Improved Gait Cycle Detection for Use in Gait Rehabilitation

Improved Gait Cycle Detection for Use in Gait Rehabilitation

PhD Thesis by

Jan Rüterbories

*Center for Sensory-Motor Interaction (SMI),
Department of Health Science and Technology,
Aalborg University, Aalborg, Denmark*

River Publishers

Aalborg

Abstract

After stroke, hemiparesis is a common problem resulting in very individual needs for walking assistance. Often patients suffer from foot drop, i.e. inability to lift the foot from the ground during the swing phase of walking. Gait rehabilitation can be supported by Functional Electrical Stimulations (FES) which requires a reliable trigger signal to start the stimulations. This can be obtained by a simple switch under the heel or by alternative sensor systems. However, the gait phase detection systems available today show limitations regarding their reliability and usability for ambulatory gait detection during overground walking.

In order to investigate present methods of gait analysis and detection for use in ambulatory rehabilitation systems, a meta analysis on research studies was carried out. Further, a new measurement system based on angular accelerations obtained by differential measurements was developed. The new system was used to investigate the potential of 3D angular accelerations of foot, shank, and thigh to characterize gait events and phases of ten healthy and ten hemiparetic subjects. Subsequently, the real time detection capability of a rule based algorithm was evaluated which detects curve features of the vectorial sum of angular accelerations and maps those to discrete gait states.

This thesis provides an overview of various sensors and sensor combinations capable of analyzing gait in ambulatory settings, ranging form simple force based binary switches to complex setups involving multiple inertial sensors and advanced algorithms. The new measurement system realized a single device setup minimizing the donning/doffing efforts. The system provided gait characteristics as modulated amplitudes of angular accelerations of foot, shank, and thigh. Increasing the gait cadence caused an amplitude increase of the vectorial sum of angular accelerations. A comparison of healthy and hemiparetic gait showed a lower mean of the magnitude of the vector

during the loading response in the hemiparetic gait, while during pre-swing and swing no significant differences between healthy and hemiparetic gait were observed. Further, no statistically significant difference between the tangential components was found for both groups.

The developed gait detection algorithm showed an overall detection rate for healthy and hemiparetic gait of 84.8(18.6) (mean(SD)). The sensitivity was 99.1(1.2) (mean(SD)) and the specificity of 99.8(1.0) (mean(SD)). The algorithm detected gait phase changes earlier than the reference system (foot switches) and showed potential to be implemented in a future FES system.

Sammenfatning

Efter slagtilfælde er hemiparese et generelt problem som resulterer i meget
individuelle behov for at få hjælp til at gå. Ofte lider patienter af dropfod,
dvs. de er ude af stand til at løfte foden fra jorden under svingfasen af
gangbevægelsen.

Gang-rehabilitering kan understøttes ved funktionel elektrisk stimulering
(FES). FES kræver dog et på lidelig trigger signal for at starte stimulation-
erne på det rigtige tidspunkt i gangcyklus. Trigger-signalet kan erhverves
ved hjælp af en simpel kontakt under hælen eller med alternative sensorsys-
temer. De systemer til detektering af gangfasen, der er til rådighed i dag har
problemer med hensyn til deres på lidelighed og anvendelighed for ambulant
gang detektion under normal gang.

For at undersøge de eksisterende metoder til ganganalyse og -detektion til
anvendelse i ambulante rehabiliteringssystemer blev en meta-analyse af den
eksisterende literatur gennemført. Endvidere blev et nyt målesystem til reg-
istrering af vinkel-accelerationer baseret pådifferentielle accelerationsmålinger
udviklet. Det nye system blev anvendt til at undersøge potentialet for brug
af 3D vinkel-accelerationer fra fod, skinneben og lårben til karakteriser-
ing af gangbegivenheder og gangfaser ved måling på ti raske og ti hemi-
paretiske personer. Efterfølgende blev anvendeligheden af en realtids de-
tektering baseret på en regelbaseret algoritme evalueret. Denne algoritme
detekterer kurvetræk for den vektorielle sum af vinkel-accelerationerne for
at kortlægge diskrete gangstadier.

Denne afhandling giver et overblik over forskellige sensorer og sensor-kombinationer
egnethed til at analysere gang under ambulante forhold. Kombinationerne
spænder fra at være baseret på simple kraftmålinger og binære kontakter til
komplekse opstillinger, der involverer flere inerti-sensorer og avancerede al-
goritmer. Det realiserede målesystem indebar en let opsætning af systemet.

Resultaterne viser, at systemet kan afdække egenskaber ved gang i form af modulerede amplituder i vinkel-accelerationerne fra foden, skinnebenet, og lårbenet.

Stigning i gangkadence forårsagede en amplitudestigning i vektorsummen af vinkel-accelerationerne. En sammenligning af sund og hemiparesisk gang viste en lavere middelværdi af vektorsummen under den vægtbærende fase i hemiparetisk gang, hvorimod der under pre-sving og svingfasen ikke blev observeret signifikante forskelle mellem sund og hemiparetisk gang. Desuden blev der ikke på vist nogen statistisk signifikant forskel mellem de tangentielle komponenter for de to grupper.

Den udviklede gangdetekteringsalgoritme viste en generel detektionsrate for sund og hemiparetisk gang på 84,8 (18,6) (gennemsnit (SD)). Sensitiviteten var 99,1 (1,2) (gennemsnit (SD)) og specificiteten på 99,8 (1,0) (gennemsnit(SD)). Algoritmen detekterede faseændringer i gangen tidligere end referencesystemet (fodkontakter) og viste samlet set et stort potentiale til at blive implementeret i et fremtidigt FES system.

This Thesis is based on the following articles:

- [1] J. Rueterbories, E. G. Spaich, B. Larsen, O. K. Andersen, Methods for gait event detection and analysis in ambulatory systems, Medical Engineering & Physics 32 (6) (2010) 545 – 552. `doi:10.1016/j.medengphy.2010.03.007`

- [2] J. Rueterbories, E. G. Spaich, O. K. Andersen, Characterization of gait pattern by 3d angular accelerations in hemiparetic and healthy gait, Gait & Posture`doi:10.1016/j.gaitpost.2012.06.029`

- [3] J. Rueterbories, E. G. Spaich, O. K. Andersen, Gait event detection for use in fes rehabilitation by 3d differential angular accelerations, not published yet

Aalborg, December 2012 *Jan Rüterbories*

Lina, Tira and Marie

Acknowledgements

Das am Rumpfe hängende Bein ist sehr beweglich (Weber and Weber 1836)

Contents

Acronyms

FES Functional Electrical Stimulation

$3D$ Three dimensional

d Distance

r Radius

$MEMS$ Micro Electro-Mechanical System

α Angular acceleration

ω Anguluar velocity

μC Micro controller

SPI Serial Peripheral Interface

PCB Printed Circuit Board

DI Digital Input

C Programming Language

FSR Force Sensitive Resistor

COR Center Of Rotation

a_t Tangential acceleration

a_r Radial acceleration

RF Radio frequency

List of Figures

1

Introduction

1.1 Introduction

Walking is one of the most common human physical activities and plays an important role in our daily activities. It can be performed in a variety of ways and directions and is furthermore a highly energy-efficient method of locomotion. For most people, walking is fully subconscious and requires no thought. This may change following a stroke, which may result on patient specific need for rehabilitation of locomotion. Restoration of walking can be supported by Functional Electrical Stimulation (FES), which has become an estimated rehabilitation method [5].

One of the first systems to correct foot drop using electrical stimulation was developed in 1961 [6] and many others followed in the years after (see review by Lyons et al. [5]). To achieve a successful timing of the stimulation, it is necessary to receive information from internal sensors such as recordings from the sensory nerves [5] or from external sensors. Nowadays, these input data is obtained from electronic sensors that measure various parameters during the gait cycle.

The definition of individual gait events and phases is a starting point for nearly all aspects of gait analysis and restoration. Hence, gait is considered in terms of temporal and spatial components. The temporal components are those periods of time during which events take place. The spacial components refer to the position and orientation of limbs and joints. For analysis of gait, it is important to consider both aspects, since a disease or trauma can affect the gait components independently [7].

The physical ways to analyze gait are based on the measurement of leg motion or

ground reaction forces to detect gait events in the signals. For all FES stimulation devices, accurate and reliable detection of the gait cycle is essential, which is traditionally obtained by a foot switch placed under the heel. Often patients do not benefit sufficiently from stimulation because false triggering is caused by irregular gait patterns or because they touch initially the ground with their forefoot instead of the heel. However, controller feedback can also be provided by sensors like goniometers, accelerometers, angular rate meters, inclinometers, or force sensitive resistors, that are suitable to determine joint angles, body segment acceleration, body segment velocity, tilt angle, and times of foot contact, respectively. Many different sensor configurations were used for gait cycle analysis or detection in humans [1]. However, they show limitations regarding their reliability and usability [8, 9], such as donning, doffing, cosmetic issues, and energy consumption for ambulatory gait detection during overground walking.

1.2 Aim of the PhD project

The aim of this PhD study was to solve the problem of accurate and reliable gait cycle detection under the perspective of a daily application within a FES system to be used during overground walking conditions.

The aims of this Ph.D project were:

A: analysis of the state of the art of methods and systems for gait detection

B: development of an improved gait detection system

C: investigation of gait kinematics of the target group, i.e hemiparetic individuals

D: development of a gait detection algorithm overground walking

1.2.1 Research Questions

1.2.1.1 Aim A:

Various measurement instruments were developed in the past decades to assess gait. Their complexity ranges from simple switches to highly integrated sensor devices used in a network of body segments. The analysis of the state of the art of methods for ambulatory gait measurement addressed the following questions:

1. What is the state of the art in ambulatory gait measurement systems?

2. How well do these systems work?

3. How complex are they regarding daily usage?

4. Which limitation do they have?

1.2.1.2 Aim B:

Gait events can be extracted out of various data such as force signals of the foot, joint angles, body segment velocities or accelerations. The development of an optimal system requires the investigation of requirements and constraints to create a dedicated system setup.

1. Which requirements should an optimal system to detect ambulatory gait events of healthy and hemiparetic gait meet ?

2. Based on the previous research results of aim A and the analysis of requirement, a new measurement system was to be created.

1.2.1.3 Aim C:

Aim C was to improve the understanding of impaired gait patterns affected by hemiparesis based on the technology chosen in Aim B which involved 3-dimensional lower limb kinematics. The kinematic measures consisted of radial and tangential accelerations of the foot, shank, and thigh. The addressed research questions were:

1. How are gait patterns represented in acceleration signals?

2. How does the signal depend on the measurement positions of foot, shank, and thigh?

3. How does the signal change due to gait velocity?

4. Is there a benefit of measuring accelerations in the frontal plane?

1.2.1.4 Aim D:

The challenge of gait detection is to develop algorithms that determine gait events while the person is walking, based on data obtained from sensor measurements. Traditionally, this has primarily involved foot switches (mainly FSR and threshold determination) to obtain data related to certain foot contact events. All alternatives to foot switches need some additional processing in order to detect gait events [1]. Aim D addressed the following questions:

1. Is it possible to detect gait events in real-time in radial and tangential accelerations?

2. How reliable is the detection of gait events?

3. How accurate and precise is the detection of gait events?

1.3   Organization of the thesis

The thesis is divided into 4 chapters. Chapter 1 introduces the topic to the reader and provides necessary background information about human gait (1.4.1) and inertial sensors (1.4.2). Chapter 2 contains the methodological considerations of the three studies: prior art (2.1), the measurement system (2.2), and the detection algorithm (2.4). Chapter 3 presents and discusses the results. Chapter 4 summarizes and concludes the thesis and provides a perspective for future work.

1.4   Background

1.4.1   Gait

The terms *gait* and *walking* are often used equivalently. However, gait describes the way or 'style' of walking and is the basis to compare the walking of different subjects. This is of interest since the earliest times tracing back to Aristoteles and his theories on the movement of humans and animals [10]. First measurements of walking were performed by the Weber brothers in Göttingen [11] who used a telescope with a calibrated graticule to assess gait by vertical movements of significant anatomical landmarks [12]. Jules Etienne Marey [13] investigated force and pressure measurements of the foot in
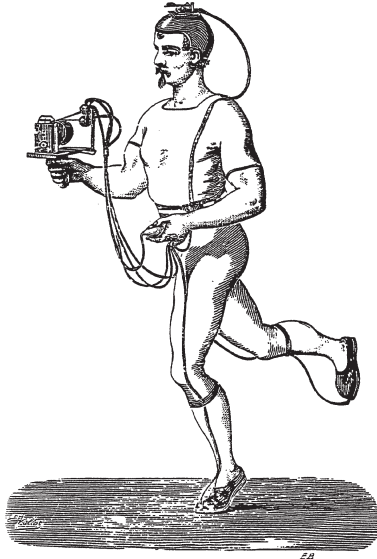
(a) Ambulatory measurement system, consisting of pressure measurement at the foot and light strobe pictures of the movement [14, 13].
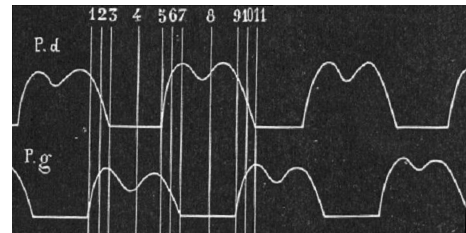
(b) Recorded pressure curves of the ground reaction force [14]

Figure 1.1: Ambulatory gait measurements in 1873

collaboration with his student Gaston Carlet. Carlet developed the first ambulatory measurement instrument in 1871 consisting of a shoe comprising three pressure transducers built into the sole (fig.1.1a) to record the forces exerted by the foot on the floor (fig.1.1b) [10]. Further, Carlet was the first who recorded the double support phase. Various other experimental techniques were used and Carlet's thesis, published in 1872, concluded with a succinct description of the normal human gait cycle which is essentially accurate [10].

Human walking consists of consecutive gait cycles. During each gait cycle, a sequence of events takes place that mark the transitions from one gait phase to another. Figure 1.2 illustrates gait events and phases on a normalized timescale.

Gait cycle A normal gait cycle ends and begins by definition with *heel strike* [15, 7]. This event is the initial ground contact of the leading limb during normal walking. This is also the beginning of the *load response phase* during which the leading limb takes over the body weight by placing the whole foot on the ground. The events *heel strike*
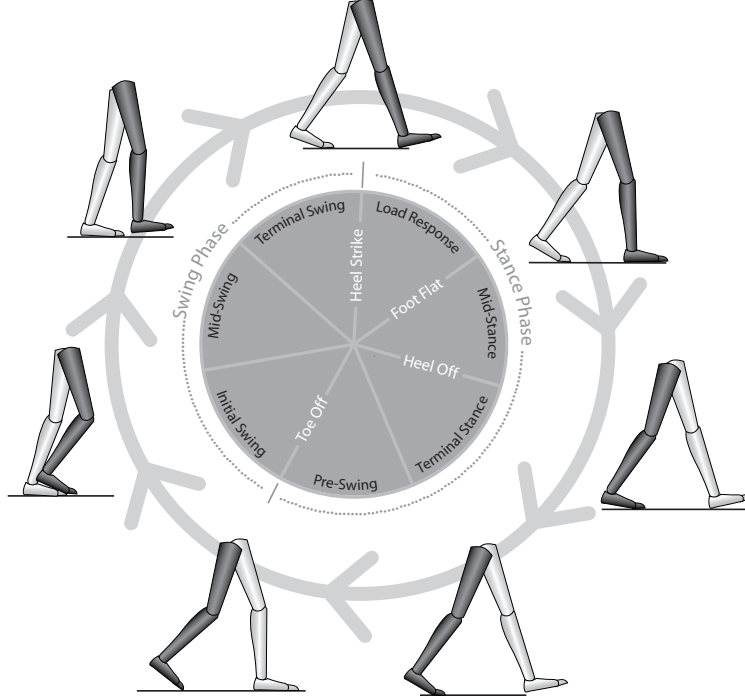
Figure 1.2: Different phases during gait on a normalized time scale modified from Whittle[4]

and *foot flat* are characterized by a rapid loading of the limb. During the *double support phase* both feet have ground contact and the walker is most stable. In the following *mid stance* phase, the body is moved forward and the opposite limb is in the swing phase. This is a position where the walker is least stable due to the small base of support and the relatively high center of gravity. The event *heel off*, where ground contact of the heel is lost, indicate the transition from the *mid stance* phase to the *terminal stance* phase. During *terminal stance*, the body is propelled forward until the *pre-swing* phase starts. This propulsive movement causes the final *toe off* event where the contact between toes and floor is lost and the *swing* phase begins. During the swing phase the swinging limb moves in front of the stance limb leading to a forward progression. The *swing* phase itself is divided into the sub phases of *initial swing* where the limb is accelerated forward, the *mid swing* phase in which the limb passes the opposite stance limb, and the *terminal swing* phase where the limb is decelerated in preparation for *heel strike*, which will terminate the *swing* phase.

Gait analysis Gait analysis refers to the assessment of walking including the medical history of the patient, the objective examination, and the impact of a disease on the walking pattern [4]. The physical methods to analyze gait are based on *kinematics* and *kinetics*. The term *kinematics* is used to describe movement not taking into account the forces that cause the movement. One approach to measure movement is the direct measurement of linear and angular displacements provided by joint angles, limb velocities or accelerations. A second established method is the indirect measurement of movement with cameras or tracking systems. Such systems capture the trajectory of markers attached to the body during a movement, and determine hereby the desired quantities by inverse kinematics. The term *kinetic* describes the study of forces and moments that cause a movement. Those are for example gravitational, ground reaction, other external forces, or forces produced by muscle contractions. By analyzing the kinetics and kinematic during the gait cycle, various disabilities can be assessed and the results used to setup the gait rehabilitation.

Pathological gait Healthy gait, as a set of measurement parameters is in a certain range of variability defined as normal. Any deviation outside the normal range is considered as abnormal or pathological gait [4]. Pathological gait affects the four walking tasks: support of the body weight during single leg stance, maintaining of balance during single leg stance, leg coordination during swing, and muscle contraction to take over the body weight [16, 17]. The pattern of gait is a result of complex interaction within the neuromuscular system and may change after an injury affecting the brain, spinal cord, nerves or muscles. In case of stroke, this leads typically to reduced posture and balance control. Muscle weakness of the affected side and compensatory muscle activation of the non-affected side result furthermore in an asymmetric walking pattern at a lower speed and with a lower step-length compared to healthy gait [18]. Further, spasticity modulates the indented movement and hinders the forward propulsion during the swing phase [19].

1.4.2 Inertial sensors

An inertial sensor is a device that uses the inertia of a test-mass to perform a measurement. Common known sensors are accelerometers to measure the acceleration of an object and gyroscopes to measure the angular velocity of a rotating object. The sensors

available today evolved from a purely mechanical system to the technology of MEMS (Micro-Electro-Mechanical Systems) which allows the development of miniature, low power, and low cost sensors.

Accelerometer The physical mechanisms underlying MEMS accelerometers are based on a miniature mass-spring system described by Hook's law:

$$F = kx \tag{1.1}$$

with the spring konstant k and the displacement x, and Newton's second law of motion:

$$F = ma \tag{1.2}$$

the acceleration can be expressed as:

$$a = kx/m \tag{1.3}$$

During motion, the sensing element moves with its inertia relative to a fixed base inside the part, which can be detected and transformed into an electric signal (fig.:1.3a). Common physical mechanisms to determine x are based on capacitive or piezo-resistive sensing. For slow accelerations the signal represents the vectorial component of the gravity vector in the sensing axis.

Gyroscopes Gyroscopes provide measurements of angular displacement. The measurement principle utilizes the Coriolis force which is the response to the rotation of a moving mass.:

$$F_c = -2m \cdot (\nu \times \omega) \tag{1.4}$$

where m is the mass, ν the vibration frequency of the moving mass and ω the angular velocity acting on the moving mass (fig.:1.3b). Due to a rotational movement of the object the sensor is attached, the vibration frequency is modulated by the induced Coriolis force F_c and converted into an output signal. This principle requires continuos energy to keep the mass vibrating.

(a) Mass-spring system of an accelerometer
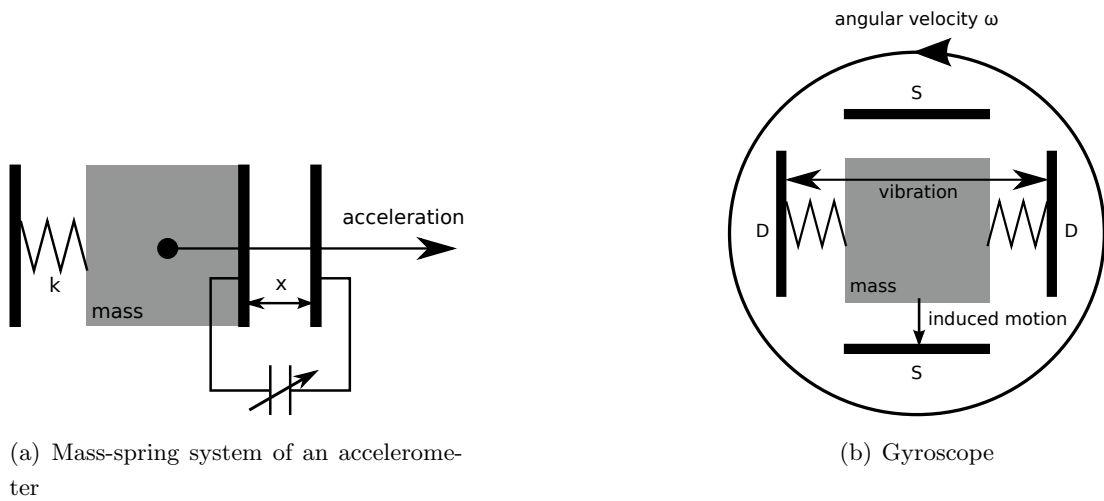
(b) Gyroscope

Figure 1.3: Functional structure of inertial sensors

2

Methodology

2.1 Study I - prior art

2.1.1 Motivation

The use of alternative sensors to foot switches, like accelerometers and/or gyroscopes require additional signal processing. Accelerometers are affected by the influence of gravity and further, a drift problem may occur with integration of accelerometer or gyroscope signals due to noise [20]. The attachment of sensors is another source of imprecision due to movement of muscles during walking [20]. However, past research has shown that heel off is detectable even with poorly defined heel contact [21], or during shuffling gait [22].

Study I was motivated by the fact that no prior reviews on the detection of gait characteristics in relation to rehabilitation systems existed. In particular, systems intended for use outside the lab had never been systematically reviewed.

2.1.2 Methods

Study I systematically reviewed all measurement systems based on the used sensor types and their associated algorithms. Further, the study covered methods for monitoring gait and detecting gait events, that could be used in an ambulatory rehabilitation system. In particular, the study discusses the aspects of sensor setup, signal processing, and evaluation of the performance. For this purpose, a meta analysis of research studies on measurement methods, matching one or more of the following search keys was performed: *gait kinematics, gait detection, gait analysis, gait events, locomotion,*

ambulatory measurements, wearable sensors in combination with: *force sensitive resistors, accelerometer, gyroscope.* Subsequently the identified systems were categorized according their used sensor types.

2.2 Measurement system

2.2.1 Motivation

Despite the commercial availability of ambulatory movement analysis systems (e.g. Xsens, Biometrics, TMSI, Noraxon) they are not commonly used in clinical environments. A major problem of such systems is the extensive amount of data they provide and the difficulty to distinguish between measures which are important and those which are not [4] combined with the effort of donning and doffing. Therefore, the decision was made to develop a system dedicated to be used for automatic gait event detection with a minimal set of sensors and minimal energy consumption.

2.2.2 Methods

An incremental development process was used to develop the gait detection system. After the requirement analysis followed the conception phase leading to a first prototype covering the core functionalities. In a proof of principle study the sensor system was verified [23] to investigate the signals obtained at the foot. The prototype was further refined until the final hardware design (Fig.2.2), which was manufactured. In a next development increment, the system was extended to record signals of foot, shank and thigh simultaneously and used to analyze healthy and hemiparetic gait in Study II [2]. Based on the findings in Study II [2] the algorithm 2.4 was developed.

2.2.2.1 Requirement analysis

The functional requirements were extracted out of interviews, records of focus group workshops performed by Neurodan A/S, and experiences of researchers stated in their publications, which where reviewed (see [23]). The following terms were used to signify the requirements [24]:

- SHALL: means that the definition is an absolute requirement of the specification.

- SHOULD: means that there may exist valid reasons in particular circumstances to ignore a particular item, but the full implications must be understood and carefully weighed before choosing a different course.

- MAY: means that an item is truly optional.

General requirements:

- The system SHALL provide a configurable output to interface to a FES device.

- The system SHOULD support to be don and doff one handed by the user to allow home-care use.

- The system SHOULD provide a 24h usage.

- The system SHOULD be able to be used outside clinical environments.

- The system MAY provide an integration into the FES device.

- The system MAY support different fixation positions.

- The system MAY support a cosmetically acceptable housing design.

Specific requirements to gait event detection

- The system SHALL be robust to initiation and shutdown of stimulation.

- The system SHALL be robust to weight transfer or changes during stance phase, standing or sitting to prevent false triggers.

- The system SHALL detect heel off/strike or a biomechanical equivalent.

- The system SHOULD detect early stance phase.

- The system SHOULD detect swing phase.

- The system SHOULD be capable of detecting gait events on various surfaces (such as cobblestone, grass, sand, carpet)

- The system SHOULD be capable to detect gait events during up/down hill walking.

- The system SHOULD be capable to detect gait events during stairs ascending/descending.

- The system SHOULD be capable to detect gait events independent of the users footwear.

- The system SHOULD be capable to detect gait events of various pathological gait patterns (such as stiff knee, or circumduction of the hip).

2.2.2.2 Concept

The measurement principle is the measurement of rotational components of body segment accelerations, since those are the result of muscular control, rotating a body segment around a joint axis. The design of differential acceleration measures was chosen to compensate for the impact of gravity (g) and due to the fact that two 3D MEMS accelerometers consume less energy compared to one 3D gyroscope (example: 3D digital accelerometer type ADXL344 140μ, 1D gyroscope ADXRS453 6mA, see measurement principle 1.4.2). The two sensors were mounted separated by a distance d on a rigid bar, which is in an ideal case, in-line with the radius vector of the body segment. Both sensors are affected in the same way by the environmental acceleration of translation and gravitation. Figure 2.1 illustrates the measurement principle including the definitions. Non angular accelerations are eliminated by vectorial subtraction of the signals based on:

$$\text{a}_{\text{T}1} = \alpha r_1 \tag{2.1}$$

and

$$\text{a}_{\text{T}2} = \alpha r_2 \tag{2.2}$$

can be calculated to:

$$\Delta \text{a}_{\text{T}} = \text{a}_{\text{T}1} - \text{a}_{\text{T}2} = \alpha d \tag{2.3}$$

where d is the sensor distance:

$$d = r_1 - r_2 \tag{2.4}$$

By this, the angular acceleration α is independent of the radius i.e. the measurement position or segment length . This applies also for the radial acceleration given by:

$$\Delta a = a_1 + g + c - (a_2 + g + c) = a_1 - a_2$$
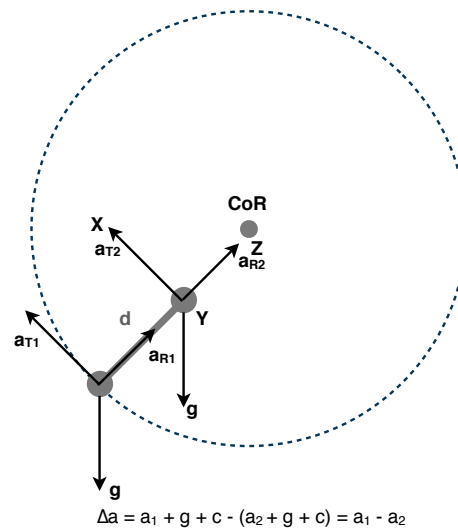
Figure 2.1: Measurement principle: Two 3D accelerometers are placed at a distance of 50 mm and provide radial and tangential accelerations by subtraction. (a_T: tangential acceleration, a_R: radial acceleration, CoR: center of rotation, g: gravity vector, c: constant, representing translational acceleration, a_1 and a_2 surrogate the measurement channels X, Y and Z of the accelerometers 1 and 2

$$a_{R1} = \omega r_1 \tag{2.5}$$

Furthermore, the radial acceleration a_R is measured by the x component (pointing along the segment axis), which allows to calculate the instantaneous angular velocity ω.

$$\omega = \sqrt{\frac{\Delta a_R}{\Delta r}} \tag{2.6}$$

The measurement of α and ω are the base of further gait parameters estimation as described in the literature [25, 26, 27] and can further be extended to 2D vectors since a_T can be seen as a composition of acceleration in the y and z directions (sagittal and frontal planes). By this, three dimensional rotational movements of body segments were acquired.

2.2.3 Hardware

The hardware design (fig.2.2) was realized after the following functional block diagram (fig.2.3) as printed circuit board (PCB). Optional digital input pins were implemented in the foot unit and connected to custom made foil switches [28]. A rechargeable Li-Ion button cell with 120mAh allowed an operating time of approximately 7 hours. The communication with the external devices (radio controller and accelerometer) was realized by both SPI (serial peripheral interface) interfaces of the micro controller, since the accelerometers and the radio controller needed different SPI settings.
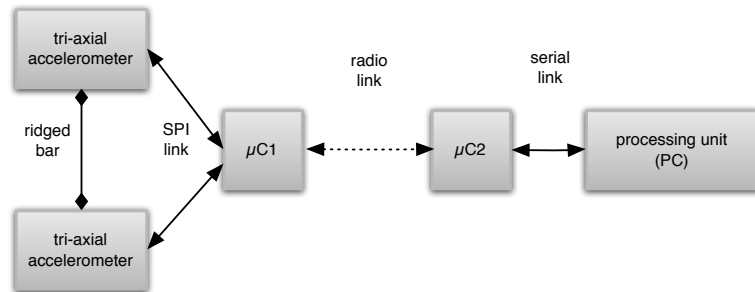
Figure 2.2: Final design comprising 2 accelerometers and interface for further processing of the data.
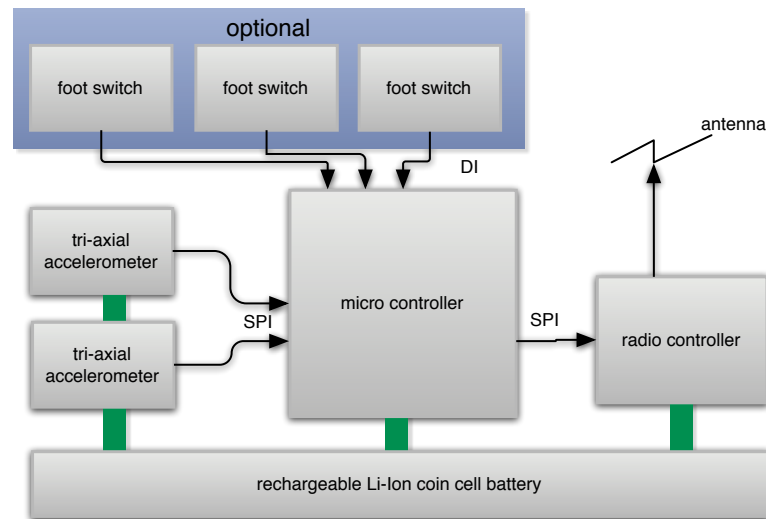
Figure 2.3: Functional block diagram of used components

2.2.4 Software

The software was created following the agile software development paradigm to allow adaptive planning and an evolutionary development based on experiences which suits best to a research based development. The architecture of the distributed system required several software parts, which are shown in the deployment diagram (fig: 2.4). Several sensor units communicate wirelessly with the receiver unit. The sensor and receiver devices share a common package radio and SPI which provides a set of functions to drive the radio chip and the SPI communication. The package Initialization is tailored to the specific device, and includes additional commands to initialize the accelerometers. The software of the sensor units contains a package (AccSensor) which provides functions to drive the accelerometers. The serialCom package of the receiver unit provides functions for the serial communication with the PC application.

The software for the sensor and receiver units was programmed in C, using the Code-Composer Studio of Texas Instruments [29]. A personal computer was used to collect the signals. Therefore, a Java application was created, which realized basic functions similar to a tape recorder and provided an online visualization of the measured signal. Java was chosen to enable a platform independent software reuse for future projects.
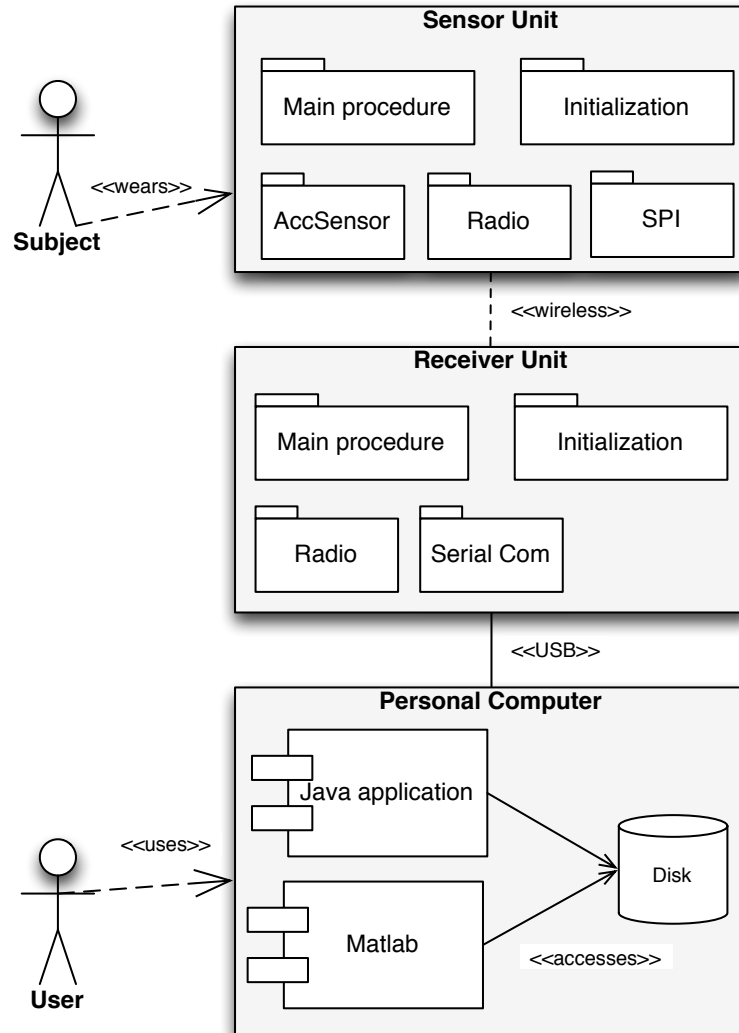
Figure 2.4: Software deployment

2.2.5 Verification

The sensor system was verified during the entire development process on different functional levels by unit test, integration tests, and whole system tests in a pilot study [23].

2.3 Study II - Characterization of gait pattern by 3D angular accelerations in hemiparetic and healthy gait

2.3.1 Motivation

The study was performed to investigate the gait kinematic characteristics of the target group in order to identify signal features, which would provide a foundation for the subsequent development of an automatic gait detection algorithm (section 2.4). The new measurement system was used to record angular accelerations of the foot, shank, and thigh of healthy and hemiparetic individuals, with the aim of obtaining information about the different gait phases. Further, the influence of sensor position and walking cadence on the signal was investigated and since past research mainly analyzed gait in the sagittal plane, the study additionally evaluated the benefit of including frontal accelerations as well.

2.3.2 Methodological considerations

While designing the experimental procedure it was empirically found that 70 steps/min was the slowest cadence that resulted in a natural gait. A slower cadence required more attention and might have caused an unnatural gait pattern. Therefore, 70 steps/min were chosen to compare healthy and hemiparetic gait, as they showed comparable gait cycle times ([2]).

Past studies analyzed self-selected gait velocities [25, 30, 31, 9], or used fixed gait velocities set by a treatmill [32, 8] over and under the prefered velocity [25]. However, in order to control gait velocity during an experiment, it must be measured and feed back to the subject in order to be able to adjust.

As the gait velocity had a direct impact on the measurement signal, it needed to be a controlled variable for comparing healthy and hemiplegic gait. A metronome was used to control the cadance. However, even if the gait cycle appeared to be synchronous, i.e. initial contact happened seemingly at the same time as the sound of the metronome, it did not guarantee a low inter-step variability. Some subjects modified the duration of their stance phase, while other prolonged or aborted their swing phase. To reduce this effect, we chose a 20m walking path and three trials per cadence and further analyzed mean accelerations of all steps for the specific gait phases. In the experimental design, the ages of the subjects did not match, despite it is known that age in the

absence of pathological gait affects the gait pattern. Elderly gait is reported as a slower version of healthy gait, affecting the temporal parameters, and not to be considered as pathological gait [33]. By comparing conditions with similar cadence, the timing effects were minimized.

2.4 Study III - Detection algorithm

2.4.1 Motivation

The problem of gait detection is to develop algorithms that determine gait events while the person is walking, based on data obtained from sensor measurements. Out of the literature review (see [1]), two main approaches for the development of an automatic gait detection algorithm were identified: Functional analysis of the measurement signal and machine learning techniques. Both approaches implement a set of rules that identify certain characteristics of the gait measurements. The purpose for this study was to develop a rule based algorithm based on radial and tangential accelerations of the foot, since the gait event features were mostly expressed in signals of the foot (see [2]) and the rule based algorithm needs less computational power.

2.4.2 Methodical considerations

The algorithm in [3] realized a hand-crafted, rule-based state machine. This approach was preferred against machine learning due to the following reasons: first, the aspiration was to develop a single sensor system, that fits to all users, without the need for a specific parameterization as required for the training of machine learning algorithms. Further, the identified signal features were present in the vectorial sum of the acceleration vector, which is a single, uniform signal. Those signals are inappropriate as input for machine learning as it was identified by a test with the inductive learning algorithm presented in [34]. The algorithmic approach was therefore to translate the identified signal features into mathematical expressions based on low level processed data.

3

Results and discussion

The overall aim of this PhD study was to overcome the problem of accurate and reliable ambulatory gait cycle detection for over ground walking under the perspective of a daily application in a rehabilitation device.

3.1 Study I - prior art - aim A

The first study, the literature review, revealed the current possibilities to detect gait events with an ambulatory system. The main finding was that the possibilities for detecting gait are manifold and offer a wide range from simple to complex solutions [1]. Various sensors and sensor combinations were capable to provide physical signals relevant for the analysis of gait. Accelerometers, [35, 36, 37, 38, 21, 39, 20] tended to be the most used sensors, often used in combination with gyroscopes [40, 41, 42, 31, 9, 43, 44, 45]. Furthermore, sensor positioning seemed less critical as placing the sensor at nearly any combination of foot, shank, thigh, and trunk of one or both legs were possible with appropriate signal processing. Several solutions were already part of FES systems for foot drop correction [22, 46, 21, 47, 44, 48, 49], while others were used solely for online or offline gait analysis.

An improved sensor device should be minimally sensitive to placement, small and light-weight and no wires should be needed in order to be worn comfortably and be cosmetically accepted. Light-weight coheres strongly with energy efficiency in battery powered devices. The fixation of the sensor devices should be done in a way that facilitates easy donning and doffing, including a certain tolerance to the exact position.

Further, the sensor system should ultimately provide the possibility to walk barefoot with the device, since this is part of life quality and perhaps culture [46]. However, the most important requirement for a successful use in an ambulatory rehabilitation system is a sufficient reliability in the detection of gait events during daily use. Detecting the trigger event with a reliability between 70 and 90% is insufficient for practical applications [48].

3.2 Measurement system - Aim B

The engineering work of this thesis was the development of a measurement system towards the fulfillment of the general requirements, which will be discussed now and specific system requirements, which will be discussed later with the results of the detection algorithm.2.2.2.1.

The concept of differential acceleration measurement was applied in previous research. Liu et. al. [50] used differential accelerations of the lower limb segments to estimate the rotational angles. They developed a sensor system based on triaxial accelerometers to obtain the pitch and yaw angles of the thigh segment with an accelerometer approximating the translational acceleration of the hip joint and two accelerometers measuring the actual accelerations on the thigh. The system showed a correlation larger than 0.99 to the camera based reference system and was wearable[50], but not wireless. Djuric et. al. [51] used differential acceleration measurements and additional bandpass filtering to estimate segment angles during walking. Despite showing high correlation coefficients 0.85-0.97 [51] for the estimated angles, the technique for filtering the signals required a periodic signal, i.e. several consecutive steps to estimate the joint angles. However, in order to detect gait events with those systems [51], several sensor devices are needed, since the angle from a single segment provides too little information about the gait cycle and might interfere with other non-walking tasks.

Modern micro-controllers and sensors are advertised as 'low power' devices [52, 53, 52, 54], however this is relative. Gyroscopes tend to consume up to several milliamperes, while accelerometers are in the range of a few micro-amperes (example: 3D digital accelerometer type ADXL344 140μ, 1D gyroscope ADXRS453 6mA, see measurement principle 1.4.2). Micro controllers and other peripherals like radio chips easily add double digit milliampere to the current consumption. Therefore, the hardware design

has a great impact on usability and user acceptance and is to be chosen to use as few components as possible.

The result is a prototype of the measurement system providing a wireless interface to a computer or FES system. A full integration into a FES system requires only an extension of the stimulator software. The housing of the sensor devices was in this early phase not designed to allow a one handed donning and doffing, however common techniques for arm watches might be applicable for different fixation positions. As the device is small in size (60x60x10 mm, fig.: 3.1) an adequate housing of the device should be possible to meet the requirement for a clinical usage and cosmetically acceptance. The usage time of the system is currently limited to 7...8 hours. A usage of 24h is technically possible: in the simplest way by increasing the battery capacity or by a further development of power saving strategies as the main power consumption is due to the radio communication. In the current setup, the device monitors the movement and transmits data packages at 40Hz, resulting in an average power consumption of 38mW. This might not be necessary if only gait events as state changes are transmitted.
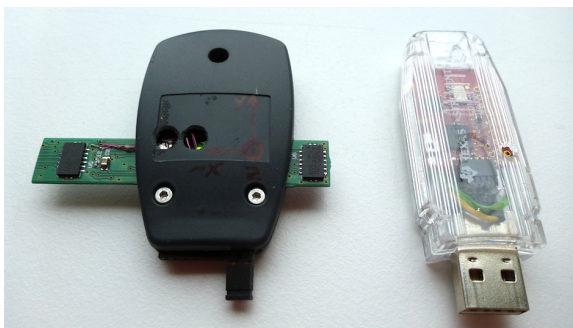
Figure 3.1: Prototype: left, measurement unit, right: USB-Receiver

3.3   Study 2 - Aim C

Aim C was to improve the understanding of impaired gait patterns affected by hemiparesis based on 3-dimensional lower limb kinematics. The kinematic measures consisted of angular accelerations of the foot, shank and thigh.

Study II [2] showed that angular accelerations were characteristically modulated across the gait cycle. The largest signal was observed in measurements at the foot,

3. RESULTS AND DISCUSSION

where the curve of healthy gait was characterized by a large increase during the loading-response after heel-strike, which was visible in the signals of the foot, shank, and thigh. During mid-stance, the angular acceleration of foot and shank was close to zero, i.e. identical to the residuals. The pre-swing phase, after heel-off, was characterized by an increasing angular acceleration leading to a local maxima when the transition into the swing phase took place. The swing phase was visible as a local minima at the foot and shank, while the signal of the thigh showed a local maxima.

Several previous studies ([1]) have shown the matching of curve features to gait events based on kinematic sensor data. However, only a few studies [46, 55, 49] used solely accelerometer data to investigate foot contact events of healthy and impaired gait. Study II showed further that an increasing walking cadence resulted in larger amplitudes of the angular acceleration and that the swing phase is most sensitive to a change of the gait cadence. This is in coherence with reported measurements of the angular velocity measured by Lau et al. [9] who used accelerometers in combination with gyroscopes.

Hemiparesis modulated the limb kinematics of the affected side in different ways. During the loading-response, the mean value of the angular acceleration was lower in the hemiparetic gait while pre-swing and swing showed no significant difference between healthy and hemiparetic gait for all measurement positions. During mid-stance, differences were observed within the measurement positions shank and thigh. This might be due to reduced weight bearing and weak push off capabilities of hemiparetic subjects[16]. Further, hemiparetic gait showed sharp acceleration peaks, caused by limited body control resulting in a jerky movement and further ground contact during swing leading to additional acceleration peaks as well as irregular gait phase timings. These relate to the acceleration values presented in [2], where the results of the present study show lower values of median, minimum and maximum accelerations for the hemiparetic gait compared to the healthy gait. The weight of the hemiparetic leg might affect the acceleration values, as it is known that lean muscle mass is rapidly lost after stroke; however, in case the subject relearned walking, no significant changes in lean muscle mass have been observed [56]. Therefore, it is unlikely that reduced muscle mass is responsible for the lower acceleration values recorded in the present study. The energy cost of hemiparetic gait showed lower kinetic energy at toe-off and

a higher kinetic energy during swing compared to non-disabled controls [57]. This correlates to the findings in study II [2], since the angular accelerations of the foot in the healthy gait were higher during pre-swing than during the swing phase, while the hemiparetic gait was characterized by lower angular acceleration during pre-swing than during the swing phase. Commonly, modern acceleration sensors provide measurement in three dimensions. However, traditionally only the two dimensions in the sagittal plane were used for gait analysis in the past [1]. Study II [2] showed that the magnitude of the tangential angular accelerations in the frontal plane was in the same order as the tangential accelerations in the sagittal plane suggesting that tangential frontal accelerations contribute also to the characterization of gait events by angular accelerations. This was further shown by the statistical analysis, which showed no significant difference within the frontal and tangential accelerations. Including accelerations in the frontal plane provided additional information which is of a great value for the relevant overall clinical assessment [58].

3.4   Study III - Gait detection algorithm - Aim D

The research Aim D addressed the questions whether it is possible to detect gait events in real-time in angular accelerations and further how reliable, accurate and precise the detection is. The detection rate was except for one subject above 100%, which indicated that the algorithm over performed [3]. Specificity, as determined for both, healthy and hemiparetic gait, was between 99.65 and 99.91% indicating that false detections resulting in a state transition rarely occurred. The best detection sensitivity, 100%, was achieved in detecting pre-swing and swing in both, healthy and hemiparetic gait [3]. The detection timings of gait phase transitions were biased towards an earlier detection compared to the reference system (Bland-Altmann analysis). Precision values and the limits of agreement indicated large variations in the timing of gait events.

The detection of hemiparetic gait was mainly limited by severe gait disorders prohibiting independent and rhythmic walking [3]. The largest detection errors where observed in subjects who had a hobbling or shuffling gait pattern, which resulted in limited foot contact information, leading to mistakingly detected state transitions. The broad timing variation ranging from early to delayed detections in hemiparetic gait is owed to two reasons: first, due to jerky movements causing an early detection, and

second, by not detecting the ordinary condition, i.e. the curve maxima during loading response. If this happens, the algorithm remains in loading response until the recovery condition 'isMStance' is true [3], which takes place later in time.

Previous research showed sensor approaches, such as 3D acceleration and the angular rate measurements of the shank in the sagittal plane obtained from the XSens [59] system, that realized a state machine consisting of swing, stance, and push-off with a detection rate of 100% for healthy gait, where the transition between states were defined by thresholding of the input data in relation to a baseline [60]. Lee Jung-Ah et. al. [32] applied a peak detection algorithm on the vectorial sum of 3D acceleration measures of the shank and were able to detect initial swing and initial contact events with a detection accuracy of 99%. Lee Jung Keun et. al. [61] detected initial contact and end contact events by functional analysis, i.e. peak and local minima detection of the angular velocity of the shank, and were able to detect 100% of all events. Pappas et. al. [8] realized a state machine and defined state transitions by handcrafted rules depending on a gyroscope and FSR signals and reached an overall reliability of 96% for subjects with impaired gait. The method of machine learning has shown that detection rules derived from training data can have a performance superior to that of a human expert [62, 21] and are able to distinguish between different walking conditions [9]. However, it remains unclear if and how often such a system would need to be retrained during therapy or chronic use.

An advantage of the presented algorithm is that it did not require a specific start condition, as reported in [61] , where the signal peak during swing was used to detect the initial contact. Further the positioning of the proposed sensor system required only an alignment parallel to the metatarsus of the foot segment providing a flexible positioning, and allowing barefoot walking which is a limitation of insole based systems [8].

A disadvantage of the current system is the tendency of the algorithm to over-perform, i.e. the detection of false and premature transitions. However, despite foot switches are the gold standard to measure foot contact events, there are not completely trustworthy [9] and might bias the results. Further, the results showed a broad variation of the transition timings. An adaptive filter design and additional suppression rules might improve this.

4

Conclusions and future work

This thesis presented a new measurement system to detect gait events in healthy and hemiparetic gait to be used in the future as part of rehabilitation technology such as FES. Along, an overview over the latest research carried out on methods of gait analysis and event detection in relation to an ambulatory use was provided. A new measurement system based on differential acceleration was used to analyze signal characteristics of healthy and hemiparetic gait and provided the foundation for the subsequent development of an automatic detection algorithm.

The first part of this work addressed the prior art in ambulatory gait measurement systems. A literature review compiled information about sensor configurations, measurement positions, and detection methods under the perspective of a possible use within a gait rehabilitation system. This implied the consideration of daily activities such as shopping or trips, and the changing conditions of the environment like ground surface and temperature, and conditions of the user like footwear, changing physical condition and weight.

The second part comprised research and development of a new measurement system based on radial and tangential accelerations. The accelerations were calculated by differential acceleration measures from 2 equidistant MEMS accelerometers. The system comprised a receiver unit establishing a wireless link to up to 3 sensor units attached to foot, shank, and thigh. The receiver unit was connected to a computer, and the received data streamed to a file for later offline analysis.

The third part addressed whether gait events and phases could be recognized in the radial and tangential acceleration signals by means of foot contact events. The

characteristic gait pattern in the angular accelerations of the leg segments were present in both healthy subjects and hemiparetic individuals and provided the basis for a gait detection algorithm.

The fourth and final aim of this thesis was to develop an algorithm for the real-time detection of gait events. The algorithm realized a hand-crafted, rule-based state machine. The rules extracted signal features based on thresholds, inflection and turning points of raw and filtered signals corresponding to the gait phase transitions. The algorithm was tested offline with data from 10 healthy and 10 hemiparetic subjects.

Future research might cover further refinement of the algorithm and the evaluation of the detection capability under ambulatory conditions, including non walking tasks. Although further research is necessary before the presented technology can be validated in a FES system, this thesis has demonstrated the feasibility of the concept of angular acceleration to detect gait events and may open new possibilities for ambulatory gait analysis.

Bibliography

[1] J. Rueterbories, E. G. Spaich, B. Larsen, O. K. Andersen, Methods for gait event detection and analysis in ambulatory systems, Medical Engineering & Physics 32 (6) (2010) 545 – 552. `doi:10.1016/j.medengphy.2010.03.007`. i, 2, 4, 20, 21, 24, 25

[2] J. Rueterbories, E. G. Spaich, O. K. Andersen, Characterization of gait pattern by 3d angular accelerations in hemiparetic and healthy gait, Gait & Posture`doi:10.1016/j.gaitpost.2012.06.029`. i, 12, 19, 20, 23, 24, 25

[3] J. Rueterbories, E. G. Spaich, O. K. Andersen, Gait event detection for use in fes rehabilitation by 3d differential angular accelerations, not published yet. i, 20, 25, 26

[4] M. W. Whittle, Gait Analysis: An Introduction, Butterworth-Heinemann, 2006. ix, 6, 7, 12

[5] G. M. Lyons, T. S. r, J. H. Burridge, D. J. Wilcox, A review of protable fes-based neural orthoses for the correction of drop foot, IEEE Transactions on Neural Systems and Rehabilitation Engineering 10 (2002) 260 – 279. `doi:10.1109/TNSRE.2002.806832`. 1

[6] W. Liberson, H. Holmquest, D. Scot, M.Dow, Functional electrotherapy: stimulation of the peroneal nerve synchronized with the swing phase of the gait of hemiplegic patients, Archives of physical medicine and rehabilitation 42 (1961) 101–105. 1

[7] M. Trew, T. Everett, Human Movement - An Introductory Text, Elsevier Ltd, 2005. 1, 5

BIBLIOGRAPHY

[8] I. Pappas, T. Keller, S. Mangold, M. R. Popovic, V. Dietz, M. Morari, A reliable gyroscope-based gait-phase detection sensor embedded in a shoe insole, IEEE Journal of Sensors 4 (2) (2004) 268–274. `doi:10.1109/JSEN.2004.823671`. 2, 19, 26

[9] H. Lau, K. Tong, The reliability of using accelerometers and gyroscopes for gait event identification on persons with dropped foot, Gait & Posture 27 (2008) 248–257. `doi:10.1016/j.gaitpost.2007.03.018`. 2, 19, 21, 24, 26

[10] R. Baker, Gait analysis methods in rehabilitation, Journal of NeuroEngineering and Rehabilitation 3 (1) (2006) 4. `doi:10.1186/1743-0003-3-4`. 4, 5

[11] W. Weber, E. Weber, Mechanik der menschlichen Gehwerkzeuge: eine anatomisch-physiologische Untersuchung, Dieterich, 1836. 4

[12] J. Paul, History and fundamentals of gait analysis, Bio-Medical Materials and engineering 8 (1998) 123–135. 4

[13] E. Marey, De la locomotion terrestre chez les bipèdes et les quadrupèdes, Impr. E. Martinet, 1873. 4, 5

[14] G. Carlet, Essai experimental sur la locomotion humaine: Etude de la marche, Annales des Science Naturelles, 1872. 5

[15] J. Perry, Gait analysis: normal and pathological function, Slack, USA, 1992. 5

[16] S. J. Olney, C. L. Richards, Hemiparetic gait following stroke. part i: Characteristics, Gait & Posture 4 (1996) 136–148. `doi:10.1016/0966-6362(96)01063-6`. 7, 24

[17] C. L. Richards, S. J. Olney, Hemiparetic gait following stroke. part ii: Re- covery and physical therapy, Gait & Posture 4 (1996) 149–162. `doi:10.1016/0966-6362(96)01064-8`. 7

[18] A.-L. Hsu, P.-F. Tang, M.-H. Jan, Analysis of impairments influencing gait velocity and asymmetry of hemiplegic patients after mild to moderate stroke, Archives of Physical Medicine and Rehabilitation 84 (8) (2003) 1185 – 1193. `doi:10.1016/S0003-9993(03)00030-3`. 7

[19] D. Winter, The biomechanics and motor control of human gait: normal, elderly and pathological, University of Waterloo Press, 1991. 7

[20] A. T. M. Willemsen, J. van Alst, H. B. K. Boom, Real-time gait assessment utilizing a new way of accelerometry, Journal of Biomechanics 23 (8) (1990) 859–863. `doi:10.1016/0021-9290(90)90033-Y`. 11, 21

[21] R. Williamson, B. J. Andrews, Gait event detection for fes using acceleromenters and supervised machine learning, IEEE Transactions on Rehabilitation Engineering 8 (3) (2000) 312–319. `doi:10.1109/86.867873`. 11, 21, 26

[22] A. Mansfield, G. M. Lyons, The use of accelerometry to detect heel contact events for use as a sensor in fes based assisted walking, Medical Engineering and Physics 25 (2003) 879–885. `doi:10.1016/S1350-4533(03)00116-4`. 11, 21

[23] J. Rueterbories, E. G. Spaich, O. K. Andersen, Representation of gait events in differential angular accelerations of the foot, in: Abstracts of the XVIII Congress of the International Society of Electrophysiology and Kinesiology, ISEK 2010, 16-19 June 2010, Aalborg, Denmark [CD-ROM], Department of Health Science and Technology. Aalborg University., 2010. 12, 18

[24] S. Bradner, Key words for use in rfcs to indicate requirement levels., Internet Engineering Task Force: http://www.ietf.org/rfc/rfc2119.txt.
URL `http://www.ietf.org/rfc/rfc2119.txt` 12

[25] K. Aminian, B. Najafi, C. Bla, P. F. Lezvraz, P. Robert, Spatio-temporal parameters of gait measured by an ambulatory system using miniature gyroscopes, Journal of Biomechanics 35 (2002) 689–699. `doi:10.1016/S0021-9290(02)00008-8`. 16, 19

[26] A. T. M. Willemsen, C. Friego, H. B. K. Boom, Lower extremity angle measurement with accelerometers - errors and senistivity analysis, IEEE Transactions on Biomedical Engineering 38 (12) (1991) 1186–1193. `doi:10.1109/10.137284`. 16

[27] D. Giansanti, G. Maccioni, V. Macellari, The development and test of a device for the reconstruction of 3-d position and orientation by means of a kinematic sensor assembly with rate gyroscopes and accelerometers, IEEE Transcations on

Biomedical Engineering 52 (2005) 1271–1277. `doi:10.1109/TBME.2005.847404.` 16

[28] Neurodan a/s.
URL `http://www.neurodan.dk` 16

[29] T. Instruments, Code composer studio, integrated development environment.
URL `http://www.ti.com/tool/ccstudio` 17

[30] M. Hanlon, R. Andersona, Real-time gait event detection using wearable sensors, Gait & Posture 24 (2009) 127–128. `doi:10.1016/j.gaitpost.2006.11.088.` 19

[31] J. M. Jasiewicz, J. H. Allum, J. W. Middleton, A. Barriskill, P. Condie, B. Purcell, R. C. T. Li, Gait event detection using linear accelerometers or angular velocity transducers in able-bodied and spinal-cord injured individuals, Gait & Posture 24 (4) (2006) 502–509. `doi:10.1016/j.gaitpost.2005.12.017.` 19, 21

[32] J.-A. Lee, S.-H. Cho, Y.-J. Lee, H.-K. Yang, J.-W. Lee, Portable activity monitoring system for temporal parameters of gait cycles, J. Med. Syst. 34 (2010) 959–966. `doi:10.1007/s10916-009-9311-8.` 19, 26

[33] M. P. Murray, R. C. Kory, B. H. Clarkson, Walking patterns in healthy old men, Journal of Gerontology 24 (2) (1969) 169–178. 20

[34] S. Dosen, D. Popovic, Rule-based control of walking by using decision trees and practical sensors, in: Neural Network Applications in Electrical Engineering, 2008. NEUREL 2008. 9th Symposium on, 2008, pp. 125–128. `doi:10.1109/NEUREL.2008.4685587.` 20

[35] J. J. Kavanagh, H. B. Menz, Accelerometry: A technique for quantifying movement patterns during walking, Gait & Posture 28 (2008) 1–15. `doi:doi:10.1016/j.gaitpost.2007.10.010.` 21

[36] K. M. Culhane, M. O'Connor, D. Lyons, G. M. Lyons, Accelerometers in rehabilitation medicine for older adults, Age and Ageing 34 (2005) 556–560. `doi:10.1093/ageing/afi192.` 21

[37] P. E. Crago, H. J. Chizeck, M. R. Neuman, F. T. Hambrecht, Sensors for use with functional neuromuscular stimualtion, IEEE Transactions on Biomedical Engineering 33 (2) (1986) 256–268. `doi:10.1109/TBME.1986.325808`. 21

[38] W. Y. Wong, M. S. Wong, K. H. Lo, Clinical applications of sensors for human posture and movement analysis: A review, Prosthetics and Orthotics International 31 (1) (2007) 62–75. `doi:10.1080/03093640600983949`. 21

[39] J. R. W. Morris, Accelerometry - a technique for the measurement of human body movements, Journal of Biomechanics 6 (1973) 729–736. `doi:10.1016/j.medengphy.2004.11.006`. 21

[40] R. E. Mayagoitia, A. V. Nene, P. H. Veltink, Accelerometer and rate gyroscope measurement of kinematics: an inexpensive alternative to optical motion analysis systems, Journal of Biomechanics 35 (4) (2002) 537–542. `doi:10.1016/S0021-9290(01)00231-7`. 21

[41] H. Dejnabadi, B. M. Jolles, K. Aminian, A new approach to accurate measurement of uniaxial joint angles based on a combination of accelerometers and gyroscopes, IEEE Transactions on Biomedical Engineering 52 (2005) 1478–1484. `doi:10.1109/TBME.2005.851475`. 21

[42] A. M. Sabatini, C. Martelloni, S. Scapellato, F. Cavallo, Assessment of walking features from foot inertial sensing, IEEE Transactions on Biomedical Engineering 52 (2005) 487–494. `doi:10.1109/TBME.2004.840727`. 21

[43] G. Wu, Z. Ladin, The study of kinematic transients in locomotion using the integrated kinematic sensor, IEEE Transactions on Rehabilitation Engineering 4 (1996) 193–200. `doi:10.1109/86.536775`. 21

[44] P. H. Veltink, P. Slycke, J. Hemssems, R. Buschman, G. Bultstra, H. Hermens, Three dimensional inertial sensing of foot movements for automatic tuning of a two-channel implantable drop-foot stimulator, Medical Engineering and Physics 25 (1) (2003) 21–28. `doi:10.1016/S1350-4533(02)00041-3`. 21

[45] R. Williamson, B. J. Andrews, Sensor system for lower limb functional electrical stimulation (fes) control, Medical Engineering and Physics 22 (2000) 313–325. `doi:doi:10.1016/S1350-4533(00)00038-2`. 21

BIBLIOGRAPHY

[46] Y. Shimada, S. Ando, T. Matsunaga, A. Misawa, T. Aizawa, T. Shirahata, E. Itoi, Clinical application of acceleration sensor to detect the swing phase of stroke gait in functional electrical stimulation, The Tohoku journal of experimental medicine 207 (2005) 197–202. `doi:10.1111/j.1525-1403.2006.00074.x`. 21, 22, 24

[47] M. D. Duric, S. Dosen, M. B. Popovic, D. Popovic, Sensors for assistive system for restoring of the walking, in: Proceedings 52nd ETRAN Conference, Society for Electronics, Telecommunications, Computers, Automatic Control and Nuclear Engineering, 8-12 June 2008, Palic, Serbia, Proceedings 52nd ETRAN Conference, Society for Electronics, Telecommunications, Computers, Automatic Control and Nuclear Engineering, 8-12 June 2008, Palic, Serbia, 2008, p. 4. 21

[48] I. P. Pappas, M. R. Popovic, T. Keller, V. Dietz, M. Morari, A reliable gait phase detection system, IEEE Transactions on Neural Systems and Rehabilitation Engineering 9 (2001) 113 –125. `doi:10.1109/7333.928571`. 21, 22

[49] D. J. Weber, B. Stein, K. M. Chan, G. Loeb, F. Richmond, R. Rolf, K. James, S. L. Chong, Bionic walkaide for correcting foot drop, IEEE Transactions on Neural Systems and Rehabilitation Engineering 13 (2005) 242–246. `doi:10.1109/TNSRE.2005.847385`. 21, 24

[50] K. Liu, T. Liu, K. Shibata, Y. Inoue, R. Zheng, Novel approach to ambulatory assessment of human segmental orientation on a wearable sensor system, Journal of Biomechanics 42 (16) (2009) 2747 – 2752. `doi:10.1016/j.jbiomech.2009.08.008`. 22

[51] M. D. Djuric-Jovicic, N. S. Jovicic, D. B. Popovic, D., Kinematics of gait: New method for angle estimation based on accelerometers, Sensors 11 (11) (2011) 10571–10585. `doi:10.3390/s111110571`. 22

[52] St mircoelectronics.
URL `www.st.com` 22

[53] Texas instruments.
URL `www.ti.com` 22

[54] Analog devices.
URL `www.analog.com` 22

[55] R. Dai, R. B. Stein, B. J. . Andrews, K. B. James., M. Wiele, Application of tilt sensors in functional electrical stimulation, IEEE Transactions on Rehabilitation Engineering 4 (1996) 63–72. `doi:10.1109/86.506403`. 24

[56] L. Jorgensen, B. Jacobsen, Changes in muscle mass, fat mass, and bone mineral content in the legs after stroke: a 1 year prospective study, Bone 28 (6) (2001) 655–659. 24

[57] G. Chen, C. Patten, D. H. Kothari, F. E. Zajac, Gait differences between individuals with post-stroke hemiparesis and non-disabled controls at matched speeds, Gait & Posture 22 (2005) 51–56. `doi:10.1016/j.gaitpost.2004.06.009`. 25

[58] M. G. Benedetti, M. Manca, G. Ferraresi, M. Boschi, A. Leardini, A new protocol for 3d assessment of foot during gait: Application on patients with equinovarus foot, Clinical Biomechanics In Press, Corrected Proof (2011) –. `doi:10.1016/j.clinbiomech.2011.06.002`. 25

[59] Xsens.
URL `http://www.xsens.com/` 26

[60] D. Kotiadis, H. Hermens, P. Veltink, Inertial gait phase detection for control of a drop foot stimulator: Inertial sensing for gait phase detection, Medical Engineering & Physics 32 (4) (2010) 287 – 297. `doi:10.1016/j.medengphy.2009.10.014`. 26

[61] J. Lee, E. Park, Quasi real-time gait event detection using shank-attached gyroscopes, Medical and Biological Engineering and Computing (2011) 1–6`doi:10.1007/s11517-011-0736-0`. 26

[62] C. A. Kirkwood, B. Andrews, P. Mowforth, Automatic detection of gait events: a case study using inductive leasrning techniques, Journal of Biomedical Engineering 11 (1998) 511–516. 26