TECHNISCHE UNIVERSITÄT WIEN
Vienna | Austria

MASTER THESIS

Development of a Prosthetic Control to Support the Rehabilitation Process

carried out for the purpose of obtaining a degree of Maser of Science (MSc or Dipl.-Ing. or DI), submitted at TU Wien, Faculty of Mechanical and Industrial Engineering, by

Lena TRAUTMANN

Mat.Nr 01526567

under the supervision of

Ao.Univ.Prof. Dipl.-Ing. Dr.med. Dr.techn. Dr.phil. Thomas Angeli

Faculty of Mechanical and Industrial Engineering

reviewed by

Ao.Univ.Prof. Dipl.-Ing. Dr.med. Dr.techn. Dr.phil. Thomas Angeli

Faculty of Mechanical and Industrial Engineering

Getreidemarkt 9, 1060 Vienna, Austria

DIPLOMARBEIT

Entwicklung einer Prothesensteuerung zur Unterstützung des Rehabilitationsprozesses

ausgeführt zum Zwecke der Erlangung des akademischen Grades einer Diplom-Ingenieurin (Dipl.-Ing. oder DI), eingereicht an der TU Wien, Fakultät für Maschinenwesen und Betriebswissenschaften

von

Lena TRAUTMANN

Mat.Nr 01526567

unter der Leitung von

Ao.Univ.Prof. Dipl.-Ing. Dr.med. Dr.techn. Dr.phil. Thomas Angeli

Fakultät für Maschinenwesen und Betriebswissenschaften

begutachtet von

Ao.Univ.Prof. Dipl.-Ing. Dr.med. Dr.techn. Dr.phil. Thomas Angeli
Fakultät für Maschinenwesen und Betriebswissenschaften
Getreidemarkt 9, 1060 Wien, Österreich

I confirm, that going to press of this thesis needs the confirmation of the examination committee.

Affidavit

I declare in lieu of oath, that I wrote this thesis and performed the associated research myself, using only literature cited in this volume. If text passages from sources are used literally, they are marked as such.

I confirm that this work is original and has not been submitted elsewhere for any examination, nor is it currently under consideration for a thesis elsewhere.

Vienna, September, 2018

Ich nehme zur Kenntnis, dass ich zur Drucklegung meiner Arbeit unter der Bezeichnung

Diplomarbeit

nur mit Bewilligung der Prüfungskommission berechtigt bin.

Eidesstattliche Erklärung

Ich erkläre an Eides statt, dass die vorliegende Arbeit nach den anerkannten Grundsätzen für wissenschaftliche Abhandlungen von mir selbstständig erstellt wurde. Alle verwendeten Hilfsmittel, insbesondere die zugrunde gelegte Literatur, sind in dieser Arbeit genannt und aufgelistet. Die aus den Quellen wörtlich entnommenen Stellen, sind als solche kenntlich gemacht.

Das Thema dieser Arbeit wurde von mir bisher weder im In- noch Ausland einer Beurteilerin/einem Beurteiler zur Begutachtung in irgendeiner Form als Prüfungsarbeit vorgelegt. Diese Arbeit stimmt mit der von den Begutachterinnen/Begutachtern beurteilten Arbeit überein.

Wien, September, 2018

Abstract

In rehabilitation, amputees need to regain mobility to participate in everyday life. Frequent repetition and specific instructions for improvement determine the success. The current training application of Otto Bock Healthcare only includes general instructions, but does not count repetitions, detect poor execution, or give feedback for improvement. Hence, a finite state machine is implemented for five rehabilitation exercises, assisting amputees in autonomous training at home. This work presents the exercises using an anthropometric model and analysing the associated sensor signals, summarises the exercises' specification, and describes the resulting structure of the implementation. Furthermore, it evaluates the functionality of feedback and compares the assessment of repetitions by a physiotherapist versus by the control in a usability test.

For all five exercises, poor execution can be detected as long as the exercise is performed without support by the arms, but compensatory movements can only be differentiated for one exercise. Additionally, amputees can improve their performance with the aid of the finite state machine's feedback after a set, but further testing is needed to assess the benefit of real-time audio feedback during execution. In conclusion, the implemented collection of finite state machines represents a first success for guided autonomous training and the collected data constitutes a good foundation for further adjustments.

KURZFASSUNG

Im Rahmen der Rehabilitation müssen Amputierte ihre Mobilität zurückgewinnen, um am täglichen Leben teilnehmen zu können. Häufige Wiederholungen und spezifische Verbesserungshinweise bestimmen den Erfolg. Die momentane Trainingsapplikation von Otto Bock Healthcare enthält nur generelle Instruktionen, zählt jedoch keine Wiederholungen, erkennt keine fehlerhaften Ausführungen und gibt keine Möglichkeiten zur Verbesserung. Daher wurde für fünf Rehabilitationsübungen ein endlicher Zustandsautomat programmiert, der Amputierte beim eigenständigen Training zu Hause unterstützt. Diese Arbeit präsentiert die Übungen anhand eines anthropometrischen Modells und einer genauen Analyse der zugehörigen Sensorsignale, fasst die Spezifikationen der einzelnen Übungen zusammen und beschreibt die daraus abgeleitete Struktur der Implementierung. Außerdem wird in einem Anwendertest die Effektivität der Verbesserungshinweise untersucht und die Beurteilung der Wiederholungen von einem Physiotherapeut und der Steuerung verglichen.

Bei allen fünf Übungen können schlechte Ausführungen erkannt werden, solange die Übung ohne zusätzliche Unterstützung durch die Arme ausgeführt wird, jedoch können nur für eine Übung die Ausgleichsbewegungen unterschieden werden. Zusätzlich kann eine Verbesserung der Übungsausführung von Amputierten aufgrund des Feedbacks, das der endliche Zustandsautomat nach einem Set zur Verfügung stellt, festgestellt werden. Der Nutzen von akustischem Echtzeit-Feedback zur Verbesserung der Übungsausführung muss jedoch genauer getestet werden. Somit stellt die implementierte Sammlung endlicher Zustandsautomaten einen ersten Erfolg für unterstütztes, eigenständiges Training dar und die gesammelten Daten bilden eine gute Grundlage für weitere Anpassungen.

Contents

List of Tables

List of Figures

List of Abbreviations

BM	Body Mass
COM	Centre of Mass
FA	Finite Automaton
FSM	Finite State Machine
JC	Joint Centre
MPK	Microprocessor-controlled Prosthetic Knee
QU	Quantisation Units
ROM	Range of Motion
(P/CL)HJC	(Prosthetic/Contra-lateral) Hip Joint Centre
(CL)KJC	(Contra-lateral) Knee Joint Centre
(CL)AJC	(Contra-lateral) Ankle Joint Centre
Signals:	
AX	Axial Load
DE	Damping Extension
DF	Damping Flexion
KA	Knee Angle
LA	Leg Angle
PA	Pitch Angle
RA	Roll Angle
YA	Yaw Angle

1. Introduction

After the amputation of a lower extremity, the patient needs to relearn walking and other daily activities like sitting down, climbing stairs, and walking backwards to open a door [1, 2]. For this, guided rehabilitation is inevitable [1, 3–6]. The effect of rehabilitation is increased by specific gait training [6, 7]. A physiotherapist should assist this process [8], increasing safety and providing the knowledge of a variety of exercises adjusted to the amputee's personal needs. A laboratory environment reduces the amount and diversity of impacts and therefore risk.

Guided rehabilitation is influenced by the repetition [9]. If the exercises are repeated on a regular basis at home, the patient attains a routine of the underlying movement patterns. Not only the number of repetitions determines the success of rehabilitation, but also the instructions given for improvement. In the training app *Fitness für Amputierte* by Otto Bock Healthcare, home training is only assisted by text instructions and small animations, but the patient receives no feedback about his performance.

The current prosthetic control optimises the gait pattern and additionally facilitates daily activities [10]. This control detects predefined patterns and supports the corresponding movement by adjusting the damping [11]. While exercising, certain muscles shall be activated and shall work against a resistance. Since rehabilitation exercises are done to improve everyday life, they resemble daily movements [1]. To increase the training effect, the prosthetic control shall enable the patient to perform the exercise, but support as little as possible. If the exercises are done in basic mode, the motion is supported, reducing the muscle activation and therefore the training effect.

The motivation for this work is to provide amputees with guidance for autonomous practice at home [12]. The main objective is to programme a prosthetic control, assisting amputees of the lower extremity while executing selected exercises. This control shall fulfil the following requirements: First of all, safety shall be guaranteed at all times, whenever it can be provided by the prosthesis. That is, the damping is adjusted to the patient's abilities. Second, all necessary movements shall be enabled. For exercises executed with a stretched leg no movement is required. During drills that involve a bending motion, the damping is controlled continuously. Third, the requested level of intensity shall be adjustable. It defines the necessary range of motion (ROM) to count a repetition as correctly, evaluating the quality of execution, and the resistance of the prosthesis, determining the increased difficulty of the exercise by additional damping. Furthermore, the number of repetitions shall be determined. Correct and incorrect repetitions are counted separately to give the user detailed feedback on what to improve. The exercises are chosen from a catalogue developed in a previous project by K. Gatas [13].

The background chapter introduces relevant knowledge on amputation levels and mobility of amputees. In addition, it presents the prosthetic knee in detail, focusing on the target user, its software, and hardware components. The rehabilitation chapter discusses the available exercise catalogue and gives instructions for correct execution of the implemented exercises. Furthermore, it analyses these exercises with a biomechanical model. The following chapter elaborates upon the technical implementation of the exercises, dictated by the available sensor signals and the software limitations. This implementation is tested with an amputee. The results are presented in the next chapter evaluating the functionality of the prosthetic control and the concordance of its classification and a physiotherapist's opinion. Finally, the main conclusions are summarised and possible future work is recommended.

2. Background

Since it is the primary goal of this project to design a control which assists amputees with MPKs (Microprocessor-controlled Prosthetic Knee), this chapter summarises fundamentals on amputations and the used MPK. The short introduction on amputations focuses on amputation heights, defined by the level of amputation, and classification of the mobility of amputees. Then, the used prosthesis is introduced in detail: First, basic functionality is described. Afterwards, hardware components are presented. Finally, the theory of finite state control is summarised, specifying definitions of the used concept, and the development-environment is introduced.

2.1 Amputations

When discussing details on amputations, it is important to keep in mind that an amputation itself is no disease, but a consequence of a disease [14]. In 2017, an overall number of 70 877 amputations are registered in Germany 25 %, affecting the lower extremity [15]. Arterial occlusive disease, mainly diabetes mellitus, caused 87 % of amputations [16–18]. Trauma, Infections, tumors, and other causes each make up less than 5 % [16]. The cause and its severity influence the amputation height which will be discussed in Section 2.1.1. Moreover, prosthetic treatment depends on the amputation height, personal needs, and a classification presented in Section 2.1.2.

2.1.1 Level of Amputation

Amputations are distinguished between macro and micro amputations [17]. In case of the lower extremity, micro amputations are distal of the upper ankle joint, macro amputations are proximal. Dealing with a knee prosthesis, all amputees have a macro amputation. Furthermore, amputation and disarticulation are differentiated [17]. An amputation is defined as a cut through the bone and a disarticulation is an amputation with a partial removal of a joint without cutting of bone. In addition to this conceptual differentiation, several amputation heights are distinguished (see Figure 2.1). Users of knee prostheses have either had a knee or hip disarticulation or a transfemoral amputation.

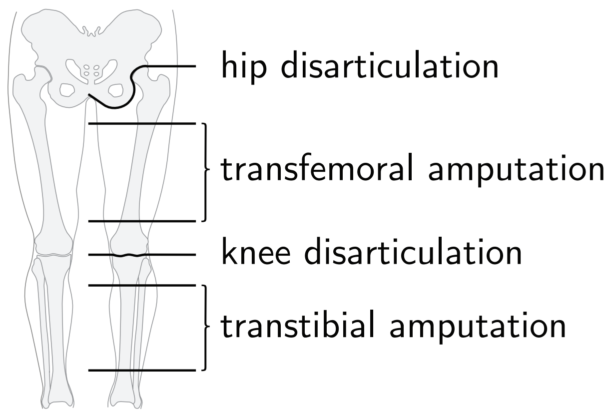

Figure 2.1: Differentiated heights for macro amputations [17], adapted from [19].

The choice of the amputation level is difficult. Several aspects need to be considered. The main goal is to amputate the leg as peripherally as possible while attaining a painless and load-bearing stump [16]. The load-bearing surface should be as large as possible which is achieved more likely if the scar is outside the load-bearing zone [14]. Nevertheless, maintaining tissue is more important than the load-bearing capacity [14, 16] since a longer stump enhances prosthetic control. Furthermore, energy expenditure increases significantly for more proximal amputations due to a higher rate of oxygen uptake and a reduced maximum aerobic capacity [20]. Conversely, a more peripheral amputation demands better skills from the surgeon and increases the risk of failure [16].

Wilde and Baumgartner [16] emphasise in their book that several factors influence the decision-making of the surgeon prior to the operation. According to them, medical history, including character and localisation of pain as well as evaluation of the skin, plays an important role. This assessment is assisted by technical examination methods. Intra-operatively, the general blood circulation can be assessed. Furthermore, tissue and muscles can be evaluated when cut through since superficial thermal and electrical burns make it difficult to assess lower tissue. They also note that all those decisions are rather subjective. Generally, anatomical conditions, aetiology, and thrombosed venes can lead to more proximal amputations, but the fear of complications should never be a reason to amputate more proximally [14, 16].

2.1.2 Mobility Classes

Mobility classes categorise the ambulatory skills of the user. Universal use by suppliers, industry, and accounting units of public health insurance companies shall facilitate to define the therapeutic goal, select appropriate treatment [17] and claim insurance.

Contrarily to this aim, many mobility predictors exist. In the United States, the *Medicare Functional Classification Level* [21, 22], also called K-levels, defined by the Human Care Financing Administration, are used to classify insurance matters. In Germany, the Medizinischer Dienst des Spitzenverbandes defined *Äktivitätsklassen* (eng. activity classes) [23]. Both systems differentiate 5 categories: not capable of walking (k0), walking ability indoors (k1), limited walking ability outdoors (k2), full walking ability outdoors (k3), and full walking ability outdoors with high demands (k4). The German system additionally defines therapeutic goals.

Furthermore, many manufacturers use their own mobility classification. This also applies to Otto Bock Healthcare's MOBIS system [24, 25] which is based on the K-levels and additionally accounts for the patients' weight. For a detailed description see Appendix A.

Additionally, output measures exist such as the Functional Independece Measure [26], Amputee Mobility Predictor [22], the Patient Assessment Validation Evaluation Test, Timed Up and Go, and several others. According to a US study [27], 90 % of practitioners are involved in the assignment of K-levels. Two thirds of respondents think that the K-levels are not representative for the functional level of the amputee and 75 % suggest to incorporate outcome measures into the assignment process to increase objectivity.

This study shows that the usage of mobility classes still needs to be improved. Reducing the number of currently applied mobility measures could improve comparability of different studies. Alternatively, settling a universal approach to assist K-level-classification could facilitate cooperation and exchange of experience of all involved experts and simplify the treatment procedure. Both approaches would ultimately serve the patient.

2.2 Kenevo – Microprocessor-Controlled Prosthetic Knee

2.2.1 Functionality

Otto Bock Healthcare developed the Kenevo, a mono-centric MPK, offering increased security to amputees with reduced motor control. This introduction in the Kenevo's functionality is derived according to an information booklet for technicians, published by Otto Bock Healthcare[10]. It is specifically designed to assist patients of MOBIS 1 & 2 in their daily activities. Therefore, it provides basic modes to ameliorate safety and to support the user during stand-to-sit, sit-to-stand, stance-phase, and while sitting in a wheel-chair. Additional safety is guaranteed by increased flexion impedance in case of stumble, aborted motion, overheating of the prosthesis, or an empty battery.

Moreover, three different activity modes enable to adapt the behaviour of the prosthesis to the current abilities of the amputee. The first mode, locked mode, guarantees maximal safety by a locked knee. Nevertheless, the freedom to move is given by the basic modes. The second mode, semi-locked mode, increases the safety while walking by easily enabling the swing-phase. Additionally, if stance-phase flexion is activated, the knee is locked during stance-phase at a flexion of 10°. This enables amputees to walk alternately on moderate ramps. The last mode, yielding mode, facilitates more dynamic movements by high impedances, for example walking on uneven terrain, ramps, and stairs.

The Kenevo weighs 1.2 kg, including the tube adapter and is approved up to a body weight of 125 kg.

2.2.2 Hardware

Damper

The Kenevo is an energetically passive device with a microprocessor-controlled two-way hydraulic damping unit (see Figure 2.2), allowing to set different flexion and extension resistances simultaneously and independently. Therefore, no delay is introduced, when switching direction between flexion and extension. The damper consists of a piston, displacing oil in the two chambers (A & D) of the cylinder. During flexion of the knee, the oil flows from chamber A through valve 1 into the additional chamber B and through check valve 3 into chamber D. The oil flowing

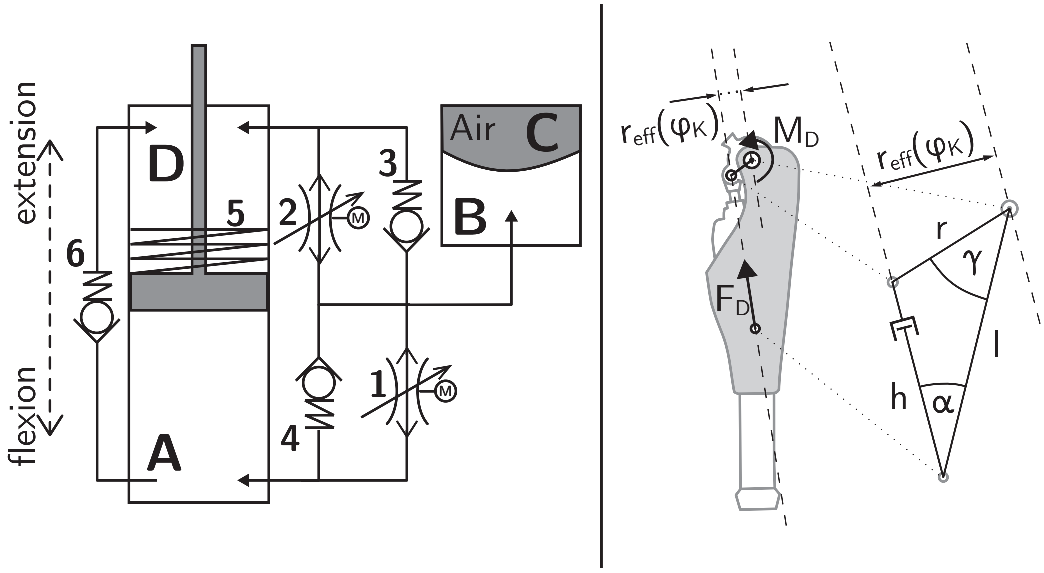

Figure 2.2: Schematic of the hydraulic circuit (left), adapted from [11, 19]. The two chambers A and D are connected by the servo valves for flexion (1) and extension (2) and the corresponding check valves (3&4). Chamber B and C define minimum flexion resistance, the vulkollan element (5) the extension stop. The additional check valve (6) opens in case of overload during flexion. Kinematic relationship of the hydraulic damping unit defined by the prosthesis geometry (right). The effective lever arm r_{eff} depends on the knee angle ϕ_K. The torque generated by the damper is defined by the damper's force F_D and r_{eff}.

into chamber B pushes against the membrane of the air pressure storage. The air pressure defines the minimum flexion resistance and is set at production. During extension, fluid flows from chamber D through the extension valve 2 and the check valve 4. Additionally, energy is stored in the air spring, pushing the oil out of chamber B. The cross-section of the check valves is larger than of the servo valves in parallel such that the resistance is defined by the servo valve in series. A vulkollan® element on top of the piston (5) allows a smooth stop of the extension motion. In case of low battery, the servo valves are adjusted to high flexion impedance and to low extension impedance, guaranteeing safety and enabling a fast extension. In case of overloading during flexion, the control valve 6 opens and the oil flows from chamber A to chamber D.

As this damping unit is installed in a piston-rod-connection, a torque is generated around the knee axis. The force of the damper F_D generates are torque M_D which are defined by

$$F_D(\phi_K) = -d \cdot \dot{x}(\phi_K) = -d \cdot \dot{\phi}_K \cdot r_{eff}(\phi_K), \tag{2.1}$$

$$M_D(\phi_K) = F_D \cdot r_{eff}(\phi_K) = d \cdot \dot{\phi}_K \cdot r_{eff}^2(\phi_K), \tag{2.2}$$

with d, the damping coefficient, ϕ_K, the knee angle, $\dot{\phi}_K$, the knee angle velocity, and r_{eff}, the effective lever arm.

The opening of valve 1 and 2 can be set by means of the valve angle. If the knee angle velocity ($\dot{\phi}_K$) is zero, the damping coefficient d increases exponentially with the valve angle, rising from 15° to a blocking position. To provide a more intuitive definition of damping values during ruleset-programming, a characteristic line maps the valve angle on to damping values for flexion and extension, d_F and d_E respectively. The damping values range from 0 (open valve) to 200 (closed valve). The resolution amounts 200 quantization units (QU). The characteristic line accounts for differences in prostheses, ensuring the exponential increase in the damping coefficient for a linear increase in the damping value. The change in valve position between the damping values 199 and 200 is larger to guarantee a locked prosthesis at 200. This step is not perceptible if the characteristic is determined correctly.

The effective lever arm $r_eff(\phi_K)$ of the damping force around the knee axis is defined by the kinematic of the Kenevo (see Figure 2.2). The shank length l measures 145.03 mm and the lever arm r is 24 mm. The angle γ depends on the preflexion of the knee, the offset of the shank axis, and the angle of the knee-head, equal to

$$\gamma = 97.8° - \phi_K. \tag{2.3}$$

Using the law of cosines, the length of the hydraulic axis h can be determined as

$$h = \sqrt{l^2 + r^2 - 2lr \cdot cos(\gamma)}. \tag{2.4}$$

Table 2.1: Description and discretisation of the sensor signals used in the ruleset library. The abbreviation refers to the sensor signal. The symbol is used in mathematical notations.

Description	Abbreviation	Symbol	Discretisation
Axial load	AX	F	10 N
Roll angle	RA	ϕ_R	0.1°
Pitch angle	PA	ϕ_P	0.1°
Yaw angle	YA	ϕ_Y	0.1°
Knee angle	KA	ϕ_K	0.1°
Knee angle velocity	VKA	$\dot{\phi}_K$	$1\,\frac{°}{s}$
Position flexion valve	DF	d_F	1 QU
Position extension valve	DE	d_E	1 QU

By applying the law of sines using l, γ, r, and α and expressing $sin(\alpha)$ with r_{eff} and l, one can derive a formula for the effective lever arm:

$$r_{eff}(\phi_K) = \frac{lr \cdot sin(\gamma)}{h}. \tag{2.5}$$

Hence, for a knee angle of $\phi_K = 97.8°$, no torque can be acting which is called the hydraulic dead-centre. For knee angles larger than the dead-centre, the lever arm is negative. Hence, the flexion damping regulates the extension movement and the extension damping the flexion movement. Trigonometric functions are not available in the programming environment and therefore the resulting function of r_{eff} is interpolated and implemented as a look up table. The maximum error is 0.3 mm at 62.2°, equivalent to a relative error of 2 %.

Sensors

Several sensors are installed in the prosthesis, giving information about the spatial position, dynamics, and forces. This is important for evaluating the progress of the current repetition. The available sensor signals are depicted in Figure 2.3. The measurement accuracy of all signals used is listed in Table 2.1. The abbreviations are used to refer to the sensor signal and to their characteristics, if specified by a subscript. That is, for example the range of axial load, referenced as AX_{range}.

3D-Sensor The 3D-Sensor is an Inertial Measurement Unit. It consists of a triaxial gyroscope, measuring the angular velocities around the axes, and a triaxial accelerometer, measuring the accelerations along the axes. Traid-Bryan roll, pitch, and yaw angles, as used in aerospace applications defined by DIN 9300 [28], are

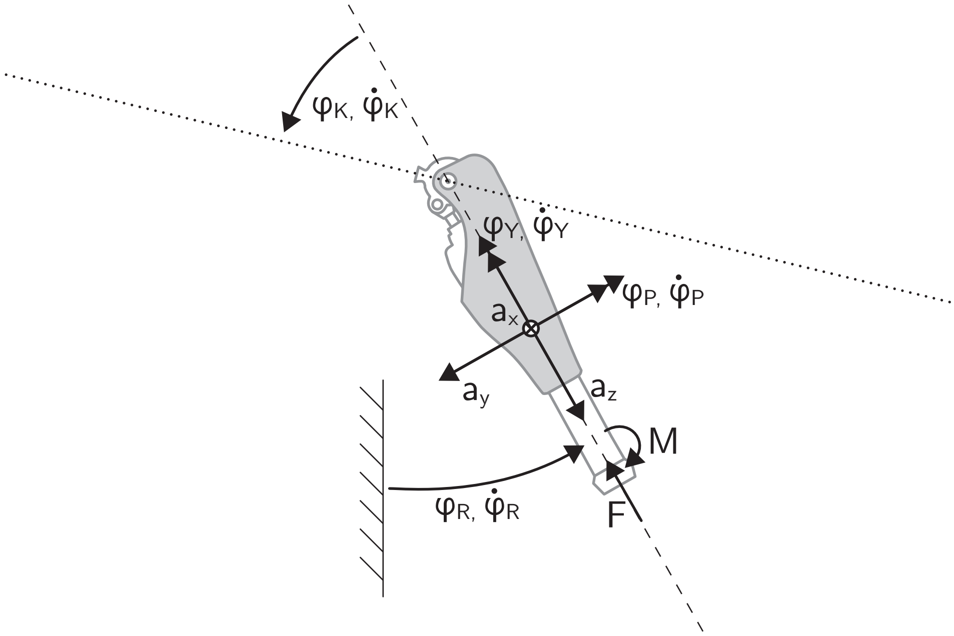

Figure 2.3: Sensor signals measured at each time step.

calculated from accelerations, as well as from angular velocities. To calculate the roll and pitch angle from accelerations, the system needs to be at rest. With this approach the yaw angle can not be determined. For the second approach, angular velocities are integrated to determine the position of the body in the global coordinate system, assuming the initial position at time $t = 0$ is known. Due to measurement errors, the integration of the signal results in a drift of the calculated angles. Hence, a fusion algorithm weighs the two approaches, favouring the acceleration-approach at rest. The yaw angle remains drifting as it can not be determined by accelerations.

Knee angle sensor The knee angle, referring to the angle at the knee joint, is measured with a magnetometer integrated into the axis of the knee. The knee angle velocity is derived by differentiation.

Strain gauge elements The force along the axis of the tube adapter and the ankle torque in the sagittal plane are determined slightly above the distal end of the tube adaptor. Strain gauge elements are positioned around the circumference. If the tube adapter is analysed in the sagittal plane, the axial load is determined by a uniformly distributed load with an amplitude equal to the mean of all elements. Consequently, the torque is determined by the remaining triangular loads multiplied with the respective lever arm.

Micro-controller

The integrated micro-controller processes the measured sensor signals and regulates the hydraulic valves according to the ruleset. The embedded software is executed on the micro-controller and runs the ruleset library.

2.2.3 Software

Finite State Control

The use of finite state control for an artificial leg was suggested in 1966 by Tomovic and McGhee [29] and further elaborated in the years after [30]. Finite automata (FA) are used to control the sequential behaviour of discrete dynamic systems non-analytically [31, 32]. Since the gait consists of recurring patterns, FA are appropriate for control. In the following, FA will be defined according to Hopcroft [33] and Kunze [32]. The tuple

$$A = (S, \Sigma, \delta, s_0, F), \tag{2.6}$$

defines an FA, with S, the finite set of states, Σ, the finite set of input symbols, $\delta \subseteq S \times \Sigma \times S$, the state transition relation, $s_0 \in S$, the initial state, and $F \subseteq S$, the set of final or accepted states.

If for each state and for each input symbol at most one state transition relation exists, the automaton is said to be deterministic. Then, the transition relation becomes a transition function $\delta : S \times \Sigma \to S, p = \delta(q, a)$, with q, the current state, a, the input symbol, and p, the next state. Hence, a deterministic FA is always in only one state at a time.

Additionally, automata can provide output. The output depends either on the reached state, called Moore automata, or on the state transition, called Mealy automata. A Moore automaton is defined by the tuple

$$A_{Moore} = (S, \Sigma, \Lambda, \delta, \lambda, s_0, F), \tag{2.7}$$

with $(S, \Sigma, \delta, s_0, F)$, a deterministic automaton, Λ, the finite set of output symbols, and $\lambda : S \mapsto \Lambda^*$, the output function that assigns an output sequence to each state.

Furthermore, FA can be extended with variables, assignments, conditioned transitions, and time. Values are assigned to variables which thereby add information to a state. Conditions allow to restrict a transition to a logic expression on input, variables, and parameters. Therefore, transitions can not only be triggered by input, but also by fulfilled conditions independent of the input. Besides, timed automata can extend the solely causal relation of automata with time-dependent transitions. In this case, a local clock is introduced, often implemented as an integer counter incremented at each time step.

Finite automata are also often referred to as finite state machine (FSM). The FSM of the Unified Modeling Language which defines an industry standard is based on the concept of extended FA.

Two more differentiations exist. Systems are classified as synchronous or asynchronous [34]. Synchronous systems are timed by a device, called clock, producing a train of equally spaced pulses. These pulses determine the rate at which the FSM is evaluated. Asynchronous systems are evaluated at the occurrence of an input and are thereby usually faster.

Finally, finite state automata can be either discrete or mixed discrete-continuous systems. In a classic finite automaton, state outputs are constant, resulting in a discrete system. In hybrid finite automata, the output can be changing within one state, resulting in a mixed discrete-continuous system [35].

The finite state control concept used for this project is a hybrid, synchronous, and deterministic FSM which will be presented in the next section.

Ruleset Library

To apply these concepts of finite state control to the control of the prosthesis, a supporting code, called ruleset library, is necessary. The ruleset library is the entirety of the programme which is run on the micro-controller. It contains several FSMs, called rulesets, which are explained in detail later on. Furthermore, the library defines syntax of the code, names of sensor signals, global variables, parameters, and events. The two available events contain commands which are executed, when a parameter is changed or when the prosthesis is switched on or reset. The library is designed in the in-house development environment Bioleg and saved as XML-file. Additionally, rulesets can be defined as modes in the XML-file, enabling the user to switch between these on a remote or mobile application. This XML-file is translated by the interpreter and the resulting file is transferred to the micro-controller via bluetooth. Due to storage restrictions on the prosthesis, the

Table 2.2: Available boolean and arithmetic operations, exemplary functions, and system commands.

Function	Description
AND, OR, NOT	Boolean operators
<, <=, ==, >=, >	Boolean comparators
+, -, *, /	Arithmetic operators
min(x), max(x)	Minimum and maximum of the argument x, within one continuous execution interval
liml(x,l), limu(x,u)	Limits the argument to the given lower limit l ([l,∞]) or upper limit u ([-∞,u]), respectively
ipol(x,{x1, y1, x2, y2, ..., x6, y6})	Interpolates the supporting points linearly and calculates the function value for the argument
select(*memset*, *ruleset*, *rule*)	Loads the rule of the ruleset on the memset
jmp(*rule*)	Transitions to the specified rule
en/dis_set(*memset*)	Activates/ Deactivates the memset
beep({*pattern*}, *freq*, *vol*)	Defines an acoustic signal by bit pattern, frequency and volume

Table 2.3: Available variable types, their naming convention, and accessibility.

Type	Naming Convention	Availability
local variable	name	rule level (while continuously active)
slider	SL_NAME	library level
global variable	@NAME	library level
adaptive parameter	AP_NAME	library level (available after a reset)

transferred file is limited to 32 kb. Hence, the library must be designed compactly and efficiently. The micro-controller can load four rulesets, one on each of its four execution positions, called memset, which are evaluated consecutively every 10 ms. Thereby, four rulesets can be executed almost synchronously. To explicitly refer to a ruleset, its name is written in small capitals, for example REFERENCE.

Before describing the ruleset in more detail, the programming language is introduced: Boolean and arithmetic operators are available, as well as basic mathematical functions and system commands (see Table 2.2). The mathematical functions are limited, not including, for example, trigonometric or exponential functions. System commands allow to change the rule or the ruleset on a specific memset, as well as enable or disable memsets at runtime.

Moreover, different types of variables are available in the programming environment (see Table 2.3). Local variables, global variables, and sliders are used in the implementation. The number of variables of each type is restricted. Local

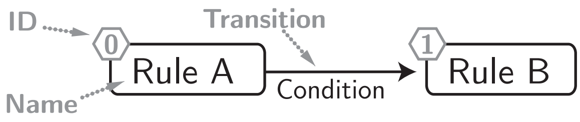

Figure 2.4: Terminology and visualisation of a ruleset.

variables are available in the currently active rule until the next transition. That is, at each entry, local variables are reinitialised.

Global variables are accessible in the complete library. They can be modified on each memset and are only reset when the prosthesis is reset or switched on. To identify global variables, they are labelled with @ at the beginning and capital letters in the following chapters. To enhance readability, further specifications are added to the variable as indices, for example @VARIABLE$_{\text{SPECIFICATION}}$.

Sliders are parameters which are set before run-time. Their authorisation to access and adjustment range is defined in the library. The orthopaedic technician sets slider values and, depending on the access settings of each slider, the user can be enabled to change them by remote. Nevertheless, the ruleset cannot modify them at runtime. Sliders are also available at the library level.

Each ruleset is a hybrid, synchronous, and deterministic FSM, consisting of several states, called rules and identified by either ID or name, which are connected by conditioned transitions (see Figure 2.4). To identify rules within the text, their names are written in the font "sans serif", for example Init. Within each rule, the behaviour of the system is defined, regulating the output variables. Since it is a hybrid FSM, the output may vary within one state, depending on current input, variables, and parameters. Generally, the damping values are the output. It is the aim of this project to assess exercise executions. Therefore, feedback counters and tones are additional output of this ruleset library. The counters are implemented as global variables to which integer values are assigned.

Each rule consists of entry, default, exit, and timer section. The entry is evaluated once when entering the rule, followed by an evaluation of the default section. The default is evaluated at each time step if no transition condition is fulfilled. If a condition is fulfilled, the exit section is evaluated, before transitioning to the next rule. Additionally, timers enable a timed condition, independent of

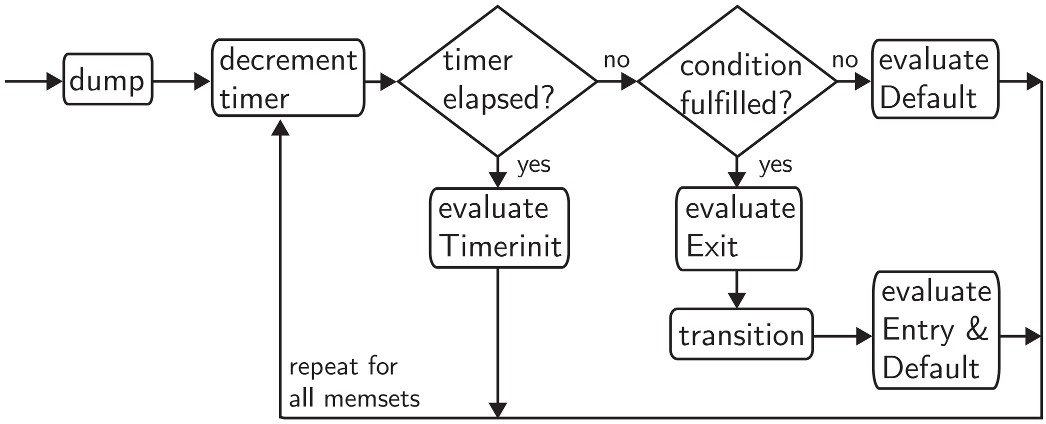

Figure 2.5: Flow chart, illustrating the interpretation of the ruleset library at each time step.

sensor signals. A timer is a variable which is initialised once and then decremented at each time step. When it reaches zero, the timer section is evaluated.

At each time step the current sensor signals are dumped. Afterwards, the memsets are interpreted consecutively according to the flow chart visualised in Figure 2.5.

3. Rehabilitation & Biomechanics

In the previous chapter, the fundamentals on amputations and the prosthesis were introduced to explain decisions made concerning rehabilitation. The available sensor signals determine the observability of the exercises and the mobility of the amputees limits the extent to which exercises can be carried out.

This chapter focuses on rehabilitation and biomechanics. After introducing some basic terminology, the existing catalogue of exercises will be described and analysed, focusing on the therapeutic view-point and considering technical feasibility for the final selection. For a better understanding of the chosen exercises, a biomechanical analysis is performed to determine the forces and moments which are overcome by the muscles to stabilise the static end position.

3.1 Terminology

The introduction of basic terminology in this thesis is not complete but rather introduces frequently used terms and can therefore be skipped by the experienced reader.

The three-dimensional body can be cut by three orthogonal planes: the frontal, the sagittal, and the transverse plane. The frontal and sagittal plane are used for the analysis of the exercises. If all motions occur in sideways, upwards, and downwards direction, the motion is happening in the frontal plane. In case of the sagittal plane, the amputee is looked at from the side, moving forwards, backwards, upwards, and downwards. In the analysis, the amputee is always analysed looked at from the front for motions in the frontal plane and from the right side for motions in the sagittal plane.

Movements in the frontal plane are pronation and supination of the foot around the ankle joint (elevating the soles to the outside or inside), abduction and adduction of the leg (raising and lowering) around the hip, and abduction and adduction of the arms around the shoulder joint. In the sagittal plane, the occurring motions are plantar- and dorsiflexion of the foot (lowering and lifting of the toes), flexion and extension of the knee (bending and stretching), and flexion, extension, and hyperextension of the hip and the shoulder (bending, stretching, and stretching further 180°). The explained motions are visualised in Figure 3.1.

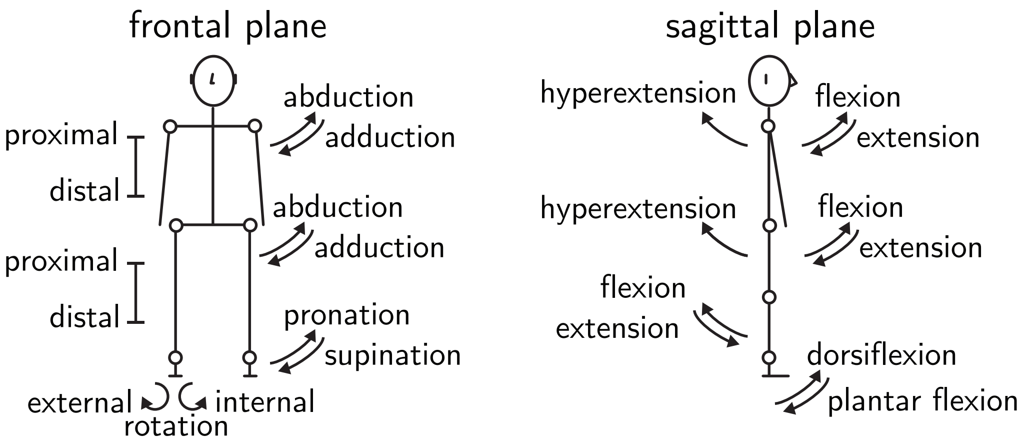

Figure 3.1: Terminology: anatomical planes, directional terms, and motions. Adapted from [36].

Table 3.1: Interrelation of movements and measured sensor signals. The global variable @LEG represents the prosthetic leg to be right (1) or left (-1).

Joint	Negative (-)	Positive (+)	Sensor signal
knee	extension	flexion	KA
hip	external rotation abduction extension	internal rotation adduction flexion	YA · @LEG PA · @LEG KA + RA

Finally, two directional terms are introduced, often used for the description of the extremities. Proximal references the direction towards the body centre and distal away from the body centre.

On the prosthetic leg, less motion is possible as foot and prosthesis are screwed tight, limiting ankle movement to the material's flexibility. Movement of knee and hip can be determined by the sensor signals. Assuming that all movements are executed in solely one plane, the roll and knee angle of the prosthesis describe motion in the sagittal plane. The pitch angle determines the inclination of the prosthesis in the frontal plane and the yaw angle determines rotation around the prosthesis' axis. The interrelation of the sign of the sensor signals and the performed motion is summarised in Table 3.1.

Table 3.2: Evaluation of the exercise catalogue. Selection of five exercises by three physiotherapists (P1, P2, and P3), technical possibility for detection (TD), and final selection (FS)

ID	Name	P1	P2	P3	TD	FS
1	Lateral Weight Shift	x	x	x	x	x
2	Anterior Weight Shift	x	x	x	x	x
3	Criss Cross	x	x	x	x	x
4	Bridging			x	x	
5	Prosthetic Hip Extension				x	
6	Contra-Lateral Hip Extension					
7	Prosthetic Abduction		x		x	x
8	Contra-Lateral Abduction					
9	Push-Up 1 (knees on the ground)				x	
10	Push-Up 2 (at elevated level)				x	
11	Trunk Stabilisation		x			
12	Squat	x		x	x	x
13	Step-Up	x		x		

3.2 Exercise Catalogue

In a previous project, a physiotherapist collected a catalogue of thirteen exercises, aiding the rehabilitation of transfemoral amputees. The aim of this project is to implement a control for these exercises. To determine which drills are performed regularly, three physiotherapists who work with amputees regularly selected their five most important exercises of the catalogue (see Table 3.2). Taking into account their reasoning and the technical feasibility, five exercises are selected for the implementation.

Push-Ups are either performed on the ground with knees on the floor (ex 9) or with extended legs and arms at an elevated level (ex 10), for example against a wall. They were not chosen by the physiotherapists, because only the trunk is trained. This is important for certain gait impairments, for example duchenne, but too basic for most patients.

In hip extension and abduction drills (ex 5 to 8), either the prosthetic or the contra-lateral leg is hyperextended or abducted, strengthening the muscles needed for stance. The difficulty of these exercises can be increased, using a Theraband® which is fixed in front of the patient to train hyperextension or on the side of the standing leg to strengthen abduction. Three of the hip extension and abduction drills (ex 5, 6, and 8) were not selected. The given explanations were controversial. One expert stated that a Theraband® is required to increase intensity to an appropriate level, whereas another reasoned that the increased resistance of the Theraband® strengthens the muscle more than needed in everyday life. If instructed to use a Theraband®, the requirement of having one might result in patients performing the exercise infrequently. Furthermore, the aim of these exercises was discussed. According to two experts, the muscles for stance can be strengthened better with other drills, whereas the third one selected Prosthetic Abduction (ex 7) as an important exercise to strengthen abductors, stressing their importance for the stance phase.

Excluding all exercises which were not selected by physiotherapists, eight drills remain for the further selection progress. Step-Up practises to stand on the prosthetic leg, while stepping on a footstool, step, or chair. Although Step-Up practises a daily problem, it was removed from the selection, because it is not suitable for examination with the available sensor signals.

The advantages and disadvantages of the remaining drills will be discussed next. Lateral Weight Shift, practising the weight shift sideways from one leg to the other, is rated as important by two experts. The third expert considers the drill as too basic but selected it nevertheless. The physiotherapists value the exercise, because it is the simplest drill, easy to practise anywhere and important for a good stance and step. The user familiarises with the limits of heel and forefoot load, as well as lateral hip movement, while maintaining balance. Hence, body awareness is improved. Better users can do the exercise more precisely.

In case of Anterior Weight Shift, the user shifts his weight with the prosthetic foot positioned in front. The user gets to know the range of balanced stance in a step position, improving trust and balance on the prosthesis. This has the same advantages as exercise one, but is more advanced, preparing the user for a step.

For Criss Cross, the user steps forwards, sideways, and backwards. This is very important to enhance balance, reaction, and coordination, while maintaining an upright posture. Furthermore, it prepares for daily activities such as swerving in a crowd.

To perform Bridging, the user lies on his back with knees bent and pushes the pelvis upwards. According to two experts, Bridging is for strengthening only and can be executed and assessed autonomously by the user. According to the third, it is versatile and strengthens various muscle groups. From the technical aspect, the working range of the knee angle includes the hydraulic dead centre, making it impossible to adjust the damping for a constant moment and resulting in a jerky motion of the prosthetic joint.

Prosthesis Abduction strengthens the abductors of the thigh which are important for stabilisation during stance phase. This exercise was only selected by one expert, because all three criticised the significance of good form and two the difficulties and restrictions caused by the use of a Theraband®.

For Trunk Stabilisation, the user holds a Theraband® which is fixed next to him to a stable object with extended arms and walks sideways against the Theraband®'s resistance. Trunk Stabilisation is the second complex exercise, improving balance, posture, and coordination. When executed correctly, it strengthens the lateral trunk muscles, promoting an upright posture. This drill is for advanced users as they need to stand freehand on the prosthesis. Rotation of the trunk cannot be detected with the available sensor signals. Hence, the possibilities to assess this exercise are limited.

By squatting, the user practises to sit down and to load prosthetic and sound leg equally.

Considering all pros and cons, both weight shifting exercises (one and two), Criss Cross (three), Prosthesis Abduction (seven), and Squat (twelve) were chosen to be implemented and tested in this project. Hence, a detailed instruction of these exercises is given now.

Instructions

Initially, all gait training exercises should be done in parallel bars, reducing the weight with both upper limbs to make the amputee feel secure [1, 26]. When amputees feel secure, they can advance to single upper limb support and finally to no upper limb support. Specific instructions for each exercise are given below.

Lateral Weight Shift

The goal of this exercise is to practise a good loading of the prosthesis, while keeping ankle, knee, hip, and shoulder in one line. The feet are positioned hip-width apart. Amputees shift their weight laterally onto the prosthetic leg trying to fully load the prosthesis without lifting the contra-lateral leg. They should move laterally as far as possible, while maintaining balance. Then, they shift their weight back onto the contra-lateral leg, unloading the prosthesis. The knees are extended at all times. During the movement the torso shall stay upright, that is, the torso may not be flexed in the frontal plane. Placing the hands on the hip can assist moving the torso in unity with the hip and maintaining balance without the arms. The whole motion is performed at moderate speed.

Anterior Weight Shift

This exercise shall prepare amputees to step fully onto the prosthesis and maintain security and balance. The feet are in step position with the prosthetic leg in the front. Amputees shift their weight forwards onto the prosthesis, trying to fully load the prosthesis without lifting the contra-lateral leg. They should try to move forwards as far as possible, while maintaining balance. Then, they shift their weight back onto the contra-lateral foot, unloading the prosthesis. During the movement, the torso shall stay upright. Lateral torso lean can be avoided by sufficient lateral weight shift. The whole motion is performed at moderate speed.

Criss Cross

During this exercise, the prosthetic leg always stays on the ground. Amputees start from a normal standing position and step forwards, sideways, and backwards. During the stepping motion, the weight shall be supported by the prosthetic side. Amputees need to stabilise the pelvis to maintain balance. This is achieved when lateral weight shift is sufficient. During the movement, the torso shall stay upright above the pelvis. The whole motion is performed at moderate speed. For lower intensities an intermediate step to the normal standing position is allowed after each step to the front, side, or back.

Prosthetic Abduction

Amputees start from a normal standing position and raise the prosthesis in the frontal plane. Good form is defined by internal rotation of the leg and hip extension. Hence, the prosthetic foot always points to the front and stays behind the frontal plane through the contra-lateral toes. The prosthesis shall be raised as high as possible, while maintaining internal leg rotation and hip extension. The leg is then adducted, still keeping good form. To end each repetition the prosthetic foot is set aside the contra-lateral foot. The exercise shall be executed with moderate speed and a controlled motion. The intensity can be additionally increased by fixing a Theraband® in a loop around the ankles or attaching light weights to the prosthetic leg.

Squat

Amputees stand with their feet shoulder-width apart. Motion is initiated by pelvis and torso. Amputees push their pelvis back, lower their torso, and extend their arms to the front, while bending their knees. Good form is achieved if the knees stay posterior to the toes, heels do not lift, knees point outwards, and weight is distributed equally on both feet, resulting in a symmetric execution. Amputees slowly bend their knees as far as they can, maintaing good form. Outward rotation of the knees can be assisted by a Theraband® looped around the knees, activating the muscles with a slight tension.

3.3 Biomechanical Analysis

Until now, the technical background and the aspects concerning rehabilitation were analysed. The next step is to gain further knowledge of the selected exercises by a biomechanical analysis. Joints are stabilised by muscles, resulting in forces and a moment of force (also referred to as moment or torque) in each joint. To estimate the joint reactions, a model is established and then applied to each exercise.

3.3.1 Anthropometric Model

The muscles stabilising a joint generate forces by contraction. This force and the muscle's attachment and origin determine the moment generated within the joint. The biomechanical analysis determines resultant forces and moments achieved by the muscles to stabilise the static position. Nevertheless, the generated forces within the muscles are larger since the joint is stabilised by a group of agonists and antagonists with counteracting forces.

To perform the analysis, a model needs to be developed. Modeling the human body is very complex since each human is individual and deviates from the mean in for example bone geometry, muscle formation, and mass distribution. The interplay of muscle groups results in complex three-dimensional motions due to changes in the instantaneous centre of rotation and the lines of action. As the muscles contract during motion, they vary their length and shape, also altering mass distribution.

The established model simplifies the human body excessively, based on the following assumptions. A transfemoral amputee's body consists of seven segments: torso and head, two arms, stump, thigh, calf, and foot. The prosthesis is treated as eighth segment. Torso and head are modelled by a single segment, assuming the head to always be in extension of the torso. The arm is assumed to be extended at all end positions. Hence, upper arm, forearm, and hand are in line and are summarised into one segment.

Although the mass distribution of muscles changes during contraction, each segment apart from the prosthesis and the contra-lateral foot is estimated as a circular cylinder with a fixed mass distribution. Hence, the centre of mass (COM) lies on the longitudinal axis at a fixed percentage of the segment's length, independent of the segment's position. Prosthesis and foot are also assumed to have a fixed mass distribution, but are estimated with an L-shaped geometry. Hence, they do not only have a COM position along the longitudinal axis, but also a COM position in the anterior-posterior direction within the sagittal plane.

Each drill is assumed to be executed in either solely the frontal or the sagittal plane and only the static end position is evaluated. This allows to calculate the resulting forces and moments in the respective plane.

The segments are connected with simple joints. Assuming that all drills are executed with good form, the legs remain constantly at the neutral position, that is, they are not rotated ex- or internally. Hence, the hip needs not to be modelled as a ball and socket joint. For the analysis in the frontal and the sagittal plane, foot, hips, and shoulders are modelled as hinge joints. The knee is simplified as a hinge joint in the sagittal plane. Furthermore, the foot is assumed to have full contact between sole and ground at all times, resulting in a constant ankle height.

To complete the model, segment lengths, position, masses, and COMs need to be determined, as well as ground reaction forces and their point of application. The physiotherapist marked joint centres with adhesive points, following the guidelines of the International Society of Biomechanics [37]. Due to limitations on the experimental setup and equipment, the positioning of the markers is inaccurate. The lengths of the segments were read from the measuring tape with an accuracy of ± 5 mm. The amputee performed the exercise on a force plate. Ground reaction force and point of application were measured while the amputee remained in the static end position of each exercise. Otto Bock Healthcare's 3D L.A.S.A.R. Posture was used as force plate and adapted with in-house software to access the necessary data. Since the two plates of the 3D L.A.S.A.R. Posture are inseparably connected and some drills requested a larger distance between the feet than the plates' distance, two measurements had to be made for the respective exercises. Although the geometry was reproduced as well as possible for the second measurement, the separation into two measurements induced errors in the ground reaction force, its point of application, and the determined position of the segments. The position of the amputee relative to the force plate and the position of the segments relative to each other were determined from one single photo per position. To reduce distortion, the camera was placed in 8 m distance. Nevertheless, the force plates were looked at slightly from above, seen as a trapezoid in the photo. To determine the ankle joint centres (AJC) relative to the force plates' coordinate system, their position in the photo was projected onto the abscissa, considering the distortion of the force plate. The position of the segments is defined by angles measured from the photograph using Kinovea [38], an open-source software to analyse motion. When two measurements were necessary to determine the ground reaction forces, the angles were determined from the photo belonging to the measurement of the prosthetic leg. The segments' mass and COM are approximated using anthropometric data.

Several anthropometric studies have been carried out in the past by Harless [39], Braune & Fischer [40], Fischer [41], Dempster [42], and Clauser et al. [43], providing a variety of data. It is important to use data in which population represents the modelled amputee [44]. Hence, the data of Clauser et al. [43] will be used, where age and height lie within the covered range of the population. Although the subject's weight was outside the range of the population, the data of Clauser et al. are used, limiting the accuracy of mass and COM. The data published by Clauser does not consistently define segment boundaries at joint centres (JC). Hinrichs [45] adjusts segment lengths and COM/segment length ratios to segments defined bei JCs. According to this study, the segment mass data can be used from Clauser without adaptation. To adjust this data to the amputee, further adaptations are necessary.

The segmental weight is related to the total body weight of non-amputees. To be able to use the segmental weights, the mass M of the amputee needs to be adjusted to his equivalent mass M^*, including the amputated calf and foot. The mass M of the amputee was measured, wearing the prosthesis and shoes. Thus, the mass of all body parts M_{Body} equals (90.7400 ± 1.0001) kg and represents (94.2 ± 0.1) % of the equivalent mass, resulting in an equivalent mass of (96.33 ± 1.07) kg.

The segment length was measured for thigh, calf, foot, socket, prosthesis, the arm from Shoulder JC to Wrist JC, and the height of the trunk between Hip Joint Centre (HJC) and Shoulder JC. Since Hinrichs only reports corrected data for single segments, missing lengths are approximated using the mean lengths given by Hinrichs and Clauser [43, 45] and assuming that the ratio of the segments is equal for the reported data and the amputee. The lengths of upper arm, forearm, and palm are calculated from the measured arm length between Shoulder JC and

Table 3.3: Anthropometric data of the segments. The COM is determined from the segments' length and the COM/Length ratio. The reference points of foot and trunk were adjusted to the points of this model.

Segment	Mass [kg]	Length [mm]	COM/Length [%]	COM [mm]
head	7.014 (±0.173)^{CM}	250.32 (±6.30)^{R}	53.58 (±0.73)^{C}	134.12 (±3.84)
trunk	48.848 (±0.771)^{CM}	603.30 (±3.51)^{R}	48.75 (±0.18)^{H}	294.11 (±2.04)
upperArm	2.534 (±0.064)^{CM}	301.89 (±2.90)^{R}	49.10 (±0.38)^{H}	148.23 (±1.82)
forearm	1.551 (±0.042)^{CM}	278.11 (±2.70)^{R}	41.76 (±0.40)^{H}	116.14 (±1.58)
hand	0.626 (±0.020)^{CM}	92.73 (±8.69)^{R}	81.98 (±0.12)^{C}	76.02 (±7.12)
thigh	9.895 (±0.247)^{CM}	435.00 (±5.00)^{M}	40.01 (±0.26)^{H}	174.04 (±2.29)
calf	4.191 (±0.107)^{CM}	385.00 (±5.00)^{M}	41.79 (±0.26)^{H}	160.89 (±2.32)
foot	1.416 (±0.033)^{CM}	110.00 (±5.00)^{M}	46.22 (±2.80)^{C}	50.84 (±3.85)
(sagittal)		275.00 (±5.00)^{M}	55.15 (±0.44)^{C}	81.66 (±5.84)
socket	1.440 (±0.010)^{M}	495.00 (±5.00)^{M}		320.00 (±5.00)^{M}
prosthesis	2.370 (±0.010)^{M}	450.00 (±5.00)^{M}		255.00 (±5.00)^{M}
(sagittal)				15.00 (±5.00)^{M}

^{H} published by Hinrichs [45]
^{C} published by Clauser [43]
^{M} directly measured
^{R} determined based on measurements and data from Hinrichs [45] and Clauser [43], assuming equal size-ratios
^{CM} derived from Clauser [43] and the equivalent mass M^*

Wrist JC. The length of trunk and head is determined by the ratio of the amputee's height and the reported mean height.

The COM is always determined from the proximal joint of the segment. In case of the trunk, the HJC are used as reference. To derive the COM, the segment length is multiplied by the published COM/segment length ratio. Since the COM for the foot in the sagittal plane is referenced to the heel, it is adjusted to be in reference to the Ankle Joint Centre (AJC). The resulting data are summarised in Table 3.3.

To simplify calculations later on, these parameters are summarised to multi-segment parameters for head and trunk, the total arm, the total upper body, stump and socket, and foot and shoe (see Table 3.4). Mass M_{MS} and COM x_{MS} of the multi-segment are determined by

$$M_{MS} = \sum_i m_i \tag{3.1}$$

$$x_{MS} = \frac{\sum_i m_i \cdot x_i}{M_{MS}} \tag{3.2}$$

Table 3.4: Anthropometric data of multi-segments.

Segment	Mass m[kg]	Length l [mm]	COM x [mm]
trunk & head	55.862 (±0.790)		349.77 (±7.05)
total arm	4.711 (±0.080)	580.00 (±5.00)^M	304.56 (±7.48)
upper body	65.285 (±0.806)		331.83 (±8.55)
foot & shoe	1.866 (±0.034)	125.00 (±5.00)^M	
stump & socket	11.335 (±0.247)	495.00 (±5.00)^M	192.59 (±6.15)

^{M} directly measured

with m_i, the mass of the segment i, and x_i, the COM of the segment i with respect to the reference joint centre of the multi-segment. This assumes a constant positioning of the segments to each other.

3.3.2 Application

The same procedure is used to apply the model to all exercises. A global coordinate system is introduced, following the convention of the International Society of Biomechanics [37]. The proximal end of segments representing an extremity defines the positive face point. Hence, forces and moments at the proximal end are assumed to be positive in the direction of the coordinate system. Therefore, the segment of the trunk has only negative face points. All segment angles are defined by the proximal JC, the distal JC, and the vertical axis in the interval $]-180°, 180°]$. According to the coordinate system, a positive angle is measured for counterclockwise rotations. The model is depicted exemplarily for Lateral Weight Shift, visualising all measurements taken in the frontal plane, and Anterior Weight Shift, visualising all measurements taken in the sagittal plane.

To calculate the two reaction forces and the moment at the JC of the contralateral ankle (CLAJC), knee (CLKJC), hip (CLHJC), and prosthetic hip (PHJC), the model is cut at the joints into the segments. The free-body diagram of each segment is drawn which depicts all acting forces and moments, including gravitational forces and the joint reaction forces and moment. A force F which is acting on a rigid body with the line of action not passing through the analysed JC generates a moment M defined by

$$M = r \cdot F \tag{3.3}$$

with r, the perpendicular lever arm being the smallest distance between the force's line of action and the JC.

Table 3.5: Lateral Weight Shift – Measured angles, ground reaction forces at the related foot segment, and point of application with respect to the related AJC.

Segment	Angle [°]	F_z [N]	F_y [N]	z [mm]
Prosthesis	$\lambda_P = 5.7$	-32	728	22.88
Contra-lateral leg	$\lambda_{CL} = 6.0$	39	232	-14.56
Trunk	$\alpha_T = 4.3$			

For each exercise, only a static position is analysed. Since the amputee pauses in the end position of each drill, before beginning the backwards motion, the end position can be assumed to be a static situation and all segments have zero acceleration. Hence, the equilibrium conditions can be applied to the free-body diagram of each segment. The equilibrium conditions state that the sum of all forces acting on the segment and the sum of all moments equals to zero:

$$\sum F = 0 \tag{3.4}$$
$$\sum M = 0. \tag{3.5}$$

The analysis of the static position estimates the acting forces and moments only roughly as multiple times larger forces and moments act on the joints during the acceleration and deceleration phases of the movements.

Nevertheless, these equations are used to estimate all JCs. Beginning with the information from the ground reaction forces, the AJC is analysed first, followed by KJC and HJC, for both, the prosthetic and the contra-lateral leg. Many inaccuracies are accepted during the calculation. The position of the body segments is measured from pictures, the data for all body parts is determined by the anthropometric model, and the posture is assumed to be ideal, for example the physiological leg is assumed to be fully extended if appropriate. To assess the accuracy of the calculations, the joint forces and moment are determined twice for the prosthetic HJC. Once, by considering the prosthetic leg (PHJC) and once, by using the contra-lateral HJC and the trunk (PHJCC). These calculations are performed exemplarily for the prosthetic leg in the exercise Lateral Weight Shift.

Lateral Weight Shift

The model of this exercise is depicted in Figure 3.2. Table 3.5 lists the measured data from the force plate and all angles, defining the geometry determined from the photograph. To calculate the forces and moment in the PHJC, the free-body

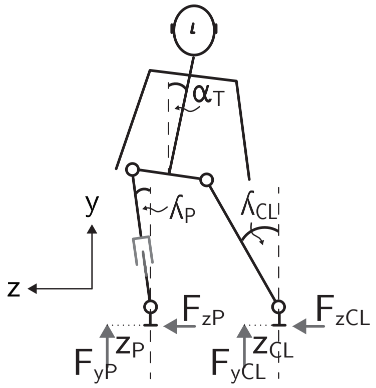

Figure 3.2: Measured angles and ground reaction forces for biomechanical analysis in the frontal plane, depicting exemplarily the model for Lateral Weight Shift.

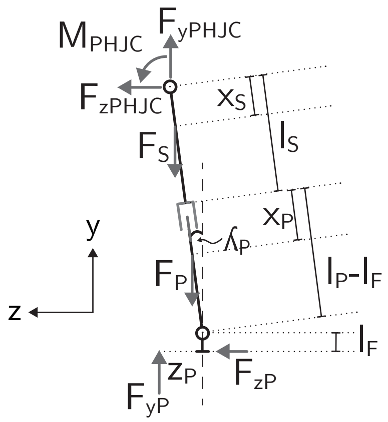

Figure 3.3: Free-body diagram of the prosthetic leg to determine joint reaction forces at PHJC.

diagram of the prosthetic leg is drawn (see Figure 3.3).

Assuming the static analysis to be adequate, Equation 3.4 is applied twice to calculate the joint reaction forces F_y and F_z at the end position as

$$F_{yPHJC} = F_S + F_P - F_{yP} \quad \text{and} \tag{3.6}$$

$$F_{zPHJC} = -F_{zP}. \tag{3.7}$$

Table 3.6: Lateral Weight Shift – Calculated joint reactions.

Joint Centre	F_y [N]	F_z [N]	M [Nm]
PHJC	-594	32	-66
CLAJC	-213	-39	1
CLKJC	-172	-39	8
CLHJC	-75	-39	19
PHJCC	-565	39	-56

Using Equation 3.3 for all acting forces and Equation 3.5, the moment generated in the PHJC can be determined by:

$$\begin{aligned} M_{PHJC} = {} & F_S \cdot x_S \cdot sin(\lambda_P) + \ldots \\ & + F_P \cdot (x_P + l_S) \cdot sin(\lambda_P) + \ldots \\ & + F_{yP} \cdot [z_P - (l_S + l_P - l_F) \cdot sin(\lambda_P)] + \ldots \\ & + F_{zP} \cdot [l_F + (l_S + l_P - l_F) \cdot cos(\lambda_P)] \,. \end{aligned} \tag{3.8}$$

After calculating all quantities for the PHJC, considering the prosthetic leg, all JC shall be determined starting from the contra-lateral ankle. All reaction forces are calculated in analogy, beginning with the ankle and moving upwards segment by segment. The MATLAB®-scripts containing all calculations are presented in Appendix B.

The calculated forces and moments are summarised in Table 3.6. Since the amputee shifted his weight towards the prosthesis, the vertical force in the PHJC is approximately eight times larger than the one in the CLHJC. The calculated muscle moments determine abduction motion in both hips and pronation in the CLAJC. The amputee stands with legs brace. Hence, both feet are pushing outwards, which is in accordance with the abduction in both hips. To stabilise the ankle and keep the foot flat on the ground, the amputee pronates his foot. The determined muscle activities are only applicable for the end position, when the amputee stabilises himself. For the dynamic exercise execution, one hip needs to be abducting and the other adducting to accelerate and decelerate the pelvis sideways.

When comparing the values determined by the prosthetic leg (PHJC) and the control values determined by the contra-lateral leg (PHJCC), an error can be determined in all three joint reactions. Adjusting the difference in F_y reduces the error in moment, whereas a correction of F_z worsens the error. Nevertheless, the model can be used to determine the direction of joint moments and to approximate their absolute value.

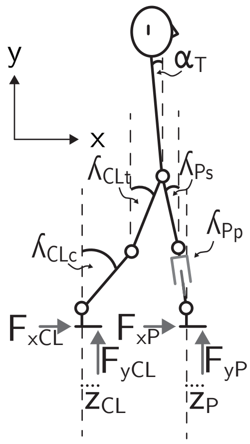

Figure 3.4: Measured angles and ground reaction forces for biomechanical analysis in the sagittal plane, depicting exemplarily the model for Anterior Weight Shift.

Anterior Weight Shift

The model of this exercise is depicted in Figure 3.4 to visualise exemplarily all measurements (see Table 3.7) taken in the sagittal plane. The resulting joint reactions are listed in Table 3.8. The error in F_y equals 79 N, in F_x 16 N, and in the moment 38 Nm with different signs. Taking into account the errors in ground reaction forces, the error in the prosthetic hip moment can be reduced. Nevertheless, the approximation in the prosthetic hip joint centre cannot be used as the absolute value of the moment is reduced and the sign remains opposite. The error might be caused by a slight deviation in posture between the two pictures. Furthermore, the amputee slightly rotated his hip, violating the assumption of movement in a single plane.

The muscle moment determines plantar flexion in the contra-lateral ankle, extension in the knee, and flexion in the hip. Physiologically, one would expect an extension moment in the contra-lateral hip to push the pelvis forwards. Since the amputee cannot increase balance at the knee and ankle at the prosthetic leg, he stands with legs braced, resulting in altered muscle activity.

Criss Cross

The measurements for all three positions are listed in Table 3.9, resulting in the forces and moment summarised in Table 3.10.

Table 3.7: Anterior Weight Shift – Measured angles, ground reaction forces at the related foot segment, and point of application with respect to the related AJC.

Segment	Angle [°]	F_x [N]	F_y [N]	x [mm]
Prosthesis	λ_{Pp} = 3.8	-40	522	20.76
Stump & socket	λ_{Ps} = 1.2			
Contra-lateral calf	λ_{CLc} = -19.4	24	331	48.63
Contra-lateral thigh	λ_{CLt} = -13.6			
Trunk	α_T = 0.1			

Table 3.8: Anterior Weight Shift – Calculated joint reactions.

Joint Centre	F_y [N]	F_x [N]	M [Nm]
PHJC	-387	40	13
CLAJC	-312	-24	-18
CLKJC	-271	-24	12
CLHJC	-174	-24	25
PHJCC	-466	24	-25

Table 3.9: Criss Cross – Measured angles, ground reaction forces at the related foot segment, and point of application with respect to the related AJC.

(a) front

Segment	Angle [°]	F_x [N]	F_y [N]	x [mm]
Prosthesis	λ_{Pp} = -17.4	24	410	85.85
Stump & socket	λ_{Ps} = -16.1			
Contra-lateral calf	λ_{CLc} = -1.8	-18	570	31.34
Contra-lateral thigh	λ_{CLt} = 20.3			
Trunk	α_T = -11.5			

(b) side

Segment	Angle [°]	F_z [N]	F_y [N]	z [mm]
Prosthesis	λ_P = -9.8	-72	486	3.95
Contra-lateral leg	λ_{CL} = 10.8	78	482	-10.34
Trunk	α_T = -1.9			

(c) back

Segment	Angle [°]	F_x [N]	F_y [N]	x [mm]
Prosthesis	λ_{Pp} = 2.4	-40	525	26.58
Stump & socket	λ_{Ps} = 7.4			
Contra-lateral calf	λ_{CLc} = -19.9	60	393	27.30
Contra-lateral thigh	λ_{CLt} = -20.1			
Trunk	α_T = 1.7			

Table 3.10: Criss Cross – Calculated joint reactions.

(a) front

Joint Centre	F_y [N]	F_z [N]	M [Nm]
PHJC	-276	-24	23
CLAJC	-551	18	-14
CLKJC	-510	18	-1
CLHJC	-413	18	-65
PHJCC	-227	-18	23

(b) side

Joint Centre	F_y [N]	F_z [N]	M [Nm]
PHJC	-351	72	-8
CLAJC	-464	-78	5
CLKJC	-423	-78	2
CLHJC	-326	-78	4
PHJCC	-314	78	-10

(c) back

Joint Centre	F_y [N]	F_z [N]	M [Nm]
PHJC	-391	40	-11
CLAJC	-375	-60	-17
CLKJC	-334	-60	8
CLHJC	-237	-60	28
PHJCC	-404	60	-22

Although two separate measurements had to be made for each stepping direction, the results seem promising. Analysing the step to the front, the error in the forces is limited to 49 N in the y-direction and 6 N in the x-direction. The moments calculated for the PHJC coincide. The calculated moments suggest a flexion motion by the muscles in the prosthetic hip, an extension in the contra-lateral hip, and a plantar flexion in the ankle. The moment determined for the knee is almost zero.

For a step to the side, the error of the force in the y-direction measures 37 N and in z-direction 6 N. The moments at the PHJC coincide as well. During the side-step, the amputee distributed his weight nearly equally. The generated muscle moments are smallest for the side-step. The direction of the muscle activity is equal to the one in Lateral Weight Shift, defined by abduction in both hips and pronation in the ankle.

When stepping backwards, the error in F_y equals 13 N and in F_x 20 N. The moment calculated at the prosthetic hip has a difference of 10 Nm between the two ways of calculation. A slight difference in body position between the two measurements can be determined from the photographs. Additionally, the error in the horizontal ground reaction F_x is larger than for the other two directions. Hence, the measurements taken for this static position do not fit together. Nevertheless, the sign of the moments coincides with the ones determined for Anterior Weight

Table 3.11: Prosthetic Abduction – Measured angles, ground reaction forces at the related foot segment, and point of application with respect to the related AJC.

Segment	Angle [°]	F_z [N]	F_y [N]	z [mm]
Prosthesis	λ_P = -33.5			
Contra-lateral leg	λ_{CL} = -3.2	-1	961	-6.49
Trunk	α_T = -19.2			

Table 3.12: Prosthetic Abduction – Calculated joint reactions.

Joint Centre	F_y [N]	F_z [N]	M [Nm]
PHJC	134	0	-21
CLAJC	-943	1	-6
CLKJC	-902	1	13
CLHJC	-805	1	34
PHJCC	164	-1	44

Shift and the deviation of 10 Nm in the approximation is accepted. The calculations describe an extension in the prosthetic hip, a flexion in the contra-lateral hip, an extension in the knee, and a plantar flexion in the ankle. Apart from the ankle joint activity, these movements are opposite to the front position, where the legs switch position.

Physiologically surprising hip moments can be explained by compensatory movements of the amputee. As in Lateral Weight Shift and Anterior Weight Shift, the amputee stands with legs braced to maintain balanced in the static position of the exercise.

Prosthetic Abduction

The parameters determining ground reaction force and body position are stated in Table 3.11. The resulting values for the joint reaction forces are presented in Table 3.12. The error in F_y equals 30 N, F_z can be assumed to be correct. Since the prosthetic leg is lifted for this exercise, no weight is carried by this leg. Hence, the resulting joint force F_y at the PHJC has a positive sign, representing tension instead of pressure. Comparison of the two moments for PHJC, suggests a strong violation of a modelling assumption. The analysis of the photo yields that the amputee did not perform the exercise in the frontal plane. The right hip is rotated backwards and the trunk is slightly tilted forwards. If the amputee were to correct these mistakes, the angle of the contra-lateral leg would decrease. This adjustment does

Table 3.13: Squat – Measured angles, ground reaction forces at the related foot segment, and point of application with respect to the related AJC.

Segment	Angle [°]	F_x [N]	F_y [N]	x [mm]
Prosthesis	λ_{Pp} = -2.0	-35	281	74.44
Stump & socket	λ_{Ps} = 34.3			
Contra-lateral calf	λ_{CLc} = -15.2	43	613	-6.97
Contra-lateral thigh	λ_{CLt} = 43.4			
Trunk	α_T = -63.9			

Table 3.14: Squat – Calculated joint reactions.

Joint Centre	F_y [N]	F_x [N]	M [Nm]
PHJC	-146	35	-47
CLAJC	-595	-43	0
CLKJC	-554	-43	43
CLHJC	-457	-43	-125
PHJCC	-184	43	-103

not influence CLAJC, it increases the moment determined for CLKJC and CLHJC, and reduces the moment of PHJCC to the value of PHJC.

The calculated moments suggest, that the muscles abduct at both hips to stabilise the pelvis and lift the prosthetic leg. Furthermore, the contra-lateral foot supinates to maintain stable in the ankle.

Squat

The measured parameters are summarised in Table 3.13 and the resulting joint reactions are listed in Table 3.14. The difference between the two calculations of PHJC equals 37 N in F_y, 8 N in F_x, and 56 Nm in M. The load is carried by both legs, resulting in only negative joint reaction forces in the y-direction. It can be concluded from the difference in F_y between the two HJC that the amputee squatted mainly using his contra-lateral leg. Furthermore, the horizontal ground reaction force shows that he pushed his prosthetic leg forwards into the ground and pulled his contra-lateral leg back, creating a moment around the y-axis which he tried to compensate at the hip. Since he could not fully compensate, the HJCs are not at the same location in the sagittal plane, violating the model's assumptions. Considering this reduces the difference in the moment at the PHJC.

Furthermore, the amputee is not able to maintain a straight back and to keep his shoulders pulled back. Slight deviations from good form change the mass distribution and due to the large lever arm of trunk and arms during this exercise the respective moment is heavily influencing the moments at the HJC.

The muscles perform an extension at the CLKJC, CLHJC, and PHJC. No moment is acting on the CLAJC. Since the amputee extended at the CLKJC which is technically not possible with a passive prosthetic knee, he could not perform the exercise with symmetric muscle activation which explains the difficulty in maintaining symmetric weight distribution and the opposite sign in the measured horizontal forces.

4. Ruleset Library

In the previous chapters finite state control and the hardware components of the prosthesis were introduced, as well as the biomechanical background of the implemented exercises. This information constitutes the ruleset library's foundation. To assess the exercise execution using the prosthesis' sensor signals a thorough analysis of these signals is necessary for each exercise. However, correct execution of the exercise is not always reflected by distinctive features of these signals. This lack of feature representation by the signals, as well as software limitations, restricts the development of an efficient ruleset.

In this chapter the exercises' implementations are presented in detail. Due to the many interdependencies throughout the rulesets, the general library structure is described first. Additionally, specifications and limitations that concern every exercise are discussed. Afterwards, each of the five exercises is discussed individually. Firstly, the sensor signals representing the specific motion are described and their implications on the ruleset structure are discussed. Secondly, the ruleset's requirements are stated and the general flow of the exercise is illustrated in detail. Finally, the intensity levels and their realisation are introduced.

4.1 Library Structure

Each exercise is implemented in a separate mode, monitoring and evaluating the execution of the current drill. At the beginning of the specific exercise an additional ruleset (see REFERENCE in Section 4.2.1) is loaded to quantify the user's normal standing posture. Thereby a general reference position is defined equivalent for all exercises but specific to the user's alignment and present posture. Moreover, two additional rulesets are loaded within REFERENCE. They run in parallel throughout the exercise. One controls the damping values of the prosthesis (DAMPING) with respect to the selected activity mode (see Section 2.2.1) and the drill executed. The other enables auditive feedback, using the same tones in each exercise (see TONES in Section 4.2.2).

4.2 Rulesets

In this Section, the rulesets will be described. First, the two general rulesets REFERENCE and TONES are introduced. Second, general specifications and

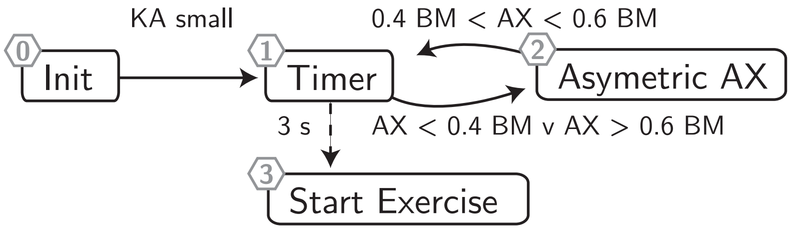

Figure 4.1: Reference – Structure of the ruleset.

procedures for defining levels of intensity are discussed. Afterwards, each exercise ruleset is derived by the sensor signals, the specifications, the resulting structure, and the design of levels of intensity. To review the exercise instructions see Section 3.2.

4.2.1 Ruleset REFERENCE

The structure of REFERENCE is illustrated in Figure 4.1. First, the DAMPING ruleset is loaded with the rule Init which uses the same control algorithm as the basic mode of the Kenevo. Additionally, the TONES ruleset is loaded and initialised in a silent mode. Furthermore, all global variables and all counters are set to zero. Upon attaining a small knee angle, the ruleset transitions from Init (0) to Timer (1). This allows users to choose an exercise while seated but the exercise and the reference only start once they are standing, assuming that amputees do not fully extend their prosthetic leg while sitting.

Whenever the measured axial load exceeds an allowed range ($0.5 \cdot \mathrm{BM} \pm 0.1\mathrm{BM}$), the ruleset switches to Asymmetric AX (2). A tone, described in the next Subsection, indicates the asymmetric weight and is replayed every second for as long as the rule is active. Once the weight is again distributed symmetrically, the ruleset transitions back to Timer. Whenever Timer is entered, a timer is set to 3 s. During these three seconds, roll, pitch, yaw, and leg angle are averaged and saved as global variables, making them available in Init of the current exercise. If the timer elapses without the weight being distributed unequally, the ruleset jumps to rule Start Exercise (3). A global variable is set, indicating the end of the setup. Hence, the current exercise ruleset leaves the rule Init and users can start to execute the drill.

Table 4.1: Definition of the available feedback tones by tone sequence and duration of sound and pause.

Name	Tone Sequence	Duration [s]
Timer	d^4 - d^4 - d^4 - a^4	0.1 $\underline{0.9}$ 0.1 $\underline{0.9}$ 0.1 $\underline{0.9}$ 0.3
Correct	d^4 - fis^4	0.1 $\underline{0.1}$ 0.1
Overall correct	d^4 - fis^4 - a^4	0.1 $\underline{0.1}$ 0.1 $\underline{0.1}$ 0.1
Wrong	d^4 - a^3	0.2 $\underline{0.1}$ 0.2
Wrong form	as^2 - as^2	0.2 $\underline{0.1}$ 0.2
End	a^4 - fis^4 - d^4 - h^4	0.1 $\underline{0.1}$ 0.1 $\underline{0.1}$ 0.1 $\underline{0.1}$ 0.1

4.2.2 Ruleset Tones

Different tones are used to facilitate the testing of the ruleset. The tones available are described as tone sequences using German notation and duration which need to be assigned to both sounds and pauses (see Table 4.1). "Timer" is used in the Reference ruleset to signal the duration of the initialisation phase and the beginning of the exercise. "Wrong form" is audible in Reference to draw attention to asymmetrically distributed weight. Furthermore, it is used in Prosthetic Abduction and Squat, as long as the exercise is not executed with good form. "Correct" is played, whenever a repetition is performed fulfilling all criteria, otherwise "Wrong" is played. "Overall correct" is only utilised in Criss Cross, when a sequence of all three positions is correctly performed. "End" signals the end of a set.

4.2.3 Ruleset Damping

The Damping ruleset consists of three modes. One mode controls the damping equivalent to the official damping of the Kenevo, the second one locks the extended prosthesis, and the third is designed specifically for the exercise Squat. The mode for the Squat will be specified in the Subsection of Squat. Generally, all exercises are executed in the normal mode if not specified otherwise.

4.2.4 General Remarks on Exercise Rulesets

Specification

Some specifications are applied to all exercise rulesets. To reduce redundancy, they are presented here and not repeated in the specifications of each exercise.

The ruleset represents the progression of the exercise and assesses the performance of each repetition. The level of intensity shall be adjustable before beginning the exercise by selecting a slider-value between one and ten. The observed sensor signals are defined in the exercise-specific requirements. For each of these signals, a necessary ROM is defined for the selected level of intensity. To perform a repetition correctly, the signal needs to exceed its defined ROM. For each exercise, the levels of intensity are defined by thresholds for all ROMs assessed. Feedback counters represent the performance and are implemented as global variables which are incremented in the respective rule and sent to the computer for post-processing.

Real-time audio-feedback evaluating the current execution shall be available. The volume has to be adjustable including a silent mode. In Lateral Weight Shift and Anterior Weight Shift, each repetition can be divided into two phases. The loading phase refers to the weight shift onto the prosthesis and the unloading phase consists of the counter movement, when the weight is shifted back onto the contra-lateral leg. In these two exercises, tones shall only be implemented for the loading phase.

A set consists of a predefined number of repetitions which have to be performed in a row. This number of repetitions shall be adjustable by the user with a slider. A tone signals the end of each set. Afterwards, the feedback counters of all loading (and if available unloading) phases are evaluated collectively.

Small deviations in the observed sensor signals shall not count as mistakes or terminate repetitions early. At the same time, the program shall detect repetitions of users with restricted mobility, hence reduced ROM. This requires a quantitative limit, differentiating an unintended deviation from an intended exercise progress.

Levels of intensity

One aim of this work is to suggest a first categorisation of the intensity levels. To simplify the design process of the intensity levels, target values are defined and generally interpolated linearly. Hence, for each signal the minimally and maximally requested ROM needs to be defined. Thus, the requirements for the first and tenth intensity level were determined with the aid of an experienced amputee in an intermediate usability test.

For some exercises, the specifications require different weighting of the sensor signals for changing intensity levels. Therefore, not all ROMs are equally important and some ROMs are only increased for higher intensities. That is, a level needs to

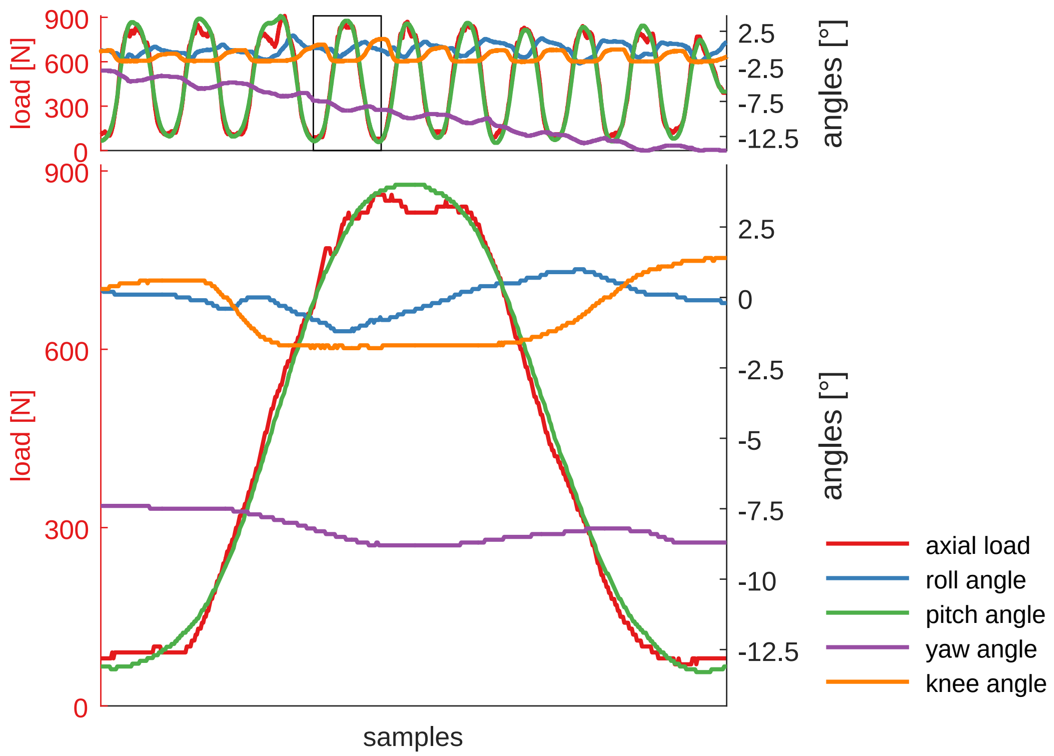

Figure 4.2: Lateral Weight Shift – Characteristic sensor signals for correct execution. The top panel depicts a whole set. The bottom panel enlarges the marked single repetition.

be chosen for each ROM after which the ROM will be increased with constant rate for successive levels. Should the rate change, the targeted ROM for an intermediate intensity level was determined in consultation with the physiotherapist.

4.2.5 Ruleset Lateral Weight Shift

Sensor signals

The sensor signals of a correct repetition (see Figure 4.2) show the following characteristics: Axial load and pitch angle increase tremendously while the weight is shifted onto the prosthesis. The measured knee angle increases slightly, while putting weight onto the contra-lateral leg. Whenever the signal is below zero, the prosthesis is extended into the extension stop.

The roll angle decreases and yaw angle increases while weight is shifted onto the prosthesis. Additionally, the yaw angle drifts due to technical limitations over the course of all repetitions. Hence, knee angle, roll angle, and yaw angle do not add information, as they are in correlation with axial load and pitch angle and their amplitude is comparatively small.

Specification

Taking into account these characteristics, the requirements for the ruleset are specified as follows: The axial load and the pitch angle are monitored to evaluate the weight on the prosthesis and the lateral movement of the pelvis respectively. For both observed phases, loading and unloading, the software differentiates between correct, insufficient weight, insufficient lateral pelvis movement, and insufficient weight and lateral movement. After each completed repetition, the symmetry of the weight shift is evaluated. In consultation with the physiotherapist it was decided to equally weigh the importance of all these feedback variables.

Repetitions shall be detected by means of changing pitch angle for lower levels of intensity, without evaluating the pitch angle during classification of performance. For these low intensity levels it has to be assumed that the weight on the prosthesis is reduced through excessive support by the upper extremity.

Structure

The structure for implementing LATERAL WEIGHT SHIFT is shown in Figure 4.3. Axial load and pitch angle are evaluated in real-time in order to monitor and classify the current progression. Hence, they are assessed in the logic expressions of the conditions, determining the transitions to the next state of the FSM.

Each ruleset has a start rule called Init where parameters and other necessary rulesets are prepared. In Init (0) of LATERAL WEIGHT SHIFT, the ruleset REFERENCE is activated (see Section 4.2.1) and all global variables specifying the requested ROM are set according to the intensity slider which is discussed later in this Section. After a successful setup, the exercise starts by putting weight on the prosthesis. Each repetition begins with the rule Loading Basis (1) which assumes weight to be predominantly on the contra-lateral leg. The minimum axial load and pitch angle during unloading is stored in two global variables, $@AX_{MIN}$ and $@PA_{MIN}$, respectively. The sum of these global variables and their corresponding ROMs defines the threshold for correct execution. The conditions, defining a transition in the FSM, assess if the sensor signals reach their threshold. Since the first repetition begins with equally distributed weight due to the initialisation phase, the reference variables $@AX_{MIN}$ and $@PA_{MIN}$ are initialised during the setup, such that only half of the ROM needs to be performed.

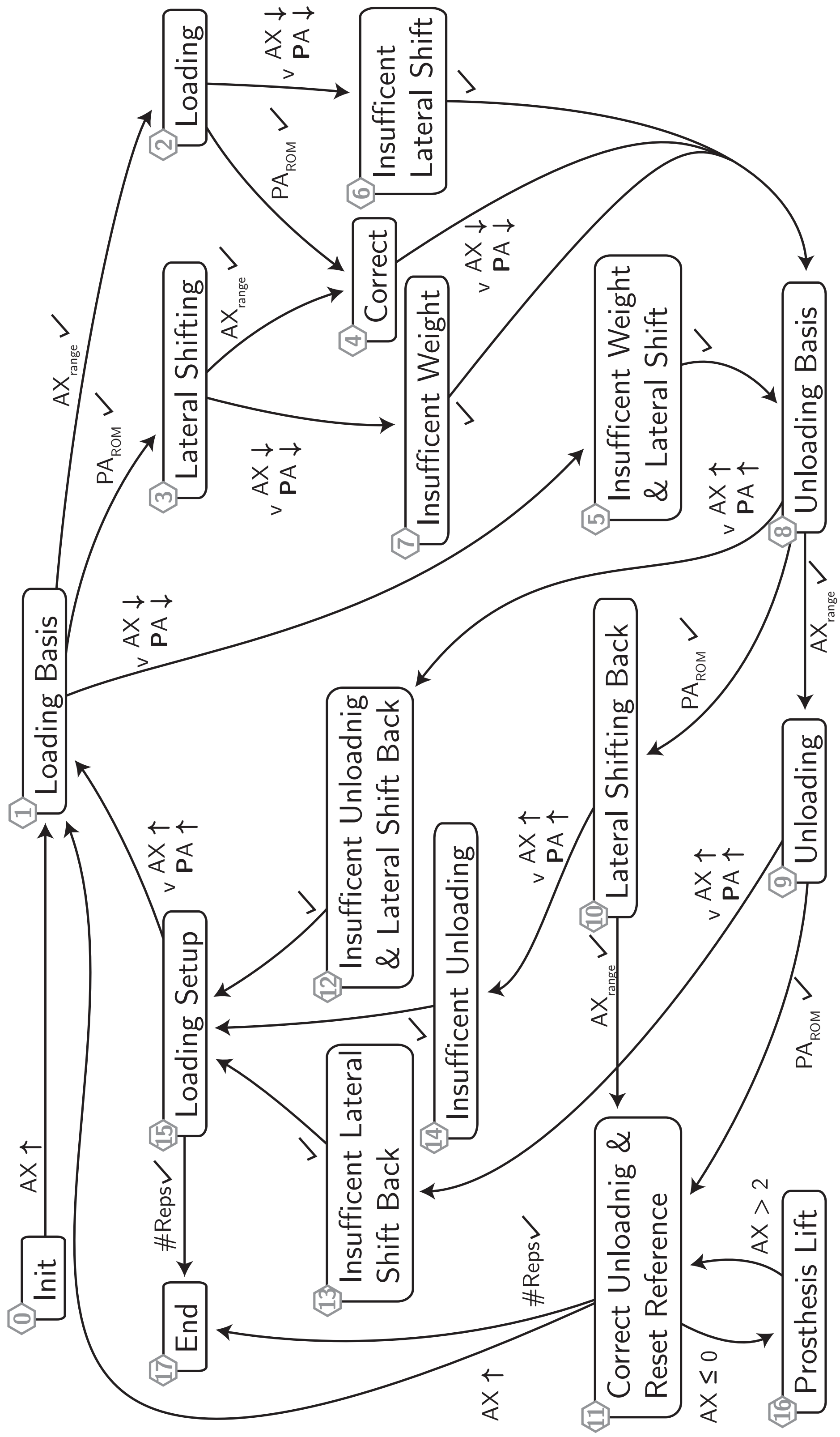

Figure 4.3: Lateral Weight Shift – Structure of the ruleset.

To enable real-time audio feedback, the conditioned transition determines when the rule is changed and the tone is played in the new active rule. Since tones shall differentiate between the four scenarios of the loading phase, a separate rule (4-7) is necessary for each. Furthermore, the ruleset transitions at the first time step at which a condition is fulfilled. In terms of combinatorial logic, the order of events is determined by the ruleset. Since it is not requested that either axial load or pitch angle reach their respective threshold first, the order of occurrence shall not affect the classification. Hence, two additional rules (2 and 3) are necessary to allow for all possible combinations. Overall, this results in the following structure and behaviour of the ruleset: If the axial load reaches its requested amount, the rule Loading (2) becomes active. Accordingly, if the requested pitch angle is reached first, the rule Lateral Shifting (3) becomes active. From these two rules Correct (4) is reached when the other signal attains its threshold. In these four rules (1-4), the maximum values of axial load and pitch angle are stored as global variables (@AX_{MAX}, @PA_{MAX}). A signal is classified as decreasing when its amplitude falls below a certain percentage of the maximum value. When either of the two signals decrease, the ruleset detects the progress and switches to the next rule: from Loading Basis (1) to Insufficient Weight & Lateral Shift (5), from Loading (2) to Insufficient Lateral Shift (6), from Lateral Shifting to Insufficient Weight (7), and from Correct (4) to Unloading Basis (8). Rules 5-7 are evaluated once, before the ruleset transitions to Unloading Basis (8).

In case of correct loading, the mean values of this repetition's @AX_{MAX} and the previous repetition's @AX_{MIN} is calculated. This mean value has to be within a predefined range around half the body weight. Otherwise, the respective feedback variable is incremented.

To emphasise the equivalent assessment of the unloading phase, the same structure is used there, represented by the rules Unloading (9), Lateral Shift Back (10), Correct Unloading & Reset Reference (11), Insufficient Unloading & Lateral Shift (12), Insufficient Lateral Shift (13), Insufficient Unloading (14), and Loading Setup (15). If memory shortage becomes an issue at a later point, the reference values can be logged within one rule. They are then evaluated on exit when the next repetition begins, reducing the total number of rules in the unloading structure.

Finally, in Correct Unloading & Reset Reference, the reference value of the pitch angle is updated. Therefore, @PA_{MIN} is stored as long as the prosthesis touches the ground. When users lift the prosthesis, the axial load vanishes and rule Prosthesis Lifted (16) is activated. Once weight is again put on the prosthesis, the ruleset transitions back to rule 11. On exit, when the next repetition starts, @PA_{MIN} is stored in the reference variable @PA_{REF}. With the detection of an increase in axial load, a

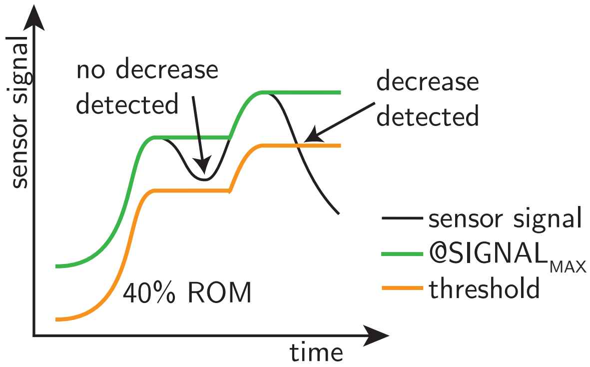

Figure 4.4: Implementation to detect a signal de- or increase. Only deviations of more than 40 % ROM define a falling or rising signal.

transition to Loading Basis (1) takes place. If the unloading movement is poorly executed and Loading Setup (15) is active, a further increase in axial load or pitch angle is necessary before transitioning to Loading Basis (1) in order to guarantee rising signals. When amputees execute the last repetition, End (17) is activated from either Correct Unloading & Reset Reference (11) or Loading Setup (15).

If a threshold value was not reached during loading or unloading, the corresponding reference variable is set to the threshold value. This reset allows users to continue the exercise correctly after a mistake. Conversely, the pitch angle reference is not reset but kept from the preceding repetition since it is stored in @PA$_{REF}$.

The specifications require that small deviations from the main trend shall not result in an early termination of the repetition. To facilitate this feature, the allowed range for deviations is limited to 40 % of the signal's ROM. An example signal is depicted in Figure 4.4. After an initial rise, the signal decreases slightly but does not drop below the limit before rising again. Afterwards, the signal reaches its global maximum and decreases by more than 40 % of ROM. Hence, the current phase is terminated, that is, users continue with the exercise. This definition is used whenever a change in the trend of the signal needs to be detected, indicated as an upwards or downwards arrow in the rulesets' schematic. In an intermediate usability test, it could be shown that the best differentiation between unintended deviation and intended proceeding is achieved with a value of 40 %. The quality of differentiation is assessed by the reproduction of the total number of repetitions counted by the ruleset compared to the user's count.

Levels of intensity

The different ROMs for all signals assessed are presented in Figure 4.5. During the first three intensity levels, the pitch angle is not assessed for classification of the execution. This allows amputees to focus on the correct loading of the prosthesis. Within these three levels, the necessary AX_{range} is increased from 40 % bodymass (BM) up to 60 %. The threshold begins at 40 % as inexperienced users usually have difficulties trusting the prosthesis and distributing their weight equally. Nevertheless, as loading of the prosthesis is the main goal of this exercise, the threshold is lifted up to 60 % BM in two steps. Since PA_{ROM} is 7.5°, changes of 3° in pitch angle terminate the current loading or unloading phase.

From level 4 to level 10 the required AX_{range} is linearly increased from 60 % to 80 % and PA_{ROM} is increased from 7.5° to 10°. During normal executions the pitch angle reaches even larger values, but the required PA_{ROM} is limited to 10° to prevent users from pure and excessive pelvis motion with the trunk flexed in the opposite direction. The AX_{range} is increased in smaller steps compared to the first levels since increasing the load on the prosthesis becomes tougher with each level.

The allowed range for the mean value of minimal and maximal axial load is continually decreased from 20 % BM down to 15 %. The amputee with whom the thresholds were tested was very experienced and had an advanced body awareness. This awareness made it difficult to determine appropriate boundaries for this ROM since the amputee never made significant errors even while imitating beginners.

4.2.6 Ruleset ANTERIOR WEIGHT SHIFT

Sensor signals

The sensor signals measured during a repetition of Anterior Weight Shift are plotted in Figure 4.6. Axial load rapidly increases and after a short plateau rapidly decreases. The pitch angle correlates with the axial load and increases while weight is shifted onto the prosthesis. This increase represents the lateral weight shift necessary to maintain balance.

The knee angle rises to a maximum when users start to shift their weight anteriorly. Afterwards, it drops below zero which means that the knee is extended into the extension stop. When the weight is shifted back again, the knee angle reaches another maximum before reducing to a small knee angle. The roll angle decreases while weight is shifted towards the prosthesis. Its signal shows two bumps, one during rising axial load when the knee angle moves into the extension

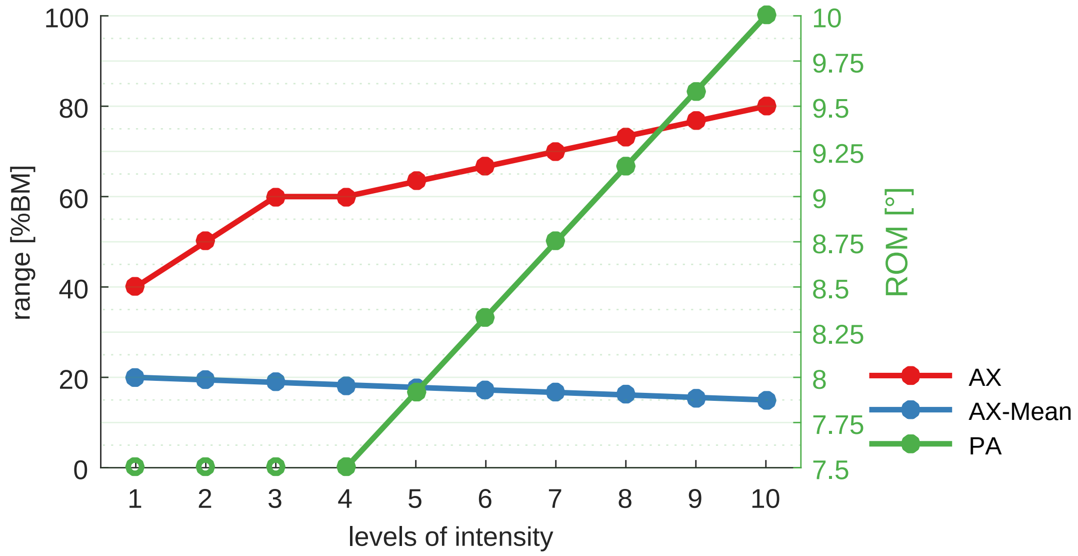

Figure 4.5: Lateral Weight Shift – Levels of intensity. At intensity levels one to three, the pitch angle is evaluated to determine the ruleset's progress, but does not influence the classification.

stop, the other one during decreasing axial load. It is difficult to evaluate the performance using only roll angle or pitch angle as the hip motion needs to be determined and assessed. Therefore, knee angle and roll angle are combined into one signal, called leg angle (LA) and defined by LA = KA + @LR · RA, where @LR is the length-ratio between thigh and calf. The leg angle is the angle between hip, foot, and the vertical line through the foot, hence, representing horizontal hip movement. It decreases when the axial load increases and vice versa. Moreover, the signal of the yaw angle has no significant characteristics.

Specification

With these signal characteristics in mind, the following specifications are defined: The axial load and the leg angle are monitored. For the loading and the unloading phase of the prosthesis a difference is made between sufficient and insufficient weight on the prosthesis. In case of correct weight, the anterior motion is assessed additionally. All three situations are logged in their own counter. Furthermore, the number of correct loadings and correct unloadings within one repetition is counted, giving a total of seven feedback counters.

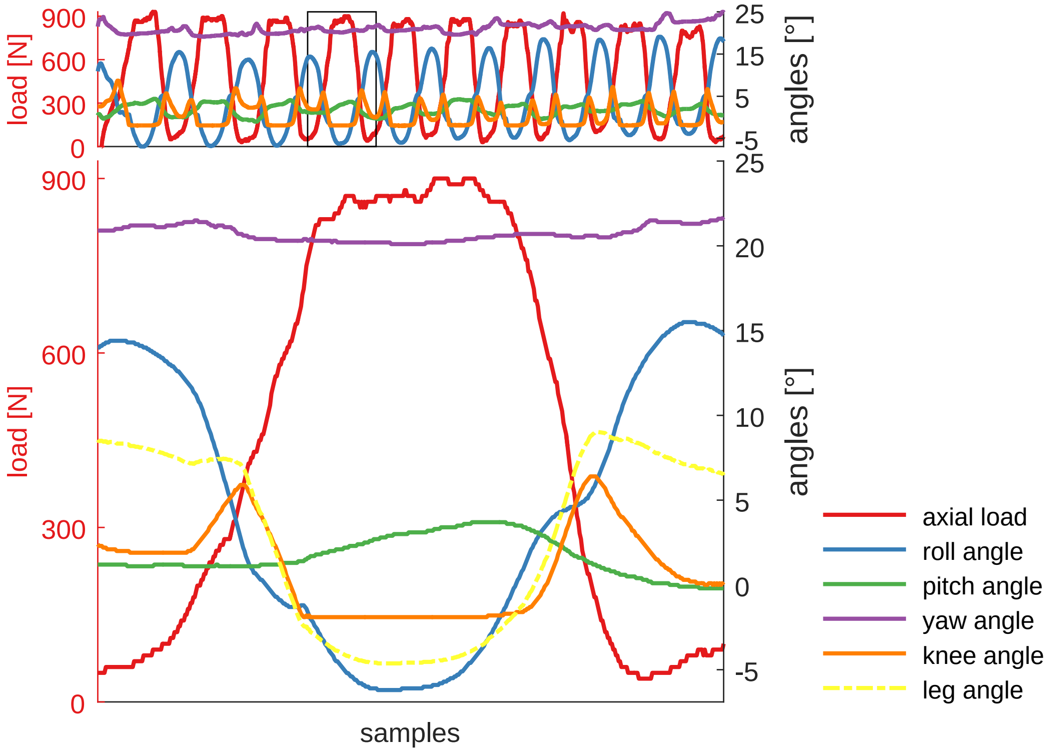

Figure 4.6: Anterior Weight Shift – Characteristic sensor signals for correct execution. The top panel depicts a whole set. The bottom panel enlarges the marked single repetition and additionally shows the leg angle.

Structure

The resulting structure is presented in Figure 4.7. The initial rule is Init (0), where the setup is started and the ROM is defined according to the intensity level. Furthermore, the reference variable of the leg angle (@LA_{REF}) is set. Since the initialisation setup is always executed with parallel feet, the set is started by positioning the prosthetic foot in front and beginning to shift weight onto it. Thereby, the ruleset transitions into rule Loading Basis (1). First, the load on the prosthesis is rated. When users increase their weight on the prosthesis, reaching the required AX_{range}, Loading (2) is activated. Second, the anterior hip motion is evaluated until amputees reduce their weight on the prosthesis. If LA_{ROM} is exceeded, Correct (3) is entered and the respective feedback counter is increased. Furthermore, if audible feedback is switched on, a "correct" tone is played. When users reduce the weight on the prosthesis and start to move backwards, the repetition is terminated. Hence, after a correct execution Unloading Basis (6) is

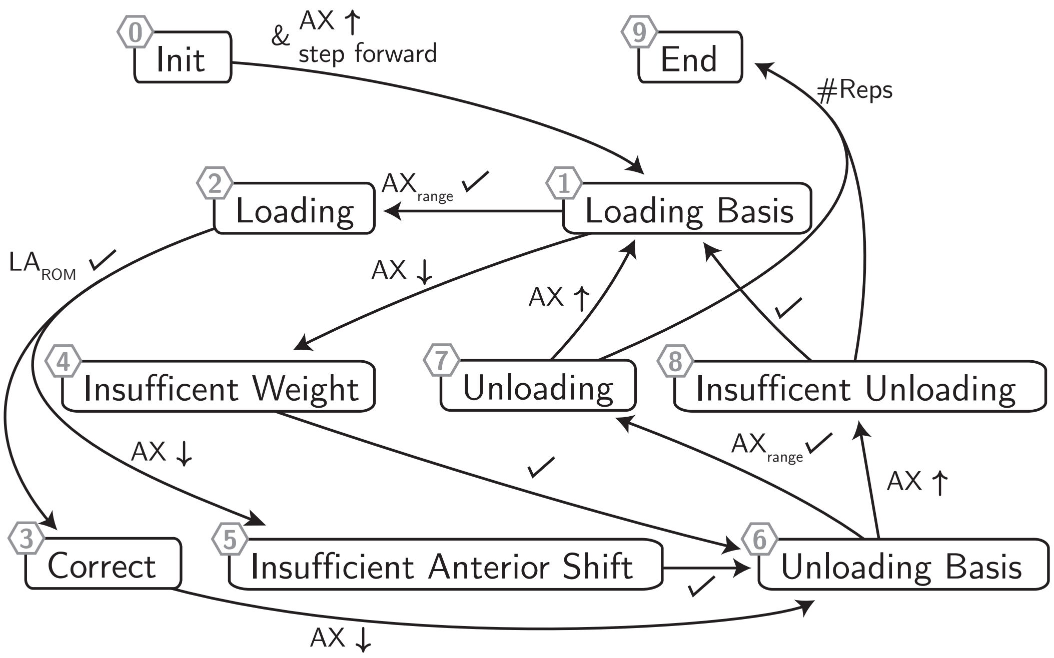

Figure 4.7: Anterior Weight Shift – Structure of the ruleset.

reached. If the load was sufficient but the anterior hip movement was insufficient, rule Insufficient Anterior Shift (5) is activated, whereas rule Insufficient Weight (4) is activated if the axial load never reached the threshold. Within both rules (4 and 5) a chime signals erroneous execution and the corresponding feedback counter is incremented before transitioning to Unloading Basis (6). Additionally, the reference for axial load @AX$_{MAX}$ is set to the threshold value in rule 4 to enable a correct unloading phase. Since @LA$_{REF}$ is set during initialisation and never changed, it needs not to be reset if the threshold is not exceeded.

The unloading movement is assessed equivalently. Since no real-time tones are requested, rules 2, 3, and 5 can be combined into Unloading (7) and rule 4 is represented by Insufficient Unloading (8). Once the number of requested repetitions is reached, the ruleset transitions to rule End (9) from either rule 7 or 8, depending on the execution of the last repetition.

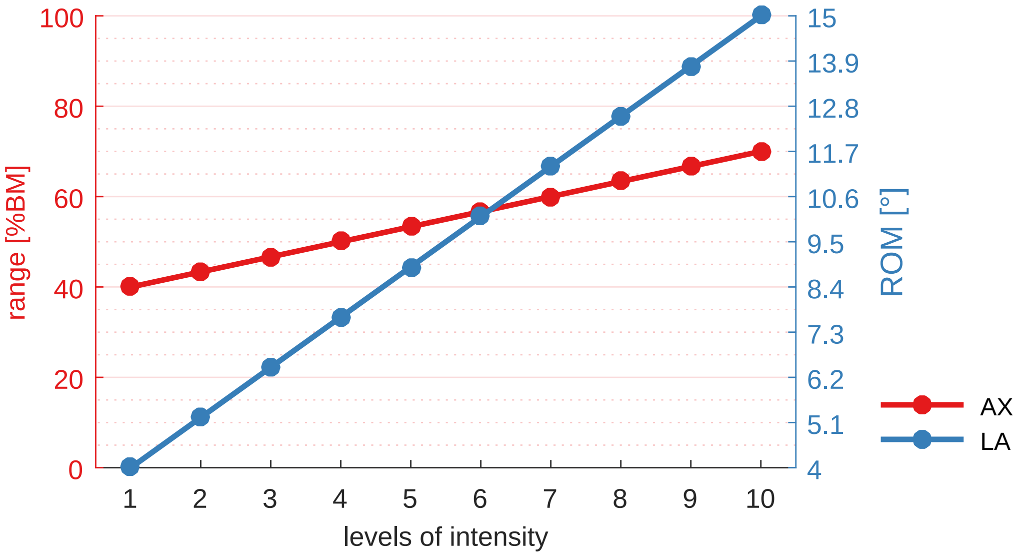

Figure 4.8: Anterior Weight Shift – Levels of intensity.

Levels of intensity

The levels of intensity are summarised in Figure 4.8. Boundary thresholds of 40 % BM and 70 % BM of axial load were determined for this exercise. Compared to the previous drill the maximal axial load is reduced since amputees cannot put more weight on the prosthesis without making a step to the front. Since the amputee performing the test had an advanced skill level, it was difficult to determine the leg angle threshold for level one. It was decided to increase both thresholds simultaneously and therefore a small LA_{ROM} of 4° is required for low levels of intensity. To increase the difficulty, the leg angle is increased up to 15°.

4.2.7 Ruleset CRISS CROSS

Sensor signals

First, the sensor signals (see Figure 4.9) are analysed. During each stepping movement, the axial load increases rapidly, reaching a large peak. In general, this peak is comprised of two local maxima, characteristic for a stepping motion. Reduced characteristics (see third peak) often occur due to a lack of confidence during a backwards step. At the minima of the axial load, users stepped towards one

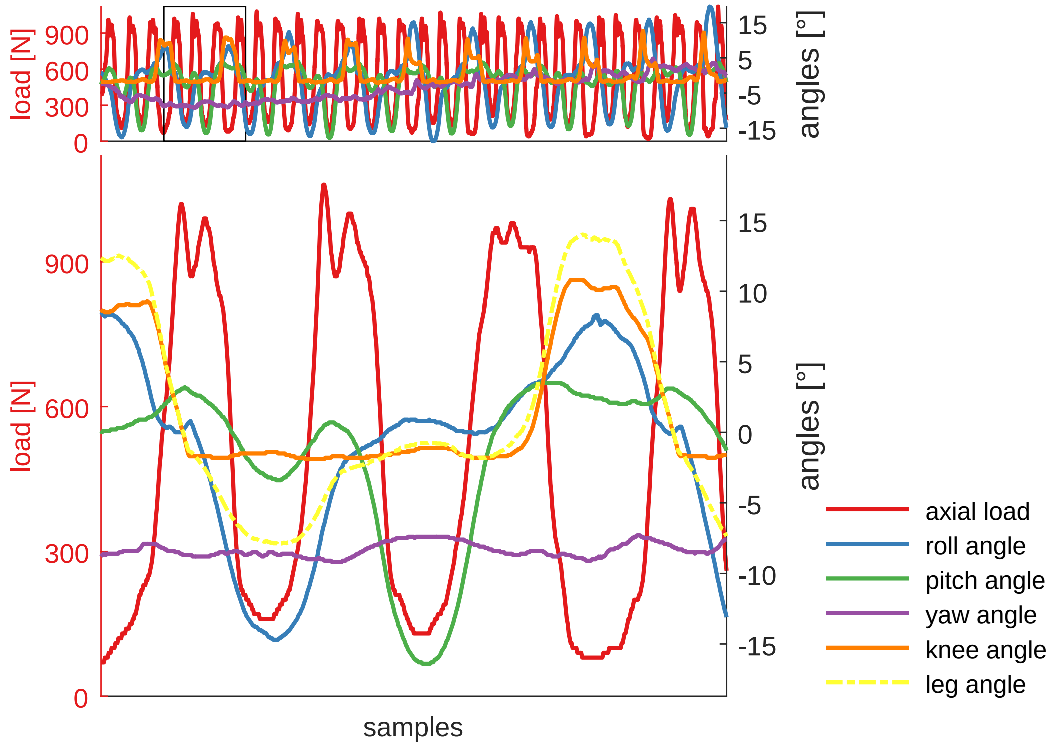

Figure 4.9: Criss Cross – Characteristic sensor signals for correct execution. The top panel depicts a whole set. The bottom panel enlarges the marked single repetition and additionally shows the leg angle.

of the three positions or took an intermediate step, standing in a normal, parallel stance.

When users step to the front (after the first decrease of axial load), the roll angle decreases excessively while the pitch angle decreases only slightly and the knee angle stays between $-1.5°$ to $-2.0°$ which means the knee is fully extended. When they step to the side, the roll angle stays nearly constant, the pitch angle decreases excessively and the knee angle maintains its value. When they step to the back, roll and knee angle increase, but the ratio between the two signals is different for each repetition. The pitch angle decreases only slightly when amputees reach the back position, but the local minimum is larger than for a step to the front or side. To reduce the number of observed signals and to be independent of the changing ratio of roll and knee angle, both measures are combined to the leg angle as in the previous exercise (see Section 4.2.6). The yaw angle contains no further relevant information.

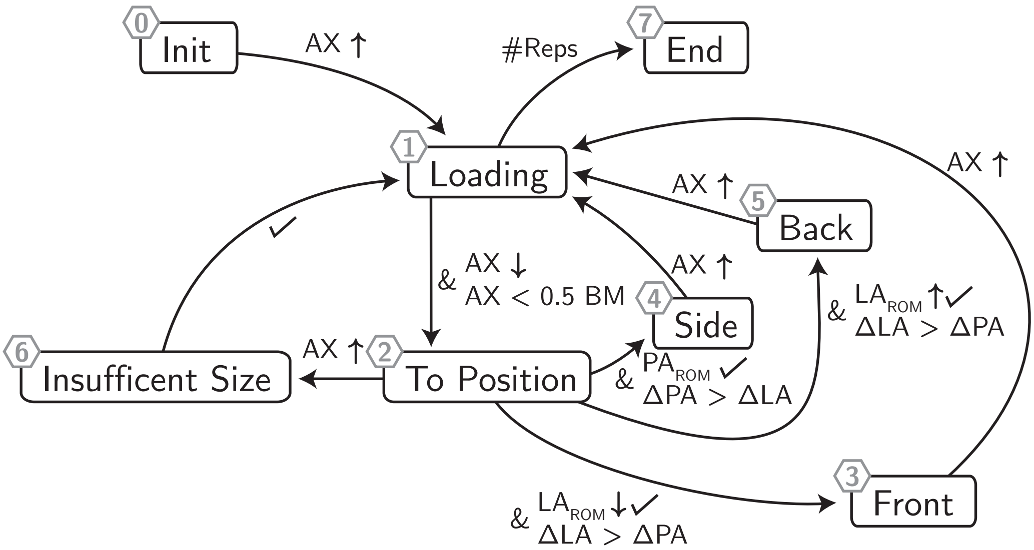

Figure 4.10: Criss Cross – Structure of the ruleset.

Specification

In this exercise axial load, pitch angle, and leg angle are monitored (the other signals do not provide further information). At the end of each stepping movement, users get audible feedback if the requested position was reached correctly or if the overall ROM was too small. At lower levels of intensity, an intermediate step at neutral standing position shall be allowed before moving on to the next position. Since this exercise is very demanding for amputees, this simplification shall allow users with reduced muscle strength and balance to perform the drill as well. The feedback counters shall differentiate between correct positions and undersized steps. Additionally, loading of the prosthesis during the step shall be evaluated. If users reach a position that is correct in general, but not required at that specific point during the overall movement (that is, for example stepping to the front instead of the back), the repetition is terminated and the error counter is increased. Overall, seven feedback counters are available for offline analysis.

Structure

As in the exercises before, the REFERENCE ruleset is activated in rule Init (0) and the ROM-variables are set according to the intensity slider. Once axial load increases, the exercise begins and the ruleset transitions to Loading (1). In Loading (1), the maximum axial load is assessed (@AX_{MAX}) and the reference for the pitch angle (@PA_{REF}) is saved. If @AX_{MAX} does not reach the requested threshold before the rule exits, the respective feedback counter is incremented. When the axial load drops below 50 % of BM, the rule To Position (2) becomes active. In this rule, the sensor signals are evaluated to determine the direction in which the amputee has stepped. Front (3) is activated, if the leg angle decreases below the threshold and the change in leg angle is larger than the change in pitch angle. Side (4) is activated, if the pitch angle decreases below the threshold and the change in leg angle is smaller than the change in pitch angle. Back (5) is activated, if the leg angle increases above the threshold and the change in leg angle is again larger than the change in pitch angle. The change is calculated as the difference of the signal and its corresponding reference variable, @PA_{REF} or @LA_{REF}. If none of the three directions is detected before the axial load increases, the step size is too small and Insufficient Size (6) is activated.

A global variable (@FLAG) is used to store the already successfully reached positions. This variable is binary, using one digit for every position. Hence, at each position (rules 3-5), the respective counter for correct position is incremented. Afterwards, @FLAG is evaluated to check if the current direction is requested. If it is requested, @FLAG is updated accordingly. Otherwise, this step is counted as a mistake. The respective feedback counter is incremented and @FLAG is reset to zero. Hence, the current overall repetition is terminated and users have to correctly reach all three positions again. Furthermore, the counter for intermediate steps is reset at each position, allowing to step back to neutral stance for lower intensity levels. If all positions are reached in the correct order, the counter for overall repetitions is incremented and @FLAG is reset. Equivalently, when Insufficient Size (6) was reached, a mistake was made and therefore @FLAG is reset.

Additionally, the leg angle reference (@LA_{REF}) is determined in Side (4) by averaging the leg angle while this rule is active. In all three positions (rule 3-5), the minimum axial load is stored to detect an increase. When the number of repetitions is reached, the current set is terminated and the ruleset switches to End (7).

Levels of intensity

The requirements for each level of intensity are depicted in Figure 4.11. The range of axial load increases from 60 % to 90 % BM and the boundaries of the ROM for pitch and leg angle are determined to be 6° to 17.5°. The LA_{ROM} has to be exceeded fully

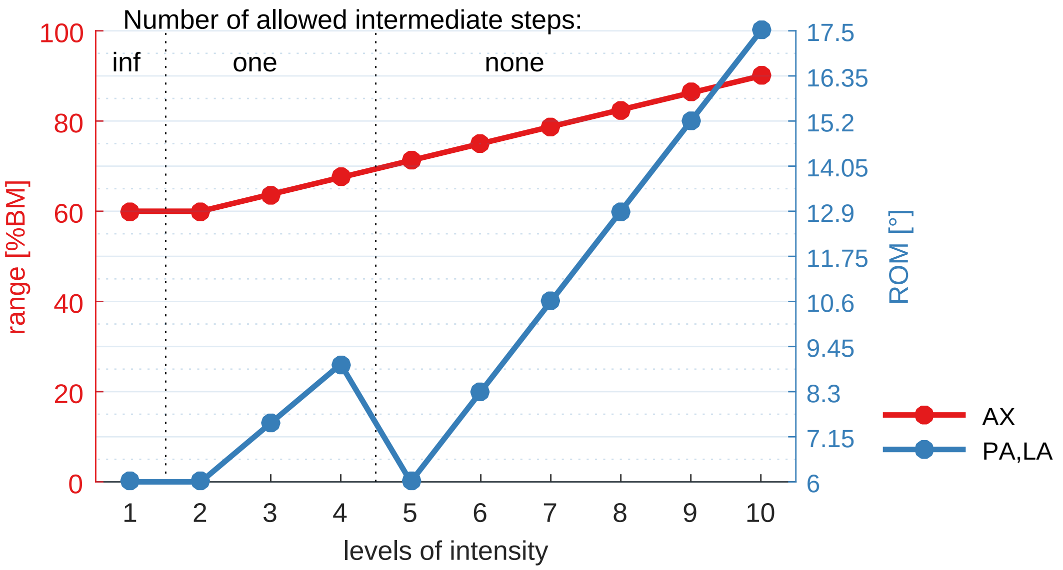

Figure 4.11: Criss Cross – Levels of intensity.

for steps to the front, but only to 80 % for steps to the back since a stepping to the back is demanding high balance and trust from a prosthesis user. The threshold in axial load is large compared to the previous exercises since the ROM variable defines the threshold in this exercise.

For the first level, an infinite number of intermediate steps is allowed and all other signals have minimal thresholds. This allows users to familiarise themselves with the exercise. Due to an infinite number of intermediate steps, less errors are counted and @FLAG is reset less often. Hence, amputees can practise more often before the set is terminated.

In levels two to four, only one intermediate step is allowed and the thresholds are linearly increased up to 9° for pitch angle and leg angle. In levels five to ten, the positions have to be reached one after another without an intermediate step. To simplify the transition, the ROM value of pitch and leg angle is reduced to 6° at level five, before rising linearly to its maximum value. From levels two to ten, the required axial load is continuously increased, because the load shifted onto the prosthesis rather depends on the familiarity with the exercise and the trust in the prosthesis than on the step size. Furthermore, higher axial load is measured for more dynamic movements due to de- and acceleration, while transitioning to the new position.

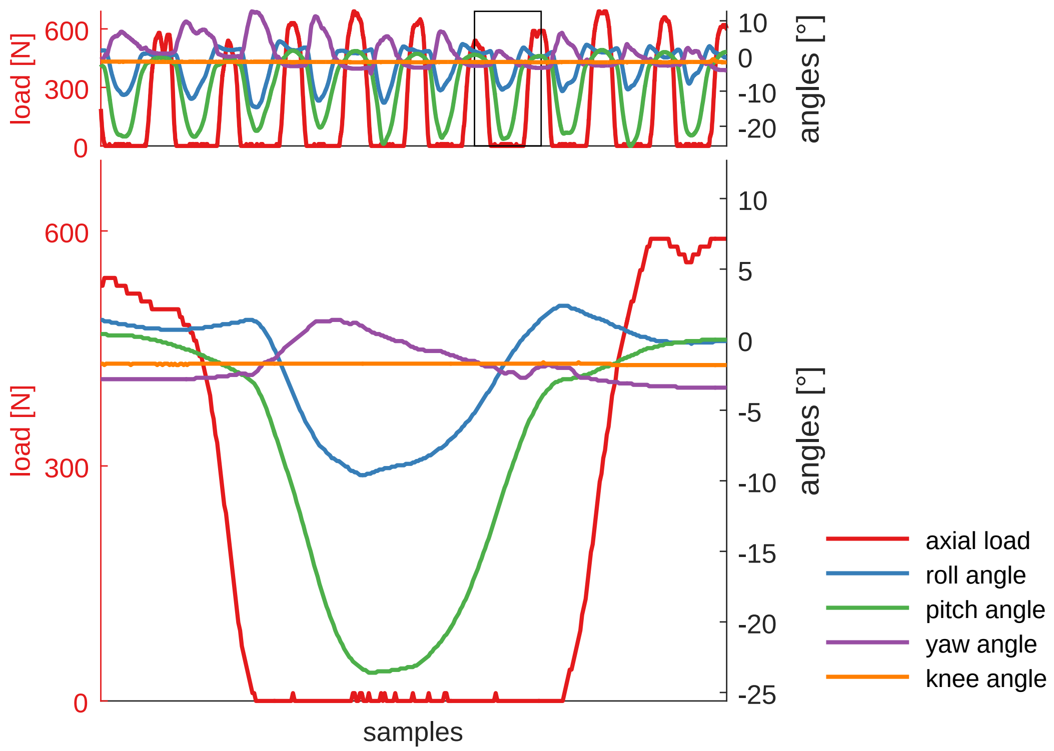

Figure 4.12: Prosthetic Abduction – Characteristic sensor signals for correct execution. The top panel depicts a whole set. The bottom panel enlarges the marked single repetition.

4.2.8 Ruleset PROSTHETIC ABDUCTION

Sensor signals

The sensor signals of a correct repetition are depicted in Figure 4.15. Since users shall always tap the ground with their foot after each repetition, an increase in axial load can be detected. In the depicted signal, the user returns to a neutral standing position after each repetition, resulting in a larger variation of the axial load than can be expected for a correct execution. During the whole exercise, the knee angle is at constant $-1.7°$ since the prosthesis is locked. The pitch angle representing the abduction height decreases during each repetition. If amputees perform the exercise with good form, the roll angle decreases and the yaw angle increases. Since these two signals depend on the form of execution, their range represents the quality of the form.

Specification

During this exercise, pitch, yaw, and roll angle are monitored. Although the yaw angle drifts and depends on the user's orientation in the room, it is used for assessment, assuming that amputees remain in the position established during the initialisation phase. The ROM of yaw and roll angle define good form, hence the maximally allowed motion in external rotation and hip flexion, whereas the ROM of the pitch angle defines the extent of motion for a correct execution. If users perform the drill with bad form during the abduction or adduction motion, the repetition shall be counted as wrong. If the motion is executed correctly, but a mistake is made at the top of the movement, the repetition is counted as correct if the mistake is corrected before beginning the adduction. Users shall get audio feedback if they lose good form until they regain good form. Furthermore, correct abduction, insufficient abduction, and abortions shall be differentiated, resulting in five feedback variables. Finally, the prosthetic knee has to be locked during the exercise to facilitate execution.

Structure

The structure (see Figure 4.13) is derived according to the signals' characteristics and the specifications. In the first rule, Init (0), the REFERENCE ruleset is started and the prosthetic knee is locked by loading rule Locked in DAMPING. Moreover, all global ROM and reference variables are prepared. The reference variables of roll and yaw angle are valid during the set, whereas the pitch angle is updated after each repetition. After these preparations, the ruleset transitions to Unloading AX (1). When the axial load is below 10 % of BM, the rule Basis (2) is activated. This condition needs to have a very low threshold to allow minor deviations of axial load in rule 1. For example, users can shift their weight and get themselves into a comfortable position, but the exercise only starts when they lift the prosthesis.

In Basis (2), the current performance is evaluated. If users lift their leg sufficiently high, rule Correct PA (3) is activated. When they start to adduct their leg again, PA increases, and rule Insufficient PA (4) is activated. In all three rules (2-4) form is assessed. When users either rotate the leg externally or flex their hip, a transition to Bad Form (5) occurs. In this rule the type of form error made is determined and the respective counter is incremented. To guarantee that only one counter is incremented each repetition a variable signals whether bad form was already reached during this repetition. This variable is updated in rule 5.

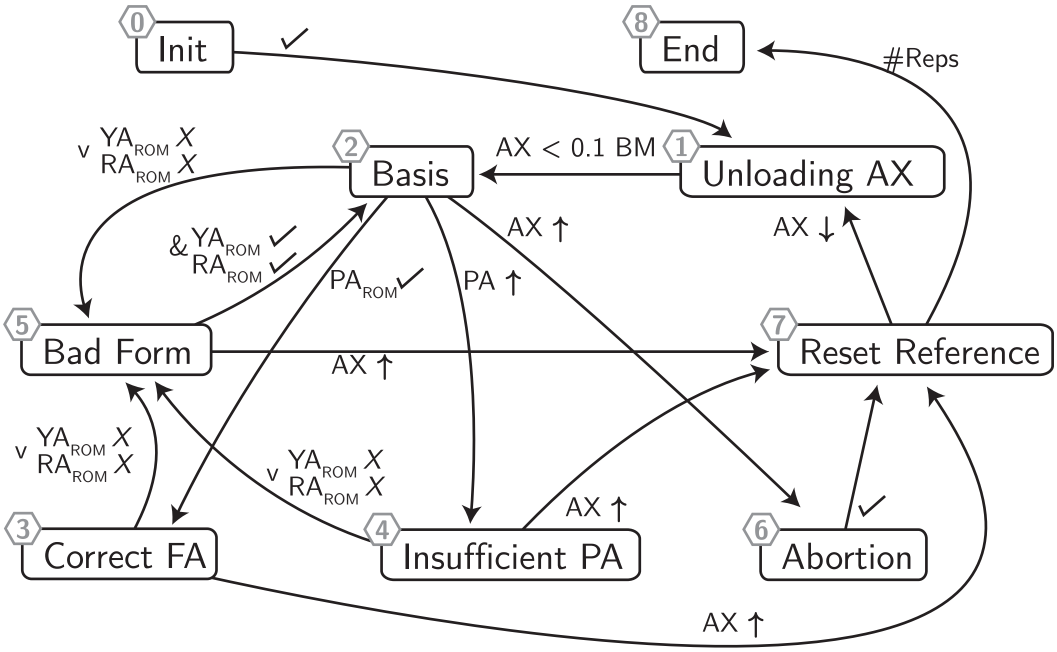

Figure 4.13: Prosthetic Abduction – Structure of the ruleset.

Furthermore, users continuously hear the "wrong form" tone (see Table 4.1) while rule 5 is active.

In rules 3 and 4, variables are incremented on the first entry if users have executed the drill with good form up to this point. Tones are only played on the first entry of either of the two rules.

In rules 2-5, the end of the current repetition is determined by a small increase in axial load. Since the maximum in axial load can be very small if the user's foot taps the ground, this threshold needs to be smaller than in the other exercises. The ruleset always transitions to Reset Reference (7), except for Basis (6), where the ruleset first switches to Abortion (6) before reaching rule 7. In rule 6, a counter is incremented which can be used to inform users that lifting their leg without abducting it is not counted as a repetition. This counter does not influence the number of repetitions.

The rule Reset Reference (7) ensures that the feedback counters for correct and insufficient PA are only incremented if the repetition was continuously performed with good form. When the total number of repetitions per set is reached, the

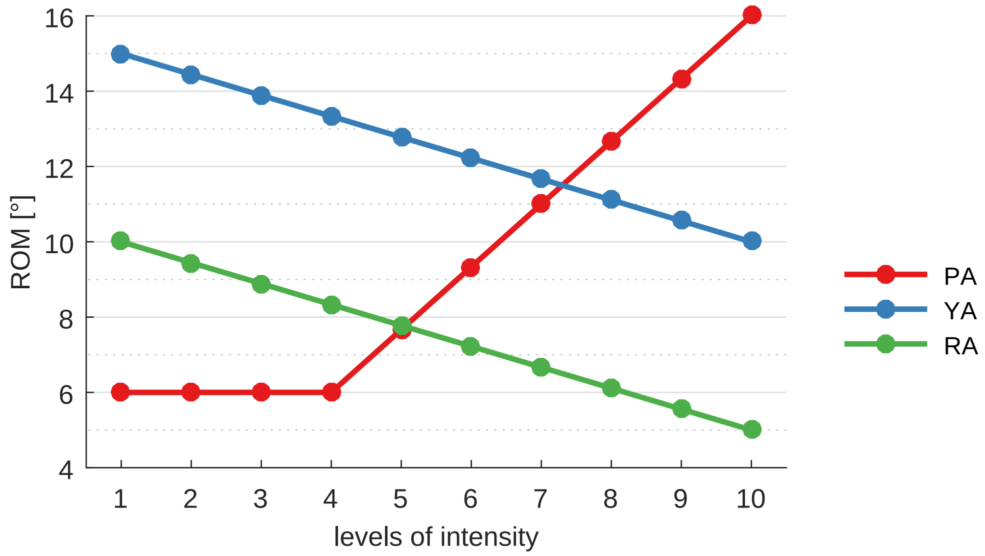

Figure 4.14: Prosthetic Abduction – Levels of intensity.

ruleset transitions to End (8). Otherwise, when a reduction in axial load is detected, it switches to Unloading AX (1) and the next repetition can start by lifting the prosthetic leg.

Levels of intensity

The different levels of intensity are visualised in Figure 4.14. The maximum allowed ROM for the roll angle is linearly decreased from 10° to 5° and for the yaw angle from 15° to 10°. The requested ROM for the pitch angle stays constant and low at 6° for the first four levels. This was requested by the physiotherapist to guarantee that users can concentrate solely on improving their form. When an acceptable form is reached, the pitch angle is increased up to 16°. The maximum pitch angle was chosen to be small, because a good form makes the exercise more strenuous than a higher PA_{ROM} does. Furthermore, low requirements for PA_{ROM} allow the usage of the ruleset while making the exercise more strenuous with a Theraband®.

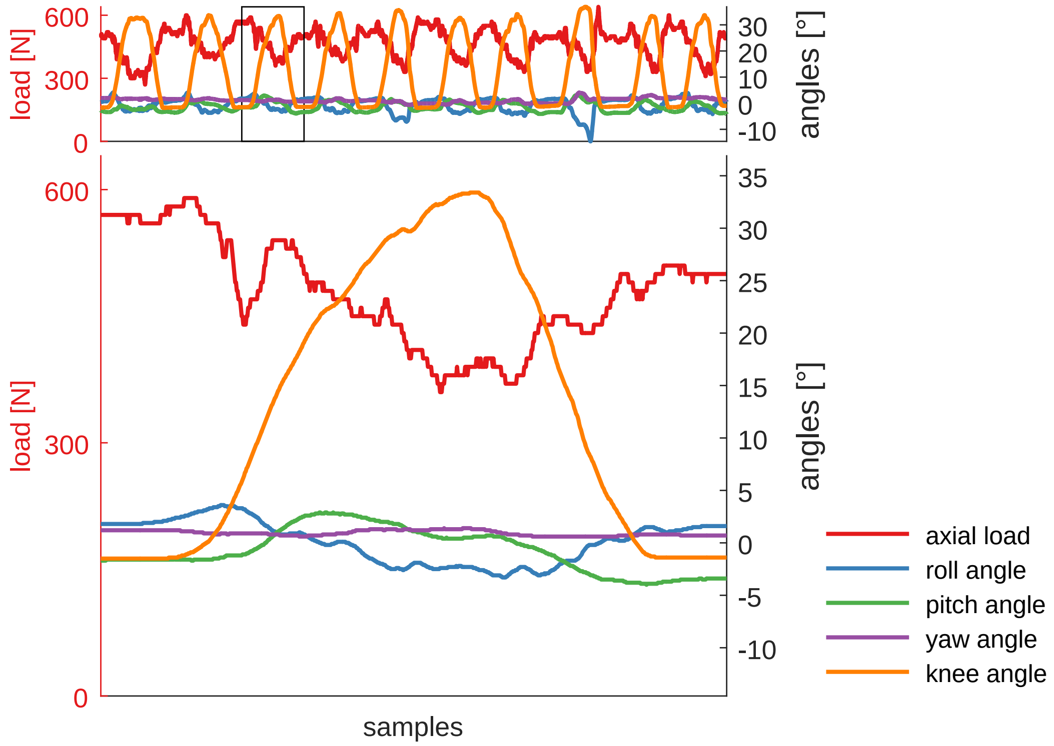

Figure 4.15: Squat – Characteristic sensor signals for correct execution. The top panel depicts a whole set. The bottom panel enlarges the marked single repetition.

4.2.9 Ruleset SQUAT

Sensor signals

The sensor signals for a good repetition are depicted in Figure 4.17. The axial load is nearly constant around half of the BM. With each repetition the knee angle increases rapidly and after reaching the maximum decreases steeply. When amputees push their pelvis back and start to bend their knee, the roll angle is first increasing, but decreases for larger knee angles. A minimum in roll angle is reached for the largest knee angle, before an increase is detected during stand-up and neutral stance. The pitch angle increases during the bending motion and decreases during stand-up, whereas the yaw angle stays rather constant and provides no further relevant information.

Table 4.2: Squat – Damping parameters.

Name	Meaning
@DF	targeted damping value for flexion (at $\phi_K = 0°$)
@KA$_{\text{DF}}$	knee angle, at which the prosthesis is locked during flexion
@DE	targeted damping value for extension (at $\phi_K = 0°$)
@KA$_{\text{DE}}$	knee angle, at which the damping value is increased during extension

Specification

In this exercise, knee angle, axial load, roll angle, and pitch angle are monitored. The ROMs are defined according to the intensity level. The range of the axial load and the ROM of the pitch angle define good form. Hence, these sensor signals must not leave their range. These two criteria are additionally evaluated for each repetition. For a correct execution the knee angle has to exceed its ROM, defining the extent to which the centre of gravity needs to be lowered. The roll angle has to remain within its ROM, defining the threshold at which the heel is classified as lifted. In total, five feedback variables are available.

Users shall get audio feedback as long as the weight is distributed asymmetrically. Furthermore, they shall be notified whenever the execution is classified as correct or incorrect.

In this exercise, the damping has to be controlled differently than in the basis mode of the Kenevo. For this purpose, an additional damping mode is implemented, fulfilling the following requirements: Four parameters (see Table 4.2) define the damping behaviour and are set by the intensity slider. During flexion, the damping value is adjusted to maintain a constant torque for constant knee angle velocity, defined by @DF. To guarantee a locked knee at @KA$_{\text{DF}}$, the damping value is implemented to increase up to 200 QU for an increasing knee angle. The extension control shall facilitate easy extension, but ensure a soft extension stop. Moreover, it shall enhance the training benefit. The gluteal muscles shall be activated by increasing the damping values for extension up to @DE, starting at the knee angle @KA$_{\text{DE}}$. If the prosthesis is not loaded, it must extend easily. Finally, a dead-centre control needs to be implemented for safety reasons, although the dead-centre (see Section 2.2.2) should never be reached according to the flexion control. This guarantees that the prosthesis can always be extended.

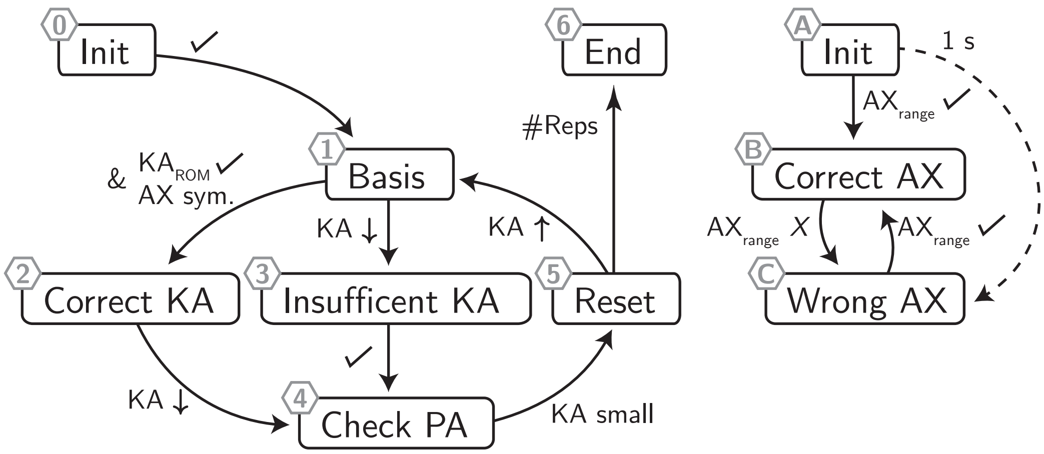

Figure 4.16: Squat – Structure of the ruleset.

Structure

Two rulesets assess the execution of this exercise: The main ruleset, SQUAT, is similar to the other exercises, evaluating the progress of the repetition. The other ruleset, SQUAT_LOAD, continuously assesses the symmetry of the weight distribution and sets the variable @AX_{SYM} accordingly.

First, the ruleset SQUAT will be discussed (see left part of Figure 4.16). As always, the REFERENCE ruleset is started in rule Init (0) and all parameters and reference variables are set. Afterwards, the automaton transitions to Basis (1). A new repetition starts with the entry of Basis (1) and the current repetition is assessed. To allow for an evaluation of roll and pitch angle later on, their minimum values are stored as global variables. If the knee angle threshold is reached, the ruleset switches to Correct KA (2), where the minima of roll and pitch angle are updated. The "correct" tone is played on entry, if the heel was not lifted until then. On exit, the roll angle is evaluated again to check if the heel was lifted during this repetition and the respective feedback counter is incremented.

When the knee angle decreases, the user is standing up and terminates the current repetition. Hence, the ruleset switches to Check PA (4). If the knee angle was never reached, the ruleset transitions from Basis (1) to Insufficient KA (3), where the counter is incremented and the corresponding tone is played. At the following time step Check PA (4) is activated. In Check PA (4), the pitch angle is evaluated to rate the symmetry of the execution, more precisely, to detect any rotation of leg and trunk to the contra-lateral side. The respective feedback counter is incremented if applicable.

When the knee angle decreases below 7.5°, rule Reset (5) is activated. In this rule, the counting of $@AX_{SYM}$ is deactivated and reactivated only when entering rule Basis (1). Furthermore, the $@PA_{REF}$ is updated. When the number of repetitions is reached, the ruleset transitions to End (8), and otherwise, if the knee angle is increasing, it transitions to Basis (1) to start the next repetition.

Second, the ruleset SQUAT_LOAD is discussed (see right part of Figure 4.16). This ruleset starts in Init (A). If the axial load is within the allowed range of AX_{range}, rule Correct AX (B) is activated, where the variable $@AX_{SYM}$ is set to true. Otherwise, the ruleset transitions to Wrong AX (C) after 1 s. Whenever the axial load reaches or leaves the allowed range, a transition to rule B or C, respectively, occurs. In Wrong AX (C), the variable $@AX_{SYM}$ is updated to wrong and on the first entry during each repetition the feedback counter is incremented. Furthermore, the tone "wrong form" (see Section 4.2.2) is played every second to draw attention to the asymmetrically distributed weight.

Levels of intensity

For the target group of the Kenevo has difficulties to execute the Squat with good form but without upper limb support. Hence, they would perform the exercise in a way that does not allow for ruleset monitoring and classification. Therefore, the levels of intensity of the Squat are defined with focus on maximum possible difficulty instead of adjusting the levels to the target user group. Since a different control of damping was designed for this exercise, many parameters need to be set (see Figure 4.17). The initial values for level one are a KA_{ROM} of 25° and a locking angle $@KA_{DF}$ of 35°. Furthermore, the damping value for flexion is set to 180 QU and to 150 QU for extension. The damping value for extension is increased for angles smaller than 17.5°. During the first three levels, only form criteria are made more rigorous. The axial load's allowed range is reduced from 60 % to 30 % BM around half the BM. That is, for level one, the axial load must be within 0.5±0.3 BM, otherwise the respective error-counter is incremented. Furthermore, the ROM for roll angle is decreased from 12° to 8°. The pitch angle ROM is constantly restricted to 5°.

For higher intensity levels, the range for axial load is kept constant since it is very demanding to keep the weight distributed symmetrically. Conversely, the range for the roll angle is slowly decreased to 6° to further improve form. Moreover, the

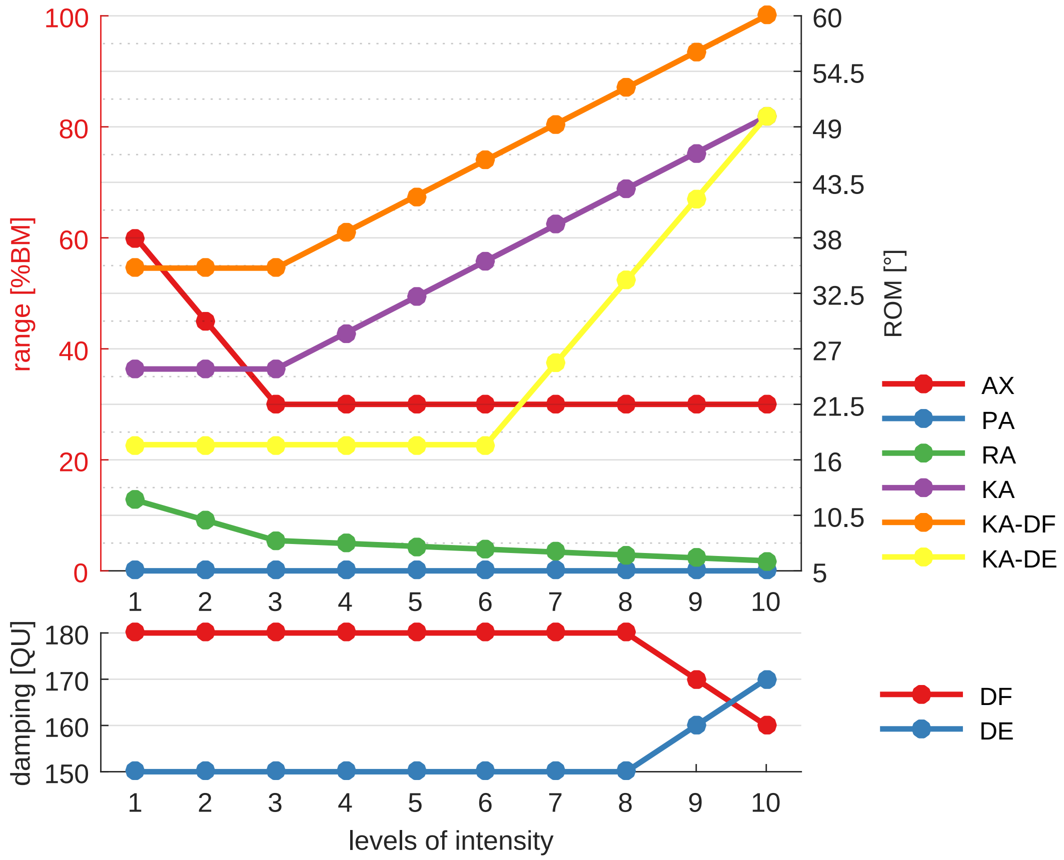

Figure 4.17: Squat – Levels of intensity.

ROM for the knee angle is increased from 25° to 50°, making the Squat physically more demanding. The angle at which the knee is locked for support and safety reasons is always 10° larger than the requested bending depth. Parameters @DF, @DE, and @KA$_{DE}$ which are responsible for the damping are kept constant up to level six. Thereby, amputees can concentrate on improving and internalising the motion before increasing the drill's difficulty through altered damping parameters. The threshold angle @KA$_{DE}$ is increased from level six to level ten. It starts at 17.5° flexion for levels six and below and increases to 50° flexion for level ten. Hence, the muscles are activated earlier. To make the exercise even more strenuous for higher levels, the damping values are modified for level nine and ten: Support during flexion is decreased by lowering the damping values to 160 QU. Additionally, the extension damping is increased up to 170 QU. The higher intensity levels mainly focus on achieving increased impedance to facilitate activating the gluteal muscles.

5. Usability Test

In the previous chapter, the structure of the rulesets was derived by the underlying sensor signals and requirements. To assess the ruleset library, it is essential to test its usability with amputees. During the development phase, one intermediate test was carried out to find conceptual errors and to hone functioning parts. Results of this test are not reported in this work. However, after completing the development, the ruleset was examined with two different tests, each test being performed by a different amputee. In the first test, the agreement between the physiotherapist's professional assessment and the ruleset's classification was validated. In the second, the functionality of the ruleset was assessed with an unbiased amputee, focusing on the feedback given by the ruleset.

This chapter introduces the two test subjects and their amputation details, summarises the used materials and methods and discusses the results of the two final usability tests. The tests analyse the ruleset's ability to correctly classify the exercise execution. Additionally, the general functionality of the ruleset and the effects of audio feedback on the exercise execution are assessed.

5.1 Participants

Amputees of high mobility class (three or four) performed the test. Although these users did not represent the target group of the Kenevo, their high mobility ensured their safety during the test. Testing with users of this mobility class had additional advantages. The users were competent to perform the exercises correctly and to imitate predefined compensatory movements. Furthermore, the amputees stayed energetic and focused despite extensive testing.

Both participants are transfemoral amputees and accustomed to MPKs and shall henceforth be designated as user A and B, respectively. User A is amputated at the right side, has a long stump and is classified as mobility class four. User B is also amputated at the right side, has a medium stump length and is classified as mobility class three. For further details, see Table 5.1.

5.2 Materials & Methods

The tests were performed in the gait laboratory of Otto Bock Healthcare (Kaiserstraße 39, Vienna, Austria). To ensure the amputees' safety, all exercises

Table 5.1: Subject characteristics.

Subject	Age	Stump length	Amputation side	Height	Weight	MOBIS
A	49	long	right	180 cm	95 kg	4
B	63	middle	right	183 cm	90 kg	3

were conducted within or next to parallel bars, facilitating upper limb support. Whenever parallel bars were not appropriate, a self-supporting crutch was used instead.

The tests were performed with the Kenevo and the ruleset library described in Chapter 4. If the amputees' tube and foot were suitable, they were used. Otherwise, an alternative was provided by the company's prosthetist. If necessary, adapters were used to connect prosthesis and socket. An in-house orthopaedic technician aligned the prosthesis, following the general guidelines.

The acquired data was recorded via BioLeg, the in-house development environment. BioLeg allows real-time logging of sensor data, system information, slider values, and global variables. Simultaneously, the amputee's performance and any oral feedback were recorded on video. The data of the prosthesis was transferred to the computer via Bluetooth. Due to restrictions in transmission rate, only a limited number of variables can be logged. If too many variables are logged, the amount of data to be transmitted exceeds the transmission rate and data points are missing in the log file. Consequently, missing data may result in an illogical sequence of active rules and restrict the possibility to simulate this data. Missing data can be detected as a jump in a global variable called cyclecounter which is incremented at each time step. Thus, only variables necessary to reproduce or evaluate the behaviour of the ruleset were logged. These included five sensor signals, that is, the axial load and the four available angles listed in Table 2.1, the feedback counters of the exercise, the IDs of ruleset and active rule of the current exercise ruleset, and the cyclecounter.

To test the agreement between professional and ruleset classification, compensatory movements were determined for each exercise by the attending physiotherapist and by video analysis of previous usability tests with less mobile amputees. These movements are described at the beginning of the results-section of each exercise. The physiotherapist instructed the amputee in these movements.

For each exercise, the amputee performed several sets, imitating one compensation movement per set to achieve similar numbers of repetition. Since the amputee sometimes imitated the compensatory movement incorrectly, each repetition was assessed by the physiotherapist. The rulesets LATERAL WEIGHT SHIFT and ANTERIOR WEIGHT SHIFT classify both loading and unloading phases of a repetition. Since annotating both phases would overextend the physiotherapist, only the loading phases were annotated and compared to the ruleset classification. During the performance, the tones (see Section 4.2.2) were audible.

Due to the knowledge gained from the usability tests, the rulesets of some exercises were slightly adapted. These changes concerned the evaluation of repetitions, but not the damping behaviour of the prosthesis. In case of changes, the logged sensor signals were used to simulate the ruleset classification of the updated library version. The simulation results were saved and used for the evaluation.

The physiotherapist's assessments were digitalised in a .csv file. Expert annotation and ruleset classification were compared using MATLAB®. For each exercise, general information necessary to interpret the physiotherapist's annotation and the ruleset's classification was stored. The analysis was automated using a MATLAB®-script which detects all classifications within one repetition and summarises them giving an overall classification of each repetition. These ruleset classifications are then grouped by the corresponding annotation of the physiotherapist and are presented as classification matrices in the following result section. Hence, each column summarises all classifications of the ruleset which were annotated by the physiotherapist with the column's label. Every column is normalised to its occurrence frequency. Higher saturation of the colour represents a higher occurrence of the respective ruleset classification. Form criteria are additionally evaluated. That is, a certain classification and a form criterion can be detected simultaneously, resulting in high saturation of two classifications (rows) for one annotation (column).

In a second set of experiments, the functionality of the ruleset's feedback was tested. The physiotherapist instructed the amputee in each exercise and the corresponding important criteria. Afterwards, the amputee performed a few repetitions in order for the technician to determine an appropriate intensity level.

For each exercise, the amputee performed four sets, consisting of seven repetitions for the Squat and ten repetitions for all other exercises. The amputee executed the first two sets without audio feedback so that he was unaffected and his current skills were evaluated. After the first set, the ruleset's counters were evaluated on the computer and instructions on how to improve the most frequent error were presented to the amputee. The second set was executed to rate his improvement due to the feedback. Afterwards, the amputee was quickly introduced to the possible tones and their meaning in the current exercise. During the third set these tones were audible. The last set was performed silently again to determine the amputee's developed proficiency without ruleset interference. Finally, the technician surveyed the amputee. The questions concerned the difficulty of the exercise, the user's performance, the comprehensibility of the ruleset's instruction, and the clarity of the tones.

5.3 Agreement between Professional and Ruleset Classification

5.3.1 Lateral Weight Shift

The following four compensatory movements were imitated and analysed:

- Rotating the trunk in the frontal plane while shifting the pelvis (*trunk flexion*).
- Rotating the trunk and supporting with the upper limb by one bar on each side (*trunk flexion & support*).
- Supporting with the upper limb in parallel bars (*support*).
- Excessively supporting with the upper limb and leaning forwards while pushing back the pelvis (*excessive support*).

Generally, repetitions which were annotated as *correct* by the physiotherapist were classified accordingly by the ruleset (see Figure 5.1). If the physiotherapist noted *trunk flexion*, *support*, or *excessive support*, the ruleset detected insufficient weight (@$FB_{\text{INS. WEIGHT}}$). Overall, the exercise was repeated sixty times and only the loading phase was annotated by the physiotherapist. To understand these classifications, the sensor signals of three loading phases are analysed. Figure 5.2 depicts the axial load and the pitch angle for correct execution and the four compensatory movements.

All repetitions with *correct* execution are classified as correct (@FB_{CORRECT}) by the ruleset. During each repetition, axial load and pitch angle first increase simultaneously and decrease afterwards, dropping approximately to the initial value. The covered ROM is large. AX_{range} measures up to 85 % BM and PA_{ROM} up to 17°. The signal of the first repetition deviates slightly before rising up to the maximum due to a short instability of the user. Since the covered range of both signals is large and their deviation is small enough, all repetitions are classified

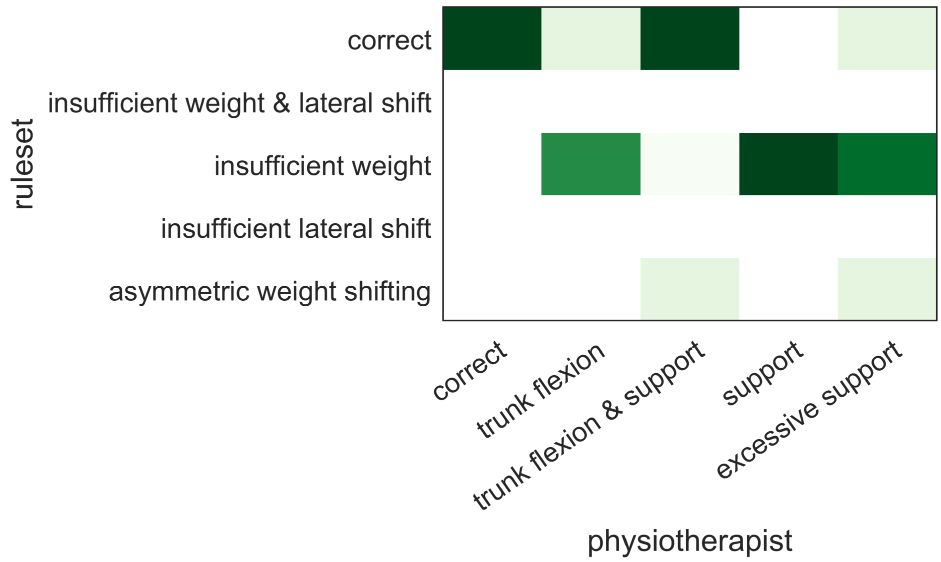

Figure 5.1: Lateral Weight Shift – Classification matrix, comparing the agreement of classification between physiotherapist and ruleset. Higher saturation represents higher concordance.

as correct and the number of detected repetitions corresponds to the number of executions.

Trunk flexion is widely detected as $@FB_{\text{INS. WEIGHT}}$ and two out of eight times as $@FB_{\text{CORRECT}}$. The axial load shows the most unsteady characteristics with lower AX_{range} of 57 % BM, rarely ranging up to 84 % BM. Stability is gained by engaging muscles which connect the trunk to the pelvis. If the trunk is flexed in the frontal plane, these muscles additionally have to coordinate the flexion. Since this supplementary task reduces overall stability, the weight is shifted with reduced balance which explains the noisier signal. In contrast to the reduced AX_{range}, the pitch angle reaches an excessive ROM of up to 26°. By flexing the trunk, the center of mass stays between the feet, but the pelvis can be pushed further laterally without falling sideways. Hence, PA_{ROM} is increased, but more weight remains on the prosthesis during the unloading phase. This higher local minimum reduces the AX_{range} and increases the threshold for the next loading phase. The classification $@FB_{\text{INS. WEIGHT}}$ reflects minor AX_{range} and sufficiently large PA_{ROM}. Nevertheless, AX_{range} was sometimes large enough to fulfil the quantitative requirements, explaining the $@FB_{\text{CORRECT}}$ classifications.

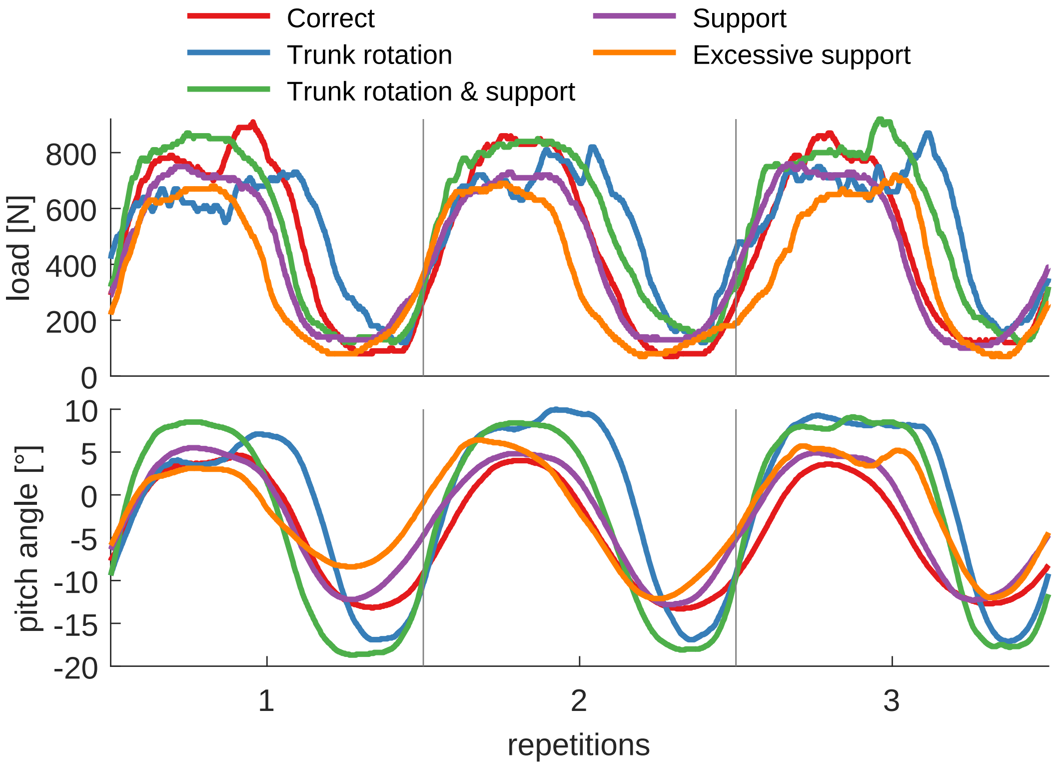

Figure 5.2: Lateral Weight Shift – Sensor signals for correct execution and the four compensatory movements, representing the loading phase of three exemplary repetitions.

Trunk flexion & support is detected as @FB$_{\text{CORRECT}}$. Compared to sole *trunk flexion*, the axial load is smoother and overcomes a greater range. The PA$_{\text{ROM}}$ is similar. As the amputee uses his arms to balance himself, he can push further laterally and thereby increase the weight shifted onto the prosthesis. Due to this increase, the axial load mostly reaches the threshold. Hence, the repetition is classified as @FB$_{\text{CORRECT}}$.

Pure upper limb *support* is classified as @FB$_{\text{INS. WEIGHT}}$. The measured axial load is smooth, but its ROM is reduced to 63 %, whereas PA$_{\text{ROM}}$ is similar to correct executions. The amputee holds some of his weight with his arms, thereby reducing the weight on the prosthesis. Nevertheless, he covers the requested lateral ROM. Accordingly, this scenario is classified as @FB$_{\text{INS. WEIGHT}}$.

In the case of *excessive support* combined with pushing the pelvis back, the ruleset labelled 80 % of the repetitions as @FB$_{\text{INS. WEIGHT}}$ and 20 % as @FB$_{\text{CORRECT}}$. Additionally, asymmetric weight shift (@FB$_{\text{ASYM. WEIGHT SHIFT}}$) is detected which will be discussed later. The *excessive support* reduces both extremal points of the axial load, the weight on the prosthesis and on the contra-lateral leg. Hence, AX_{range} covers 66 % BM. The PA_{ROM} meets the requirements and ranges from 10° to 15°. For this compensatory movement, as well as for pure *support* discussed before, the amputee did not support himself to increase balance, but to reduce the weight on the prosthesis, imitating a beginner who does not trust the prosthesis yet. Consequently, this scenario is categorised as $@FB_{\text{INS. WEIGHT}}$.

Overall, insufficient lateral shift ($@FB_{\text{INS. LAT. SHIFT}}$) is never detected, as PA_{ROM} meets the requirements in all scenarios. Additionally, the design specification is met, in that repetitions are not terminated by small signal deviations (as for example in the correct signal's first loading phase). In the case of sole *trunk flexion* or sole *support* by the arms, the ruleset can detect the reduced weight. Hence, the instructions should specifically target these two compensatory movements and contain hints to keep the trunk above the pelvis, moving them in one unit and using the bars only when starting to lose balance. Asymmetric weight shift ($@FB_{\text{ASYM. WEIGHT SHIFT}}$) was only annotated for 20 % of the repetitions of *trunk rotation & support* and *excessive support*. Since both movements are characterised by reduced balance, weight is shifted in a less controlled manner which can result in an asymmetric movement. The benefit of this feedback counter has to be reconsidered and tested with users of lower mobility. Finally, if errors in the ruleset classification are to be reduced, parallel bars should be used only while the amputee learns to shift his weight onto the prosthesis. When the threshold for axial load is reached, the amputee should execute the drill without support, touching the bars only if necessary.

5.3.2 Anterior Weight Shift

For Anterior Weight Shift, correct execution and two compensatory movements were tested:

- Dynamically pushing the prosthesis into the ground and quickly moving the pelvis forwards and backwards. The weight is never balanced on the prosthesis for a longer time (*dynamic loading*).
- Reducing the weight on both legs by excessive upper limb support. The movement rather looks like a Push-Up (*excessive support*).

Since the classifications by ruleset and physiotherapist agree for all repetitions (see Figure 5.3), all three scenarios can be differentiated by their sensor signals. Both compensatory movements can be distinguished as they violate different

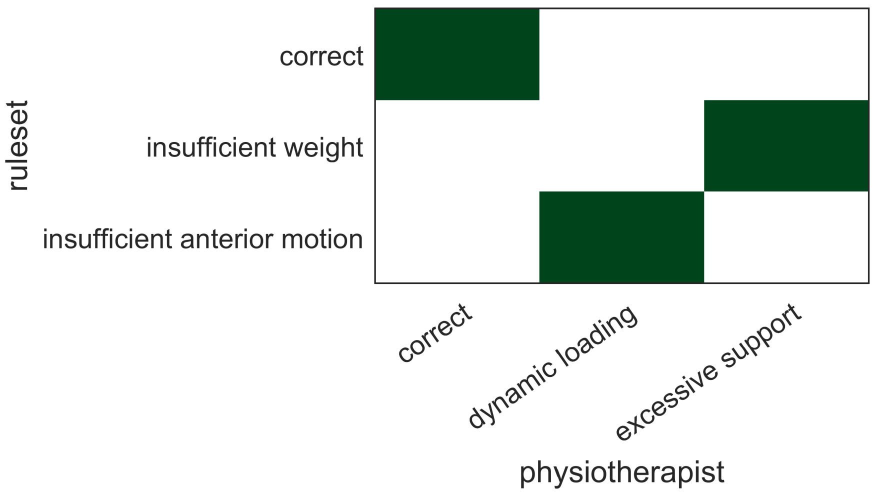

Figure 5.3: Anterior Weight Shift – Classification matrix, comparing the agreement of classification between physiotherapist and ruleset. Higher saturation represents higher concordance.

requirements. Figure 5.4 illustrates the sensor signals of three loading phases which are used to explain the classification.

Correct repetitions show a steep increase in axial load, a slightly uneven plateau while the pelvis is maximally moved forwards, and a steep decrease again while weight is shifted backwards onto the contra-lateral leg, resulting in an AX_{range} of 86 % BM. The leg angle shows a large ROM of 22°.

Dynamic loading of the prosthesis without proper pelvis movement leads to a similar increase in axial load, but reaches a maximum of only 77 % BM and decreases directly after the peak without any plateau. The large peak is measured due to the superposition of a static and a dynamic force component. The static component results from the weight shift and the dynamic one acts due to acceleration of the weight by the dynamic movement. The leg angle changes only by 7.8° due to minor hip movement. It has to be noted that the measured leg angle of this compensatory movement is consistently about 20° larger than for all other movements. The amputee bent his contra-lateral leg and therefore moved the pelvis down and backwards, resulting in larger absolute values for the leg angle. Consequently, *dynamic loading* is classified as insufficient anterior motion ($@FB_{\text{INS. ANTERIOR MOTION}}$).

For the second compensatory movement (*excessive support*), the axial load maximally reaches 42 % BM. The ROM of the leg angle fulfils the requirements with

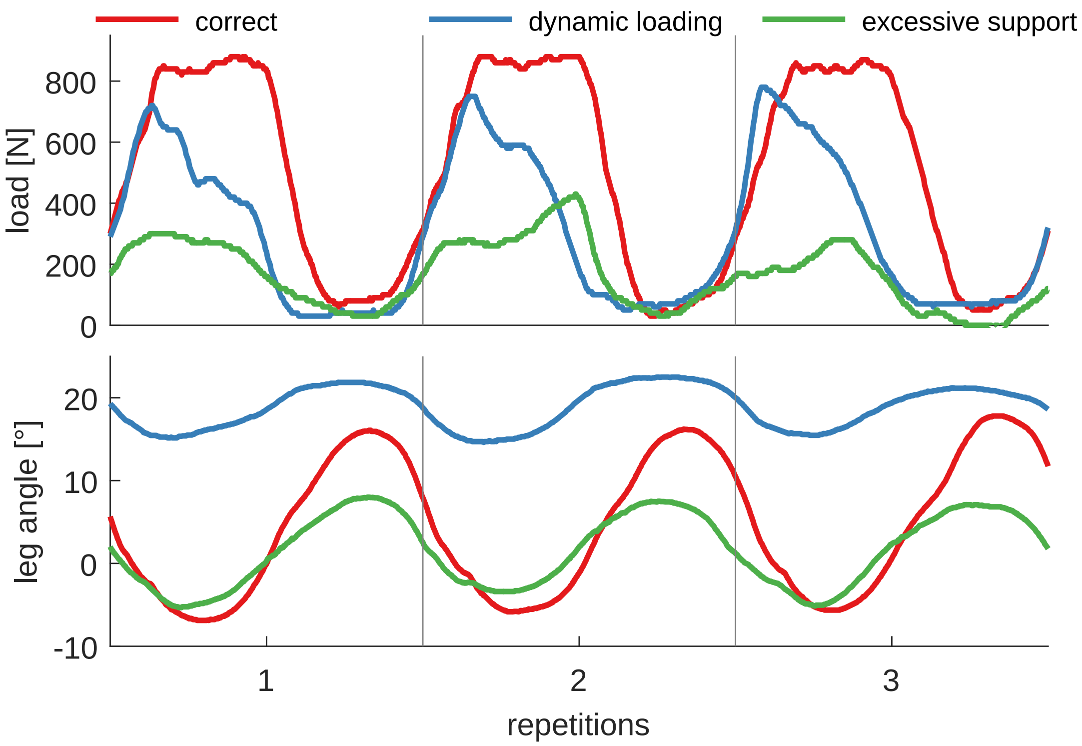

Figure 5.4: Anterior Weight Shift – Sensor signals for correct execution and the two compensatory movements, representing the loading phase of three exemplary repetitions.

values of only 11°, but is smaller than for the correct execution. This behaviour is expected as the patient limits his pelvis movement by the Push-Up position. Thus, these repetitions are classified as insufficient weight (@$FB_{\text{INS. WEIGHT}}$).

5.3.3 Criss Cross

Three different compensatory movements were imitated:

- Stepping with insufficient step size (*insufficient step size*).
- Using a self-supporting crutch and stepping with insufficient step size (*support*).
- Trying to step to the different positions without putting weight on the prosthesis. The movement looks like limping on the spot (*limping*).

All three directions and *insufficient step size* are classified in accordance with the physiotherapist (see Figure 5.5). It has to be noted that the different situations were imitated in an exaggerated manner. These excessive motions did not only

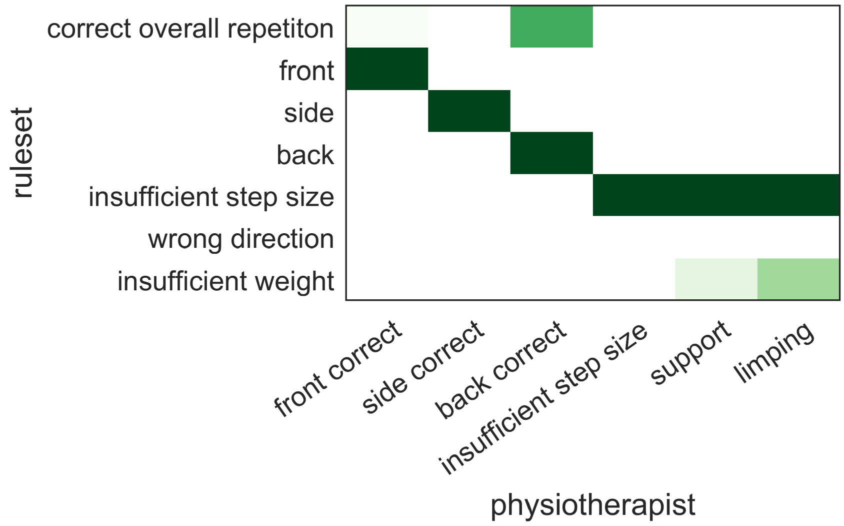

Figure 5.5: Criss Cross – Classification matrix, comparing the agreement of classification between physiotherapist and ruleset. Higher saturation represents higher concordance.

help the physiotherapist to annotate each repetition, but also resulted in extreme signal characteristics which are clearly distinguishable by the quantitative criteria defined in the ruleset. All sets are evaluated at intensity level eight. The related sensor signals are visualised in Figure 5.6.

Correct repetitions show characteristics familiar to walking in the axial load. This includes a sharp increase, two maxima with only a slight reduction in between, and a fast decrease afterwards. The reached maximum measures at least 100 % BM. The leg angle largely decreases while stepping forwards, covering a ROM of 15°. It stays rather constant or increases only slightly while stepping sideways. Stepping backwards, it increases by 13°. The pitch angle increases when the weight is shifted onto the prosthesis. This increase is smaller when the user steps from the front to the side. When reaching a stepping position, the pitch angle always has a local minimum. This is smallest for the step back ($PA_{ROM} \approx 1°$), indicating that the amputee positions his foot nearly behind the prosthesis. When stepping forwards, the pitch angle shows a greater decrease to the minimum ($PA_{ROM} \approx 7°$). This is visible because he positions his feet one foot apart from each other. The largest decrease is noticeable for lateral steps with a covered ROM of 16°.

Smaller steps could be detected as insufficient size (@$FB_{INS.\ SIZE}$). If the amputee makes smaller steps but performs the exercise without additional aids, the axial load reaches the same maximal value. The characteristic of two local maxima is

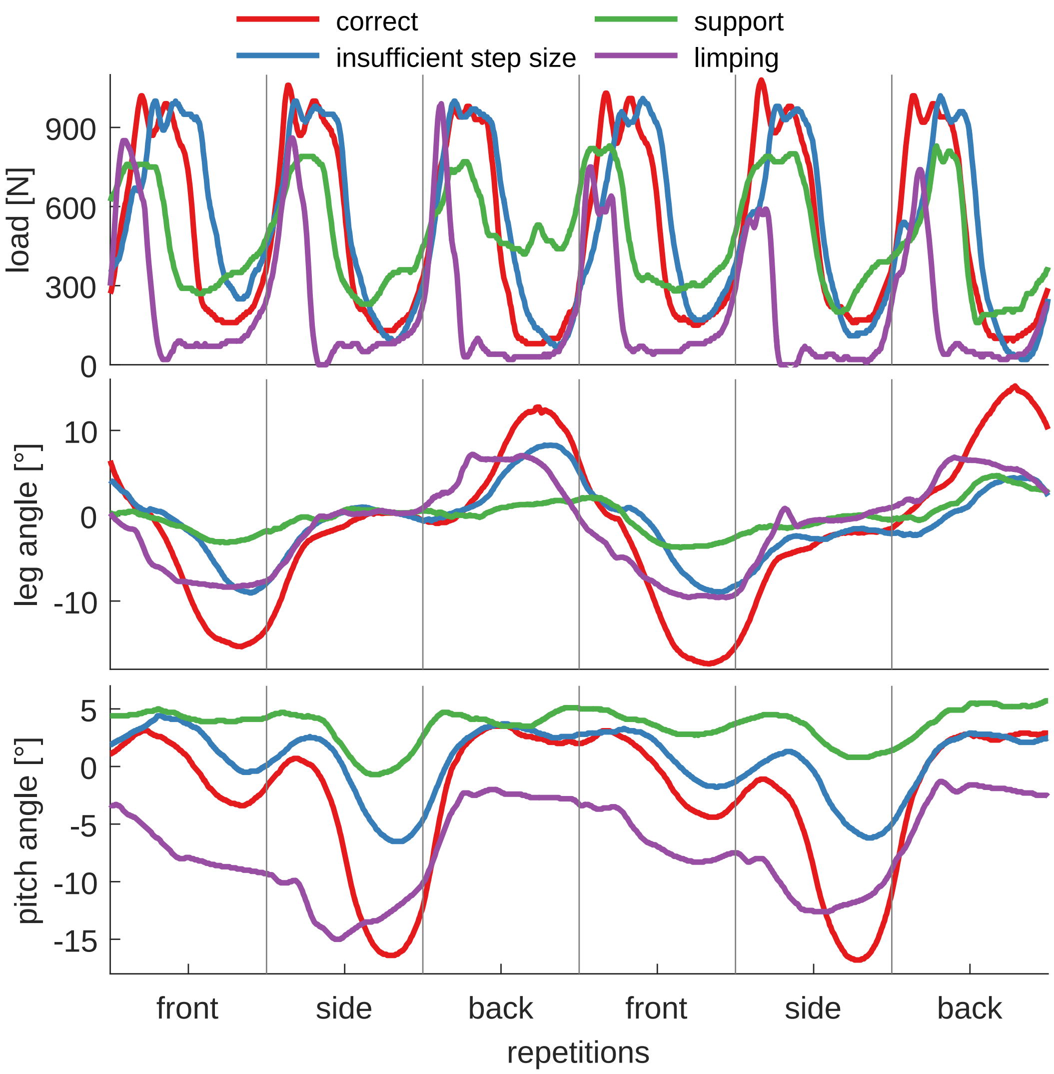

Figure 5.6: Criss Cross – Sensor signals for correct execution and the three compensatory movements, representing all three stepping directions of two exemplary repetitions.

still visible, but the difference between local minimum and maxima is smaller. As the user makes smaller steps, deceleration and acceleration are lower and therefore the respective force component is smaller, reducing the detected difference. Pitch and leg angle show the same features as correct repetitions, only with reduced amplitude. The LA_{ROM} reaches up to 9° for the front position and up to 8° for the back position. The PA_{ROM} covers up tp 9°. Hence, the quantitative criteria for the intensity level used are not met and the repetitions are classified as $@FB_{INS.\ SIZE}$.

The repetitions with additional *support* were detected as steps too small, that is, $@FB_{\text{INS. SIZE}}$. Insufficient weight ($@FB_{\text{INS. WEIGHT}}$) on the prosthesis while stepping was only detected in 22 % of cases. The three signals suggest having the same characteristics, but with drastically reduced amplitude. The axial load only reaches maxima of up to 86 % BM. The pitch angle has larger absolute values compared to correct repetitions since the patient was leaning over the prosthesis, supporting his weight with the crutch located on the prosthetic side. Constantly holding on to the crutch, he did not move the hip a lot which explains the higher measurements but reduced ROM. The leg angle shows equal characteristics to all other movements, but limited ROM.

During *limping*, the repetitions were detected as $@FB_{\text{INS. SIZE}}$. $@FB_{\text{INS. WEIGHT}}$ on the prosthesis while stepping was only detected in 43 % of the cases. For the limping movement, the axial load increases rapidly, has only one maximum, and immediately decreases again. Only one peak in axial load is observed as the user gives only one impulse to shift his weight instead of distinguishable deceleration, balance, and acceleration phase. The absolute value ranges from 62 % to 104 % BM. High values are measured for highly dynamic limping and lower values when the user did not support his weight with the prosthesis, jumping with the contra-lateral leg. Leg and pitch angle show the shape of correct repetitions, but their curves are less smooth as the patient moved jerkily and the signal is resolved over time. Additionally, the pitch angle has lower absolute values as the amputee had less time to step from one position to the next. Therefore, he reduced his sideways motion, never shifting his weight fully back onto the prosthesis. The classification is in accordance with the signal's characteristics. The axial load reaches the requested threshold in 55 % of all repetitions, though its course is different. The exceeded ROM of leg angle and pitch angle was insufficient. Therefore, the detection of wrong loading of the prosthesis must not be detected by a quantitative limit, but by qualitative characteristics.

5.3.4 Prosthetic Abduction

Correct repetitions as defined in Section 3.2 were performed, as well as the following compensatory movements:

- Solely rotating the leg externally (*external rotation*).
- Combining external rotation and hip flexion (*both*).

Although not specifically imitated, some repetitions were annotated as pure *hip flexion* by the physiotherapist.

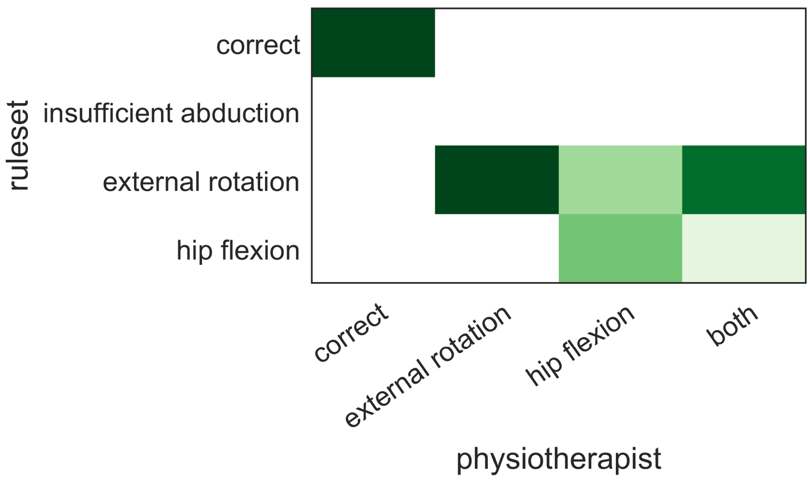

Figure 5.7: Prosthetic Abduction – Classification matrix, comparing the agreement of classification between physiotherapist and ruleset. Higher saturation represents higher concordance.

According to the classification matrix in Figure 5.7, correct repetitions and external rotation can be detected, whereas hip flexion and the combination of external rotation with hip flexion cannot be classified as such. To understand these classifications, the signals of three repetitions depicted in Figure 5.8 are analysed.

Correct repetitions show a reduction in pitch angle with PA_{ROM} of 22°, representing the abduction movement of the prosthetic leg. During the abduction, the yaw angle increases due to an inwards rotation of the foot. The roll angle decreases by 14° since the amputee followed the instruction to extend his hip and lift his leg behind the plane defined by his torso. Hence, the pitch angle fulfils the requirements and pitch and roll angle vary only in the allowed direction, resulting in correct (@$FB_{CORRECT}$) classifications.

Pure *external rotation* could also be categorised by the ruleset. The pitch angle covers a larger ROM than for correct repetitions, the yaw angle decreases drastically, whereas the roll angle slightly increases. Compared to the correct execution, yaw and roll angle behave contrarily. The amputee could increase the abduction height as it is less exhausting to abduct the prosthesis with bad form. Due to the large external rotation of the leg, the front side of the prosthesis is turned upwards. Hence, lifting of the prosthesis is measured not only by a decreasing pitch angle, but also by an increasing roll angle. According to the requirements (see Section 4.2.8), an external rotation must be detected more urgently. Therefore, all repetitions are

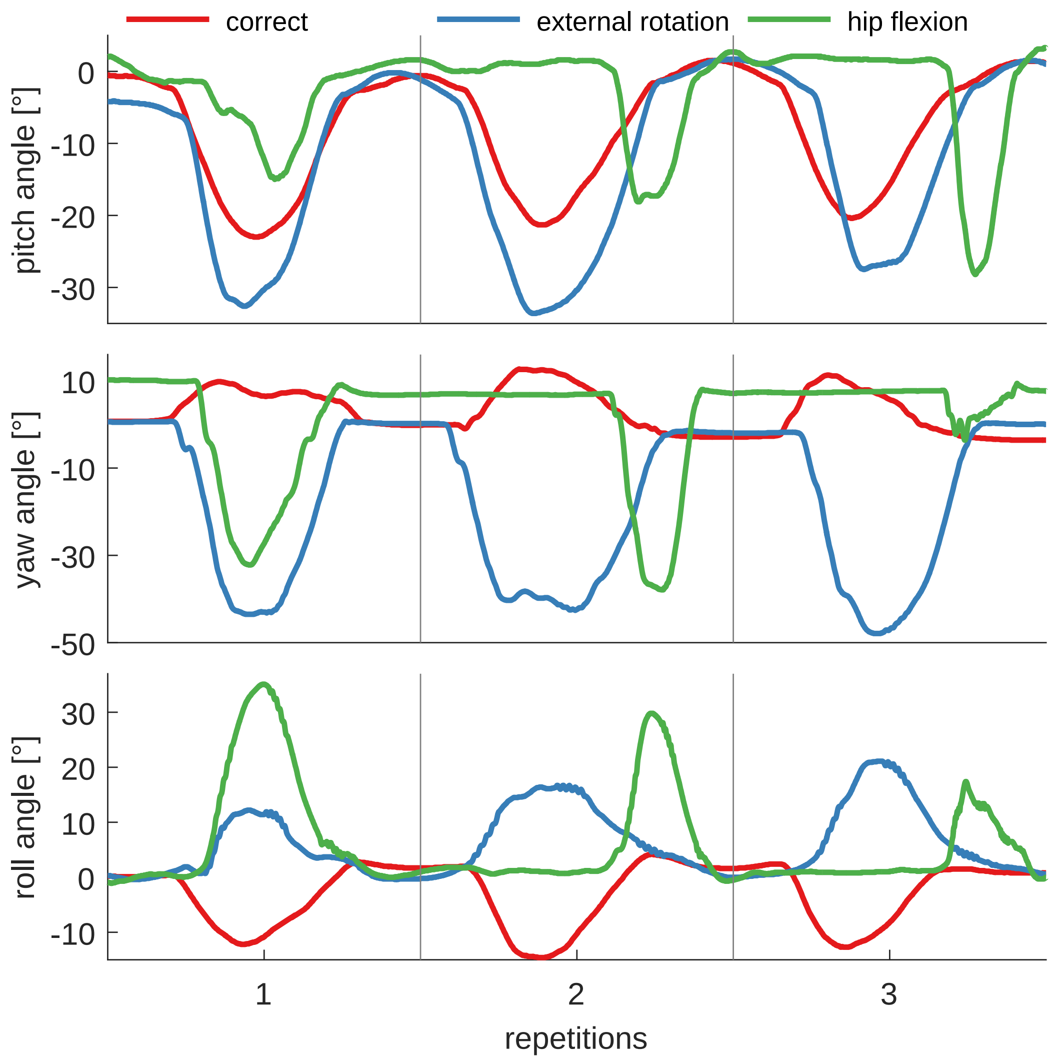

Figure 5.8: Prosthetic Abduction – Sensor signals for correct execution and the two compensatory movements, representing three exemplary repetitions.

labelled with external rotation (@FB$_{\text{EXT. ROTATION}}$), although the roll angle increases by up to 11°.

Repetitions where the physiotherapist criticised *both* form criteria were categorised as @FB$_{\text{EXT. ROTATION}}$ in 80 % of the cases. If the amputee rotates his leg externally and simultaneously flexes his hip, the measured pitch angle decreases less than for correct executions. The yaw angle exceeds the allowed threshold except for the third repetition, where external rotation is reduced and the yaw angle stays within the allowed range. For the first two repetitions, the roll angle increases

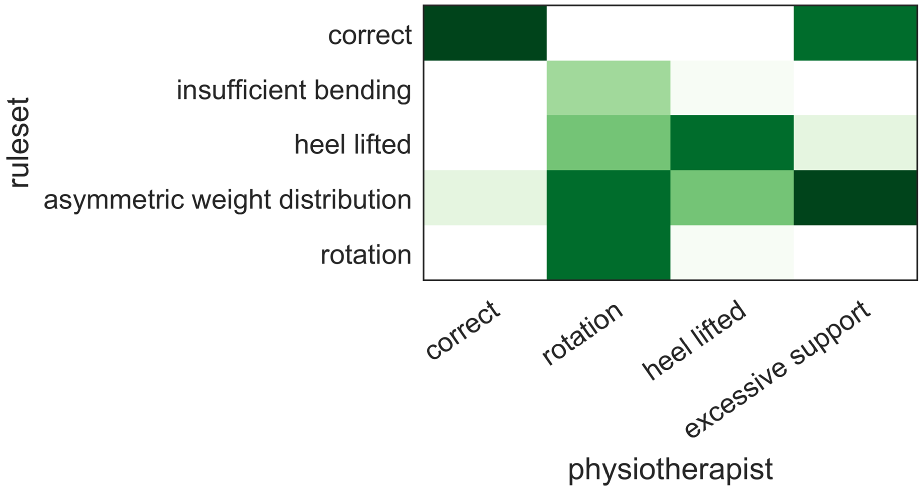

Figure 5.9: Squat – Classification matrix, comparing the agreement of classification between physiotherapist and ruleset. Higher saturation represents higher concordance.

drastically since the prosthesis is externally rotated as well as lifted before the plane defined by the torso. For the third repetition, the roll angle increases the least since the foot points to the front and the roll angle is increased solely by hip flexion. Hence, the first two repetitions are labelled as @FB$_{\text{EXT. ROTATION}}$ and the third as hip flexion (@FB$_{\text{HIP FLEXION}}$) which represents the results of the classification matrix.

5.3.5 Squat

The Squat was performed correctly and three compensatory movements were imitated:

- Lifting the heel (*heel lifted*).
- Rotating leg and torso towards the contra-lateral leg (*rotation*).
- Excessively supporting with the upper limb. The amputee holds on to a bar in front of him and leans back, holding his weight with extended arms (*support*).

Correct repetitions, *heel lifted*, and *support* are detected in accordance with the physiotherapist (see Figure 5.9), whereas the ruleset classification of *rotation* needs to be analysed in more detail. The measured signals are depicted in Figure 5.10.

Correct performance was recognised by the ruleset. Furthermore, a quarter of the correct repetitions are additionally annotated as asymmetrically distributed weight (@FB$_{\text{ASYM. DIST. WEIGHT}}$). Correct executions show a moderate increase in knee angle, that is, flexion of the knee. The roll angle decreases slightly during

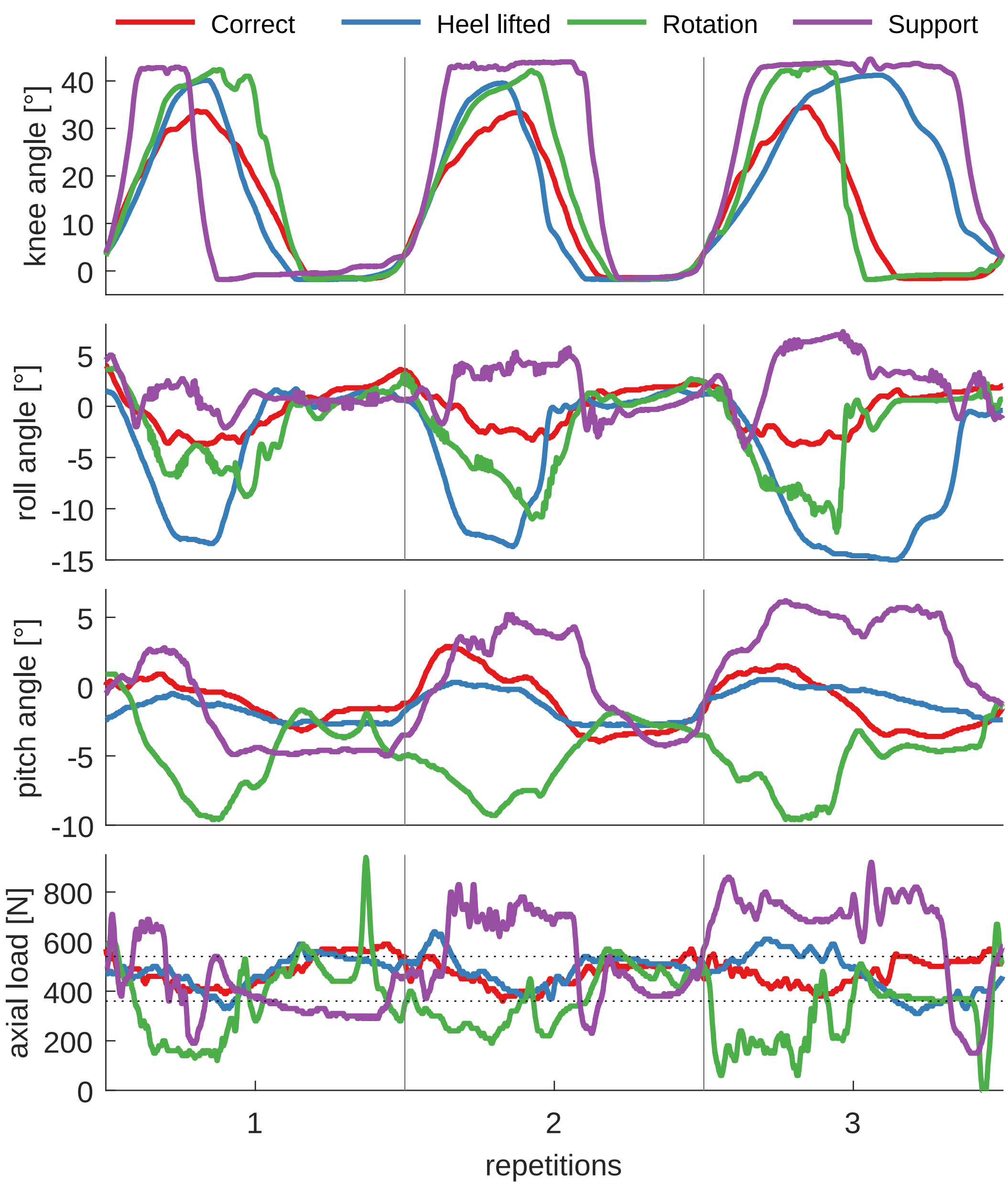

Figure 5.10: Squat – Sensor signals for correct execution and the three compensatory movements, representing three exemplary repetitions. The two dotted lines mark the range of symmetric weight distribution.

flexion, representing a slight dorsiflexion of the ankle and an anterior movement of the knee. The pitch angle increases at the beginning of the bending motion. This increase resembles the lateral movement of the knee relative to the foot. While squatting, the axial load remains between the requested boundaries during the depicted repetitions, having minimum value at maximum flexion. Between the first and second repetition, the axial load exceeds the threshold which has no influence on the assessment of the repetition. Since the knee angle is already small, the repetition is considered to be finished and weight can be shifted arbitrarily to get into a comfortable position before the next repetition starts. Overall, these signal characteristics results in a classifications as $@FB_{\text{CORRECT}}$. A few times, the user put less weight onto the prosthesis than required, violating the additional form criterion (see requirements in Section 4.2.9) and explaining the respective labelling as $@FB_{\text{ASYM. DIST. WEIGHT}}$.

Heel lifted is well detected. A larger knee angle is measured compared to correct execution and the roll angle decreases, covering a RA_{ROM} of 14°. The pitch angle slightly increases during flexion of the prosthetic knee. The weight shifted onto the prosthesis first exceeds the allowed threshold and then drops below the minimum requested threshold at maximum flexion. Hence, the weight is not distributed symmetrically during the execution and $@FB_{\text{ASYM. DIST. WEIGHT}}$ is detected additionally to $@FB_{\text{HEEL LIFTED}}$.

Most of the repetitions with leg and trunk *rotation* towards the contralateral leg are detected as rotated ($@FB_{\text{ROTATION}}$) and additionally labelled with $@FB_{\text{ASYM. DIST. WEIGHT}}$. A large knee angle is reached, again larger than for correct execution. Roll and pitch angle decrease, but the amount depends on the exact execution of the compensatory movement. All repetitions illustrated are additionally classified as lifted heel ($@FB_{\text{HEEL LIFTED}}$) by the ruleset, although only the last one was annotated as such. The axial load is always below the limit requested for symmetric weight distribution. The overlap of rotation and heel lift is reasonable since the physiotherapist annotated them simultaneously three times and the ruleset detected them simultaneously five times.

Executions with excessive *support* are often (80 %) classified as $@FB_{\text{CORRECT}}$ and 90 % of the cases are additionally detected as $@FB_{\text{ASYM. DIST. WEIGHT}}$. Only 20 % of repetitions are detected as $@FB_{\text{HEEL LIFTED}}$. When the patient lifts weight off the prosthesis through leaning away from the bar holding on to it with extended arms, the knee angle reaches the maximum knee angle possible again at which the knee is locked. Since the amputee maintained balance by holding on to the bar in front of him, he could push his pelvis further back. Therefore, his knees were not moving forwards, but rather backwards, resulting in an increase in roll angle during each repetition. The pitch angle increases more during each repetition compared to the other three situations. The amputee pushed his knees outwards as balance was

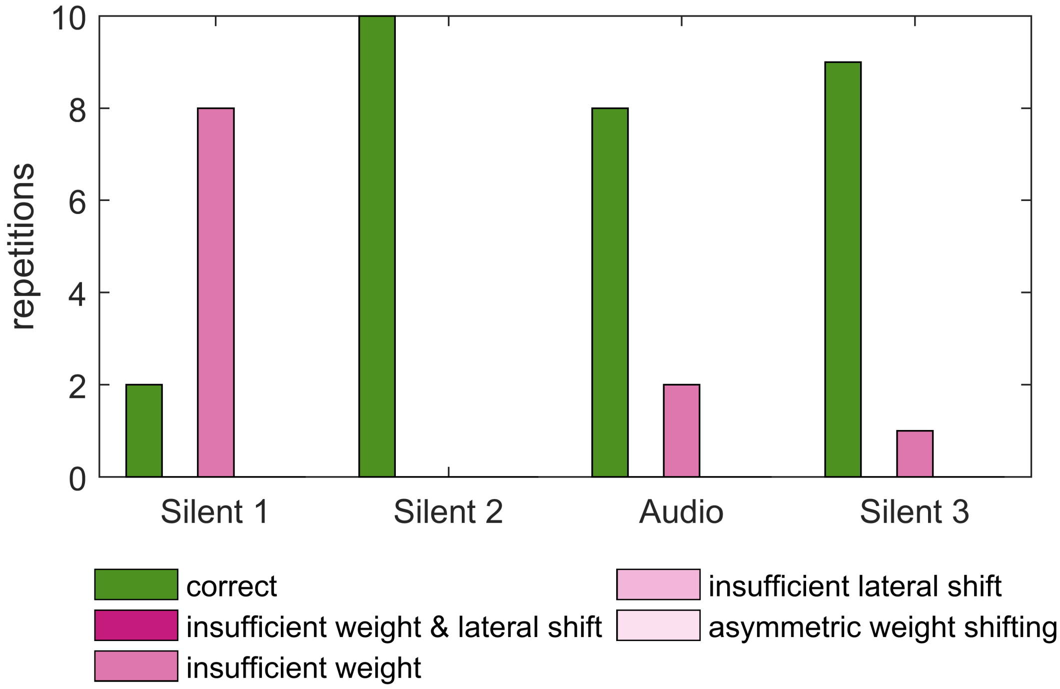

Figure 5.11: Lateral Weight Shift – Performance of all four sets.

guaranteed by the arms. The axial load always exceeds the upper load boundary during flexion of the knee and falls below the lower limit before the amputee is fully standing again. Therefore, excessive support can be detected by means of @FB$_{\text{ASYM. DIST. WEIGHT}}$. Hence, the user is advised to concentrate more on equal loading of the prosthesis and reducing the support by his arms.

5.4 Functionality of the Ruleset Feedback

The classification of the four sets of Lateral Weight Shift are depicted in Figure 5.11. The exercise was executed at intensity level ten. During the first silent set, the amputee performed the exercise correctly twice and shifted insufficient weight onto the prosthesis eight times. Afterwards, he got the instruction to trust the prosthesis, fully stand on it, and shift the pelvis as far as possible, while maintaining balance and feeling comfortable. The physiotherapist could never see insufficient loading and had nothing to criticise about the first set. Then, the amputee performed the second silent set without any mistakes according to the ruleset. The physiotherapist noted a slight rotation of the trunk compared to the first set, but judged it to still be

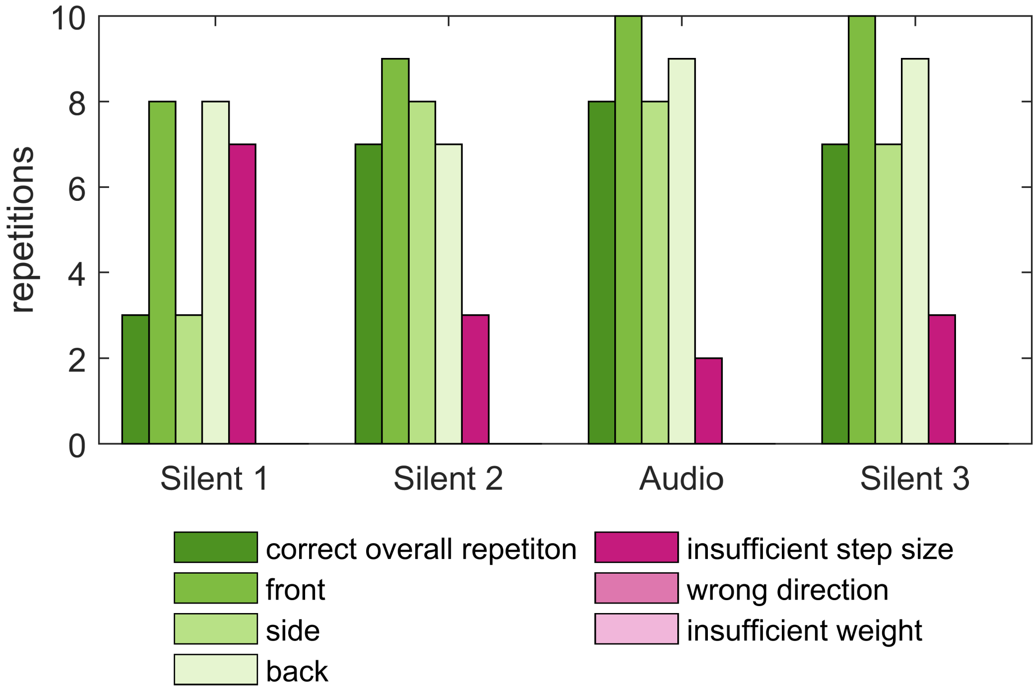

Figure 5.12: Criss Cross – Performance of all four sets.

correct. According to the ruleset, an improvement was achieved. During the third set with audible tones and the final silent set the amputee performed similarly, putting insufficient weight onto the prosthesis twice and once, respectively.

The second exercise, Anterior Weight Shift, was also performed at intensity level ten. Since the amputee is advanced in walking with a prosthesis, he had no difficulties with this drill, executing all repetitions correctly. Therefore, no instruction for improvement could be given and the testing of this exercise was stopped.

Criss Cross was executed at intensity level nine. The details of all four sets are shown in Figure 5.12. During the first set, the amputee performed three correct repetitions and made seven steps with insufficient step size. Furthermore, the amputee executed eight steps to the front and back correctly. Based on this evaluation, the ruleset concludes an insufficient step size to the side. The ruleset's instruction was therefore to step further, especially sideways. This instruction was not appropriate as the user had already made large steps. The physiotherapist noted that the executions were correct from her point of view, noticing only minor instability during two steps to the back. Her suggested instruction was to execute the exercise more slowly, especially the loading and unloading of the prosthesis. This guidance was explained to the patient. In the following set, overall correct repetitions increased to seven and therefore only three steps were detected as too small. Furthermore, more steps to the side were classified as correct such that sufficiently large steps to the back were detected the least. This is in accordance with the physiotherapist who annotated an overall good performance and insecurity while stepping backwards twice. Performance improved according to the ruleset's labelling.

During the audio set, ten steps to the front were correct, eight sideways, and nine backwards, resulting in eight overall correct repetitions and two detections of steps with insufficient size. The final set is classified similarly to the preceding one. Since the exercise was repeated only ten times per set and the number of occurrences of each class varies only by one or two between the last three sets, no conclusions regarding performance due to audible feedback can be drawn from these small variations.

In summary, resetting of the pitch angle needs to be reviewed and the general instructions of this exercise need to be improved. With the instruction to perform the exercise more slowly and to shift more weight onto the prosthesis while stepping, the performance of the amputee enhanced. Nevertheless, according to the ruleset, he never performed all repetitions correctly, although the steps were large enough and the execution speed was appropriate according to the physiotherapist. Hence, the intensity level is adapted in several ways. It was decided to replace level ten's requirements with those of level nine. According to the amputee's feedback, maximal ROM can be reached when intermediate steps are allowed. Therefore, the ROM could be increased from levels two to six, while allowing for intermediate steps, requiring the largest ROM for level six. Furthermore, intensity levels for dynamic motion should have a reduced ROM which is increased from level seven up to level ten.

Prosthetic Abduction was performed at intensity level ten (see Figure 5.13). During the first set, the amputee executed three repetitions correctly, rotated the leg externally six times, and flexed the hip once. Thus, the amputee was instructed by the ruleset to actively rotate his foot internally. Although noting that an increased internal rotation is preferable, the physiotherapist accepted the external rotations and considered the repetitions of the first set as correct. During the next set, external rotation was limited to one occurrence. The amputee executed the remaining nine

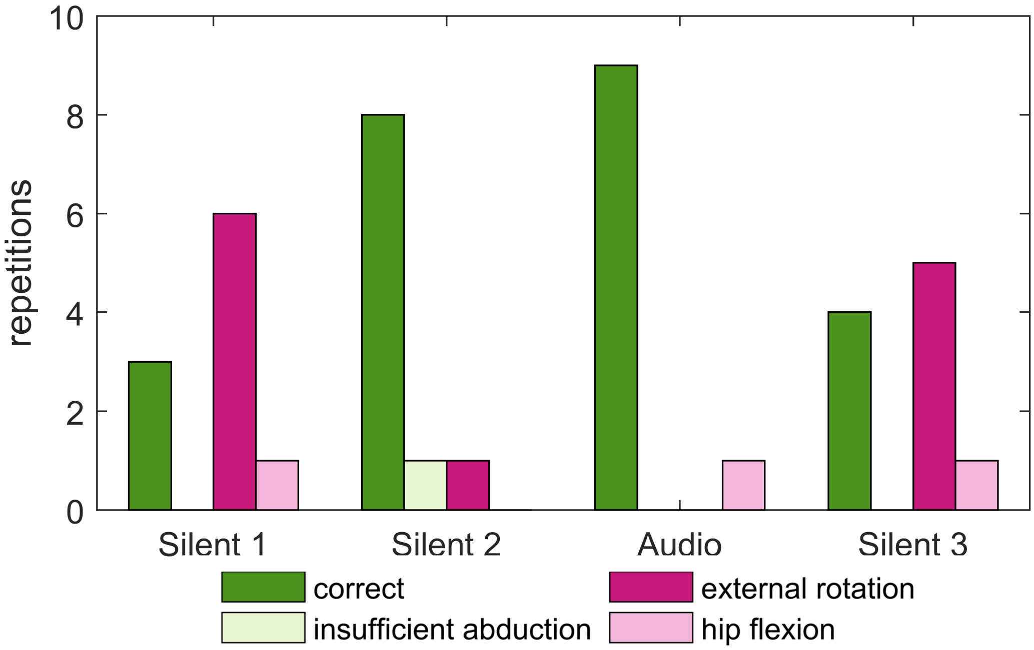

Figure 5.13: Prosthetic Abduction – Performance of all four sets.

repetitions with good form, eight of them also with sufficient abduction height. Hence, performance was improved.

During the audio set, the amputee flexed his hip once. All other repetitions were performed correctly. According to the amputee, he was distracted by the tones and tried to ignore them. Afterwards, the amputee executed the final set. Four executions were correct, five with externally rotated leg, and one with flexed hip. Hence, the performance deteriorated during the last set. Two external factors were noted for the test: Firstly, a few details about the ruleset were already discussed. Therefore, the last set was logged five minutes later and the amputee did not maintain the same focus on the exercise instructions. Secondly, a break was requested after this set due to general inattention.

The amputee executed the Squat at intensity level four. All classifications are summarised in Figure 5.14. Six repetitions were executed correctly during the first set and one was performed with insufficient knee flexion. The weight was distributed asymmetrically during all seven repetitions. A leg and trunk rotation was detected once. Hence, the ruleset instructed the amputee to distribute the weight equally on both legs. This is in accordance with the physiotherapist who noticed a

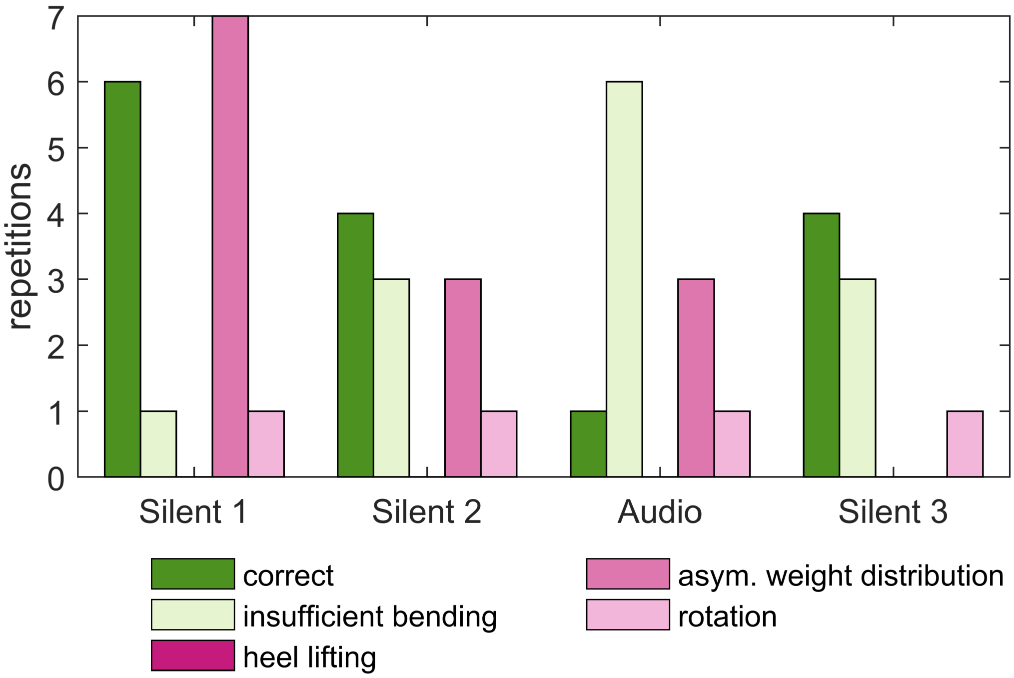

Figure 5.14: Squat – Performance of all four sets.

rotation of the foot, signalling insufficient weight onto the prosthesis and therefore suggested to concentrate on the weight distribution instead of the bending depth. In the next set, the amputee flexed his knee less, concentrating on weight symmetry. Hence, only four repetitions were classified as correct by the ruleset and three as insufficient bending, whereas weight distribution was improved. Asymmetries were only detected during three repetitions. Again, a leg rotation was detected once.

With tones, the execution was very similar, only the bending angle was further reduced. One repetition was classified as correct, the other six as insufficient bending. The physiotherapist remarked that the requested knee angle is too difficult. The amputee noted that he was distracted by the amount of tones which explains his performance during this set. In the final set, the amputee distributed his weight symmetrically during all repetitions. Of the seven repetitions, four were labelled as correct, three as insufficient bending, and one was additionally classified as leg and trunk rotation. Thus, he further improved his performance presumably by developing body awareness of weight distribution with every repetition. According to the amputee, he exerted himself to keep the prosthetic heel on the ground and distribute the weight equally on both legs.

Furthermore, this exercise needed to be tested at a low intensity level of four, although the amputee had higher mobility than the target group. Hence, the difficulty needs to be adjusted to the target group of this prosthesis.

Concluding from all exercises, a general improvement due to the ruleset's instructions was deduced from the changes in classification between the first and second set. It is difficult to draw any conclusions from the audio set as the tones were developed on the fly and introduced very briefly without time for the amputee to ask questions. Furthermore, the perception of tones is extremely subjective and the test was performed with only one amputee performing few repetitions. Nevertheless, the amputee stated that tones could be helpful to support independent training at home. Hence, the ruleset's classification and resulting instruction were helpful in general, but the auditive feedback needs to be revised with a larger test group, representing Kenevo users.

6. Conclusion

The amputation of a lower extremity strongly impedes the mobility of a human. After the operation, the amputee needs to relearn basic activities during rehabilitation [1, 2]. Frequent repetition of rehabilitation-exercises and specific feedback on possibilities for improvement are necessary for an effective rehabilitation. This work represents a first attempt to use prosthetic sensor signals to assist amputees during practice at home.

A collection of finite state machines, called ruleset library, was developed to facilitate autonomous training. Five exercises were analysed with the aid of physiotherapists, biomechanical analysis, and examination of the available sensor signals. A hybrid, synchronous, and deterministic finite state machine, called ruleset, was implemented for each exercise based on specific requirements. These rulesets counted exercise repetitions and assessed the quality of the user's performance. Specific feedback was given to the user at the end of each set of repetitions.

In general, the ruleset classifies repetitions in accordance with the physiotherapist. The ruleset's instructions, derived from the quality of execution during one set, result in improved performance of the amputee. Furthermore, the usability tests show that the ruleset can assess different aspects simultaneously, whereas a human rather focuses on the evaluation of one aspect. Additionally, incorrect weighting of the prosthesis is rarely annotated by the physiotherapist, but easily detected by the ruleset.

Nevertheless, the ruleset can be reviewed in various ways to improve its performance. Either, by enhanced ruleset classification, that is focusing on exercise-specific compensatory movements during further development of a ruleset which simplifies the structure of the ruleset and improves the specificity of classification. Or, by making the ruleset's knowledge available to users by audible feedback. According to the conducted usability test, real-time tones are not perceivably improving performance. This can be explained with the user's feedback that tones were too numerous and difficult to differentiate. However, it was only tested with one user and therefore the aid of audible feedback needs to be further examined. The amputee suggested fewer and better distinguishable tones.

To assess the performance of the ruleset library, each exercise has to be discussed separately as the concordance of classification with the physiotherapist's opinion varies from drill to drill. In general, the main difficulty for ruleset classification is additional support by the arms, influencing the executed motion and hence the measured sensor signals.

In the exercise Lateral Weight Shift, all compensatory movements, apart from a trunk rotation while supporting oneself with the upper limbs, can be detected as mistakes by the ruleset. The current structure of the ruleset evaluates all four combinations of the two sensor signals assessed with respect to the requested threshold. As a consequence, this drill's implementation can be simplified in the future if it is focused on occurring compensatory movements during revision.

The ruleset of Anterior Weight Shift classifies in complete conformity with the physiotherapist. This is a great achievement as the exercise prepares amputees for the stance phase. Nevertheless, this exercise can be improved by evaluating the movement in a more differentiated manner. Corresponding to physiological gait, a bending motion in the knee should be requested first, before assessing the anterior weight shift.

The three stepping directions in the drill Criss Cross can be differentiated by the ruleset. The classification of step size for side-steps is error-prone, depending on the speed of execution. Thus, the reference position of the pitch angle and the associated feedback need revision. In addition, the usability tests showed that the intensity levels should be adapted. Possible changes in the requirements include bigger step size from a neutral standing position and smaller step size for a more dynamic execution without intermediated steps at neutral stance.

The control of Prosthetic Abduction can detect errors, but hip flexion and external rotation are difficult to differentiate with sensor signals. The current ruleset focuses on the detection of external rotation, achieving complete conformity with the physiotherapist for rotation and only 50 % conformity for hip flexion. Improved classification might be achieved by calculating the exact position from the sensor signals to obtain a more detailed evaluation of the prosthesis' position.

Conformity of the ruleset's classification with the physiotherapist has not been reached yet for the exercise Squat. On the one hand, a rotation of the leg and trunk is strongly linked to the lifting of the prosthetic heel. On the other hand, upper limb support changes the execution, resulting in different characteristics of the sensor signals. Only asymmetric weight distribution can be assessed easily by the current ruleset. During development, the difficulty of this exercise was increased to its full potential by changing damping values of extension and flexion. Tests with the resulting ruleset showed that experienced and strong amputees can use the Squat for strengthening, but the current requirements are overwhelming and inappropriate for the actual targeted group. Hence, specifications and difficulty need to be revised to ensure greater usability for the user.

Future work could include additional sensors, monitoring the movement of the trunk and the contra-lateral leg. Currently, some exercises from the catalogue cannot not be implemented due to missing information of segments' movements. Using additional sensor signals might enable to monitor these drills and allow for a more detailed assessment of the current exercises. Though, the reliability of sensors attached to the body is doubtful and therefore must be tested and assured beforehand.

Another improvement can be obtained by varying difficulty in terms of execution speed. That is, slower executions are more demanding. Additionally, the rehabilitation can be improved by implementing drills which prepare the user for specific functions of the prosthesis.

Currently, a mobile application is used to communicate between user and prosthesis, enabling the user to switch between basis mode and exercises. In the future, it could be extended to facilitate real-time evaluation of the exercise, visualise amputees' overall performance, and display the resulting feedback.

Finally, a clinical trial should be conducted. Such a study would include a larger user group, especially representing the targeted user group. As a result, the ruleset library could be adapted more specifically. As part of the study, the levels of intensity could be revised and audio feedback could be further developed. The study could be used to compare the benefits achieved by instructions received from physiotherapists and from prosthetic control. Likewise, the usability of the ruleset could be analysed by surveying amputees and physiotherapists.

In this work, a finite state control was implemented for five exercises. This control determines the number of repetitions and detects erroneous executions. The resulting feedback improves the amputees performance.

References

[1] A. Esquenazi and R. DiGiacomo, *Rehabilitation After Amputation*, Journal of the American Podiatric Medical Association **91**, 13 (2001), ISSN 8750-7315.

[2] G. Fiedler, J. Akins, R. Cooper, S. Munoz, and R. A. Cooper, *Rehabilitation of People with Lower-Limb Amputations*, Current Physical Medicine and Rehabilitation Reports **2**, 263 (2014), ISSN 2167-4833.

[3] B. Rau, F. Bonvin, and R. de Bie, *Short-term effect of physiotherapy rehabilitation on functional performance of lower limb amputees*, Prosthetics and Orthotics International **31**, 258 (2007), ISSN 03093646.

[4] C. Sjödahl, G.-B. Jarnlo, B. Söderberg, and B. Persson, *Kinematic and kinetic gait analysis in the sagittal plane of trans-femoral amputees before and after special gait re-education*, Prosthetics and Orthotics International **26**, 101 (2002), ISSN 0309-3646.

[5] S. Millstein, D. Bain, and G. A. Hunter, *A review of employment patterns of industrial amputees—factors influencing rehabilitation*, Prosthetics and Orthotics International **9**, 69 (1985), ISSN 17461553.

[6] L. E. Pezzin, T. R. Dillingham, and E. J. MacKenzie, *Rehabilitation and the long-term outcomes of persons with trauma-related amputations*, Archives of Physical Medicine and Rehabilitation **81**, 292 (2000), ISSN 00039993.

[7] V. Agrawal, R. Gailey, C. O'Toole, I. Gaunaurd, and A. Finnieston, *Influence of gait training and prosthetic foot category on external work symmetry during unilateral transtibial amputee gait*, Prosthetics and Orthotics International **37**, 396 (2013), ISSN 03093646.

[8] A. Bär, *Mit Prothese (um)gehen*, physiopraxis **3**, 30 (2005), ISSN 1439-023X.

[9] M. J. Highsmith, C. R. Andrews, C. Millman, A. Fuller, J. T. Kahle, T. D. Klenow, K. L. Lewis, R. C. Bradley, and J. J. Orriola, *Gait Training Interventions for Lower Extremity Amputees : a Systematic Literature Review*, Technological Innovation **18**, 99 (2016).

[10] OttoBock, *Kenevo Modernste Technologie Für neue Sicherheit : Kenevo*.

[11] J. Thiele, B. Westebbe, M. Bellmann, and M. Kraft, *Designs and performance of microprocessorcontrolled knee joints*, Biomedizinische Technik **59**, 65 (2014), ISSN 00135585.

[12] L. G. Glynn, P. S. Hayes, M. Casey, F. Glynn, A. Alvarez-Iglesias, J. Newell, G. OLaighin, D. Heaney, M. O'Donnell, and A. W. Murphy, *Effectiveness of a smartphone application to promote physical activity in primary care: The SMART MOVE randomised controlled trial*, British Journal of General Practice **64**, 384 (2014), ISSN 09601643.

[13] K. B. Gátas, Masterthesis, FH Campus Wien (2017).

[14] G. Taeger and D. Nast-Kolb, *Amputationen und Prothesenversorgung der unteren Extremität*, Der Unfallchirurg **103**, 1097 (2000).

[15] Statistisches Bundesamt, *Gesundheit Fallpauschalenbezogene Krankenhausstatistik (DRG-Statistik) Operationen und Prozeduren der vollstationären Patientinnen und Patienten in Krankenhäusern (4-Steller)*.

[16] B. Wilde and R. Baumgartner, *Physiotherapie und Sport nach Beinamputationen* (Thieme, 2000).

[17] I. Matthes, M. Beirau, A. Ekkernkamp, and G. Matthes, *Amputation und Prothesenversorgung der unteren Extremität*, Der Unfallchirurg **118**, 535 (2015).

[18] J. Gille, H. W. Kranz, W. Ulsaß, C. Jürgens, and K. Seide, *Amputationen und Prothesenversorgung der unteren Extremitäten*, Trauma und Berufskrankheit **8**, 108 (2006), ISSN 14366274.

[19] D. B. Seifert, Masterthesis, Technischen Universität Wien (2011).

[20] R. L. Waters, J. Perry, D. Antonelli, and H. Hislop, *Energy cost of walking of amputees: the influence of level of amputation.*, The Journal of bone and joint surgery. American volume **58**, 42 (1976), ISSN 0021-9355.

[21] Centers for Medicare and Medicaid Services and US Department of Health and Human Services and others, *HCFA Common Procedure Coding System (HCPCS)*, Springfield (VA): US Department of Commerce, National Technical Information Service (2001).

[22] R. S. Gailey, K. E. Roach, E. B. Applegate, B. Cho, B. Cunniffe, S. Licht, M. Maguire, and M. S. Nash, *The Amputee Mobility Predictor: An instrument to assess determinants of the lower-limb amputee's ability to ambulate*, Archives of Physical Medicine and Rehabilitation **83**, 613 (2002), ISSN 00039993.

[23] C. Elio, Ph.D. thesis, Westfälischen Wilhelms-Universität Münster (2010).

[24] Otto Bock HealthCare, *MOBIS® System Overview for Otto Bock Modular Lower Leg Pros theses* (2004).

[25] P. G. Trafton, *Lower Extremity* (2003).

[26] G. M. Berke, N. C. Buell, J. R. Fergason, R. S. Gailey, B. J. Hafner, S. M. Hubbard, D. G. Smith, and L. L. Willingham, *Transfemoral Amputation: The Basics and Beyond*, 361 (2007), ISBN 9780615268705.

[27] D. Borrenpohl, B. Kaluf, and M. J. Major, *Survey of U.S. Practitioners on the Validity of the Medicare Functional Classification Level System and Utility of Clinical Outcome Measures for Aiding K-Level Assignment*, Archives of Physical Medicine and Rehabilitation **97**, 1053 (2016), ISSN 00039993.

[28] *DIN 9300: Luft- und Raumfahrt; Begriffe, Größen und Formelzeichen der Flugmechanik (ISO 1151)*.

[29] R. Tomovic and R. McGhee, *A Finite State Approach to the Synthesis of Bioengineering Control Systems*, IEEE Transactions on Human Factors in Electronics **7**, 65 (1966), ISSN 0096-249X.

[30] R. Tomovic, D. Popovic, and R. B. Stein, *Control, Nonanalytical Methods for Motor.* (1995), ISBN 9810220901.

[31] D. Zlatnik, B. Steiner, and G. Schweitzer, *Finite-state control of a trans-femoral (TF) prosthesis*, IEEE Transactions on Control Systems Technology **10**, 408 (2002), ISSN 10636536.

[32] M. Kunze and M. Weske, *Behavioural Models* (2016), ISBN 978-3-319-44958-6.

[33] J. E. Hopcroft, R. Motwani, and J. D. Ullman, *Einführung in die Automatentheorie, formale Sprachen und Berechenbarkeit* (Pearson Studium, 2002), 3rd ed., ISBN 3-86894-082-0.

[34] Z. Kohavi and N. K. Jha, *Switching and Finite Automata Theory* (2009), ISBN 1139483080.

[35] T. A. Henzinger, *The Theory of Hybrid Automata*, in *Verification of Digital and Hybrid Systems*, edited by M. K. Inan and R. P. Kurshan (Springer Berlin Heidelberg, Berlin, Heidelberg, 2000), pp. 265–292, ISBN 978-3-642-59615-5.

[36] P. M. McGinnis, *Biomechanics of sport and exercise* (Human Kinetics, Champaign, Ill. [u.a.], ????), 2nd ed., ISBN 0736051015; 9780736051019.

[37] G. Wu, S. Siegler, P. Allard, C. Kirtley, A. Leardini, D. Rosenbaum, M. Whittle, D. D. D'Lima, L. Cristofolini, H. Witte, et al., *ISB recommendation on definitions of joint coordinate system of various joints for the reporting of human joint motion–part I: ankle, hip, and spine. International Society of Biomechanics.*, Journal of biomechanics **35**, 543 (2002), ISSN 0021-9290.

[38] J. Charmant and Contributors, *Kinovea* (2017).

[39] E. Harless, *The static moments of the component masses of the human body*, Trans. Of the Math-Phys., Royal Bavarian Acad. Of Sci., 8 (1): 69-70, 1860.(Unpublished English Translation, FTD-TT-61-295, Wright-Patterson AFB, Ohio.) (1860).

[40] W. Braune and O. Fischer, *The center of gravity of the human body as related to the German infantryman*, ATI **138**, 452 (1889).

[41] O. Fischer, *Theoretical fundamentals for a mechanics of living bodies with special applications to man as well as to some processes of motion in machines*, BG Teubner, Berlin. ATE 153668. Available from Defense Documentation Center. (1906).

[42] W. T. Dempster, *Space requirements of the seated operator*, WADC-TR-55-159 (1955).

[43] C. E. Clauser, J. T. McConville, and J. W. Young, Tech. Rep., ANTIOCH COLL YELLOW SPRINGS OH (1969).

[44] R. Dumas, L. Chèze, and J. P. Verriest, *Adjustments to McConville et al. and Young et al. body segment inertial parameters*, Journal of Biomechanics **40**, 543 (2007), ISSN 00219290.

[45] R. N. Hinrichs, *Adjustments to the segment center of mass proportions of Clauser et al. (1969)*, Journal of Biomechanics **23**, 949 (1990), ISSN 00219290.

Appendix

A Definitions for the MFCL Classification

These definitions are introduced by the US federal government [21] and also presented in reference [22].

K-Level 0 Does not have the ability or potential to ambulate or transfer safely with or without assistance, and a prosthesis does not enhance quality of life or mobility.

K-Level 1 Has the ability or potential to use a prosthesis for transfers or ambulation in level surfaces at a fixed cadence. Typical of the limited and unlimited household ambulator.

K-Level 2 Has the ability or potential for ambulation with the ability to transverse low-level environmental barriers such as curbs, stairs, or uneven surfaces. Typical of the limited community ambulator.

K-Level 3 Has the ability or potential for ambulation with variable cadence. Typical of the community ambulator who has the ability to transverse most environmental barriers and may have vocational, therapeutic, or exercise activity that demands prosthetic use beyond simple locomotion.

K-Level 4 Has the ability or potential for prosthetic ambulation that exceeds basic ambulation skills, exhibiting high impact, stress, or energy levels. Typical of the prosthetic demands of the child, active adult, or athlete.

B Calculations of the Biomechanical Analysis

This Section presents the MATLAB®-scripts, calculating the forces and moments at all JCs. One script is used for evaluation of exercises performed in the frontal plane, the other for exercises performed in the sagittal plane. Ground reaction forces and angles are used as listed in the Tables in Section 3.3. Additionally, the scripts use the variables listed in Table 1.

Table 1: Variables used in the calculations.

Variable	Description
llegP	length of the prosthetic leg between HJC and AJC
lfootP	height of marker at the prosthetic AJC
lpelvis	distance between CLHJC and PHJC

Analysis in the Frontal Plane

```
Fy = struct();
Fz = struct();
M = struct();

%% Prosthetic HJC
Fy.PHJC = -FyP + 9.81*sum([TMS.Mass('stumpSocket'),T.Mass('prosthesis')]);
Fz.PHJC = -FzP;
M.PHJC = 9.81*TMS.Mass('stumpSocket')*TMS.COM('stumpSocket')*sind(lambdaP)+...
     + 9.81*T.Mass('prosthesis')*(TMS.Length('stumpSocket')+T.COM('prosthesis'))*sind(lambdaP)+...
     + FzP*(llegP*cosd(lambdaP)+lfootP) -...
     FyP*(llegP*sind(lambdaP)-zP);
%% Contra-lateral AJC
Fy.CLAJC = -FyCL + 9.81*TMS.Mass('footShoe');
Fz.CLAJC = -FzCL;
M.CLAJC = FzCL*TMS.Length('footShoe') + FyCL*zCL;

%% Contra-lateral KJC
Fy.CLKJC = Fy.CLAJC + 9.81*T.Mass('calf');
Fz.CLKJC = Fz.CLAJC;
M.CLKJC = M.CLAJC + 9.81*T.Mass('calf')*T.COM('calf')*sind(lambdaCL)+...
     Fy.CLAJC*T.Length('calf')*sind(lambdaCL)-...
     Fz.CLAJC*T.Length('calf')*cosd(lambdaCL);

%% Contra-lateral HJC
Fy.CLHJC = Fy.CLKJC + 9.81*T.Mass('thigh');
```

```
Fz.CLHJC = Fz.CLKJC;
M.CLHJC = M.CLKJC + 9.81*T.Mass('thigh')*T.COM('thigh')*sind(
    lambdaCL)+...
    Fy.CLKJC*T.Length('thigh')*sind(lambdaCL)-...
    Fz.CLKJC*T.Length('thigh')*cosd(lambdaCL);

%% Prosthetic HJC Control
Fy.PHJCC = -Fy.CLHJC - 9.81*TMS.Mass('upperBody');
Fz.PHJCC = -Fz.CLHJC;
M.PHJCC = -M.CLHJC - 9.81*TMS.Mass('upperBody')*(.5*lpelvis*cosd(
    alphaT)-TMS.COM('upperBody')*sind(alphaT))-...
    Fy.CLHJC*lpelvis*cosd(alphaT)-...
    Fz.CLHJC*lpelvis*sind(alphaT);

%% SI units & Table
JointCentre = table(zeros(5,1),zeros(5,1),zeros(5,1),'VariableNames
    ',{'Fy','Fz','M'},'RowNames',{'PHJC','CLAJC','CLKJC','CLHJC','
    PHJCC'});
JointCentre.Fy = struct2array(Fy)';
JointCentre.Fz = struct2array(Fz)';
JointCentre.M = struct2array(M)'*10^(-3); % M = {M}.*10^(-3);
```

Analysis in the Sagittal Plane

```
Fy = struct();
Fx = struct();
M = struct();

%% Prosthetic HJC
Fy.PHJC = -FyP + 9.81*sum([TMS.Mass('stumpSocket'),T.Mass('
    prosthesis')]);
Fx.PHJC = -FxP;
M.PHJC = 9.81*TMS.Mass('stumpSocket')*TMS.COM('stumpSocket')*sind(
    lambdaPs)...
```

```
    +9.81*T.Mass('prosthesis')*(TMS.Length('stumpSocket')*sind(lambdaPs)+T.COM('prosthesis')*sind(lambdaPp)+T.COM_sagittal('prosthesis')*cosd(lambdaPp))...
    -FxP*(TMS.Length('stumpSocket')*cosd(lambdaPs)+(T.Length('prosthesis')-lfootP)*cosd(lambdaPp)+lfootP)...
    -FyP*(TMS.Length('stumpSocket')*sind(lambdaPs)+(T.Length('prosthesis')-lfootP)*sind(lambdaPp)+xP);

%% Contra-lateral AJC
Fy.CLAJC = -FyCL + 9.81*TMS.Mass('footShoe');
Fx.CLAJC = -FxCL;
M.CLAJC = - FxCL*TMS.Length('footShoe') - FyCL*xCL...
    +9.81*TMS.Mass('footShoe')*T.COM_sagittal('foot');

%% Contra-lateral KJC
Fy.CLKJC = Fy.CLAJC + 9.81*T.Mass('calf');
Fx.CLKJC = Fx.CLAJC;
M.CLKJC = M.CLAJC + 9.81*T.Mass('calf')*T.COM('calf')*sind(lambdaCLc)...
    +Fy.CLAJC*T.Length('calf')*sind(lambdaCLc)...
    +Fx.CLAJC*T.Length('calf')*cosd(lambdaCLc);

%% Contra-lateral HJC
Fy.CLHJC = Fy.CLKJC + 9.81*T.Mass('thigh');
Fx.CLHJC = Fx.CLKJC;
M.CLHJC = M.CLKJC + 9.81*T.Mass('thigh')*T.COM('thigh')*sind(lambdaCLt)...
    +Fy.CLKJC*T.Length('thigh')*sind(lambdaCLt)...
    +Fx.CLKJC*T.Length('thigh')*cosd(lambdaCLt);

%% Prosthetic HJC Control
Fy.PHJCC = - Fy.CLHJC - 9.81*TMS.Mass('upperBody');
Fx.PHJCC = - Fx.CLHJC;
M.PHJCC = -M.CLHJC ...
    +9.81*TMS.Mass('upperBody')*TMS.COM('upperBody')*sind(alphaT);

%% SI units & Table
```

```
n=5;
JointCentre = table(zeros(n,1),zeros(n,1),zeros(n,1),'VariableNames
    ',{'Fy','Fx','M'},'RowNames',{'PHJC','CLAJC','CLKJC','CLHJC','
    PHJCC'});
JointCentre.Fy = struct2array(Fy)';
JointCentre.Fx = struct2array(Fx)';
JointCentre.M = struct2array(M)'*10^(-3); % M = {M}.*10^(-3);
```