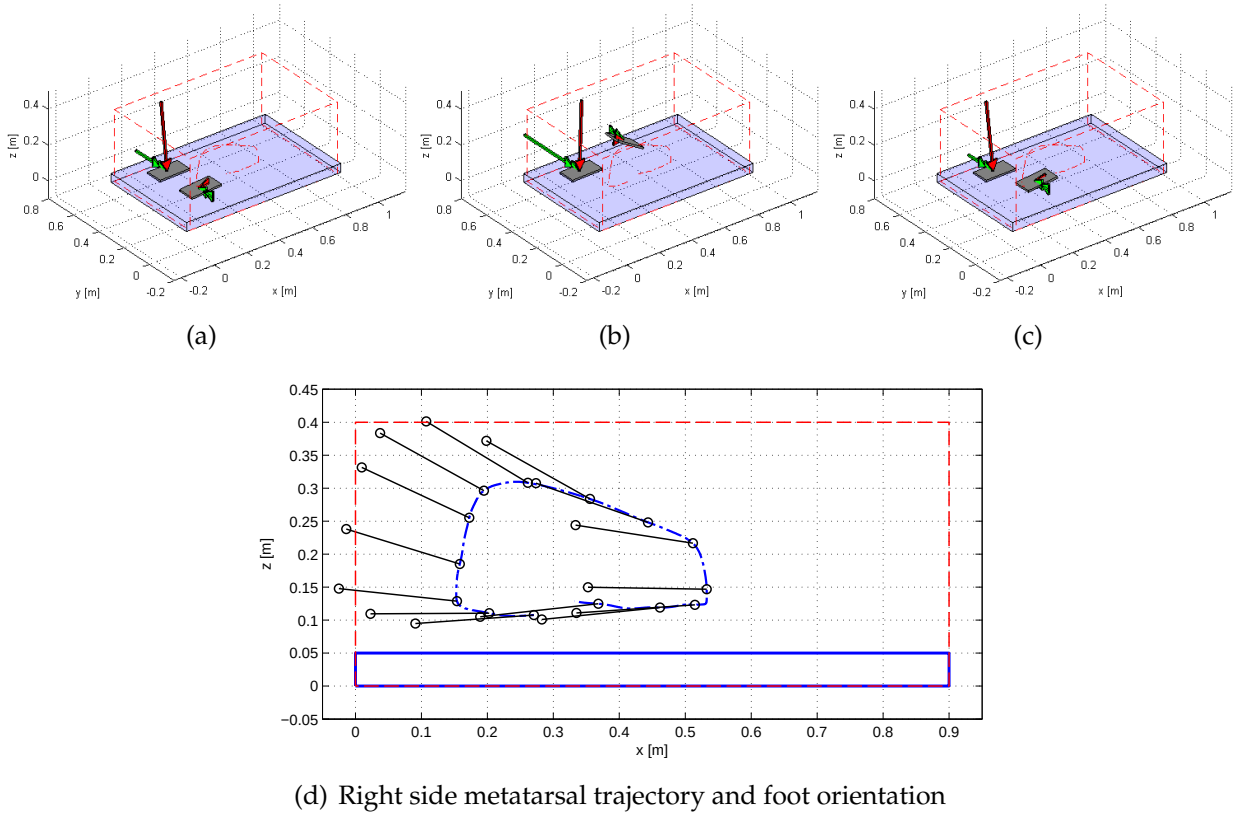


Figure 6.8: MATLAB[®] visualization of simulated scenario and measured operator trajectory and force for minimum impedance (free) motion experiment – exemplary 3D snapshots at three instants (a – c) and 2D metatarsal trajectory (d)

The first example of voluntary motion evaluated is free motion, i. e. minimum impedance simulation, on the haptic gait interface. The environment consists of the free space and a floor tile to allow the subject to stand on the device in haptic control mode. For the free motion experiment the user was standing on the virtual floor with his left leg and performed a motion with minimum resistance above the floor with his right leg. Fig. 6.8 a – c presents three snapshots of a 3D visualization of the experimental virtual scene and the voluntary operator motion in this trial. Fig. 6.8 d depicts the trajectory of the operator’s right foot during the trial. The blue dash-dotted line is the trajectory of the TCP (metatarsal joint) and the solid black lines represent the probe, i. e. user’s foot, at discrete times. They provide the probe orientation. Fig. 6.9 shows the remaining forces and torques acting on the user’s right foot during the free movement. The resistance felt by the user in free motion strongly depends on the velocity of the movement as a result of the damping given in Tab. 5.3. Still the forces seen in the experiment ranging up to 50 – 80 N are relatively low considering the device’s high inertia being close to 300 kg.

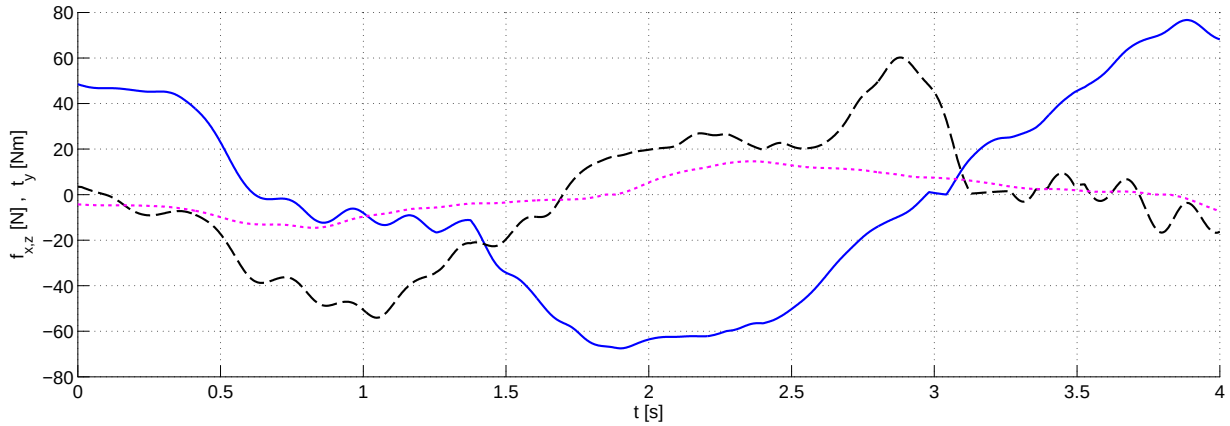


Figure 6.9: Operator forces during free motion experiment: — f_x , -- f_z , $\cdots$ t_y

From a second free motion experiment the actual minimum impedance parameters $\hat{m}_t$ and $\hat{b}_t$ were identified using a least squares estimator and the known coupling transfer function structure. Therefore the device was moved by hand randomly. Tab. 6.1 shows the identified parameters for all three Cartesian degrees of freedom. The results show, that the effective mass, identified from the manual excitation, is higher than the target mass m_t of the controller parameter sets. The position controlled actuators are acting as low pass filters with a cutoff frequency equal to their bandwidth. This leads to a decrease of the gain, and therefore responsiveness, in the complete plant consisting of coupling controller and position controlled device for higher frequencies. High frequency inputs to the control system result from motion patterns with high acceleration and jerk. Hence the operator perceives an increased mass when strongly accelerating the footplates.

Cartesian direction	Mass $\hat{m}_t$	Damping $\hat{b}_t$	α
x	7.13 kg	1232.8 $\frac{Ns}{m}$	0.2
z	7.10 kg	1234.9 $\frac{Ns}{m}$	0.2
Φ	0.435 kgm ²	126.4 $\frac{Ns}{rad}$	0.2

Table 6.1: Implemented free motion impedance parameters estimated from the voluntary free motion experiment

6.2.2 Ascending and descending a static stair

A voluntary movement within a scenario, composed from several static objects, is the next level of complexity for the haptic rendering task. For this evaluation the environment consists of a floor tile and two steps forming a stair. In the beginning the subject stands on the floor in front of the stair. He ascends the two steps and stands on the upper step with both feet. Then he descends the steps backward ending with both feet on the floor again. Fig. 6.10 shows the scenario.

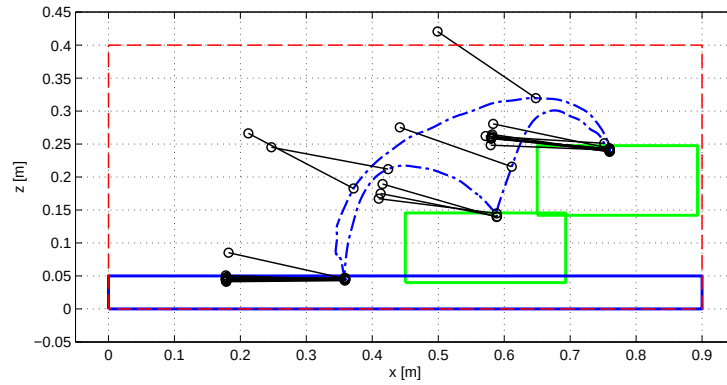
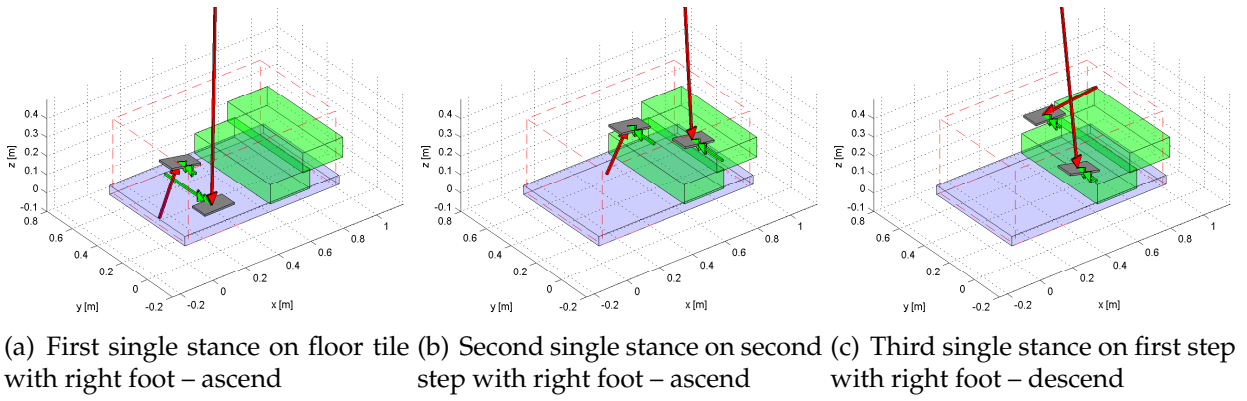


Figure 6.10: 3D MATLAB[®] visualization of haptic simulation and measured operator trajectory and force for ascending and descending a static stair – exemplary snapshots at three instants (a – c) and metatarsal trajectory (d)

The forces measured at the right footplate during the experiment are shown in Fig. 6.11. No bodyweight support was provided for the experiment, but the subject was allowed to hold on to the handrails to support balance. Although he was not supposed to support his bodyweight using the handrails, this side effect could not be completely avoided. Accordingly the forces in stance phase do not always completely reach the subject's bodyweight as could be expected. The force plot shows the transition from double stance in the beginning to single stance when the left leg is lifted onto the first step. After that the swing phase follows, in which the right leg is set on the second step. Finally there is another single stance phase when the left leg follows the right one and the double stance as the subject stands on the second step. After that the motion is reversed and the subject descends the stair.

6.2.3 Unassisted walking on virtual treadmill

To assess voluntary motion of a healthy operator within dynamic environments, a healthy subject walked through the virtual treadmill simulation at varying treadmill speeds between 0.4 km/h and 1.2 km/h . The results were published in [93] on ICRA 2011. The

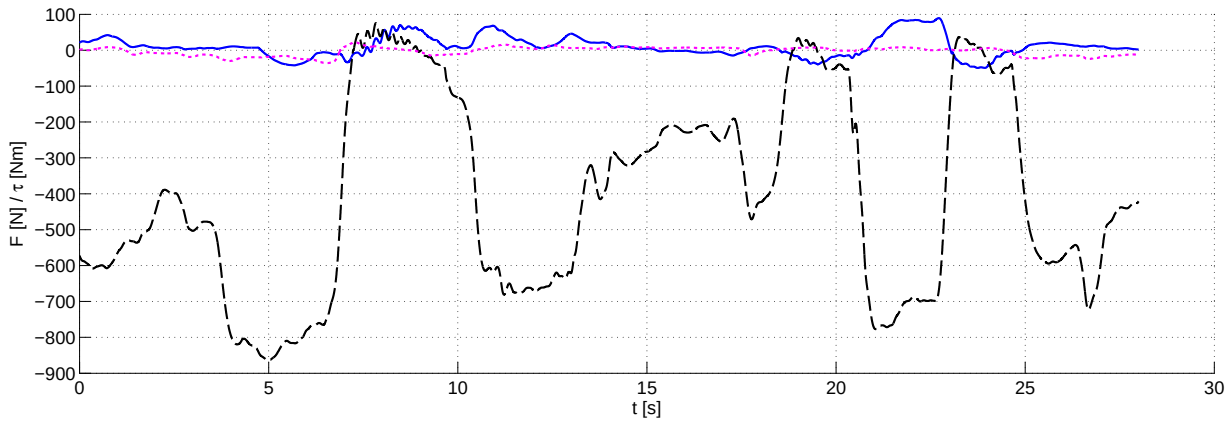


Figure 6.11: Right side operator forces during ascending and descending a static stair: — f_x , -- f_z , ··· t_y

experiment involves eight strides. Fig. 6.12 a - c shows three snapshots of an offline MATLAB[®] 3D visualization which were taken approximately at left initial contact, right toe off and right mid swing. Besides the virtual environment (i. e. treadmill) and the probes (i. e. footplates) the trajectory of all strides performed is plotted.

In Fig. 6.13 the results of the virtual treadmill walking experiment are presented. Fig. 6.13 a shows the horizontal (solid blue) and vertical (dashdotted black) forces measured during the experiment. The subject's bodyweight of approx. 80 kg is indicated as a reference by the dotted red line. Fig. 6.12 d shows the corresponding cyclic trajectory of the right leg. The step length was freely chosen and is approx. 0.55 m. As can be seen from the vertical forces, the virtual treadmill objects bear the user's complete bodyweight during stance phase. Each stride involves the transition from free motion to contact and vice versa. Considering the probe position, i. e. the trajectory from Fig. 6.13 c, it is seen that the contact transitions remain stable without loss of contact. The resistive force felt in free motion, when lifting the foot off the ground and moving it forward during swing phase, remain about 50 – 100 N, which is consistent with the free motion experiment. These forces are more evident in horizontal direction when swinging the foot forward because of the higher velocity of the motion in that direction.

The coupling controller characteristics, needed to stabilize the interaction and the resulting resistive forces in free motion, respectively, lead to a distorted gait pattern in unassisted treadmill simulation. This becomes evident when the measured foot trajectory is compared to the nominal training trajectory for assisted treadmill training, which is shown later in Fig. 6.14. This nominal trajectory is directly derived from healthy persons' gait patterns measured via a marker based motion tracking system.

This experimental setup illustrates the importance of the controller parameter adaptation proposed in Sec. 4.2.3. In the swing motion the free motion controller parameter set with reduced damping is active which is only adapted in the vicinity of contact. Despite the influence of the haptic device on the walking pattern, the operator is able to move through the dynamic environment according to its motion. The use of only one parameter set,

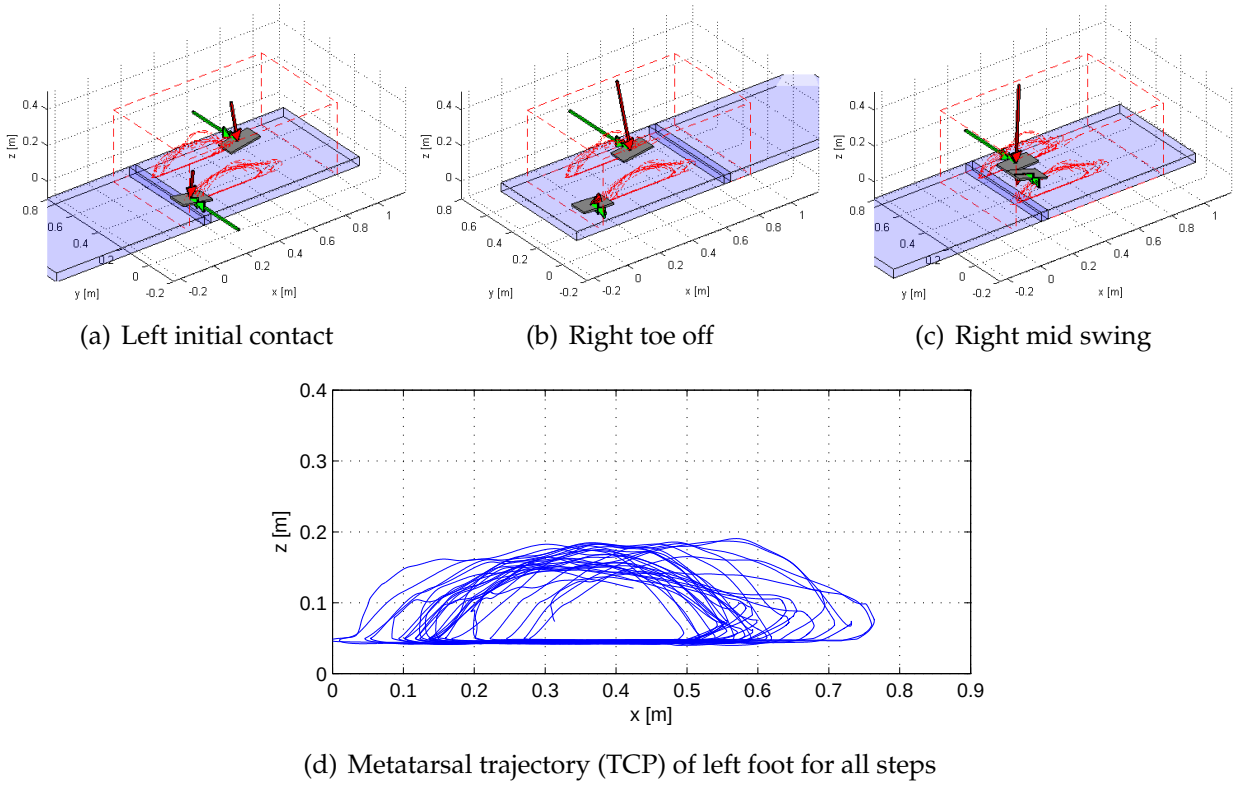


Figure 6.12: 3D MATLAB® visualization of haptic simulation and measured operator trajectory and force for unassisted walking on the virtual treadmill – exemplary snapshots at three instants (a – c) and metatarsal trajectory (d)

which is designed to stabilize the interaction with weight bearing environment objects, would imply a considerably higher damping in swing phase. In a setup, in which the operator needs to voluntarily move the device according to the dynamic environment, the forces needed to do so would be excessively high and hence render the system useless.

6.3 Assisted training modes

Three assisted training tasks were experimentally evaluated with one healthy subject utilizing assistance mode A, i. e. the moving window mode:

- A. treadmill walking,
- B. stair climbing upstairs,
- C. stair climbing downstairs.

These training scenarios are based on the available gait trajectories for floor walking and climbing up- and downstairs, which were derived from real motion tracking data obtained from healthy subjects (see Sec. 5.1). These training patterns can be considered relevant in every day's life and are desirable for real clinical rehabilitation training. The

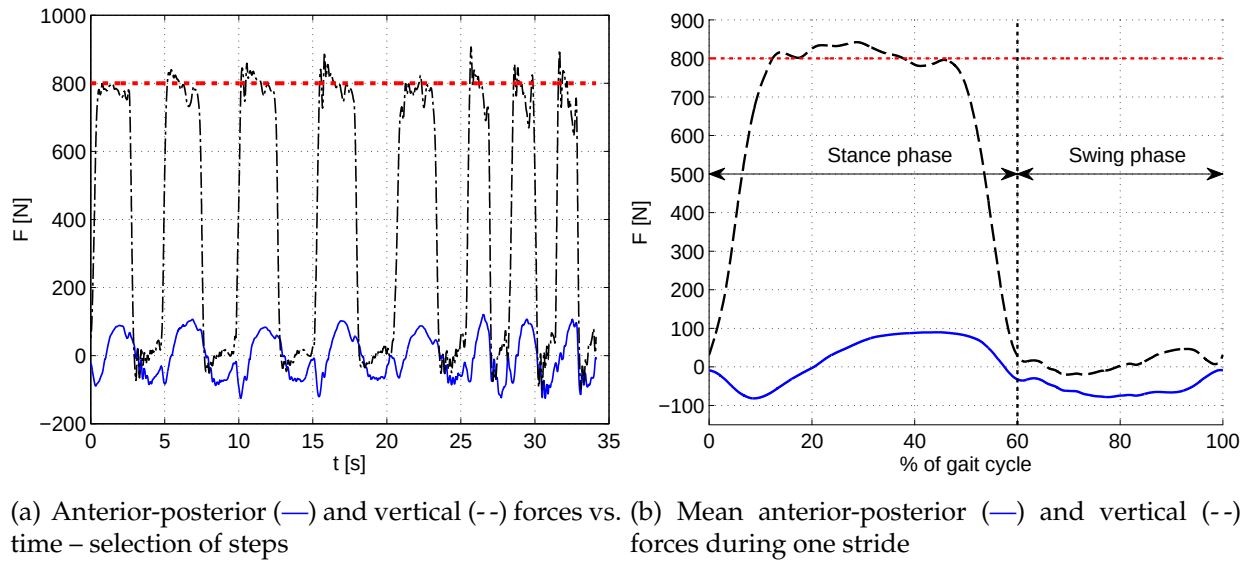


Figure 6.13: Measured operator interaction forces at left footplate during unassisted walking on virtual treadmill simulation – force vs. time selection of steps (a), mean force over all steps (b)

assistance window moves on these trajectories and the objects in the training environment, i. e. floor tiles and steps, are modified to fit the trajectory geometry.

Assisted training was tested both with constant assistance level λ (scenarios A - C) and with iterative assistance level adaptation (scenarios A and B).

6.3.1 Moving assistance window with constant assistance level

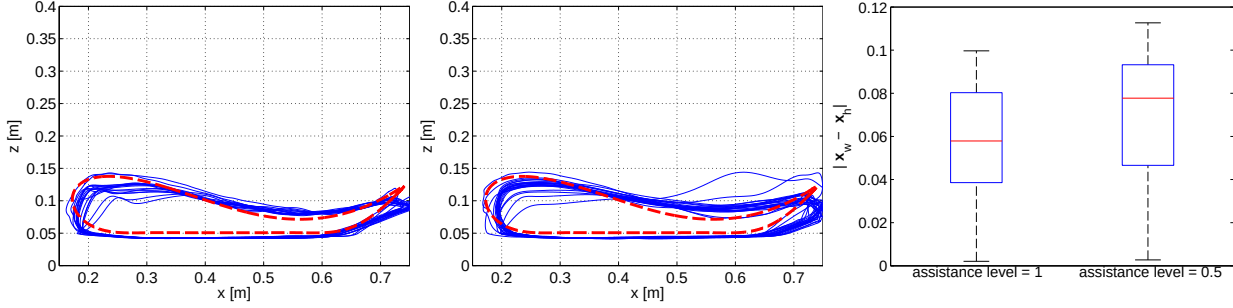
All three training scenarios were performed twice, firstly with assistance level $\lambda = 1.0$ and secondly with $\lambda = 0.5$. Referring to the concept in Sec. 4.4 a reduction of the assistance level by 50 % means reducing the window stiffness by half and hence only half the guidance and support force for the same deviation. To assess the effect of the compliant assistance with adjustable assistance level, footplate positions and GRF in each training scenario are used.

The nominal as well as the actually performed trajectory of the metatarsal joint are plotted and the deviation of the training motion from the nominal training path is given in a box plot. The box plot denotes the median value, the upper and lower quartile and maximum value range. The median is marked by the horizontal line, interquartile distance by the box and the maximum range by the whiskers.

The second interesting point is the rendering of so called haptic cues, i. e. the ground reaction forces and assistive forces as proprioceptive input to the subject (refer to [71, 72]). To assess this haptic component the anterior-posterior and vertical interaction forces measured at one footplate are plotted. The force data is for each gait cycle is resampled and averaged over all gait cycles performed.

A Treadmill

Fig. 6.14 shows the nominal and actual trajectory of the metatarsal joint for $\lambda = 1.0$ (a) and $\lambda = 0.5$ (b) as well as a box plot of the deviation between the actual training motion and the nominal trajectory. The red dashed line is the nominal training trajectory and the solid blue line the actual training motion for all steps.



(a) Nominal and actual training trajectory: $\lambda = 1.0$ (b) Nominal and actual training trajectory: $\lambda = 0.5$ (c) Box plot of the deviation from nominal training trajectory

Figure 6.14: Nominal and actual training trajectories and deviation from nominal motion for treadmill walking

The comparison between unassisted and assisted treadmill walking clearly shows, that the assistance successfully guides the user in the vicinity of the nominal path. Moreover smaller forces during swing phase are observed, compared to those evident in unassisted treadmill walking. The remaining interaction forces during swing phase can be interpreted as the assistance provided for the training task. The compliant assistance by the moving window allows variation of the actual movement, which is also evident for natural human movement. The box plot in Fig. 6.14 c shows that the average deviation from the nominal trajectory as well as the maximum deviation range is higher for $\lambda = 0.5$ than for $\lambda = 1.0$, as would be expected.

The ground reaction forces measured during the treadmill training in the subject's left foot are given in Fig. 6.15 for both assistance levels. The vertical force surpasses the subject's body weight, which indicates a complete weight shift to the stance leg. The distinct force peaks, which normally occur in heel strike and push off cannot be discerned during this assisted training trial.

B Stair climbing upstairs

The second training scenario considered is climbing upstairs. For this setup Fig. 6.16 provides the real movements compared to the nominal training trajectory and the deviation between real and nominal motion as box plot for both aforementioned assistance levels.

Similar to treadmill walking the performed movements vary with each step but remain consistent with the nominal trajectory for both investigated assistance levels. Other than expected the average deviation in this scenario does not noticeably vary between the

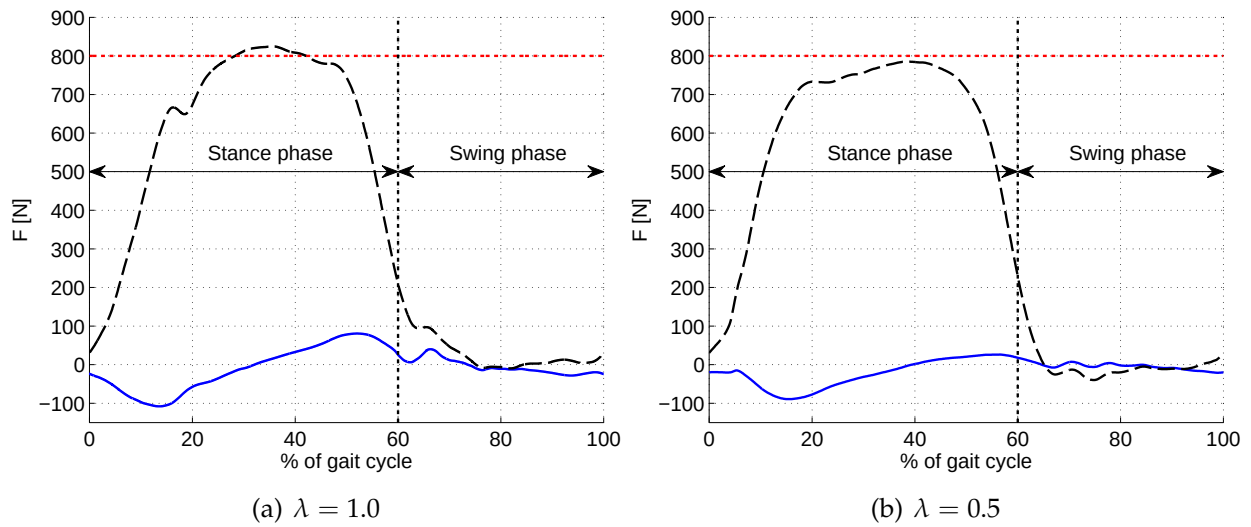


Figure 6.15: Mean anterior-posterior (—) and vertical (--) ground reaction forces during treadmill walking

assistance levels. However, as expected, the range of maximum deviations increases for lower assistance.

During this trial mean vertical GRF in Fig. 6.17 barely reaches the subject's bodyweight level of approx. 800 N. As stated at the beginning of this chapter the subject was allowed to hold on to the handrails mounted on the device to secure his balance. Although he was not supposed to support his bodyweight thereby, this effect cannot be completely eliminated and is evident in this trial. The force graph shape resembles data from natural walking, e. g. presented in [73], more closely than in the treadmill simulation. Still shape lags the characteristic peaks during heel strike and push off for the investigated subject.

C Stair climbing downstairs

The last training task presented here is climbing downstairs. Corresponding to the previous tasks Fig. 6.18 shows the foot, i. e. metatarsal, trajectories and their deviation from the nominal trajectory for both assistance levels.

As expected and already observed in the treadmill trial the average deviation as well as the deviation range increases noticeably for the lower assistance level.

For the last training task C the GRFs are provided in Fig. 6.19. The observation in the vertical GRF during training task B is valid for this trial, too. The amplitude of force indicates that the subject involuntarily supports not only his balance but also his bodyweight by holding onto the handrails.

Common results and findings for training tasks A - C

The vertical GRF curves, especially in the treadmill trial, miss the distinct peaks a heel strike and push off compared to natural walking data from literature (e. g. [77, 73]) and data recorded on the *HapticWalker* in position controlled mode from [71, 72], which was

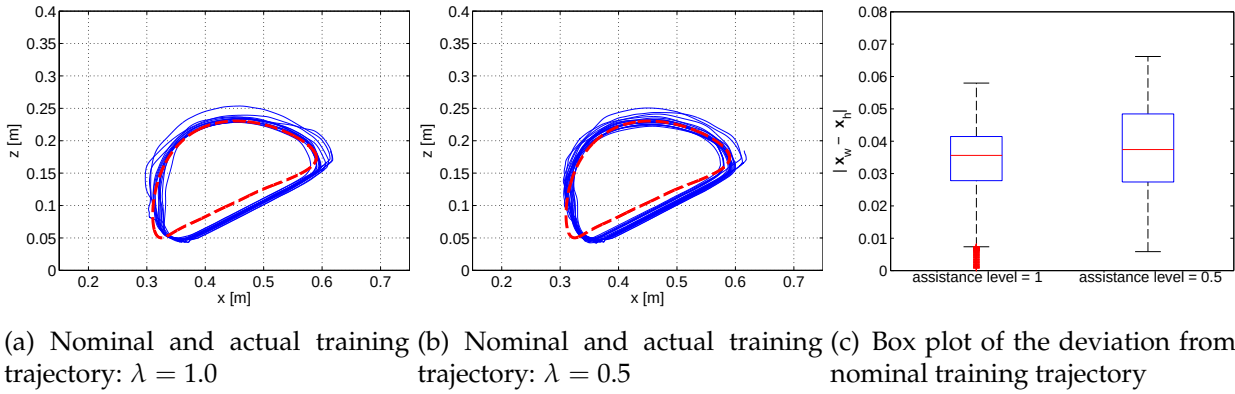


Figure 6.16: Nominal and actual training trajectories and deviation from nominal motion for stair climbing (upstairs)

also mentioned in Sec. 2.4. In [71, 72] already differences between haptic cues during walking on the *HapticWalker* in position controlled mode and free walking were investigated. The data represents mean results from ten subjects. The experiments showed considerable variability of the walking patterns on the gait interface between the subjects. Therefore from the single trials performed in the haptic control mode cannot be deduced if the inadequate force trajectory results from a systematic effect of the assistance mode or from the subject specific behavior. Additionally the walking speed has significant influence on the intensity of the aforementioned haptic cues. The cadence, which was used in the presented experiments, was considerably lower than the walking speeds, which the literature data was acquired with.

The investigation in position controlled mode showed a smaller inclination of the slopes in the force trajectory for robot assisted walking. This results in a different timing of the force peaks in heel strike and push off. Consistent with this finding the slopes in the measurements offer a smaller inclination than observed in natural walking.

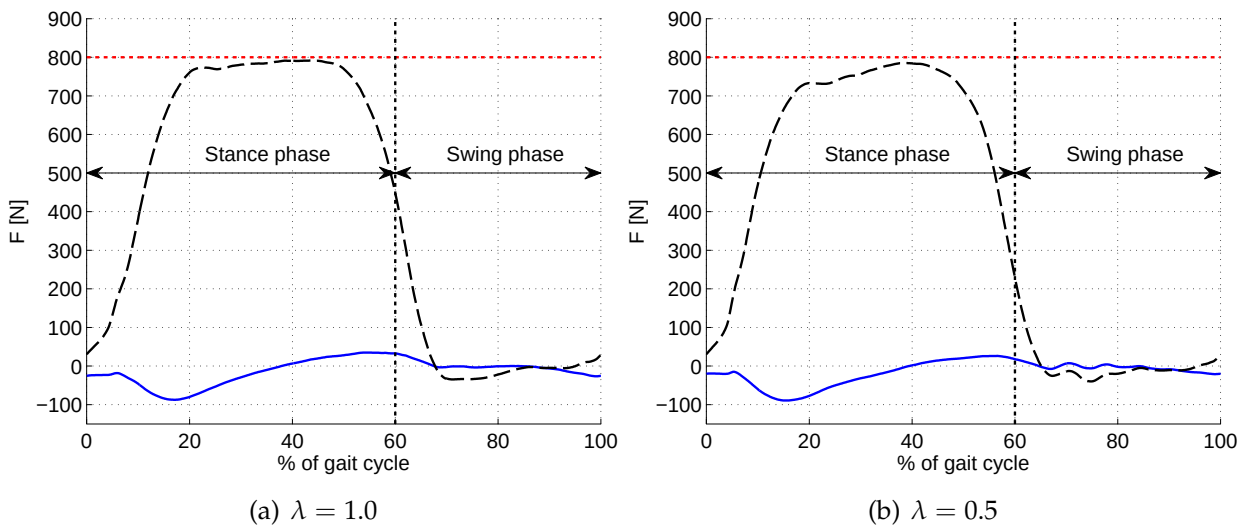
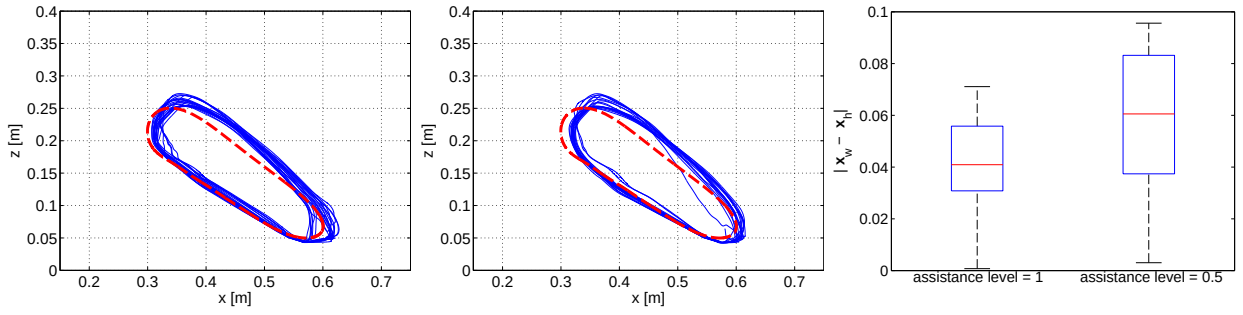


Figure 6.17: Mean anterior-posterior (—) and vertical (--) ground reaction forces during stair climbing (upstairs)



(a) Nominal and actual training trajectory: $\lambda = 1.0$ (b) Nominal and actual training trajectory: $\lambda = 0.5$ (c) Box plot of the deviation from nominal training trajectory

Figure 6.18: Nominal and actual training trajectories and deviation from nominal motion for stair climbing (downstairs)

As indicated in the section for task A (treadmill) the remaining forces in the swing phase represent the assistive forces rendered by the robot. Here the largest forces appear in early swing. This is supposed to be due to a prolonged stance phase, which means a delayed push off. Hence in this moment the proceeding assistance window pushes against the bodyweight, which is not yet shifted. Except for example task C the observed assistance forces decrease together with the assistance level.

6.3.2 Moving assistance window with iterative assistance level adaptation

The training scenarios A and B, i. e. treadmill walking and climbing upstairs, were tested with the automatic, performance based adaptation of the assistance level proposed in Sec. 4.4.3. The training parameters, such as step length and height, were the same as in

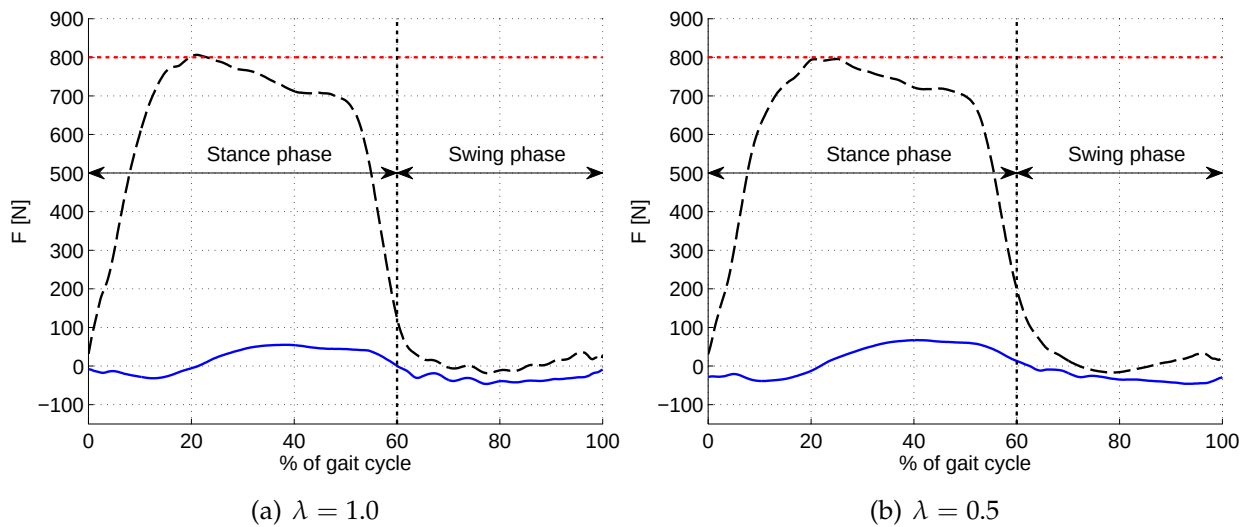


Figure 6.19: Mean anterior-posterior (—) and vertical (--) ground reaction forces during stair climbing (downstairs)

the previous trials with constant assistance level. In these experiments the assistance level is set to $\lambda = 1.0$ at the beginning. After that it is only modified by the ILC algorithm [88].

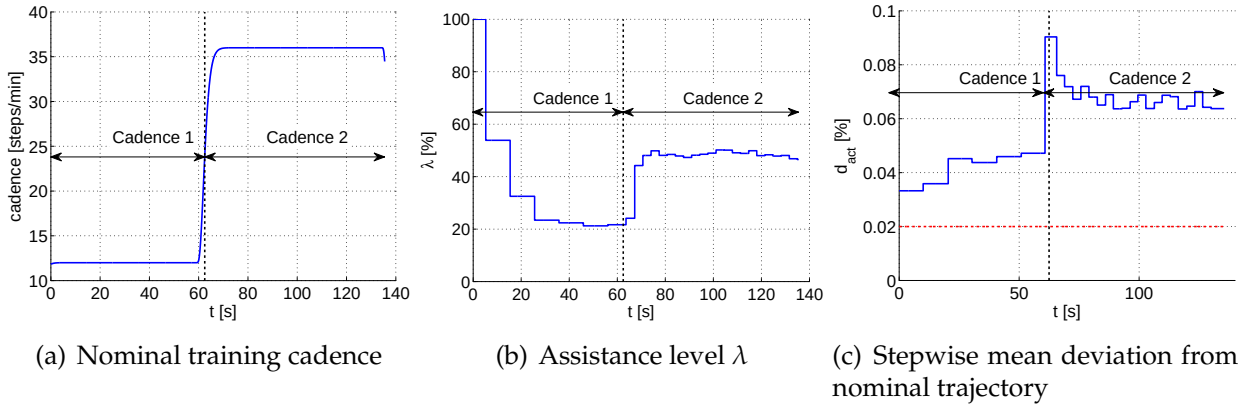


Figure 6.20: Nominal training cadence and assistance adaptation during treadmill training

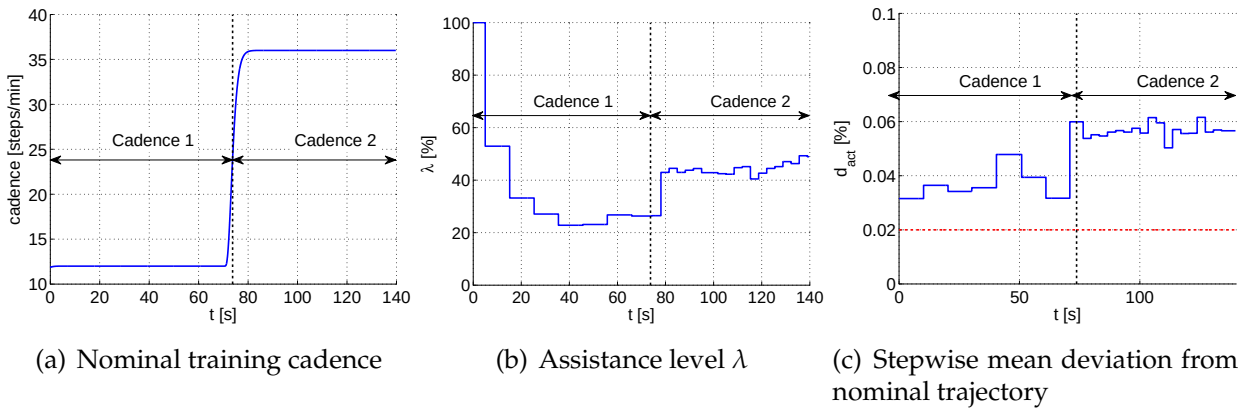


Figure 6.21: Nominal training cadence and assistance adaptation during stair training (upstairs)

Fig. 6.20 and Fig. 6.21 show the nominal cadence, assistance level and stepwise mean deviation from the nominal trajectory for the treadmill walking and stair climbing tasks, respectively. The assistance level decreases fast at low training speeds due to a good tracking of the nominal path. The observed tracking accuracy reduces with increasing training speed. The mentioned free motion within the assisting window is still considerably damped to ensure interaction stability. Therefore with increasing training speed the effort for the subject to keep inside the window increases. As a result the deviation from the nominal position, i. e. penetration of the window, increases, which means that the window provides stronger assistance. Thereby it acts as feed forward to track the nominal motion. An adaptation of the subject to the increased walking speed and thereby a reduction of assistance level is observed only in the mean deviation during floor walking, yet not at all in the assistance levels. There might be several reasons for this. Firstly the number of steps could be too low to see this effect. Further the window is rather small to prevent hitting the ground too early in late swing. The parameters of the ILC were cho-

sen empirically. To find more optimal parameters data of the learning behavior of healthy subjects for the respective task would be needed.

7 Conclusions and future work

7.1 Summary

In this thesis a haptic control framework was developed, which presents the first systematic and unified approach for interaction control design and haptic simulation of complex training environments on end-effector based haptic gait interfaces. Thereby it firstly provides means to render sophisticated compliant, adjustable and patient-adaptive training assistance on these devices.

The concept consists mainly of three parts:

- A selection of control system architecture and control design for stable rendering of virtual walking environments,
- B flexible synthesis of complex, modular walking environments from object primitives and
- C integration of adjustable training assistance into the haptic control framework.

A Control system architecture and design

Considering the characteristics of the class of end-effector devices, e. g. the necessity to carry a user's bodyweight during stance and hence the need for powerful drives and strong mechanical structures, an admittance display system architecture was selected. It uses an underlying velocity or position control loop, which is assumed to be present in these devices. The coupling controller, which ensures interaction stability, comprises a hybrid structure combining small damping for low frequencies and increasing damping for high frequencies. Moreover it connects an impedance type environment to the admittance display. For the control parameter design a robust control approach based on [18] is used. It utilizes linearized dynamics and includes known worst case environment impedance bounds in the control design. To achieve a low minimum impedance for the haptic display in free motion and at the same time ensure contact and interaction stability with bounded impedance virtual environments, the coupling controller parameter set is adapted according to the contact state using gain scheduling. Therefore for each object type present in the virtual training environment an appropriate controller damping level is designed. The damping is then adapted in the vicinity of contact situations according to the object properties.

B Synthesis of virtual training scenarios

Based on the low level haptic control system from the previous section, which ensures stable interaction, a synthesis method for virtual walking environments was developed. It allows to compose training scenarios in a flexible and modular way from simple polygonal primitives. For the walking environments, considered here, cubic primitives were selected, which comprise the mechanical properties mass, stiffness and damping. Additionally the simulation of tangential surface friction and stiction is possible. The complex training scenarios are superposed from those primitives, which can be static or moving objects. The result is a parallel network of spatially confined impedance type objects.

C Adjustable training assistance

With the proposed haptic control framework exemplary adjustable and patient-adaptive training assistance modes were developed in this thesis, which are directly integrated as objects within the virtual training scenario. These elliptic window objects move along a predefined, task-specific trajectory and provide a region of free motion inside. Outside of this area they comprise a progressive stiffness. This springlike behavior draws the footplate back toward the center of the window, when it leaves the inner region of tolerated deviation.

The concept presents two approaches based on the aforementioned window objects, which provide adjustable training assistance. In the first approach, similar to [1], the windows move along the predefined training trajectory at a therapist selected speed. If the subject on the device deviates too much from the nominal trajectory, i. e. lags behind the nominal motion or moves too fast, the movement is corrected by window's assistive forces. A factor λ , the assistance level, scales the window stiffness and thereby the amount of assistive force provided. The second approach implements a paradigm similar to the path control, developed for the *Lokomat* in [2], using the elliptic window. The window is always placed on the nearest point to the current footplate position on the nominal trajectory. Thereby the window provides guidance for deviations from the trajectory. The support helping the subject to fulfill the training task, is implemented as a force field tangential to the nominal trajectory within the tunnel.

7.2 Conclusions

The proposed framework provides end-effector based devices, like the *HapticWalker*, with the foundations to implement sophisticated training strategies, which are able to render adjustable and patient-adaptive training assistance. The developed approach was implemented on the end-effector based robotic walking simulator *HapticWalker*. Its functionality was evaluated both in simulations and in experiments. The moving window assistance mode was preliminarily tested with a healthy subject.

The ability of the robust interaction controller, to stabilize the contact and interaction with bounded impedance environments was validated in simulation and experiments. Compared to the robust coupling controller in [18] the proposed controller turns out to be more robust against environment parameter variations. This is probably due to the frequency dependent damping, which increases for higher frequencies.

With the developed synthesis method complex, weight bearing training scenarios, static as well as dynamic, were composed and successfully rendered to the subject. Unassisted movement through dynamic environments was investigated for the example of walking on a virtual treadmill. This shows the feasibility of the haptic simulation. However, even with the adaptive controller the remaining coupling impedance, which is needed to stabilize the interaction, strongly altered the walking pattern. The effect is supposed to increase with movement speed as the effect of damping is proportional to the velocity of the movement.

The application of compliant and adjustable training assistance, which is integrated into the training environment, was evaluated for the assistance mode A, i.e. for the moving window. Three example training tasks were available for the experiments: treadmill walking, climbing upstairs, and climbing downstairs. In contrast to the unassisted mode the subject's feet were successfully guided near a physiological trajectory during the assisted walking experiments. The assistance allowed a variability between the various gait cycles, which is also present in natural gait, and still kept the gait pattern close to the nominal foot trajectory. In the sequel of trials two different, constant assistance levels were used in each of the three available training tasks. The average deviation and deviation range between the actual motion and the nominal path increased when the assistance level was reduced, as was expected. The haptic cues, as the induced forces are termed in [72], however deviated from those observed in natural gait and on the *HapticWalker* in position controlled mode. Due to the high variability between subjects it cannot be deduced from this single case trials, if the differences originate in the design of the assistance or from the specific motion patterns of the subject. Consequently this remains subject to further investigations. The ILC based adaptation of the assistance level was tested for the first two of the aforementioned training tasks. The parameters were chosen empirically as the development of a theoretical basis for the adaptation algorithm was not in the scope of this work. The assistance converged to a constant level, which was depending on the training velocity. However no distinct learning or adaptation effect could be observed in the subject with these parameters. Otherwise the assistance level, which increased when the training was sped up, should have decreased again over time.

7.3 Future work

It was previously pointed out that this thesis provides a framework as a first systematic basis for patient-adaptive gait training on end-effector based devices in the first place. On

top of this framework an exemplary concept for the integration of adjustable assistance was developed and tested. This exemplary assistance concept claims not to be perfect nor complete. Hence following points should be subject to further research and development.

In the proposed low level control scheme gain scheduling is used to adapt the controller parameters to the contact state between virtual environment and probe. This approach leaves room for improvements either to extend the gain schedule adding more and specialized cases or to apply different adaptive control schemes, e. g. with parameter estimation.

In this thesis a simple, constant tangential force field was proposed in the virtual tunnel concept (assistance mode B) to render training support. Other work suggests to design the force field to be passive [2], which would be an enhancement of the presented approach. Likewise the moving window assistance mode only considers a therapist selected training velocity. An automatic performance based training speed adaptation or even voluntary training speed control are next steps to add valuable features to the system.

In the current system the parameters for the ILC based adaptation of the assistance level were selected empirically. To achieve a theoretical background, an in depth investigation of gait patterns and learning behavior of healthy subjects under assisted gait training is necessary.

7.4 Potential impact of the presented framework in other applications

The presented haptic control framework was inspired by the constraints, requirements and potential benefits in the considered target application, i. e. end-effector based gait rehabilitation. However the developed system, with its modular nature and high portability to different devices, is not restricted to the use in rehabilitation. Its combination of stable haptic control design, modular virtual scenarios and integration of adaptive movement assistance can prove useful in a wide range of other applications.

7.4.1 Haptic simulation and the *assist as needed* paradigm

Transferring the *assist as needed* paradigm into productive environments enables the integration of training and safety aspects within working processes. The basic idea is that the assistance ensures a known task to be fulfilled with an allowable tolerance and thereby even unpracticed operators can safely perform the work. On the other hand if the assistance is automatically reduced depending on the operator's performance there will be a

training effect, eventually leading to a point where the operator performs the work steps on his own. In the following two possible application fields are described.

Robotic training and assist systems in surgery

Robot assisted surgery is already an active field of research. The work in that field ranges from systems assisting in the insertion of minimally invasive tools to fully automated robotic systems, which perform tasks such as milling of bones on their own. An approach which already incorporates the interaction between the surgeon and a robotic assist system is the control of the activity of a manually operated tool depending on its position during the operation.

The haptic control framework providing adjustable movement assistance can be used in assisted surgery to define a nominal operation task and assist the surgeon in this task as needed. The nominal task could be taught to the assist system by an experienced specialist in the respective field. This approach offers to unexperienced surgeons a safe way to *learn by doing* certain operation techniques. The *assist as needed* approach ensures, that the task is performed correctly within an allowable tolerance but if the operator performs well the assistance reduces to zero.

Assembly and power assist systems

To date the automation of small batch production processes becomes more relevant for small and medium enterprises and the same is valid for handling increasing numbers of variants in mass products. Therefore the worker, who provides the necessary flexibility, is closely cooperating with assisting systems. The result is a partial automation of the process making use of the human's flexibility and the performance increase from automation technology.

Active power assist systems to handle and assemble heavy parts, such as *Kobots* presented in [94, 95], can profit from the presented framework. It allows to implement the power assistance and operator guidance as well as virtual environments to define boundaries, e. g. to avoid collisions. Further the adaptation of the device impedance in the vicinity of real, intended contacts might improve the usability of the handling system.

The *assist as needed* paradigm in cooperating automation systems could allow knowledge transfer from specialists to other workers in the future, similar to the vision of robot assisted surgery given above. In that case the nominal task is again recorded from the specialist and can then be performed with the automation system's assistance from any other worker.

7.4.2 Application fields for haptic gait interfaces

Haptic interfaces in varying complexities are used as man-machine-interfaces in various applications and improve usability or extend the experience of virtual worlds.

Weight bearing haptic gait interfaces featuring the presented framework are expected to offer new possibilities in high-end virtual reality gaming applications. They allow the user to move within the virtual world, feel it and experience its topology. Modern video game consoles already attempt to combine games and physical exercises. The idea is to enhance training motivation in the playful atmosphere. The use of haptic gait interfaces together with 3D virtual reality techniques will allow to take the approach of high-end physical exercise games to a new level. On such a device a new gaming experience will be possible. Additionally by inverting the *assist as needed* approach to a *resist as needed* paradigm the physical effort can be controlled and thereby the training effect is optimized.

Likewise such systems could enable us to travel in time. We would be able to experience historical buildings and scenes visually and haptically as if we were in fact there. Imagine e. g. walking through the soft desert sand and watching the old Egyptians building the Pyramid of Cheops. Or you might climb the steps of the Colosseum in antique Rome to reach your seat and enjoy the games.

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