Universidade do Minho
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ASBGo: A Smart Walker for mobility Assistance and monitoring System Aid

PhD thesis in Biomedical Engineering

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Abstract

Locomotion is an important human faculty that affects an individual's life, bringing implications not only in social and personal development but also in the aspect of employment. Thus, it becomes necessary to find means and tools to improve or help to restore and increase the mobility of the affected people, so they can recover their independence.

For several years, researchers have been addressing the needs of persons with mobility disabilities through alternative, e.g. wheelchairs, or augmentative devices, e.g. canes, and walkers. Among augmentative devices, walkers play an important role, due to the large number of potential users, its simplicity and ambulatory potential. They were designed to improve pathological gait, through a support base for the upper limbs that improves the balance of the individuals and reduces the load on their lower limbs.

Over the past years, technological advances allowed the incorporation of sensors and actuators in conventional walkers. A new class of devices, the smart walkers, emerged to provide a better stability, without affecting the resultant naturalness of the users' gait patterns.

In this context, this thesis aims to develop a smart walker (SmartW) for mobility assistance in hospitals and clinics for people with balance problems. This work is structured in five stages as follows.

A complete survey regarding the current state-of-art of walker-based studies is presented. The advances in the walkers' and SmartWs' have been enormous and have shown a great potential. Thus, it is presented a review of the available literature of walkers and SmartW and it is discussed major advances that have been made and limitations to be overcome.

Then, it is presented the design specifications of the proposed SmartW based in an "end-user" approach, i.e. concerns and needs of end-users are the main focus for these specifications.

Functionalities is the next stage where different information gathered by several built-in sensors is used to characterize the assisted human gait and the interaction user-walker. Thus, three sensory systems are developed: (i) a system that captures the relatives evolutions between the lower limbs of the user and the walker as well as the trunk, giving information related to gait pattern and stability for further clinical evaluation; (ii) an intuitive interface for direct acquisition of navigation commands; and (iii) sensory systems to ensure the user safety

during assisted gait, by identifying situations like the fall of the user.

These systems are validated by performing real experiments with healthy and pathological users. This data is converted into gait parameters to be used in the evaluation of the user. Results are used in the assistive-movement analysis.

This latter stage turns possible the evaluation and finding of important characteristics that define the main effects of the use of a walker in comparison to other assistive devices.

Finally, a clinical validation with patients with ataxia are performed. Results enabled the clinicians to differentiate gait deviations objectively and evaluate quantitatively the evolution of the rehabilitation process of these patients. This also validated the use of the SmartW as a diagnostic tool that enables clinicians to monitor the medical conditions of their patients.

The proposed research is relevant because introduces a new concept in terms of rehabilitation, since the SW may effectively work as a rehabilitation tool, by monitoring objectively the patients' motor state.

In the future, the proposed SmartW will serve not only as a measure of a treatment outcome, but also as a useful tool in planning ongoing care for various gait disorders. It is noteworthy that currently there is no device in the national market that offers the same possibilities as the one proposed.

Resumo

Locomoção é uma importante faculdade humana que afeta a vida de um indivíduo, trazendo implicações não só no desenvolvimento social e pessoal, mas também no aspecto do emprego. Assim, torna-se necessário encontrar meios e ferramentas para melhorar ou ajudar a restaurar e aumentar a mobilidade das pessoas afetadas, para que eles possam recuperar a sua independência.

Durante vários anos, investigadores têm endereçado as necessidades de pessoas incapacitadas através de dispositivos alternativos (cadeiras de rodas) ou aumentativos (bengalas e andarilhos). Entre os dispositivos aumentativos, os andarilhos têm um papel importante devido ao grande número de potencial utilizadores, à sua simpliciade e potencial ambulatório. Foram construídos de maneira a melhorar a marcha patológica, através de uma base de suporte para os membros superiores, que melhora o equilíbrio dos indivíduos e reduz a carga nos seus membros inferiores.

Nos últimos anos, os avanços tecnológicos permitiram a incorporação de sensores e atuadores nos andarilhos convencionais. Uma classe nova de dispositivos, os andarilhos inteligentes (AI), emergiu para fornecer melhor estabilidade, sem afetar a naturalidade do padrão de marcha dos seus utilizadores.

Neste contexto, esta tese tem como objetivo desenvolver um AI para dar assistência à marcha em hospitais e clínicas a pessoas com problemas de equilíbrio. Este trabalho está dividido em cinco fases, descritas em seguida.

Uma pesquisa completa acerca do estado de arte de estudos sobre andarilhos é apresentada. Os avanços na área dos andarilhos convencionais e AI tem sido grande e tem mostrado um enorme potencial. Deste modo, é apresentado uma revisão de literatura sobre estes dispositivos onde são discutidos os avanços alcançados bem como as limitações a ultrapassar.

Depois, são apresentadas as especificações do design do AI proposto, sendo baseadas numa abordagem "utilizador-final", ou seja, as preocupações e necessidades dos utilizadores finais são o principal foco das especificações.

A próxima fase baseia-se no desenvolvimento de funcionalidades, onde é adquirida informação através de vários sensores integrados no AI. Esta informação é usada para caracterizar a marcha humana e a interação utilizador-máquina. Três sistemas foram criados: (i) um sis-

tema que captura a evolução relativa dos membros inferiores do utilizador ao andarilho, bem como o movimento do tronco, dando informação relativa do padrão de marcha e estabilidade do utilizador para futura avaliação clínica; (ii) uma interface intuitiva para aquisição direta dos comandos de navegação; e (iii) um sistema sensorial que assegura a segurança do utilizador durante a marcha assistida, identificando situações de perigo, tais como a queda do utilizador.

Estes sistemas foram validados em experiências reais com indivíduos saudáveis e patológicos. Os dados são convertidos em parâmetros de marcha que serão usados na avaliação do utilizador. Os resultados são usados na análise de movimento assistido.

Esta última fase torna possível a avaliação e procura de características importantes que definem os efeitos dos andarilhos em comparação com outros dispositivos de assistência.

Finalmente, é feita uma validação clínica com pacientes atáxicos. Os resultados permitiram aos clínicos diferenciar os efeitos da marcha objetivamente e avaliar quantitativamente a evolução dos pacientes. Foi também possível validar o uso do AI como uma ferramenta de diagnóstico que permite aos clínicos monitorizar o estado dos pacientes.

Este trabalho é relevante pois introduz um novo conceito em termos de reabilitação, dado que o AI funciona como uma ferramenta de reabilitação.

No futuro, o AI proposto será não só usado para tratamento, mas também como ferramenta útil no planeamento de cuidados continuados para vários distúrbios de marcha. É de evidenciar que correntemente não existe nenhum dispositivo no mercado nacional que ofereça as mesmas possibilidades que o dispositivo aqui desenvolvido.

Contents

Chapter 1

Introduction

This thesis presents the work developed during the past four years in the Industrial Electronics (IE), Control, Automation and Robotics (CAR) group of the Department of Industrial Electronics, Algoritmi Center from University of Minho.

This work addresses the field of rehabilitation robotics based on the development of a smart walker and its funcionalities. The ultimate goal of this work is to enable a technological breakthrough in the field of human pathological gait assistance, by providing functional rehabilitation with higher safety and quality. The developed smart walker device can provide an alternative to mobility, with an eventual recovery and enhancement of the residual capacity of the user's movement.

1.1 Motivations, scope and problem statement

Stability in ambulation is fundamental to independent activity and quality of life. However, among the disabled population, such stability is impaired, bringing implications not only in social and personal development but also in the aspect of employment [1]. These individuals present widely different and heterogeneous functional profiles, like impairments that range from extremely moderate (people able to walk with a cane) to extremely severe (persons bedridden). In the first case, we are in the domain of prevention, functional compensation and rehabilitation, and, in the second case, palliative care or institutionalization.

People in the middle stage of impairment are prone to fall injuries and show a dependence on caregiver's attention and support from society [1, 2]. As in most cases the fall-injuries cannot be treated medically or surgically, they are expected to have the best results and improvements through the use of proper assistive technology and are the target of the assistive intervention, such as rehabilitation. In this sense, emerging therapies are necessary to help and improve the quality of life of these individuals. Some diseases have no current solution in terms of physiotherapy, and because of that, some patients have to choose the wheelchair as their locomotion device.

Rehabilitation involves the management of disorders that alter the motor abilities and performance of patients. It is a combination of medication, physical manipulation, therapeutic exercises and adaptation to technical aids. Thus, it involves assistive intervention that is based on assitive devices that can improve balance control by providing mechanical advantages and help to reduce lower-limbs loading and thereby alleviate joint pain [3]. In cases like the elderlies, the assistive devices can be used as a functional compensation tool to compensate for weakness or injury and reduce fatigue in their daily routine in home or in the care centre [4].

For that purpose, new devices for assisted mobility appeared by acting as augmentative devices. In intermediate situations the individuals have residual capacity of locomotion, so they can use augmentative devices, represented by orthoses, crutches, walkers, and others. These devices contribute to the improvement of physical and cognitive abilities of the user. The goal is to avoid the inappropriate use of wheelchairs that may present a negative effect on locomotion, leading to lower limb muscle atrophy [5].

One of most common used augmentative device is the walker. Walkers have the potential to improve mobility by decreasing weight bearing on one or both lower limbs. Using walkers may also compensate for weakness or impaired motor control of the lower limb, and can increase user's confidence and feeling of safety, which may raise their levels of activity and independence. It is very likely that walkers can increase stability and improve balance by providing an increased base of support [6].

However, some considerations have to be taken into account, since walkers are not suitable for every patient. Walkers are more appropriate for patients with low/medium balance problems. Individuals with motor disability with high imbalance and coordination problems, such as ataxic gait caused by tumours or stroke, cannot use walkers due to many problems. The light weight of the walker (mainly the four-legged walker) provides unstable and fragile support to them. In addition, it does not provide a natural gait pattern to its users, forcing them to stop in order to place it correctly on the floor, and then continue walking. Also walkers lack of adaptation to the user's different necessities in terms of handling and stabilization. There is also the problem with the rollator (walker with four wheels) that requires strenght and good coordination capabilities to be maneuvered, transmiting a sense of instability to its users. Moreover, studies have identified other several factors explaining the correlation between walkers and risk of fall due to destabilizing biomechanical effects, interference with limb movement during balance recovery, and metabolic and physiological demands.

Thus, there are people that despite having residual capacities of locomotion, prefer to use wheelchairs, since they do not feel safe with a walker, or any other augmentative device.

In addition, it is a very high manpower demand on the healthcare professionals in nursing home to maintain a reasonable mobility for the needed patients that require assistive devices, like walkers. Since we are facing a future aging society, nursing and home care problems will become more severe and demand urgent attention [7].

To improve or even prove their efficiency, the influence of walkers on gait needs to be studied, explored and quantified. By examining which variations in the design can influence clinical and functional features of a certain disease, one may pave the way for further research in other specific diseases and more generic rehabilitation populations.

Thus, it is intended in this thesis to give the first steps in directing the walkers' research on standard methodology that includes assessment and evaluation of gait patterns of patients while they are being assisted by a walker. A preliminary approach will be proposed in order to conceptualize and improve the investigation and developments of walkers, in terms of design and effectiveness in the rehabilitation and functional compensation programs.

Following this argument, another problem that appears is the non-standardization of clinical guidelines for prescribing walkers and other assistive devices that often depends on the clinicians experience. Selecting an appropriate walker should depend on objective assessments of a person's functional requirements and physical capabilities. Also, individual care and evaluation is necessary. There is still little research about these methodological considerations [8]. Therefore, in this thesis, a preliminary proposal to classify the funtional features that should be taken in high consideration before prescribing a walking aid will be presented. It will also be verified if individual evaluation is needed when prescribing an assistive device, i.e., do not generalize the individuals and group them by disorder/disease, since each individual has his/her own characteristics and problems.

The general opinion of the researchers is that active devices can free medical staff of demanding tasks and give them the possibility to improve their concentration and help on patients, improving the quality and duration of the rehabilitation exercises [4, 9]. Therefore, technological aids can reduce the number of persons around the patient, provide the execution of repetitive basic movements, assist the patient in therapeutic movements, grid movements to be as natural as possible, keep control of therapeutic movements, develop new rehabilitation protocols, and bring a gait and posture evaluation thanks to robot sensors acquisition.

Thus, over the past years, technological advances allowed the improvement of walkers [4, 9]. Due to problems with the design and stability of walkers, robotics have been investigated in the last few years as a mean to change and augment walkers in order to provide higher stability, postural stability and safety to its users.

Thus, smart walkers appeared providing the stability of four-legged walkers, without affecting the naturalness of gait patterns of the users, assistive navigation systems with sensors that detect obstacles, help in auto-localization and provide an easy manoeuvrability[4, 9]. Through the existence of a smart interface, the maneuver of these devices does not require a great cognitve, physical and metabolic effort to the user, since it reads user's interaction information and transforms this knowledge into motor commands. Smart walkers also present the potential to be introduced in an acute phase of rehabilitation, in which the patient needs to get out of bed and give his/her first steps.

Despite existing many smart walkers in the state-of-art [4, 9], this research area still presents many problems.

One of the problems that investigators should pay more attention is the lack of "end-users" research with smart walkers. There is a high concern in developing complex control strategies with high cost interfaces, that often are not user-friendly and do not work with walker-users. The goals and research questions of smart walkers' projects tend to do not direct their attention to the patients' real needs. The following open questions remain: What does a walker-user needs in order to improve the patient's recovery? Which functionalities are important to them? How can a smart walker help efficiently a physical therapist?

Research should aim to transform the smart walkers into multi-funtional devices that provide different functionalities, either mechanical/structural or electronic, for the therapy of their patients. Such funtionalities should provide different options for the physical therapists to work with their patients, with more quality, objective and optimal recovery results, depending on the problem of the patient. This will decrease the burden required in a conventional therapy, both mental and physical. Such goal will be studied in this thesis.

Also, there should exist a tool, incorporated in the smart walker, to objectively evaluate, monitor and diagnose the patient's treatment, since nowadays, such process is made subjectively or through clinical scales. These tools will enable the clinicians to differentiate gait deviations objectively, and thus check the effectiveness of the rehabilitation treatments and validate the clinical benefits of the smart walker as a rehabilitation tool. This point is crutial to improve the quality of physiotherapy with smart walkers and is included in the goals of this thesis.

It this context, the present thesis aims to develop a smart walker that provides different functionalities to their users and physical therapists. The developement of such functionalities will be focused on the end-users and validated with them. Also, a gait and posture assessment tool will be developed and tested in order to capture the relative evolution between the lower limbs of the user and the walker, as well as posture and balance evaluation. The device of this thesis, in its final state, will be a simple, intuitive and multi-functional system with the capability of being used by patients with high balance disorders.

1.2 Overview of the research

In the state-of-art [4, 9], walkers play an important role on giving balance and partial weight support to its users. However, there is still the necessity to better understand which are the real walker-user's needs; which are the specific effects of the interaction user-walker, and which are the benefits that conventional and smart walkers can bring to the gait and posture of the patient regarding other assistive devices, to improve the aimed results on the rehabilitation process. More specifically in this thesis, there is a special concern in the potential of four-wheeled walkers with forearm supports. Until this date there are no studies focusing on them.

The author of this thesis, in previous work [10], built a four wheeled smart walker with two rear wheels driven by motors. This smart walker had a joystick, as a smart interface, placed in the upper base (handlebar) whose signals were treated and processed into user's command intentions. A fuzzy control system converted this knowledge onto the required motor commands to drive the walker. However, despite achieving good results with this smart walker, there are concerns and improvements which have to be tackled that rise several multi-disciplinary challenges in terms of scientific contributions, design implementation and robotic methodologies. For instance, it was not performed a detailed description and analysis of the main parameters that describe the user-walker interaction (spatiotemporal parameters, for example). Additionally, the design was not ergonomic and not concerned with the user confort and support.

In order to address these advantages, this thesis is composed by five main stages of research: survey, design, functionalities, assistive-movement analysis and clinical validation.

The first stage is one of the most important in this thesis, being related to the survey of the walker's research field. In the begining of this thesis, there was no complete survey about studies that focus on evaluating the walker's efficiency and their influence on gait. Moreover, there was also no survey about the many potential advantages of smart walkers and their functionalities. Such review is essencial to organize ideas and present to the scientific community which are the main advances in this field. So, it is crucial to present and discuss the current state of this research area and this will be done in this thesis.

After evaluating all the state-of-art about walker, the next stage, design, must be carried out. It aims developing a smart walker for mobility assistance in hospitals and indoor houses for people with balance problems. A necessary first step is the improvement and re-design of a smart walker capable of supporting the weight of the user with high stability. This device will be driven through a smart interface based in low cost sensors and simple strategies. Also, the smart walker should be safe to drive with high manoeuvrability. For instance, a user-friendly handle bar design must be addressed and it must be verified the best way to dispose the base support of the user's upper limbs, in order to improve the manoeuvrability of the walker. Thus, there is a great concern in the end-user of the smart walker in order to improve his/her confort and rehabilitation recovery.

At the end of this stage, a new smart walker with an ergonomic and economic design will be presented.

The next stage, funcionalities, is concerned with the functionalities that a smart walker should contain to answer the walker-users' needs as well as decrease the burden of the physical therapists. In this thesis, the developed smart walker will be integrated with four operation modes (autonomous, manual, security and remote control) and will be turned into a measurement tool for evaluating the walker's user gait.

This latter funcionality will be handled in detail. A system based on an active depth sensor and a laser range finder that captures the relative evolutions between the lower limbs of the user and the walker will be developed. It is intended to characterize the gait by extracting the gait features related to the users' gait patterns. Posture will also be explored in detail, to infer the balance problems of the walker users. An accelerometer sensor will be used for such task. Thus, new algorithms will be explored to give as outcomes a set of spatiotemporal and postural stability parameters.

At the end of this stage, it will be possible to characterize the signals gathered by the different sensors, by performing an exhaustive analysis of the main features involved in the user's gait pattern.

Different assessments will be considered at different points, such as, the way the user's lower limbs and center of mass move. These assessments lead this thesis to the next stage, the assistive-movement analysis, which puts the previous developed systems in practice.

This stage is based on two great concerns: assessment and evaluation of gait patterns of patients while they are being assisted by a walker and the correct prescription of walking aids to a specific patient. Thus, it investigates which are the gait and posture stability parameters that should be considered when deciding what is the most suitable device for the recovery of a given patient. For this, a group of patients (recovering from total knee arthroplasty) was selected to study which are the specific effects of the user-walker interaction and the benefits that walkers with forearms can bring to the gait and posture of the patient regarding other walking aids. Due to the type of studied disorder, uni-lateral disfunction, a special focus will be given to inter-limb symmetry and postural stability measurements. The study and characterization of the assisted-movement was done with different assistive devices, including a four wheeled walker with forearm supports.

To help in the decision making of the prescription of a given assistive device, it was proposed a complete gait classification approach that verifies differences in the acquired walking patterns. Multivariate statistical analysis was employed to determine relationships between the features, the patients and the assistive devices. Clinicians were consulted to help in the interpretation of such relationships.

Results from this stage were decisive in the definition of different patterns of performance of different users with different assistive devices and can be used to modify and personalize the rehabilitation program accordingly to the user disorder.

The final stage is based on the validation of the clinical benefits of the smart walker's as a rehabilitation and functional compensation tool. It is intended to objectively validate the use of the developed smart walker as a diagnostic tool that will enable clinicians to monitor the medical conditions of their patients. For this a group of patients with ataxia were selected and evaluated through months, giving the possibility to evaluate the long-term effects of a smart walker in a rehabilitation program.

This proposed research is relevant because introduces a new concept in terms of rehabilitation and prevention of risks with the use of a smart walker and a four-wheeled walker, both with forearm supports. Additionally, provides means to monitor the patients' motor state in hospitals and clinics. Finally, an evaluation of the clinical benefits of the usage of the walker and smart walker both with forearm supports will be possible.

The motivation is that this will contribute towards better rehabilitation purposes by promoting ambulatory daily exercises and thus extend users' independent living.

1.2.1 Goals and research questions

The ultimate goal of this thesis is the development of a robot for mobility assistance that it is easy to drive with a smart interface and capable of monitoring and evaluate gait characteristics. The smart walker will be based on a conventional four-wheeled device and will be driven by the movement intentions of the user, online interpreted by a user-walker interface for command and control. Human gait with the developed device will be studied and characterized in order to provide assisted locomotion adapted to the user's needs and to short- and long-term changes in mobility capacities and modifications on gait patterns.

To achieve this, it is necessary to achieve the following goals:

Goal 1: To conduct two extensive surveys on the state of the art related with walkers. The first survey will be a systematic review that focus on the effects and influence of the walker device on its user's gait, routine, and behavior. The second survey will review the importance of smart walkers in maintaining mobility, will discuss their potential in rehabilitation and their demands as robotic assistive devices.

This goal will unify the current goals and limitations of the reported studies and highlights the main topics that should be present on walker-related studies.

Chapter 2 addresses this first goal, which is related with the first stage (survey) of this thesis. This chapter led to one conference publication [11] and two journal publications [4, 9].

Goal 2: To develop and assembly a smart walker platform based on a conventional four-wheeled walker with special attention on end-users's concerns. Different operation modes and functionalities will be integrated on the proposed smart walker in order to turn it into a adaptative and versatile device that is capable to answer to the different user's needs.

This goal will allow a better understanding of the main concerns that a smart walker design should address.

Chapter 3 addresses the second goal of this dissertation, which is related with the second and third stages (design and functionalities) of this thesis. This chapter led to six conference publications [12–17]and one journal publication [18].

Goal 3: To develop a gait assessment system capable of different assessments at different points as follows: the pattern followed by feet and legs and the characteristics of body balance of a walker user.

This goal will give the necessary information to spatiotemporal, posture stability and fall risk estimation.

Chapter 4 addresses this third goal, which is related with the third stage (functionalities) of this thesis. This chapter led to five conference publications [19–23] and one journal publication [24].

Goal 4: To address additional safety issues, to detect possible falls of the user or other mishaps that can arise while the user is guiding the walker and to deal with the possibility that the walker may roll away from the user. Henceforth, additional sensor systems including infrared and force sensors will be included onto the system.

This goal will turn possible to predict danger situations for the user and as such stop in time the walker.

Chapter 4 addresses this forth goal, which is related with the third stage (functionalities) of this thesis. This chapter led to a conference publication [14].

Goal 5: To characterize the assisted human gait, through the processing of the information obtained by the gait assessment systems. Gait characteristics data presents high-dimensionality and therefore there is a critical need for data reduction and to determine which parameters actually contain useful information within a specific clinical context. Multivariate analysis approaches and multi-classification will be addressed in order to assess parameters related with patient's gait in assisted gait with three different ADs: crutches, standard walker and rollator with forearm supports.

This goal will allow to identify differences and similarities in gait performance between three different assistive devices and to understand how gait patterns of patients recovering from total knee arthroplasty differ from person to person and how they are influenced by the type of device that is prescribed during their recovery. Such understanding might help in physical therapy for improving the recovery of such patients.

Chapter 5 addresses this fifth goal, which is related with the fourth stage (assistive-movement analysis) of this thesis. This chapter led to two conference publications [25, 26] and four journal publications [27–30].

Goal 6: To clinically validate the smart walker through experiments with specific users (patients with ataxia) in a rehabilitation program.

This goal will verify if the proposed smart walker may be prescribed as a rehabilitation tool to correct specific gait disorders.

Chapter 6 addresses this sixth goal, which is related with the fifth stage (clinical validation) of this thesis. This chapter led to one conference publication [31] and four poster presentations in medical conferences [32–35].

Goal 7: To establish new quantitative measures for the assessment of the progression of ataxic patients' gait. The main expected result is to obtain quantitative information that objectively indicate the functional motor recovery of the patients based on their assisted gait performance (spatiotemporal data), balance, posture and symmetry.

Chapter 6 addresses this sixth goal, which is related with the fifth stage (clinical validation) of this thesis.

Therefore, the current thesis aims to allow in the near future the use of the proposed smart waler in physiatric treatment of patients with high balance disorders requiring gait training, in order to contribute to gaining functional autonomy and improving the quality of life of these patients.

The following research questions (RQ) are expected to be answered:

RQ1: How do walkers with forearm supports influence and modify the walking gait pattern and posture of their users, in comparison with crutches and standard walkers?

RQ2: Is there a need to individualize the gait pattern evaluation in order to prescribe a walking aid?

RQ3: Can the smart walker with forearms be prescribed as a rehabilitation tool to correct specific gait disorders (ataxia)?

RQ4: Which parameters are important to evaluate and diagnose the recovery of a patient with balance that is performing gait training with a smart walker?

1.3 Contributions to knowledge

This thesis approaches five distinct but complementary studies: survey about the advantages and disadvantages of the conventional walkers and smart walkers; design and assembly of a smart walker; development of different functionalities for such device with implementation of a gait and posture assessment tool integrated in the smart walker; analysis of the assisted-movement with walker with forearm supports and other assistive devices; and finally validation of the developed smart walker in the reabilitation of patients with ataxia.

In the overall, this thesis developed a new robotic motorized walker for gait assistance, which is missing either in the national and international market. The new design, functionalities and assessment tool can provide several advantages to its users, mainly to people with high balance disorders: (1) The motorization allows systematic control and progression of the speed at which walking is performed; (2) The repetitive training of a complete gait cycle with user involvement enables a more appropriate gait pattern; (3) Allows the physical therapist to provide manual assistance during assisted gait to help the patient simulate a more normal walking pattern; (4) The patient performs double tasking, training his/her coordination and cognition; (5) Walk in a real environment; (6) Provides the physical therapist with quantative clinical information; (7) Provides axial support, with the forearm supports, decreasing tremor and dysmetria, which is very important for patients with ataxia, parkinson, etc; (8) Allows the physical therapist to challenge the patient with different velocities and directions; (9) The device can be used in acute stages, allowing the patient to start early (regarding conventional therapy) his/her gait training; (10) Decreases the burden of physical therapist work with a safety system that monitors the patient's state, by predicting danger situations for the user.

The following statements point out the main contributions of this work:

- Presentation of systematic review that unifies the current goals and limitations of the studies that focused on walker investigation. It also highlights them so that the investigation on this area can be more focused on the walker users' complaints, disorders and the design limitations of the walkers, as well as the necessities of the physicians (described in Chapter 1).

- Presentation of a descriptive review that highlights major advances that have been made and limitations to be overcome in smart walker's field, in terms of gait analysis and user-machine interface (described in Chapter 1).

- New smart walker design developed with the participation of physicians, physical therapists and end-users, achieving an ergnomic and comfortable model that meets the needs of its users (described in Chapter 3).

- Development of a smart interface based on a mechanic handlebar adapted to the user: with position adjustment of the handles and lateral and rotational course. Economic and simple solution with a versatil and adaptative potential, being suitable for different patients (described in Chapter 3).

- Implementation of four different operating modes that allow the physiotherapist to choose the most appropriate one for the actual type of difficulty of the patient (described in Chapter 3).

- Development of an innovative methodology to analyse precisely the status of a patient gait. The proposed device is integrated with sensors that evaluate, in real-time, the progress of the patient in terms of spatiotemporal and postural stability parameters and safety. This information is then analyzed to follow the evolution of the patient and helps on deciding when the patient should leave the smart walker, to go to the next stage of treatment (described in Chapter 4 and 6).

- Development of an innovate methodology that applies a feature reduction method, based on multivariate analysis techniques, to identify the differences and common features in evaluating the gait performances by using three different assistive devices (described in Chapter 5).

- An understanding about how gait patterns of post-surgical patients differ from person to person and how they are influenced by the type of device that is prescribed during their recovery might help in physical therapy (described in Chapter 5).

- Findings concluding that inter-limb symmetry and postural stability features can be evaluated in an outpatient setting, supplying important additional information about individual gait pattern, which is not represented by gait velocity, cadence and scales usually used (described in Chapter 5).

- The features calculated in this study are able to provide complementary information to gait velocity, cadence and clinical scales to assess the functional capacity of patients that passed through total knee arthroplasty (TKA). The selected parameters make a new clinical tool useful for tracking the evolution of patients' recovery after TKA (described in Chapter 5).

- Findings concluding the potential of the developed smart walker and its long-term benefits in ataxia rehabilitation treatment (described in Chapter 6). Despite the clinical consensus in relation to the usefulness of physical therapy exercises in patients with ataxia, documentation of the effects of different protocols on functional performance of such subjects is scarce in the scientific literature. In these patients the potential of using a smart walker was showed to be promising. The combination of impaired balance and incoordination in lower-limb dynamics of ataxic patients suggests a strong rationale for the use the proposed smart walker for gait training.

- Findings from this thesis provided possible support for research demonstrating the importance of cerebellar structures in gait adaptation and in practice-dependent motor learning. The correct intensity and duration of gait training using smart walker, or other therapies, required to achieve functional gains is not known nor standardized. However, with the clinical validation presented in this thesis, the preliminary steps were given towards the determination of the optimal intensity and duration of gait training to optimize functional walking outcomes following cerebellar pathology (described in Chapter 6).

The resulting product is very significant throught a medical perspective and its use results in clinical benefit based on a qualitative analysis. Such validation is itself innovative and does not exist today.

In the future, it will serve not only as a measure of a treatment outcome, but also as a useful tool in planning ongoing care for various gait disorders. The validation of the clinical benefits of the walker‘s as a rehabilitation and functional compensation tool validates the use of the smart walker also as a diagnostic tool that will enable clinicians to monitor the medical conditions of their patients.

1.4 Publications

The work here described allowed the publication of the following journal articles, conference papers and poster communications.

Journal Articles

- M. Martins, L. Costa, A. Frizera, and C. P. Santos, “Feature reduction with pca/kpca for gait classification with different assistive devices,” International Journal of Intelligent Computing and Cybernetics, vol. 8, no. 4, 2015.

- M. Martins, L. Costa, A. Frizera, and C. P. Santos, “Feature reduction and multiclassification of different assistive devices according to the gait pattern,” Disability and Rehabilitation: Assistive Technology, 2015.

- M. Martins, C. P. Santos, and A. Frizera, “A review of the functionalities of smart walkers,” Medical Engineering and Physics, pp. 1–12, 2015.
- A. Tereso, M. Martins, and C. P. Santos, “Evaluation of gait performance of knee osteoarthritis patients after total knee arthroplasty with different assistive devices,” Research on Biomedical Engineering, 2015.
- M. Martins, A. Elias, C. Cifuentes, M. Alfonso, A. Frizera, C. P. Santos, and R. Ceres, “Assessment of walker-assisted gait based on principal component analysis and wireless inertial sensors,” Research on Biomedical Engineering, vol. 30, no. 3, pp. 1–12, 2014.
- M. Martins, L. Costa, C. P. Santos, A. Frizera, and R. Ceres, “Hybridization between multi-objective genetic algorithm and support vector machine for feature selection in walker-assisted gait,” Computer Methods and Programs in Biomedicine, vol. 113, no. 3, pp. 736–748, 2014.
- M. Martins, C. P. Santos, A. Frizera, and R. Ceres, “Real time control of the asbgo walker through a physical human-robot interface,” Measurement, vol. 48, pp. 77–86, 2014.
- M. Martins, C. P. Santos, A. Frizera, and R. Ceres, “Assistive mobility devices focusing on smart walkers: Classification and review,” Robotics and Autonomous Systems, vol. 6, pp. 548–562, 2012.

Conference Papers

- M. Martins, T. Pereira, A. Matias, M. Cotter, F. Pereira, A. Frizera, and C. P. Santos, “Smart walker use for Ataxia’s rehabilitation: Case study,” in Proceedings of the 2015 IEEE International Conference on Rehabilitation Robotics (ICORR). IEEE, Aug 2015.
- M. Martins, S. Page, L. Saint-Bauzel, C. P. Santos, V. Pasqui, and A. Mézière, “Real-time gait assessment with an active depth sensor placed in a walker,” in Proceedings of the 2015 IEEE International Conference on Rehabilitation Robotics (ICORR). IEEE, Aug 2015.
- S. Page, M. Martins, L. Saint-Bauzel, C. P. Santos, and V. Pasqui, “Fast embedded feet pose estimation based on a depth camera for smart walker,” in Proceedings of the 2015 IEEE International Conference on Robotics and Automation (ICRA). IEEE, May 2015, pp. 4224 – 4229.

- M. Santos, L. Santana, M. Martins, A. Brandão, M. Sarcinelli-Filho, "Estimating and Controlling UAV Position Using RGB-D/IMU Data Fusion with Decentralized Information/Kalman Filter," in Proceedings of the IEEE International Conference on Industrial Technology (ICIT). IEEE, March 2015, pp. 232 - 239.

- A. Tereso, M. Martins, C. P. Santos, M. Silva, L. Rocha, and L. M. Gonçalves, "Detection of gait events and assessment of fall risk using accelerometers in assisted gait," in 2014 Proceedings of the 11th International Conference on Informatics in Control, Automation and Robotics (ICINCO). IEEE, Sept 2014, pp. 788 – 793.

- M. Martins, L. Costa, C.P. Santos, A. Frizera,"Gait feature selection in walker-assisted gait using nsga-ii and svm hybrid algorithm," in 2014 Proceedings of the 22nd European Signal Processing Conference (EUSIPCO). IEEE, Sept 2014, pp. 1173 – 1177.

- M. Martins, C. P. Santos, A. Frizera, and R. Ceres, "Legs tracking for walker rehabilitation purposes," in 2014 Proceedings of the 5th IEEE RAS/EMBS International Conference on Biomedical Robotics and Biomechatronics (BioRob2014). IEEE, Aug 2014, pp. 387 – 392.

- M. Martins, C. P. Santos, A. Frizera, and E. Seabra, "Design, implementation and testing of a new user interface for a smart walker," in Proceedings of the 2014 IEEE International Conference on Autonomous Robot Systems and Competitions (ICARSC). IEEE, May 2014, pp. 217–222.

- V. Faria, J. Silva, M. Martins, and C. P. Santos, "Dynamical system approach for obstacle avoidance in a smart walker device," in Proceedings of the 2014 IEEE International Conference on Autonomous Robot Systems and Competitions (ICARSC). IEEE, May 2014, pp. 261 – 266.

- M. Martins, E. Seabra, L. Basílio, C.P. Santos, "Conceção, projeto e desenvolvimento de um "guiador" para um andarilho motorizado," in International Conference on Engineering UBI, ICEUBI Nov 2013, pp. 27-29.

- M. Martins, C. Cifuentes, A. Elias, V. Schneider, A. Frizera, and C. P. Santos, "Assessment of walker-assisted human interaction from lrf and wearable wireless inertial sensors," in 1st International Congress on Neurotechnology, Electronics and Informatics. INSTICC, Sept 2013, pp. 143–151.

- M. Martins, C. Santos, E. Seabra, L. Basilio, and A. Frizera, "A new integrated device to read user intentions when walking with a smart walker," in 2013 Proceedings of the 11th

IEEE International Conference on Industrial Informatics (INDIN). IEEE, July 2013, pp. 299–304.

- E. Seabra, L. F. da Silva, P. Flores, J. Machado, M.H. Vu, M. Martins, R. Campos, “Mechatronic medical device for wrist rehabilitation,”in 2013 Proceedings of the 11th IEEE International Conference on Industrial Informatics (INDIN). IEEE, July 2013, pp. 331 – 336.

- M. Martins, L. Costa, C. P. Santos, A. Frizera, and R. Ceres, “Multivariate analysis of walker-assisted ambulation,” in 2013 Proceedings of the 3rd Portuguese Meeting in Bioengineering (ENBENG). IEEE, Feb 2013, pp. 1–4.

- M. Martins, C. Santos, A. Frizera and R. Ceres, “A Preliminary Tests of Joint Angles Measurements with Wireless Inertial Sensors during assisted gait”, in Denmark-South America Workshop on Sustainable Technologies, Research and Innovation, 2012, pp. 1-3.

- M. Martins, C. P. Santos, A. Frizera, and R. Ceres, “Smart walker control through the inference of the user’s command intentions,” in 2012 Proceedings of the 9th International Conference on Informatics in Control. IEEE, Sept 2012, pp. 458– 463.

- M. Martins, C. P. Santos, A. Frizera. “Online control of a mobility assistance smart walker,” in 2012 Proceedings of the 2nd Portuguese Meeting in Bioengineering (ENBENG). IEEE, Feb 2012, pp. 1–6.

- M. Martins; C.P. Santos, A. Frizera, R. Ceres and T. Bastos, “A Novel Human- Machine Interface for Guiding: The NeoASAS Smart Walker,” in 2012 Proceedings of 3rd IEEE Biosignals and Biorobotics Conference - BRC, ISSNIP 2012, pp. 1-7.

- M. Martins, C. P. Santos, A. Frizera, and R. Ceres, “Review and classification of human gait training and rehabilitation devices,” in 2011 Proceedings of the 11th European Conference for the Advancement of Assistive Technology in Europe. AAATE, Setp 2011, pp. 774–781.

- M. Martins, C. P. Santos, A. Frizera, R. Ceres, “Revisão e Classificação de Dispositivos de Treino e Reabilitação da Marcha Humana, “ in VI Congreso de Tecnologías de Apoyo a la Discapacidad IBERDISCAP, Actas del VI Congreso de Tecnologías de Apoyo a la Discapacidad IBERDISC, July 2011, pp. 149-160.

Poster communications

- T. Pereira, M. Martins, C. P. Santos, and A. Matias, “Avaliação da eficácia de um andarilho motorizado numa paciente com tetraparésia atáxica,” 2015, poster presented at 9th Congresso Nacional de Fisioterapeutas, Fisioterapia é Saúde, July, Cascais.

- M. Cotter, M. Martins, T. Pereira, C. P. Santos, A. Matias, and F. Pereira, “Motorized walker gait training: Gait and balance improvement for cerebellar ataxia,” 2015, poster presented at 9th World Congress of the ISPRM. Germany, Berlin.

- M. Cotter, M. Martins, T. Pereira, C. P. Santos, and A. Matias, “Motorized walker for gait training in patients with ataxia,” 2015, poster presented at 9th World Congress of the ISPRM. Germany, Berlin.

- M. Cotter, M. Martins, T. Pereira, C. P. Santos, and A. Matias, “Integração de um andarilho motorizado no programa de reabilitação de doentes com ataxia cerebelosa,” 2015, poster presented at XVI Congresso SPMFR.

- R. Ceres Ruiz, A. Frizera and M. Martins, “Desarrollos evolutivos de andadores avanzados en la mejora de la seguridad, la autonomía y el guiado intuitivo,” 2015, poster presented at II Encuentro de Investigadores ‘Investigación y Envejecimiento’. Fundación General de la Universidad de Salamanca.

1.5 Thesis outline

The thesis is organized in seven chapters, as illustrated in figure 1.1.

Chapter 1 (current chapter) introduces the topic of this thesis, through the presentation of the smart walker developed in this thesis. The overview of the research work is described together with the goals, main contributions and outline of the thesis.

Chapter 2 presents two reviews. First it is presented a systematic review that presents 37 studies that focus on the effects and influence of the walker device on its user’s gait, routine, and behavior. It also presents the testing protocols used to acquire the evaluation parameters, data processing and statistical tools. This review intends to unify the current goals and limitations of the reported studies and highlights the main topics that should be present on walker-related studies. Then, a review that surveys the importance of smart walkers in maintaining mobility, discusses their potential in rehabilitation and their demands as assistive devices. It also presents related research in addressing and quantifying the smart walker’s efficiency and influence on gait. Besides, it discusses smart walkers focusing on studies related to the concept

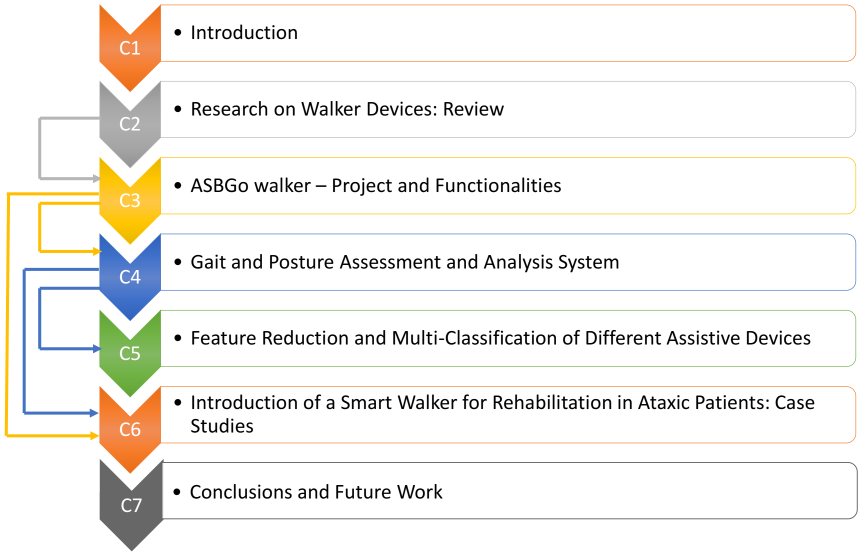


Figure 1.1: Schematic of the thesis outline.

of autonomous and shared-control and manual guidance, the use of smart walkers as personal helpers to sit-to-stand and diagnostic tools for patients' rehabilitation through the evaluation of their gait.

Chapter 3 presents the development and assembly of the proposed smart walker. The design of the smart walker is presented and was planned for specifically help prescribed walker patients for gait therapy. Then, since a smart walker is intended to be a device that can act as a versatile rehabilitation and functional compensation tool, it should be adaptive considering the necessities of its user and its use should be safe. Patients present different necessities according to their intrinsic characteristics, their diseases and therapies. In order to help them, a smart walker should provide different functionalities. This chapter aims the implementation of four diferent operating modes that allow the physical therapist to choose the most appropriate one for the type of difficulty of the patient. In addition, a brief overview of the gait assesment system tool is presented.

Chapter 4 intends to present different assessments: the pattern followed by lower limbs and the patterns of body balance of a walker user. Different systems were created and will be presented in detail. One active depth sensor will provide position and orientation of the feet center and one laser range finder will provide position and orientation of the legs. After validating these systems, a method for spatiotemporal parameters calculation is proposed as well as a multi-sensor data fusion based on these systems. Other system that will be presented is the use of one accelerometer placed at the trunk to indicate the stability of the user regarding his centre of mass position, giving posture and balance information. Finally, it is important to monitor the user safety while walking with the walker device. For this force sensors and infra-red sensors will infer different security states of the user in order to alarm him and advise for dangerous situations.

Chapter 5 presents a gait analysis approach based on feature reduction tecniques associated with multivariate analysis and classification. Such techniques aim to identify differences and similarities in gait performance between three different assistive devices. Also, it is studied how gait patterns differ from patient to patient and are influenced by the assistive device that is prescribed. Standard walker, crutches and rollator are tested with patients recovering from total knee arthroplasty.

Chapter 6 presents a study that verifies the potential of the smart walker and its long-term effects in rehabilitation therapies. This study introduced a smart walker in the rehabilitation of six patients with ataxia. Their gait patterns and postural stability was acquired and clinically evaluated.

Finally, Chapter 7 provides a general conclusion on the achievements of the thesis and the perspectives for future research.

Chapter 2

Research on Walker Devices: Review

People with motor disabilities represent a relatively small minority of the population with disabilities, but their importance transcends their numbers [36]. Stability and balance in ambulation are fundamental for independent activity and quality of life.

Individuals who do not have such stability and balance require the help of assistive devices that may improve their state [1, 4]. Given the importance of assistive devices and the impact they have on the functional ability of the user, research needs to conceptualize and to improve investigation on this area. In the first stage of rehabilitation, treadmill exercises are often prescript since they have a partial weight relief that uses belts or suspension systems. However, the patient should start walking on the floor as early as possible, using parallel walking bars, walkers, canes or crutches. Parallel walking bars restrict the patient's movement to a small area, canes are unilateral, not providing enough support for the muscles and crutches are ver unstable [4]. Thus, different types of walkers appear as a good option to improve mobility and independent performance in mobility-related tasks [4].

Individuals requiring walkers present a decreased ability to provide the supporting, stabilizing, propulsive or restraining force necessary for forward progression [1]. Walkers may help these individuals by releaving their pain through the decreasing weight bearing on one or both lower limbs. Static equilibrium is maintained when the body's center of pressure (COP) is positioned over the base of support. Loss of balance can result when the center of mass (COM) is displaced in relation to the base of support because of voluntary movements or external perturbations. The use of a walker increases the base of support, thereby allowing a greater tolerated range for COM positions. It can also prevent instability by allowing stabilizing reaction forces such as holding on or pushing against the ground [37].

However, several issues have been raised, since a large number of walker owners have reported problems related to the use of a walker related to its design, and to the great number of accidents [2, 38]. The most commonly reported walker related accident is a fall [4].

In addition, it is important that medical staff ensures that patients walk for a reasonable time during the rehabilitation process, to adjust their walking speed, according to their coordination ability and strength, as well as to define the individual limits for walking distance exercises [39]. This is a very demanding and subjective task. Thus, it is crutial to find means to turn this process more effective, objective and less demanding.

For such purpose, research should first evaluate the effects caused by the use of different types of walkers on the mobility of its users. Then, investigation may proceed to the standardization of protocols and methodologies to objectively assess a person's functional physical capability. Then, the adequability and long-term effects of walkers should be investigated in order to help with their prescription. Finally, new functionalities and design of walkers should be reviewed with the help and use of robotic technology.

In the last years, many studies have appeared to show the many methodologies to select the adequate walker type for different patients. Such studies will be presented and discussed in section 2.1 in format of a systematic review. In such section, it will be discussed the related research with the quantification of the conventional walker's efficiency and their influence on gait. These studies are expected to pave the way for further research in specific disorders. Related studies about the benefits and the possible demands associated with walkers will also be examined, by making important research questions. It is essential to determine the effectiveness of walker device interventions in terms of activity and participation for people with mobility limitations, since prescription often does not take into account the influence that walkers may have on the user's resultant gait pattern.

Additionally, in order to help healthcare professionals to execute their work with more efficiency, research began to find a solution to improve walkers' design and funcionalities. Robotics emerged with particular interest to promote safe mobility, specifically considering the prevention of falls through precise motor function evaluation and additional help in the patient's rehabilitation process. Therefore, researchers invested in creating smart walkers. Due to the many potential advantages of smart walkers, it is crucial to present and discuss the current state of this research area. Many examples of studies related to smart walkers development and their different functionalities will be presented in section 2.2 in the format of a descriptive review. It is intended to focus on the relevant achievements that smart walkers had in the past few years, demonstrating that although this area is growing, there are still fundamental questions in terms of usability, design and multi-modal functionalities to address and validate.

2.1 Assisted Gait Evaluation: A systematic review about conventional walkers

Given the prevalence of walkers it is clear that their impact on the health care system is increasing, as well as the functional ability of the user [40]. Therefore, it is crutial to conceptualize and improve the investigation and developments in walker's field, in terms of design and effectiveness of the device in the user's rehabilitation process and functional compensation.

In order to design a useful system for clinicians, research has been focused on this problematic by addressing the characterization of human gait parameters and other aspects with the use of walkers [8]. Moreover, some studies started to relate clinical and functional features of disorders to the specificities of the type of walker [41–44]. However, there are still many issues to address in terms of protocols, data analysis, sample data, etc.

Thus, this systematic review intends to present some walker-related studies in order to answer the following research questions. (A) What are the purposes and goals, in general, of the walker-related studies? (B) Which type of disorders are currently being the focus on walker-research? (C) How are the test experiments and data collection in walker-related studies performed? (D) What are the relevant parameters (outcomes) in walker-assisted gait evaluation? (E) What kind of processing and data analysis is done in walker-related studies?; (F) How do walkers influence and modify the walking gait and are there differences between different types of walkers? A wide range of testing protocols, data processing and statistical tools will be reported in this review. These research questions are summarized in table2.1.

The key criterion for this review was to include studies that incorporated a research design explicitly examining the evaluation of the use of a walker. A second important criterion was to include only those studies which specified the population, testing protocol, the framework of data processing and the evaluated parameters. Other inclusion criteria were the type of devices included in the studies: standard walker, two, three and four-wheeled walker (Figure 2.1).

The results of the selection process, that took into account the aforementioned criteria, are shown in figure 2.2. A total of 1609 citations were retrieved from the search of electronic databases. After inspecting the title and abstract, 100 articles were assessed for full text review. Based on the previous inclusion criteria, 37 articles were suitable for full review. Tables of Appendix A summarize the selected studies.

The selected studies present the effects and influence of different walker types on the user's gait pattern, routine, and behavior and report the testing protocols used to acquire the outcome parameters, data processing and statistical tools.

Eleven studies compared the impact of different walkers to determine their limitations and risk factors. The results of these studies may help to explain the condition and injury of

Figure 2.1: Conventional Walkers: A. four-wheeled walker; B. three-wheeled walker; C. two-wheeled walker; D. standard walker.

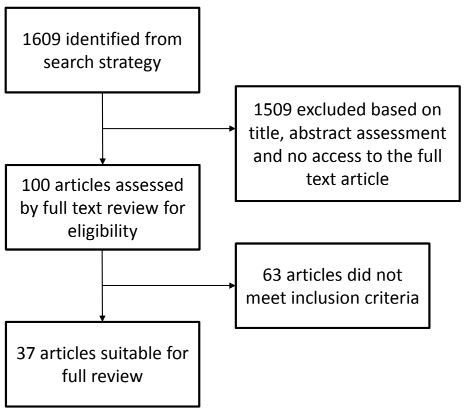


Figure 2.2: Search results through the review process.

patients using walkers. In addition, these studies became a good reference to improve walkers' design in order to facilitate their users in holding and handling. These articles are divided into four types of studies: attentional demands, metabolic cost, walker risks and gait modifications provoked by walker use.

Another eleven studies explored the standard walkers (SWs). SWs are recommended and prescribed for patients with hip fracture and a variety of musculoskeletal and neurological problems [1, 8, 39]. Studies show that a SW is an attractive option following surgery or trauma to the lower extremities (LEs) as it may provide mechanical support and inherent stability. However, this walker has some drawbacks related with the discrete, unnatural and slow gait motion that is imposed on its users [1]. The physical therapist may evaluate the patient for abnormalities and identify the correct assistive device (AD) to prescribe and the type of gait pattern to be learned. Also, the SWs have not changed in their basic design along the past years [1, 39]. Thus, the role of a SW has to be revised and explored in order to consider design modifications and enhancement, as well as, define which type of patients should actually use this type of walker. Those eleven studies have explored these issues. They present studies related to the evaluation of spatiotemporal, kinematic, kinetic patterns and muscle activity of SW users while ambulating with the device. In addition, they focus on the effect and demands on the LEs and/or the upper extremities (UEs). Another concern is related to the postural stability and reaction of the perturbations while ambulating with the walker.

Two articles explored the two-wheeled walker (2WW). The 2WW is designed for people who have little muscle strength in UEs or tendency to fall back when trying to lift a SW. This walker promotes a natural gait, not forcing the person to stop in order to move forward. It is particularly useful for patients with Parkinson's Disease (PD) or Paraparesis, because they have a strong tendency to fall backwards [1]. However, very few articles studied this walker. This review, only identified two articles that explore this type of walker and compare its potential with crutches, with two different concerns: LEs and UEs' demands.

Three-wheeled walker (3WW) is the least explored and studied walker in the literature. Only one study was found with this type of walker related to exercise evaluation. Because of this, the study will not be analyzed in detail.

Twelve articles were selected to present the typical studies about the effects that four-wheeled walker (4WW) have on its users. 4WWs are preferred by patients, as compared with the 2WW, for example [1]. These walkers do not need to be raised and have a wheel system that allows these devices to roll in a smooth and gentle way with little effort. The 4WW promote a natural gait and are easier to use, when compared with other types of walker. However, they provide less stability to the user and require great cognitive effort [1, 8]. This happens because they have manual brakes, i.e. controlled by the user when he wants to stop

Table 2.1: Summary of the research questions proposed on the current systematic review.

What are the purposes and goals, in general, of the walker-related studies?(A)	• Attentional demands; • Metabolic costs and functional exercise capacity; • Risk factors (falls); • Gait modifications (spatiotemporal, kinematic and kinetic changes); • LEs and UEs demands (biomechanical effects); • Postural adjustments; • Stability (balance).	***What kind of processing and data analysis is done in walker-related studies?(E)***	• Data filtering; • Parameters normalization; • Exploratory data analysis; • Analysis of variance and covariance; • Post-hoc analysis; • Parameters correlations; • Level of significance: 5%. • Software: SPSS, SAS, Minitab, MATLAB, Labview and Ariel.
Which type of disorders are currently being the focus on walker-research?(B)	• Young and healthy; • Elder and healthy; • Elder (walker users, potential walker users); • Chronic Obstructive Pulmonary Disease (COPD); • Huntington's Disease (HD); • Parkinson's Disease (PD); • Amputation patients; • Spinal Cord Injury (SCI); • Hip or knee joint replacement surgery.	***How walkers influence and modify the walking gait and is there differences between different types of walkers?(F)***	**Comparison of multiple walker types:** • Avoid SW if the patient has attention problems and need to concentrate on its ambulation; • Healthy elders should use 4WW to reduce their fatigue, and COPD elders should use the 2WW; • Walker design should facilitate elders in holding and handling; • Walker use might change the body mechanics of its user; • Any type of walker is useful as a tool to overcome freezing on PD. However, 4WW can be taken into account for promoting safe ambulation and easy maneuverability; • 4WW more acceptable to HD; **Standard walker:** • Limits velocity; • Provokes asymmetrical step length and time; • Elevated vertical accelerations; • More stability and balance; • Flexed trunk posture; • Supports great percentage of body weight; • Lower height alleviates the load on LEs, but increases demand on UEs; • Demanding for shoulders and elbows; • Prevents stepping compensatory reaction in loss of balance situations; • Improves postural control when visual information is compromised; • Maintains/increases muscle strength; • Caused psychological and physical consequences; **Two-wheeled walker** • Great base of support; • Reduces percentage of body weight load; • High load on shoulder for SCI patients; • Decreases step length. **Three-wheeled walker** • Provides high exercise capacity **Four-wheeled walker** • Improve functional capacity; • Patients with airflow limitations may benefit; • Increase hip joint flexion; • Reduce range of motion of LEs joints; • Not considered a safe device; • Relieve muscular loads in UEs; • Reduce attentional demand in narrow surfaces.
How are the test experiments and data collection in walker-related studies performed?(C)	• Depend on the study purpose; • Different test conditions: different walker types and without assistance; • Pre-determined time and/or distance to walk forward on a walkway; • Questionnaires, motor and cognitive scores; • Sensors placed both on the subject and walker; • Test pre-established different gait patterns, velocities, body inclination and weight.		
What are the relevant parameters (outcomes) in walker-assisted gait evaluation?(D)	• Spatiotemporal parameters; • Voice response time (VRT); • Oxygen cost, heart rate and blood pressure; • Questionnaire results; • Observational results; • Kinematic parameters; • Kinetic parameters; • Electromyography parameters; • Motor and cognitive score results.		

the device. This implies that the user has to be able to learn to press the brakes, simultaneously (press the right and left brake at the same time), and react quickly to dangerous situations. As it will be presented, the main focuses of these articles are the effects on the balance and UEs demands, the changes on the LEs' control, attentional demands and the effects on functional exercise capability on patients with Chronic Obstructive Pulmonary Disease (COPD).

After analysing such studies in a format of research questions, a discussion about the limitations and problems of the reported studies will be presented.

2.1.1 Research Questions

(A) What are the purposes and goals, in general, of the walker-related studies?

Walker-related studies focus mainly on seven concerns regarding the ambulation with a walker as presented on table 2.1.

Regarding attentional demands, Wright and Kemp [45] provided a preliminary examination of the attentional demands of ambulating with a SW and 4WW and introduced a dual-task methodology. They stated that one factor that might influence a patient's ability to use an AD is the cognitive demand required to properly use the device. Later, Wellmon et al. [46] and Miyasike-daSilva et al. [47] studied the attention required to walk with 4WW among the elderly population and the execution of two tasks at the same time.

Another main topic is the metabolic costs induced by the use of walkers. This point is important for subjects with breathing problems. Three articles [48–50] compared metabolic measures between unassisted and assisted gait. They hypothesized that the lower the cardiorespiratory and metabolic costs of ambulation per distance traveled, the greater the efficiency. Other five articles worked with COPD patients [43, 44, 51–53], since these patients have a breathe limitation which need to be decreased. They focused on the evaluation of the effect of 4WW on disability, oxygenation, and functional exercise capacity in these patients in order to improve their quality of life. Finally, Grant and Capel [42] evaluated the exercise capacity with a 3WW in people with pulmonary emphysema.

Another important concern is the fact that elderlies usually get injured when using ADs. Thus, four articles [54–57] discussed whether the potential risk factors causing injury exist or not and which are the difficulties while using walkers. The most studied topic is the gait modifications (spatiotemporal, kinematic and kinetic changes) and the effects caused by walker use in specific diseases like Huntington's Disease (HD) [58] and Parkinson's Disease (PD) [41, 59]. Gait and balance impairments lead to frequent falls and injuries in individuals with HD and PD. Walkers are often prescribed to prevent falls, but their efficacy is unknown. Thus, in [41, 58, 59], they examined the effects of different types of walkers on quantitative gait measures. Also, Smidt and Mommens [60] and Crosbie [61] focused on describing gait with spatiotemporal parameters, but only with SWs. Smidt and Mommens [60] compared assisted gait with unassisted gait to discuss the clinical implications and modifications evoked by the SW use. Crosbie [61] evaluated two specific different gait patterns and observed which one was the best to ambulate with a SW. Later, Fast et al. [62], Deathe et al. [63], Melis et al. [64], Ishikura [65], Bachschmidt et al. [66], in addition to focusing on spatiotemporal parameters modifications, they studied the forces that are transmitted through the frame of the walker during ambulation. Parameters like walker height [63], kinematic excursions [64] and muscle activity [65] were also considered.

Other studies assessed the influence that 4WW has in gait and mobility of elders [67, 68] and studied the differences in gait parameters among True 4WW users (TUR) and Potential rollator users (PUR) [57].

Regarding biomechanical effects and demands on both extremities of the body, it is important to verify if the long-term use of a walker can lead to injury due to excess effort on the limbs. Youdas et al. [69] proposed a study to determine if subjects can offload the right LE to a targeted amount of weight bearing using the 2WW. They evaluated whether the limitation of weight bearing is essential to recover in the rehabilitation program. Another article studied the changes on the LEs' control, in terms of biomechanical effects with the comparison between the use and non-use of the 4WW [70]. Bachschmidt et al. [66] and Mcquade et al. [71], on the other hand, focused on the demands on the UEs, studying the forces that were applied on the UEs joints. Haubert et al. [72] conducted a study, which purpose was to compare the reaction forces of the shoulder joint and stride characteristics during 2WW ambulation in subjects with Spinal Cord Injury (SCI). They stated that the functional integrity of the shoulder joint is vital for achieving independence in subjects with SCI. Two articles [73, 74] studied the effects on the balance and UEs demands.

Takanokura [73] developed a 2D mechanical model to optimize the height of the 4WW to reduce the muscular loads in the UEs with various road conditions. Later, Tung et al. [74] characterized the way in which UEs may be used for balance control during walking with a 4WW, and investigated the consequences of using the UEs for these tasks. Finally, other kind of study was performed by Bateni et al. [75] and Vennila and Aruin [76] whose focus relied on the postural adjustments and stability during assisted ambulation. Bateni et al. [75] focused on the lateral movement and the compensatory reaction during lateral loss of balance. Vennila and Aruin [76] related vision with support of the anticipatory and compensatory postural adjustment and their interaction and studied muscle activity during these events.

(B) Which type of disorders are currently the focus of walker-research?

The most studied population type are the elderlies [46, 47, 49, 54–57, 61, 67, 68, 73, 74, 77]. Elderlies can be healthy (participants with no acute medical illness), walker users or amputated patients [63]. Then, COPD [43, 44, 51–53, 62], HD [58], PD [41, 59], SCI patients [64, 72] and hip or knee joint replacement surgery patients with good health and functional UEs [71] are other type of disorders that are addressed in walker-based studies. It is noteworthy that Fast et al. [62] evaluated various diagnoses leading to gait dysfunction (hip fracture, hemiparesis, knee amputation, COPD, cerebral palsy, hemiplegia ...). However, Fast et al.

[62] just evaluated one patient per disease.

Major populations that are also studied are the healthy and young subjects [45, 48, 50, 60, 65, 66, 69, 70, 75, 76]. Although there is no accepted justification for using healthy subjects in studies with walker, some studies [70] justified "The reason for studying a group of healthy subjects was that it was both unethical and difficult to ask actual walker users to walk without their 4WW" and "the results may be used as a model for general changes in the joint moment pattern and the kinematics during 4WW-walking in healthy subjects".

The sample size goes from 1 [62] to 47,312 [55] participants. There is a need to standardize the number of samples that is adequate for these type of studies. Since there is no specification, each study selects the sample size according to the availability of resources. In Tables of Appendix 1, one can find inclusion and exclusion criteria, as well as demographic and sample size information for each selected study.

(C) How are the test experiments and data collection in walker-related studies performed?

Depending on the purpose of the study, different protocols were established. Regarding attentional demands, the protocol consists in three main phases [45, 46]: (1) rapid respond to a stimulus with a vocal or manual response (secondary probe reaction-time task); (2) walk with and without assistance (single task); (3) walk with and without assistance in conjunction with the secondary probe reaction-time task (double task). Miyasike-daSilva et al. [47] only performed two tasks (single and double task).

To evaluate metabolic cost and functional exercise capability the following protocol with three phases was presented: (1) Oxygen consumption test that consists on running on a treadmill until volitional exhaustion; (2) metabolic measurement devices during ambulation tests with and without different types of walkers; and (3) determine test-retest reliability by performing more testing sessions. Despite using different types of walker and subjects, the protocol was very similar across studies. Priebe and Kram [50] was the only study to introduce two gait patterns (normal walk pattern and repeatedly step forward with one foot and then step-to the same position with the other foot). On the other hand, Roomi et al. [51], Honeyman et al. [44], Solway et al. [52], Probst et al. [53] and Gupta et al. [43] selected elderly patients with COPD and only performed phases (2) and (3). Phase (2) was specified in all studies as a 6-minute walk test (6MW).

In order to evaluate risks of the walkers' for the users, studies reported protocols based on questionnaires, which are different across studies. Leung and Yeh [54] proposed a questionnaire of 37 items divided into four parts: background of participants; walker using state; activities in using walker and the opinion of user (5-point Likert Scale). The study assess whether the potential risk factors causing the injury exist or not while elders are using walkers. Liu [57] performed a Mini-Mental Status Exam, that collets demographic data, background about the walker usage, walker maintenance and fall occurrences. Andersen et al. [56] performed a Short-form 36 and fall inquiry questionnaires prior and after training with a walker. Stevens et al. [55] based their results on observation of surveillance data of injuries treated in hospital emergency departments.

When the purpose is to evaluate the gait modifications produced by the walker, there is a great concern about the correct choice of sensors. In the articles the following sensors were used: accelerometers, placed posterior to the sacrum [60] to be sensitive to velocity changes; pressure foot switches, attached to the heel and forefoot of each feet [60, 63] to provide stance and swing times, through the detection of heel contact and toe off events [63]; GAITRite walkway [58, 59, 68, 77]; and video cameras that film while the subject is ambulating and record data that is later digitalized [61, 64]. When using cameras, reflexive markers usually are chosen to be placed on the user in order to identify specific joints and limbs and can also be placed on the walkers' frame [61, 64] for orientation purposes.

Another great concern with these protocols is the gait pattern. Some articles [60, 61, 65, 66] specified the type of gait pattern that the user should perform. Smidt and Mommens [60] tested and compared nine different walking patterns and explained the characteristics of each one of them (delayed two point, delayed three point, four point, etc). Crosbie [61] tested two walking patterns, as conditions: gait D (the SW was advanced followed by the right foot and then the left) and gait S (the walker and right foot moved simultaneously forward, followed by the left foot). Bachschmidt et al. [66] selected three count, delayed and five point gait pattern. For more details about these patterns see [60, 78]. Ishikura [65] selected the partial weight bearing gait (for more details consult [79, 80]).

The most used test experiment was to ask the subject to walk on a flat pathway during a determined time/distance at a self-preferred speed with different types of walkers and without assistance [41, 58–66, 68, 69, 77]. Then there are some differences among the studies. Smidt and Mommens [60] tested the influence of different velocities, Deathe et al. [63] the influence of different walker heights, Ishikura [65] performed tests with the hip joint at various flexion angles and Bachschmidt et al. [66] tested different load weight percentage over the walkers. Kloos et al. [58] and Kegelmeyer et al. [59], in order to test maneuverability of the different walkers around obstacles, timed the subjects while walking as fast as they could in a figure-of-eight pattern around two chairs. They also performed clinical scales on the subjects. In Cubo et al. [41], the subjects had to perform different tasks: rise from a chair, walk through a doorway, down a hallway, turn, walk through the doorway again and sit down. This walking course was designed to reproduce conditions that increase freezing on the PD subjects. The patients were also rated using motor and cognitive clinical scales. Vogt et al. [67] designed a study that included physiotherapy, ergotherapy, ergometer exercises and different motor clinical tests. Schwenk et al. [68] also performed motor scale tests. In order to evaluate the UE and LE demands, video cameras or infra-red with reflexive markers on the target limbs were used [66, 70–72, 74] as well as load cells under the handles of the device to acquire forces on the walker and calculate the amount of body weight that the walker is bearing [72–74]. Protocol for this purpose was also based on asking the subject to walk at a preferred-speed in a pre-determined time/distance [66, 70–72, 74] with different types of walker and without assistance. Haubert et al [72] also performed motor clinical scores to determine which limb would experience the largest weight-bearing load during ambulation. Takanokura [73] reported a different study where he created a mechanical model to optimize the height of the 4WW in order to evaluate UE efforts. To acquire the model parameters, they asked subjects to use the walker in dry asphalt and dry gravel roads with imposed posture characterized by stretched UEs and not bended elbows.

Regarding posture control and stability, accelerometers [76] and video cameras [75] were used. Bateni et al. [75] asked the subjects to stand in a standard position representing a typical self-selected stance and were instructed to push down the walker with constant force. Vennila and Aruin [76] asked the subjects to maintain a vertical position with upright stance during the test. In Bateni et al. [75], subjects were placed on a movable platform to provoke perturbations for postural reactions. Subjects had to wear a harness, for safety. The platform moved unpredictably and the subjects performed a distraction task to impede ability to engage proactive strategies. In Vennila and Aruin [76] a pendulum impact was used to cause perturbation. On both studies, tests were performed with and without a walker.

For more detailed information about the protocols, consult Tables in Appendix A.

(D) What are the relevant parameters (outcomes) in walker-assisted gait evaluation?

In walker-related studies, the purpose of the study, type of walker and the type of subjects almost do not influence the choice of relevant parameters to acquire and evaluate. The typical parameters that almost all studies calculate are the spatiotemporal parameters - velocity, stride length and time, swing and stance time, double support time, etc [41, 45, 50, 58–61, 64, 66, 68, 69, 72, 74, 77]. Then some variations appear depending on the purpose. When studies want to evaluate attentional demands, voice reaction time (VRT) is acquired [45–47]. When evaluating the metabolic cost, the oxygen consumption and saturation is essential to be measured [43, 44, 48–53]. Another essential parameter is to record the distance made on 6MW [43, 44, 51–53] to obtain the functional exercise capacity. Some of these latter studies also performed fatigue scales and questionnaires [43, 44, 52, 53].

Potential risks of the use of a walker are evaluated through different questionnaires among studies and video observations [54–57].

Regarding gait modifications, studies with such purpose differ most with the type of disorder. Gait modifications in PD caused by walker usage were characterized by freezing time, number of freezes, mean duration of freezes through video recording [41] and GaitRite® system [59]. Kegelmeyer et al. [59] also calculated the number of observed stumbles (loss of balance from which the subject recovered without assistance) and falls (loss of balance for which the investigator provided assistance to prevent the subject from coming to the ground). Kloos et al. [58] calculated the same parameters as Kegelmeyer et al. [59] for HD minus the freezing related parameters. The remaining studies with the same type of patient (elderlies and young subjects) evaluated kinematic and kinetic parameters [61–65, 69]. LE and UE demands studies are very similar to the latter articles (gait modification purpose), however focusing more on the kinematic effects on the LEs and UEs [66, 70–74], through the acquisition of the position and angular displacements of the joints (ankle, knee, hip, elbow, wrist and shoulder) as well as joint moments and angular impulses. Finally, the evaluation of postural adjustments focused on calculating the centre of pressure (COP) [75, 76]. However, other studies with other purposes also calculated these parameters [63, 65, 74].

(E) What kind of processing and data analysis is done in walker-related studies?

The data processing and analysis is independent of the study purpose and type of walker and patients. In terms of pre-processing, data from the video records was filtered with a fourth-order, zero-lag Butterworth filter [61, 64, 72]. Sensors data can be filtered by low-pass digital filter [63, 64, 70, 71, 74, 76]. The electromyography (EMG) data was high-pass filtered [65] or low-pass filtered [76]. Before filtered, parameters have to be normalized [61, 64–66, 71, 75]. Then, some studies performed exploratory data analysis. Smidt and Mommens [60], Youdas et al. [69], Andersen et al. [56], Crosbie [61], Deathe et al. [63], Bachschmidt et al. [66] and Mcquade et al. [71] calculated the average and standard deviation for each parameter, and Kloos et al. [58] and Kegelmeyer et al. [59] calculated coefficient of variation values for all parameters.

In terms of statistical analysis, Wright and Kemp [45], Holder et al. [48], Foley et al. [49], Deathe et al. [63], Bateni et al. [75], Youdas et al. [69], Leung and Yeh [54], Gupta et al. [43], Liu et al. [77], Solway et al. [52], Wellmon et al. [46], Miyasike-daSilva et al. [47], Tung et al. [74], Liu [57], Priebe and Kram [50], Kloos et al. [58] and Kegelmeyer et al. [59] used a one-way analysis of variance (ANOVA) for repeated measures. Holder et al. [48] and Vennila and Aruin [76] performed a two-way ANOVA repeated measures. If the sample is not normal distributed, non-parametric tests like, Wilcoxon signed-rank test [53, 66, 68], Krustal-Wallis test [67] and Mann-Whitney U test [68] are used.

When significant differences were found Wright and Kemp [45], Holder et al. [48], Bateni et al. [75], Priebe and Kram [50] and Tung et al. [74] used the Tukey's post hoc procedure to determine which values differed significantly. On other hand, in Foley et al. [49] and Youdas et al. [69] studies, when a difference was found, a Newman-Keuls post hoc analysis was used to determine where the difference occurred. Paired t-tests were performed to compare differences in the study parameters [44, 51, 68, 70, 74, 77]. If the results were significant, Leung and Yeh [54] used least significant difference method as post-hoc test. Stevens et al. [55] used a direct variance estimation procedure that accounted for the sample weights and complex sampling design. Cubo et al. [41] used mixed models and Friedman's test. Tests performing multiple comparisons of pairs of conditions were performed using Signed Rank tests and with a Bonferroni correction for multiple tests [69, 71, 76]. To verify how strong the relationship of variables was, Tung et al. [74] used analysis of covariance (ANCOVA).

Then, some articles used statistical tools to search for differences and correlations between the variables. Ishikura [65] used Chi-square test for goodness fit and to examine differences between variables. A Spearman's correlation coefficient was performed to correlate variables [65, 71]. To identify correlations, it was performed a one-way ANOVA and Fisher least significant difference method. To model the relationship between one variable with one or more explanatory variables univariate can be used [52, 73] and multiple linear regression [52], stepwise multiple regression [52, 53] and coefficient of multiple correlations [72].

The level of significance was set at 5%. The statistical analysis was run in the SPSS statistical software package[46, 54, 56, 68, 69, 71, 72, 76, 77], SAS software [41, 53, 58, 59], Minitab [61], Matlab [70], Labview [52] and Ariel Performance analysis system [64].

(F) How walkers influence and modify the walking gait and are there differences between different types of walkers?

In order to answer this research question, it was divided into: (i) Comparison of multiple walker types studies, (ii) SW, (iii) 2WW and (iv) 4WW studies.

Comparison of multiple walker types

The main findings of the selected articles that compared multiple walker types [41, 45, 48–51, 54, 55, 57–59] and the organization of the study purpose will now be addressed.

Wright and Kemp [45] compared the attentional demands between SW, 4WW and no walker. Their findings indicate that greater attentional demand was required when ambulating with the SW, since voice reaction time (VRT) was greater when walking with this device. The usefulness of the dual-task methodology was emphasized, since it is simple to administer as an evaluation tool to examine the progress of the patient in terms of attentional demands.

In Holder et al. [48], unassisted ambulation was compared with SW and 4WW. Findings indicate that unassisted ambulation resulted in the lowest oxygen consumption per minute, and the use of SW resulted on the greatest consumption. This may happen because SW requires to be lifted causing more fatigue to the patient. Foley et al. [49] also compared SW with 4WW and the results demonstrated that ambulation with a SW in elderly required greater oxygen consumption than in unassisted ambulation, as in [48]. The values obtained by each study are different, since the target subjects (elderlies and young subjects) were different as well as the speeds performed on each study. Foley et al. [49] also found that the use of any AD with an elderly patient with a history of myocardia infarct or who has been identified as having an increased risk, should be undertaken with caution and appropriate monitoring. Use of a SW with these patients requires a greater degree of caution. Priebe and Kram [50] obtained the same conclusions as Foley et al. [49], in terms of oxygen consumption. In addition, they reported that 2WW requires greater consumption than 4WW. Roomi et al. [51] also studied metabolic costs, but with COPD patients. They reported that 4WW improves exercise capacity in elderly patients with COPD whereas a SW reduces it. They also concluded that the need to lift the SW requires an extra metabolic requirement, since it is a work involving arm elevation. Then, they concluded that, in those elderly patients with COPD who remained disabled despite optimum medical therapy and following a pulmonary rehabilitation program, a trial with a 2WW is worthwhile since it is associated with reduced oxygen consumption during exercise. This conclusion is different by the one reported by Priebe and Kram [50]. However the study population is different. Another conclusion is that SW should be avoided if possible for these patients, as it was also recommended by Foley et al. [49]. If a patient finds the 2WW unacceptable, a 4WW should be the next choice.

Regarding walker use risk evaluation, Leung and Yeh [54] suggested three issues that might be beneficially addressed by future studies in this area to reduce walker risks: posture to hold the walker; elderlies' strength of lifting up; observe elderlies while using the walker in e.g., sit to stand, stand to sit, and walk. In summary, according to the results of this study, a walker should be designed in order to facilitate elderlies in holding and handling. Liu [57] analysis shows that older walker users experience falls, despite the use of a walker. Problems commonly identified among walker users were: lack to consult a medical professional when obtaining a walker, incorrect walker height, poor walker maintenance, improper gait initiation and forward-leaning posture. More specifically, the results from this study indicated that incorrect walker height might not lead to increased fall incidence, but might be an important factor in causing forward-leaning posture during ambulation. This forward-leaning posture could be a significant factor in the higher incidence of falls among walker users. Thus, using a walker might change the body biomechanics of its user.

Many factors should be considered to determine if an elderly may need an ambulatory device. The risk of fall could be greater if combined with other walker problems, such as incorrect walker maintenance and height. Gait and posture pattern while using a walker should be evaluated. Stevens et al. [55] found that injuries increased seven times when walker assisted. Wrist and forearm fall injuries are seen more often in healthy elderlies who are able to use their arms to cushion a fall. People using ADs had almost five times as many fall injuries from tripping as from slipping. The frequency of fall injuries, especially in people using walkers, suggests that people may have problems using ADs effectively. Approximately 4% of injuries occurred when people were sitting, standing, or transferring and an additional 5% to 6% while people were bending, reaching, or carrying objects.

Considering the results of gait modification studies, Cubo et al. [41] focused on PD subjects and demonstrated that most walkers are useful to overcome freezing. Only SWs aggravated freezing, since lifting and replacing the walker on the ground impedes smooth walking, increasing freezing. However, they confidently discourage PD patients with predominant freezing from using a walker to overcome this clinical problem. In their view, the most likely explanation is that the walker acts as a visual obstacle. Because doorways, narrow passages and other restricting environmental elements typically aggravate freezing, the presence of the walker with the patients' close extra-personal space may have contributed to the poor outcome on freezing. Then, Kegelmeyer et al. [59] stated that maneuverability is an important factor to consider when prescribing an AD as many individuals with PD fall when turning or avoiding obstacles. The added support of an AD that allows for smooth turns with larger steps and radius may be beneficial in PD. Safety was best with 4WW and worst with SW and 2WW use, considering number of falls and stumbles. SW produced the most variable gait and high attentional demands, which is consistent with Wright and Kemp [45]. During turns, the 2WW provoked the higher number of freezing episodes. Fall prevention is a critical component of care for individuals with PD. Given the frequency which ADs are prescribed, it is critical that clinicians are provided with evidence on which to base their recommendations. This study provides evidence that gait with the 4WW produced a pattern most similar to the individual's spontaneous pattern with no AD and did not decrease velocity or increase variability, as did the other devices. In addition, this walker produced a safer and smoother gait when making turns. Thus, 4WW appears to be a good choice of device for promoting safe ambulation in individuals with PD. However, as it was mentioned before, Cubo et al. [41] stated that PD subjects with high freezing episodes are not indicated for any type of walker.

In Kloos et al. [58] walking with a 4WW produced a more efficient, consistent and safe gait pattern than other commonly prescribed ADs in individuals with HD both on a straight path and during turns. The greater stability, ease of use, and maneuverability of the 4WW over other devices may account for its better performance. The SW and 2WW required the user to lift the device in time with their stepping whereas the 3WW and the 4WWs allow the person to push the device without lifting it.

Caution when prescribing ADs for individuals with HD who have difficulties with learning sequences of movements and performing a second task during walking. Gait with the 3WW was equivalent to the 4WW across several measures. Thus, these observations are likely to make the 4WW more acceptable to patients and increase likelihood that the device will be used. Based on these findings, they recommend that clinicians consider prescribing 4WWs over other devices for gait impairments and fall prevention for individuals with HD.

Standard Walkers

The use of SWs is common in the rehabilitation process. However, it may impose predictable limiting effects on the gait pattern. Many conclusions were obtained by the eleven selected studies [56, 60–66, 71, 75, 76].

First of all it is important to point out that most studies were conducted with healthy people. But Crosbie [61] reported a justification for not using subjects with limitations: “although being healthy, the age of the subjects are comparable with subjects for whom SW is prescribed. The lack of any predisposing gait abnormality supports the fact that spatiotemporal and joint kinematics characteristics observed in the gait pattern are attributable to the walker rather than pathological limitations.”

The overall findings for spatiotemporal and kinematic characteristics were that subjects walked slower with SW [64] than without it, which implicates that the use of a SW limits velocity. Crosbie [61] obtained the same conclusions with gait D (the SW was advanced followed by the right foot and then the left), as well as Melis et al. [64] with SCI patients, which have the same pattern as the one tested by Smidt and Mommens [60], the five point pattern. This lower velocity caused a higher gait cycle time. Stance, swing and double-stance time were symmetrical, but step length and step time were asymmetrical [60]. It also provides a slower cadence, higher stance, decreased hip excursion and reduced step length [64]. Vertical accelerations were disproportionately elevated, since assisted gait tends to increase the vertical loading on the structures of the body [60]. However, it causes less perturbation of balance [61]. It may offer added security and stability to the user as a consequence to its slowness. However, this pattern causes no benefit to forward linear momentum of the body, since it is a discrete movement. Thus, this type of gait (gait D or five-point) may be chosen to subjects with physical limitations, since it does not comprise balance [61]. But the fact that the walker is advanced forward while the LEs remain in place, it provokes a flexed trunk posture [61, 64, 65]. This characteristic should be noticed by the therapists since this can lead to the alteration of the tissues around the hip [65].

Crosbie [61] also tested another pattern with the walker, gait S (the walker and right foot moved simultaneously forward, followed by the left foot) and concluded that this pattern imposes a less flexed posture on the protected hip joint during the period of weight transfer forward onto the frame than gait D. In addition, this pattern is faster than gait D. Thus, if the subject has no problems in balance and coordination, gait S should be preferred.

In terms of articles that dedicated their study to the evaluation on the forces transmitted to the SW, in the overall they concluded that the walker is a stable device and high forces can be supported by the walker (vertical support) [62–65]. This caused a reduction of the load/weight of the LEs since the walker supports great percentage of the body weight. However, the UEs can be injured since the load is supported by them [63–66, 71].

Fast et al. [62] tested various patients with disturbances of gait, and detected two patterns in the overall patients: use of the walker to unload one of the LEs (fracture on one side, hemiparesis, amputation, stroke) and use of the walker to enhance balance (progressive supranuclear palsy, cerebral palsy). On the first pattern, the patients ambulate on a synchronous, rhythmic pattern, transmitting a significant portion of their body weight to the walker, thus SW evokes a reduction of the load/body weight on the LEs and supports a large percentage of the body weight. On the second pattern, the patients ambulate with an asynchronous and less regular pattern and a random loading pattern prevailed. The rate of progression was slow and the patient covered short distances. Thus, they demonstrated that different patients require different needs and use the walker differently, which means that different designs and specifications are necessary.

Reinforcing the idea that the walker design and specifications should be different from patient to patient, Deathe et al. [63] tested different heights of the walker in amputee patients, and concluded that lower height alleviates the load on the prosthetic leg but increases the demand on UEs. Thus, subjects with UE limitations need a higher walker and subjects with LE limitations need a lower walker. If the patient has both problems (limitations on the LEs and UEs), the therapist has two choices: recommend another type of walker or find a trade-off for the height that benefits both limitations.

Bachschmidt et al. [66] and Mcquade et al. [71] studied in more detail the UEs and both concluded that the arms of the walker support the body against ground reaction loads. However there is a high demand on the elbow and shoulder joints to support body weight, being more demanding for the shoulder than for the elbow. Other concern is that many walker users are frail, elderly individuals and may not have the UE strength to meet the higher demands of walker use during rehabilitation of hip fracture. Thus, parameters that can influence the demand on the UEs were reported: walker height, UE length, subject weight, forward placement of walker, cadence, velocity and stride length.

In terms of postural control, Bateni et al. [75] and Vennila and Aruin [76] concluded that SW causes a biomechanical stabilization (reaction forces generated by the user's hands) and the reaction forces prevent instability and recovering balance, in the event of a disturbance. Which is in accordance to Crosbie [61], Fast et al. [62], Deathe et al. [63], Melis et al. [64] and Ishikura [65]. However, Bateni et al. [75] reported that the SW can sometimes prevent lateral movement of the legs and consequently disabling the implementation of the stepping compensatory reactions' in situations of loss of balance. On the other hand, Vennila and Aruin [76] suggested that the walker can be a valuable strategy to improve posture control when visual information is not available or compromised.

In relation to the muscle activity, Ishikura [65] demonstrated that the use of the walker provokes a high muscle activation level that maintains/increases muscle strength, helping on the everyday exercise.

Finally, Andersen et al. [56] with a more different approach to the demands of the SW concluded that the use of walker causes a weakening of physical functioning and general health and these limitations can cause psychological and physical consequences. However, it should be recognized that AD use may positively affect both mobility and fitness levels, as demonstrated by Ishikura [65].

Two-Wheeled Walkers

2WW is a type of walker with little research. However, it is usually a substitute for the SW for subjects with little muscle strength [1].

Youdas et al. [69] concluded that 2WW has a great base of support and subjects maintained a low percentage of body weight for a long time during stance with the walker. However, speed, cadence, step width and stride length are reduced when ambulating with the walker. On the other hand, Haubert et al. [72] concluded that the load on the shoulder by the walker is low, however is still very high for the SCI patients. They also observed a decreased stride length during ambulation with the walker, which can be attributed to the differences in structural design. Its design can provide a barrier to limit mobility of the LEs.

Rollators (Four-wheeled walker)

Typically, 4WWs are not used in rehabilitation environment [39]. However, this type of walker has many advantages over other types. Twelve articles [43, 44, 46, 47, 52, 53, 67, 68, 70, 73, 74, 77] were found.

Functional capacity improves with the use of a 4WW. An increase in 6-minute walking distance was reported, demonstrating that this is a valid outcome to measure clinical trials in patients with airflow limitations. There was also a reduction in dyspnea, increase in walking ability and high sense of security, allowing an increased stride length and speed [43, 44, 52, 53]. The decrease on oxygen consumption may be related to the fact the arms are supported on the walker. When the arms are unsupported, the accessory muscles increase their participation in the postural support of the chest wall. To reduce dyspnea, subjects often stabilize their arms. This stabilization enables the arm and shoulder muscles to help the muscles of respiration [52, 53]. In addition, the 4WW allows the subjects to lean in a forward position which improves diaphragmatic function. This can improve distribution and reduction of ventilation [44, 53]. Thus, 4WW can provide additional benefit in daily life for patients with airflow limitations.

Outcomes in the effects provoked by 4WW assistance on LEs' control consist on a reduction in knee moment and a reduction of the range of motion of LEs' joints [70]. This can be related to the reduction of the load on the LEs [77]. However there is an increase on the hip joint flexion that can be harmful to the subject, as reported by Ishikura [65]. For some elderlies the 4WW is not considered a safe device. This is reflected on the changes of the spatiotemporal parameters, where their cadence, walking speed, swing time, step length and stride length decreased and stance and double support times increased [77]. However, Vogt et al. [67] and Schwenk et al. [68] reported an increase on the confidence and safety for its elderly subjects. They also reported that 4WW does not interfere with rehabilitation and improves the balance and mobility.

Relatively to the demands on the UEs, the outcomes were that subjects with UEs' limitations should have a higher walker and push it in the perpendicular direction by leaning their upper body on the walker. This is in accordance with Deathe et al. [63]. This finding reveals that maintaining an upright posture and gait pattern characteristics will relieve muscular loads in the UEs [73]. In addition, during assisted gait, UEs play a key role in the control of balance by compensating for the limitations of the LEs [74]. By this, there is the necessity of a correct adjustment of the height of the 4WW and body posture [73].

The attentional requirements to walk with a 4WW can be demanding [46] while performing a second task. However, Miyasike-daSilva et al. [47] and Tung et al. [74] reported that there is a reduction of attentional demand and increased stability of gait in conditions where the equilibrium is 'challenged' (narrow surface).

2.1.2 General Discussion

The aim of this review was to synthesize and discuss research related with walker's evaluation, its effects and demands on its users. After analyzing the 37 selected articles, many concerns were raised about proper use and research of walkers. The studies presented here show that there is no agreed standardized protocol when evaluating assisted gait as evidenced by inconsistencies in distance and time walked, instrumentation, sample size and analytical algorithms. Moreover, different and interchangeable terminology is used. The lack of standardization of measurement protocols and knowledge of gait parameters limits the interpretation and comparison of gait modifications with the use of the walker within the evaluative, diagnostic and prognostic studies.

More important is to verify that most walker users have never been instructed on its proper use and often have a type of walker that is inappropriate [55]. Assistance and training in the use of walkers are essential because inappropriate use is even associated with an increased risk of falls [8]. The selection of a suitable walker depends on the patient's strength, endurance, cognitive function, vision, balance, height, weight, gender and environmental demands [39, 54, 55, 57]. Thus, physicians need to provide walker's checkup for their patients during routine visits. These services should include routine walker maintenance inspections for appropriate walker height, handle-grips, tips of walker legs and brakes of wheels [57]. Moreover, it might be beneficial to increase the amount of time devoted to fitting aids and educating people how to use walkers safely, especially when performing these types of activities [55]. Research is needed to understand the physical and cognitive demands that walking aids place on users. Additional studies are needed to identify potential design problems so as to improve walkers and reduce the incidence of fall injuries in this high-risk population. In addition, specific tests and effective strategies to prevent fall injuries in people who use walkers are needed.

Through this systematic review it was possible to identify a series of limitations, non-studied research topics that need to be highlighted for further studies, since these are still unclear: (1) Posture and Gait pattern, e.g. holding gesture, trunk inclination, amount of supported weight on the walker, sequence of UEs and LEs movements, need to be studied in detail for different type of gait disorders. Crosbie [61] and Smidt and Mommens [60] started researching different gait patterns and Deathe et al. [63] tested different postures. However, these studies were performed with healthy users [60, 61] and amputees [63]. Thus, this needs to be extended for more types of gait disorders to infer which posture and gait pattern should different walker users present to make use of the potentialities of walkers, and help in the walkers selection; (2) Subjects' strength: evaluate the strength needed to maneuver/lift/push the different types of walker, to help in the selection of suitable subjects for each type of walker or conclude that the subject needs a different AD such as cane/crutches or wheelchair. Despite existing studies [66, 71] that identify parameters that can influence the demand on the UEs and point out that one type of walker was not suitable for SCI patients due to the overload on their shoulders [72], there are still no studies reporting the percentage of strength that a user should have to properly execute different tasks with the walker [54]. Only Ishikura [65] performed a study about muscle activity on LEs, but only to infer muscle capacity improvement; (3) Discussion of the biomechanical results: study, evaluate and discuss with clinicians the functional consequences and clinical evaluation of the biomechanical changes and the long-term effects of walker-walking to understand the indications for recommending the use of a walker. None of the selected studies on this review evaluated the biomechanical effects of a long-term usage of a walker, nor with different gait disorders' patients; (4) Limited study sample: the majority of the works present a small number of subjects. The ideal number of subjects should be related to the disorder and consider the respective prevalence of cases in the country where the study was made [81]; (5) Manual records and inspection of video records [54–57]: since this task is time-intensive, automated machine vision algorithms are necessary; (6) No standard analysis techniques and walking courses (walk tests): Through research questions (C and D from Table 2.1), the authors verified that there is a need to define and select the appropriate analysis techniques and ambulatory measurement protocols to observe behavior in its natural context. To control the frequency of activities, environmental conditions, time and distance course, trajectory, sampling frequency, etc. Explore the interactions between intrinsic factors (gait parameters) and extrinsic factors (obstacles) to define and establish specific behaviors (attention, collisions, falls) to form hypothesis. In addition, find tools (data treatment designs) to manipulate and evaluate the relevant factors (parameters, behaviors) that are applicable to everyday life protocols; (7) Lack of variety in the type of studied disorders: Through research question D (Table 2.1) it was identified that the type of studied subjects are healthy young and elder's subjects, PD and COPD patients. There is the need to study subjects with decreased cognitive level (Mini-Mental Status Examination (MMSE)<24) and other gait disorders such as ataxic gait, hip post-surgery, severe myopathy. Defining which disorders are the main targets of walker use and the respective modifications that are expected to be seen. A wide range of outcome measures were used which did not discuss validity and reability for the population under study. Development/adoption of valid and reliable outcome measures for a specific type of population would improve methodological rigor and interpretation of research; (8) Lack of attentional demand studies (this review only identified three articles) related with time of practice: longitudinal examinations of changes in attention as walker users acquire skill at walking with the walker. Provide insights about the effects of practice and understanding about how much time might be required to ensure security. This understanding can help to guide recommendations for follow-up care [45]; (9) Lack of perturbation studies during ambulation; (10) Study performance on real world and outdoor environments; (11) Lack of detail on subjects' demographic data, inclusion and exclusion criteria; (12) No specific gender studies: some studies only address one gender, however do not refer this as an outcome; (13) Limited studies about 3WW and 2WW. Thus, further studies should focus on these current limitations in order to clarify and answer them; (14) Lack of detailed descriptions of interventions made it impossible to accurately document the protocol strategies used; (15) Long-term effects were not investigated.

Moreover, as it was stated before, the results of the current studies suggest that more research is needed to improve the design of walkers. More information is also needed about the circumstances proceeding falls, both to better understand the contributing fall risk factors and to develop specific and effective fall prevention strategies. In addition, proper matching in combination with gait training strategies and appropriate AD progression should help to optimize patient function and independence. Thus, three main steps were considered interesting to be taken into account when prescribing a walker:

1. Select the appropriate device, which depends on patient's strength, endurance, cognitive function, vision, balance, and environmental demands;

2. Give instruction to patients:

    (a) Correct height (define if the patient has UEs or LEs limitations and find a trade-off to avoid extremities' damages);

    (b) Proper use:

        i. Both feet should stay between the posterior legs or wheels;
        ii. Posture upright without forward or lateral leaning;
        iii. Take the time they need to turn;
        iv. Define a proper gait pattern to learn.

3. Monitoring:

    (a) Clinicians should routinely assess whether the device is appropriate;

    (b) Walker maintenance (proper height, verify tips, wheels, brakes, etc).

In addition to addressing these limitations and as a way of standardizing all studies, future studies for further walkers evaluation should include their main goals (article purpose), the criteria for choosing the participants on the study (number of samples, gender, age, body mass, height, inclusion criteria, exclusion criteria, informed consent) as well as the description of instrumentation (all used devices, sampling frequency, devices' models). Then a complete protocol/procedure should be described (conditions, tasks, specify if there was a training session/explanation procedure session/rest interval, walker height adjustment and at which level, test design, conditions execution order, gait pattern, walk test where distance should be specified, time, location, number of trials) as well as the parameters to evaluate (outcomes and how they were obtained specifying the used instrumentation, definition and purpose) and data processing (data filtering, normalization, software). Finally, the statistical tools (procedures, techniques, confidence interval and software), if used, should be presented.

The results of this systematic review have highlighted the need for high quality research investigating the effectiveness of walker-use programs. Specifically, authors should employ a study design with a comparison group and describe the protocol in detail. Clinicians and researchers should work together to design and undertake clinically relevant research projects in this area so that the future medical staff can prescribe the correct walkers which are known to be effective.

With this systematic review, one can unify the current goals and limitations of the studies that focused on walker investigation and highlight them so that the investigation in this area can be more focused on the walker users' complaints, disorders and the design limitations, as well as the necessities of the physicians.

2.1.3 Conclusions

Walkers have high potential to become feasible substitutes for personal mobility-related activities and confidence enhancement during rehabilitation and functional compensation if their use is correct and proper to the user disorder.

Potential future directions are motor rehabilitation with the objective of identifying predictors of rehabilitation outcome and the development of training programs that will potentially involve rehabilitation technology, like smart walkers.

2.2 Smart Walkers: A descriptive review about new robotic funcionalities

Smart walkers (SmartW) (Figure 2.3) started to emerge in rehabilitation research by integrating motors and sensors in a conventional walker. In addition, different funcionalities were developed to be integrated in these devices.

Patients with no cognitive and visual problems have the capacity to guide the walker independently. However, if the patient has cognitive, coordination and/or visual problems, other solutions are necessary. Thus, the ability for navigating in environments with obstacles should one of the functionalities of SmartW to assist and guide its patients. Another problem that arises is the heavy burden on the physiotherapists, both mentally and physically, during the rehabilitation process of patients using a walker. If a SmartW is turned into an autonomous training machine with autonomous navigation for rehabilitation that does not require direct assistance of a physiotherapist, then the burden will be reduced and patients will gain independence. Moreover, the patient can concentrate on correcting his gait through physiotherapy.

Research on SmartW-based systems is also concerned with balance and stability of the device's frame. The system must be secure and stable, not putting the user's health and physical state in danger. Some SmartW addressed this challenge by relying on significant weight to lend stability to the system and by putting the electronics on the lower base of the structure [4]. This requires a design trade-off since lightweight and/or affordable walkers have generally been preferred for their portability and ability to be carried up and down stairs. Also, SmartW may integrate a system that helps its users when sitting and/or standing, in order to provide an extra support care.

The other concern relies in the integration of a shared-control framework, meaning that the SmartW is designed to continuously evaluate and correct its actions based on its perception of the goal and needs of the user. However, corrective actions taken by the SmartW should be in accordance with the users' desires. The user must feel in control of the guidance of the device, by giving him a sense of independence.

Additionally, SmartW research is also focused on the manual guiding and braking system, that has to be easily operated, intuitive and effective, to avoid dangerous situations, such as, great accelerations of the SmartW on descending surfaces, the fall of the user and the possibility that the device may roll away from the user.

Due to the high stability of the device, SmartW can also be introduced on an acute phase of rehabilitation, in which the patient needs to perform the first steps. In such situations it is not possible with conventional walkers.

A new topic has emerged recently concerning the analysis of the user state (gait and pos-

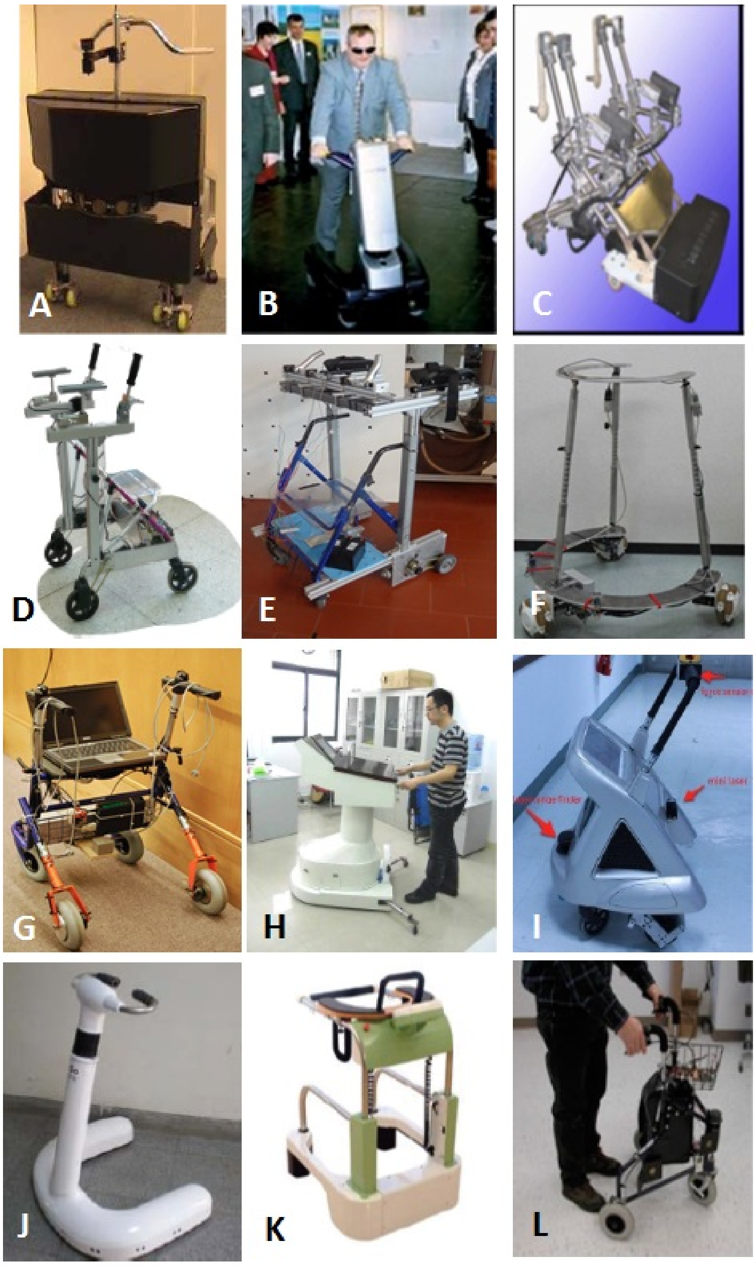


Figure 2.3: Smart Walkers. a) PAMM [46]; b) GUIDO [8]; c)MONIMAD [41]; d)SIMBIOSIS [10]; e)ASBGO [36]; f) JAROW [52]; g)i-Walker[22]; h) Ye et al. [37]; i) CAROW [30]; j) i-go [39]; k) ODW[11]; l) MARC [20].

Table 2.2: General Advantages, Demands and Functionalities of Smart Walkers.

Advantages	Disadvantages	Functionalities
• Motorization; • Dynamic support; • Gait and posture stability; • Require little or no effort from the user; • User-friendly; • Different functionalities; • Adaptable to the user; • Decreases work effort of medical staff; • Higher user independence; • Reduces clinic visits.	• Weight; • Size; • Transport; • Social and medical acceptance; • User clinical validation; • High cost sensors.	• Autonomous control; • Shared-control; • Manual guidance; • Sit-to-stand help; • User state monitoring; • Safety; • Sensorial feedback.

ture) in order to verify the patterns acquired during the assisted gait. This analysis may be fundamental while using the SmartW in a rehabilitation program, being helpful for the physiotherapists and clinicians to obtain clinical outcomes.

Thus, the development of a SmartW presents unique challenges to researchers in this area, mainly to the issues related to specific demands of people with impaired mobility and balance.

In the next subsections some of the main concerns and challenges in SmartW will be presented, giving some studies as examples of such concerns that try to fulfill the demands of conventional walkers to improve the life quality of their users. Table 2.2 lists the general advantages, demands and functionalities that will be presented throughout this section.

The following subsections are divided by SmartW' functionalities: autonomous and shared-control of SmartW, manual guidance, sit-and-stand, security and user's state monitoring.

2.2.1 Autonomous and Shared-Control smart walkers

In case the SmartW has to provide navigation and guidance it should maintain a natural and predictable motion response. Such functionality is important for patients with cognitive and visual problems that do not have the capacity to guide the walker independently.

Thus, they need this functionality to have more independence. In addition, it is very important to give the possibility, from a rehabilitation point of view, for the patient to concentrate on its mobility and gait pattern. Some patients do not have the capability to concentrate on the guidance of the SmartW and at the same time to concentrate on their gait patterns. Thus, the ability for navigating autonomously in environments with obstacles should be added to the functionalities of SmartW to assist and guide its patients during rehabilitation, when needed.

In addition, the burden for the physiotherapists will be reduced.

Many studies [82–98] addressed this functionality in different ways.

The SmartW might only avoid obstacles or navigate to a specified destination (or both). This can be achieved with the implementation of algorithms such as potential fields, like ORTW-II [86] and Graf [93], adaptive control model, such as Tan et al. [87], Clean Sweep obstacle-avoidance algorithm, like GUIDO [84] and VFH+ model [99], like in PAMM [82, 83].

Another possibility to be included in the autonomous SmartW is the shared-control concept [82, 83, 89, 91–94, 97]. This gives the patient some control in decision-making. Since the goals of the user and SmartW may often misalign, the shared-control system must determine whether user or machine yields control. Thus, a machine's manual control interface is motorized to allow a human and an automatic controller to simultaneously exert control [88].

In order to decide whether the user or the embedded controller commands the SmartW movement, haptic and force interfaces might be integrated in the SmartW. By monitoring sensor haptic or force data, it is possible to infer the user intent. Griffiths et al. [88], Morris et al. [90] and Barrué et al. [91] used probabilistic techniques with a shared-control haptic interface. In terms of force sensors, Huang et al. [92] implemented a system based on a heuristic logic that exploits a dynamic model of the walker to detect sliding and loss of walker stability. Graf [93] and Song and Jiang [95] implemented a mass model to generate velocities, depending on the applied force by the user. More than one option of navigation can be integrated. In [95], Walbot SmartW was integrated with three options: obstacle avoidance, wall following and goal seeking through a fuzzy kohonen clustering network fusion. In Wasson et al. [89], it is emphasized that an on-board control system must be taken into account because the more collaborative and loosely coupled relationship between a walker and user, the higher the number of fall episodes. Thus, they integrated several layers of control systems in their SmartW, from simple warning sounds (and no corrective action) to corrective actions that consist of a combination of braking and steering commands away from obstacles, to path planning that gently keeps the user on his way even when no obstacles are present. This last level is achieved with correction times that are long enough and corrective forces that are subtle enough to give the user the impression of full control rather than the feeling of being steered by the device.

Other options are to include walker training programs in the SmartW, like in Tan et al. [87]. The walker has to precisely follow the paths defined in the walking training programs to guarantee the effectiveness of rehabilitation. Also, Barrué et al. [91] and PAMM [82] in their SmartW offer a map of the environment and user's localization and a cognitive support that consists in memory reinforcements, like agenda of daily activities and auditory messages.

However, none of these systems address the safety issue, related to the risk of falls or other mishaps while the user is walking with the device, and, as previously mentioned, such situations have a great probability to occur. Also, some of these navigation and obstacle avoidance systems were not tested in complex environments full of obstacles nor in long courses [84, 87, 91, 93] and others have local minima problems [82] and present oscillations in narrow passages [86].

2.2.2 Manual guidance

If the SmartW's movement only depends on the user, the SmartW exhibits a passive behaviour, and does not proceed to any corrections. In this way, the patient is responsible for deciding the SmartW movement while not getting any feedback from the controller to avoid the obstacles in front of the SmartW. As the movement is defined by the patient, this mode is only recommended for patients with visual and cognitive capabilities, as well as motor coordination and strength to manipulate the handlebar. This functionality only provides for power assistance. Patients that are cognitively able to make command decisions, but do not properly control their walking velocity, gait initiation and finalization nor gait pattern, need to use this functionality.

In this manual guidance, user interaction is usually provided through force sensors. These are normally placed at the handles. This way user's intentions can be easily transmitted through physical interaction. This interface is expected to "read" and interpret all kind of intended motions, to follow the user's movement, and to provide a good walking support.

The idea consists in detecting the intention through different grip forces. When the user wants to go with more/less velocity he grabs with greater/little force, or he pulls/pushes the handle bars, depending on the force sensors integration and configuration on the handles. When the user wants to turn, he exerts more/less force on one of the handles, depending on the side he wants to turn. Projects like [84, 89, 96, 100–102] implemented this type of control.

Walkmate [103] also implemented this concept, but, in addition, he added a negative feedback loop on the SmartW's motion control. SIMBIOSIS [104] and Cifuentes et al. [105] also innovate by integrating force sensors on the forearm supports, detecting the user intentions to guide the walker. This was done by implementing adaptive filters and algorithms [105].

GUIDO [84] in its manual mode provides the user with complete control over the direction of the walker, while the information gathered by the sensors is presented to the user through auditory messages.

A low-cost and alternative way to these force sensors configuration were used by Egawa et al. [106]. The force sensors consist of a pair of U-shaped members joined by four rubber springs and four gap sensors on a U-shaped handle-bar that detect the relative displacements between the arms.

Other low-cost work was based on the use of a joystick [18]. This interface consists on placing, at the centre of the upper base support, a joystick associated with a spring that moves according to the user's manipulation. When the user applies force to the handles, a slight movement is transmitted to the upper base support, mechanically coupled to the joystick that reads the user's intentions. When the user begins his gait, he has to slightly move the handlebar through the handles, moving the joystick to inform the walker which direction and velocity he wants to take. The joystick is a robust and low cost device that does not require excessive use of electronics, and reduces the risk of failure.

Later, Martins et al. [13, 15, 17] proposed a handlebar that incorporates two-axis sensors (potentiometers) to detect the forward and turning forces. The control system uses the forward and turning forces for forward and turning-speed control. With this system, the user can intuitively manipulate the walker at his own pace. If the user pushes or forces to a side the handgrips, the walker moves forward or turns accordingly. It is not allowed to walk backward.

To interpret these interaction signals that transmit user intention, the use of algorithms that can read/classify/determine them are needed. In literature, the most common algorithms apply mass/damped models [96, 107], fuzzy logic control [18, 104], or simple threshold/proportional algorithms [103, 108, 109].

Additionally, in a recent study by Huang et al. [110] using handles force sensors (13 piezoresistive force sensors), were developed. They used Lasso model to infer the relationship between the user's intentions and the measured pushing/pulling forces. Then, a PCA algorithm was applied to obtain the weighting for each intention, from each force sensor. Finally, a fuzzy-neural network controller classified the user intention and determined the proper walker velocity, accordingly.

There are, however, still some concerns regarding this topic. There is still a lack of interpretation and explanation about the signal processing of the sensors, which carry noise and misreads. Also, there is no focus on developing interfaces and controllers that can adjust to the different specifications of each user. How can the different necessities, strengths and patterns of each user/disorder be handle by the walkers? Most of these studies just validated their system with healthy individuals. Frizera [111], for example, validated their processing and control strategies with a target population (Spinal Cord injury). Further validation with the same and other target populations is required using other strategies and SmartW types.

2.2.3 Sit-to-Stand

The sit-to-stand motion demands a lot of effort from the patient. Thus, it is of the most importance to design the robotic assistance device so that this effort is minimised. In addition, this will decrease the working load of the medical staff by freeing them to other technical actions.

Sit-to-stand support can be evaluated by the interaction between robot and human body based on sit-to-stand analysis. In an initial stage, Hirata et al. [112] developed a braking system to ensure that the walker did not move during sit-to-stand assistance. Then, Miró et al. [85] introduced a control that pulled the user in the forward direction. Besides being simple, this solution was not ideal since this motion can be dangerous or unconfortable for the user since it does not provide a natural sit-to-stand motion.

Recently, some studies [113–115] started to implement more complex movements to help in sit-to-stand. In order to implement such movements, sit-to-stand was first studied in the sagittal plane. They agreed that by adding one actuator in the upper part of the SmartW in combination with the forward motion, the SmartW could provide an infinite number of trajectories to assist the sit-to-stand motion.

To determine the correct trajectory, two procedures were employed. Firstly, body postures from different subjects and different trajectories had to be captured off-line using a motion capture system. Secondly, the guiding is based on the observation of the postural state of the user, i.e. imitate the trajectory performed by the user through a developed model.

Studies that followed the first procedure only used one or two subjects. However, Jun et al. [115] concluded that the inclination of the arm support leads to a more stable and natural motion, then, the second procedure requires the development of a model. The problem consists in determining a model of posture that predicts a specific intention of motion, i.e. anticipation of the postural adjustment. To determine an anomaly in the human behaviour, as balance loss, the personal normal postural state has to be known [113]. Pasqui et al. [113] integrated six axis forces/torques sensors on the walker handles to interpret the postural movements in case of incorrect posture, correcting the postural equilibrium. The postural state can be determined by a fuzzy controller [7] or by a sit-to-stand algorithm [113]. Thus, methodology used in this study is user-centred, i.e. specific trajectories are generated for each particular user. However, more validation is required for the parameterization of the model proposed by these latter studies.

Jun et al. [115] also performed the second approach in order to create a model. The required parameters and performance index for sit-to-stand model were evaluated by the analysis of force reflection between the operator and the system. The sit-to-stand motion is generated by the force and moment interaction between the arms of the operator and the supporting platform. A force model was created to then convert force in linear and angular velocities of the walker, describing the interaction. The trajectory of the supporting element is a critical factor to determine sit-to-stand support performance. Two types of trajectories were evaluated, differing in the angle of the body during transition. During the motion capture, the force plate's responses of the seat and foot of the operator and the force/torque sensor responses in the supporting element for arm rest were simultaneously measured. The chosen trajectory was the one that presented better performance in balancing considering the inclination of the body trunk, i.e., where the centre of pressure was the key performance index.

Sit-and-stand approach is still under great attention, since there are many users that need help to transfer to the walker, not having enough strength to stand up alone, i.e. without the help of the physiotherapist nor family.

2.2.4 User's State Monitoring and security

The SmartW can assist the therapis in monitoring the user's motor capabilities and supervising the execution of daily exercises. In general, it is difficult for therapists to continuously attend their patients and the self-assessments of patients are often unreliable, due to poor memory or due to the patients will to avoid therapeutic interventions. Therefore, smart walkers can help therapists to obtain a complete and valid assessment of the user's condition. In order to do so, the walker needs to have the ability to collect and recognize the user activity.

Therefore, these two research topics are based on the type of sensors used and their location. The developed walker systems can monitor user's state through their upper limbs and/or lower limbs. Thus, depending on the localization, different parameter's assessments can be done. The security of the user while walking with the walker can also be inferred by different approaches, focusing mainly on the acceleration of the body and the distance to the walker.

2.2.4.1 Upper limbs: Force sensors approach

The majority of these walker studies focus on developing systems based on force sensors located in the handles [116, 117], or in the frame of the device. These sensors detect the bending force that is applied on the walker [118] and identify the body weight load of the user on the walker. Table 2.3 summarizes the details of these studies.

Alwan et al. [116] implemented a method that passively assesses basic walker-assisted gait characteristics, including heel strikes and toe-off events, as well as double support and right/left single support phases using only force-moment measurements from the walker's handles. The walker is a standard three-wheeled commercial walker augmented with a stepper motor and two 6-DOF load cells, to provide the load/moment transfers between the walker and the user. The author hypothesized that the walker's handles will have cyclic changes reflecting the gait cycle, and from these changes basic gait characteristics could be identified. Results have shown that peaks in vertical direction are related to heel initial contact and in the forward direction to the toe-off event. With the detection of these two events, stride time, double support and single supports were estimated. This information enables control actions to be

Table 2.3: Upper limb studies: Force sensors.

	Alwan et al. [116]	Henry et al. [117]	Abellanas et al. [118]
Sensors' Location	Handles		Forearm supports
Signals/ Algorithms	Force moments / Peak detection	Time and frequency force signals/ K-mean classifier	Forearm reaction forces/ Weighted Frequency Fourier Linear Combiner
Calculated Features	Heel strike and toe-off events, double support, single support	Correlate force signals with gait condition	Cadence and speed
Number of patients	22 healthy subjects	8 subjects with gait disorders	Not specified
Results	97% sensibility and 98% sensitivity	Not quantified	Mean square error= 3.7%
Drawbacks	Expensive sensors		

taken in due time. These passively derived gait characteristics were validated against motion capture gait analysis, testing 22 healthy subjects. Results showed good correlations.

In Henry et al. [117], they used signal processing methods to extract features from the robot's force sensors and correlated them to the subjects' gait condition. Time- and frequency-domain features were extracted and then clustered using a K-Mean classifier. With 8 patients, they concluded that vertical force peak and lateral torque center frequency were the best pair of features. The results indicate that a smart walker can be used as a diagnostic tool that will enable clinicians to monitor from a distance the medical conditions of their elderly patients, thus dramatically reducing clinical visits.

On the other hand, Abellanas et al. [118] calculated cadence and speed through two force sensors placed on the forearm supports of the walker. Through a method based on Weighted Frequency Fourier Linear Combiner, they could infer the user's state, and moreover they could control the SmartW's movement. They presented a mean square error of 3.7% in calculating cadence.

2.2.4.2 Lower limbs: gait tracking

Other potential application of integrating sensors on the walker is to infer the user's legs and feet trajectory to compute gait characteristics and monitor their state.

Some of the existing SmartW aim at tracking the trajectory of gait in order to acquire clinical insight. The great advantage of such systems is that the user stands at a known position with regards to the walker and lower limbs tracking is then made in an easier way. Direct measurement of lower limbs' segments may be obtained with sonar sensors [119, 120], accelerometers [83], laser range finder sensors (LRF) [22, 121, 122], infra-red sensors [123] or cameras [20, 21, 124–126]. Table 2.4 summarizes these studies.

Frizera et al. [120] presented a subsystem to determine the relative position of the user in relation to the walker, monitoring his gait and his safety. Two sonar transmitters were positioned on the user's feet and one sonar receiver was installed onto the walker. They scaned the space between the user and the walker using a direct transmission technique in order to determine the specific spherical coordinate of each leg. The information obtained by this subsystem automatically modulated the velocity of the motors and also stopped the device in case of excessive separation user-walker, avoiding the risks of falling.

Wu et al. [119] tried to develop a walker with dynamic support by identifying a relationship between the person's ankle and the walker position. Consequently, the distance between the ankle and walker was detected and then used to control walker movement. They also used sonar range finders to detect the distance user-walker. Results showed that this walker could keep a constant distance.

In PAMM project [82], the SmartW was integrated with encoders and accelerometers to record user's speed and calculate stride-to-stride variability. Through a power spectrum analysis of speed, user's stride length and frequency can be computed, as well as the gait asymmetry. However, these approaches require adding markers to the patient. Other approach can be based on LRF sensor without the need of adding any markers on the patients' limbs.

The RT-Walker [112] is equipped with a LRF and performs an estimation of the kinematics of a 7-link human model. The model is only used to estimate the position of the users' centre of gravity (CoG) in 3D. A LRF acquires the position of the knee with regard to the walker. Despite the fact that the model had already been tested in the real world environment, no real walker users tested this system.

The JaRoW (JAIST Active Robotic Walker) [121, 123] was developed to provide potential users with sufficient ambulatory capability in all directions and easy-to-use features. In 2010, a preliminary walker prototype was developed which integrated a pair of rotating infrared sensors to detect the location of the user's lower limbs. Based in the results, two main control algorithms were proposed to estimate the location of the user's lower limbs in real time, and allow the users to walk naturally. In addition, it enables the JaroW velocity to be controled automatically. However, results showed that the walker was moving intermittently and the proposed algorithms that were not able to deal with the nonlinearity of the human gait and with the fact that gait parameters vary across users.

An upgrade was made and instead of infrared sensors, LRFs were integrated into the SmartW [121]. In this new study, Kalman and particle filters were applied to estimate and

Table 2.4: Lower limb studies: gait tracking.

	Frizera et al. [120]	**Wu et al. [119]**	**Spenko et al. [83]**	**Lee et al. [121]**	**Cifuentes et al. [105]**	**Martins et al. [22]**	**Pallejá et al. [122]**	**Hu et al. [124]**	**Paolini et al. [126]**	**Joly et al. [125]**	**Martins et al. [20, 21]**
Sensors' type and location	Ultrasonic sensors / Ankles .		Accelerometers on feet and wheel encoders	Laser range finder sensor pointed to the legs				Camera			
Signals/ Algo-rithms	time-of-flight		power spectrum signal/ peak detection	Raw laser signal / self-developed algorithm				Color image / particle filters	Depth image	Depth image / Segmenta-tion, model parameteri-zation and Kalman filter	Depth image / Segmenta-tion
Calculated Features	Feet distance to walker / fall risk		Stride length, gait symmetry and frequency	Legs' distance to the walker	Legs' distance to the walker	Spatiotemporal parameters	Legs' distance to the walker	Step width and length	Feet position and orientation	Gait phases and feet position and orientation	
Number of patients	N/A		N/A	1 healthy subject	N/A	10 walker users	6 healthy subjects	N/A			3 healthy and 13 patients
Results	N/A				Errors were lower than 10 %	Average error of 10% for leg's estimation	10 mm error of distance	Error : 33.6 mm for step length and 25.5 mm for step width	Position error: RMSD: 4.9 to 12.1 mm in x and z, 19.4 to 26.5 mm in y. Orientation error: %RMSD 5.6% to 8.8% in x and z, 15.5% to 18.6% in y.	5 degrees of error	Error : 28.5 mm for step length and 17.6 mm for step width
Drawbacks	User needs to wear the sensors			Expensive sensor				High processing cost and needs markers	Use of markers attached to the feet	High processing cost	Error when patient is too close to the camera

predict the locations of the user's lower limbs and body, in real time. Both filters showed very good estimating results. Despite the good results, the algorithm was only tested with one subject, and this does not prove that it is efficient for different subjects. In addition, they did not perform a gait analysis study. Also, in Lee et al. [121], the authors state that since elderly people with poor posture are the ones who tend to lean their upper body on the upper frame, a more sophisticated controller should be developed to cope with unpredictable changes in the JARoW dynamics.

In Martins et al. [22], a leg detection method to estimate legs position during assisted walk, detects gait events and calculates the corresponding spatiotemporal parameters. The method is based on the detection of the legs' patterns to calculated the position of each leg and then calculate the spatiotemporal parameters for user's state monitoring. Preliminary results obtained on ten walker users show that relevant data using a LRF can be extracted for gait analysis with small error.

In Cifuentes et al. [105], an interface that extracts navigation intentions from a novel combination of two sensors were developed. A LRF sensor estimates the users legs' kinematics and a wearable Inertial Measurement Unit (IMU) sensor captures the human and robot orientations.

However, the use of LRF system may fail in the case of having large pants or skirts, which will lead to false detections and make the algorithm impracticable. A possible solution is to adjust the LRF sensor base to capture the feet movement. However, Pallejà et al. [122] showed that foot detection with laser leads to incorrect gait measurement.

Thus, new studies appeared suggesting a camera approach for feet detection.

Paolini et al. [126] showed that virtual reality for the provision of motor-cognitive gait training may effective for a variety of patient populations. The interaction between the user and the virtual environment is achieved by tracking the motion of the body parts and replicating it in the virtual environment in real time. They presented the validation of a novel method for tracking foot position and orientation in real time, based on the Microsoft Kinect technology, to be used for gait training combined with virtual reality. The validation of the motion tracking method was performed by comparing the tracking performance of the new system against a stereo-photogrammetric system used as gold standard.

In Hu et al. [124] a camera is mounted on the frame and observes markers on the toes. This marker based toe tracking algorithm allows calculating step width and length and provides an accurate assessment of foot placement during walker use. However, the user needs to wear markers, which makes this approach uncomfortable to the users. In addition, it is a solution unviable to be used in a daily routine.

Joly et al. [125][125] uses a Kinect sensor for biomechanical analysis to measure and estimate legs and feet position during an assisted walk. Despite being a good approach, the actual set up was unable to capture the data during all the phases of a cycle gait. This is due to the range of the sensor which is not able to deal with very close data and its orientation on the system. Monitoring legs and feet will allow an estimation of the gait characteristics and information that other systems cannot provide.

As these two methods use legs, they require two separated sets of points for legs, thus large clothes and skirts will lead to false detection. Moreover, because of complex segmentation, the image processing is long, not allowing a real time processing of data. In addition, no tests were performed with actual walker users as elderlies.

In [20, 21] it is presented a better visualization of both feet in all phases of gait cycles and presents a much more simple and effective approach for feet tracking. To improve the reliability against environmental conditions (especially clothes), we propose to extract main data about the walk (feet position and bearing angle) only by segmenting feet. By this way, the processing time is reduced and the algorithm can be used in real time application. When [124] has a processing time around 15s, the proposed algorithm [20, 21] takes around 0.1s to process with similar computers.

2.2.4.3 Upper limbs and Lower limbs

In other studies [127, 128], in addition to the force sensors, they have integrated other sensors on the SmartW to try to infer more information about the whole body, in an attempt to monitor the user's state. Table 2.5 summarizes these studies.

Sinn and Poupart [127] developed a four-wheel rolling walker equipped with four load sensors (one in each leg) to measure the ground reaction forces, a wheel encoder to measure the walking distance, a 3D accelerometer to measure the instantaneous acceleration, and two video cameras (facing forwards and backwards, respectively) to record the environment and the position of the lower limbs relative to the walker. In order to recognize user's activities from the sensor measurements, they presented and evaluated different methods based on Conditional Random Fields. Experiments with real user data showed that these methods achieve good accuracy (85-90%) to detect different activities like sitting on a chair, not touching the walker, walking forward/backwards and execute left/right turns. In addition, the authors raised fundamental questions for their future research, such as, how the accuracy achieved by the algorithms compares to the agreement among different human labelers. In cooperation with clinicians, they proposed to evaluate how much accuracy is actually needed for providing robust measures, e.g., of the user's stability, and the development of strategies to deal with the basic uncertainty about the "true" user activity.

Postolache et al. [128] proposed a SmartW (without motors) based on technologies ex-

Table 2.5: Lower limb and upper limb studies: whole body analysis.

	Sinn and Poupart [127]	**Postolache et al. [128]**
Sensors' type and location	Load sensors on each walker leg, wheel encoders, 3D accelerometer	Microwave Dopler radar, accelerometers and flexible force sensors
Signals/ Algorithms	Ground reaction forces, acceleration / Conditional Random Fields	Frequency signals
Calculated Features	Different user's activities (turn, sit, lift, walk...)	Walker usage, upper kinetics and kinematics
Number of patients	12 healthy young subjects	Not specified
Results	85-90% of accuracy	Not quantified
Drawbacks	It does not provide stability to the user	Not specified

pressed by microwave Doppler radar, accelerometers and flexible force sensors and Bluetooth communication that can remotely collect data on walker usage, upper kinetics and kinematics during walker-assisted gait. It allows automatic or semi-automatic gait analysis that can be used by physiotherapists to extract information related to patient's functional disability in walking as a result of motor or sensory dysfunction, but also to evaluate the gait recovery progress.

2.2.4.4 Monitoring user's safety

A very important aspect of SmartWs is to provide security so that the user feels safe while controlling the SmartW. Otherwise, the user will not use this device and resort to others devices such as wheelchairs.

Different forms of feedback were already presented in the previous sections to inform the user about his/her state [83, 108] and safety [15, 108, 112]. This can be achieved through haptic sensors, auditory and visual information. Table 2.6 summarizes the forms of feedback.

Some actuation from the SmartW can be performed [15, 108, 112] in order to avoid dangerous situations.

In PAMM project [83], the SmartW can record the user's activity level which over time

Table 2.6: Feedback Types.

Feedback types	Functionalities	Studies
Auditory	Agenda information, Medication, Warning messages, alarms	[82, 84, 89, 91, 129]
Haptic sensor	Collision risk	[88, 89, 91, 129]
Visual	Gait information, maps, localization	[82, 91, 93, 129]

can help the physician to better monitor the user's health. ECG-based pulse monitor was used to monitor the user's state. It also records user's speed and calculate stride-to-stride variability, as well as gait asymmetry. Both variability and asymmetry are indicators of physical injury and predictors of fall. However, this recorded data is only to inform physicians, no actuation is performed by the SmartW.

In Kai et al. [108] typical gait disturbances of parkinsonian patients and a walk supporting monitoring system suited for such gait disturbances are evaluated. They also presented a model of walk supporting and monitoring system tested with five healthy subjects and one Parkinson patient.

Major characteristics of gait disturbances in parkinsonian patients are anteriorly tilted posture with little arm swinging, short strides, quick short steps, and reduced lifting of the toes from the ground. All of these characteristics make them prone to stumbling. Typical symptoms include frozen gait, hesitation to start walking, and festination and pulsion symptoms. Thus, the walking aid must prevent such symptoms. The SmartW design includes forearm supports in order to be possible to measure the anterior tilting of the posture. Furthermore, force sensors are applied to detect the intention of the user to move. By forcing a fixed speed, the walker can prevent festination, pulsion symptom and prevention of falling. Such fixed speed is set when the patient exceeds the threshold force on the handles and his distance to the walker is safe. Detection of anterior tilting of the posture and brachybasia are done through position sensors. These sensors detect the relative distance between the patients' feet and the walker, and stop the device if needed (to prevent the patient from falling when his feet were left behind the device). Detection of frozen gait and hesitation to start walking is done according to pressure information on the plant of the foot, measured through pressure sensors.

These functionalities collect the required data such that the medical staff can evaluate the state of recovery of patients and use it for guidance in rehabilitation.

In RT-Walker [112] three different states are inferred: a walking state, a stopped state and an emergency state. A LRF acquires the position of the knee with regards to the walker. User velocity is estimated from the walker velocity obtained by encoders. The stopped state occurs when both the walker and the user velocities are zero. To distinguish the walking state from the emergency state, user-walker distance is used. A normal distance is determined to infer the walking state that differ from the emergency state.

In ASBGo project [15, 18], additional safety was envisaged by several sensorial subsystems that complement each other and can free medical staff. Such safety is ensured by a acting brake. To detect possible forward falls of the user it monitores the approximation of the user with infrared sensing at the height of the chest. If the user is falling forwards, the distance between the user's chest and the walker decreases. An algorithm was developed to detect abrupt changes on the signal, to then detect if the user is falling forward and stop the walker in time accordingly. To detect if the user is falling backwards three procedures were introduced. First, the walker cannot move backwards. So, if the user pushes the upper structure in his direction, the walker stops. Another subsystem ensures the user is guiding the walker grasping the two handlebars. The proposed safety system is compounded by two force sensor resistors, one on each handlebar. If the two handlebars are not being held by the user, the walker will immediately stop. The third procedure is based on two force sensor resistors, one on each forearm support that will verify if the user is with his forearms properly supported on the base supports.

Thus, the security of the SmartW is based on the inference of different states that differentiates a normal non-dangerous situation from a risk situation. These states help the braking system of the SmartW to be quick, efficient and independent of the user's reaction time. In addition, there should exist a concern about backward movement. Such "direction" should be avoided, or done with supervision of a third person.

2.2.5 Discussion and Challenges

A SmartW should provide support whenever required, and it should be an easy-to-use device, presenting the following features: (i) provide dynamic support whenever the user is walking, standing or sitting, and provide a relatively stable support for the user to recover from losing balance, (ii) require little or no effort to use, i.e. to move and change direction, (iii) to be user-friendly, the movement speed and direction is controlled by the user subconsciously, not requiring special training.

Research on autonomous and shared-control systems has been extensively described. This is a very useful functionality for the SmartW, since, for example, the elderly have cognitive impairments, often forgetting their goals and localization. It is also possible to include the participation of both parts (user and device) in achieving a goal task. Other problem is when patients have visual problems, requiring an autonomous help, that can, with safety, drive them to their goal. Navigation algorithms and techniques are being improved, however more real experiments with real users are necessary in the current state of art. Moreover, most SmartW only provide obstacle avoidance when moving forwards. Such point needs to be addressed in terms of the safety of the patient. In addition, for the safety of smart walker users, the backward movement should be avoided, or done with supervision, since generally walker users do not have the capability to walk backward without falling.

A problem that can be risen in smart walkers is the sensor coverage. Relative to wheelchairs, walkers present the advantage that sensors can be placed in front without concern for blockage by the user's legs; but coverage in the rear can be problematic. Through mechanical and design solutions, such problem may be handled.

Another concern, that is still a major challenge, is the user adaptation. The key problem is choosing, implementing and validating algorithms for the recognition of user activities from the stream of sensor measurements that can effectively detect and interpret user's intent, comfort and sense of control.

Other factor that remains a challenge in current research is, for example, system's validation. This is, in general, done by simulation or with people with no gait dysfunction nor cognitive impairments or blindness. Results are not sufficient to prove the effectiveness of the system, since research is interested in addressing the market to compete with the conventional walkers.

Another problem is the employed protocols for system validation. These protocols are created for validation of user-friendliness and user's comfort but have to be standardized, such that researchers can compare their works with each other. There is also a lack of published work on clinical outcome measures.

As a clinical and diagnostic assistant, the SmartW should provide clinicians with longitudinal data of the physical conditions of walker users. In contrast to tests performed in a clinical setting, this data should be collected in the users' everyday-life environment and over continuous periods of time. Thus, a smart walker can be used as a diagnostic tool that will enable clinicians to monitor at a distance the medical conditions of their patients, dramatically reducing clinic visits.

Also, SmartW should present different options, either mechanical/structural or electronic, to give different possibilities for the physiotherapists to work with their patients, with more quality and optimal recovery results. Another possibility is to control the SmartW remotely, where the physiotherapist can test different velocities and directions during the therapy sessions, analyzing the behavior of their patients. Moreover, the patient can concentrate on correcting his gait through physiotherapy. This latter was not addressed by any study and should be addressed on future studies to verify its potential and importance during the rehabilitation process.

Moreover, to meet usability and compete with the conventional walkers' market, SmartW' investigation has to use low cost but effective sensors, improve algorithms, actuators, walker design and user interface showing they are safe to use. Most people do not like walkers because of their size. SmartW may present a bigger size than conventional walkers, complicating social acceptance. In a first stage, authors think that SmartW should be used in a clinical environment. Comfort, stability and safety provided by SmartW may be a deal breaker for some users. This can turn SmartW more accepted than conventional walkers. It is important to state that many devices are only prototypes and aesthetics, an important factor in any commercial product that is not addressed.

Other problem may be the weight and transport of the SmartW. Such problem must be handled by designers and mechanical engineers by creating a SmartW model that can be easily transported. In terms of weight, even wheelchairs are heavy and are transported by patients. Thus, a SmartW can be equally heavy, if its transport is possible to be carried out.

The high cost of a SmartW compared with a conventional walker is impossible to escape. However, such comparison is the same as motorized and non-motorized wheelchairs. The comfort and quality of treatment may be better guaranteed by a SmartW. Its functionalities increase its cost, but improves the quality of life of its user. Some patients are not capable of using a conventional walker, thus they have no other option than to use a wheelchair. With the extra functionalities given by a SmartW, such patients may have the opportunity to walk and do exercise. It targets a population with severe problems that needs rehabilitation or functional compensation. It even targets specific diseases with particular motor problems. In the long run, these other design considerations should be tackled. Also, other less expensive alternative methodologies and technologies should attempt to replace the more expensive components such as the LRF.

Researchers have to demonstrate to clinicians that this can be a powerful tool to help them in diagnostic, better rehabilitation process, and so on.

Therefore, if the developed technologies are to gain user acceptance and widespread adoption, control interfaces must be intuitive, seamless, and non-obtrusive. Component advancements will achieve seamless and non-obtrusive interfaces. Control algorithm advancements will achieve intuitive control. However, only persons with disabilities can provide specifications for intuitive, comfortable and easy design. If we do not make consulting persons with disabilities a priority, we will not meet the demands of the end user and the technology will be abandoned.

There is still a long way to go with this assistive device before its commercialization. However, the authors think that the first product must be simple with few functionalities and sensors to first gain medical and user acceptance. Then, the necessary upgrades will be made.

2.2.6 Conclusions

Even though major market segments of assistive mobility technology products are based on communication and vision aids, the mobility aids are becoming more important, since the society is ageing and technology needs to help them to improve their quality of life, autonomy and the efficacy of rehabilitation efforts.

Walkers are devices with great potential for rehabilitation, helping in terms of stability and mobility. Advances in these devices can achieve transformative changes in mobility and decrease cognitive efforts.

However, there's still a lack of acceptance and adoption by the patients of the developed devices. This survey addressed the requirements that SmartW have to address in order to strengthen their position as rehabilitation or functional compensation tools. The survey allowed to conclude that the design has to be attractive, ergonomic and comfortable. Control interfaces must be intuitive and non-obtrusive. In addition, there are still missing studies considering persons with disabilities, which are the only that can provide this feedback and specifications. To improve and achieve total acceptance, a continuous user involvement is essential, ensuring that the developed devices match user needs and desires, as well as capabilities.

Chapter 3

ASBGo walker – Project and Functionalities

This thesis is part of the development project of a 4-wheeled motorized walker (smart walker). This project, included in the working group - ASBG (Adaptive System Behaviour Group) - culminates in the development of a smart walker (SmartW) with the ability to ensure safety and natural handling of the device to the user. This device is equipped with multiple sensors, and the selection was based on simplicity of implementation and diversity of functionalities. The use of multiple sensors allows the walker to provide information about the user gait pattern, identify the movement intentions by evaluating the direction and speed, and ensure security conditions by detecting possible falls. Thus, the final stage of the project is to develop a product with realibility that may help improve subject's health conditions during their rehabilitation period, providing daily exercise and better quality of life [4, 9].

A SmartW is intended to be a device that can act as a versatile rehabilitation and functional compensation tool. It should be adaptive considering the necessities of its user and its use should be safe. Patients present different necessities according to their intrinsic characteristics, their disorder and therapies. In order to help them, a SmartW should provide different functionalities.

This chapter aims to present the project in general, focusing on design considerations, walker's system, the implementation of four different operating modes, and, finally, an overview of the gait assessment tool development.

3.1 Design considerations

For the creation and development of a medical device such as a SmartW, it should be taken into account for whom it is intended. This brings crucial characteristics and limitations to the development of the final prototype. Therefore, it is important that first of all a list of goals is specified before any other point of prototype creation is set.

The first goal is to guarantee the safety of the device to its user. The walker should be robust and reliable in order to reduce to the maximum any risk of injury to its user. Second goal is the attractiveness of the device, which means that it has to be economic and comfortable. Other goal is to provide multifunctionality to the walker, being adjustable to the user and able to incorporate and solve various problems such as being motorized and help its user in various tasks (e.g. sit and stand from a chair). Also, the SmartW's design must be suitable to the aim of use, i.e. as a functional compensation and rehabilitation tool. Thus, the device must have an ergonomic design that can provide the necessary support for the patient's treatment. Since it will be a product for mass manufacturing it should be noted that its design, assembly and manufacture has to be as simple as possible, thus saving not only costs to the manufacturer, but also to be less confusing and expensive for the user. Finally, its use must be practical, easy to transport, store and adjust.

With these goals well-defined, it is necessary to find and define the main functions and requirements that the SmartW should present.

In figure 3.1, it is presented the main functions that are proposed to integrate in the SmartW. These main functions are structure, motor conection, sensor location, adjustments, extra-help components. Each main function has several sub-functions that were considered throughout the project. In this way, several options were considered to be designed and developed, so the designer could get a better sense of the most reliable option for the final prototype. Options not considered during the project were rejected due to the price, availability of manufacturing processes and limitations imposed by existing machinery in the factory.

So, in order to understand which is the best way to create the defined functions, it is necessary to define the main requirements. These requirements will help in the development of several hypotheses for a possible final prototype, with the creation of different prototypes. To set such requirements, a team of medical staff, patients and ASBG members were consulted, and the final list was defined as: to allow a comfortable position to the patient; to allow a gait without obstructions; to ensure the successful implementation of the associated electronics; to have easy access to the various electronic components; to have easy and intuitive use; to have suitable dimensions for hospital use; to be stable enough to support partial body weight.

It is also important to note that this process of selecting the necessary functions and requirements for the development of the SmartW took into account the disadvantages presented

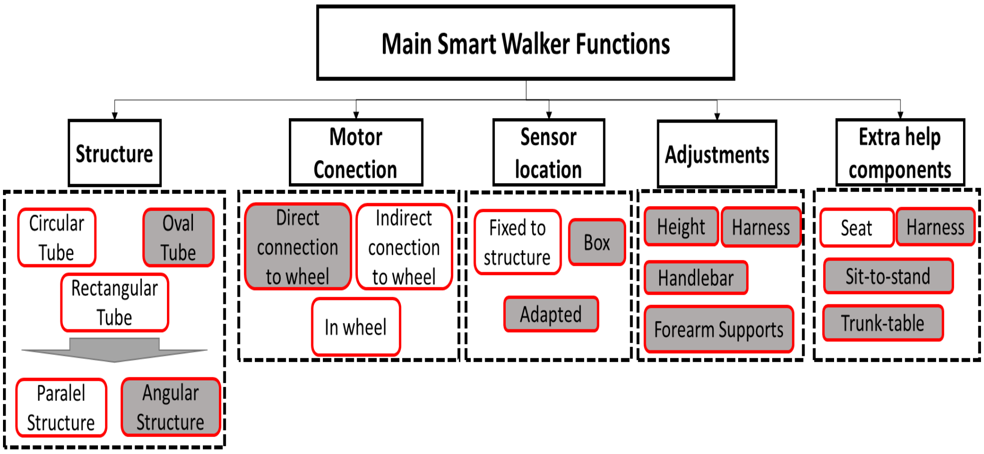


Figure 3.1: Main functions proposed to the SmartW prototype. The dark boxes represent the final decisions.

by conventional walkers. One of the main problems observed in the conventional four-wheeled walkers is their instability, fragile structure and low weigth. These disadvantages may cause problems such as falls rather than help in the treatment of patients. Throught the identification of these problems and all aforementioned topics, the team was able to initiate the development of the SmartW.

The evolution of the project is shown in figures 3.2, 3.3 and 3.4 in terms of the different prototypes that were created until the final version. First the name ASBGo (Assistance and monitoring System Aid) walker was given. Then, the initial version of the ASBGo (Figure 3.2) was projected as a proof of concept in order to verify some requirements and functions. The structure was too rudimental, made of rectangular tubes and parallel structure, needing improvements, and composed by iron materials, which are very heavy. The motorization system of this SmartW was done by two independent motors and a system of pulleys and belts for each rear wheel (indirect conection to wheel). Its front casters freely rotate. Sensor locations were tested, mainly the handlebar mechanics and electronics, however the components were fixed, and a more adjustable position was required. A handlebar integrated with a joystick was developed [18, 130], however it presented many problems and disadvantages, since this approach uses rubber springs. Because of that, the system presents a delay and hysteresis behavior, which decreases the precision of the system. The offset should be zero or around zero, which sometimes did not happen, making the walker to move arbitrarily without manipulation of the user. Since the handlebar constitutes the direct contact/interaction between the SmartW and the patient, it should present a reliable functioning. Thus, a new approach was studied and it will presented in this section.

Then, height adjustment was integrated but it was too rudimental. Forearm supports were considered very important for a stable support of the patient, however better and more confortable supports were needed.

Thus, the first main considerations and modifications for the creation of the second prototype were the height adjustment, extra-help components such as a harness support, confortable forearm-supports, attractive design, easy to store and transport.

The second version (Figure 3.3) was designed with a circular tube base with a parallel structure that can pass through any environment (elevators, doors, etc) and to have a small area to have an easy storage. However this latter characteristic turned out to be a bad option for its users since most of them present a gait with a wide base of support, making them to trip over the walker structure.

The motorization system of this SmartW was the same as the first prototype. Also, this device has only a type of adjustment (height), that it was soon realized that was not enough, as there are several factors that differ from person to person besides height. One factor that soon was identified was the need for level adjustment of the forearm supports and distance to the handlebar. This lack of adjustment led to the limitation of some important wrist movements to guide the walker as well as the disconfort to be supported on the walker. A box compartment was placed on the front of the walker in order to store the electronics and integrate the necessary sensors.

Another concern was the SmartW materials. The inferior part was composed by steel, since is a low price material, rigid and ease of production; and the top and the front were constructed of aluminum because of its lower electromagnetic driving, not damaging the electronic components.

This second prototype was tested with different patients and important modifications were set for the third prototype (Figure 3.4).

First, a more robust and stable structure was necessary to give a greater sense of confidence and safety to the user. In terms of the base structure, oval tubes were designed and instead of parallel tubes, they were angled in 10º for each side. After extensive field research (Chapter 2) and several discussions with medical staff and physiotherapists of the Hospital of Braga it was possible to conclude that the users of walkers, especially users with ataxia and cerebellum lesions, tend to have a wider gait base of support. After measuring different patients, one concluded that in the mid stance phase of gait a width exceeding 52cm was necessary. However, on the other gait phases, such width could be more narrow. Thus, a triangle shape base was designed.

With regard to the motor position, in this version it was directly coupled to the wheels (direct conection), to save space and give more torque to the wheels.

One of the major goals of this project is to make a multifunctional and adaptable SmartW to users with different degrees of disability and different body structures. This, it became necessary to allow multiple adjustments that do not concern only with an adjustable walker height, as the vast majority of existing conventional walkers. There are several points that can vary from user to user, as the distance between the arms, forearms and the length of the distance from the shoulders to the elbows. In view of these points, the design of the SmartW considered some characteristics that could enable all these adjustments.. A lateral adjustment was added to the handles as well as an adjustment of the distance to the walker structure. These settings are important since it was observed in the second prototype an incorrect distance among support arms and handles depending on patients. This often prevented an adequate rotation and flexion of the wrists as required to control the walker.

Two types of grasping and support were set: forearm support with vertical handgrips, for users with extension problems on their arms; horizontal handgrips for users with shoulder problems (Figure 3.5). The configuration of the handles can provide adequate stability levels and may also be used in man-machine interactions, such as detection of users movement intentions (details in section 3.2). Also for better support, more ergonomic and rigid forearm supports with the possibility of being adjusted in both width and length by a Velcro system were created (Figure 3.5). These supports have also the possibility to be integrated with sensors, as it will be detailed in the next section.

The required box compartment to accommodate all the electronic part of this walker was also modified and it was designed to be easy to integrate into the structure and be aesthetic and functional, allowing easy access to electronic components. Also, a structure to hide wires was added as well as a protection to encoders and handlebar (Figure 3.6).

Another aspect that was observed in some patients was the assymetry of support in the walker. They have a tendency to choose one of the arms and therefore have decentralized gait forcing on one of the upper limbs, creating an incorrect and harmful posture. Therefore, an abdominal surface area with a curvature in the contact area with the user was added to center the user and correct his posture, independently of his anatomy. Such surface is presented in figure 3.5 and was built of wood because it is a cheap material, attractive and easy machining.

Some extra-help components were also added. In order to give more autonomy and safety to patients it was added two bars with handles on the back of the walker to assist the transition of sit-to-stand. A harness struture was also reinforced to help some patients with higher balance disorders and weak lower limbs. This will force these patients to maintain the correct position while walking, increasing their level of confidence, and giving them a greater support

Figure 3.2: First ASBGo prototype.

to avoid undesirable imbalance and falls. Finally, it should be mentioned that the seat was removed since it caused obstructions and restrictions to gait.

3.2 Electronic System Overview

The final version of ASBGo walker (Figure 3.4) was integrated with multiple sensors and other electronic components given it different functionalities and characteristics that are presented in table 3.1.

Two motors drive its right and left rear wheels independently. Each rear wheel is installed

Figure 3.3: Second ASBGo prototype.

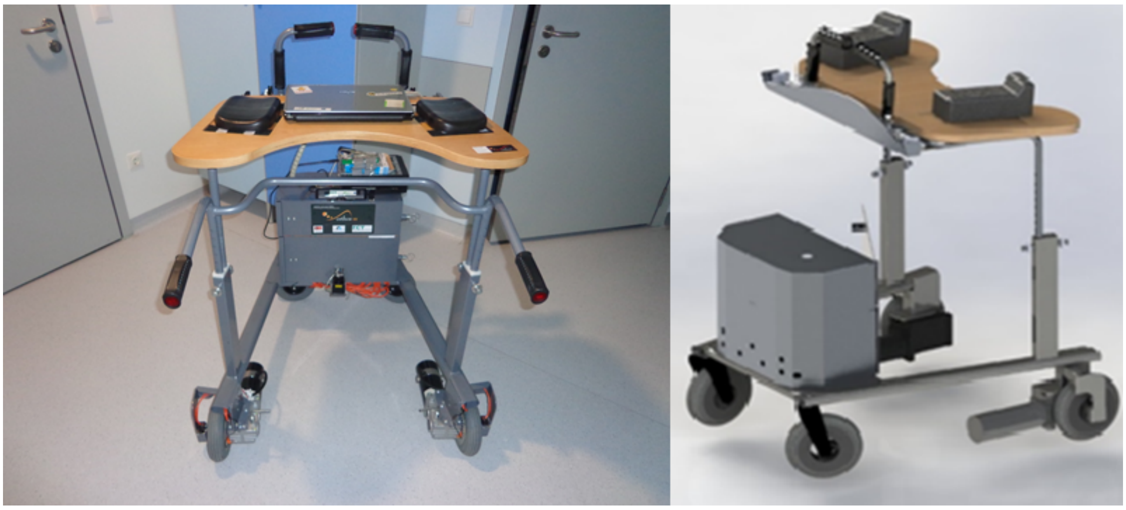


Figure 3.4: Third ASBGo prototype.

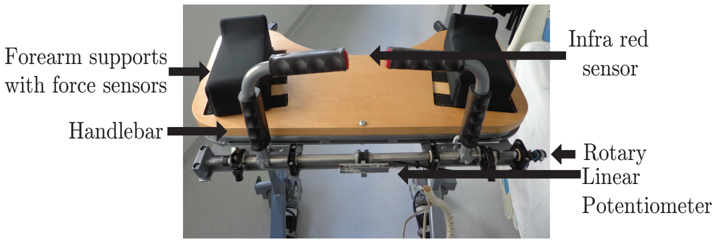


Figure 3.5: ASBGo handlebar

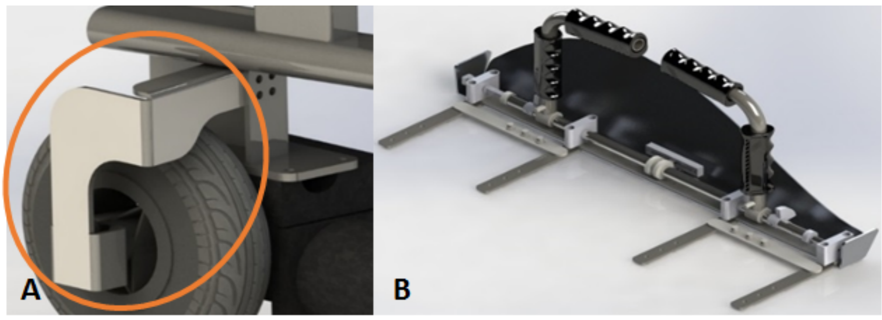


Figure 3.6: Protection of A. encoders and B. handlebar.

Table 3.1: ASBGo charateristics and functionalities.

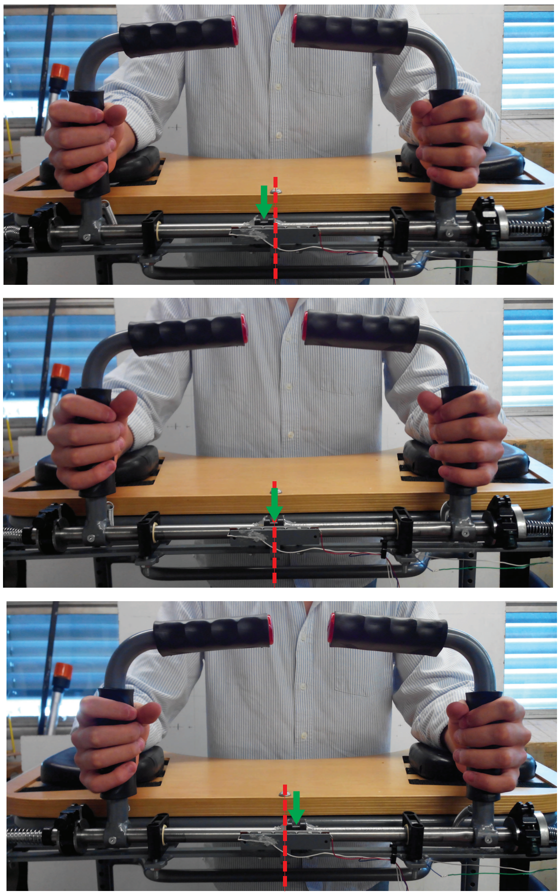
Walker	ASBGo
Type of Device	Motorized Walker
Target Population	High balance disorders and coordination problems
Key Functionalities	Intention recognition, adaptation to the user, obstacle avoidance, gait pattern evaluation
Modes	Autonomous, manual, secure and remote control
Physical Interaction	Potentiometers
Steering	Motorized rear wheels
Indirect Interaction	Force resistive sensors, Infra-red sensor, Laser range finder, Active depth sensor and Sonar sensors
Safety Functionalities	Braking and falls detection
Communication and Programming	Arduino Platform and Portable computer

with an encoder. The electronics and heavy components were installed in a lower level of the walker to improve the general stability of the ASBGo.

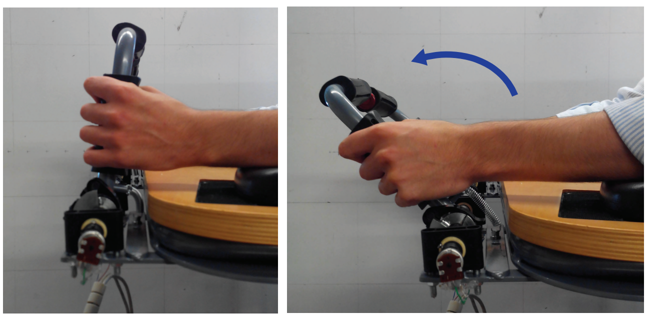
The handlebar (Figure 3.5) acts as a direct interface with the SmartW and is based on low cost electronics composed by potentiometers. To acquire user's commands, the proposed handlebar has two potentiometers to detect the forward and turning directions. The control system will use these forces for forward and turning-speed control. Thus, two commercial potentiometers were embedded into the handlebar: a linear potentiometer (0-10kΩ linear) to detect directional changes in speed and a rotary potentiometer (0-470kΩ linear) to detect forward changes in speed. With this system, the user can intuitively manipulate the SmartW at his own pace. If the user pushes or forces to a side the handgrips, the SmartW moves forward or turns accordingly. The SmartW interprets these two basic motions and controls the motors speed and direction, accordingly. It is not allowed to walk backwards. The explained movements of the handlebar are illustrated in figure 3.7. The handlebar has translational motion to the left and right, as the green arrow represented in figure 3.7 indicates, and has rotational movement represented by the blue arrow. These sensors will be actuated by the user to command the walkers movement [14, 15, 17].

For safety measures, uni-axial force sensors were installed in the forearm supports as shown in figure 3.8. Such sensors detect possible instabilities and falls from the user (details in section 3.3.3).

The walker also has 9 sonar sensors distributed in a three layer configuration to maximize the detection area (see configuration in figure 3.9). A low ring of 6 sonars mounted forward-oriented detects the majority of ordinary obstacles, like people, walls or other low obstacles.



(a)



(b)

Figure 3.7: Schematic conguration of the two movements of the handlebar: a) linear and b) rotary potentiometer.

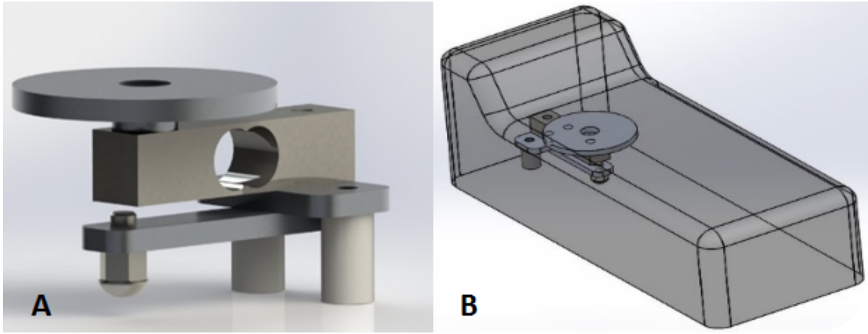


Figure 3.8: A. Uni-axial force sensor strutucture installed on B. the forearm supports.

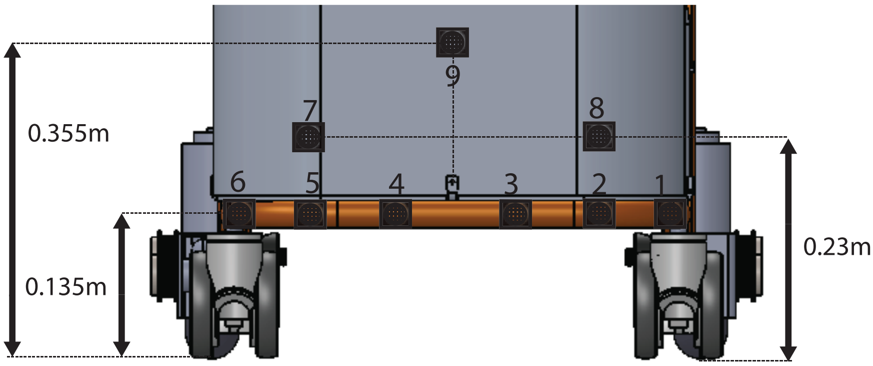


Figure 3.9: Frontal view of ASBGo. Conguration of the sonar sensors (Low ring, High ring and Stairs sonar)

High obstacles such as tables or shelves are more difficult to detect than ordinary obstacles since their support to ground can be undetected by the forward oriented sonars. They can lie in front of the walker and provoke a collision. Thus, a high ring of 2 sonars pointing upwards is mounted to detect high obstacles. These 8 sonars are meant specifically for obstacle avoidance. An extra sonar pointing downwards is mounted on the walker to detect stairs. This sonar does not contribute to the obstacle avoidance task, but stops the walker when changes in the ground, such as stairs or holes are detected.

Other important requirement of a SmartW is the possibility of doing clinical evaluation during walker-assisted gait. This is the first step to assess the evolution of a patient during rehabilitation and to identify his needs and difficulties. Advances in robotics made it possible to integrate a gait analysis tool on a walker to enrich the existing rehabilitation tests with new sets of objective gait parameters. The team of this study developed a legs detection method to estimate legs position during assisted walking using a Laser Range Finder (LRF) and a feet detection method was developed with an active depth sensor. More details will be presented in section 3.4.

Besides acting as a gait assessment tool, ASBGo also has four operating modes: autonomous mode, manual mode, safety mode and remote control mode. The autonomous mode allows the user or the physiotherapist to set the desired position to which the SmartW should autonomously move while avoiding any obstacles in the environment. The manual mode is characterized by the SmartW's movement under the guidance of commands defined on the handlebar. As the movement is defined by the patient, this mode is only recommended for patients with minimum visual capacities and/or cognitive, that have sufficient motor skills on the upper limbs. The safety mode is characterized by a warning system that alerts the presence of obstacles in front of the walker as well as the monitoring of users fall risk. However, the SmartW's movement is controlled by commands set by the patient, as in manual mode. Finally, remote control mode has been developed in order to allow the physiotherapist to control the orientation and velocity of the SmartW. Physiotherapist have here the opportunity to examine the behavior of the patients and possible gait reactions and corrections from the patient to different directions and velocities given by him. These operation modes are presented with more detail in section 3.3.

With such functionalities, ASBGo is versatile, adaptive and safe as a rehabilitation and functional compensation device for patients with mobility problems prescribed for its use. Versatile since it can be used for a variety of patients that present difficulties in mobility associated with other personal limitations such as visual problems and/or cognitive). Adaptive since it allows adapting the parameters of control systems (such as minimum and maximum speeds) depending on the physical limitations of the patient. Safe because the structure of the presented SmartW was developed with a design that provides for a more stable movement and safety for the patient.

3.3 Operation Modes

The main goal of the developed SmartW (ASBGo) is the rehabilitation and functional compensation of patients with mobility and balance problems. Since patients can present different types of difficulties and disorders associated with locomotion, the SmartW has to be adaptable to these different limitations. Thus, through four operating modes is possible to adapt the operation of ASBGo depending on the difficulties of the patient and provide a safer, comfortable and efficient rehabilitation. This section presents the operating modes of ASBGo: autonomous, manual, safety and remote control modes. Figure 3.10 schematizes the four operation modes.

	Operation Modes			
	Autonomous	Manual	Safety	Remote Control
Goals/Target patients	• Decrease physiotherapist work effort; • To monitor the patient's behavior; • Ideal for patients with visual and motor coordination problems.	• Double-task training; • Recommended for patients with visual and cognitive capabilities and enough motor coordination to manipulate the handlebar.	• Provide safety such that the user feels safe while controlling the smart walker; • Low visual and cognitive capabilities.	• The physiotherapist analyzes behavior, compensations and reactions against sudden changes in speed and orientations; • It allows the patient to focus on his gait pattern; • Patients in initial stages of walker use.
Functioning	The physiotherapist indicates a target location for the walker. It will avoid obstacles that appear along the way. Neither the patient nor the physiotherapist have control over this mode. Decisions are made by the walker itself.	The walker is conducted independently by the user. Commands are defined on the handlebar.	The patient guides the walker and a warning system is activated if a dangerous situation is detected.	The physiotherapist defines the commands to control the smart walker's movement (speed and orientation)

Figure 3.10: Four operation modes: Autonomous, Manual, Safety and Remote Control.

3.3.1 Autonomous Mode

Autonomous mode allows the user or physiotherapist to define the desired position coordinates while guiding the SmartW in the environment.

This operation mode is suitable for patients with visual and/or cognitive limitations, or/and cannot control the SmartW manually due to weakness or lack of upper and lower limbs coordination. In the case of locomotion recovery in the hospital, the physiotherapist initially defines the type of training (with different targets to be achieved) and the walker starts the process without any intervention of the patient. The locomotion recovery continues without the need for outside help, such as physiotherapists or family. Simultaneously, the autonomous mode allows monitoring the patient's behavior, so that the physiotherapist can assess his progress in recovery. To turn the ASBGo autonomous is necessary to integrate a module to ensure obstacle avoidance and movement to the target. A local navigation module based on a Nonlinear Dynamical Systems Approach (DSA) [131] was implemented.

In [16], our team presented an obstacle avoidance technique for ASBGo based on DSA and in [132] the stability of this approach for obstacle avoidance was addressed. Different simulation environments were tested. The simulated hospital environment contained typical obstacles usually found in real hospitals, such as beds, wheelchairs, litters, food trays, etc. The SmartW was not provided with the location ofknow the distribution of the obstacles over the hospital environment. Good performance of the SmartW was achieved when navigating in pure obstacle avoidance mode and when combining obstacle avoidance and target attraction. Results demonstrated that the sonar configuration mounted on the SmartW had successfully detected several types of hospital obstacles, including dynamic obstacles. In addition, the ability for detecting high obstacles, ramps and stairs was also simulated and performed with success. This is innovative since had not been considered in previous works [131, 133]. Thus, with DSA the SmartW successfully avoided obstacles with safety and smooth movements.

After implementing DSA in simulation, real experiments were performed in a lab environment (http://youtu.be/wfmFeA60B0o). Before testing with patients it is fundamental to verify how the system behaves in a real environment. Thus, in order to be faithful to a Hospital environment, three different experiments were performed, with both static and dynamic obstacles. Results showed that the ASBGo behaved very well in all experiments, being efficient in all situations.

Finally, the ASBGo was brought to the hospital for the final tests with patients. In (http://youtu.be/LQdsFFZCiJw) it is possible to watch some seconds of the autonomous mode with a patient. After moving for 30 minutes with different patients and velocities, the autonomous mode showed to be successful in detecting all obstacles. It is noteworthy that sometimes individuals appeared in front of the walker, provoking an emergency stop of ASBGo. This stop was efficient and did not put the patient in danger. Also, stairs were also put on the trajectory of ASBGo. When the stairs were closed, the ASBGo stopped in time, since the stairs' ring detected with enough margin distance the stairs, not putting the patient in danger. Patient's and physiotherapists opinions were very positive about this operation mode. Patients said: "I felt very safe being conducted by the walker" and physiotherapists said: "This mode is very useful and it seems to work well on the environment where we execute the gait training. With this mode we can observe the patient and it gives more independence for the patient".

This mode will not be further discuss since it is not in the scope of this these. Details can be found in [134].

3.3.2 Manual Mode

The Manual mode is characterized by the movement of the ASBGo under the guidance of commands defined on the handlebar. In this way, the patient is responsible for supervising the ASBGo movement while not getting any feedback controller to avoid the obstacles in front of the smart walker. As the movement is defined by the patient, this mode is only recommended for patients with visual and cognitive capabilities, as well as enough motor coordination to manipulate the handlebar.

After placing the hands on the two handgrips, the user will act on them accordingly to the command he wants to perform: start to walk, accelerate, slow down and turn left or right. Thus, if the user intends to: (1) increase the walking speed, he has to turn the handlebar in a counterclockwise direction (Figure 3.7b); (2) decrease the walking speed, he has to turn the handlebar in a clockwise direction (Figure 3.7b); (3) turn to the right, he must move the handlebar to the right side (Figure 3.7a); (4) turn to the left, he must move the handlebar to the left side (Figure 3.7a).

The pre-processing of both potentiometers is presented in detail in [13, 15]. In order to achieve adaptation to different patients due to different arm strengths, the sensors are calibrated for each patient. The authors predict that patients with greater strength and coordination prefer a higher range of velocities, while patients with lower coordination or tremor, prefer a lower range of velocities on the handlebar to gain more stability and to feel more secure. Thus, the physiotherapist can change the range of velocities, as well as the maximum velocity that the SmartW can move. This is done by means of a graphic interface (Figure 3.14a)

The control strategy for this operation mode is based on fuzzy logic to classify the signals sent by the potentiometers and transform them into motor inputs, in such way that the SmartW drives the motors according to the user's commands [15].

Comparing the situations illustrated in figure 3.11 with the data from graphs of figure 3.12, it can be seen that the experiment starts with increasing angular position of the potentiometer (see the situation of figure 3.12a) which results in increasing the linear velocity of the ASBGo, since both motors equally increase their speed (see graphic shown in figure 3.12b). From t = 13s, the position of the potentiometers is constant, i.e. the ASBGo keeps its movement straight forward. The situation B (Figure 3.12) represents the time interval in which the angular potentiometer maintains its position, however, the linear one shifts to negative (see figure 3.12a), which changes the movement of both wheels. Since the right motor speed is higher than the left motor (see figure 3.12b), the ASBGo with turn left. Between the t = 22s and t = 26s, ASBGo movement is the same as described in situation A, where the ASBGo motion has constant speed. The situation C represents a change of direction to the right side, because the linear potentiometer has a positive value and the left motor increases its speed at time as the right motor reduces. In the final part of this experience, t = 37s of figure 3.12a, the potentiometers are in position zero and the motors speed drops to zero, stopping the SmartW. Therefore, this operation mode allows the SmartW's movement to be controlled by the patient. However, this mode is prescribed by the physiotherapist only for patients without visual and/or cognitive difficulties, with motor co-ordination and sufficient strength to the manipulation of ASBGo handlebar. During rehabilitation, this mode can be used in a later stage, for instance. In (http://youtu.be/LQdsFFZCiJw) it is possible to watch some seconds of the manual mode with a patient.

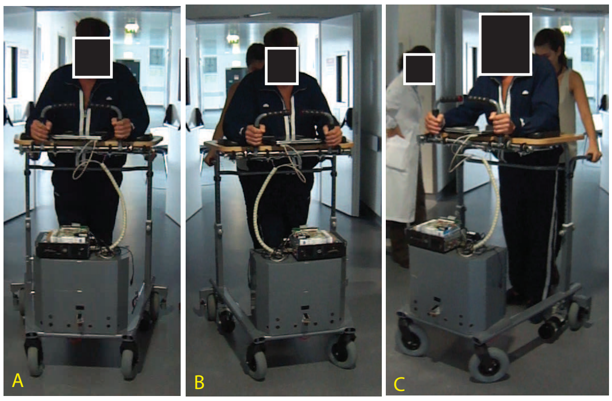
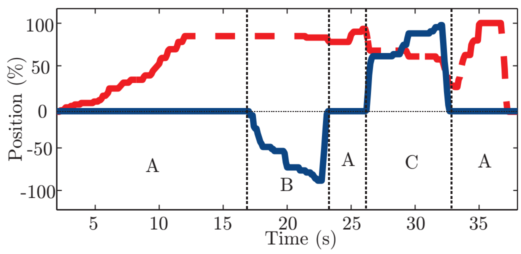
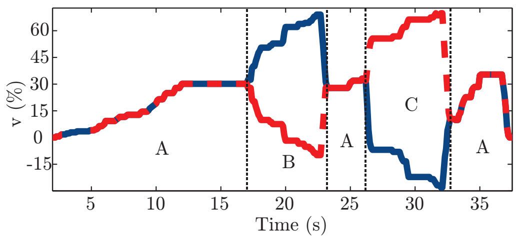


Figure 3.11: Patient controlling the movement of the ASBGo through the handlebar. A: Walking forward; B: turning left; C: turning right.



(a)



(b)

Figure 3.12: a) Percentage of rotary potentiometer movement (red dashed line) and percentage of linear potentiometer movement (blue line). b) Percentage of left motor (red dashed line) and right motor (blue line) velocity.

This mode will not be further discussed since it is not in the scope of this thesis.

3.3.3 Safety Mode

A very important aspect of SmartWs is to provide safety such that the user feels safe while controlling the SmartW. Otherwise, the user will not use this device and resort to others devices such as the wheelchairs.

On the ASBGo safety mode, the patient guides the SmartW and a warning system is activated if a dangerous situation is detected. Both the environment and the patient are monitored.

The monitoring of the environment is characterized by a warning system that alerts the presence of obstacles in front of the SmartW. The warning system of this operation mode consists of three led lights: green, yellow and red. Figure 3.13 represents three situations detected by the safety mode. The green light is lightened in the absence of obstacles in front of the SmartW, i.e. when all the sonar sensors measure a distance greater than a predefined minimum distance (*mindist*). When SmartW is at a distance of less than a pre-defined maximum distance (*maxdist*), the red light is activated to warn the patient that there is an obstacle near the SmartW. The yellow light signal is triggered when SmartW approaches obstacles in the environment. This light maintains its state if at least one of the ultrasonic sensors detects a distance between *maxdist* and *mindist*. Depending on the patient type, there may be the need to adjust *maxdist* and *mindist*. The physiotherapist has the possibility to adapt these parameters, because patients with longer reaction time need more time to avoid the obstacles. Additionally, an audible alarm system, with different sound frequencies associated to these different distances, may also be triggered if the patient is visually impaired. Another option, is to stop the device if a pre-defined *mindist* is exceeded.

On other hand, the monitoring of the patient is characterized by a group of sensors that monitor the risk of fall. This latter type of monitoring of the safety mode will be presented in detail in chapter 4.

The user behavior will be analyzed in terms of distance to the walker, support loading and motion intention. For this, three sensors installed on the upper base of the SmartW will be used: infrared sensor (IR), force sensors and potentiometer (Figure 3.5). The ideia is to infer different user states and accordingly with the detected state, one actuation of the SmartW is performed in order to avoid a fall.

3.3.4 Remote Control Mode

The remote control mode was developed to allow the physiotherapist to monitor the user behavior and control the velocity and orientation of the SmartW accordingly (Figure 3.14a). In

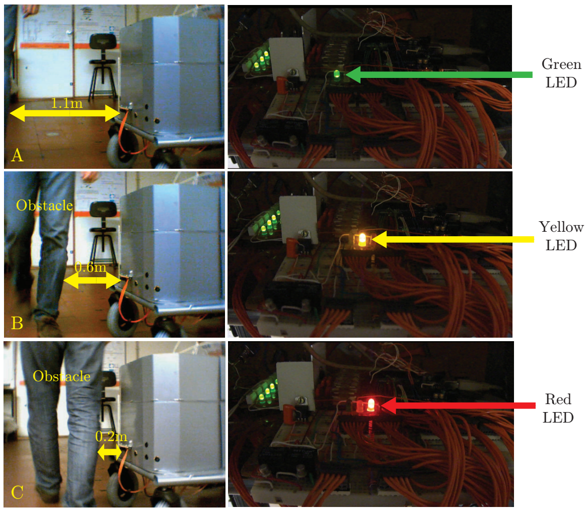
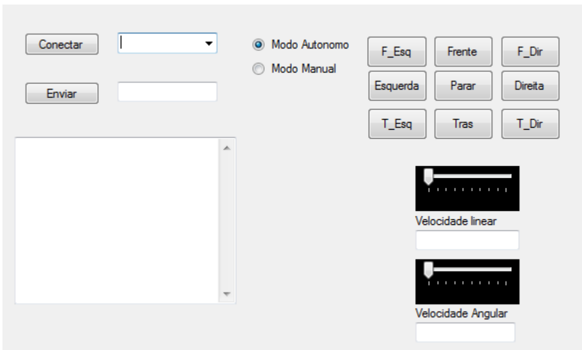
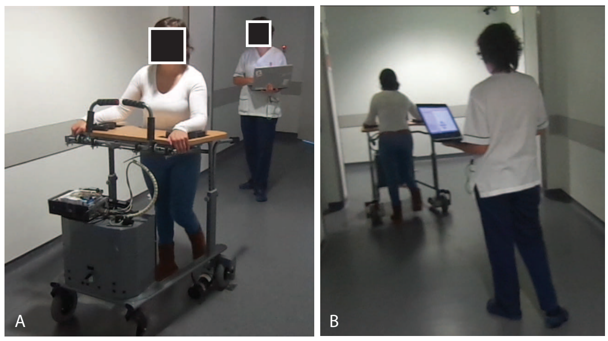


Figure 3.13



(a)



(b)

Figure 3.14: a) Remote Control interface and b) Physiotherapist controling the ASBGo at distance.

this mode, the physiotherapist analyzes the behavior, compensations and reactions of the patient against sudden changes in speed and orientations and defines the commands to control the SmartW's movement. In addition, it allows the patient to focus on his gait pattern and balance and not on the guidance of the SmartW. A graphic interface (Figure 3.14a) was developed to allow the SmartW remote control. It can be used on a portable computer or on a smart-phone (Figure 3.14b).

Due to the insecurity of patients in the movement to the rear, in this operation mode the movement of the walker to the rear is possible. However, this feature requires the supervision of a physiotherapist to ensure the safety of the patient.

In addition to this remote control, a feedback feature was added. This feedback consists on showing the feet movement and/or trunk of the patient while using the walker (Figure

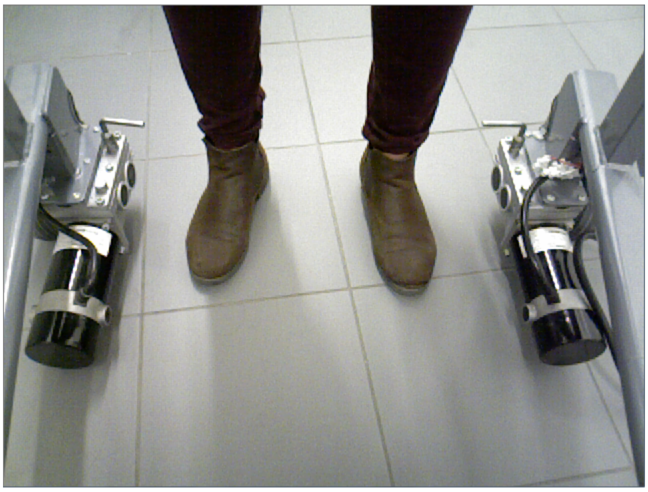


Figure 3.15: Feet Feedback

3.15). With such visual information the patient demonstrated will have the opportunity to auto-correct his movements, having the sense of his problem and solving it automatically.

In (http: //youtu.be/LQdsFFZCiJw) it is possible to watch some seconds of the remote control mode with a patient.

3.4 Gait Evaluation system

A SmartW is not only a device to give support and guide its user. It should also have the functionality of evaluating the recovery of its user. By this, a gait assessment and analysis tool was integrated on ASBGo.

Precise motor function evaluation in rehabilitation programs is a major challenge in clinical practice and has gained widespread interest with recent technologies. Further, assistive device rehabilitation is becoming popular, since these devices present characteristics that make use of the residual capabilities of its users, maintaining and enhancing motor strength capabilities.

Nowadays, in physiotherapy, physiotherapists evaluate motor function and patients' performance based on visual information and personal expertise about the movement patterns. Such information is qualitative in nature and the final clinical decisions are strongly empirical and subjective. This evaluation can be more objective and quantitative, if it applies gait evaluation techniques that allow a systematic study and characterization of the human locomotion.

Thus, gait analysis has been an important research field for rehabilitation purposes. A common technique in gait analysis is gait tracking and there are many precise systems with such purpose, e.g. optical motion capture systems. However, these systems may present occlusions when acquiring data, need a considerable workspace, require expensive processing devices, and demand dedicated personnel to conduct the measurements. Therefore, often these systems are limited to special laboratories for analysis, which results in both economical and practical disadvantages.

Nowadays, many studies focus on research of gait tracking with portable sensors placed on the subject. Inertial [135] and pressure/force sensors [136, 137] are very well known examples to measure joint rotations, dynamics and spatiotemporal parameters.

Regarding gait tracking systems integrated on external devices like walkers, there are a large variety of examples [4, 9]. This research is very important since clinical evaluation of walker users is the first step to decide the degree of assistance they require. This evaluation is only performed once and by observation, using standard scales and questionnaires [138]. Advances in robotics made it possible to integrate sensors on conventional walkers to act as portable gait analysis systems. Further, these systems allow evaluating the evolution of some disorders and enhance diagnostics in ambulatory conditions.

A sensory system that captures the relatives evolutions between the lower limbs of the user and the walker, providing information related to gait pattern and stability for further clinical evaluation was developed in this thesis. Such system is composed by two sensors and will be the focus of chapter 4. One sensor is a Laser Range Finder (LRF) that captures the legs' movement regarding the walker. Then, an active depth sensor caputes the feet of the user. These two systems can work separately depending on the disorder and/or clothes of the walker user, or together through a sensory data fusion system. With these systems it is possible to identify gait events in order to calculate the corresponding spatiotemporal parameters. With these parameters, it is possible to calculate stride-to-stride variability, which is a strong indicator of risk of fall. Other important indicator is the symmetry of parameters. This can tell us if the coordination between legs is improving or not.

Besides the lower limbs, it is possible to monitor the trunk in terms of stability and postural control. This system is composed by one acceleromenter that will also be presented in chapter 4.

With this main functionality as a gait evaluation tool, the ASBGo walker can give important information about the user's state to the physiotherapist. With such information the physiotherapist can verify quantitatively the evolution of his patient, can decide which treatment is more appropriated and which are the actual movement problems that have to be enhanced and exercised.

Chapter 4

Gait and Posture Assessment and Analysis System

Clinical gait analysis is the process by which quantitative information is collected to aid in understanding the etiology of gait abnormalities and in treatment decision-making[139]. Such analysis is often divided in (1) patient qualitative observation, (2) a description phase and (3) a biomechanical analysis. Description phase and biomechanical analysis are facilitated through the use of technology such as specialized, computer-interfaced video cameras to measure patient motion, electrodes placed on the surface of the skin to appreciate muscle activity, and force platforms imbedded in a walkway to monitor the forces and torques produced between the ambulatory patient and the ground [140]. Since most clinics and hospitals do not have access to this equipment, simple and portable gait analysis systems should be available [39, 141]. Such availability of equipment is important to allow an objective assessment of a person's functional physical state [39].

This problem affects the clinical gait analysis in training with walkers. Thus, research has been focused on this problem by addressing the characterization of human gait parameters and other aspects with the use of walkers [8].

In order to well diagnose and follow rehabilitation with the use of a walker, a gait assessment system has to be accurate but also affordable to reduce unequal access to health care and to improve clinical follow-up (i.e. to allow to be used in physiotherapists'/physicians' office). Similarly, the gait assessment system should be portable and adaptive to the majority of walkers. The system should be contactless to be used in daily routine, improve comfort and decrease the time of analysis. If the system enables realtime analysis, data could be used directly during the consultation by the physician and eventually at home for motivational purposes or monitoring the quality of walk (to predict any forthcoming deterioration of a user's gait). Commercial motion capture systems do not fit with the previous requirements. They are expensive, not portable and use markers [142]. In addition, in assisted gait, occlusions often occur. Thus, inclusion of embedded and portable systems on the walker seems to be more appropriate for building a gait assessement system to characterize and analyze walker-assisted gait.

Advances in robotics made it possible to integrate sensors on conventional walkers to act as portable gait assessment systems. Further, these systems allow evaluating the evolution of some disorders and enhance diagnostics in ambulatory conditions.

However, the majority of these walker studies focus on developing systems based on force sensors located in the handles [103], or in the frame of the device to detect the bending force that is applied on the walker [62] to identify the body weight load of the user on the walker.

Thus, this chapter intends to present different assessments at different points as follows: a) the pattern followed by feet and legs and b) the variables to evaluate body balance of a walker user. These assessments will give the necessary information to spatiotemporal evaluation, posture and fall risk estimation.

Different systems were created and will be presented on the following subsections.

One active depth sensor (ADS) will track the feet (section 4.1), to provide position and orientation of the feet center and one Laser range finder (LRF) will track the legs (section 4.2). After validating these systems, a method for spatiotemporal parameters calculation is proposed (section 4.3). However, since ADS and LRF systems are prone to errors, a multi-sensor data fusion based on these systems will be presented (Section 4.4). Other system that will be presented is the use of one accelerometer placed at the trunk to indicate the stability of the user regarding his centre of mass (COM) position, giving posture and balance information (section 4.5). Finally, besides analysing the gait pattern, it is important to monitor the user safety while walking with the walker device. For this force and infra-red sensors will infer different security states of the user in order to alarm him and advise for dangerous situations (section 4.6).

4.1 Feet Position Tracking with an Active Depth Sensor

4.1.1 Active Depth Sensor System

Recently, technologies using depth images have considerably evolved. Kinect and Xtion PRO Live sensors, the main cameras referenced on literature, reached an acceptable price compared with stereo vision systems [143], with similar results. These sensors use a weak infrared laser to project a predefined pattern of dots of varying intensity [143]. This pattern provides a source of easily extracted features. The variation of these features compared with the known

Figure 4.1: Xtion Pro Live sensor

pattern for a fixed distance provides a method for depth reconstruction, giving a depth map. In addition, this depth map is not sensitive to light changes.

Thus, in this study a Xtion PRO Live sensor (www.asus.com), shown in figure 4.1, an active depth sensor (ADS), is used to capture the position of feet placed on a walker-type device. In figure 4.2 it is shown the sensor integrated in different walkers.

It has a resolution of (640x480) pixels, field of view of 58° H, 45° V, 70° D (Horizontal, Vertical, Diagonal) and 16 bits depth precision. It processes 30 frames per second (33 ms per frame), time that meets the requirements for the measurement of parameters associated with the pathological human gait [140, 144].

The specific location/orientation requirements are such that it has be at the center of the walker capturing the feet in the coronal plane and to have a good visualization of both feet over all gait cycle, i.e. minimum distance of 0.4m (sensor specifications) between feet and camera field of view. The sensor should also presents an inclination, whose angle depends on the walker (structure and design) where it is integrated.

The Institute for Intelligent Systems and Robotics (Marie et Pierre Curie University, Paris, France), AGATHE group, and the ALGORITMI research center (Minho University, Guimarães, Portugal), ASBG group, collaborated in a research project based on the integration of this ADS on a smart walker. AGATHE purpose was to develop an ADS based control and ASBG intended to use this sensor for gait monitoring purposes. To achieve these two goals, it is first necessary to evaluate the efficiency of the feet position estimation with an ADS placed on the smart walker. As a result of this collaboration, two scientific contributions were published [20, 21].

4.1.2 Brief Review of Feet' Tracking Methods with Active Depth Sensor

Paolini et al. [126] proposed real time feet position and orientation traking for treadmill use. The camera was placed 1m above the ground and 1m in front of the treadmill (which is within the range of operation of the depth sensor of the Kinect) so that it had an unobstructed view of

(a) (b) (c) (d)

Figure 4.2: Different walkers with ADS system. a) ISIR's smart walker. b) ASBGo. c) and d) two four-wheeled walkers.

the treadmill belt surface. In addition, a marker was placed on each foot. Their position errors are lower than 27mm (Root Mean Square Deviation - RMSD). Their orientation error (bearing angle) is below 10% RMSD and it is calculated between two fixed points of the marker. They obtained 3D orientations of the feet, but angles around vertical axis should be sufficient for gait analysis and control purposes. Although these results are good, the proposed method is not suitable for our application since it is based on the use of markers attached to the feet.

Two teams proposed the use of a camera depth sensor without markers.

Hu et al. [124], placed a Kinect under a four-wheeled walker, at the center capturing the legs in the coronal plane. They proposed to estimate 3D poses from depth images of the lower limbs in the coronal plane in a dynamic, uncontrolled environment. They employed a probabilistic approach based on particle filtering, with a measurement model that works directly in the 3D space and another measurement model that works in the projected image space. Only position errors are reported by the authors (less than 60mm) and they are considerably bigger than the ones obtained with markers (27mm) [126].

Joly et al. [125] proposed a 3D skeleton on the partial Kinect data. The sensor is also placed on a four-wheeled walker in the axial plane. Despite presenting the same principle as in [124], in [125] the model is simpler, representing the two legs as two rigid bodies linked with a ball joint and the correspondence with the model is made with the depth map. Simmilarly orientation error is reported by the authors to be less than 15º. Such orientation is calculated through the lines that represent the feet. However, due to occlusion problems, data are not available in all the gait cycle phases.

As these two methods [124],[125] use legs, they require two separated sets of points for legs, thus large clothes and skirts will lead to false detection. Moreover, because of complex segmentation in both studies, the image processing is long (more than 15s) and cannot currently be implemented in real-time, making it unsuitable for control and gait analysis applications.

Based on these studies, the ADS system, in this section, was placed in a specific position of the walker, as shown in figure 4.2, to have a better visualization of both feet in all gait cycle phases. A new algorithm was implemented to present a simpler and effective approach for feet tracking. To improve the reliability against environmental conditions (especially clothes), it is proposed to extract feet data by segmenting the feet. Since no model will be used, it is believed that the time processing will be reduced. This redution will allow to apply this method in real-time applications, such as gait analysis. Also, it will be possible to integrate this system in different walkers, which means that the system will be independent regardless of which device is placed. Such consideration was not detailed, nor tested, by the previous studies.

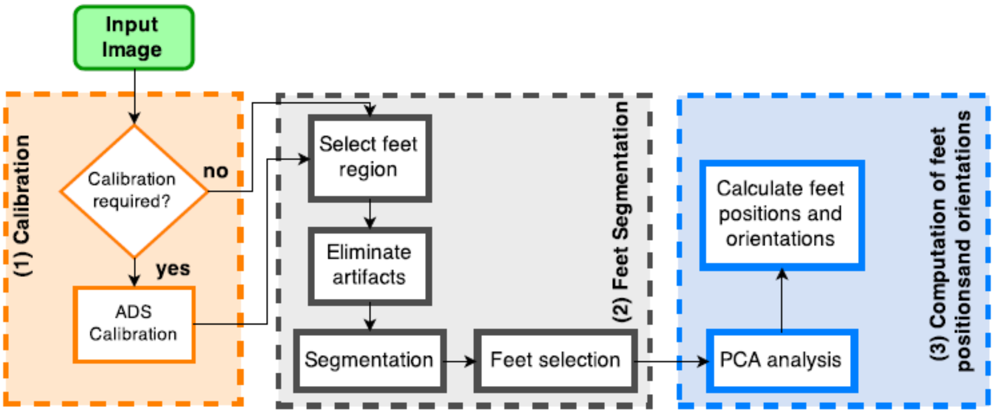


Figure 4.3: Schematic of the proposed feet detection algorithm.

4.1.3 Algorithm for Feet's Tracking

In order to detect and track the feet with the active depth sensor (ADS), three main steps have been implemented: (i) Calibration, (ii) Feet segmentation and (iii) Computation of feet distances and orientations. The workflow is presented in figure 4.3. The algorithm and image processing was done using OpenCv library.

4.1.3.1 Calibration

The calibration aims at converting (u,v) and depth coordinates on the ADS frame (X,Y,Z) into $(X_w,Y_w;Z_w)$ coordinates of points in meters in the walker frame (see figure 4.4). The calibration of ADS for feet tracking is done in 2 steps: Calculation of (1) ADS intrinsic parameters and (2) extrinsic parameters. Step (1) is related to the characteristics of the ADS, thus it is only necessary to do it once. Step (2) depends the orientation/position of the ADS, thus it is necessary to perform this step every time the camera changes its orientation/position.

(i) Camera Intrinsic parameters

To express the point coordinates in meters in the ADS frame, it is necessary to know the intrinsic parameters of the ADS, gathered in the following matrix:

$$\mathbf{F} = \begin{bmatrix} f_x & 0 & c_x \\ 0 & f_y & c_y \\ 0 & 0 & 1 \end{bmatrix} \tag{4.1}$$

where f_x and f_y are the focal lengths and c_x, c_y the coordinates of the center of the image. To get this matrix, the OpenCV method of calibration was used with the RGB image of the

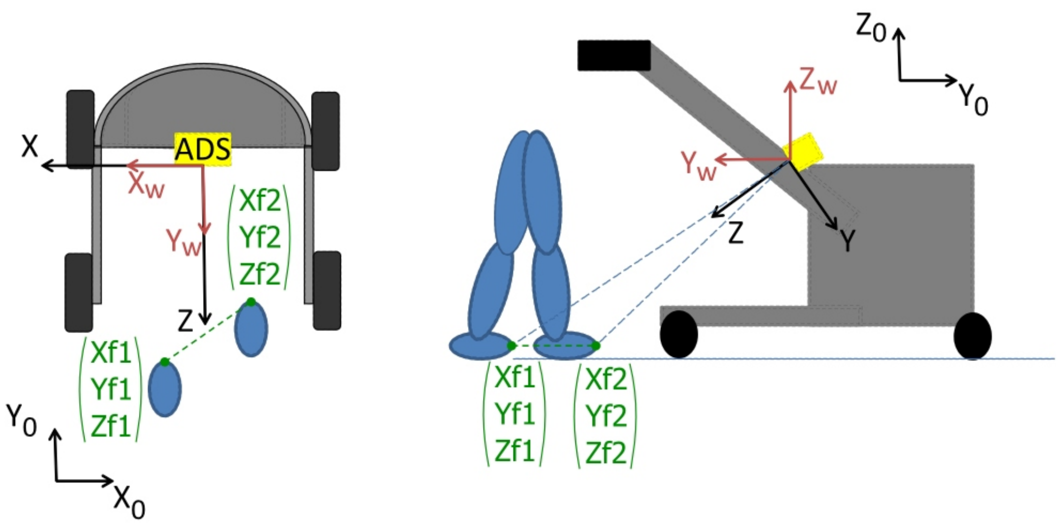


Figure 4.4: Schematic of the ADS (X,Y,Z), walker $(X_w,Y_w;Z_w)$ and world $(X_0,Y_0;Z_0)$ frames. $(Xfi,Yfi;Zfi)$, where $i=[1,2]$, are the 3D positions of the center of each foot in the walker's frame.

ADS. It assumes that the intrinsic parameters of the RGB camera are close enough to those of the camera depth sensor. With the intrinsic parameters matrix, points in the ADS frame (X,Y,Z) are obtained by multiplying the vector $(u,v,1)$ by inverted F and by the depth (s), given by the depth map of ADS, associated with the (u,v) coordinates:

$$\begin{bmatrix} X \\ Y \\ Z \end{bmatrix} = \mathbf{F}^{-1}.\begin{bmatrix} u \\ v \\ 1 \end{bmatrix}.s(u,v) \tag{4.2}$$

(ii) Conversion of ADS's frame to walker's frame

The geometric relation between the ADS frame and the walker frame is assumed to be a simple rotation along X. To find the angle of this rotation two methods can be used: to measure the angle manually (poor precision) or use the Point Cloud Library (PCL) to segment the ground plane. With this, a rotation matrix R is obtained to calculate the ADS coordinate points in meters in the walker frame.

$$\begin{bmatrix} X_w \\ Y_w \\ Z_w \end{bmatrix} = \mathbf{R}^{-1}.\begin{bmatrix} X \\ Y \\ Z \end{bmatrix} \tag{4.3}$$

4.1.3.2 Feet Segmentation

The image issued from the ADS contains feet but also legs (with perturbation of the clothes) of the user, background, floor and wheels of the walker. The aim is to extract the feet so the following process is applied to the image.

(i) Select feet region

The ADS depth image is selected to display only the "feet region". The "feet region" is defined in Z (height) direction. It is the set of data between a few millimeters above the ground to a threshold height (*thresh*) that is obtained experimentally (trial and error).

(ii) Feet segmentation

A binary image, based on the depth map, is computed to select points in the feet region according to the distance of the measured points from the ground. The binary "blob" technique (based on the identification of different regions [145]) labels the different objects in this region.

(iii) Feet selection

The centroid of each foot candidate is processed and the two closest centroids from the center of the image are labeled as right and left foot. This selection methods eliminates candidates close to the walker wheels and other objects in the background (due mainly to the imperfect estimation of the ground plane).

Elderly's gait usually presents decreased stride and step length and an increase in the walking base [140]. However, walking base is reduced when ambulating with the walker [69]. This pattern causes the feet to touch during the walk and so to appear together on the images captured by ADS. When such situation is detected through area information, the "blob" is eroded until two "blobs" appear. If this method does not allow to identify the two feet (the image does not contain two blobs), the image is not taken into account.

4.1.3.3 Computation of feet distances and orientations

The feet data will be used for gait analysis during walker use. Positions in Y_w direction (Figure 4.4) contain most of the necessary information to calculate spatiotemporal parameters (step lenght, step time, and others). In X_w direction it is possible to calculate step width which can be used to know if users have a tendency to increase their support polygon and thus give clues to predict falls [146]. Feet orientation is an important parameter for gait pattern correction.

(i) Monitoring feet positions

Two points are candidates to monitor the positions of the feet: the centroids of the feet and the toe tips. However, none of these two solutions are perfect since their position varies a lot due to changes (noise) on the blobs limits. Thus, it was selected the feet positions (Xf, Yf) as coordinates in the walker frame (see figure 4.4) as the mean of the abscissa of the foot and the ordinate of the closest point of the foot to the ADS. Despite the presented disadvantages, it also takes the centroid of each foot into account for further purposes.

(ii) Calculate feet orientations

Accordingly with [124, 125], the orientation (bearing angle) can be calculated through a line connecting to points of interest in the foot, e.g. centroind and tip toe points. However, in this approach such points presented an instable behaviour, as it was mentioned before. Thus, to calculate the orientation of the feet it was chosen to use Principal Component Analysis (PCA) [147]. This approach is used to find the highest variance of the points that correspond to each foot, thus obtaining the information on the orientation of each foot, in each frame. PCA is used for many different applications where correlation between variables is required and creates a new space where this correlation is defined.

Consider p variables and n samples. The first step of this method is to calculate the covariance matrix (size pxp) that takes the n samples into account. Then, the eigenvectors of this matrix are calculated and give the principal directions of the correlation (p is the maximal number of directions that can be found) and the eigenvalues will give the importance of the resulting correlations. In this specific application, PCA will be used to find the axis of inertia of each foot. It is applied with two parameters ($p = 2$) that correspond to the feet coordinate points (x_i, y_i), the n samples correspond to the number of detected points (i) of each foot and $\bar{x}$ and $\bar{y}$ correspond to their means. The covariance matrix ($\mathbf{S}$) for each foot (j) is defined as follows:

$$\mathbf{S_j} = \begin{bmatrix} \sigma_x^2 & \sigma_{xy} \\ \sigma_{xy} & \sigma_y^2 \end{bmatrix} = \begin{bmatrix} \frac{1}{n}\sum_{i=1}^{n}(x_i-\bar{x})^2 & \frac{1}{n}\sum_{i=1}^{n}(x_i-\bar{x})(y_i-\bar{y}) \\ \frac{1}{n}\sum_{i=1}^{n}(x_i-\bar{x})(y_i-\bar{y}) & \frac{1}{n}\sum_{i=1}^{n}(y_i-\bar{y})^2 \end{bmatrix} \tag{4.4}$$

The covariance matrix $\mathbf{S}$, may be reduced to a diagonal matrix $\mathbf{L}$ by premultiplying and postmultiplying it by a particular orthonormal matrix $\mathbf{U}$ such that

$$\mathbf{U}^T\mathbf{SU} = \mathbf{L}. \tag{4.5}$$

The diagonal elements of $\mathbf{L}$, λ_1, λ_2, ..., λ_p are eigenvalues of $\mathbf{S}$. The columns of $\mathbf{U}$, u_1,u_2,..., $u_p$are the eigenvectors of $\mathbf{S}$. The characteristic roots may be obtained from the solution of the following equation, called the characteristic equation:

$$|\mathbf{S} - \lambda \mathbf{I}| = 0 \tag{4.6}$$

where $\mathbf{I}$ is the identity matrix. This equation produces a pth degree polynomial in λ from which the values λ_1, λ_2, ..., λ_p are obtained. For this work, there are $p = 2$ variables and hence,

$$|\mathbf{S} - \lambda \mathbf{I}| = \begin{bmatrix} \sigma_x^2 - \lambda & \sigma_{xy} \\ \sigma_{xy} & \sigma_y^2 - \lambda \end{bmatrix} \tag{4.7}$$

$$0 = \sigma_x^2 \sigma_y^2 - \sigma_{xy}^2 - \lambda(\sigma_x^2 + \sigma_y^2) + \lambda^2.$$

Then, after solving the above equation, the eigenvalues (λ_1, λ_2) associated with the axis that characterizes the line the connects the center of the foot with the tip toe, are used to give the orientation (γ_{PCA}) of each foot:

$$\gamma_{PCA} = tan^{-1}\left(\frac{\lambda_2}{\lambda_1}\right). \tag{4.8}$$

(iii) Double support instants (DSI) detection

In human walking, the feet orientations vary according to the gait phase (Swing and Stance) in the gait cycle. In this study, it will defined that on the swing phase the orientation of the feet is not significant and on the stance phase is significant. By this, it was defined a specific "event" to calculate this orientation. In addition, different subjects have different feet orientations, and even the same subject can present different orientations for the same trajectory. So, the determination of the orientation of the front foot when it is on the ground (stance phase) it is not sufficient. Therefore, the orientation will only be calculated when both feet are on the ground - double support instant (DSI). This event is adequate since both feet are stable and with fixed orientations. To detect DSI, the distance, in Y_w direction, between the feet is calculated (Figure 4.5). The modulus of the distance is maximum before the derivative of the depth signal (Y_w direction) turned negative or positive. If this condition is verified, a DSI is detected. (squares in figure 4.5) and the orientation is calculated.

The height signal (Z_w direction) would be good to estimate if a feet was on the ground or not. However, in this study such signal was not used for double support instant detection since even if heigth is well estimated, the DSI are not as clear on the Z-direction signal as on the Y-direction signal, in terms of signal processing. Y-direction signal is more predictable and simpler to process than Z-direction signal.

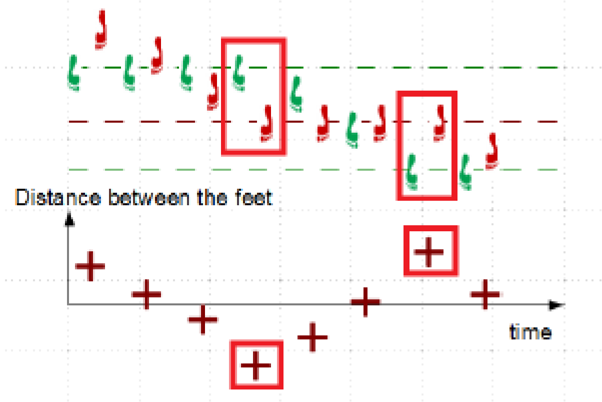


Figure 4.5: Detection of double support instants.

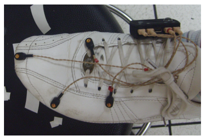


Figure 4.6: Codamotion markers on the foot.

4.1.4 Results

4.1.4.1 Experimental setup

One trajectory, consisting in a U-turn, was performed by three healthy subjects and eleven elderlies with various diagnoses leading to gait dysfunction (Table 4.1). These patients were colleted from Braga Hospital in Braga, Portugal and from Charles Foix Hospital in Paris, France. Each subject performed the trajectory two times. Different walkers were used (Table 4.1), in order to demonstrate that the system is flexible regarding the walker that is integrated. For validation of the obtained data with the proposed ADS system, ground truth trajectories are provided by the motion capture system Codamotion System (www.codamotion.com). The technology uses miniature infra-red active markers positioned on each foot (Figure 4.6) to track the key positions on any subject. The spatial error of this sensor is less than 0.33 mm. The sampling period was set to 100Hz and data processing was made on MATLAB (version 2012b). Positions and orientations of the feet are then compared between data obtained from ADS and from a ground truth (Codamotion).

Regarding the ADS system, the sensor was placed 0.35 m from the ground and around 40° of inclination in the walkers of figures 4.2a, 4.2cand 4.2d. In the smart walker of figure 4.2b, the sensor was placed 0.8 m from the ground with 20° of inclination.

Table 4.1: Subjects' Demographic Data (ADS) and walker type used.

	Age	Weight (kg)	Diagnosis	Walker device
1	26	63	Healthy.	Figure 4.2a
2	28	79	Healthy.	Figure 4.2a
3	31	76	Healthy.	Figure 4.2a
4	86	69	Recurrent falls and compression of vertebrae T12.	Figure 4.2d
5	86	62	Recurrent falls and oedema.	Figure 4.2d
6	88	65	Fall with head trauma.	Figure 4.2d
7	90	75.3	Confounding syndrome that led to a loss of autonomy.	Figure 4.2d
8	84	75	No autonomy and has cognitive limitations.	Figure 4.2d
9	71	82	Post-surgical Knee Osteoarthritis.	Figure 4.2b
10	89	39	Psychomotor agitation.	Figure 4.2c
11	91	51.5	Brain breed.	Figure 4.2c
12	91	68	Lower limbs ulcer and acute decompensated heart failure.	Figure 4.2c
13	85	40	Heart failure and femoral head fracture.	Figure 4.2c
14	79	93	Pulmonary abscess.	Figure 4.2d

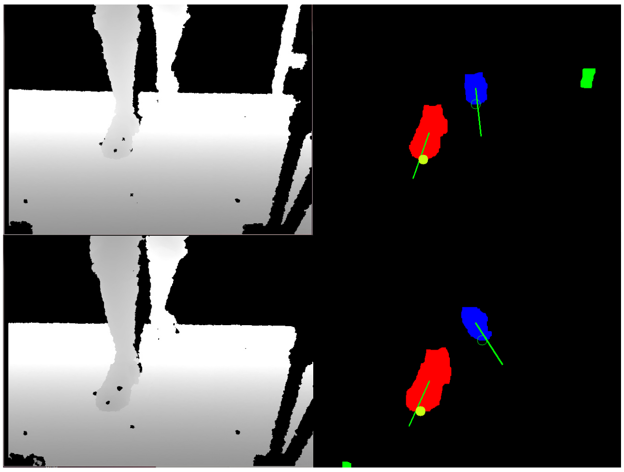
4.1.4.2 Feet detection and segmentation results

Figure 4.7 shows two frames of different phases of the performed trajectories with one healthy subject (Figure 4.7a) and one elderly subject (Figure 4.7b). The first frame shows the moment that the right foot is beginning to cross with the left foot in a straight line. The second frame shows the feet performing a curve for the right. The first image of each frame corresponds to the original input image captured by the ADS. After applying the feet tracking algorithm, the second image of each frame is obtained. Some unknown objects can appear in the image, while the subject is walking, leading to a feet false detection. However, the algorithm was capable of discarding such objects. Each foot is labeled as right foot (red) or left foot (blue). With the detection of the feet, the point of interest of each foot can be calculated. Then, PCA is applied to calculate the orientation of each foot. The image shows a representation of the axis of inertia (line of each foot) that allows the calculation of the angle of orientation of the feet. It can be seen that different foot orientations are well identified by the PCA algorithm.

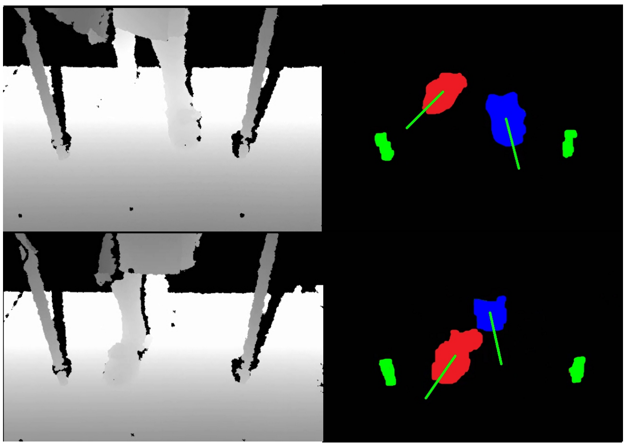
Regarding *thresh* (expressed in mm), it was defined $10 < thresh < 80$ and in terms of processing time, it takes only 0.17 s to qacquire and process each frame and obtain these results.

4.1.4.3 Position errors results

Since the goal is to extract gait parameters from the collected images, it is represented the signals obtained with the position coordinates in (X_w, Y_w) frame when the subject is walking forward. To validate these positions, coordinates of the feet from Codamotion system in the



(a)



(b)

Figure 4.7: Two Frames representing the feet tracking algorithm result. First frame represents the subject walking forward. Then, second frame represents the same patients turn right. Right foot is red and left foot is blue. The line represents the result of PCA, giving the orientation of each foot. a) healthy subject; b) elderly subject.

world frame (X_0, Y_0) were transformed into coordinates in (X_w, Y_w) frame to be compared with the ADS signals

As it can be seen in figure 4.8 the positions (X_w, Y_w) estimated with the ADS are very similar with the ones captured by Codamotion system. As it was previously said the walker may bring motion occultation, which means the Codamotion system will miss data that will increase the difficulty to study the gait. Such situation is observed in figure 4.8 where Codamotion system signals present missing data and errors due to occlusions.

Another situation where occlusion of data appears is during U-turn (Figure 4.9). The Codamotion system loses the markers, and it is not possible to acquire data in such event. However, our proposed system it is capable of tracking the feet even in such situation. It is possible to verify that the ADS system detects both feet through the all trajectory and CODA loses the feet at $t \approx 10s$, when the turn begins.

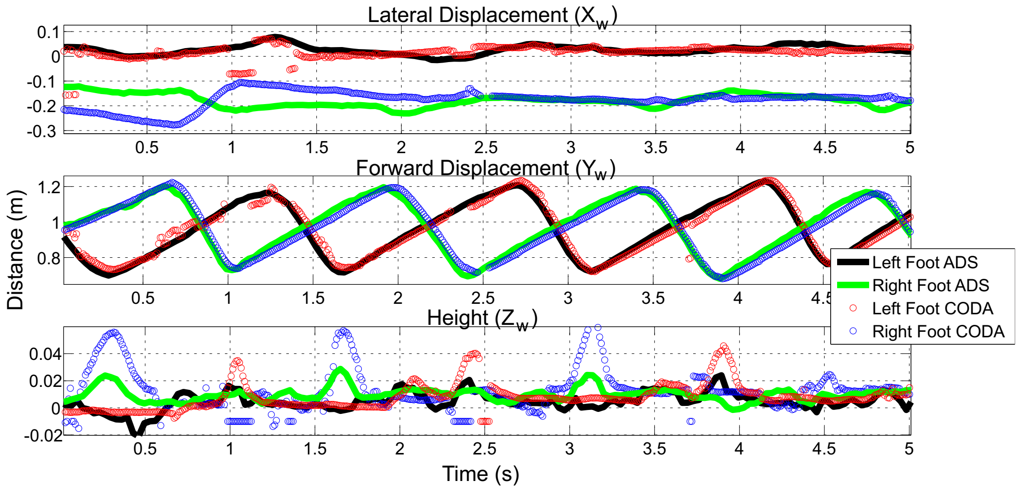
The results allow concluding that the proposed system is better suited for gait motion analysis in assisted gait with a walker device than Codamotion system.

As in [126], Root Mean Square Deviation (RMSD) is used to quantitatively compare the Codamotion and ADS position data. The results are shown in table 4.2. On position precision, the Z-signal is the more precise and precision on X and Y-signals are about the same. In figure 4.8, in Z_w, main differences between ADS data and ground truth (CODA) occur when the feet are far away from the ground. Indeed in this case, a smaller part of the foot is in the feet region leading to less precise measurement. In Y_w, the pattern is detected with ADS data as well as with the Codamotion data. In X_w, even if the difference seems bigger due to the scale, position error is about the same as for Y_w.

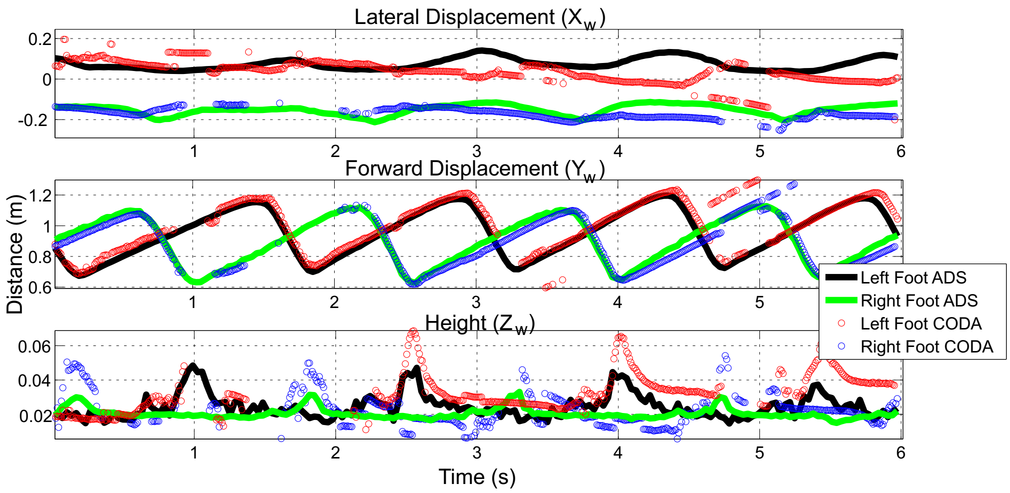
By comparing the results with [126] (method with markers) for all positions, in table 4.2, it can be seen the proposed algorithm presents worse results. Even if the precision of the proposed method is not as good as in [126], it seems acceptable for gait analysis.

In table 4.2, it can also be seen results with elderlies. Despite the error increase, it is still acceptable that the proposed algorithm is suitable to be used for feet tracking, acquiring the necessary data to be used as a gait assessment system. It is now necessary to perform gait analysis (spatiotemporal evaluation) to assess whether these errors are admissible for such application. Such verification is done in section 4.3.

In figure 4.10, it can be seen that double support instants are difficult to extract from z-signal as no threshold in altitude between “on the ground” and “out of the ground” can be easily defined. Double support instants are well detected by the proposed method (green picks in figure 4.10) with a precision around 0.1s. This signal will be used for controlling a smart walker in future work.



(a)



(b)

Figure 4.8: 3D feet positions acquired by CODA motion capture system and ADS sensor with two patients when walking in straigth line.

Table 4.2: Comparison between [126] and the proposed method on positions errors

	RMSD (mm)		
	X	Y	Z
Healthy subjects	28.9±2.03	30.8±2.82	13.4±3.37
Elderly	35.2±2.87	40.1±7.03	13.9± 4.56
[126]	4.9±1.40	19.4±6.10	8.4±1.70

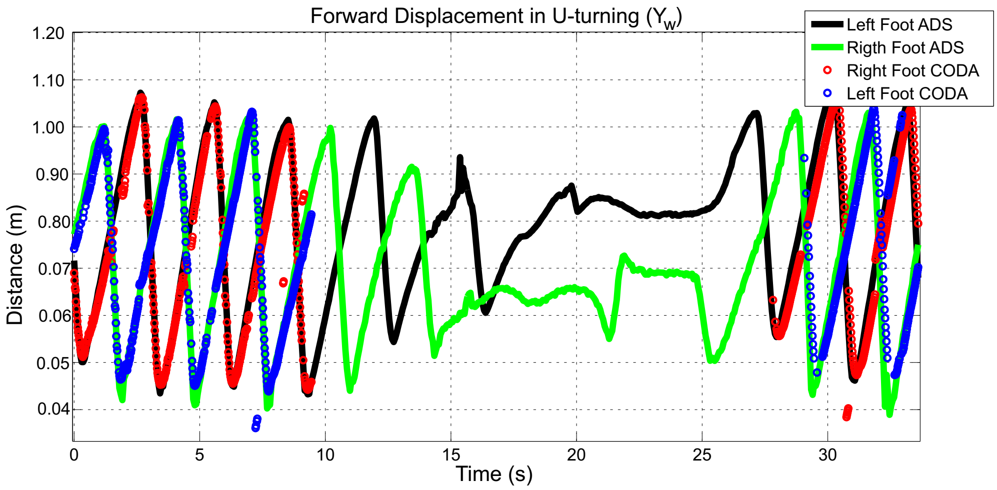


Figure 4.9: Feet position acquired by ADS sensor and CODA of one patient during a U-turn.

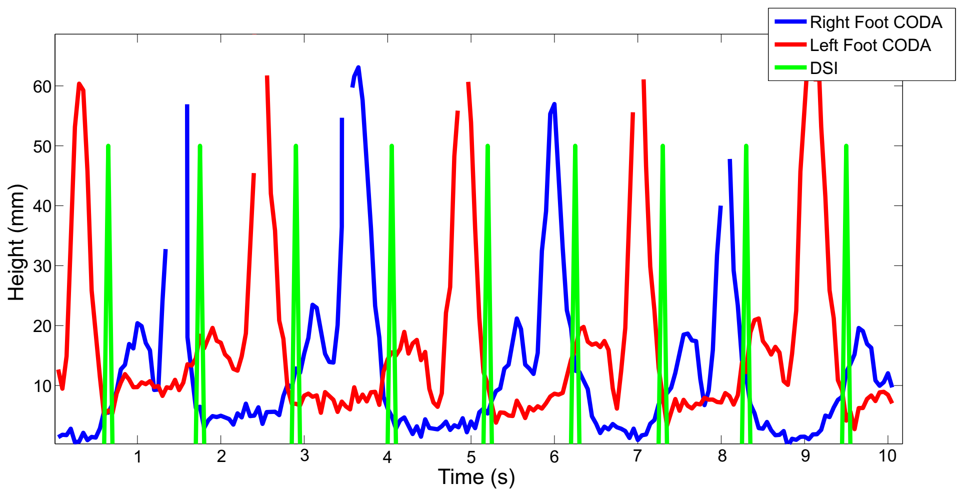


Figure 4.10: Z_w position tracking with Codamotion system with DSI detection by ADS sensor.

Table 4.3: Comparison between [125] and the proposed method on orientation errors.

	RMSD (%)
Healthy subjects	21.1±4.33
Elderly	25.2±5.30
[125]	7.9±2.6

4.1.4.4 Orientation errors results

Figure 4.11a and 4.11b represent angle measurements during a left and a right turn, respectively. Forward and turn phases (long pink dashed lines) are identified as well as DSI instants (little black dashed lines). Joly et al. [125] presented discontinuous angle measurements, but as it is shown in figure 4.11, the proposed method gives continuous measurements. ADS data and data from the Codamotion system display the same behavior. The error in orientation of this purpose is about 20% (see table 4.3) that should be sufficient to assess big changes. Indeed, it represents $\pm 7^o$ and it is possible to see that during straight phases, angle could change $\pm 10^o$. During turn phases, angle variations increase. Looking at Paolini et al. [126] results, 7.9% o error is very low. But once again, they use markers on their approach, which is not suitable for this application.

From figure 4.11, it is observed that, in turn phase, both signals present opposite slopes, which are correlated to the turn direction. This is a feature that can be used to discriminate between a left turn motion and a right turn motion.

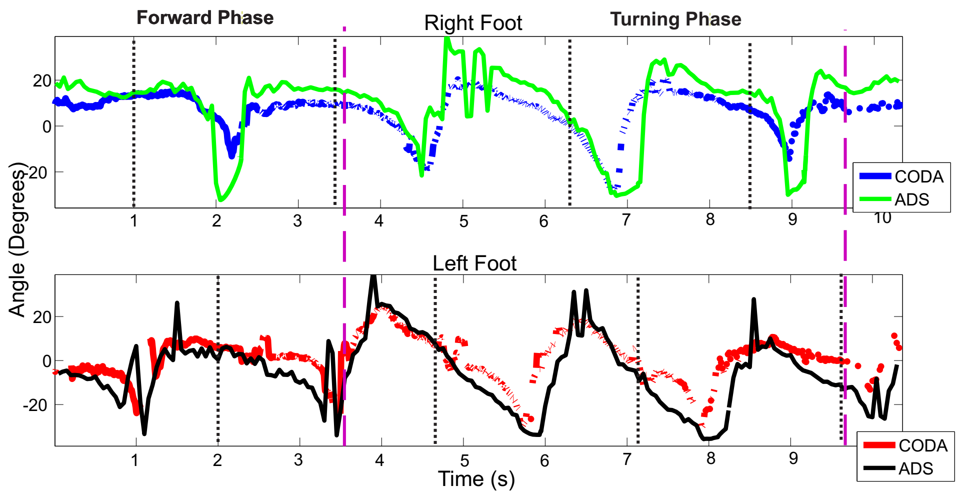
Looking into detail at the DSI instants, it can be verified that at these instants the angle obtained through ADS is very similar with the one obtained with CODA. The error orientation decreases to 10%, which means that this instant is ideal to turn the feet orientation acquisition more accurate.

Looking at a complete U-turn, in figure 4.12, for example, it can be observed that the algorithm detects the orientation of the feet.

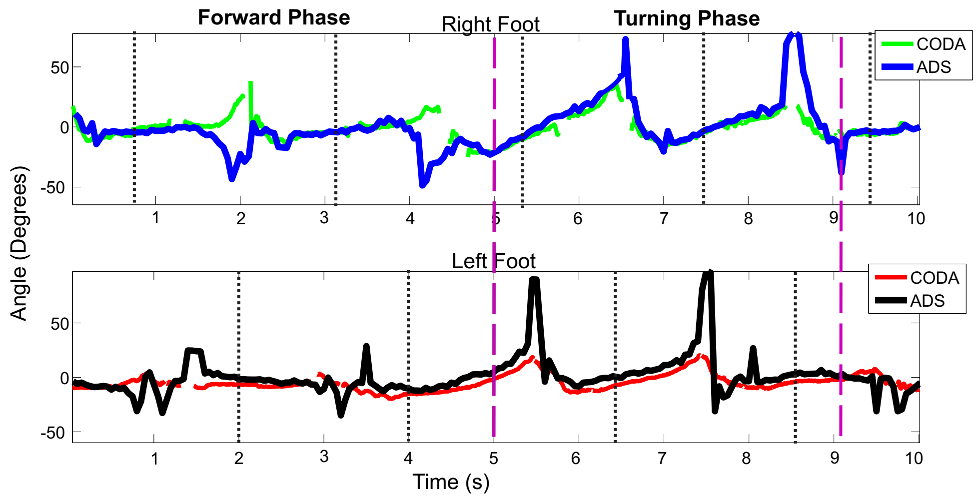
These results show that the proposed algorithm can be used to monitor the feet orientation, and perhaps be used to a walker movement control purpose with the feet detection.

4.1.5 Conclusions

This section presents a system able to track the feet position and orientation during an assisted walk without equipping the user. An active depth sensor was used with a new detection algorithm that suits for all subjects and walkers like rollator. The main advantages of our method compared to others realized with this kind of sensor is that it is markerless, faster than using 3D models, reliable against clothes conditions and detects continuously orientations of the



(a) Left Turn



(b) Right Turn

Figure 4.11: Orientation signal given by ADS and Codamotion systems in forward path followed by a (a) left and (b) right turn. Dashed lines represent the time interval when the turn occurs.

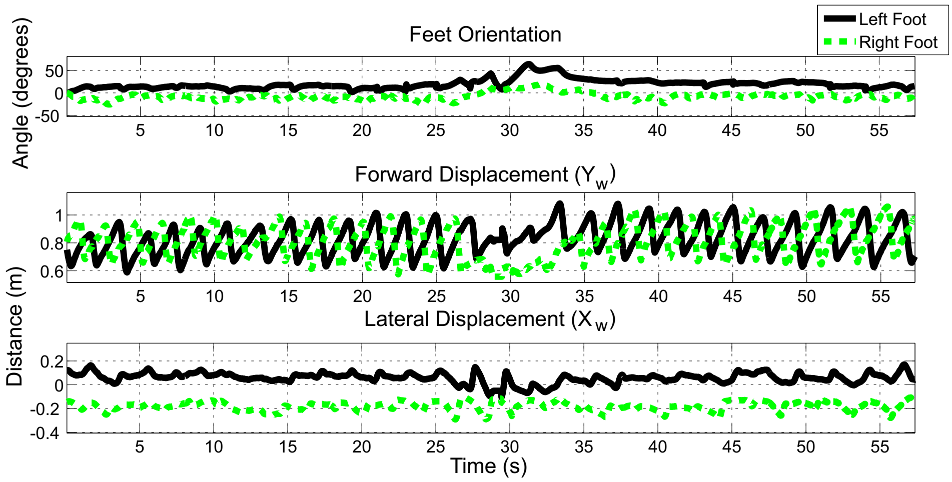


Figure 4.12: 2D Feet position and orientation acquired by ADS sensor of one patient describing a U-turn.

feet. The precision of the presented method is better than other markerless methods [125] and seems sufficient for gait analysis. However, validation of such precision is still necessary to be done.

Preliminary results show that this system has high potential to be used on clinical trials with patients to give clinical insight to the clinicians. Other application of this system may be the control of the smart walker movement. Since this system gives the orientation and position information of the feet, it can be used to translate patients' intention of velocity and direction. Further studies are necessary on this matter.

4.2 Legs Position Tracking with Laser Range Finder Sensor

4.2.1 LRF System

The LRF (URG-04LX URG01), in figure 4.13, performs a scan of 240° with an angular resolution of 0.36°. The time spent in each scan is 100 ms, time that meets the requirements for the measurement of parameters associated with the pathological human gait [140, 144]. In a full scan, the sensor acquires 682 points (approximately one point per 0.36°) from left to right.

This system aims to acquire the distance between the legs and the walker. It can be deduced, mistakenly, that the most appropriate position for fixing the sensor would be a few centimeters from the ground so that the feet of the user can be intercepted by the scanning plane of the laser. However, during the gait process, the user's feet rise above this plane, so that, in these moments, no information regarding the lifted foot is detected.

Pallejà et al. [122] showed in their work that it was not possible to detect accurate information of feet movement with LRF sensor. Thus, to prevent undesired detection of the feet, the sensor is positioned to scan a plane, which is distant 30 cm from the ground and parallel to it (Figure 4.13). This plan was chosen according to [121, 123].

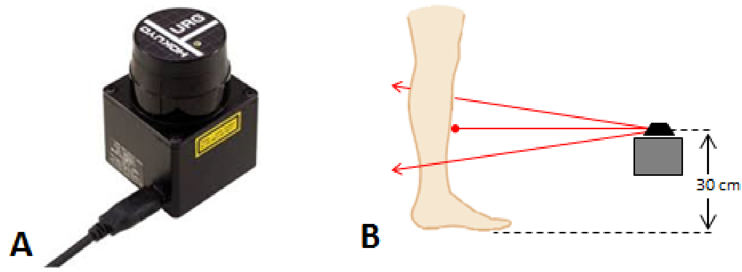


Figure 4.13: A. Laser Range Finder sensor (URG-04LX URG01); B. Location of the LRF sensor.

The proposed system in this section was designed in collaboration with the colleagues of the Electric Engineering Department, from Federal University of Espirito Santo, Vitória, Brazil. As a result of this collaboration, two scientific contributions were published [19, 22].

4.2.2 Brief Review of Legs' Tracking Methods with LRF

For the development of an algorithm to track the legs, a state-of-art research was made. Many studies already presented methods of legs tracking with a LRF sensor.

Kheyrur et al. [148] used the geometric approach with the Bounding box method. This is a method for checking geometric features of a set of candidate data that is to be classified as "human legs". The classification is based on the length of the diagonal of an imaginary rectangle, among other features, which has two opposite vertices that correspond to legs points. However, according to [148], it is not able to capture critical information for efficient detection of legs.

Chalvatzak et al. [149] proposed the detection and tracking of user's limbs using the range data for the feature extraction. Towards this end, the authors apply a combination of K-means clustering along with Kalman Filtering (KF). However, postures that may lead to false detections, such as closed legs, are not addressed in this study.

Another method is known in the literature as Circle fitting [150]. This method assumes that the data from laser scanning concerning to human legs appears with a curved shape. Although other objects during scanning may also have curved forms, it is considered that the radius of curvature of human legs is normally between two specified limits. This builds up the classification method for verifying the radius of curvature of the detected shape. The disadvantage of this approach lies in the fact that the type of clothing can change the geometry.

Belloto and Hu [151] used a LRF to identify patterns of legs which can be separated legs, legs together or not parallel legs, in order to allow interaction between a person and a mobile robot. Despite dealing with different legs postures, these patterns are pre-defined with the help of features whose values are found off-line. Since each person has its own characteristics, the pre-definition may lead to errors in the detection of patterns. The approach presented in [129] does not classify the legs posture into pre-defined patterns. They divide the space into two sub-regions (right and left) and classify the legs as right or left by observing the sub-region in which they operate. This division of regions is made by an imaginary line passing through the centre of the LRF scanning. This approach could work well if implemented on a walker. However, it would only be successful in straight-line paths when the legs are enough apart. If the user walks with the legs very close to each other or if one leg invades the sub-region of the other, the algorithm may fail.

In this section it will be presented the development of a technique for detection of legs similarly to [151] since it deals with the problem of different legs postures. The difference relies on the self-calibration of the system to be able to detect legs of different subjects and the features that characterize the legs. Besides, the system is able to deal with noise and situations of non-pattern. In addition, it does not depend on pre-defined sub-regions for each leg.

4.2.3 Proposed Algorithm for Legs' Tracking

The legs' detection method presented in this section is based on [151] and develops improvements since [151] does not deal with noise, non-patterns and different user's legs dimensions and clothes. The developed detection algorithm is divided into five parts represented in figure 4.14.

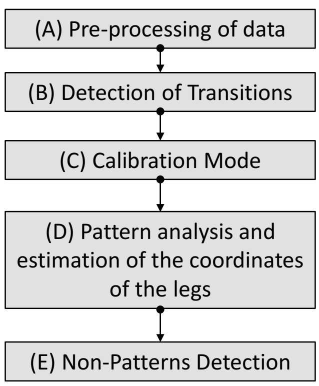


Figure 4.14: Proposed leg's detection algorithm.

4.2.3.1 Pre-processing of Data

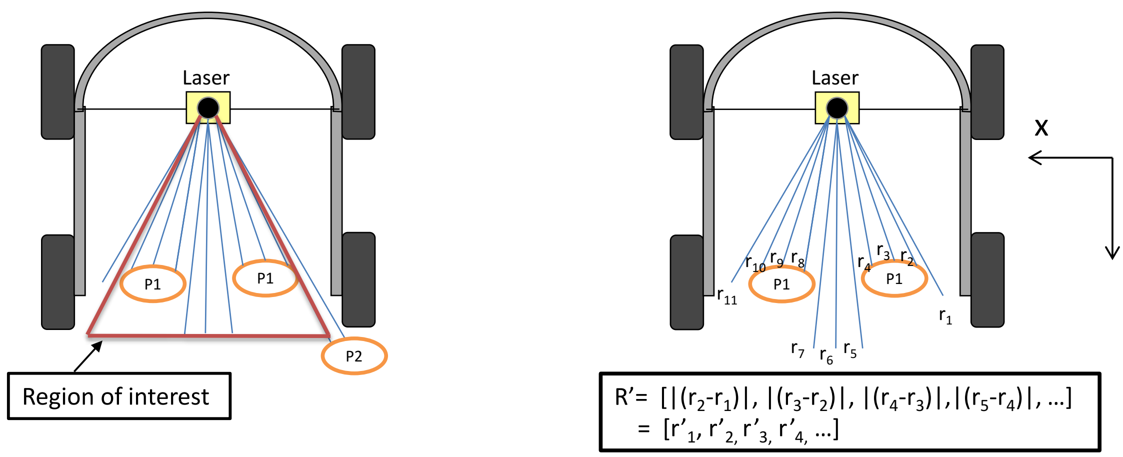
Each point distance is represented by m_i, where i is the index of the point of acquisition. Thus, i varies from 1 to the maximum number of points of the scanning (682 points for a full scan). Each measurement point i, in each scan, is represented as follows:

$$m_i = (\alpha_i, r_i), i = [1, ..., 682] \tag{4.9}$$

where α_i is the angle calculated from the i index and $r_i$corresponds to the measured distance (mm). Thus, the point set that is acquired in a full scan can be represented by:

$$U = \{m_1, m_2, ..., m_{682}\} \tag{4.10}$$

In order to limit the background, a boundary of the region of interest (ROI) is performed. This ROI seeks to address the whole area where the legs will be positioned during walking. All measurements that are outside the ROI will not be considered ($256 < i < 426$ ($30^o < i < 30^o$) , $r_{max} = 1000\ mm$). This procedure aims to make the LRF to only identify the person who is using the walker, thus preventing people and objects that are near the walker to interfere with the detection of the user's legs. Figure 4.15a shows the top view of the walker in a situation where a person P1 is with the legs inside the defined region and a person P2 is outside that region.



(a) Delimitation of the region of interest.

(b) Representation of the transitions' detection procedure.

Figure 4.15: Top view of the walker.

4.2.3.2 Detection of Transitions

In this section the calculation of transitions in each scan of the LRF signal will be presented.

These transitions are defined as the difference between two consecutive i points of scan j. After the delimitation of the ROI, vector $R_j = [r_{256j}, r_{257j}, ..., r_{426j}]$ is created and contains the distances measured in scan j. For the transitions' detection $R'_j = [r'_{256j}, r'_{257j}, ..., r'_{425j}]$ is created, which contains the transitions. Each element is calculated as follows:

$$r'_{ij} = \left| r_{(i+1)j} - r_{ij} \right|, 256 < i < 425 \tag{4.11}$$

Vector R'_j is then used to infer which transitions correspond to the bounds of a leg. For this, each value of the vector R'_j is compared to a threshold λ (this constant is calculated online as it will be explained in the next section). If a transition value r'_{ij} is higher than λ, it corresponds to a bound of a leg. Figure 4.15b shows an example of this detection, where r'_1 and r'_4 of vector R' correspond to leg bounds.

4.2.3.3 Calibration Mode

For the correct detection of the legs, some features must be assessed to determine if there is one leg, two legs or not in vector R' of the ROI. This is herein called the problem of different legs postures. This evaluation aims to distinguish legs from other objects that could be in the ROI between the user and the walker. In [151], the leg's width (a), the maximum step length (b) and the width of two legs together (c) are the features selected to detect legs' boundaries. However, the authors proposed to use the following features: opening angle of the leg (lp) to check if it is one or two legs; and maximum step length (λ) to detect transitions that correspond to legs' boundaries (LB). These features are illustrated in figure 4.16.

In [151] the features are pre-defined off-line. This work proposes an online calibration (OC), during which the individual only needs to take two steps with the walker, at his own pace and without time limit, to estimate lp and λ. λ is the difference between r'_{ij}. lp is calculated as the difference between i of two consecutive transitions that correspond to the extremies of a leg.

Please note that OC should be performed with clothing that allows distinguishing the two legs and both legs must be spaced from one another during OC. It is also noteworthy that as more acquisitions are obtained during OC, the better will be the outcome of the OC, since the values of the evaluated features are based on average values of all scans. These values are used for the same person. If the person and/or conditions change, a new OC has to be done.

Figure 4.16: Features for calibration.

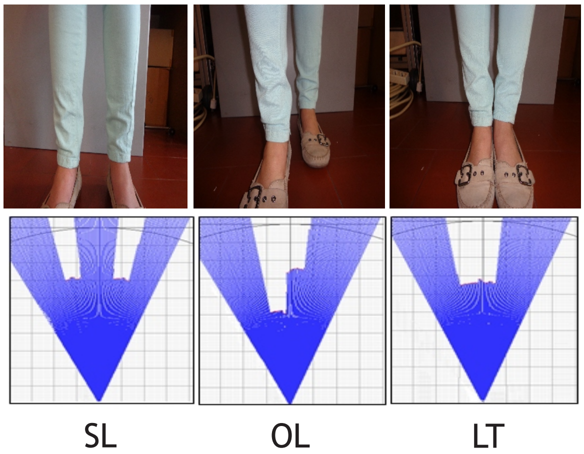


Figure 4.17: Legs' Patterns represented by the LRF: Separated legs (SL), legs together (LT) and overlapping legs (OL).

4.2.3.4 Pattern Analysis and Estimation of the Coordinates of the Legs

During assisted gait, the user can present different legs' patterns. This makes the laser to capture different data patterns, and thus the calculated center of each leg will be different.

The detection of patterns is based on the classification of the position of the legs according to the number of detected transitions. Three different patterns can be identified: separated legs (SL), legs together (LT) and overlapping legs (OL). Figure 4.17 illustrates these three patterns and the corresponding acquired raw LRF data.

To classify the patterns a flow chart is presented in figure 4.18. First the number of candidate of leg boundaries (LB') that might correspond to a leg is calculated through the λ value (calculated on OC). Then, if the number ($b = 1, 2 \ldots B$) of LB' corresponds to one of the values of table 4.4, the pattern is classified and the center of each leg is transformed onto polar coordinates (r, α). Later, for spatiotemporal parameters calculation, these coordinates are converted to cartesian coordinates, since those parameters are calculated accordingly to x and y directions.

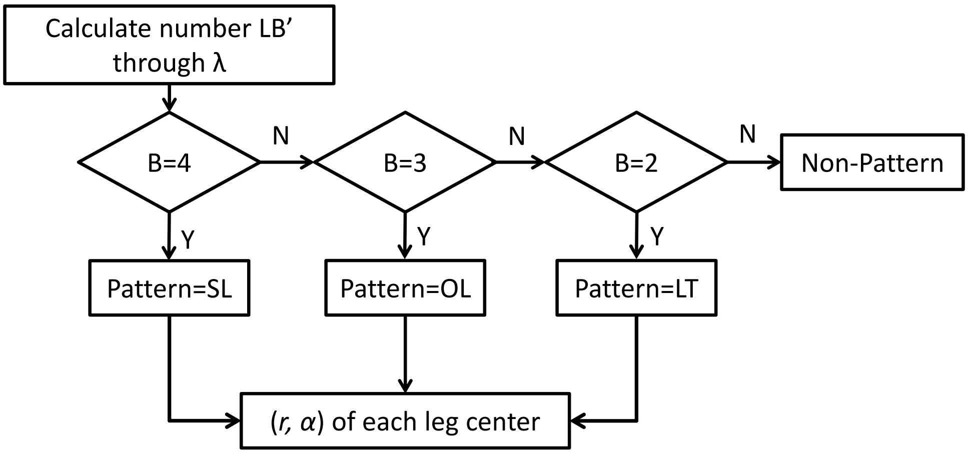


Figure 4.18: Flow chart of the classification of leg patterns.

	Pattern		
	SL	OL	LT
Number of LB' (B)	4	3	2

Table 4.4: Number of transitions for each leg pattern.

4.2.3.5 Non-Patterns Detection

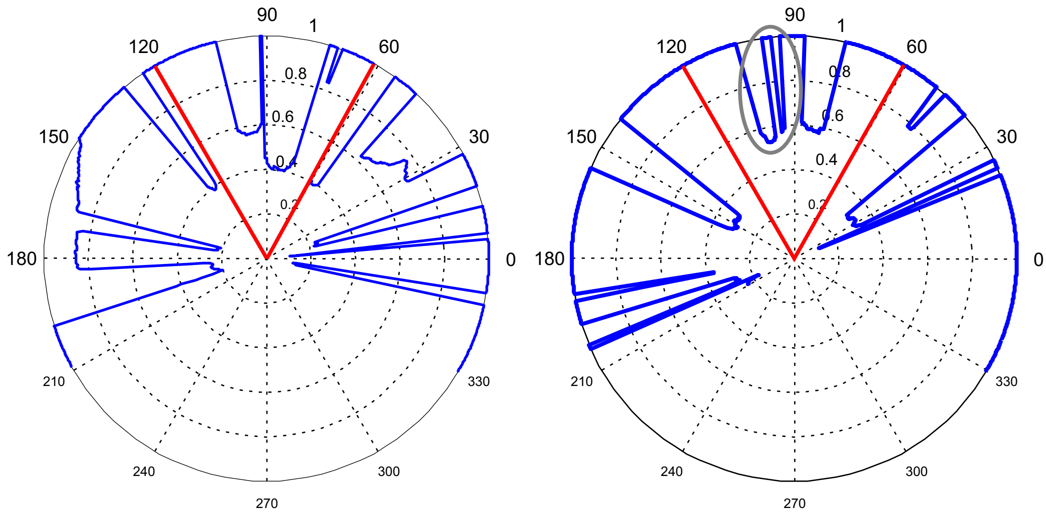
In case more than 4 transitions are acquired a non-pattern is detected. The occurrence of this situation appears when the laser detects an object or noise in the ROI. In case 5 transitions are detected, it means that the laser detected noise or OC was not properly carried out. In case 6 or more transitions are detected, it means that an unknown object was detected on the ROI (Figure 4.19a) or a noise occurrence divided one leg in two parts (Figure 4.19b). If these situations are detected the following procedures are executed:

(i) Transition Pairs Verification

First, the pairs of transitions are verified to check which pairs correspond to a leg. Two conditions are verified and both have to be valid as presented in figure 4.20:

(1) Is the difference between i of two consecutive transitions higher then lp? If yes, it means that probably a leg was found and if not it is not a leg, like the situation illustrated in figure 4.19a; then (2) Is r_{i+1} (i corresponds to the position of a transition) lower than 1000mm (r_{max})? This condition eliminates false legs, since the space between the legs can present a distance greater than lp.

If both conditions are verified, only the pairs of transitions that correspond to legs are saved.



(a) (b)

Figure 4.19: Situations where the algorithm detects 6 transitions.

(ii) State Sequence Verification

If verification (i) is confirmed (Figure 4.20), the number of detected transitions (B_j) is compared with the number of transitions of the previous scan (B_{j-1}). This comparison is based on a state sequence verification. A state is characterized by a number of transitions. This state sequence is composed by four states: 4T (4 transitions), 3T (3 transitions), 2T (two transitions) and 0T (no transitions detected, which means no legs). All states are bidirectional and from scan to scan the same state can be verified. Observing figure 4.21, possible transitions between states are identified by arrows.

In case incorrect state sequences are detected, a flag is set with value 1 and the current legs' coordinates acquire the value of the coordinates of the past state.

An example of the importance of the state sequence verification is illustrated in figure 4.19b. In this case, 6 transitions were identified and through verification (i) one leg (right leg) would have been eliminated since this verification (i) would have considered that this acquisition only has the presence of one leg. This elimination would pass 4T to 2T, which is an incorrect state sequence. Thus, state sequence verification becames important to verify if a correct sequence of patterns is followed.

(iii) System Error

To verify if the two last verifications worked (Figure 4.20), the algorithm verifies if each r between two consecutive samples is greater than 200 mm (the distance walked between two

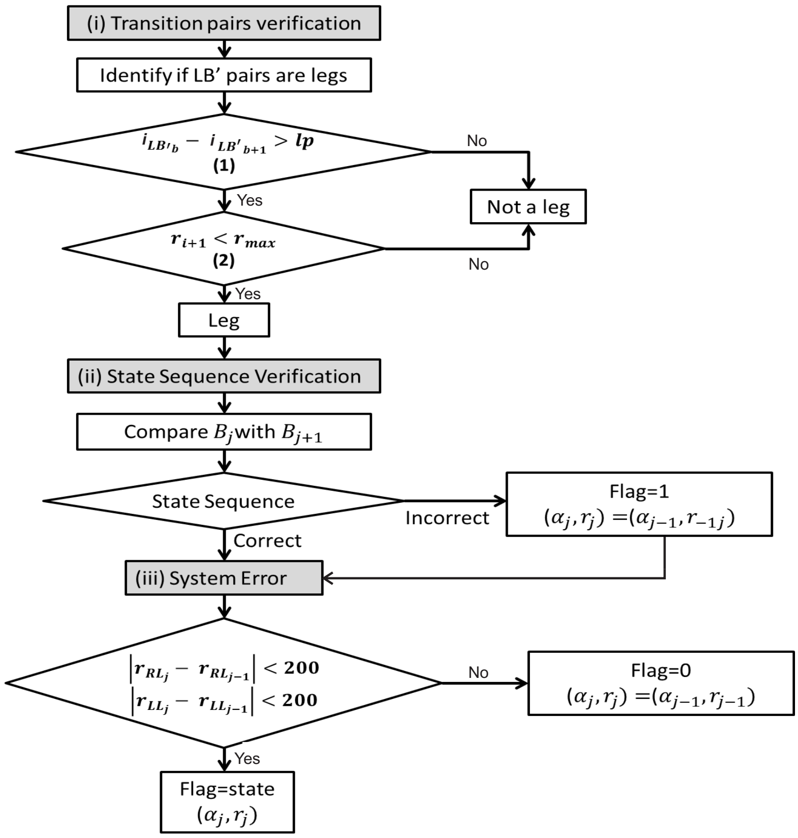


Figure 4.20: Flow chart to verify pairs of transitions and state sequence.

samples is never higher than this value). If that happens, the distance of scan j is equal to scan $j-1$ and a flag is set equal to zero (the flag is only recorded to count how many times the system detected a verification error, as itl will be shown in results.

4.2.3.6 Conversion of LRF frame to walker's frame

Looking at figure 4.22, it is necessary to know the cartesian coordinates (X_L and Y_L) of the subject's leg relative to the walker through the polar coordinates (r, α) given by LRF.

Thus,

$$X_L = rcos(\alpha) \quad (4.12)$$

$$Y_L = rsin(\alpha) \quad (4.13)$$

However, this result would not be accurate since there is an error between the distance measured by the laser and the distance that the subject actually walks [105]. As it can be

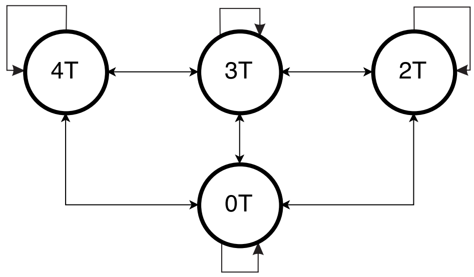


Figure 4.21: The correct state sequence. 4T (4 transitions), 3T (3 transitions), 2T (two transitions) and 0T (no transitions detected).

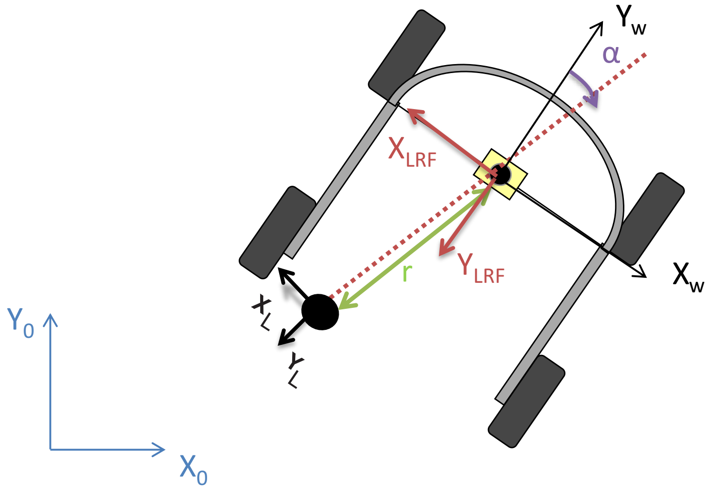


Figure 4.22: Axis scheme for the calculation of legs' coordinates on the world reference (X_0, Y_0).

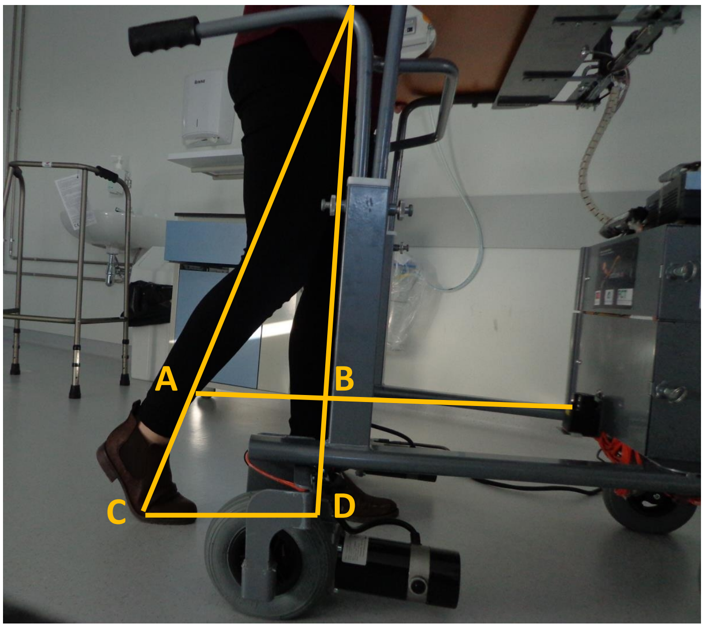


Figure 4.23: Ratio between the reading of the laser and the person walks.

seen in figure 4.23, r_{C-D} corresponds to the real step length and r_{A-B} corresponds to the step length measured by the laser that is 30 cm above the ground. Through the followed relation, it is possible to find the correction coefficient k between these distances:

$$r_{C-D} = k.r_{A-B} \tag{4.14}$$

and thus calculate r_{C-D}.

4.2.4 Results

4.2.4.1 Experimental Setup

A total of 14 patients (12 elder patients subjected to total knee replacement surgery and 2 ataxic patients, indicated for the use of walker) were asked to perform the tests that will be presented on the following subsections. Their gait is irregular and unbalanced. These subjects, listed in table 4.5, were patients from Braga Hospital and signed the informed consent. The tests with the walker were video-recorded for temporal validation and the steps were marked on the floor for spatial validation.

Table 4.5: Subject's Demographic Data (LRF)

Subject	Age	Height (m)	Calf length (m)
1	66	1.70	0.55
2	65	1.71	0.52
3	67	1.69	0.54
4	74	1.67	0.51
5	65	1.66	0.50
6	41	1.71	0.51
7	54	1.68	0.49
8	58	1.60	0.35
9	52	1.65	0.48
10	62	1.62	0.38
11	62	1.85	0.63
12	28	1.61	0.47
13	42	1.67	0.53
14	40	1.60	0.47

Table 4.6: Mean and Standard Deviation values of Legs' Features.

OC Features	Values (Mean ± Standard Deviation)
lp	$15.5^o \pm 11^o$
λ and *b* [151]	31 cm ± 20 cm
a [151]	10 cm ± 6 cm
c [151]	20 cm ± 9 cm

4.2.4.2 Online Calibration Mode and Patterns' Detection Results

In order to evaluate the effectiveness of the proposed method to detect legs' patterns and the achieved error reduction, the proposed method in [151] has been implemented for comparison with the proposed system (with and without OC). To acquire the selected features (Table 4.6) a set of tests where the patients had to walk in straight line for data collection involving the 14 subjects (Table 4.5) were conducted. After measurements with data from LRF, it was concluded that features' values are different from subject to subject. In table 4.6 it can be seen that the standard deviation is slightly high. The three approaches were tested: [151] approach was tested with *a,b,c* values of table 4.6; the proposed system without OC was tested with *lp* and λ values of table 4.6; and the proposed system with OC found *lp* and λ values for each patient.

The total number of laser scans recorded for each patient was 1800 (180 s, an average of 180 steps). In table 4.7 it is presented the results of how accurate the detection algorithms clearly classifies the transition state into SL, OL and LT. True Positive means that true SL,

Table 4.7: Comparison of leg detection errors using [151], proposed system without OC and with OC. Errors are expressed as ratios of false positives, false negatives and true positives versus total detectable patterns. True Positive are related to patterns correcty classified and false negative and positive are related to wrongly classified patterns as false and true, respectively. Mean±Standard deviation values.

	[151]	**Proposed system without OC**	**Proposed system with OC**
False Positive (%)	2±2	8±1	0.01±0.5
False Negative (%)	45±2	83±1	0.49±0.5
True Positive (%)	53±2	9±1	99.50±0.5

OL, LT data are classified correctly, and false negative and positive means the patterns are wrongly classified as false and true, respectively. Our algorithm with OC presented a percentage of 99.5% of correct detections compared with 53% and 9% of the other approaches (Table 4.7). The error of false negatives (missed patterns) is very high both for [151] and for the proposed system without calibration. This shows that OC is needed to decrease the error of pattern detection. It is noteworthy that if more steps are done by the subject during OC, more effective are both features to detect legs and it is necessary that during OC both legs are visible. In general it has been required an average of 1-2 steps, which corresponds to an average of 20 samples (*i.e.* 2s for OC since LRF has 0.1s of acquisition period). Figure 4.24 illustrates a compilation of the three patterns and two situations of non-patterns that were detected throughout the experiments with the 14 subjects using OC.

4.2.4.3 Estimation of the Coordinates of the Legs Results

In order to test if the coordinates of the detected legs are being well calculated two types of experiences were performed with the 14 patients: (i) Subjects stand in front of the walker; and (ii) Subjects push the walker while executing a straight forward trajectory. The achieved results are described next.

(i) Subjects Stand in Front of the Walker

The distance in X-axis and Y-axis between LRF and the legs was measured through a metric tape (ground truth) and compared to the distance calculated by the proposed system. Root Mean Square Deviation (RMSD) is used to compare quantitatively the distance measured by ground truth with the one measured by the proposed system. Results can be seen in table 4.8. RMSD values show that the algorithm is correctly detecting the center positions of the legs with small error in standing position.

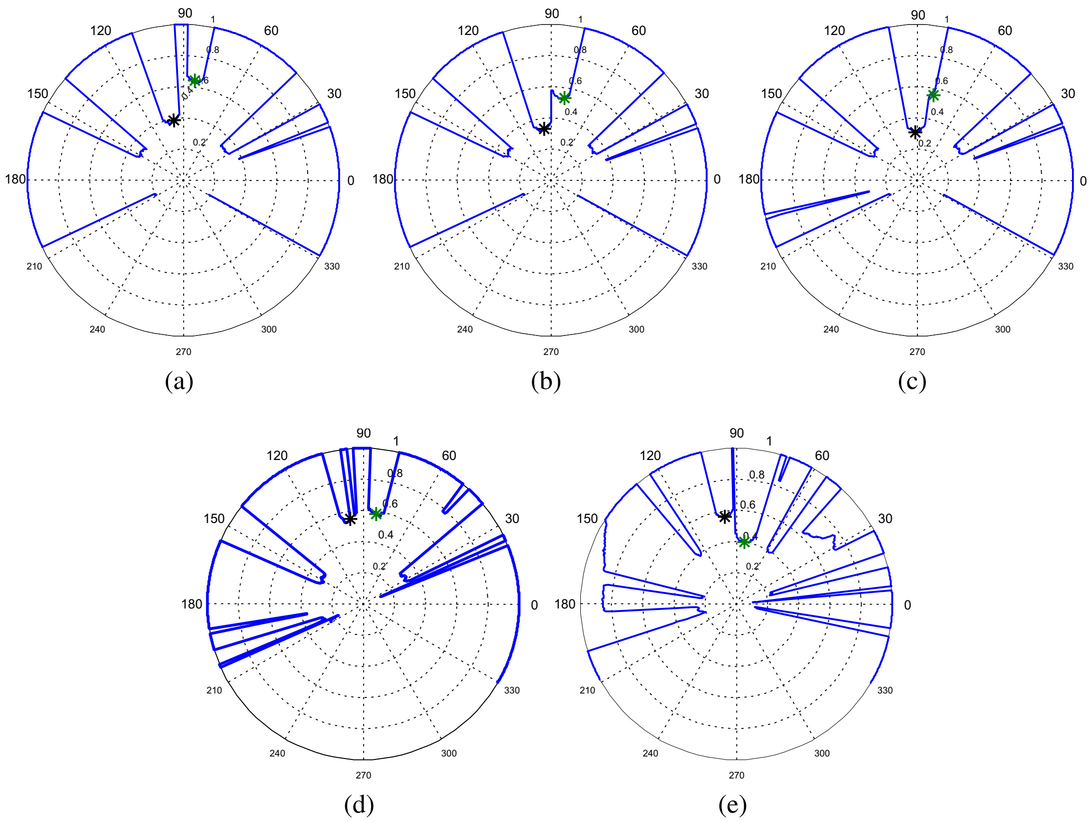


Figure 4.24: Patterns results: a) SL; b) LT; c) OL and Non patterns on d) and e).

Table 4.8: RMSD error between the proposed LRF system and the ground truth. Mean±Standard deviation is presented.

	RMSD (mm)	
	X-axis	**Y-axis**
(i) Stand in Front of the Walker	20.1 ± 0.15	27.2 ± 0.09
(ii) Walking with the Walker	21.2 ± 2.35	28.5 ± 2.18

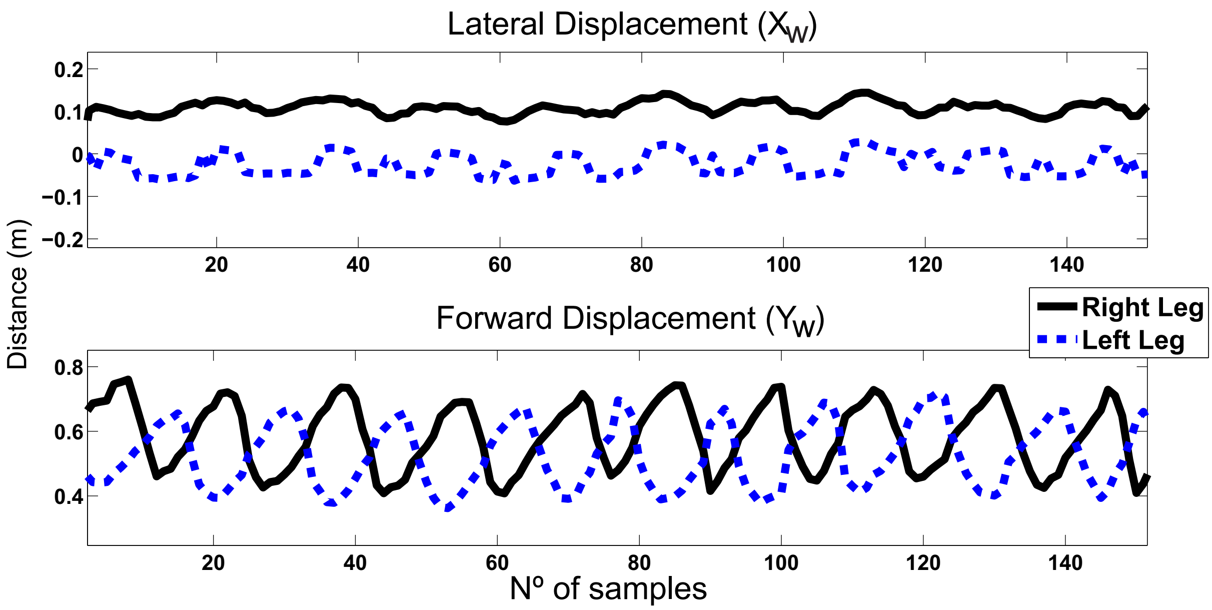


Figure 4.25: LFR distance signal in Y- and X-axis.

(ii) Subjects Walking with the Walker

In order to verify the error of the proposed system when the subjects are moving, a Vicon (www.vicon.com) motion capture system (ground truth) was used in order to calculate RMSD. As it can be seen in table 4.8, the error increases a little in comparison with the standing situation but is still small regarding the application of this method. Since it is intended to calculate gait parameters that usually are expressed in mm, the error is acceptable. In figure 4.25, it is shown an example of the LRF distance signal in a straight-forward trajectory.

4.2.4.4 Conversion of LRF coordinates to real world coordinates

(i) Estimation of error between the displacement estimation of the proposed LRF sensor's method and the real displacement.

In order to estimate the correction coefficient between the displacement estimation of the proposed LRF sensor's method and the real displacement, the distance measured on a straigth forward walkway was compared with the distance measured by the LFR. It is noteworthy that 14 different subjects (Table 4.5) with different calf lengths were chosen to verify if this length has some influence on the measured correction coefficient. With this information, the correction coefficient can be calculated as follows:

$$\bar{k} = \frac{1}{N}\sum\frac{\text{RealDistanceLength}}{\text{LRFDistanceLength}} \tag{4.15}$$

Table 4.9: Calf length and the corresponding mean error for each subject

Subject	Calf length (m)	$\bar{k}$
1	0.55	1.86
2	0.52	1.65
3	0.54	1.55
4	0.51	1.60
5	0.50	1.68
6	0.51	1.59
7	0.49	1.60
8	0.35	1.77
9	0.48	1.64
10	0.38	1.67
11	0.63	1.66
12	0.47	1.63
13	0.53	1.65
14	0.47	1.62
Total Mean		1.66±0.08

where N corresponds to the total of trials. After data analysis of the 14 subjects, $\bar{k}$, for each subject, was found and it is presented in table 4.9.

This mean correction coefficient $\bar{k}$ will correspond to k from eq. 4.14 and its average value is equal to 1.66. Analyzing all distances' length measured by the LRF with the correction coefficient k the general error is ±20%.

Since each person has his own style of walking and the relation between calf length and mean correction coefficient does not seem to follow a pattern, it is difficult to find a model that can define this relation. In addition, although the error variability (0.08) is small, it was still obtained an error of 20% and since this estimation is very important for the calculation of spatial parameters, a solution is required. Thus, it is recommended to execute, before the rehabilitation program, a calibration test (it can be integrated during OC) where the subject walks through a known distance. Then this distance is introduced on the program that will calculate the corresponding k to each walker user. With this calibration, the error was reduced to ±10% with these subjects.

4.2.5 Conclusions

This paper presented a system able to track the legs position during an assisted walk without equipping the user. A LRF sensor was used by a new detection algorithm that suits for all subjects through a calibration mode. Preliminary results show that this system has high potential to be used on clinical treatment with patients on the hospital to give clinical insight to the clinicians. Further work is necessary to run more tests and reduce the errors associated with the calculation of spatiotemporal parameters.

4.3 Spatiotemporal Evaluation in Walker Rehabilitation

This section aims to present the calculation of some specific spatiotemporal parameters while the user is walking with the walker. These parameters are important to evaluate the state of the user and infer his evolution in the rehabilitation program, for instance. Many disorders are characterized by spatiotemporal parameters, and their modification can bring insight into the diagnostic of the user. Many examples can be presented [152]: after a fall, people tend to enlarge their support base and present higher stance duration; Parkinsonian festination corresponds to an inconstant speed and short steps; multiple infarcts syndromes are related to small steps; Ataxic patients increase their support base, small steps, low velocity and very insecure gait; arthroses' patients present asymmetric gait and small steps.

Thus, the following spatiotemporal parameters should be calculated (Figure 4.26): step and stride length, stride width, gait cycle, cadence, velocity, stance and swing phase duration, double support duration and step time [144, 153]. In order to calculate these parameters it is needed to detect three main events: Heel strike, toe off and legs crossing (Figure 4.26). This is possible through the use of the aforementioned sensor systems: ADS and/or LFR.

The gait cycle (Figure 4.26) has its beginning and end in successive events of the same foot, thus identifying repetitive events that can be characterized by one cycle time. During the gait cycle, the relative distance between the lower limbs and walker varies as follows: increases when the lower limb moves more slowly than the walker or when it is stopped (stance phase); decreases when the lower limb moves forward with a faster speed than the walker (swing phase). These variations are illustrated in figure 4.26 for one lower limb. It is also possible to observe that the distance between the lower limb and the walker is maximum at the end of the stance phase and before the beginning of the swing phase (time when the lower limb goes from the state of being stopped to the state of approaching the walker).

Figure 4.27 illustrates the relevant events detection. The maximum values (squares) correspond to toe-off (TO) events; the minimum values (white circles) correspond to heel-strike (HS) events; and when the signals are crossing (d_cross) it means that the lower limbs are also crossing (blue circles) [19, 154]. td_R , d_R (time and distance right lower limb), td_L and d_L (time and distance left lower limb) are generic points chosen to exemplify how to calculate the parameters. These events enable to calculate the following parameters ($i = R, L$, where R corresponds to the right leg and L corresponds to the left leg). Also, the same events in

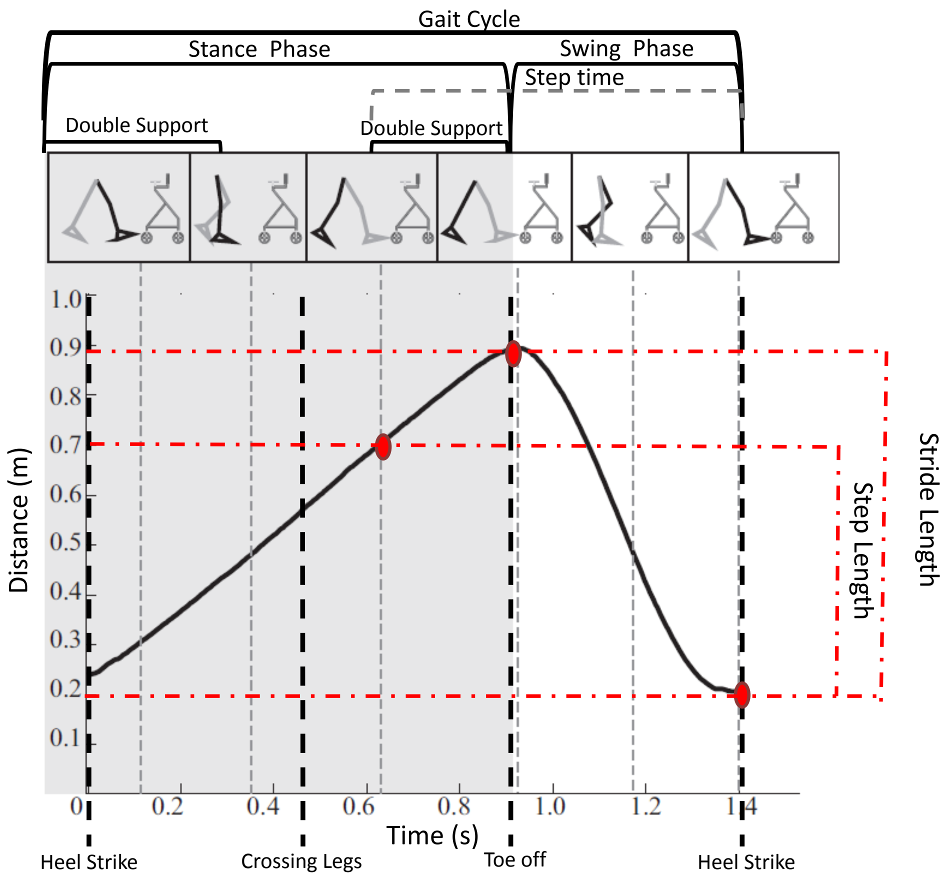


Figure 4.26: Relative distance between one leg (black) and the walker, during one gait cycle.

consecutive gait cycles are represented by *d'* notation and time instant is represented by *t*.

- Stride Length (*STR*) and Gait Cycle (*GC*): are the distance and time, respectively, between toe-off events from the same foot. It is calculated through the difference/instant between the maximum distance and consecutive minimum.

$$STR_i = d_{iTO} – d_{iHS} \tag{4.16}$$

$$GC_i = |td_{iTO} – td'_{iTO}| \tag{4.17}$$

- Step Length (*STP*) and Step Time (*STPT*): is the distance and time, respectively, between heel strike of one foot and consecutive heel strike of the other foot.

$$STP_R = d_{LHS} – d'_{RHS} \tag{4.18}$$

$$STP_L = d_{RHS} – d_{LHS} \tag{4.19}$$

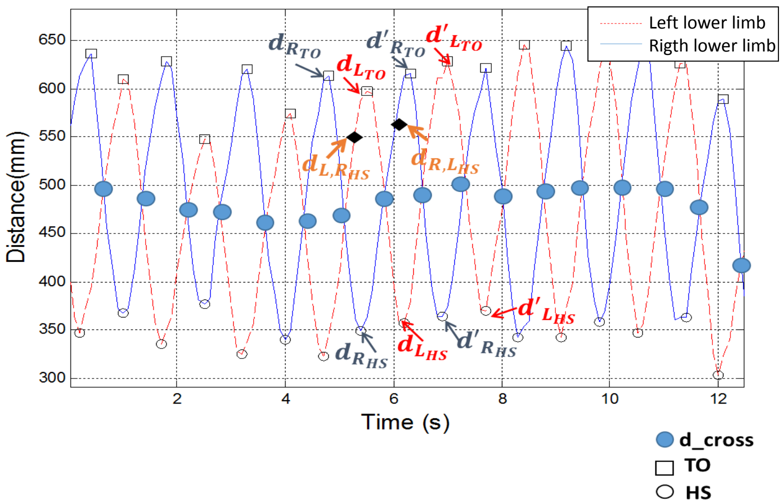


Figure 4.27: Distance LRF signal of both lower limbs walking in forward direction. The squares correspond to toe-off events (TO), the white circles to heel-strike events (HS) and the blue circles to lower limbs' crossing events (d_cross). The same events in consecutive gait cycles are represented by *d'* notation.

$$STPT_R = |td'_{RHS} – td_{LHS}| \tag{4.20}$$

$$STPT_L = |td_{LHS} – td_{RHS}| \tag{4.21}$$

- Cadence (frequency of the signal -*CAD*) and average velocity (*Avspd*):

$$Avspd_i = (d_{iTO} - d_{iHS})/(td_{iHS} - td_{iTO}) \tag{4.22}$$

- Stride width (*WIDTH*): is the distance between both lower limbs in width.

- Swing duration (*SWD*): is the time correspondent to the oscillation phase, when the foot is not on the ground. It is calculated as the time between toe off and and heel strike of each foot.

$$SWD_i = td_{iHS} - td_{iTO} \tag{4.23}$$

- Stance duration (*STAD*): is the time correspondent to the support phase, when the foot is on the ground. It is calculated by the time between heel strike and toe off of each foot.

$$STAD_i = td'_{iTO} - td_{iHS} \tag{4.24}$$

- Double support time (DS): is the time when both feet are on the ground. It is calculated when both signals present a positive derivate. This happens between the heel strike of one foot and the toe off of the other foot:

$$DS = td_{TO} - td_{HS} \tag{4.25}$$

However, these values correspond to the distance between the user and the walker. In order to convert these parameters into distances walked on the world a new method is proposed.

4.3.1 Conversion of Lower limbs coordinates to world displacement

If one calculates the distance-based parameters directly from the forward direction signal, their values will correspond to the distance between the user and the walker. To obtain the spatiotemporal parameters relative to the world it is necessary to estimate the displacement of the subject in the world reference (X_0, Y_0). Thus, the displacement of the subject's lower limbs (LWL) is needed regarding the displacement of the walker in the world. Such displacement of the walker can be calculated through two approaches: encoders' information (odometry) or stance LWL distance [155]. Since odometry presents cumulative errors, stance LWL distance method is adopted.

For obtaining the real displacement of the subject, it is necessary to estimate the displacement of the subject (dX_{p0} and dY_{p0}) in the world reference (X_0, Y_0). Looking at figure 4.22, it is necessary to know the displacement (dX_w and dY_w) of the walker in world axis (X_0, Y_0) and the displacement (dX_{pw} and dY_{pw}) of the subject's LWL (the one in swing phase) relative to the walker axis.

By this:

$$dX_{p0} = dX_w + dX_{pw} \tag{4.26}$$

$$dY_{p0} = dY_w + dY_{pw} \tag{4.27}$$

The displacement of the subject's LWL in swing phase (dX_{pw} and dY_{pw}) as well as the displacement of the walker (dX_w and dY_w) are obtained through the LRF and/or ADS signal, accordingly to the proposed approach, described in the following subsection.

Lower limbs's coordinates are expressed as (X_{pw}, Y_{pw}). It is noteworthy that: $X_L = X_{Pw} = Xf$ and $Y_L = Y_{Pw} = Yf$ (aforementioned nomenclature in ADS system).

(ii) Calculation of walker's and subjects' Displacement

By measuring the relative distance between the walker and the LWL that is on the stance phase, it is possible to infer how much the walker displaced. Since the LWL in the stance phase is fixed on the floor, the variation in distance measured by the sensor (gray region of the graph in figure 4.26) is due only to the displacement performed by the walker. Since there is always a foot on the ground during gait, it is possible to measure such displacement. By this, walker displacement is obtained with the consecutive coordinates in time (t) of the LWL that is on stance phase:

$$dX_w = X_{Pw(t)} - X_{Pw(t-1)} \tag{4.28}$$

$$dY_w = Y_{Pw(t)} - Y_{Pw(t-1)} \tag{4.29}$$

The subjects displacement (dX_{pw} and dY_{pw}) is obtained by the same equations 4.28 and 4.29, but with the coordinates of the LWL on swing phase.

4.3.2 Results

4.3.2.1 Gait Events Detection

Figure 4.28 (first graph) illustrates the gait events detection (HS and TO) that enable to calculate the spatiotemporal parameters (section 4.3). In the second graph, consecutive strides are represented for one patient in % of gait cycle (left and rigth lower limbs are represented by a dashed line and continuous line, respectively). On this figure, intra-individual variability can be observed in space and time. It can be seen that the subject presents an asymmetrical gait as his LWL's do not cross each other. Despite these variabilities, the proposed method algorithm has still the capacity to track both LWLs and the developed gait assessment tool detects all gait events.

Now that the gait events are correctly detected, calculation of the real displacement of the user with the walker is possible for spatiotemporal parameters' calculation.

4.3.2.2 Calculation of the Walker Displacement

Figure 4.29 shows two trajectories in the walker axis, each acquired with one type of sensor. The trajectories consist in walking forward, turn and then walk forward again. The detection of the LWL in stance phase, where a flag is 0 when it detects the left LWL and 1 when it detects the right one. In figure 4.29a the trajectory was acquired with LRF. The turn is performed

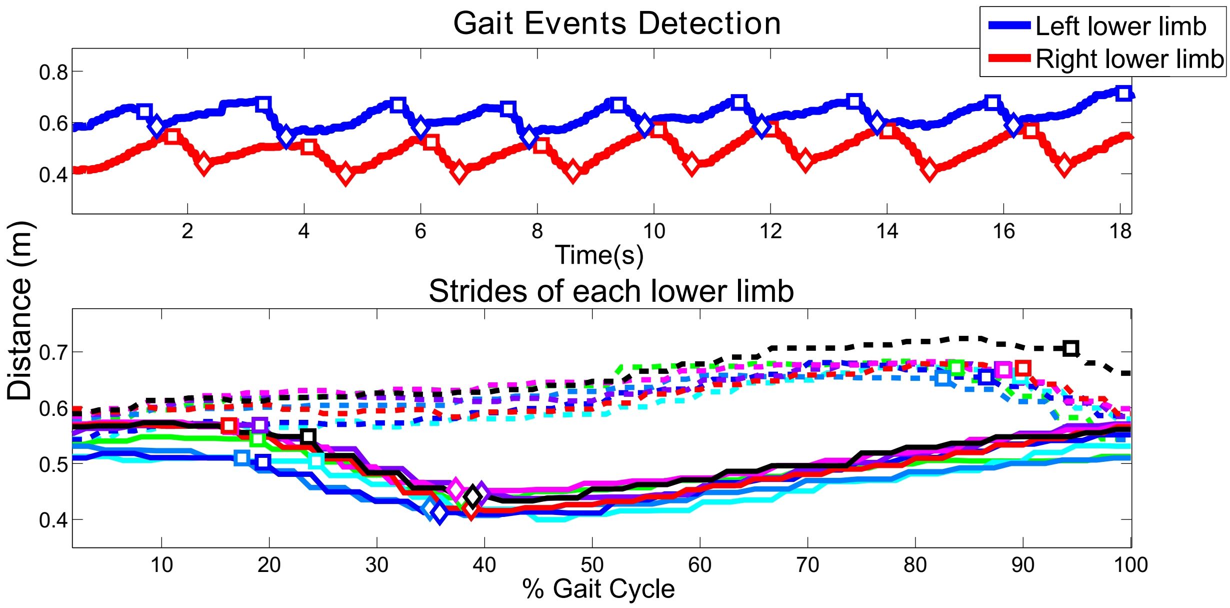


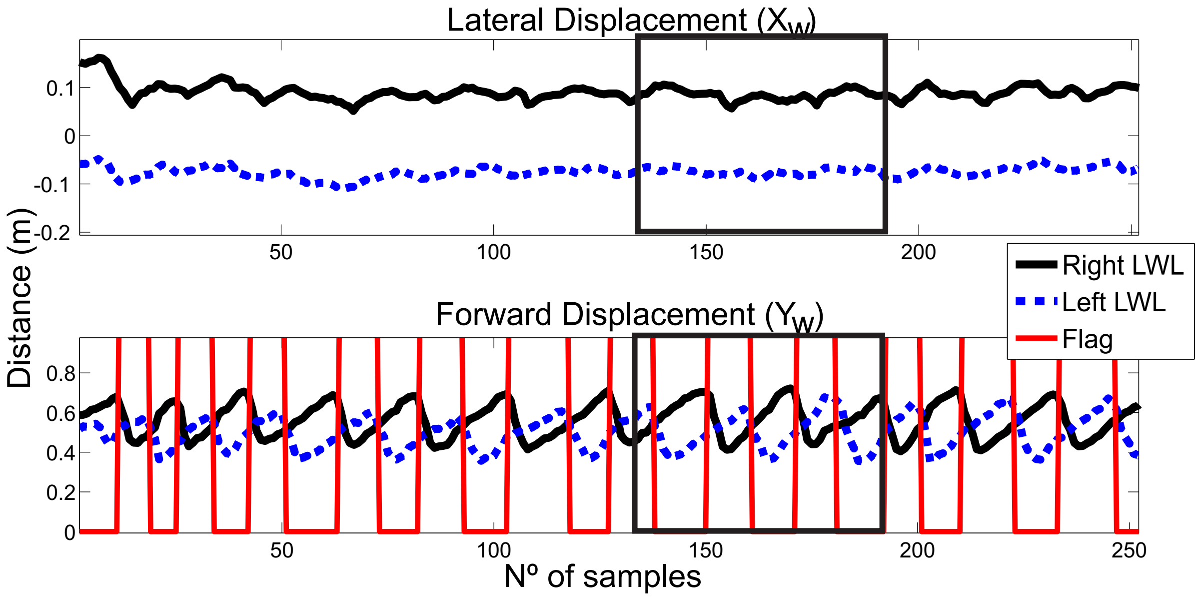
Figure 4.28: Distance signal with the detected gait events results. Squares identify HS events and Diamonds identify TO events. Left and rigth lower limbs are represented by a dashed line and continuous line, respectively, on the above graph.

between sample 140 and 190, however no change in the signals are visualized. In figure 4.29b the trajectory was acquired with ADS. The turn is performed between sample 220 and 300, as it can be slightly seen on the X -direction.

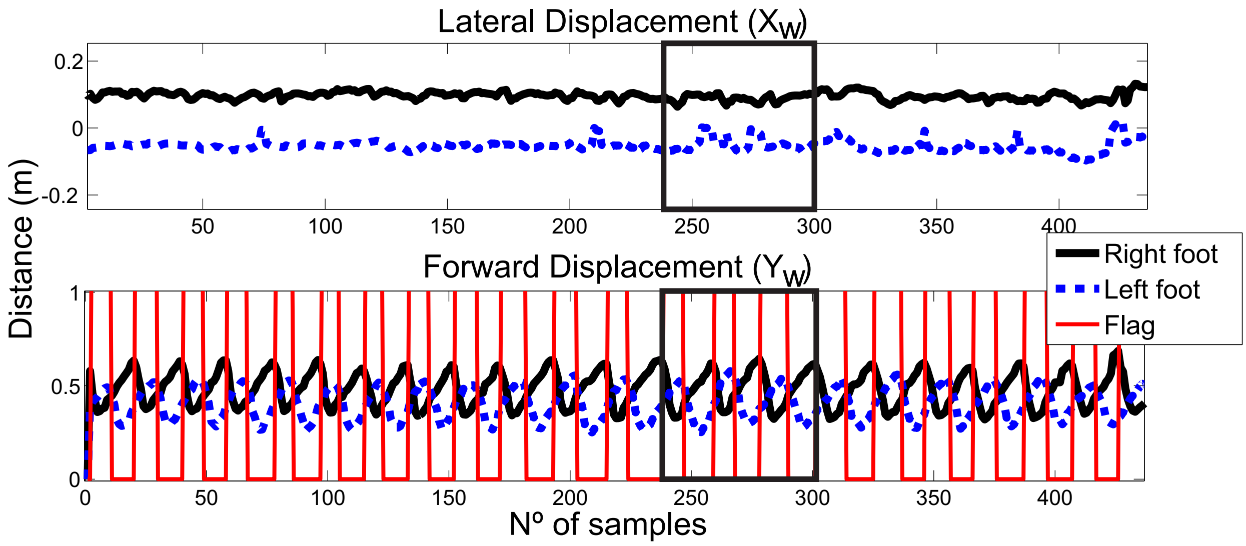
With the detection of such events, it is possible to estimate the displacement of the walker. The LWL which is not considered to be on stance phase, is used for the estimation of walker user displacement.

In figure 4.30, it can be seen two trajectories, in the world axis, each acquired with one type of sensor. The two legs are represented by dots: red dots represent the movement of the leg in swing phase and green dots represent the stance leg (which is stopped). The black line represents the trajectory of the walker in the real world, calculated through the model presented in section 4.3.1. In figure 4.30a it is represented a trajectory acquired with LRF sensor. It consisted on walking 3 m forward and perfom a curve of $\pm 45^{\circ}$. By using the k-correction and online calibration methods an error of ± 35 cm was obtained. However, as it can be seen in figure 4.30a, the curve is not detected. It was concluded that with the legs' movement it is difficult to detect a curve.

In figure 4.30b it is represented the same trajectory with ADS. It is possible to verify that the turn is detected with this system. Thus, with feet movement the intention to perform a curve is better expressed. This was expected since the feet present an orientation that contains the "intention" of the direction the user wants to follow. This demonstrates that the feet's



(a)



(b)

Figure 4.29: Two trajectories in the walker axis: a) Stance phase detection with LRF; b) Stance phase detection with ADS. A right turn is represented by a black box. Flag is 0 when the left leg is on stance phase and 1 when it is the right one.

postion may be used in control purposes.

4.3.2.3 Validation of the Spatiotemporal Parameters acquired with LRF

In order to validate the calculated spatiotemporal parameters with the LRF system and the proposed algorithm and calibration, the tests with the walker were video-recorded for temporal validation and the steps were marked on the floor for spatial validation. 14 subjects (4.5) were asked to walk straight-forward under a plastic walkway and wore special shoes that marked the walkway with their steps. Then, the spatioparameters were measured to be used as ground-truth for this validation.

For each patient, gait analysis was performed calculating the spatiotemporal parameters presented in section 4.3 with LRF signal and compare them with the ground-truth parameters.

A one-way ANOVA (analysis of variance) was performed to the calculated gait parameters and their errors regarding the ground truth in order to verify differences among/within patients. The level of significance was set to $p<0.05$.

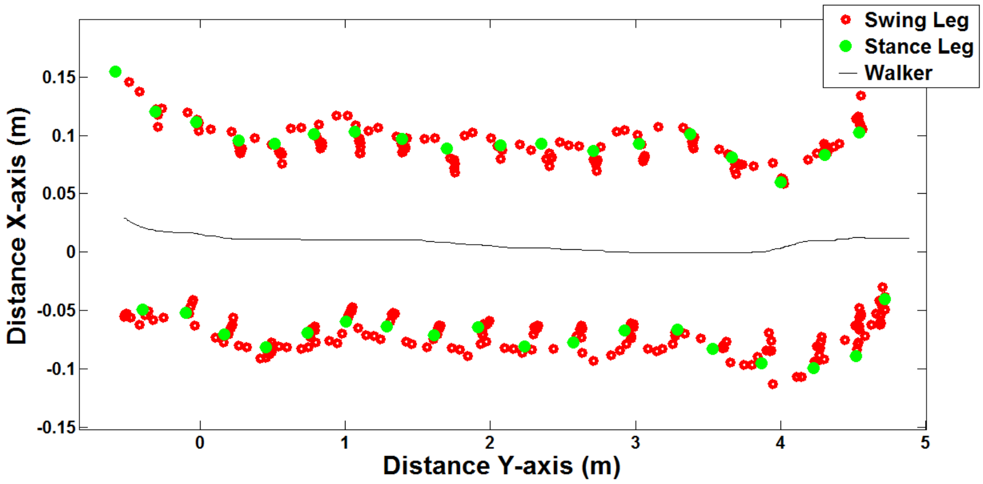
Results found that gait parameters were significantly different among patients ($p<0.05$). Thus, the authors concluded that the addressed sample group was significant to test the proposed system, due to the strong variability in terms of gait parameters between the 14 patients. They also present different heights, age, calf lengths and were dressed differently (since this can influence the algorithm detection).

It was also verified that errors between the measures with LRF system and ground truth are small. In the overall of the 14 patients, the greatest error for a parameter was 17%. Between different patients the temporal errors are not significantly different ($p>0.05$), which means that they are systematic among patients. However, the spatial errors are significanly different among patients ($p<0.05$). By making an analysis within each patient, it is verified that the errors obtained for spatial parameters are not significantly different ($p<0.05$). This means that the spatial errors are systematic for the same user, since it depends on the k parameter's choice.

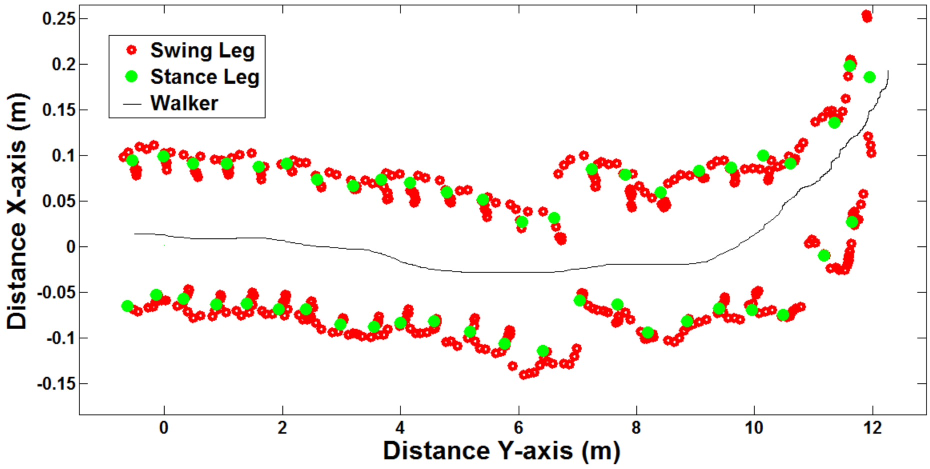
Results show that for gait analysis purposes, the presented algorithm with LRF sensor has great potential to be further used to provide clinical insight during walker rehabilitation with very small error.

4.3.2.4 Validation of the Spatiotemporal Parameters acquired with ADS

In order to validate the spatiotemporal parameters obtained with ADS system, such parameters were compared on a step-by-step basis against the data derived from the Codamotion as a reference, using Intraclass Correlation Coefficient (ICC(2, k) as reported previously [156]. The mean true error between these two systems was also examined on a step-by-step basis.



(a)



(b)

Figure 4.30: Two trajectories in the world axis: a) Trajectory obtained with LRF signal; b) Trajectory obtained with ADS signal.

Such error was calculated for all spatiotemporal gait parameters as the difference between the ADS and Codamotion values. Then, ANOVA (analysis of variance) will be performed for each parameter error to verify if there are differences among patients in terms of errors and parameters. The result of this analysis will inform if the errors can be considered systematic. The level of significance was set to $p<0.05$

After calculating the errors between ADS and Codamotion for each gait parameter of the fourteen subjects (4.1), it is verified that the average errors of spatial parameters varied between 2% and 4% and the temporal parameters varied between 1% and 3%. The greater errors correspond to patients with greater stride lengths. This happens because their feet in toe-off and heel strike are too far and too close from the camera, respectively. This will change feet's shape, which lead the centroid to change its position more often, increasing the error. By performing ANOVA, all parameters presented significant differences between patients ($p<0.05$) showing that patients present significant different gait patterns between each other. Then, we verified that errors of most spatiotemporal parameters have no significant differences between patients ($p>0.05$). Only step width (*WIDTH*), stride length (*STR*) and step length (*STP*) present significant differences ($p<0.05$), not being considered as systematic errors. Despite this remark and the errors being significantly different among patients, they are small (less than 4cm and 0.8s) regarding the purpose of this study, which is the calculation of spatiotemporal parameters. In addition, the ICC for temporal parameters *GC*, *STPT*, *DS*, *STAD, SWD* was found to be greater than 0.86, indicating good agreement. Similarly, for spatial gait parameters—*STR, STP, WIDTH, Avspd*—the ICC was found to be greater than 0.90. These results show good to excellent concurrent validity in spatiotemporal gait parameters demonstrating the great potential of using ADS as a gait analysis tool.

4.3.3 Conclusions

The main goal of this study was to develop two techniques based on LRF and ADS for accurate gait parameter determination.

First a technique for detection of legs, based on LRF, for different legs postures during assistive device rehabilitation. Comparing to the existing system in literature, the main novelty of the proposed one is the ability to calibrate parameters for each subject online, rather than fixing a single set of values for all subjects. Results show small errors and the capability of being used in real-time approaches.

Then, a feet position and orientation detection algorithm is proposed. It is based on a ADS and does not required the use of any marker. The obtained results are compared with a ground truth provided by a motion tracking system to experimentally assess the performances of the proposed algorithm. Comparing to the existing systems in literature, the proposed algorithm is faster and simpler, based in image segmentation. Also, it can be used in real-time approaches, which does not happen with the compared systems. Finally, the errors are small compared with the non-markers systems.

With the estimation of lower limbs position during assisted walk, it was possible to calculate the corresponding spatiotemporal parameters. Experimental results are provided to show relevant data can be extracted for assisted gait analysis with small error.

Between those systems, LRF is more precise (21 mm in X-direction and 28 mm in Y-direction in terms of RMSD) when acquiring the trajectories than ADS (35 mm in X-direction and 40 mm in Y-direction in terms of RMSD). However, when calculating the spatiotemporal parameters, ADS presents the lower error (under 5%). This happens because the transformation of the coordinates from the walker axis to the world axis with LRF depends on a correction coefficient k. This coefficient brings more error to the LRF trajectory estimation. Since ADS does not need any factor, this system is more accurate.

However, both systems present algorithms that may fail in some specific situations. Thus, it is not possible to exclude one of them for gait analysis. In section 4.4 it is presented a preliminary approach of how can these two systems be used together in gait analysis.

By this, physiotherapists can make routine analysis of their patients and infer the evolution and recovery of the patients. Thus, they can quantitatively assess their performance. Nowadays, elderlies and other patients in their recovery process are sometimes evaluated with observational and timed scales like Time Up and Go (TUG) or 6-minute walk distance (6MWD) [8, 41, 42, 135, 141, 157]. These scales are very poor in terms of evaluation parameters, since they provide only speed information. With ADS and LRF systems it is possible to extend this evaluation for more parameters, since they are capable of acquiring feet/leg position through any trajectory.

The same did not happen with Codamotion analysis that was only capable of acquiring data when the patients were walking forward. Thus, occlusion problems are eliminated with ADS and LRF systems. Other advantage is to collect relevant data about the gait evolution and the adequacy of the device use. These data will be stored and processed to assess any misuse of the aid and to foretell any decline of walking capabilities.

The main contribution of this study is that the proposed method was tested with walker end-users showing to be feasible and that can help in rehabilitation.

4.4 Multi-sensor Data Fusion based on Laser Range Finder and Active Depth sensors' data

This chapter has been focused on addressing embedded gait analysis systems installed on walker types. The great advantage of such systems is that the user stands at a known position with regards to the walker and lower limbs (LWL) tracking is then made in an easier way. Direct measurement of LWL's segments may be obtained by placing laser range finder sensors (LRF) (section 4.2), or active depth sensors (ADS) on a walker device (section 4.1).

LRF approach does not need to add any markers on the patients' limbs. In section 4.2, this sensor was used as a gait assessment system, showing that the proposed algorithm based on pattern detection was suitable for different subjects. Despite the good results, one problem that can appear is the situation of having large pants or skirts, which will lead to false detections and make the algorithm to be impracticable. Also, this algorithm needs an online calibration that if is not done properly, the algorithm detection errors increase.

Other possibility for LWL's tracking was proposed in section 4.1 using ADS. The feet tracking algorithm was tested with healthy subjects and patients and satisfied the required qualities to be suitable as a gait assessment system (portable, marker-less and low computing cost). However, problems arise when feet are too close or too far from ADS, increasing the measurement errors.

In order to be able to adapt to different situations (environment, trajectories, clothes, step lengths, step widths, etc.), a fusion between the LRF legs detection algorithm output and ADS feet detection algorithm output is proposed. This will enable to fuse advantages of both approaches and minimize disadvantages. Results from this approach will be used to decrease the errors of calculating spatiotemporal parameters.

In addition, both systems are portable which allows the clinician to perform gait analysis whenever he wants and in the type of walker he wants. In figure 4.31, both systems are installed on the ASBGo walker.

To perform such data fusion a fusion filter - Decentralized Information Filter - tuned by a fuzzy logic supervisor is designed to get a more reliable estimation of the LWL positions in different situations.

4.4.1 Data Fusion

Through LRF and ADS, it is possible to estimate the 2D positions of LWL in relation to the walker, as it was mentioned above. Since each sensor is prone to errors, and in order to improve such data information, sensorial data fusion is adopted. Thus, 2D coordinates

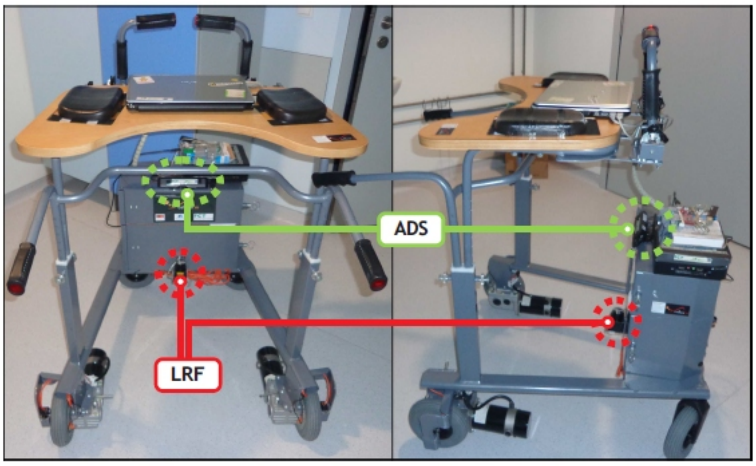


Figure 4.31: ADS and LRF systems placed on the ASBGo.

(x, y) for each right (R) and left (L) legs will be estimated through the implementation of Information Filter (IF) and the Decentralized Information Filter (DIF) [158]. The outputs of the LRF system will be represented as $(x_{LLRF}, y_{LLRF}; x_{RLRF}; y_{RLRF})$ and for the ADS system as $(x_{LADS}; y_{LADS}; x_{RADS}; y_{RADS})$.

4.4.1.1 Information Filter

The Information Filter (IF) is essentially a Kalman Filter (KF) expressed in terms of information measures on the states of interest, rather than estimates of states and their associated covariances [158]. It has been shown that IF have advantages over the KF in multisensor data fusion applications. These include reduced computation, algorithmic simplicity and easy initialization. In particular, these attributes make the IF easier to decouple, decentralize and distribute. These are important filter characteristics in multisensor data fusion systems.

The Information filter equations will be presented in the following.

The information matrix, Y_k, is the inverse of the covariance matrix,

$$Y_k = P^{-1}_{(k|k-1)} \tag{4.30}$$

it relates $P^{-1}_{(k|k-1)}$ at the previous time step $k-1$ to Y_k at the current time step k.

The information state vector, $\hat{y}_k$, is a product of the inverse of the covariance matrix (information matrix) and the state estimate,

$$\hat{y}_k = P^{-1}_{(k|k-1)} \hat{x}_{(k|k-1)} = Y_k \hat{x}_{(k|k-1)}, \tag{4.31}$$

The information state contribution $i_{(k)}$ from a measurement $z_{(k)}$, and its associated information matrix $I_{(k)}$ are defined, respectively, as follows:

$$i_k = H_{(k)}^T R_{(k)}^{-1} z_{(k)} \tag{4.32}$$

$$I_k = H_{(k)}^T R_{(k)}^{-1} H_{(k)} \tag{4.33}$$

$H_{(k)}$ is the matrix that relates the measurements with states (measurement model) and $R_{(k)}$ is the measurement error covariance matrix.

The information propagation coefficient $L_{(k|k-1)}$, which is independent of the observations made, is given by the expression

$$L_{(k|k-1)} = Y_{(k|k)} A_k Y_{(k|k-1)}^{-1} \tag{4.34}$$

where $A_{(k)}$ is a state transition matrix.

With these information quantities well defined, the Kalman filter can now be written in terms of the information state vector and the information matrix.

The equations correspondent to the IF prediction are:

$$\hat{y}_{(k|k-1)} = L_{(k|k-1)} \hat{y}_{(k-1|k-1)} \tag{4.35}$$

$$Y_{(k|k-1)} = [A_{(k)} Y_{(k-1|k-1)}^{-1} A_{(k)}^T + Q_{(k)}]^{-1} \tag{4.36}$$

where $Q_{(k)}$ is the process error covariance matrix.

IF estimation equations correspond to:

$$\hat{y}_{(k|k)} = \hat{y}_{(k|k-1)} + i_k \tag{4.37}$$

$$Y_{(k|k)} = Y_{(k|k-1)} + I_k \tag{4.38}$$

Thus, the output estimation state is equal to:

$$x_{(k|k)} = Y_{(k|k)}^{-1} \cdot \hat{y}_{(k|k)} \tag{4.39}$$

4.4.1.2 Decentralized Information Filter

A decentralized system consists of a network of filters, each one with its own processing unit. In such a system, the fusion occurs locally on each node, based on local information and information transmitted from neighbor filters. For a decentralized system of data fusion, the

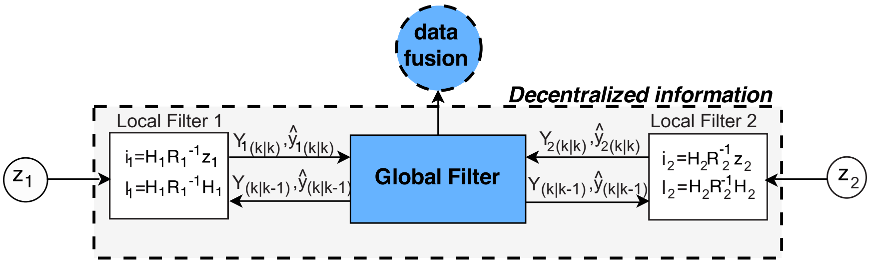


Figure 4.32: Decentralized Information Filter.

local filter is a sensor node, which distributes measurements and local information to other fusion nodes. Later, it assimilates this information and computes a local estimation, i.e., in the decentralized system, the local filter/node uses the information to generate a fused local output. In figure 4.32 it is presented an example of a scheme of a general DIF with *n-th* local filters.

In this work, the information ($i_{n(k)}$ and $I_{n(k)}$) comes from N sensors (LRF and ADS, i.e. N = 2). Thus, the formulation of the DIF run in the filters or nodes in the instant k, for the *n-th* local filter one has

$$\hat{y}_{n(k)} = \hat{y}_{(k|k-1)} + i_{n(k)} \tag{4.40}$$

$$Y_{n(k)} = Y_{(k|k-1)} + I_{n(k)} \tag{4.41}$$

whereas for the global filter one has

$$\hat{y}_{(k)} = \sum_{n}^{N} \hat{y}_{n(k)} - (N-1)\hat{y}_{(k|k-1)} \tag{4.42}$$

$$Y_{(k)} = \sum_{n}^{N} Y_{n(k)} - (N-1)Y_{(k|k-1)}. \tag{4.43}$$

4.4.1.3 State Estimation

The global filter (DIF) has the following configuration:

$$x_{(k|k-1)} = Ax_{(k-1|k-1)} + w_{(k-1|k-1)} \tag{4.44}$$

$$z_{(k|k-1)} = Hx_{(k-1|k-1)} + v_{(k-1|k-1)} \tag{4.45}$$

where $w_{(k-1|k-1)}$ and $v_{(k-1|k-1)}$ are the noise vectors. The $x_{(k|k-1)}$ and $x_{(k-1|k-1)}$ are state vectors. It relates the state at the previous time step $k-1$ to the state at the current time step k. In this work, the output variables for each sensor i:

$$x_i = [x'_{Ln}, y'_{Ln}, x'_{Rn}, y'_{Rn}], \tag{4.46}$$

where x and y represents the 2D position of the LWL, and the indices l and r represent left and right, respectively. As aforementioned, the information obtained to estimate the LWL position comes from the LRF ($n = 1$) and ADS ($n = 2$) sensors. Thus, DIF has two local IF. For each local IF, n, the general configuration of measurement vector, observation model, state transition matrix, measurement and process error covariance matrices are, respectively:

$$z_n = [x_{Ln}, y_{Ln}, x_{Rn}, y_{Rn}] \tag{4.47}$$

$$H_n = \begin{bmatrix} 1 & 0 & 0 & 0 \\ 0 & 1 & 0 & 0 \\ 0 & 0 & 1 & 0 \\ 0 & 0 & 0 & 1 \end{bmatrix} \tag{4.48}$$

$$A = \begin{bmatrix} 1 & 0 & 0 & 0 \\ 0 & 1 & 0 & 0 \\ 0 & 0 & 1 & 0 \\ 0 & 0 & 0 & 1 \end{bmatrix} \tag{4.49}$$

$$R_n = \begin{bmatrix} \sigma_{lx}^2 & 0 & 0 & 0 \\ 0 & \sigma_{ly}^2 & 0 & 0 \\ 0 & 0 & \sigma_{rx}^2 & 0 \\ 0 & 0 & 0 & \sigma_{ry}^2 \end{bmatrix} \tag{4.50}$$

$$Q = \begin{bmatrix} \sigma'^2_{lx} & 0 & 0 & 0 \\ 0 & \sigma'^2_{ly} & 0 & 0 \\ 0 & 0 & \sigma'^2_{rx} & 0 \\ 0 & 0 & 0 & \sigma'^2_{ry} \end{bmatrix} \tag{4.51}$$

where σ^2 is the variance of the measurement noise and σ'^2 is the variance of the process noise.

Output from these filters will result in data whose variance is smaller than the smallest variance associated to the data measured by each sensor. In order achieve this, special attention to matrices R_n and Q have to be taken. These matrices are used to specify the quality of information that flows in the filter.

Matrix Q is built based on the process error of the global system and matrix R is builtd based on the sensors' measurement error. They have a fundamental role in the IF performance since they define the weight by which the process and each sensor measurement contribute to the estimated state. If a sensor provides information with a low level of noise, the matrix R should be constructed to take into account these considerations.

The determination of the process error covariance matrix Q is generally more difficult as it is difficult to have the ability to directly observe the process that is being estimated. Sometimes a relatively simple process model can produce acceptable results if enough uncertainty is "injected" into the process via the selection of Q. Therefore, let us assume that the process matrix Q is constant, i.e., the error presented in the system does not change as the user is walking. We assumed the matrix Q (through trial and error) as follows,

$$Q = \begin{bmatrix} 150 & 0 & 0 & 0 \\ 0 & 150 & 0 & 0 \\ 0 & 0 & 150 & 0 \\ 0 & 0 & 0 & 150 \end{bmatrix} \tag{4.52}$$

These values were intuitively assumed to be the amount of noise variance that the process could add. This was based on trial and error.

Regarding R_n matrix, it stores the weight relative to the LRF ($n = 1$) and ADS ($n = 2$) data. If R_1 values are higher than R_2 values, then LRF data weights more in the estimation than ADS data. On the other hand, if R_2 values are higher than R_1 values, then ADS data contributes more to the user's walking trajectory estimation. Since the contribution of each sensor may change due to different situations, a supervisor was designed to change R_1 and R_2 matrices. The default (initial) values of R matrices are:

$$R_1 = \begin{bmatrix} 100 & 0 & 0 & 0 \\ 0 & 120 & 0 & 0 \\ 0 & 0 & 100 & 0 \\ 0 & 0 & 0 & 120 \end{bmatrix} \tag{4.53}$$

$$R_2 = \begin{bmatrix} 9 & 0 & 0 & 0 \\ 0 & 64 & 0 & 0 \\ 0 & 0 & 9 & 0 \\ 0 & 0 & 0 & 64 \end{bmatrix} \tag{4.54}$$

Therse matrices were defined based on the calculation of the sensors' measurement error, and then adjusted through trial and error.

4.4.1.4 Fuzzy Logic Supervisor

Changing the value of the R matrix allows to define which sensor has more weight to contribute to the estimation of LWL's positions. Therefore, in order to obtain an estimation with low and bounded error and with smooth transitions, it was decided to merge data from LRF and ADF through the DIF with adaptive R_n matrices. To decide if R matrix has to be changed, a fuzzy logic supervisor [159] has been implemented.

The supervisor will act when:

- The ADS does not track the user's feet when they are too close of the ADS;
- The light is poor and the ADS camera detection is not so trustful;
- Detection of strange objects from the ground by the ADS, acquiring false feet;
- Some person or object appears near the person less than 1 m from the walker, and thus are detected by the LRF making the LRF to detect them;
- If the user has large pants, short pants or pants that change the leg's format, its format on the leg while the user is walking, the LRF detection becomes less trustful, prone to errors and more noisy;

A fuzzy logic supervisor (FLS) was used to decide the reliability of the sensor and such reliability is changed by changing matrices R_n. In figure 4.33 it is represented the sensor fusion scheme using both FLS and DIF.

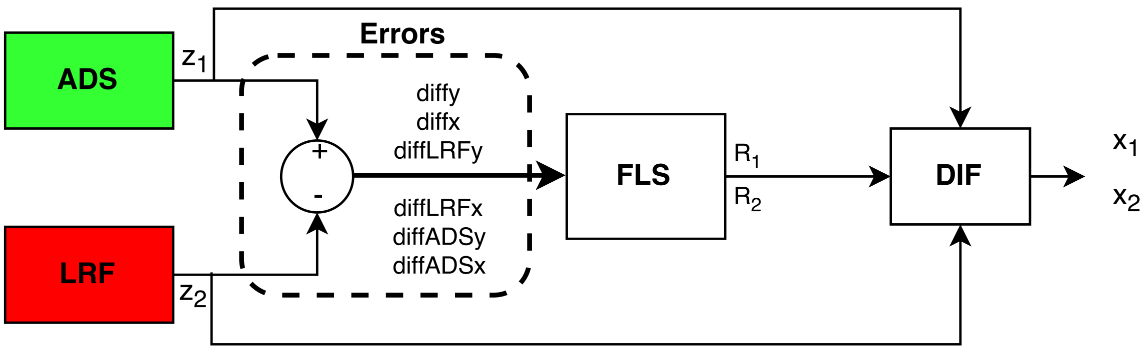


Figure 4.33: Sensor fusion with FLS (Fuzzy logic supervisor) and DIF (Decentralized Information Filter).

The aforementioned errors are detected, for each LWL side. The appearence of those errors makes the signals of each sensor to diverge from each other. Thus, to detect such divergences it will be calculated: absolute difference between y-direction signals from both sensors (*diffy*);

Table 4.10: Fuzzy logic rules set for each leg: underlined to σ_{j1} ; not underlined to σ_{j2}, $j = x, y$ directions.

Inputs		*diffLRFj*		*diffADSj*	
		AC	NAC	AC	NAC
diffj	AC	$\underline{\text{No}}$	$\underline{\text{AN}}$	No	AN
	NAC	$\underline{\text{No}}$	$\underline{\text{AN}}$	No	AN

Table 4.11: Membership functions.

Input	**Membership Functions**	**Function type**	**Range [a,c]**
diffy	Acceptable (AC), NotAcceptable (NAC)	Sigmoid ($f(x,a,c) = \frac{1}{1+e^{-a(x-c)}}$)	AC: [32.62 64.76], NAC: [34.92 151]
diffx			AC: [40.8 89.81], NAC: [57.8 141.7]
diffLRFy			AC: [23.1 80.56], NAC: [22.4 189.8]
diffLRFx			AC: [19.9 54.5], NAC: [27 71.16]
diffADSy			AC: [18.1 73.15], NAC: [22.4 132.4]
diffADSx			AC: [16.5 44.97], NAC: [21.1 77.78]
Output	**Membership Functions**	**Function type**	**Range [a,c]**
σ_x^2	Normal (N), Abnormal (AN)	Sigmoid ($f(x,a,c) = \frac{1}{1+e^{-a(x-c)}}$)	N: [116 147.9], AN: [120 352]
σ_y^2			N: [67.9 133], AN: [93.3 162.1]
$\sigma_x'^2$			N: [8.7 20], AN: [15, 150]
$\sigma_y'^2$			N: [60 100], AN: [90, 150]

absolute difference between x-direction signals from both sensors (*diffx*); absolute difference between samples in each sensor and direction (*diffLRFy*, *diffLRFx*, *diffADSy*, *diffADSx*). Thus, these inputs will detect abnormal situations, changing the matrices R_1 and R_2. They are classified as Acceptable (AC) and NotAcceptable (NAC). Then, through a rule set presented in table 4.10, R_n matrices will be classified as Normal (No) or Abnormal (AN). Each rule receives a corresponding normalized membership value from the input membership functions (Table 4.11). From the rules' result for a given instant, the output membership functions (Table 4.11) are chosen. The membership value was used to find the crisp value of the outputs. These membership values were defined based on several tests made with the available signals samples (trial and error). The output represents the decision made as to whether ADS or LRF is given higher priority or if both should have a change on its priority. Defuzzification was done using the center of gravity method. The crisp value of the decision is taken as the measurement noise covariance values for each matrix R_n. It is noterworthy that the aforementionated procedure is done for each lower limb.

4.4.1.5 Results and Discussion

In order to evaluate the performance of DIF with fuzzy logic supervisor, 14 different walker users (10 elders with knee osteoarthrosis and 4 young ataxic patients) were asked to walk in a straight-forward trajectory and perform a curve. Different situations were identified, and for

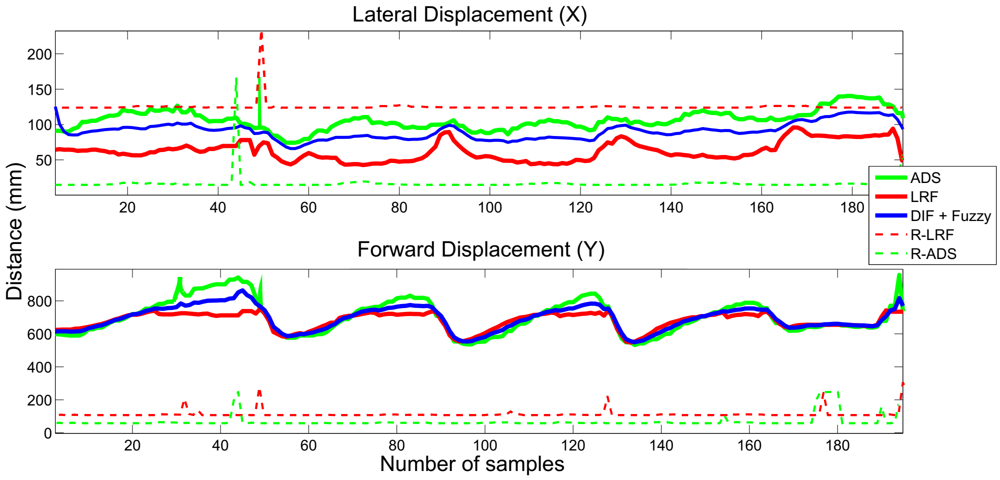


Figure 4.34: ADS, LRF and DIF signals with the adapted R.

simplification reasons, only one LWL signal will be shown. Signals were acquired at 10Hz of sample rate.

In figure 4.34 it is shown one example of the result obtained with DIF and fuzzy logic supervisior. It can be seen the values of matrices R_1*(R-LRF)* and R_2*(R-ADS)* change along the samples accordingly with the behaviour of the signals. For example, between sample number 30 to sample number 50, ADS fails twice, increasing R_1*(R-LRF),* in order to give more weight to LRF signal. In the both times this happens because the feet are too close from each other, causing a disturbance in ADS on the y-direction signal of the foot. In this same interval. LRF fails once, presenting the same value during some samples and not following the same behavior as ADS. This triggers the increase of R_2*(R-ADS).* When the errors disapear, R_1*(R-LRF)* and R_2*(R-ADS)* are reset to its "normal" values.

In figure 4.35a, a full trajectory of an elder subject is presented.

It can be observed that the suitable sensor is chosen for each signal by the supervisor, divergence does not occur, and large variations are dampened. In this case, the calibration of LRF was not perfectly done, causing the LRF algorithm often to fail (e.g. around sample nº 100 and nº 320). Comparing DIF with and without supervisor fuzzy, it is possible to verify that some information is lost without the supervisor.

Other situation that occurred is shown in figure 4.35b with some steps of an ataxic patient. Since this patient presented high step lengths, he often closed his base and his feet approximated too much from ADS, causing the system to fail. However, DIF with supervisor was able to damp such errors.

In section 4.3, spatiotemporal parameters were calculated through LWL positions. In order to verify if the presented data fusion algorithm is suitable for such purpose, pre-defined

Table 4.12: Errors for each system expressed in *mm*.

	ADS	LRF	DIF	DIF + Fuzzy	[124]
Step Length	28.50±16.51	20.2±6.18	22.41±4.96	21.30±4.51	33.6±22.00
Step Width	22.46±6.82	20.7±5.85	20.10±3.01	15.57±3.63	25.5±20.60

step lengths' and step widths' (marked on the floor) were performed and calculated with the three systems: LRF, ADS and DIF with supervisor. Table 4.12 shows the error values, demonstrating that DIF+ fuzzy presents the smallest error as well as the smallest variation of the variables' values. It is also shown the result of an approach presented by Hu et al.[124], that used a parametric model of legs and feet adapted to camera images using point clouds. By observing the obtained values for each parameter, it can be concluded that the approach DIF+Fuzzy has better results, with decreased error.

Thus, such method is promising in terms of improving the gait measurement error.

4.4.1.6 Conclusion

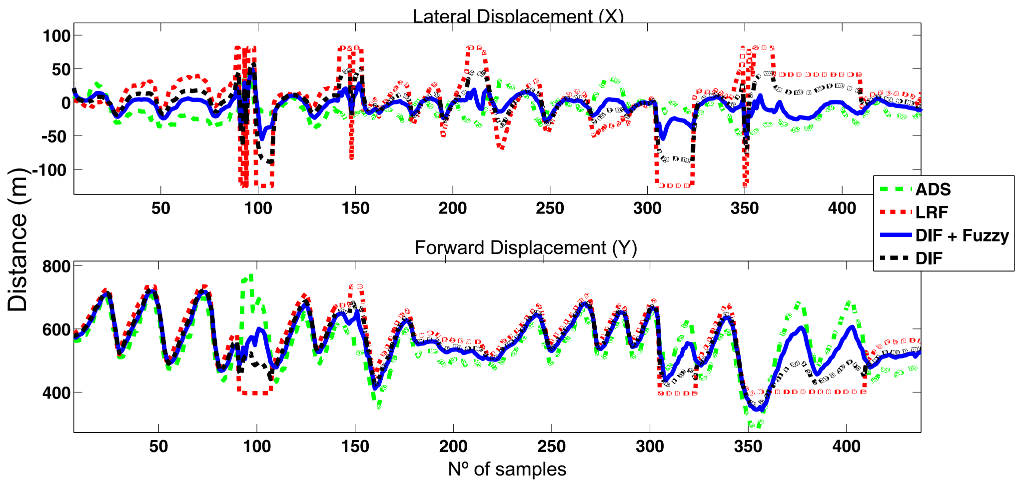
This subsection described a motion capture device integrated on a rollator walker type. 2D motion of lower limbs is obtained by fusing data of LRF and ADS sensors through an information filter supervised by a fuzzy logic supervisor that adapts the measurement error covariance matrix. The performance of the system in gait analysis is quantified through the calculation of step length and width and compared with other work. It is possible to see the error decrease associated to the fusion algorithm, demonstrating that such method has a great potential to be used. However, more experiments with different patients and situations are required to better infer the system performance.

4.5 Assessment of Posture Stability and Fall Risk using one Accelerometer

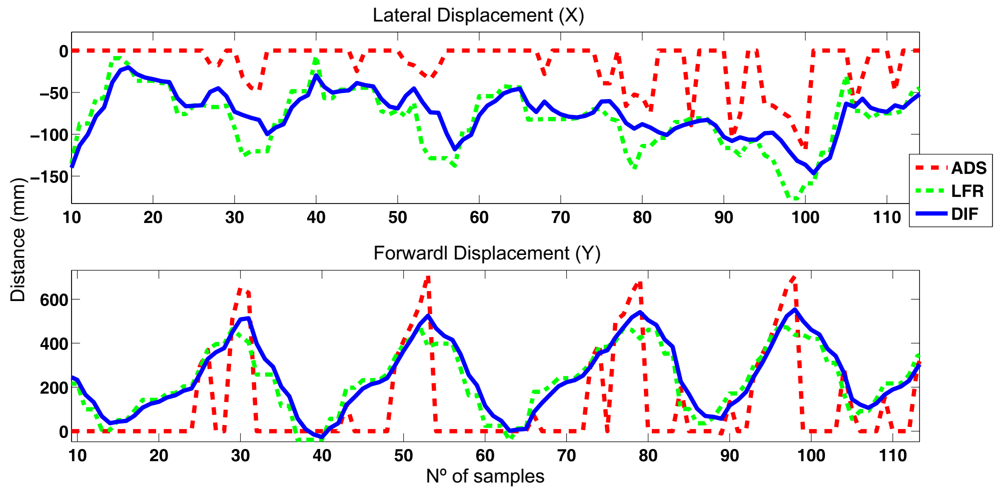
An important assessment to perform in walker assisted gait is the posture stability and risk of fall. Since the common problem of walker users is usually lack of balance, such assessment is fundamental through their recovery.

To assess posture stability and risk of fall, an accelerometer, for example, may be located near the center of mass (COM) of the subject since it is the best place to evaluate with accuracy such outcomes [153].

In order to evaluate the important recovery outcomes with the use of a walker device, the aforementioned assessment, was taken into account. Thus, a research about the different types



(a)



(b)

Figure 4.35: (a) Elder performing a U-shape trajectory with walker. ADS, LRF, DIF+Fuzzy and DIF are compared. (b) Four strides of an ataxic patient. ADS, LRF and DIF+Fuzzy are compared.

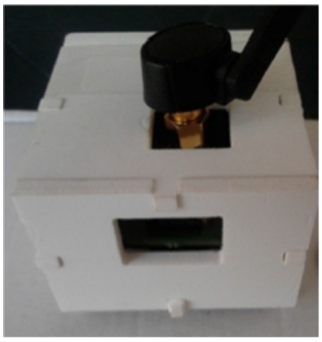


Figure 4.36: Inertial Sensor (SMI, InvenSense MP6000) encapsulated in a white box (for protection).

of algorithms that evaluate the COM displacement in order to assess the posture stability and risk of fall was performed and it is presented on the following subsections.

As a result of this work, two scientific contributions were published [23, 24].

4.5.1 Accelerometer system

In this study, it was used one inertial sensor (SMI, InvenSense MP6000) that includes one gyroscope and one accelerometer, both triaxial. However, on this application only the accelerometer will be used, since it is engough for this study. These sensors require a computer and a base station (Texas Instrument CC2530) for acquiring signals. Each inertial sensor includes a SD memory card (Secure Digital) which stores data. In figure 4.36 the inertial sensor used in this project is presented. For more information on the specifications and calibration method consult [160].

4.5.2 Brief Review of the Assessment of Posture Stability and Risk of Fall Methods with Accelerometers

The use of accelerometers for the evaluation of posture stability and risk of fall has been shown to be accurate, being considered the one of the best devices for evaluation of human posture, surpassing the force platforms [161]. In this subsection some of the studies about the use of accelerometers for the evaluation of posture stability and risk of fall will be presented.

In 2004, Bonnet et al. [162] sought to evaluate in a quantitative manner the static and dynamic balance, in order to develop strategies for regulating the posture. To this end, they used only one inertial sensor, which includes accelerometer and magnetometer (Maxicube integrated with 6 sensors). After performing the sensor calibration, developed a rotation matrix, which defines the orientation of the anatomical axes. The sensor is located in the trunk at the level of the sternum in order to estimate the 3D orientation of the trunk by applying an efficient algorithm. However, such algorithm was considered to be a very complex approach.

Gietzelt et al. [163] developed an automatic and objective method to determine the risk of falling of patients. For this purpose, they used a triaxial accelerometer placed at hip level close to the center of mass. They developed a machine learning algorithm to classify models. At the end, they concluded that the presented method allows to distinguish between those with high or low risk of falling. This study showed to be inconvenient since it requires the use of thresholds as a discriminative factor for the different levels of risk of falling, which are specific to the studied participants.

In 2011, Ishigaki et al. [164], via a position monitoring system, intended to measure the pelvic movement in the elderly and to identify characteristics associated with an unsteady gait. The monitoring system located at the hip, incorporates an accelerometer and gyroscope, both triaxial, enabling the measurement of angles, angular velocity and acceleration of the pelvic movement. They concluded that the pelvic movement characteristics during the gait associated with unstable elderly can be identified, which is the main risk factor associated with falls.

Also in 2011, Rigoberto [161] developed a method with the goal of measuring postural balance through acceleration signals, describing the processing of these signals and calculation of sixteen parameters for comparison between young and elderly. Additionally, he intended to determine the presence of possible undesirable sources of variability in these signals and evaluated the sensitivity of the various parameters calculated in order to detect changes in the postural stability with advancing age. For this study, he used an inertial sensor (accelerometer and gyroscope triaxial). Regarding the parameters, the set included RMS, jerk, circular area of trust, among others. He concluded that the acceleration signals are actually effective for the detection of differences between healthy elderly and young, even better than with force platforms [161].

Doheny et al. [165] sought to identify what parameters and what better tests distinguishes participants (elderlies), considering its fall risk condition. More specifically investigated the usefulness of range of acceleration amplitudes and COM displacement in the identification of risk of falling. For this purpose, they used one triaxial accelerometer at the level of vertebra L4. The calculated parameters were the ranges of acceleration towards anterior-posterior (AP) and medial-lateral (ML) directions, sway length at AP, ML and horizontal directions, RMS at AP, ML and Horizontal directions and finally COM displacement.They concluded that the calculated parameters can identify people with high risk of falling [165].

Greene et al. [166] aimed to develop a model able to accurately classify people with greater or lesser tendency to fall using multiple sensors and then compare their results with the Berg Balance Scale. The sensors were intended to quantitatively determine the equilibrium. For the tests, they used a platform with pressure sensors, an inertial sensor (placed at the level of vertebra L3) and a video camera. Through the obtained signals with the inertial sensor it was obtained the RMS amplitude of acceleration signals in the direction AP and ML, to quantify the postural balance in each direction, the frequency domain of the variability of the acceleration and angular velocity signals using Spectral Edge Frequency (SEF) and also the calculation of spectral entropy. Through the force platform it was determined the center of pressure range of motion, including the average distance, RMS, sway length, among others. They came to the conclusion that both through the use of force platform, as the inertial sensor it is possible to distinguish people with higher and lower tendency to fall.

In 2013, Riva et al. [167] studied the relationship between certain parameters and the risk of falling in a large sample of people aged over 50 years. Participants had to use an inertial sensor located below the shoulder blade. With this test, they calculated the harmony rate, harmonicity index, multiscale entropy (MSE), recurrence quantification analysis (RQA) of trunk accelerations. For the determination of these parameters the authors applied the discrete Fourier transform, power spectral density and coarse-grained time series. The authors concluded that the MSE and RQA parameters are positively related to fall history, and can be used as tools for the prevention of falls.

Finally, in general all mentioned authors refer to the inertial sensor as a suitable tool for achieving an objective and quantitative assessment. This sensor enables the discrimination of people with major and minor risk of falling, determination of pelvic motion characteristics associated with posture stability and consequently the risk of fall, static and dynamic balance, relation between variability with the risk of fall, among others.

4.5.3 Algorithm for Assessment of Posture Stability and Risk of Fall

To assess posture stability and risk of fall, it was found as obstacles in the literature the high cost of equipment, complexity on some algorithms and the use of specific thresholds to study subjects. It was noted that the pelvic movement during unstable gait is the main risk factor for falls [164], hence the relevance of the assessment of this movement.

Since the objective of this work is to evaluate the stability during walker assisted gait, the previous studies were analyzed. It is intended to implement a simple and low cost system, with minimum included sensors. Among all, it was found that Doheny et al. [165] study was the most suitable. The large sample of subjects (101 elderly), the various test types, the parameters and objectives presented, led to conclude that the assessment of posture stability and risk of fall of this work will be based on the study conducted by Doheny et al. [165]. In this study, the inertial sensor is placed on the trunk, located at hip level, vertebra L4. Thus, based on the processing of data and parameters calculated in this article is intended to extrapolate the findings obtained for this project test conditions. In order to compare the results obtained in [165], the team conducted two studies presented in [23, 24]. With these studies it was allowed to conclude that the implemented system is effective for our purpose. Thus, such a system will be implemented in chapter 5 with patientes recovering from Total Knee Arthroplasty.

In the following, a brief presentation of the algorithm is presented. In chapter 5, the acquisition protocol will be then presented in detail.

As it was mentioned above, the development of an algorithm for the evaluation of posture stability and risk of fall based in [165]. The first phase of processing is the acceleration signals filtering with a Butterworth band-pass filter, 5^{th}order, between 0.1-10 Hz and a sample frequency of 113Hz, indicated by [160]. The cut frequency selected by [165] was chosen since the trunk movement is associated with low frequencies [168]. Then, it is advised to remove the first and last samples of data, since they correspond to the adaptation phase to the movement/position and to the slowdown/fatigue phase, respectively.

The first parameter to be calculated is the cumulative horizontal acceleration, *AccCOM*, using the medial-lateral (ML) and anterior-posterior (AP) acceleration vectors (Figure 4.37) using:

$$AccCOM = \sqrt{AccML^2 + AccAP^2} \tag{4.55}$$

Then, being one of the main goals the calculation of the COM displacement (D), it is necessary to twice integrate the acceleration signal, using a trapezoidal method. The associated error with these integrations (low frequency drifts) is reduced by using a second-order polynomial fit and subtracting the mean amplitude of the signal before and after each integration procedure [169]. In addition, to reduce the error derivated by the integration, the signals were then high pass filtered with Butterworth, 5^{th}order, at 0.1 Hz. Then, cumulative horizontal displacement, *DCOM*, was calculated:

$$DCOM = \sqrt{DML^2 + DAP^2} \tag{4.56}$$

Displacement range in vertical (V), ML and AP direction , *ROMV*, *ROMAP* and *ROMML*, respectively, were also calculated.

Hereupon, the Root Mean Square (RMS) of the accelerations in V, AP, ML and horizontal (HOR) directions, *RMSV, RMSAP, RMSML* and *RMSHOR*, were also calculated. These latter parameters are a dispersion measure of the acceleration relatively to zero [170], allowing the quantification of the postural sway in each direction [156].

Figure 4.37: Representation of the accelerometer axis: x: Medial-Lateral, y: Vertical, z: Anterior-Posterior).

Additionally, V, ML, AP and cumulative horizontal sway length (SL) were calculated:

$$SLx = \sum |\Delta Dx|, \ldots x = V, ML, AP, HOR \tag{4.57}$$

These parameters enable to assess the patient's risk of fall. The behavior of these values will state if the risk of fall is high or not and the higher the values, the higher is the risk of fall [165, 171].

4.6 User Security State Estimation based on multi-sensor configuration

The ASBGo project has as one of its main goals to ensure user safety while walking with the walker. This section will present the sensory system designed to evaluate the safety and detection of different states of a person during assisted gait with ASBGo. It is intended to extract patterns and behaviors of the user along his/her gait, characterizing them in 4 different states regarding the intended movement, the weight applied on the walker, forward and backward falls and distance between the user and the walker. For this, a sensory system able to evaluate the various states was developed and comprises in force sensors, infrared sensors and potentiometers. Finally, a state machine that acquires and processes in real time the signals from the mentioned sensors to an evaluation of the user behavior was designed. Depending on the detected state, the walker will perform a different action. Therefore, each state will provide enough information so that the waker can make a decision.

The detection of the intention to move on or turn determines the course of the walker. Information regarding the weight applied to the walker reports the need to adjust better the device to the user or problems regarding the support of the device or subject's posture. Then, if the situation of falling forward is detected, the walker can support the fall, by stopping. However, if a backward fall is detected, this will only suit as an alarm information to the therapist because there is no back security to prevent the person from falling backward. Nevertheless, when detecting this state, the walker automatically stops. If the distance from the user to the walker is reduced, the walker must accelerate and when the distance increases, it should slow down in order to establish the normal distance of the person to the device.

In general, the states' detection allows to adjust and adapt the behavior of the walker to different needs of the subject during therapy.

4.6.1 Multi-sensor system

Different equipments were included on this approach:

- Arduino platform (Arduino Mega 2560, Interface Processing and Software);
- 2 force sensors on the forearm supports (Chapter 3);
- 1 infrared sensor on the walker center (Chapter 3);
- 2 potenciometers (linear e rotary) (Chapter 3).

For a correct identification of the states considered in the study, the signals from the sensors must be properly acquired, processed and analyzed. To this end, some considerations should be weighed for a careful and strict examination, such as the implementation of protection circuitry for sensors, signal filtering, signal fluctuations, among others. Thus, the complementarity between the implementation of both hardware and software is essential in order to achieve the expected results in a more efficient and satisfactory manner.

4.6.2 Definition and Implementation of States

The definition of states began with an evaluation of different conditions with the walker by acquiring signals from the different sensors: rotary potentiomenter (PotRot), 2 force sensors (Force) and infrared sensor (IR). A fuzzy logic supervisor [159] will be thus implemented. For

Table 4.13: Membership functions of fuzzy logic.

Input	Membership functions	Function Type	Range
PotRot	Stop	Sigmoid ($f(x,a,c) = \frac{1}{1+e^{-a(x-c)}}$)	[3.841 145.6]
	Move		[111 235.8]
Force (2)	Zero		[2.67 67.86]
	Heavy		[563.4 656.4]
	Normal	Gaussian ($f(x,\sigma,c) = e^{\frac{-(x-c)^2}{2\sigma^2}}$)	[94.1 326.1]
IR	TooClose	Sigmoid ($f(x,a,c) = \frac{1}{1+e^{-a(x-c)}}$)	[267 316.9]
	TooFar		[38.2 90.48]
	Close	Gaussian ($f(x,\sigma,c) = e^{\frac{-(x-c)^2}{2\sigma^2}}$)	[15.9 231.5]
	Far		[8.435 107]
	Normal		[10.1 160.6]
Output	Membership functions	Function Type	Range
States	SlowDown	Peaks	2
	Alarm		3
	Normal		4
	Accelerate		5
	Stop		6

each sensor input different membership functions were defined accordingly with the different conditions that it is intended to evaluate. Such membership functions are defined in table 4.13. These are related to:

– Greater or lesser proximity of the user to the walker, which can be too close or too far meaning that the subject is in a dangerous situation;

– Intention to stop or move;

– Load on the walker and if it is to heavy or null may mean that the subject is in a dangerous situation.

To define the range of values for each membership function, 10 different young subjects were asked to simulate these different conditions. After analyzing the different signals and the corresponding values for each condition, an average of values was performed and the final values are presented in table 4.13.

After defining the different conditions for each sensor, the output states were set as shown in table 4.13. The 'If/and/then" rules to define which is the current state are shown in table 4.14.

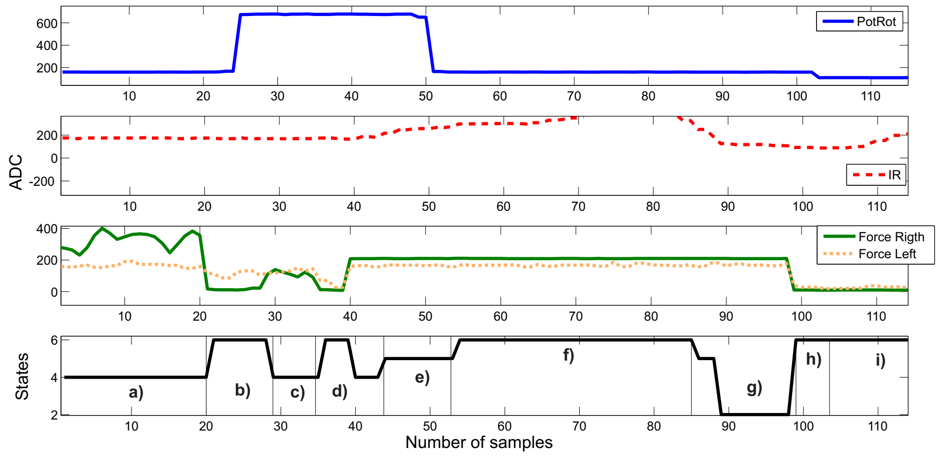
Accordingly with the output state an actuation can be associated as shown in table 4.15.

Table 4.14: Rules definition.

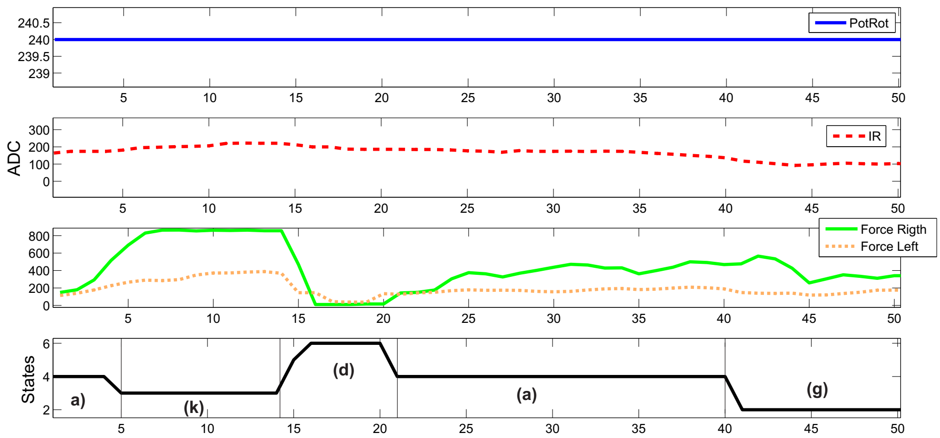
<table>
<tr><td rowspan="9">If</td><td>PotRot</td><td rowspan="9">and</td><td>Force</td><td rowspan="9">and</td><td>IR</td><td rowspan="9">then</td><td>State</td></tr>
<tr><td>Stop</td><td>-</td><td>-</td><td>Stop</td></tr>
<tr><td rowspan="7">Move</td><td rowspan="5">Normal</td><td>Normal</td><td>Normal</td></tr>
<tr><td>TooClose</td><td>Stop</td></tr>
<tr><td>TooFar</td><td>Stop</td></tr>
<tr><td>Close</td><td>Accelerate</td></tr>
<tr><td>Far</td><td>Slowdown</td></tr>
<tr><td>Zero</td><td>-</td><td>Stop</td></tr>
<tr><td>Heavy</td><td>-</td><td>Alarm</td></tr>
</table>

Table 4.15: Actuations associated with the defined states.

State	Actuation
Normal	Rotary and Linear Potentiomenters set the velocity of the walker.
SlowDown	Velocity is deacreased 10%.
Accelerate	Velocity is increased 10%.
Stop	Walker is stopped.
Alarm	Light/sound alarm is on.



(a)



(b)

Figure 4.38: Identification of states.

4.6.3 Results

An experiment was performed to verify if the states were being correctly identified: a) Start to move normally; b) Leave one forearm support; c) Move normally; d) Leave both forearm supports; e) Close to the walker; f) Too close to the walker; g) Far from the walker; h) Too far from the walker; i) Stop moving; k) heavy load in the forearm support. By looking at figure 4.38a and table 4.16, it is possible to verify that the states were correctly identified. Situations (a) and (c) are associated with state Normal (4), situations b), d), f), h) and i) are associated with Stop (6), situations e) is associated with Accelerate (5) since the person is close to the walker and situation g) is Slow Down (2) because the person is far from the walker. In figure 4.38b, state Alarme (3) was correctly identified in situation k).

Accordingly to the detected states, the actuations presented in table 4.15 were performed. In figure 4.39 it can be visualized the results of motor actuation regarding the detected states.

Table 4.16: Correspondence between situations of the experiment and states.

Situations	States
(a),(c)	Normal (4)
(b), (d), (f), (h), (i)	Stop (6)
(e)	Accelerate (5)
(g)	Slow Down (2)
(k)	Alarme (3)

It is noteworthy, that the experiment was done in a forward movement, *i.e.* both motors receive the same actuation.

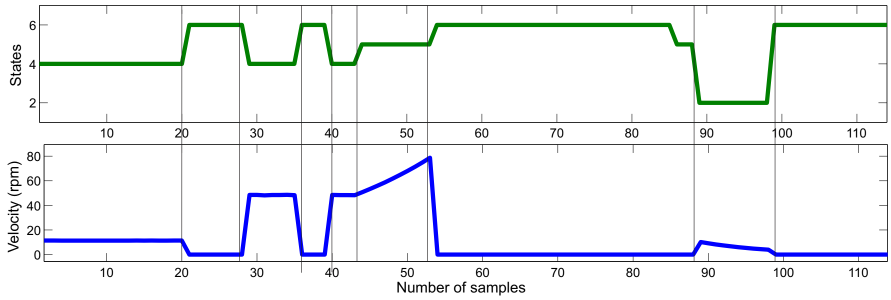


Figure 4.39: Results of motor actuation regarding the detected states.

4.6.4 Conclusions

The definition and implementation of user states were correctly performed, achieving with success the expected results.

With this Fuzzy logic it is possible to give the security functionality to the walker, so that it can act accordingly to the user needs. However, this system has to be validated with real walker users, since it was just tested in a lab environment.

Chapter 5

Feature Reduction and Multi-Classification of Different Assistive Devices

Nowadays, walkers are prescribed without considering an objective criterion, resulting in different interpretations of the disorder and evolution of the patients' treatment. This happens because there is no specific knowledge and understanding about the interaction user-walker, as well as the benefits that rollators with forearm supports (RFS) and other conventional assistive devices (ADs) can bring to the patient's gait in rehabilitation.

To achieve this knowledge, it is necessary to identify which characteristics are more affected by the use of ADs. Findings will lead to the first steps of knowing if the therapy is being or not effective to the patient, and if the prescribed AD is adequate for the patient.

For such identification, a complete processing of gait characteristics data has to be done. However, these data present high-dimensionality which brings a lot of ambiguous and irrelevant information. In order to avoid such information, data reduction methods are crutial to determine which parameters actually contain useful information within a specific clinical context. Then, a classification approach is necessary to distinguish which AD is the most suitable for a giving patients, based on the selected gait characteristics.

Thus, it is proposed in this chapter to develop a study based on data reduction with a multi-classification approach for helping in the decision making of which characteristics are more affected by the use of ADs and which device is more suitable for a specific patient.

To perform this study, two approaches were developed with different methods. In this study, it will be used as ADs, a RFS, crutches and a standard walker.

First, differences and similarities in gait performance between different ADs will be identified. For this, pre-processing will be performed for extracting specific gait features. It will be applied multivariate statistical methods, such as Principal Component Analysis (PCA) and Kernel-PCA (KPCA) [172], which are suited to data reduction and to the search shared relationship among the calculated parameters. The help of the clinicians will be extremely important in the interpretation of these results. The transformation of parameters can determine the features that could be used to quantify differences in gait parameters among the ADs. Such transformation might not only be capable of extracting relevant gait features, but can also provide additional discriminative information for improving classification performance.

This approach will enable us to conclude if different ADs influence differently the gait characteristics of the group of studied patients.

Then, it is intended to select which are the most significant parameters, i.e which exactly parameters are most affected by the use of the RFS and the other ADs. Thus, the work will focus on the type of device that can be more helpful and proper for a certain kind of disorder. It is expected to be given the first steps to construct a model that can be used to help the physicians on deciding which AD can be more adequate to a certain patient based on gait biomechanics. For such selection, it will be investigated different techniques for feature reduction and selection. Among them, F-ratio ranking [173], evolutionary multi-objective optimization techniques [30], backward, forward and stepwise selection [174] and shrinkage methods incorporated on the classifier [175].

In both approaches, in order to distinguish between different ADs, Support Vector Machine (SVM) [176] will be used as classifier.

At the end of this chapter, it will be extracted gait features to be used for interpretation of the clinical benefits of walker with forearms supports and other ADs. In addition, some of these parameters may one day be used to control the movement of the walker and others will be sent to the physician so that he can assess the progress of the patient.

More specifically, the study will be conducted with Knee Osteoarthristis (KOA) patients who had Total Knee Arthroplasty (TKA) surgery.

Osteoarthritis (OA) affects the joints responsible for supporting the body weight, such as the knee, producing restrictions on movement in patients usually over 65 years [177]. Typically, these patients present slow gait and short step length [178]. TKA is a successful surgical procedure to relieve knee pain that results in an improvement in functional capacity of patients. Generally, after 3 days of surgery, patients leave bed, can stand up and start to use the wheelchair. Usually, these patients' recovery is made with the help of crutches. However, this device might not be the correct one for these patients, since some postural and symmetrical problems might not be well corrected and compensated by this device. TKA has been reported to improve the gait patterns of patients with OA but a successful recovery is necessary. Shakoor et al. [179] observed in post-surgical patients gait alterations including gait asymmetry, suggesting that an asymmetric gait pattern may be a subconscious compensation strategy to reduce the load in the operated limb. This may increase loading in the contralateral limb, which can lead to a non-random progression of OA on its knee joint [179]. These assumptions strongly suggest that the contralateral knee is at an elevated risk of OA initiation and progression while the patient is recovering from TKA. Thus, inter-limb analysis is necessary to provide useful clinical and recovery insight as well as identifying gait asymmetries.

For such analysis, spatiotemporal variables, symmetrical indexes and postural control parameters will be acquired and calculated.

The choice of the spatiotemporal parameters relies on the fact that such parameters provide an objective measurement tool and can help in evaluating KOA severity, effectiveness of treatment and might help in disease management [178, 180, 181]. To analyze inter-limb coordination, symmetry indexes need to be calculated through the acquisition of spatiotemporal parameters of each leg. Also, postural control measures are fundamental to monitor the risk of fall, patient stability and balance assessment, which has to be preserved and/or enhanced [165].

Gait of TKA patients is characterized by slow speed, short step length, short single limb support [178] and the consequent increase on the duration of the double support phase. Recovery of these patients is made with the help of crutches. However, this type of AD provides an unnatural gait performance. Thus, it is intended to verify if different ADs influence differently the gait pattern of TKA patients. Findings will help to infer which AD is the best solution for the recovery process. Ultimately, this information may also be used to identify some benefits and limitations of these devices on the rehabilitation of TKA patients and to evaluate the benefit of their use.

This chapter will be divided as follows. First, preliminary work will be presented in order to present the first steps taken by the author in study and implementation of multivariate analysis. Then, since the two approaches have the data acquisition and classification methods in common, such information will be presented first. A brief introduction of the proposed feature reduction techniques, design and methodology, results, discussion and conclusions for each approach will be finally presented.

5.1 Preliminary Work in Multivariate Analysis

In the beginning of the feature extraction and selection research, we performed a preliminar work about multivariate analysis techniques in order to verify the potential of PCA and genetic algorithms with the use of SVM classifier in walker-assisted gait analysis. Through two works, [25] and [30], the first steps were taken in order to gain the knowledge to develop the studies presented in the next sections of this chapter.

In [25], the PCA technique was expanded to provide a comparison of the kinematic and spatiotemporal gait parameters of two groups of subjects: assisted and non-assisted gait. This analysis was performed as a proof of concept, to assess if PCA can be used to identify the main effects in gait performance during assisted gait, when a subject is using the RFS. PCA was used as a data reduction tool, as well as a preliminary step for further analysis to determine differences between assisted and non-assisted gait groups. These further analyses included statistical hypothesis testing to detect group differences, and discriminant analysis to quantify overall group separation. Thus, the goal was to determine the main features of these gait parameters that are related to effects of assisted gait and therefore be used to emphasize a comparison between assisted and non-assisted gait.

Results with healthy young subjects with no locomotion limitations demonstrated the effectiveness of the proposed technique in identifying both the main effects in gait performance during assisted gait and the use or non-use of the RFS. If PCA shows differences in these healthy cases, authors believe that when the technique is applied with actual RFS users (e.g. elderlies), that is, users that can effectively benefit from the RFS usage, these differences in gait parameters will be greater and get a better performance of PCA.

Three principal components (PCs) explained about 63% of the variation in gait measures across different persons without any gait dysfunction, during unassisted and assisted gait with a RFS. The first PC was found to be related to support; the second to speed and posture and the third with balance. These results are very satisfactory since aspects regarding posture, balance and support enhance the rehabilitation potential of the use of RFS. It was also possible to conclude that the main characteristics that distinguish both groups can correctly classify the overall groups with 96.2% of accuracy. The proposed work allows data reduction in gait analysis by investigating, in an objective and statistical way, the relationships between the vast quantities of variables and numerical information. Further, PCA provided useful insight to clinical information necessary and relevant to find some limitations and benefits of RFS.

Other approach was performed by addressing the problem of selecting the most relevant gait features required to differentiate between assisted and non-assisted gait. For that purpose, it was presented in [30] an efficient approach that combines evolutionary techniques, based on genetic algorithms (one objective and multiobjective), and support vector machine algorithms, to discriminate differences between assisted and non-assisted gait with a RFS. For comparison purposes, other classification algorithms were verified (Neural networks and Naives-Bayes). The aim was to restrict the number of features to the smallest subset capable of discriminating normal locomotion from walker-assisted locomotion.

Results with healthy young subjects showed that the main differences are characterized by double support duration and hip joint excursion in the sagittal plane. These results, confirmed by clinical evidence, allowed concluding that this technique is an efficient feature selection approach.

After exploring and performing these studies, a protocol with elderly patients was created to explore the gait patterns' differences that exist during the use of different ADs. Some states of recovery require gait training with a walking aid, that sometimes is not objectively selected and prescribed by the clinician. Thus, this process needs to be studied in detail in order to give more clinical insight to objective decisions regarding ADs.

5.2 Methods and Data acquisition

5.2.1 Participants

A group 13 elderly patients (5 men and 8 women), aged 67.3 ±5.1 years and diagnosed with osteoarthritis in the knee and subjected to TKA were considered. These patients do not have experience using crutches (CRT), standard walker (SW) and rollators with forearm supports (RFS). The inclusion criteria were patients on the third day after TKA, hemodynamically stable that have already lift from the bed, with good cognitive capabilities, presenting appropriate flexion control of some muscles of the hip, wrist, elbow and knee, and Berg Balance Scale (BERG) test score [182] of such patients should be less than 45. The exclusion criteria were cardiac, vascular, respiratory, neurologic or metabolic diseases that affect the gait; neurologic diseases that affects balance; pathologies in the ear and recent clinical history of trauma in the limbs. The study was conducted at Hospital of Braga, approved by the Ethical Committee, and all the patients signed the informed consent. All trials were filmed with a video camera.

5.2.2 Protocol and parameters

In order to assess the effect of the ADs on gait, tests were conducted using three conditions: (1) CRT, (2) SW, and (3) RFS. In these tests, subjects had to walk for approximately 10m, or perform, at least, 15 gait cycles, with the different 3 ADs, along a corridor at the Hospital of Braga. Three walking trials for each subject and condition were realized. Then, the mean and standard deviation were computed for each feature. All trials were filmed by a video camera. Before the realization of the trials, the height of each AD was adjusted for each subject, by a physiotherapist. To measure the spatiotemporal parameters, floor markers were used as well as the video camera. For the postural control features, one tri-axial accelerometer (SMI, MP6000, of InvenSense) was used. This sensor will provide information regarding the movement of body center of mass (COM). It was attached to the lower back at approximately

Figure 5.1: ADs used in this study. Left image: Crutches; center image: standard walker; and right image: RFS ASBGo walker. In the left image it is represented the axis of the sensor, used in all experiments.

the level L4 vertebrae. The used system configuration and the coordinates of reference are shown in figure 5.1. The x-axis, y-axis and z-axis correspond to the medio-lateral (*ML*), vertical (*V*) and anterior-posterior (*AP*) accelerations, respectively.

The spatiotemporal features considered for this study are Gait Cycle (*GC*), Stride Length (*STR*), Step Length (*STP*), Stance and Swing duration (*STAD* and *SWD*), Leg Speed (*SP*), Double Support Duration (*DS*) and Step Time (*STPT*). These features were calculated for both legs, operated (*OL*) and non-operated legs (*NOL*). Thus, each leg is considered to be independent. However, this approach will be compared in terms of redundancy and better gait assessment evaluation with the inter-limb consideration, by calculating symmetry indexes of the parameters [183]. Symmetry indexes (*SI*) were calculated for each feature using:

$$SI = \frac{U_{\mathrm{OL}} - U_{\mathrm{NOL}}}{U_{\mathrm{NOL}}} \tag{5.1}$$

where U_{OL} and U_{NOL} are any aforementioned features for the operated and non-operated leg, respectively. Perfect symmetry results if *SI* is equal to 0., larger Larger positive and negative deviations would indicate a greater symmetry towards the operated or non-operated leg, respectively. In addition, velocity (*Avspd*) and cadence (*CAD*) are calculated.

The postural control parameters consist on a range of measures of COM displacement along with standard measures of accelerometer-derived postural sway, *RMS* acceleration [165]. Thus, the root mean square of *V*, *AP*, horizontal and *ML* acceleration (*RMSV*, *RMSAP*, *RMSHOR* and *RMSML*), horizontal sway length in *V, AP, ML* and horizontal directions (*SLV, SLAP, SLML* and *SLHOR*), range of motion of *V, AP* and *ML* directions (*ROMV, ROMAP* and *ROMML*), and cumulative displacement and acceleration of COM (*DCOM* and *ACCCOM*) were calculated. The data processing of the accelerometer and respective calculation of the presented features is explained in detail at [160] and chapter 4 (section 4.5).

5.3 SVM Classification

SVM was chosen since they are powerful nonlinear classifiers based on statistical learning theory, which have been successfully used in various classification problems [175, 176]. In addition, many classical multi-class and multivariate analysis methods have difficulties in handling such data because of the curse of dimensionality.

SVM was originally designed for binary classification. However, it was successfully applied in learning large feature size with small sample size [184]. There are several methods available to extend binary SVM to multi-class SVM (MSVM) [175]. One common method is to decompose the multi-class problem into multiple binary problems [185], using one-versus-rest or one-versus-one schemes, and to combine the multiple binary rules by a voting method. These approaches are useful in practice but have some limitations [186].

First, the one-versus-rest (MSVM OvR) approach may fail if no class dominates the union of the others. Second, this approach tends to yield unbalanced classification problems, especially if one class is much smaller than the union of the remaining classes. Third, the one-versus-one approach trains each classifier based on only a portion of samples and may increase the solution variability. Fourth, these procedures do not effectively capture the correlation between different classes.

A better method for handling multi-class problems is to separate all the classes by estimating J discriminant functions $(f_1(x), f_2(x), \dots, f_J(x))$ simultaneously (MSVM DF). The decision rule is then defined as $\Phi(X) = \text{argmax}_{j=1}^{J} f_j(X)$, assigning the label r to an input x if $f_r(x)$ gives the highest value. Several generalized loss functions have been proposed for multi-class SVMs, including [175]. Among those available, the loss function used by Crammer and Singer [187] gives a natural extension of the hinge loss from binary to multi-class problems.

Given a training set $\{(x_i, y_i), i = 1, \dots, N\}$, where $x_i \in R^p$ (value of sample i), $y_i \in \{1, 2, \dots, J\}$ (class of sample i) and N corresponds to total number of i samples, the goal of multi-class classification is to learn the optimal decision rule $\Phi : R^p \to \{1, 2, \dots, J\}$ which can accurately predict labels for future observations. For the MSVM, we need to learn multiple discriminant functions, $f(x) = (f_1(x), f_2(x), \dots, f_J(x))$, where $f_j(x)$ represents the strength of evidence that a sample x belongs to class j. The decision rule $\Phi(X) = \text{argmax}_{j=1}^{J} f_j(X)$, and the classification boundary between any two classes j and l is $\{x : f_j(x) = f_l(x)\}$ for $j \neq l$.

In the case of this study, $J = 3$ and the label y is coded as $\{1,2,3\}$, for CRT, SW and RFS, respectively. Consider the linear classification, $f_j(x) = \beta_j 0 + x^T \beta_j$, for $j = 1,2,3$. In the situation of separable case, the discriminating functions are required to satisfy the constraint (eq.5.2) for all observations: if x belongs to class y, then $f_y(x)$ is greater than any $f_j(x)$ with $j \neq y$ by at least margin 1. In the non-separable case, $\xi \geq 0$ is introduced to get the relaxed constraint (eq.5.3).

Separable case:

$$f_{yi}(x_i) - \max_{(j \neq y_i, j=1,..,J)} f_j(x_i) \geq 1 \tag{5.2}$$

Non-separable case:

$$f_{yi}(x_i) - \max_{(j \neq y_i j=1,..,J)} f_j(x_i) \geq 1 - \xi_i \tag{5.3}$$

The MSVM of Crammer and Singer [187] solves:

$$\min_{\beta.\beta_0.\xi} \sum_{i=1}^{N} \xi_i + \lambda \sum_{j=1}^{J} \| \beta_j \|^2 \tag{5.4}$$

subject to $f_{yi}(x_i) - \max_{(j \neq y_i, j=1,..,J)} f_j(x_i) \geq 1 - \xi_i, \xi_i \geq 0, \forall i.$

To avoid estimation redundancy, the constraint $\sum_{j=1}^{J} f_j = 0$ is often invoked. In (eq.5.4), $\sum_{i=1}^{N} \xi_i$ bounds the training error, and $\sum_{j=1}^{J} \|\beta_j\|^2$ controls model complexity. This formulation is a quadratic programming (QP) problem and can be solved by a QP solver. This formulation has a natural interpretation of minimizing a generalizing hinge loss $[1 - \min_{j \neq y} g_j(f(x), y)]$, where $g_j = f_y(x) - f_j(x)$. The generalized function margin of f is defined as the vector $g = (g_1, \ldots, g_{y-1}, g_{y+1}, \ldots, g_J)$.

Crammer and Singer [187] imposed L_2 penalty on the coefficients β in (eq.5.4). The resulting solution utilizes all parameters, which may diminish the prediction accuracy when there are many redundant noise parameters.

The choice of the tuning parameter λ is crucial in the regularization problem, since it controls the trade-off between the training error and generalization performance of classifiers. To select the optimal λ a five-fold cross validation is made. A fine grid search is conducted over a wide range of values of λ, and the best λ is defined as the one which gives the least cross validation error.

5.3.1 Cross-validation of the classifier

A cross-validation method was adopted to evaluate the generalization ability of the classifier. Cross-validation (CV), a powerful technique schemed to capture the more useful information that is adopted in the training process, is a standard method to evaluate the generalization ability of the classification model using different combinations of the testing and training data sets [188]. Here, the CV technique was adopted to eliminate any dependence on the selection of gait features for test set, allowing testing each gait feature after the procedure was finished.

It is important to select subsets of the data to be used as training and test in the classification stage. In this study, a SIXfold CV resampling approach is used to construct the learning and test sets for the MSVM classifier. Initially, the three-group samples (3 ADs) are randomly divided into six non-overlapping subsets of roughly equal size, respectively. A random combination of the subsets for the three groups constitutes a test set (6 sets) and the total remaining subsets are used as the learning set (6 sets). Thus, the SIXfold CV resampling produces a total of 36 pairs (6x6 combinations) of learning and test sets. Each individual of the group of patients is evaluated over the 36 pairs, *i.e.* SVM is executed 36 times, and then is calculated an average of these 36 results.

5.3.2 Classification Evaluation

Classifier evaluation is one of the fundamental issues in the machine learning and pattern recognition societies, especially when a new classification method is introduced and compared with other possible candidates [189].

Given a classification problem of N samples and J classes, define the two functions $tc, pc : N \to \{1, \ldots, J\}$ indicating for each sample n its true class *tc(n)* and its predicted class *pc(n)*, respectively. The corresponding confusion matrix is the square matrix C whose iw-th entry C_{ij} is the number of elements of true class i that have been assigned to class w by the classifier: $C_{iw} = |\{n \varepsilon N : tc(n) = i, pc(s) = j\}|$. Accuracy, ($ACC$), is the most common and simplest measure to evaluate a classifier. It is just defined as the degree of right predictions of a model:

$$\text{ACC} = \frac{\sum_{k=1}^{J} C_{kk}}{N} \tag{5.5}$$

An ACC of 100% means that the measured values are exactly the same as the given values.

The Confusion Entropy measure, CEN, was introduced for evaluating classifiers in the multi-class case. By exploiting the misclassification information of confusion matrices, the measure takes into consideration both the classification accuracy and class discrimination power of classifiers. Besides this, CEN also tries to measure if samples are classified into true classes with high probabilities and into other classes.

CEN for a confusion matrix C is defined in [189] as :

$$\mathrm{CEN} = \sum_{w=1}^{J} P_w \sum_{k=1,k\neq w}^{J} (P_{wk}^{w} \log_{2(J-1)} P_{wk}^{w} + P_{kw}^{w} \log_{2(J-1)} P_{kw}^{w}) \tag{5.6}$$

Where the misclassification probabilities P are defined as the following ratios:

$$P_{iw}^{w} = \frac{c_{iw}}{\sum_{k=1}^{J}(c_{wk}+c_{kw})}, P_{ii}^{i} = 0, P_{iw}^{i} = \frac{c_{iw}}{\sum_{k=1}^{J}(c_{ik}+c_{ki})}, P_w = \frac{\sum_{k=1}^{J}(c_{wk}+c_{kw})}{2\sum_{k,l=1}^{J} c_{kl}} \tag{5.7}$$

This measure ranges between 0 (perfect classification) and 1 for the extreme misclassification case.

Matthews Correlation Coefficient (*MCC*) is a metric used in machine learning as a measure of the quality of binary classifications. This measure is generally regarded as a balanced measure, which can be used even if the classes are of very different sizes [189]. In terms of the confusion matrix, the above equation can be written as presented:

$$\mathrm{MCC} = \frac{\sum_{k,l,m=1}^{J} c_{kk}c_{ml} - c_{lk}c_{km}}{\sqrt{\sum_{k=1}^{J}(\sum_{l=1}^{J} c_{lk})(\sum_{f,g=1,f\neq k}^{J} c_{gf})}\sqrt{\sum_{k=1}^{J}(\sum_{l=1}^{J} c_{kl})(\sum_{f,g=1,f\neq k}^{J} c_{fg})}} \tag{5.8}$$

This coefficient can be seen as a correlation coefficient between the observed and predicted binary classifications. It outputs a value between −1 and +1. A coefficient of +1 represents a perfect prediction, 0 no better than random prediction and −1 indicates total disagreement between prediction and observation. These three metrics will be used on this study as metrics for the multi-class SVM performance.

5.4 Feature Reduction and Extraction Approach

5.4.1 Proposed Approach

In this section, a gait analysis technique that aims to identify differences and similarities in gait performance between three different ADs is proposed.

In [168, 190, 191], it was shown that gait analysis yields redundant information that is often difficult to interpret and it is not always clear what measurements or which analysis are the most appropriate for a particular purpose.

A multivariate analysis avoids the bias introduced by a particular perspective, because it examines the shared relationship among all the measured parameters. Therefore, the authors propose to apply two pre-processing techniques: Linear Principal Component Analysis (PCA) and Kernel-PCA (KPCA) [168, 190–193] to provide a comparison of the spatiotemporal, symmetrical indexes and postural control parameters between the three different ADs. The transformation of parameters can determine the features that could be used to quantify differences in gait parameters among the ADs. Such transformation might not only be capable of extracting good gait features, but can also provide additional discriminative information for improving classification performance.

Linear PCA yields a set of orthogonal bases that captures the directions of maximum variance in the training data [194]. Consequently, PCA has been successfully applied in the field of gait analysis with the purpose of reducing redundant information. Daffertshofer et al. [195] highlighted the application of PCA in studying coordination and variability in human movement. Olney et al. [190] applied PCA to gait parameters (spatiotemporal, kinematic and kinetic parameters) to reduce redundant information. However, gait parameters usually interact in a complex and nonlinear relationship [196] and linear PCA is based on second-order correlations between data. This motivates the use of a nonlinear analysis technique that can capture the nonlinear structure of gait patterns. This can be achieved by means of a kernel function [197]. Thus, the gait data can, firstly, be mapped into a higher-dimensional feature space through a kernel function. Then, linear PCA methods can be applied to extract gait features in the transformed space corresponding to nonlinear feature extraction in the input space.

These pre-processing techniques can, thus, discern important features and, often, reveal relationships that were unsuspected, thereby, allowing interpretations that would not ordinarily result. Such features may improve a classifier to differentiate between different ADs. Thus, a Multi-class Support Vector Machine (MSVM) with different approaches [175] will be designed to evaluate the potential of PCA and KPCA to extract relevant gait features that can differentiate among the ADs. Furthermore, to examine whether the combination of KPCA or PCA with MSVM produces a superior classification performance, we also compared classification performance of original data with MSVM (*i.e.* without feature extraction from input gait parameters).

Hence, the main goal is to determine the main features of the proposed gait parameters that are more affected in assisted gait and can be used to emphasize the differences of using the proposed ADs. The results will reduce information about motion-assisted analysis, indicating which features are important to be evaluated in order to know which device is best suited to a patient.

5.4.2 PCA analysis

PCA technique is expanded to provide a feature transformation of the original data features of spatiotemporal gait parameters and postural parameters from the three conditions (ADs).

PCA transforms the original data space to an orthogonal set of principal components (PCs). The main idea behind PCs is that the direction of the principle component is the direction of maximum variance in the data [147, 192]. The algorithm first centers the data by subtracting the mean and, then, chooses the direction with largest variance as the principle component. It then looks in the variance that remains and places an axis in the direction that is orthogonal to the first axis and has largest variance. This is performed iteratively and creates orthogonal basis for the dataset. Thus, PCs are extracted to determine features of variation that could be used to quantify differences in gait parameters between conditions. In this particulary approach, the goal is to determine the main features of these gait parameters that are related to effects of each device and, therefore, be used to emphasize a comparison between ADs and enhance the performance of multi-class SVM. PCA is used as a data reduction tool, as well as a preliminary step for further analysis to determine differences based on clinical insights between the three conditions.

PCs (u_i) with the highest eigenvalues (λ_i) represent the vectors with maximum variance in the data set. The eigenvalue problem to be solved is defined as

$$\left(\frac{1}{N} \sum_{j=1}^{N} x_{\text{norm},j} x_{\text{norm},j}^{T} \right) u_i = \lambda_i u_i \;\; \text{with } i = 1, \ldots, M \tag{5.9}$$

with N observations of x_{norm} (features set). The term within the brackets in (eq. 5.9) is the covariance matrix. Original data are mapped on up to a maximum of M PCs. Dimension reduction is achieved by using the coefficients of the first m PCs in classification, with $m<M$. In this application, PCA was applied to an N x M matrix, where N=39 is the number of participants (3x13) and M=29 (without SI features) is the number of features, with prior normalization [192]. The number of selected PCs (m) is calculated by keeping the first PCs that retain the most variation of data, according to two criteria as follows. In order to select PCs, it was used Kaiser's criterion that selects PCs with eigenvalues greater than one.

PCA is an unsupervised technique. Therefore, it is not guaranteed that the projection that maximizes the variance in the data also maximizes the affected representations in the transformed feature space. Furthermore, PCA is a linear technique and does not consider underlying nonlinearities.

Standard statistical techniques can be applied to perform hypothesis tests regarding group differences in the PC scores, and thus in the gait parameters. A Student's t-test for paired data was used to identify statistical differences between ADs in the selected PCs. The level of significance in all statistical tests was set to 5%.

5.4.3 KPCA analysis

Kernel based principle components analysis is a nonlinear PCA created using a kernel. KPCA maps the original inputs into a high dimensional feature space using a kernel method [197]. Its advantages are nonlinearity of eigenvectors and a higher number of eigenvectors. KPCA maps the original data vector x_{norm} (features set) into a feature space F by using a nonlinear map Φ as follows:

$$\Phi : R \to F, x_{\text{norm}} \to X_{\text{norm}}. \tag{5.10}$$

Then, it performs linear PCA in the high-dimensional space F, which corresponds to a nonlinear PCA in the original data space. The covariance matrix C is given by:

$$C = \frac{1}{N} \sum_{j=1}^{N} \Phi(x_{(\text{norm},j)}) \Phi^{T}(x_{(\text{norm},j)}) \tag{5.11}$$

Applying the kernel trick, the eigenvalue problem becomes

$$N\lambda\alpha = K\alpha \ \text{ with } K_i j := \Phi(x_{(\text{norm},i)})^T \Phi(x_{(\text{norm},,j)}) \tag{5.12}$$

where the scalar product of Φ can be substituted by a kernel function $K(x,y)$. In this study, a polynomial kernel, $K_p = (1 + x.y)^d$, with degree d, and a gaussian kernel, $K_g = \exp((-\|x-y\|^2)(2\sigma^2)^{-1})$ are used in KPCA analysis.

KPCA is computationally intensive, takes more processing resources and, consequently, increases the computational time compared to PCA. The reason being that the number of training data points in KPCA is much higher than PCA. Therefore, the number of principle components that need to be estimated is also much larger.

5.4.4 Design/Methodology

In order to understand the proposed methods in this approach, a workflow is presented in figure 5.2. Through patient's gait analysis, different parameters (spatiotemporal, postural control parameters and symmetric indexes) are calculated for the 3 proposed ADs. Then, feature reduction is performed with PCA and KPCA (in a separate analysis) in order to obtain new features that will be used in classification. The set of new features, obtained with PCA/KPCA, is divided into new subsets of features (cross-validation). This is done to evaluate the generalization ability of the classifier. Finally, classification is handled by MSVM, to determine

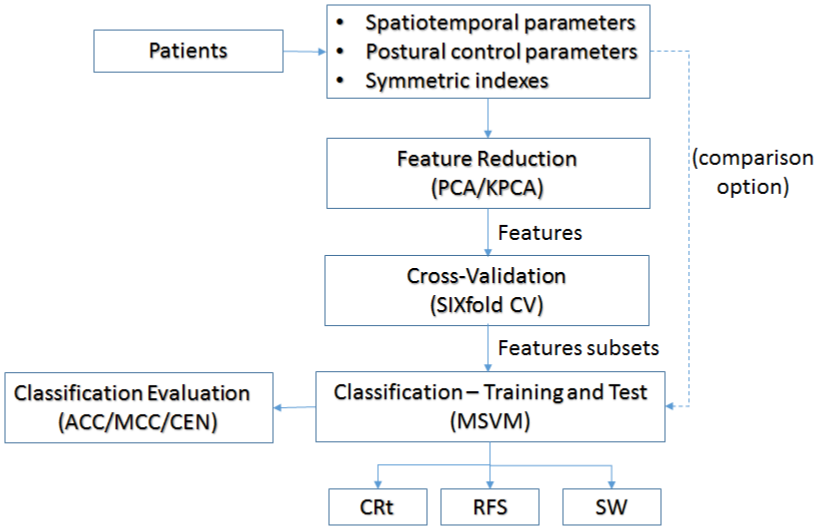


Figure 5.2: Proposed workflow of the porposed feature extration study.

which AD (RFS, CRT and SW) corresponds to the patient's gait pattern, e.g. a subset of features of one subject's gait pattern walking with one device will enter on the classifier, and such classifier has to identify which device corresponds to such gait pattern. This will determine if each AD has a different effect on the gait pattern of patients before TKA.

More specifically in the feature reduction stage, after calculating the gait parameters, KPCA and PCA were performed in order to obtain new features, PCs. All the extracted PCs were normalized by calculating their z-scores with zero-mean and unit variance before training and testing the MSVM [188].

Both approaches of the MSVM classifiers were implemented: MSVM OvR and MSVM DF with L_2 penalty. The latter approach was only implemented with the selected m PCs and compared with MSVM OvR.

Training the MSVM classifier includes initialization of the training set and optimization of parameters. Thus, PCA and KPCA (different kernel functions, i.e. linear, polynomial and gaussian functions) and different parameters (use different degree d of the polynomial kernel and the width σ of the gaussian kernel of the KPCA) were used to compare MSVM performances. Also, the first m PCs, from PCA and KPCA that improve the classifier performance were chosen.

Then, the training procedure of MSVM is described as follows. First m PCs are chosen (based on Kaiser's criterion), constructing an initial training set through PCA and KPCA, and then cross validation method (see section 5.3.1) was used to evaluate the generalization ability of the classifier. The performance of the trained MSVM model was determined by taking an average of each of the proposed three metrics (see section 5.3.2) after testing the

MSVM model. For comparison purpose of the PCA and KPCA performance, the original gait parameters were also used as MSVM input.

5.4.5 Results

Parameters results

Measurements were performed on a total of 13 patients with 3 ADs (CRT, SW and RFS) each. The mean and standard deviation of the gait parameters for each patient were determined using data of a total of 30 gait cycles for each patient. These values are shown in table 5.1 for each AD.

The spatiotemporal and *SI* parameters were calculated and analyzed in detail. As it can be seen in table 5.1, all symmetry indexes, except *SI-STP* (positive asymmetry), did not indicate a great asymmetry between both legs with RFS (mean values are close to zero). Only *SI-STR, SI-STAD* and *SI-SP* showed a symmetrical behavior with CRT. The others showed to be asymmetrical positive. With respect to SW, the parameters *STR*, *GC, STAD, STPT* and *SP* point out a symmetrical behavior. *SI-SW, SI-DS*, and *SI-STP* showed to be asymmetrical positive. The positive asymmetrical parameters indicate that values of these parameters are larger for the *OL* than for the *NOL*. These observations are clearly expressed by the spatiotemporal parameters calculated for each leg. When *SI* is positive, values of the respective spatiotemporal parameter are larger for the *OL* than for the *NOL*. When *SI* is negative, the opposite happens.

In terms of postural control parameters, SW presents decreased *RMS* accelerations, *DCOM, ACCCOM, ROMML* and *ROMV*. RFS presents decreased sway length in all directions and CRT presents decreased *ROMAP*.

Analysis of symmetrical parameters

In order to inspect the importance of symmetrical indexes on the evaluation of TKA patients and to analyze redundancies in relation to the spatiotemporal parameters calculated for each leg, a PCA was applied to the spatial parameters data. A set of 18 parameters, 3 ADs and 13 individuals was considered, corresponding to a matrix with dimensions 39x18. Taking into account the Kaiser's criterion, only 3 principal components (*PC*) were retained. Since with just two PCs, 80% of the variation of the parameters can be explained, these are the ones taken into analysis (Figure 5.3).

Looking at the loadings of *PC1* (5.3a), it is possible to verify that parameters from opposite legs are strongly correlated with each other, since they present similar contributes for each *PC*. The ones that show positive correlation like *SP, SWD, STAD*, *STR* and *GC* are parameters that do not differ from both legs, which means, that they are maybe symmetrical across subjects.

Table 5.1: Mean±standard deviation of each parameter for each device.

Parameters		RFS	CRT	SW
Symmetrical Ind.	SI - STR	0.001±0.01	-0.002±0.02	0.005±0.04
	SI - STP	1.480±3.44	4.881±6.50	14.200±16.02
	SI - GC	0.009±0.03	0.026±0.05	0.018±0.03
	SI - STAD	-0.019±0.08	0.026±0.07	-0.025±0.07
	SI - SWD	0.140±0.20	0.231±0.31	0.373±0.44
	SI - DS	-0.086±0.36	0.889±1.04	2.346±1.32
	SI - STPT	0.136±0.24	0.173±0.72	0.318±1.06
	SI - SP	0.009±0.03	0.014±0.06	0.018±0.05
Postural Parameters	ROMAP (mm)	2.510±2.15	2.470±2.48	2.755±6.90
	ROMML (mm)	0.847±0.57	1.559±1.29	0.810±1.26
	ROMV (mm)	4.947±0.37	4.099±1.55	2.588±0.46
	SLAP (mm)	17.420±11.86	25.339±19.34	19.084±36.48
	SLML (mm)	7.713±4.77	14.677±9.74	9.423±12.92
	SLHOR (mm)	18.870±12.80	29.602±20.97	21.519±38.15
	SLV (mm)	24.750±1.83	26.031±4.73	14.263±2.39
	RMSAP	0.618±0.19	0.798±0.38	0.391±0.19
	RMSML	0.335±0.12	0.392±0.22	0.229±0.13
	RMSV	0.137±0.06	0.169±0.10	0.076±0.09
	RMSHOR	0.704±0.20	0.875±0.40	0.457±0.20
	DCOM(mm)	3.290±1.30	4.07±1.50	1.860±2.10
	ACCCOM (m/s^2)	0.504±0.13	0.614±0.21	0.376±0.20
Spatiotemporal Parameteres	STROL (cm)	51.679±14.32	48.492±10.14	41.179±13.92
	STRNOL (cm)	51.818±14.47	49.232±10.43	41.606±14.28
	STPOL (cm)	31.836±8.39	36.796±7.71	34.140±12.69
	STPNOL (cm)	20.005±9.34	13.729±9.24	8.5102±9.02
	GCOL (s)	2.482±0.70	4.389±1.96	4.071±1.33
	GCNOL (s)	2.479±0.68	4.455±2.06	4.101±1.35
	STADOL (%)	67.937±8.61	79.921±5.40	80.330±7.62
	SwDOL (%)	32.043±8.59	20.078±5.41	19.237±7.73
	STADNOL (%)	70.503±5.84	81.161±7.51	84.330±5.39
	SwDNOL (%)	29.496±5.84	18.838±7.56	15.663±5.39
	DSOL (%)	25.777±6.86	41.473±13.95	51.522±14.50
	DSNOL (%)	31.368±5.98	27.207±8.27	17.998±4.58
	STPTNOL (s)	1.237±0.37	2.213±0.91	2.206±1.24
	STPTOl (s)	1.2532±0.33	2.061±1.07	1.806±1.02
	SPOL (cm/s)	23.034±10.81	13.492±6.18	11.779±6.69
	SPNOL (cm/s)	23.064±11.05	13.629±6.19	11.823±6.95
	CAD (step/min)	50.504±21.42	45.14±13.96	41.42±14.81
	Avspd (m/s)	0.2504±0.13	0.159±0.06	0.143±0.078

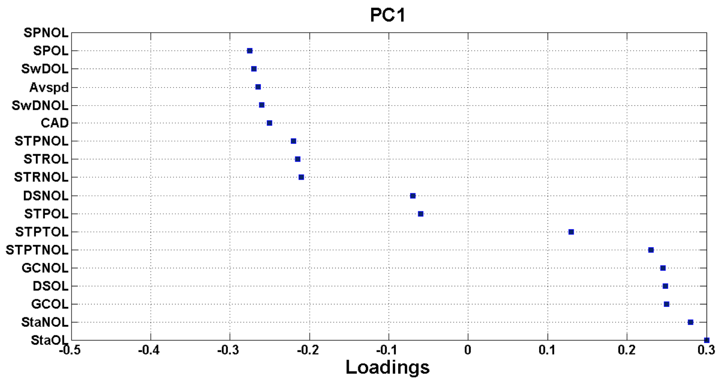
Then, the parameters that present a negative correlation, *STP, STPT* and *DS*, should be the ones with higher asymmetry between legs.

Now, analyzing *PC2* loadings (5.3b), they characterize the effect that *STP, STR* and *STPT* parameters (high positive loading) have in *CAD* (high negative loading). The shorter the steps (*STP*), stride (*STR*) and time step (*STPT*), the greater is the cadence.

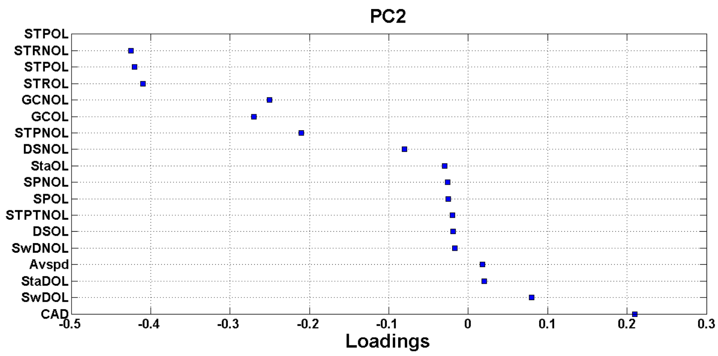
A Student's t-test was performed with *PC1* and *PC2* in order to infer if such PCs were able to distinguish between ADs. PC1 contributes significantly for the difference between ADs ($p<0.05$) and it is related to inter-limb symmetry. However, *PC2* has no significant contribution in distinguishing all the groups. It just distinguishes RFS from the other ADs, since patients tend to walk faster and with greater cadence with RFS. These results led to conclude that the calculation of symmetrical indexes is crucial to distinguish the effect that each AD has on the TKA patients, being important to analyze the inter-limb coordination. Thus, *SI* parameters will be considered for further evaluation.

Experimental results with MSVM

In order to test the effect of the extracted features on classification performance, a search for the optimal number of *PCs* is necessary, since some of these components do not offer all



(a)



(b)

Figure 5.3: a) PC1 and b) PC2 loadings.

Table 5.2: Classification performance (with MSVM OvR) with all dimensions of features (M) extracted from different sets of m gait and posture features.

Feature Extraction with MSVM	Gait Parameters and performance (ACC, MCC,CEN); All dimensions				
	SpTemp (m=16)	SI (m=8)	PCont (m=13)	SpTemp + PCont (m=29)	SI + PCont (m=21)
Original	(0.74,0.43,0.59)	(0.75,0.47,0.53)	(0.77,0.49,0.52)	(0.87,0.81,0.22)	(0.90,0.82,0.16)
PCA	(0.76,0.47,0.53)	(0.77,0.50,0.51)	(0.89,0.84,0.15)	(0.88,0.83,0.20)	(0.91,0.85,0.14)
KPCA(linear)	(0.90,0.82,0.21)	(0.90,0.83,0.15)	(0.91,0.84,0.14)	(0.91,0.84,0.14)	(0.92,0.85,0.13)
KPCA($poly, d = 2$)	(0.91,0.85,0.14)	(0.91,0.85,0.14)	(0.91,0.84,0.14)	(0.91,0.85,0.14)	(0.92,0.86,0.13)
KPCA($poly, d = 3$)	(0.86,0.71,0.27)	(0.91,0.85,0.14)	(0.86,0.69,0.34)	(0.91,0.84,0.15)	(0.91,0.85,0.14)
KPCA($gauss, \sigma = 1000$)	(0.90,0.82,0.19)	(0.91,0.85,0.14)	(0.93,0.87,0.13)	(0.92,0.86,0.14)	(0.94,0.88,0.12)
KPCA($gauss, \sigma = 500$)	(0.91,0.84,0.16)	(0.91,0.85,0.14)	(0.92,0.85,0.14)	(0.92,0.85,0.15)	(0.92,0.85,0.14)
KPCA($gauss, \sigma = 100$)	(0.91,0.84,0.16)	(0.91,0.85,0.14)	(0.92,0.85,0.14)	(0.92,0.85,0.14)	(0.92,0.85,0.13)
KPCA($gauss, \sigma = 50$)	(0.91,0.84,0.16)	(0.91,0.85,0.14)	(0.92,0.85,0.14)	(0.92,0.85,0.14)	(0.92,0.85,0.13)

the necessary information for ADs classification [195]. Thus, a selection of *PCs* was performed. First, the principal component obtained from the maximum eigenvalue was adopted as input for MSVM OvR to perform the classification. The number of principal components was then increased, one by one, according to a descending order of eigenvalues at each step [172]. Thus, the classification performance of MSVM OvR varied with the number of principal components. When the best metric (section 5.3.2) results of MSVM OvR were produced according to the cross-validation method, the optimal number of *PCs* was obtained. This framework was done for different sets of parameters: Spatiotemporal parameters (*SpTemp*), Symmetrical Indexes (*SI*), Postural Control parameters (*PCont*), *SpTemp* with *PCont* and *SI* with *Pcont*.

The classification metrics against all dimensions of features extracted from different sets of parameters, with original data, PCA and different kernels is shown in table 5.2. It shows that *PCs* extracted by using the polynomial ($d = 2$) and gaussian ($\sigma = 1000$) kernels PCA led to significantly better classification performance on the test set than linear PCA and original data.

From table 5.2, it is observed that transforming the original data set of parameters improves the classifier performance. Also, it is observed that when *PCs* extracted from all 16 *SpTemp* with KPCA were used as the inputs of MSVM to perform the classification, the best accuracy was obtained with polynominal ($d = 2$) kernel achieving 91% of *ACC*, 0.85 of *MCC* and 0.14 of *CEN*. However, this was the set of parameters with worse classification comparing with the other sets. *PCont* seems to be the best set of parameters for achieving the greatest classifier performance. The maximum rate was 93% of *ACC*, 0.87 of *MCC* and 0.13 of *CEN*. This result demonstrated that balance assessment data offers more useful information than the *SpTemp* and *SI* data.

However, the sets are combined in two new sets (*SpTemp + PCont and SI + PCont*), the performance of MSVM OvR is further improved compared to using individual gait parameter sets, suggesting that different data types combined could provide more useful information about the different effects that the 3 ADs have on TKA patients. Although there were marked

Table 5.3: MSVM OvR classification results with *SI + PCont* with dimension reduction.

Classifiers	Dimension	SI + PCont (ACC, MCC,CEN)
PCA-MSVM OvR	6	(0.91,0.85,0.14)
KPCA(linear)-MSVM OvR	11	(0.93,0.87,0.12)
KPCA($poly, d = 2$)-MSVM OvR	18	(0.92,0.87,0.13)
KPCA($poly, d = 3$)-MSVM OvR	20	(0.92,0.85,0.14)
KPCA($gauss, \sigma = 1000$)-MSVM OvR	12	(0.94,0.89,0.10)
KPCA($gauss, \sigma = 500$)-MSVM OvR	12	(0.94,0.88,0.11)
KPCA($gauss, \sigma = 100$)-MSVM OvR	14	(0.92,0.86,0.13)
KPCA($gauss, \sigma = 50$)-MSVM OvR	15	(0.92,0.85,0.13)

differences in the classification performance when KPCA was applied to the three sets of parameters, especially between polynominal ($d = 3$) and gaussian ($\sigma = 1000$), the kernels provided almost the same classification performance when all gait parameters were combined into the two sets (92% of *ACC,* 0.85 of *MCC* and 0.13 of *CEN*). However, there is one result that stands out from the others, KPCA with gaussian ($\sigma = 1000$) kernel, obtaining 94% of *ACC*, 0.88 of *MCC* and 0.12 of *CEN*. Therefore, KPCA-based MSVM has the best performance, followed by PCA-based MSVM and then the original MSVM. In addition, the best set of parameters was *SI + PCont*. This result demonstrates that the features extracted by KPCA with *SI + PCont* dataset provide more additional discriminatory information for improving classification performance.

In order to verify the number of *PCs* which influences the classification performance, a dimension reduction was performed by taking from the set of all *PCs*, one *PC* at a time, according to a descending order of eigenvalues, until performance of MSVM was improved (Table 5.3). The first *PC* to be removed was the one that represented less variance of original data, and so on for KPCA-based MSVM, the best classification *ACC* reached was 94%, with 0.89 of *MCC* and 0.1 of *CEN* for kernel gauss ($\sigma = 1000$), while the number of features chosen was 12, as it can be seen in table 5.2. This result was better than the one obtained with all *PC*s dimensions. PCA only obtained 91% of *ACC,* 0.85 of *MCC* and 0.14 of *CEN* with 6 chosen features, however greater than with all dimensions. In sum, these results demonstrate that classification performance largely depended on the selection of the number of features extracted by the KPCA or PCA algorithm, and that additional discriminatory information could be obtained by using KPCA than PCA. In addition, it is clear that KPCA extracts more *PC*s than PCA and it was verified that the type of kernel function of KPCA, regardless of linear, polynominal or gaussian, did not influence the classifier performance much.

Table 5.4 demonstrates a greater improvement of performance using MSVM DF but only with PCA and KPCA with polynominal kernels. Polynominal ($d = 3$) was the kernel that provided the higher performance of MSVM DF obtaining 98% of *ACC*, 0.93 of *MCC* and 0.01 of *CEN*. The gaussian kernel transformation worsened the results.

Table 5.4: MSVM DF classification results with *SI* + *PCont* with dimension reduction.

Classifiers	Dimension	SI + PCont (ACC, MCC,CEN)
PCA-MSVM DF	6	(0.95,0.88,0.21)
KPCA(linear)-MSVM DF	11	(0.82,0.55,0.55)
KPCA($poly, d = 2$)-MSVM DF	18	(0.97,0.91,0.02)
KPCA($poly, d = 3$)-MSVM DF	20	(0.98,0.93,0.01)
KPCA($gauss, \sigma = 1000$)-MSVM DF	12	(0.55,0.06,0.88)
KPCA($gauss, \sigma = 500$)-MSVM DF	12	(0.56,0.01,0.80)
KPCA($gauss, \sigma = 100$)-MSVM DF	14	(0.59,0.11,0.86)
KPCA($gauss, \sigma = 50$)-MSVM DF	15	(0.63,0.19,0.77)

5.4.6 Discussion

This study suggests that KPCA with MSVM is more capable of addressing the gait parameters that better distinguish the effects that different ADs have on TKA patients. As previously mentioned, KPCA is capable of extracting more nonlinear features, which could contain more relevant information about the intrinsic nonlinear behavior of the gait parameters, than PCA.

Gait data analysis is a challenging problem since it is difficult to select which set of parameters can enhance or deteriorate the generalization performance of a machine classifier [196]. Therefore, it is essential to develop appropriate pre-processing techniques that are capable of effectively extracting the relevant gait features.

In this work, two different techniques were addressed: PCA and KPCA, to extract features useful for classification purposes. First, it is important to select which type of gait parameters is more suitable to be pre-processed and then discriminate the different ADs. It has been reported that different types of gait parameters have different impacts on classification performance [188]. In this paper, the same conclusion was observed. As presented in table 5.1, the best performance of classification could not be obtained by using the individual gait parameters set, and the classification performance could be evidently improved when different gait parameters were combined to develop the MSVM model.

This demonstrates that the improved performance of classification largely depended on the more useful gait information that was offered. In particular, symmetrical indexes and postural control parameters obtained the best performance, being better suited to provide useful information about the different gait patterns that TKA patients have with different ADs. Thus, inter-limb analysis is necessary to provide useful clinical and recovery insight. Identifying gait asymmetries is particularly important as abnormal patterns can lead to an increased risk for initiation and progression of the disease. Therefore, asymmetric gait measures may be monitored after surgery to assess treatment outcome and recovery. In addition, postural control measures are fundamental to monitor the risk of fall of the patient, considering stability and balance assessment, which has to be preserved and/or enhanced. This result indicates that it is necessary to employ this type of gait data into the training of a machine classifier with a learning algorithm to automatically classify which AD the person is using, or which AD is more suitable for a patient.

Through table 5.1, RFS demonstrated to be the AD that provides for a more symmetrical gait. CRT demonstrated to be the worst AD. This result may be influenced by the different type of gait that is adopted by the patient when using the ADs. Regarding CRT, the patient learns a three points gait, *i.e.* he always has one foot and two crutches on the ground while moving forward [60]. With SW, the patient first lifts and moves the device, placing it in front of him, and then he moves his legs. Finally with RFS, the movement is continuous by pushing the walker. We believe that the type of gait that the patient presented for each AD influences the distinction of ADs. *SI-STP* is the *SI* feature with higher variability and with high positive values demonstrating that *OL* does not have sufficient strength to endure a large step of the *NOL*. Thus, the step of the *NOL* will be shorter. In addition, *NOL* presents a higher step time (*STPT*) and swing duration (*SWD*) and a lower stance duration (*STAD*). RFS is the only device that presents all symmetrical gait phases (*DS, SWD, STAD* and *GC*). This might indicate that if a person has to recover in terms of time of support of the knee, and create confidence to walk as natural as possible, RFS seems to provide the necessary help.

In general, there are more positive *SI* features that indicate greater asymmetry towards *NOL*. This means that patients put more weight on the non-operated leg throughout the gait cycle. These results are in agreement with those of Talis et al. [198] and Hurwitz et al. [199]. The asymmetry of weight bearing might also depend on small changes in the body configuration. Therefore, the additional stress on *NOL* may develop osteoarthritis in that leg.

Relatively to postural control features, the lower they are the more support the AD is providing. In general, SW is the more stable, and CRT is the worst. SW results in decreased *RMS* acceleration and *ROM* values. On the other hand, RFS provides more stability in sway length and CRT stabilizes the *ROMAP*. These results demonstrate that SW stabilizes the acceleration of the trunk, and its users are more "static" than with the other ADs. RFS stabilizes the position of the trunk, not allowing the user to sway the trunk. CRT is better to give a more erect posture to its users, decreasing the anterior-posterior range of motion, *i.e.* flexion and extension of the trunk.

It is also noteworthy that RFS present a greater velocity and cadence, and SW provides the slowest and more stable gait, as referred in [24].

Besides this, it was primordial to select features (PCs) that could provide additional discrimination information for improving the performance of classification, since some PCs are not suitable to provide the necessary information for AD discrimination. The generalization performance of the MSVM classifier mainly depends on successfully selecting features that are representative of the maximal separation between classes [172]. Thus, the number of features that are chosen can enhance or deteriorate the performance of the classifier. In this study, the results demonstrated that the selection of features was necessary to improve classification. The improvement was low (2-3%), but significant, being selected 12 features by KPCA with gaussian (σ=1000) kernel, which obtained the best result (94% of *ACC*, 0.89 of *MCC* and 0.10 of *CEN*) for MSVM OvR. When greater than 12, additional features, which may be redundant, caused deterioration of overall classification. Similar dependence of classification performance on the selected features has been reported [200].

In terms of KPCA, there is no study that reported which kernel is the best. However, the choice of kernel function has an effect on the performance of the MSVM classifier. In this experiment, when the MSVM with three different kernel functions (linear, polynomial and gaussian kernel) were compared, these kernels performed well and did not affect the classification performance of the SVM. However, gaussian (σ=1000) kernel obtained the best result with and without selection of features.

The classification results of the comparison between MSVM OvR and MSVM DF suggested that the combined classifier shows significantly better performance than MSVM DF when using PCA and gaussian kernel. When using polynomial kernel, the performance is excellent, being obtained 98% of *ACC*, 0.93 of *MCC* and 0.01 of *CEN* for 3rd degree. By this, MSVM DF with 3^{rd} polynomial kernel is the one chosen to classify different ADs gait patterns. This finding suggests that the TKA patients are differently affected by the device. If this happens, the physician should carefully analyze the gait pattern of the patient to infer which AD is better for the patient.

5.4.7 Conclusions

In this study, PCA and KPCA extraction techniques were applied. Symmetrical indexes and postural control parameters were selected as the most suited parameters to provide useful information about the different gait patterns that TKA patients have with different ADs. Also, it has been shown that TKA patients have different gait patterns regarding different ADS. The combination of KPCA and MSVM DF could identify ADs gait patterns with 98% *ACC*, 0.93 of *MCC* and 0.01 of *CEN*, resulting in a markedly improved performance compared to the combination of KPCA and MSVM OVR or PCA and MSVM OVR. Thus, it was demonstrated that with symmetric indexes and postural control parameters, the proposed technique is able to efficiently extract nonlinear gait features for automatic classification of assistive devices gait patterns with high accuracy, and carries considerable potential for future applications in rehabilitation.

The next step, presented in section 5.5, will consist in analyzing each individual subject in detail (in case of identifying differences among the devices), and give a recommendation for a better recovery. Such analysis will be done with the help of a physician.

Thus, this method can be used as pre-prescribing an assistive device and evaluate the outcome of treatment and rehabilitation. More studies should be conducted to evaluate this approach and validate the selected parameters.

The information obtained with the proposed technique could be used to identify benefits and limitations of assistive devices on the rehabilitation process and to evaluate the benefit of their use in TKA patients.

5.5 Feature Reduction and Selection Approach

5.5.1 Proposed Approach

It is intended to understand how gait patterns of TKA patients differ from person to person and how they are influenced by the type of device that is prescribed during their recovery. Such understanding might help in physical therapy for improving the recovery and preventing KOA.

Thus, this approach will focus on the type of device that can be more helpful and proper on the recovery of the TKA patients. It is intended to infer between different assistive devices (crutches, standard walker and rollator with forearm supports) which one presents more capability for providing the required support and correct posture for post-surgical TKA patients. With this study, it is expected to give the first steps to construct a model that can be used to help the physicians on deciding which assistive device can be more adequate to a certain TKA patient based on gait biomechanics.

Symmetric indexes of spatiotemporal parameters and postural control parameters, like in the previous approach, will be calculated and compared between 3 ADs. No spatiotemporal parameters will be used, since in the previous approach it was concluded that symmetric indexes contain more relevant information in terms of clinical context.

From these parameters, the most important and relevant ones will be selected to evaluate the differences between ADs. Since these parameters will not be processed, they will be called as features.

For the selection of features, it will be investigated different techniques for feature reduction and selection based on multivariate analysis.

Thus, to improve the classification it can be used techniques that select important features, since including too many redundant variables in a model may negatively impact its prediction performance [191, 201]. The redundant features include both noise features and features which are highly correlated with the predictor variables. Thus, selection techniques can discard irrelevant and redundant information that may not only affect the classifier's performance, but also describe system's efficiency [202]. However, developing a robust and efficient approach for extracting and select useful information from gait features is a significant and challenging task. Feature selection techniques with flavors of machine learning have been a focus of methodological development in recent years [202]. These techniques are often used to reduce the dimension of features and to avoid the curse of dimension. More importantly, it is to improve the performance of the resulting classifiers and to exclude the interference of a large number of irrelevant features by seeking for the relevant features to differentiate between different groups.

F-ratio ranking [173], evolutionary multi-objective optimization techniques [30, 203], backward, forward and stepwise selection [174] will be investigated and compared. Despite the popularity of these latter techniques, these procedures separate feature selection and classification in two stages, and hence selected features are not guaranteed to contribute significantly to the final classifier. Thus, in addition with these methods, in this approach it will be applied a method that instead of providing an optimal solution based on classification performance, it will impose shrinkage penalties in the learning process of the classifier to enforce solution sparsity, selecting the most relevant features for classification [175].

In order to distinguish between different ADs, the important features will be classified with SVM [176, 201].

The selected features will evaluate which device should be prescribed for TKA patients. Also, in this study, it is taken into account that each patient needs a different help since each case is different from the others, although all are post-TKA patients. Thus, each individual data set of the study will be analyzed with a one-way analysis of variance and confidence intervals.

Other important outcome is to verify if the selected features are related with standard clinical outcomes currently used by the physicians while prescribing the proper assistive devices for the patient's recovery.

5.5.2 F-ratio ranking

F-ratio can be a measure of marginal association between each feature and AD type. The F-ratio is the ratio of the variance between groups to the variance within groups *i.e.* the ratio of the explained variance to the unexplained variance. This metric will be used to rank the features and select the ones that will be used on the multi-classifier SVM. In this paper, F-ratio is calculated between selections of features on the basis of the ratio of their between-classes to within-classes sum of squares [173]. For feature l, the F-ratio:

$$\mathrm{F-ratio}(l) = \frac{\sum_{i=1}^{N}\sum_{j=1}^{J} I(y_i = j)(\bar{x}_l^{(j)} - \bar{x}_l)^2}{\sum_{i=1}^{N}\sum_{j=1}^{J} I(y_i = j)(x_{il} - \bar{x}_l^{(j)})^2} \tag{5.13}$$

Where $y_i \in \{1, 2, \ldots, J\}$ (class of sample i) and N corresponds to total number of i samples. $I(y_i = j)$ is 1 if, for feature l, y_i belongs to class j; $\bar{x}_l^{(j)}$ indicates the average value of feature l for class j samples, and $\bar{x}_l$ is the overall mean value of feature l in the training set.

5.5.3 Evolutionary multi-objective optimization approach using NSGA-II

An evolutionary multi-objective optimization approach based on Non-Domination based Genetic Algorithm (NSGA-II) is designed both for discovering good features subsets and for final feature selection and classification. This will be done by determining the best compromises between the several conflicting objectives (performance metrics). For that purpose, a multi-class SVM is used to ensure the fitness evaluation of each candidate feature subset by classifying them during the successive generations. On previous studies from the authors [30], NSGA-II-SVM approach has demonstrated its higher potential, in comparison with genetic algorithm-SVM, as a powerful tool for mining high dimension data.

NSGA-II starts from a random population of binary individuals (chromosome) representing the subset of features for classification [203]. In order to compare the individuals, the population is sorted based on the domination relation according to several (two or three) conflicting classification performance criteria. Three metrics are adopted as evaluation criteria of the performance of each feature subset: *ACC*, *CEN* and *MCC*. These metrics were explained on section 5.3.2. The maximization of these performance metrics plus the minimization of the number of features (M) allow comparing feature subsets. Consequently, better feature subsets have a greater chance of being selected to form a new subset through crossover and mutation. Crossover combines different features from a pair of subsets into a new subset and mutation changes some of the values (thus adding or deleting features) in a subset randomly. The NSGA-II-MSVM algorithm is an iterative process in which each successive generation is produced by applying genetic operators to the members of the current generation. In this manner, good subsets are "evolved" over time until the stopping criteria are met. The flowchart of the method is presented in figure 5.4:

(1) *Calculate gait features:* read the matrix N x M (N samples and M features) from data set; (2) *Generate parent population P0*: Generate n individuals (parent population) randomly. Each individual is a fixed-length string with M-length of bits of either 1 or 0 (binary-coded).

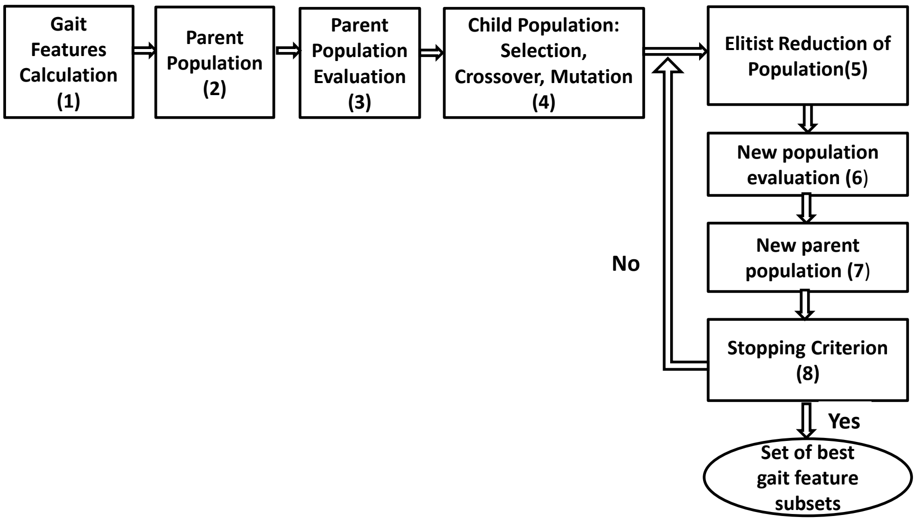


Figure 5.4: Flow chart of the proposed NSGA-II-MSVM combination.

Features are binary-coded within each string as either presence (1) or absence (0); (3) *Parent population evaluation*: (i) Fitness Calculation - each solution in population representing a combination of features is evaluated in terms of the evaluation criteria (one or two performance metrics presented and the number of selected features). The population is sorted according to the domination relation defining several domination fronts; (ii) Crowding - crowding distance is calculated for each individual. The crowding distance is a measure of how close an individual is to its neighbors. Large average crowding distance will result in better diversity in the population. To compute crowding distance for an individual, we average the distances to its immediate neighbors along the same front in every dimension (dimensions correspond to objective functions). Then, we put a rank value based on its nondomination level; (4) *Generate child population Q0*: form a child population Q0 on the basis of P0 by performing the followed genetic operators (i) Selection- Selection operator in NSGA-II is composed of picking child population from the parent population with the same size. The binary tournament selection [204] runs a tournament between two individuals and selects the winner; (ii) Crossover Operator- Crossover combines two parents, to form children, for the next generation. Then a scattered crossover is used [204]. This type of crossover creates a random binary vector. So, the genes are selected from the first parent where the vector is a 1, and from the second one where the vector is a 0, and combines the genes to form the first child, and vice versa to form

Table 5.5: NSGA-II features for the features subset selection.

Number of participants (P)	35x2
Size of population (n)	100
Stopping Criterion	100 generations
	All objectives values = 0
	Stall Gen. Limit = 10
Length of the chromosome (G)	30
Crossover probability	0.8

the second one; (iii) Mutation Operator- Adaptive Feasible Mutation adds a randomly generated number to each element in the child population. The direction (positive or negative) of the random number is adaptive with respect to the last successful or unsuccessful generation. The feasible region is bounded by the relative constraints and inequality constraints (0 and 1) [204]. (5) *Elitist Reduction of Population*: At the th generation, produce population Rt of size n by integrating parent population Pt with child population Qt; (6) *Rt population evaluation*: The new population Rt is sorted on the basis of domination and evaluated as described in (3). Assign a corresponding rank. (7) *Create new parent population Pt+1*: by filling the highest ranked front set until the size of the population size exceeds N'; (8) *Stopping criterion verification*: Goes to (5) until the stopping criterion is satisfied.

The used features on the NSGA-II are presented in table 5.5. The population size was 100 individuals. The evolution process ends if 100 generations are performed and/or fitness values reach zero and/or stall generations limit reaches 10.

5.5.4 Forward, Backward and Stepwise selection

Forward Selection: The forward selection method is simple to define. You begin with no candidate features in the multi-class SVM. Select the feature that has the highest classifier performance. At each step, select the candidate feature that increases classifier performance the most. Stop adding features when none of the remaining features are significant. Note that once a feature enters the model, it cannot be deleted [174].

Backward Selection: This method begins with a model in which all candidate features have been included. However, because it works its way down instead of up, one always retain large values of classifier performance. The problem is that the classifier may include variables that are not relevant. At each step, the feature that decreases the classifier's performance is eliminated [174].

Stepwise Selection: It is a combination of the forward and backward selection techniques. Stepwise regression is a modification of the forward selection so that after each step in which a feature was added, all candidate features in the classifier are checked to see if they increase or decrease the classifier performance [174].

5.5.5 Shrinkage Methods

As discussed in section 5.3, the standard SVM is equipped with L_2 penalty for regularization. Since L_2 penalty shrinks the fitted coefficients towards zero, it effectively controls the model variability and improves prediction performance especially when many variables are highly correlated [175]. However, L_2 penalty cannot set small coefficients to exactly zeros, so all M features are used in the learned model.

For the goal of feature selection, different penalty forms were suggested and reviewed by Huang et al. [175] to control model complexity and achieve sparse solutions. By shrinking small coefficients to exact zeros, a parsimonious model can be built. It is noteworthy that the suggested penalty forms use the same loss function as Crammer and Singer [187] (eq. 5.4).

Wang et al. [205] introduced L_1 penalty (also known as LASSO penalty) for MSVM for achieving sparsity in the solution. Then, Huang et al. [175] suggested that instead of applying the same penalty to coefficients, they can be adaptive: large coefficients receive small penalties, while small coefficients receive large penalties. Thus, Adaptive L_1 penalty appears where large coefficients can be protectively preserved during the selection process and small coefficients are decreased to zero, resulting in more sparse models.

Zhang et al. [206] proposed a new penalty form called sup-norm penalty, and Huang et al. [175] proposed an adaptive form (Adaptive sup-norm penalty) of this penalty with the same motivation as Adaptive L_1 penalty. For more details about these penalties consult [175].

In order to select the frequency of selection of each individual feature among cross-validation sets, Yamashita et al. [207] proposed Selection Counting Value (SC-value). The basic idea is that features that are repeatedly selected with good classification performance among a variety of training data sets could be important, so high SC-values should be assigned. Thus, with cross-validation we will obtain numerous coefficients for each feature with their corresponding measures of classification ACC. Then, SC-value can be defined by the total frequency of each feature selected, weighted by classification ACC. Let $l(k)$ and $p(k)$ denote the estimated feature vector and ACC, respectively, resulting from the k-th set division (cross-validation sets). Then the SC-value for the m-th feature is defined by

$$\mathrm{SC}-\mathrm{value}(m) = \sum_{k:p(k)>p_{\mathrm{thres}}}^{K} I(l_m(k) \neq 0).\, p(k) \quad m = 1,\ldots,M \tag{5.14}$$

where $I(.)$ denotes an indicator function that takes the value of 1 if the condition inside

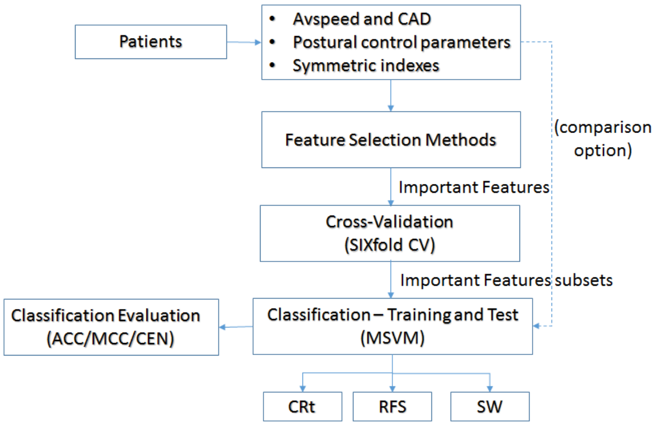


Figure 5.5: Proposed workflow of the porposed feature selection study.

the brackets is satisfied, 0 otherwise. K is the number of cross-validation sets (accordingly with section 5.3.1, K=36) and $l_d(k)$ is the estimate of the m-th element of $l(k)$. In order to exclude the results of poor classification performance, only data sets with classification performance exceeding a threshold level p_{thres} (80%) are included in the summation. It should be noted that an additional data set that is not used for calculating SC-values is used to evaluate generalization performance of the selected features.

5.5.6 Design/Methodology

The workflow of the methodology of this approach is presented in figure 5.5.

In this approach *SI* and postural control parameters were chosen to be evaluated as gait parameters. In addition, only *Avspd* and *CAD* were chosen to be evaluated in the group of spatiotemporal parameters. The others were not considered taking into account the previous results (subsection5.4.5).

Then, each feature selection method was applied to the gait parameters dataset, selecting a subset of important features. Such subset of features was then introduced in the SVM classifier. But, firstly, it is important to select subsets of the data (13 patients) to be used as training and test in the classification stage. In this study, a SIX fold cross-validation (CV) resampling approach is used (section 5.3.1).

The result of the classification, for each method, was evaluated by three metrics: *ACC, MCC* and *CEN* (section 5.3.2). NSGA-II was performed twice with different combinations of fitness: *ACC, MCC* and number of features (M); and *MCC, CEN* and number of features (M).

5.5.6.1 Statistical analysis

Descriptive analysis of spatiotemporal symmetry indexes and postural control features for each device was carried out, including the calculation of central tendency and dispersion such as the mean and standard deviation. Confidence intervals (CI) for the mean of symmetry features for each device were calculated and the distribution of the data was summarized and plotted using boxplots. Two-way and one-way ANOVA tests were performed with the spatiotemporal symmetry indexes and postural control features to determine if different devices, as well as different subjects, have significant effects on the features. Additionally, post hoc Tukey tests were performed to identify, for each feature, which combinations of ADs show statistical significant differences. One sample t-tests were performed to assess whether the symmetry indexes mean is different from zero. In all statistical tests, the significance level was set at 0.05. The software used was R (Version R-3.1.3) and MatLab 2012b (Natick, MA).

Linear regression models and Pearson's correlation coefficients between features were calculated to inspect the existence of correlations among them. In addition, multiple linear regression analysis was performed taking the selected important features as independent variables and the clinical outcomes, Berg Balance Scale (*BERG*) [182], *Avspd* and *CAD* as dependent variables.

5.5.7 Results

Spatiotemporal and Postural control features analysis

Measurements were performed on a total of 13 patients with 3 ADs (CRT, SW and RFS) each. The mean and standard deviation of the gait features for each patient were determined using data of a total of 30 gait cycles in steady gait. These values are shown in table 5.1 for each AD (Symmetrical Indexes and Postural Parameters).

The *SI* were calculated and analyzed in detail for each spatiotemporal feature. Boxplot of the *SI* for different ADs and parameters are plotted in figure 5.6. It can be seen that *SI-STR, SI-STP, SI-GC* and *SI-SWD* present positive means for all ADs. *SI-STAD* mean is only positive for CRT, *SI-DS* and *SI-SP* have negative mean for RFS and *SI-STPT* has negative mean for CRT.

In order to verify if each *SI* has a mean that is significantly different from zero (indicating lack of symmetry), one sample t-test was applied and the results (p-values) are presented in table 5.6. Based on these tests, it can be observed that all symmetry indexes, except *SI-SPT* (positive asymmetry seen in figure 5.6), did not show asymmetry between both legs with RFS ($p>0.05$). Only *SI-STR, SI-STAD* and *SI-SP* showed a symmetrical behavior with CRT ($p>0.05$). *SI-STP, SI-GC, SI-STPT, SI-DS, SI-SWD* showed to be positive (Figure 5.6). In re-

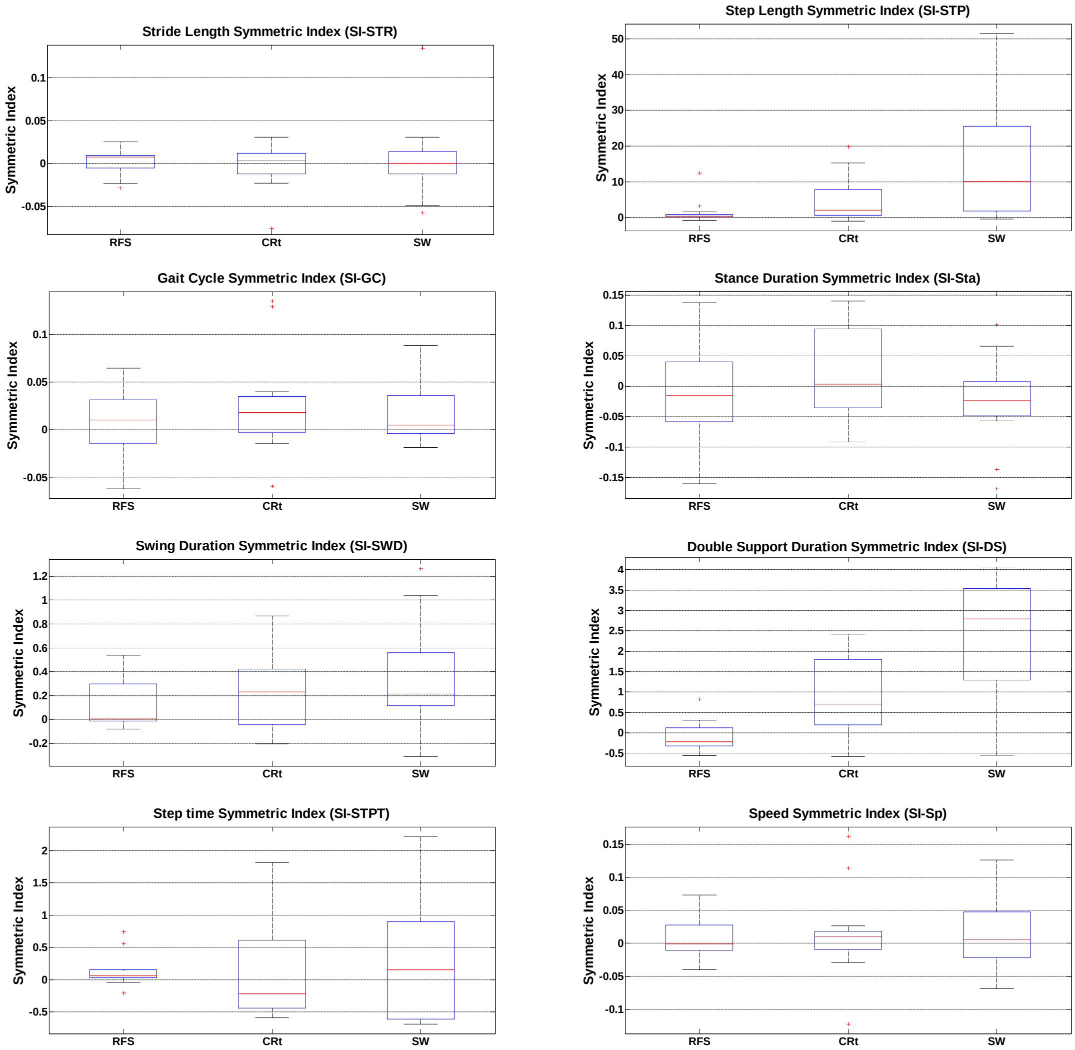


Figure 5.6: Boxplot and confidence intervals for each symmetric index.

Table 5.6: One sample t-test for *SI*.

Variables	One-sample t-test		
	RFS	CRT	SW
SI - STR	>0.05*	>0.05*	>0.05*
SI - STP	<0.05	<0.05	<0.05
SI - GC	>0.05*	<0.05	>0.05*
SI - STAD	>0.05*	>0.05*	>0.05*
SI – SWD	>0.05*	<0.05	<0.05
SI – DS	>0.05*	<0.05	<0.05
SI - STPT	>0.05*	<0.05	>0.05*
SI – SP	>0.05*	>0.05*	>0.05*

*statistical significance

lation to SW, the features *SI-STR, SI-GC, SI-STAD, SI-STPT* and *SI-SP* showed a symmetrical behavior ($p>0.05$). *SI-SW, SI-DS* and *SI-SPT* showed to be positive in figure 5.6 ($p<0.05$). The positive asymmetrical features indicate that values of these feature are larger for the *OL* than for the *NOL* ($p<0.05$).

Boxplots show a high variability of *SI-STP* in relation with the other features, being the most asymmetrical feature for all devices. The behavior of this index is consistent through all ADs, being positive for all ADs ($p<0.05$).

In terms of postural control features, SW presents decreased *RMS* accelerations, *DCOM, ACCCOM, ROMML* and *ROMV*. RFS presents decreased sway length in all directions and CRT presents decreased *ROMAP*.

The one-way ANOVA results (p-values) to test if there are significant differences between ADs for each feature are presented in table 5.7. It can be observed that there are statistically significant effects of ADs on *SI-STAD, SI-SWD, SI-STP, SI-DS, ROMV, SLML, SLV, RMSML* and *ACCCOM*. Then, Tukey post hoc allows concluding that features *SI-STP, SI-SWD, SI-DS, ROMV, SLV, RMSML* and *ACCCOM* have significantly different values for each AD and therefore should be able to distinguish which AD is the patient using. *SI-STAD* only distinguishes CRT from SW, and *SLML* distinguishes RFS from the other ADs.

With the purpose of a consistency analysis, two-way ANOVA was performed. Since feature selection and classification will be presented with the mean values of each patient, the authors want to verify if no variability information is lost. Also, we want to verify if there are differences between subjects, showing if each subject is different from the others when using a certain AD. Thus, two-way ANOVA analysis was performed using the 30 gait cycles data for each feature and patient between ADs. Thus, a total of 30 repetitions for each patient (n=13) was used for each feature and AD. Results are shown in table 5.7. Results among ADs showed consistency between two-way ANOVA and one-way ANOVA, since the features

Table 5.7: Symmetry indexes for each spatiotemporal features and postural control features from TKA patients (N=13) using 3 ADs. ANOVA and Tukey post hoc test results.

Parameters	One-way ANOVA	Tukey Post hoc (one-way ANOVA)	Two-way ANOVA			Tukey Post hoc (two-way ANOVA)
			ADs	Subj.	Interaction	
SI - STR	>0.05	-	>0.05	>0.05	>0.05	-
SI - STP	<0.05*	ALL ADs	<0.05*	<0.05*	<0.05*	ALL ADs
SI - GC	>0.05	-	>0.05	>0.05	>0.05	-
SI - STAD	<0.05*	CRT - SW	<0.05*	<0.05	<0.05	CRT - SW
SI - SWD	<0.05*	ALL ADs	<0.05*	<0.05*	<0.05*	ALL ADs
SI - DS	<0.05*	ALL ADs	<0.05*	<0.05*	<0.05*	ALL ADs
SI - STPT	>0.05	-	>0.05	<0.05	<0.05	-
SI - SP	>0.05	-	>0.05	>0.05	>0.05	-
ROMAP (mm)	>0.05	-	>0.05	>0.05	>0.05	-
ROMML (mm)	>0.05	-	>0.05	>0.05	>0.05	-
ROMV (mm)	<0.05*	RFS - CRT	<0.05	<0.05*	<0.05*	ALL ADs
SLAP (mm)	>0.05	-	>0.05	>0.05	>0.05	-
SLML (mm)	<0.05*	RFS-ALL	<0.05	<0.05*	<0.05*	ALL ADs
SLHOR (mm)	>0.05	-	>0.05	>0.05	>0.05	-
SLV (mm)	<0.05*	ALL ADs	<0.05*	<0.05*	<0.05*	ALL ADs
RMSAP	>0.05	-	>0.05	>0.05	>0.05	-
RMSML	<0.05*	RFS - ALL	<0.05	<0.05*	<0.05*	RFS-ALL
RMSV	>0.05	-	>0.05	>0.05	>0.05	-
RMSHOR	>0.05	-	>0.05	>0.05	>0.05	-
DCOM(mm)	>0.05	-	>0.05	<0.05*	<0.05*	-
ACCCOM (m/s^2)	<0.05*	ALL ADs	<0.05*	<0.05*	<0.05*	ALL ADs
CAD (step/min)	<0.05*	ALL ADs				
Avspd (m/s)	<0.05*	RFS-ALL				

*statistical significance

that are statistically significant in one-way ANOVA are also in two-way ANOVA (*SI-STAD, SI-SWD, SI-STP, SI-DS, ROMV, SLML, SLV, RMSML, ACCCOM*). Among subjects the same features are statistically significant. Also, *SI-STPT* and *DCOM* showed to be statistically different among subjects. The group of features that are statistically significant on the interaction between subjects and ADs include all the latter features. Through Tukey post hoc, it can be observed that the results are similar to one-way ANOVA. Only *SLML* showed some loss of information, since the one-way ANOVA (more general) only distinguishes between RFS and the other ADs, and two-way ANOVA (more specific) distinguishes between all ADs.

In figure 5.7, the results of a correlation analysis between the features are graphically presented. It is possible to verify that, in general, there are low correlations between features. Only some postural control features present positive high correlations: *ROMAP* and *ROMML* are strongly correlated to sway length features (*SLAP, SLHOR, SLML, SLV)*, since all are based on COM displacement measures. However, they are not strongly correlated to *RMS* measures. *CAD* and *Avspd* show a high positive correlation between them. In relation to the other features, *CAD* and *Avspd* showed very low correlations. *BERG* has low correlations with all features.

Feature Selection results

In this study, three classes were analyzed related with three ADs – RFS, Crt, SW. It is essential to determine which gait features are influenced by these ADs and that affect the behavior, and consequently the recovery of the TKA patient.

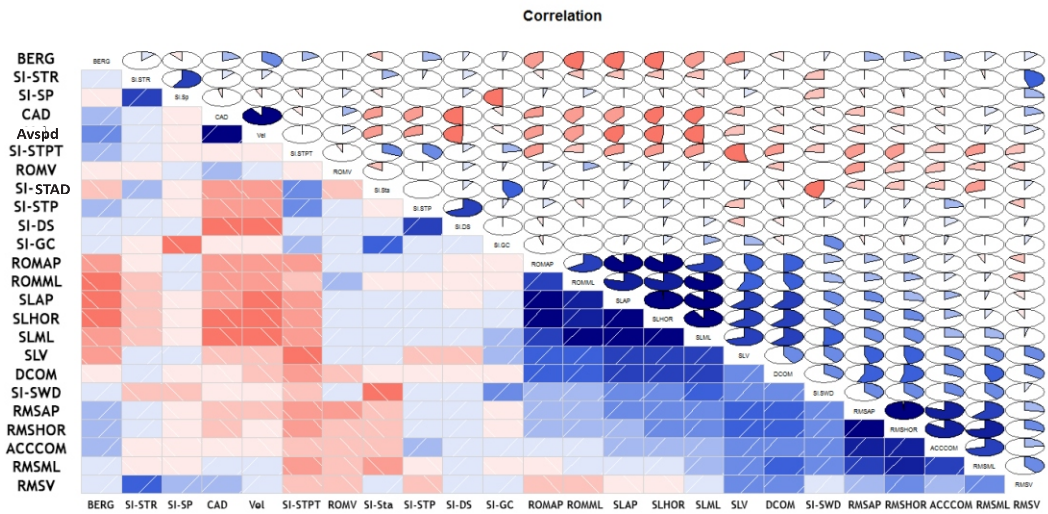


Figure 5.7: Linear correlation among all features (Pearson method). If the pie increases in clockwise direction (blue) the correlation is positive, if increases anti-clockwise (red) the correlation is negative. The ascend lines (blue) represent positive correlations and descend lines (red), negative correlations. Strong color corresponds to high corretation, and vice-versa.

Two different approaches of MSVM were performed: One vs rest (MSVM OvR) and discriminant functions (MSVM DF). Univariate statistics F-ratio, forward, backward and stepwise selection and NSGA-II were performed with MSVM OvR. Then, L_2, L_1, Sup-norm, Adaptive L_1 and Adaptive Sup-norm penalties were performed with MSVM DF.

In table 5.8, it can be observed the results of the different approaches considering classifier performance and number of selected features (m). The best set of features that result in the best classification performance was obtained with Sup-norm penalty and Adaptive Sup-norm Penalty. However all approaches, except the MSVM OvsR, L_2 and L_1 penalties, obtained an *ACC* performance greater than 91%, *MCC* greater than 0.8 and *CEN* lower than 0.2. L_1 penalty presents zero features, since SC-value obtained zero selected features with performance higher than 80%. In table 5.9, it is verified that each approach selected different sets of features with high performance of the classifier. Thus, the ones (features) that were more chosen between the approaches were evaluated with the two types of MSVM with test subsets. Table 5.10 shows the obtained results. The selected features obtained a performance equivalent to the ones obtained in table 5.8. The set selected with Sup-norm penalty and Adaptive Sup-norm was also evaluated with the test subset and obtained worse classification performance. It is noteworthy that the feature that differs from the two sets is *SI-STPT*.

It is also verified that *SI-DS, ROMV, SLV* and *ACCCOM* were the most chosen through the different approaches. These variables were also targeted by ANOVA as statistically significant features.

Table 5.8: Classification performance and number of selected features (m) by the different feature selection techniques.

Approaches	ACC	MCC	CEN	m
MSVM OvR	90.90%	0.820	0.160	21
Forward Selection	93.14%	0.843	0.180	5
Backward Selection	94.10%	0.908	0.090	10
Stepwise selection	93.14%	0.843	0.180	5
NSGA-II (MCC, CEN, N)	91.83%	0.817	0.190	7
NSGA-II* (MCC, ACC, N)	91.05%	0.812	0.190	7
Sup-norm Penalty	95.10%	0.850	0.130	16
L_2penalty	89.84%	0.790	0.220	21
L_1penalty	53.00%	0.100	0.810	0
Adaptive Sup-norm Penalty	95.40%	0.859	0.130	16
Adaptive L_1 penalty	91.00%	0.800	0.200	16
F-ratio	92.50%	0.840	0.150	8

Table 5.9: Selected features for each feature selection approach.

Approach	1	2	3	4	5	6	7	8	9	10	11	12	13	14	15	16	17	18	19	20	21
	SI-STR	SI-STP	SI-GC	SI-STAD	SI-SWD	SI-DS	SI-STPT	SI-SP	ROMAP	ROMML	ROMV	SLAP	SLML	SLHOR	SLV	RMSAP	RMSML	RMSV	RMSHOR	DCOM	ACCCOM
Forward			x			x				x	x							x			
Backward						x		x				x	x	x	x		x		x	x	x
Stepwise			x			x				x	x							x			
NSGA-II						x		x			x			x	x	x			x		
NSGA-II*						x			x						x		x	x		x	x
Sup-norm		x		x	x	x	x		x	x	x	x	x	x	x	x	x			x	x
Adap.L1		x		x	x	x	x		x	x	x	x	x	x	x	x	x			x	x
Adap. Sup-norm		x		x	x	x	x		x	x	x	x	x	x	x	x	x			x	x
F-ratio		x		x	x	x					x		x		x		x				x
FINAL SET		x		x	x	x			x	x	x	x	x	x	x	x	x			x	x

x - selected variables

Table 5.10: Final set of Selected features with SVM performance.

Classifier	ACC	MCC	CEN	m	Features
SVM OvR	94.5%	0.859	0.120	15	2,4, 5, 6, 9, 10, 11,12,13,14,15,16,17,20,21
MSVM L_2	92.0%	0.840	0.130		
SVM OvR	90.5%	0.80	0.190	16	2,4, 5, 6, 7, 9, 10, 11,12,13,14,15,16,17,20,21
MSVM L_2	89.0%	0.80	0.200		

Table 5.11: Multiple linear regression analysis.

	m	R^2	F	*p*-value	Error Variance
Avspd	15	0.642	2.759	0.014	0.005
	16	0.643	2.481	0.025	0.006
CAD	15	0.732	4.203	0.001	14.538
	16	0.731	3.769	0.002	21.872
BERG	15	0.611	2.315	0.034	43.244
	16	0.601	2.076	0.056	45.207

Table 5.12: Symmetrical indices of spatiotemporal features of patient #1. Confidence Intervals (CI) and mean values. One-way ANOVA (*p*-values) and Tukey post hoc results.

Features	RFS			CRT			SW			One-way ANOVA	Tukey Post hoc
	CI		Mean	CI		Mean	CI		Mean		
SI-STP	0.11	0.32	0.218	0.99	3.18	2.087	0.05	0.35	0.206	<0.05	CRT-ALL
SI-STAD	-0.02	0.23	-0.106	-0.28	-0.05	-0.169	-0.13	0.03	-0,.047	>0.05	-
SI-SWD	0.21	0.68	0.432	0.11	0.63	0.374	0.13	0.47	0.305	>0.05	-
SI-DS	-0.44	-0.17	-0.309	-0.75	-0.34	-0.547	0.03	0.54	0.286	<0.05	SW-ALL
SI-STPT	-0.01	0.25	0.119	-0.30	-0.13	-0.220	-0.01	0.31	0.147	<0.05	Rol - CRT

Figure 5.7 presents the correlations among the different features. A multiple linear regression was performed considering 6 different models, *i.e.* 3 different dependent variables (*Avspd, CAD* and *BERG*) and 2 different independent variable set (16 features selected by Sup-norm and Adaptive sup-norm penalty and 15 features selected by the overall approaches). Results are depicted in table 5.11. The R^2 suggests that the models *Avspd, CAD* and *BERG* explain approximately 64%, 73% and 60% of the variability in the response variables, and all of them are statistically significant, except the model *BERG* with 16 features ($p>0.05$). The general results reveal that there is a significant linear association between the selected features and the dependent variable (*Avspd, CAD* and *BERG*). However, with low variability explained.

Individual analysis results

In tables 5.12-5.17 are presented the individual analysis of 3 different patients with the selected features. For *SI* features, confidence intervals and mean values were calculated, as well as one-way ANOVA and Tukey post hoc. For postural control features, mean and standard deviation were calculated plus one-way ANOVA and Tukey post hoc. As it can be observed, each patient presents different values for the features, and the influence of the ADs is different among them.

5.5.8 Discussion

Nowadays, clinicians use several standardized and validated clinical scales filled out by the clinician or physiotherapist, such as Berg balance scale (*BERG*) [182]. *BERG* uses a range of test conditions, aiming to provide a holistic measure of balance. This evaluation is not based

Table 5.13: Postural control Features of patient #1. Mean ± Standard Deviation. One-way ANOVA (*p*-values) and Tukey post hoc results.

	RFS	CRT	SW	One-way ANOVA	Tukey Post hoc
ROMAP	5.98±2.30	1.07±0.23	0.27±0.01	<0.05	RFS-ALL
ROMML	1.25±1.04	0.58±0.23	0.44±0.11	>0.05	-
ROMV	5.90±.44	6.97±3.34	2.49±0.12	<0.05	ALL
SLAP	28.46±4.03	12.44±6.34	0.70±0.06	>0.05	-
SLML	16.69±3.76	7.64±0.34	1.51±0.34	>0.05	-
SLHOR	32.95±3.45	14.61±2.07	1.67±0.22	<0.05	-
SLV	26.09±6.45	25.94±5.03	14.39±1.03	<0.05	SW-ALL
RMSML	0.25±0.02	0.22±0.02	0.02±0.00	<0.05	SW-ALL
DCOM	0.26±0.03	0.15±0.02	0.17±0.00	>0.05	-
ACCCOM	0.54±0.23	0.40±0.01	0.19±0.05	<0.05	SW-ALL

Table 5.14: Symmetrical indices of spatiotemporal features of patient #2. Confidence Intervals (CI) and mean values. One-way ANOVA (*p*-values) and Tukey post hoc results.

Features	RFS			CRT			SW			One-way ANOVA	Tukey Post hoc
	CI		Mean	CI		Mean	CI		Mean		
SI-STP	0.01	0.23	0.122	0.64	1.09	0.869	0.61	3.40	2.005	<0.05	RFS-SW
SI-STAD	-0.10	0.07	-0.015	0.01	0.14	0.073	-0.07	0.27	0.101	<0.05	RFS-ALL
SI-SWD	-0.13	0.04	-0.045	-0.23	-0.04	-0.133	0.03	0.37	0.198	<0.05	SW-ALL
SI-DS	0.04	0.19	0.118	0.76	1.48	1.124	0.86	2.21	1.538	<0.05	ALL
SI-STPT	0.02	0.16	0.093	0.25	0.80	0.531	0.03	1.09	0.562	>0.05	-

Table 5.15: Postural control Features of patient #2. Mean ± Standard Deviation. One-way ANOVA (*p*-values) and Tukey post hoc results.

	RFS	CRT	SW	One-way ANOVA	Tukey Post hoc
ROMAP	2.50±0.12	2.43±0.26	2.55±0.25	>0.05	-
ROMML	0.73±0.09	1.24±0.08	1.94±0.09	>0.05	-
ROMV	2.78±1.30	4.78±1.04	4.86±0.85	<0.05	RFS-ALL
SLAP	15.51±1.23	17.11±4.03	23.83±7.85	>0.05	-
SLML	6.41±0.32	6.96±4.01	16.66±3.77	<0.05	SW-ALL
SLHOR	16.78±2.25	18.47±3.45	29.08±14.92	>0.05	-
SLV	20.54±4.34	24.80±10.32	32.93±15.03	<0.05	ALL
RMSML	0.57±0.01	0.37±0.04	0.41±0.09	>0.05	-
DCOM	0.35±0.03	0.48±0.13	0.36±0.08	>0.05	-
ACCCOM	0.41±0.00	0.51±0.25	0.63±0.18	<0.05	ALL

Table 5.16: Symmetrical indices of spatiotemporal features of patient #11. Confidence Intervals (CI) and mean values. One-way ANOVA (*p*-values) and Tukey post hoc results.

Features	RFS			CRT			SW			One-way ANOVA	Tukey Post hoc
	CI		Mean	CI		Mean	CI		Mean		
SI-STP	-0.22	8.09	3.934	0.02	-0.41	-0.215	4.69	16.99	10.840	<0.05	SW-ALL
SI-STAD	-0.08	0.05	-0.014	-0.08	0.09	0.004	-0.18	0.14	-0.022	>0.05	-
SI-SWD	0.09	0.36	0.231	-0.14	0.15	0.004	0.09	0.36	0.211	<0.05	CRT-ALL
SI-DS	1.31	2.08	1.702	0.49	1.15	0.825	1.52	3.06	2.292	<0.05	ALL
SI-STPT	-0.63	-0.54	-0.589	-0.82	0.81	-0.01	-0.37	-0.03	-0.203	>0.05	-

Table 5.17: Postural control Features of patient #11. Mean ± Standard Deviation. One-way ANOVA (*p*-values) and Tukey post hoc results.

	RFS	CRT	SW	One-way ANOVA	Tukey Post hoc
ROMAP	2.49±1.02	1.91±0.01	3.30±1.02	<0.05	ALL
ROMML	0.52±0.21	0.23±0.01	0.40±0.20	<0.05	ALL
ROMV	6.45±1.32	3.73±0.03	4.85±0.14	<0.05	ALL
SLAP	16.03±4.02	13.76±0.78	21.24±4.03	<0.05	SW-ALL
SLML	5.71±2.30	2.45±0.09	11.69±4.05	<0.05	ALL
SLHOR	16.25±3.02	14.80±1.02	24.24±3.08	<0.05	SW-ALL
SLV	24.19±9.23	21.03±3.04	26.06±5.05	<0.05	ALL
RMSML	0.21±0.01	0.14±0.03	0.29±0.08	<0.05	CRT-ALL
DCOM	0.39±0.20	0.23±0.08	0.22±0.09	>0.05	-
ACCCOM	0.41±0.10	0.37±0.02	0.46±0.11	>0.05	-

on objective physical measurements but depends on the subjective opinion of the patient, the physiotherapist or clinician, so it is difficult to perform an accurate and objective assessment [39]. Moreover, there is no objective assessment of which assistive device is more appropriate for the recovery of a certain patient [39]. Normally, clinicians tend to generalize the patients as, in this case, a group of post-surgical patients of TKA, and then prescribe the same treatment for all. However, depending on the degree of balance and symmetry problems; patients may require different devices for their recovery.

The aim of this study is to investigate which mobility characteristics (gait features) during assisted walking are more affected by three types of ADs, and if these ADs have different effects on the gait features of the same patient and between subjects. After identifying these gait features, we want to know if they are related to gait velocity, cadence and clinical outcomes (*BERG*) of TKA post-surgical patients. For this purpose, we evaluated as gait features the symmetrical indexes and postural control features for quantitative assessment. Martínez-Ramírez et al. [183] demonstrated that inter-limb asymmetry provides important additional information about individual gait pattern, which is not represented by gait velocity and clinical outcomes. In addition, postural information is important to infer stability and safety during the recovery of the patients, to avoid falls and the sense of insecurity.

Studies relative to rollators revealed that this device is safe and stable, providing improve in balance and mobility [57, 59, 67]. This AD also causes a lower variability in gait and more natural gait [46, 58, 59]. In contrast, there are several authors indicating that this AD causes changes in posture and, increased risk of falling [57, 58, 77]. However, on this study forearm supports were added to the device in order to provide more stability to the gait, better posture and increased support [25, 65, 69]. This could result in an upper limbs' excessive effort [74], however since this device will just be used during recovery, not being a long term device, upper limb problems would not be a concern. On the other hand, SW is known as the more stable device, supporting a greater percentage of body weight [64]. Still, it provides a slower and varied gait, less mobility with higher metabolic cost due to reduced speed and repetitive motion for lifting it while moving forward [50, 51, 58, 59, 64]. CRT have upper limbs' support, however, these devices require some energy cost, excessive upper limb effort and do not provide a natural gait [1].

On this study, RFS demonstrated to be the AD that, in general, provides for a more symmetrical gait. CRT demonstrated to be the worst AD in terms of providing a symmetrical gait for their users. This result may be influenced by the different type of gait that is adopted by the patient when using the ADs. Regarding CRT, the patient learns a three points gait, *i.e.* he/she always has one foot and two crutches on the ground while moving forward [60]. With SW, the patient first lifts and moves the device, placing it in front of him, and then he moves his legs.

Finally with RFS, the movement is continuous by pushing the walker. It is believed that the type of gait that the patient presented for each AD influences the distinction of ADs.

SI-STP is the SI feature with higher variability, being the one that differs more between subjects and is significantly affected by ADs (p<0.05). The high positive values demonstrate that *OL* does not have sufficient strength to endure a large step of the *NOL*, thus the step of the *NOL* will be shorter. In addition, *NOL* presents a higher step time (*STPT*) and swing duration (*SWD*) and a lower stance duration (*STAD*). The *SI* of these features will be positive, positive and negative, respectively. However, CRT present a positive average *SI-STAD* which means that *OL* is in effort for more time than *NOL*. Maybe crutches can provide a higher support on the upper limbs, decreasing the load and pain on *OL*. However, this does not happen with all patients.

SI-SWD and *SI-DS* are only symmetrical with RFS, maybe because it provides for a more natural motion, which does not oblige the person to stop in order to move forward. *SI-DS* on the other ADs, tend to be positive, which means that the patient takes more time in double support when the progress-leg (leg that initiates the movement to go forward) is *NOL* since *OL* has to support the beginning of the motion. RFS is the only device that presents all symmetrical gait phases (*SI-DS, SI-SWD, SI-STAD* and *SI-GC*). This might indicate that if a person has to recover in terms of time of support of the knee, and create confidence to walk as natural as possible, RFS seems to provide the necessary help. However, *SI-STAD* result of CRT indicates that it provides for a greater load of weight on the upper limbs, decreasing the weight supported by the operated knee, and consequently decreasing knee pain. Moreover, gait phases *SI* features (*SI-DS, SI-STAD* and *SI-SWD*) seem to be the ones that are more affected by ADs and are differently affected in each person by all ADs. This means that patients require different types of help and respond differently to each AD.

On the overall, there are more positive *SI* features that indicate greater asymmetry towards *NOL* meaning that patients put more weight on the non-operated leg throughout the gait cycle. These results are in agreement with those of [198, 199]. The asymmetry of weight bearing might also depend on small changes in the body configuration [198, 199]. Therefore, the additional stress on *NOL* may develop osteoarthritis in that leg. For this reason, it is important to identify which devices are more appropriate for each person, based on its initial performance.

Relatively to postural control parameters, the lower they are the more support the AD is providing. In general, SW is the more stable, and CRT is the worst. Results indicate that *ROMV, SLML, SLV, RMSML* and *ACCCOM* provide significant discrimination between ADs and subjects, which may have relevance to unsupervised balance assessment and comparison between ADs, inferring stability and safety. In addition, since no high correlation was identified between *RMS* acceleration features and the other postural control features, normally not used for balance assessment, they may provide information not captured by *RMS* acceleration features.

SW results in decreased *RMS* acceleration and *ROM* values. By other hand, RFS provides more stability in sway length and CRT stabilize the *ROMAP*. These results demonstrate that SW stabilizes the acceleration of the trunk, and its users are more "static" than with the other ADs. RFS stabilizes the position of the trunk, not allowing the user to sway the trunk. CRT is better to give a more erect posture to its users, decreasing the anterior-posterior range of motion, *i.e.* flexion and extension of the trunk.

It is also noteworthy that RFS present a greater velocity and cadence, and SW provides the slowest gait and more stable, as referred in [23].

From these features, the most important and relevant ones were selected to evaluate the differences between ADs. The ones that can better distinguish the pattern acquired by the ADs will be selected to evaluate which device should be prescribed for TKA patients. In order to select the important features which can discriminate the differences among the ADs, we investigated different types of techniques for feature reduction and selection. The most important features were identified in table 5.9. It is interesting to verify that all identified features in ANOVA (Table 5.7) were considered to be important by the feature selection techniques. The best performance obtained was with the set of 16 features (*SI-STP, SI-STAD, SI-SWD, SI-DS, SI-STPT, ROMAP, ROMML, ROMV, SLAP, SLML, SLHOR, SLV, RMSAP, RMSML, DCOM* and *ACCCOM*). However, the final chosen set was *SI-STP, SI-STAD, SI-SWD, SI-DS, ROMAP, ROMML, ROMV, SLAP, SLML, SLHOR, SLV, RMSAP, RMSML, DCOM* and *ACCCOM*, which does not include *SI-STPT*. This feature can be controversial, since it detects differences among subjects and ADs, but it is not significantly affected, in general, by the ADs. By this, the two sets were tested to verify how the inclusion of this feature affects the performance of the classifier with the test subset samples. Results conclude that the performance was good (>90%) but poorer with *SI-STPT*, so it was not considered on the final set.

Testing for correlations between the set of selected features and *Avspd, CAD* and *BERG*, it was concluded that selected features were not highly correlated with *Avspd* and *BERG*. Being more correlated with *CAD*, but with explained variance of 73%. Thus, the set of features cannot predict gait velocity, cadence and the score of the *BERG*. The opposite is not possible also, since *SI* and postural control features showed low correlation with *Avspd*, *BERG* and *CAD*. Therefore, SI and postural control features provide important additional quantitative information about the functional mobility performance, which is not represented by *BERG, CAD* and *Avspd* . Moreover, the great variability in the selected *SI* and postural features within our patients indicates that asymmetry and posture differs between patients and is, therefore, important as independent measures.

The scales usually reflect different aspects of functionality of patients to develop activities, but not how to perform them [208]. Patients may try to maintain their functional capacity as normal as possible despite the pain and discomfort. In the current study, the authors did not find a relation between gait features and clinical outcomes, which supports the findings of Vissers et al. [208]. It is, therefore, important to measure gait features in addition to scales to understand how patients walk after surgery in order to tailor rehabilitation programs for potential recovery of normal walking patterns [198].

Finally, in order to verify the obtained results that the selected features are important to provide the information necessary to help on prescribing an assistive device, data from three patients were shown in tables 5.12-5.17. Looking at these tables, patient #1 presents lower values for SW, patient #2 for RFS and patient #11 for CRT. Thus, these devices should be taken into consideration for the recovery of these patients.

5.5.9 Conclusions and Originality

Inter-limb asymmetry and postural control features can be evaluated in an outpatient setting, supplying important additional information about individual gait pattern, which is not represented by gait velocity, cadence and scales usually used. The features calculated in this study are able to provide complementary information to gait velocity, cadence and clinical scales to assess the functional capacity of patients that passed through TKA. The selected parameters make a new clinical tool useful for tracking the evolution of patients' recovery after TKA.

Further studies should collect more data from TKA patients creating a database with the proposed features on this study, to then create a model that can be used to help the physicians on deciding which assistive device can be more adequate to a certain TKA patient.

Chapter 6

Introduction of a Smart Walker for Rehabilitation in Ataxic Patients: Case Studies

The term ‘ataxia’ refers to movement coordination disorders and is frequently caused by cerebellar injuries [209]. The functional role of cerebellum in motor control is the adjustment of movements, playing a critical role in balance and locomotion. Cerebellar damage can produce oculomotor disturbances, speech deficits, disturbances in limb movements, deficits of posture and gait, deficits of cognitive operations or subtle autonomic signs [210].

Ataxic gait has been characterized by a widened or alternatively variable base of support, inappropriate timing of foot placement, reduced step frequency, increased step width, and prolonged time in double-limb support. Both impaired postural stability and decomposition of multi-joint leg movements appear to be factors in cerebellar gait ataxia [211].

Posture and balance involve both the ability to recover from instability as the ability to anticipate and move in order to help avoiding instability. Lesions in the cerebellum can result in postural sway and backward balance reactions. The use of vision may not be effective in preventing loss of balance. These individuals are thus very prone to falls [211].

There are genetically medicated ataxias (spinocerebellar ataxia, Freidreich’s ataxia, for example), with symmetrical and slow progression, and acquired ataxias (caused by injury, stroke, hemorrhage, infeccious/inflammatory processes, metabolic or toxic derangements, neoplastic/mass effect), that may have a more sudden onset and may be asymmetrical or focal [212].

The treatment of the underlying disease is presently possible only for a little subgroup of cerebellar ataxias with metabolic dysfunction. To all the others, ataxia treatment is primary based on physical medicine and rehabilitation (PMR) intervention [213].

This chapter proposes a new PMR intervention with the introduction of the developed smart walker, ASBGo walker (SmartW). Taking into account that the use of walkers can be problematic for patients suffering from ataxia, ASBGo walker was included on the rehabilitation program of six ataxic patients to infer its potential.

In addition to introducing ASBGo walker on the treatment of ataxic patients, the gait assessment tool developed in chapter 4 was used, in real-time, to evaluate the patient's progress by assessing spatiotemporal and postural stability parameters. This information is then analyzed to help deciding when the patient can leave the SmartW and to monitor the progression of the patient. It is investigated the relationship between the static and dynamic recovery balance throughout gait training with a SmartW. Also, an intra-individual comparison is performed, since all cases are different. The goal is to evaluate the recovery of each patient in time, by comparing with his/her own results.

Therefore, it will be first presented a brief background about the current types of ataxia treatment. Then, a description of the experimental setup and parameters acquisition, which includes the presentation of the advantages of using ASBGo walker for the rehabilitation of patients of ataxia as well as details of clinical, gait and stability assessments, is presented. The participants under study and the protocol are also described. Finally, the results and discussion of the evolution of the patients and clinical relevance are presented.

The study presented in this chapter resulted in one conference publication [31] and four poster communications in conferences of the medical and physical therapy field [32–35].

6.1 Ataxia Treatment: Background

Ataxia treatment is primarily based on PMR intervention. The physiotherapy exercises promote postural stability, functional balance and stimulate precision movements of the limbs (without losing stability), aiming to teach the patient to reduce postural sway (frequency and amplitude) and to better control the position and body alignment and improve gait pattern. In addition, it is essential to promote controlled mobility activities (weight shift, swing, in and out of postures or movement transitions), perform tasks such as reaching an object with the hand while maintaining stable posture, as semi-kneeling or even standing [214].

Physiotherapy interventions found in current literature include training of balance and protection reactions, gait and coordination exercises; muscular strengthening and hydrokynesis therapy. Water offers graduate resistance that slows the ataxic patient motion and can help the patient in recovering the coordination and control of movements [211]. More recent techniques such as trans-cranial magnetic stimulation, virtual reality, biofeedback, treadmill exercises with supported bodyweight and torso weighting appear to have potential, however their specific efficacy has to be further investigated [215].

In Appendix C it can be found a table review of the state of the art of physical therapy studies with patients with ataxia [211, 216–241].

Despite the clinical consensus in relation to the usefulness of physical therapy exercises, documentation of the effects of different protocols on functional performance of subjects with ataxia is scarce in scientific literature [209]. However, recent literature suggests that individuals with ataxia may benefit from motor and functional long-term training. After intensive treatments in this area, patients with cerebellar disorders have shown improvement in motor and functional tasks [216]. Ilg et al. [213, 216] found that frequency was an important factor in retaining changes.

The timeline recovery depends on many factors, that may be not very well established. It can take one month or one year for an ataxic patient to recover. It seems to depend on the patient's state, on the degree of disorder and on self-motivation. Many physical therapy exercises are proposed but the recovery duration is not standard. That is why every rehabilitation program should be tailored to each patient.

Some ataxic patients use walking aids to gain some independence. The use of a walking stick or a walking frame can improve postural stability when balance is poor [213]. However, its use is difficult when cerebellar syndrome also affects upper limb movement control, because placing and controlling such device may be as difficult as trying to accurately place legs during swing phase. In Marquer et al. [242], a study with healthy adults using walking aids was performed. They found that these devices compromise the ability to respond to balance disturbances through impeding lateral compensatory stepping and thus can affect safety [242]. Assistance and training in the use of walking aids are essential because the inappropriate use is associated with an increased risk of falls [214]. In addition, the use of walking aids decreases the weight bearing over lower limbs through the upper limbs [216].

Other important concern is to select the appropriate measurement and assessment methods. In the literature [211], clinicians use scales, observational methods and computerized systems to assess patient's balance and functional capacities. Although the observational methods are easy to use, their ability to provide standardized measurements is limited, and the results can be very subjective, varying with the person who has done the observation [211]. Although computerized methods are highly reliable, they are costly systems, which require working within the laboratory environment.

Because of such limitations, the author proposes in this chapter to test ASBGo walker with patients with different types of ataxia. As it was already mentioned, this SmartW (ASBGo walker) was specifically designed for adapting to different patients. In addition, it was developed to be use as an ambulatory tool for evaluating the status of a patient during his/her treatment. In subsection 6.2.1, it is specified the advantages that ASBGo walker can bring to the rehabilitation of patients with ataxia regarding its design and funcionalities.

Quantitative evaluations were carried out using the sensory systems presented in chapter 4, enabling to quantify the gait evolution and to identify the problems that benefited from this treatment, as well as the best outcome measures. This experiment provides the first steps towards the creation of a clinical evaluation protocol whose results will show continuously the progression of the patients using a ASBGo walker with forearms.

6.2 Experimental Setup and Parameters Acquisition

6.2.1 Advantages of ASBGo Walker

The combination of impaired balance and discoordination in lower-limb dynamics of ataxic patients suggests a strong rationale for the use of ASBGo walker for gait training. We hypothesized that using this ASBGo walker has several advantages: (1) the existence of a table-support of the walker allows for postural control and increase in stability; (2) the motorization allows systematic control and progression of the speed at which walking is performed; (3) the repetitive training of a complete gait cycle enables a more appropriate pattern of sensory input associated with the different phases of gait to stimulate the gait pattern; (4) the design allows the physiotherapist to provide manual assistance to help the patient simulate a more normal walking pattern; (5) the patient performs dual tasking (guides the walker and corrects his/her gait at the same time); (6) the patient walks in real environment, avoiding obstacles; (7) the patient is provided with body feedback; and (8) with axial support, decreasing his/her tremor and dysmetria.

As it can be seen, the device includes several functionalities in order to include the main advantages of conventional physical therapy exercises in one device.

Controversial points in literature were also included in order to verify their efficacy.

Point (5) was reported in [243] showing that patients with cerebellar ataxia present difficulty in performing dual tasks, mainly when cognitive tasks were requested as secondary task. Scarce data is available in literature that relates the difficulty showed by people with ataxia during the performance of actions that require association between cognitive structures and musculoskeletal functions, configuring dual task. Studying this association is important in ataxic patients because of the role of the cerebellum in motor memory storage and the possibility of correction and adaptations during a motor act performance. With the SmartW treatment here proposed, such results can be visualized in a long-term point of view (during the entire rehabilitation process), verifying the long-term effects of walking with dual task, such as guiding the walker while correcting the gait pattern and avoiding obstacles (point 6).

Another controversial point may be point (7). Some studies reported the importance of body feedback in ataxic patients [244, 245]. This feedback provides a continuous input concerning the posture and balance, as well as position, speed, pace and strength of the slow movements in the peripheral body segments. In order to verify the potential of this point, the feet will be filmed and showed to the patients in real time.

Finally point 8 consists in providing some support and "fictional" weight on the upper limbs of the patient with the forearm supports. Gibson-Horn et al. [217] showed that with weighting, the patient demonstrated less sway in quiet standing, increased stability when perturbed, improved body alignment, and less ataxia during gait. The patient was able to accomplish more challenging activities with better balance while weighted. Thus, placing small amounts of weight asymmetrically on the torso, based on directional loss of balance and alignment, seemed to assist the patient in maintaining balance during static and dynamic activities. Morgan et al. [246] found improvement in gait in 11 out of 14 patients with ataxia when they were weighted at the waist and the lower extremities. One severely disabled subject was able to walk while weighted, but could not walk without the use of the weights. Clopton et al. [247] reported improvement in some gait parameters when 10% of a subject's body weight was placed on the subject's shoulders and then on his or her waist. Thus, axial weighting could improve some aspects of balance and mobility. These studies suggest that increasing sensory input via the application of additional weight may increase afferent input from deep pressure receptors, thus facilitating co-contraction of the muscles and increasing stability. Changing the center of mass (COM) by placing additional mass at appropriate locations alters the moment of inertia, affecting movement. Patients with ataxia often have difficulty controlling their body movements. Weighting may provide increased afferent input regarding a change in the biomechanical relationship, thus resulting in increased body control. Therefore, forearm supports were included in the SmartW to impose some weight on the trunk and arms of the patients.

6.2.2 Clinical, gait and postural stability assessment

Postural disorders in cerebellar ataxia can be evaluated both qualitatively and quantitatively. Qualitative evaluations are based on a precise assessment of clinical symptoms. Also, certain generic evaluations of balance disorders and ordinal scales evaluating the various components of ataxia can be used to quantify the severity of postural disorders in cerebellar ataxia. The generic evaluations of balance include the Berg Balance Scale (BBS), time standing tests, like the Time Up and Go (TUG) and posturography [213]. Generic gait assessments are also useful and include basic spatiotemporal gait parameters (stride length, stance duration, etc) [213].

The two most frequently used specific scales for the evaluation of cerebellar ataxia are the recently developed Scale for the Assessment and Rating of Ataxia (SARA), and the older International Cooperative Ataxia Rating Scale (ICARS) [211]. These scales were not used on this study, since the patient is Portuguese, and such scales are not validated for the Portuguese population.

The aforementioned scales measure simple indoors activities and present a “ceiling effect”. The Activities specific Balance Confidence (ABC) Scale [248] was designed to evaluate balance more thoroughly in several activities of daily living with different levels of difficulty.

In this study, (a) balance was evaluated with BBS, TUG and ABC, (b) spatiotemporal gait parameters (stance and swing duration, stride and step time and length, double support duration, step width and cadence) were measured with the developed sensorial system based on active depth sensor (ADS) and laser range finder (LRF) sensor (chapter 4) and (c) postural stability (trunk range motion, sway length, center of mass displacement and acceleration) was evaluated with accelerometers placed at the trunk (chapter 4).

6.2.2.1 Berg Balance Scale (BBS)

BBS was developed to measure balance among older people with impairment in balance function by assessing the performance of functional tasks [182]. It is a valid instrument used for evaluation of the effectiveness of interventions and for quantitative descriptions of function in clinical practice and research. The BBS has been evaluated in several reliability studies [211].

The test takes 15–20 minutes and comprises a set of 14 simple balance related tasks, ranging from standing up from a sitting position, to standing on one foot. The degree of success in achieving each task is given a score of zero (unable) to four (independent), and the final measure is the sum of all of the scores (56).

This test presents objectivity good test-retest, and can discriminate who is more prone to falls. The decline in scores on this scale was associated with a high risk of falling, but this relationship is not linear. In the range of 56 to 54 at each point less Berg scale is associated with an increase of 3 to 4% risk of falls. However, in the 54-46 range, a change from a point in that range was associated with an increase of 6 to 8% risk of falls. Below 36, the risk is close to 100%. Therefore, a change point in BBS can lead to a very different prediction of the probability of falls. In this study there were subjects with score in BBS from 5 to 35, which means that the risk of falls approaches 100%.

BBS was performed every 5 sessions.

6.2.2.2 Timed up and Go

The Timed Up and Go test (TUG) is a simple test used to assess a person's mobility and requires both static and dynamic balance [209]. It uses the time that a person takes to rise from a chair, walk three meters, turn around, walk back to the chair, and sit down. During the test, the person is expected to wear their regular footwear and use any mobility aids that they would normally require.

BBS was performed every 5 sessions.

6.2.2.3 Activities specific Balance Confidence (ABC) Scale

Activities specific Balance Confidence (ABC) Scale [248] is a subjective measure of confidence in performing various ambulatory activities without falling or experiencing a sense of unsteadiness. It consists of a 16-item self-report measure in which patients rate their balance confidence for performing activities. This stem is used to lead into each activity considered: "How confident are you that you will not lose your balance or become unsteady when you...". Items are rated on a rating scale that ranges from 0-100, where score of zero represents no confidence and a score of 100 represents complete confidence. The overall score is calculated by adding item scores and then dividing them by the total number of items.

By following the study of Lajoie and Gallaghe [249], a score lower than 67% indicates a risk for falling.

ABC was performed at the beginning and end of ASBGo walker treatment.

6.2.2.4 Spatiotemporal Gait Parameters

Clinical evaluation during walker-assisted gait is the first step to assess the evolution of a patient during rehabilitation and to identify his needs and difficulties. Advances in robotics made it possible to integrate a gait analysis tool on a walker to enrich the existing rehabilitation tests with new sets of objective gait parameters.

In chapter 4, it was presented a sensorial system that tracks the lower limbs in order to evaluate the gait pattern.

On this chapter, the following spatiotemporal parameters will be calculated: step and stride length (*STP* and *STR*), stride width (*WIDTH*), gait cycle (*GC*), cadence (*CAD*), velocity (*Avspd*), stance and swing phase duration (*STAD* and *SWD*), double support duration (*DS*) and step time (*STPT*).

With these spatiotemporal parameters, it is possible to calculate stride-to-stride variability [198]. This is a strong indicator of risk of fall. Other important indicator is the symmetry of parameters. This can tell us if the coordination between legs is improving or not [198].

Symmetry indices (*SI*) were calculated in chapter 5 (section 5.2.2, eq. 5.1). However in this case:

$$SI = \frac{U_R - U_L}{U_L} \tag{6.1}$$

U_R and U_L are any aforementioned features for the right (R) and left (L) leg, respectively. Perfect symmetry results if SI is zero, larger positive and negative deviations would indicate a greater symmetry towards the right or left leg, respectively.

Thus, these two indicators will be calculated. It is hypothesized that when gait training with the ASBGo walker, the gait symmetry will tend to zero and variability will decrease. Moreover, the gait pattern will improve in all alternative devices, since the ASBGo walker will help training the gait pattern, confidence and stability. This improvement will allow the patient to change for other walking aids, gaining more and more independence, until he/she can walk alone.

Spatiotemporal evaluation was performed every 5 sessions.

6.2.2.5 Postural Stability

To assess postural stability, an accelerometer is located near to the center of mass (COM), as suggested in chapter 4. Tests were performed in two situations: static position, which consists in 3 conditions (comfortable stance (CS), right and left semi-tandem stance (RSS and LSS)) as shown in figure 6.1, and dynamic position (the patient walks with ASBGo walker and other assistive device). These two situations will help to infer the evolution of the static and dynamic postural stability of the patient as well as his risk of falling.

The calculated postural stability parameters in this chapter are the root mean square of anterior-posterior (AP), horizontal (HOR) and medio-lateral (ML) accelerations (*RMSAP, RMSHOR* and *RMSML*), range of motion of AP and ML directions (*ROMAP* and *ROMML*) and sway length (*SLML, SLAP* and *SLHOR*). The calculation of these parameters is presented in chapter 4. In addition, the COM trajectories in AP and ML directions were also acquired. The variability of these parameters will be also calculated to infer risk of fall [165].

Postural stability evaluation was performed every 5 sessions.

Figure 6.1: Test Conditions: Comfortable stance (CS) on the left and semi-tandem stance (SS) on the right [165].

6.3 Methods

6.3.1 Cases Description

- Case 1: Male patient, 64 years-old. In 2014, he was admitted in the hospital with sudden right ataxic hemiparesis, due to neurobrucellosis. He started antibiotic therapy and rehabilitation program. At the beginning of the therapy, he scored 6/56 points on Berg Balance scale (BBS) [182], and required the assistance of 2 people to walk. He was as an inpatient patient at PMR department.

- Case 2: Female patient, 28 years old, with cerebellar pilocytic astrocytoma, surgically removed in 2005. She presented gait ataxia, dysarthria, nystagmus; bilateral upper limb dysmetria and intention tremor. She did not show significant improvement since the recovery period after surgery, albeit doing physiotherapy regularly. At the beginning of the therapy, she scored 12/56 on BBS [182] and was able to walk with 2 crutches, but with close supervision of a third person. She was an outpatient patient.

- Case 3: Female patient, 46 years old, with ataxic tetraparesis and lesion of cranial nerves VI, VII, IX and X, due to a petroclival meningioma surgically removed in 11/2014. She underwent surgical tracheostomy and percutaneous endoscopic gastrostomy. At the beginning of the therapy, she scored 4/56 on BBS [182] and required assistance to remain standing and walk. She was also not capable to sit and stand-alone. She was an inpatient patient at PMR department.

- Case 4: Female patient, 50 years old, was post hospitalized for cerebellar vermis syndrome with no identified etiology. At the beginning of the therapy, she scored 30/56 on BBS [182] and required the assistance of one person to walk. She was an inpatient patient at PMR department.

- Case 5: Female patient, 42 years old, with Friedreich ataxia presenting a high fall-risk (falls 1/2 times per day). At the beginning of the therapy, she scored 34/56 on BBS [182]. She presented difficulty in moving from sitting position to standing position; difficulty with balance in standing position and coordination. She was an outpatient patient and walks with a four-wheeled walker with brake and basket system, daily.

- Case 6: Female patient, 48 years old, with ataxic tetraparesis and lesion of cranial nerves due to a retroclival meningioma surgically removed in 02/2015. At the beginning of the therapy, she scored 0/56 on BBS [182] and required assistance to remain standing and did not walk. Her trunk balance was insufficient in standing position with anterior tilt of the trunk and difficulty in stabilizing the pelvis. She was an inpatient patient at PMR department.

An informed consent was signed by the patients and the study was approved by Braga Hospital Ethical Committee (Appendix B).

6.3.2 Protocol

6.3.2.1 Examination/Evaluation

Before beginning the gait training with ASBGo walker, all baseline data was collected. Patients were evaluated by the application of BBS [182] and with static and dynamic tests the information was gathered by several sensors integrated in the device, which allowed characterizing the assisted gait and stability.

Static tests consisted on 2 conditions: (1) static stance and (2) static semi-tandem stance (right and left). Dynamic testes consited on 2 conditions: (3) walk with ASBGo walker and (4) walk alone and/or with an alternative assistive device. In each condition several parameters were acquired (see section 6.2.2.4 and 6.2.2.5). Conditions (1) and (2) consisted on 3 trials with 1 minute of duration each and in conditions (3) and (4) the patient had to walk 20 meters. It is noteworthy that condition (4) is done in order to verify/compare which gait and postural modifications/evolution have been achieved with the ASBGo walker training.

6.3.2.2 Intervention

Despiste being all patients with ataxia, the atheology is different. Given this difference, each patient received a different treatment plan, i.e. the intervention was adapted to the patient's recovery goals. In table 6.1 each intervention is presented in detail. However, some specifications have been designed to be common to all cases. The maximum velocity and time of intervention was set by the physiotherapist. Such velocity and time of intervention were only increased when the patient felt comfortable to do so and set to the limit of his ability to retain adequate control over leg movements [250]. In addition, during the training session visual feedback (Figure 6.2) for foot placement was shown for all patients. A camera integrated in the ASBGo walker films the feet and the image is displayed to the patient. Whenever the patient felt comfortable, this feedback was removed, allowing the patient to independently walk without visual guidance of foot placement.

- Case 1: For three weeks, the patient trained, 5 days a week, his gait with the ASBGo walker. Since he had enough cognitive capacity to guide the walker, such task was handled by him. The first sessions were set to last 15 minutes. In addition to the ASBGo walker therapy, he performed tonus training.

- Case 2: The patient trained assisted gait with the ASBGo walker, twice a week, a total of 40 sessions. 20 sessions were done in addition with hydrokynesis therapy and the final 20 sessions were done without hydrokynesis therapy. Between week 2 and 3, the patient stopped her treatment for 3 weeks. On the first 5 sessions, remote control mode was used to guide the walker, since the patient was not capable to concentrate in guiding at the same time she corrected her gait patterns. Then, since she already had enough cognitive capacity to guide the walker, such task was handled by her. The first sessions were set to last 15 minutes.

- Case 3: The patient was submitted to conventional physiotherapy for four weeks (20 sessions). After that, the ASBGo walker was added to her treatment, 5 days a week. During the first 2 weeks (10 sessions) with ASBGo walker, remote control was used to guide the walker, since the patient was not capable to concentrate in her gait pattern and guiding at the same time. Then, the remaining weeks, manual guidance was set since the patient was already capable to handle such task. The first sessions were set to last 10 minutes.

- Case 4: Before initiating the gait training with the ASBGo walker, her recovery was based on conventional therapy, and her progression had stabilized (she was not capable of walking without help of a third person and demonstrated some static instability). Therefore, the ASBGo walker was precribed to infer if more improvements could be achieved on the recovery of the patient. In a total of 20 sessions, 5 days a week, the patient trained her gait with the ASBGo walker. She had enough cognitive capacity to guide the walker. In addition to the ASBGo walker therapy, she performed conventional physiotherapy but with no gait training.

- Case 5: The patient was submitted to hydrokynesis therapy for 20 sessions. After such sessions, she performed 20 sessions with the ASBGo walker. She had enough cognitive

Table 6.1: Parameters for gait training.

Cases	Week	ASBGo walker training time (min)	ASBGo walker velocity (m/s)	nº of rest breaks	Alternative walking aid (AWD)	AWD training time (min)	Visual feedback	ASBGo walker mode
Case 1	1^{st}	15	0.1/0.2	1	Standard Walker	5	Yes	Manual
	2^{nd}	20	0.3	0	1 crutch/alone	10/5		
	3^{rd}		0.5				No	
	4^{th}	0	-		Alone	20		
Case2	1^{st}	15	0.2	0	2 crutches	2	Yes	Remote control
	2^{nd}	20	0.3					Manual
	3^{rd}	25	0.4				No	
	4^{th}		0.5			3		
	5^{th}		0.6		1/2 crutches	5		
	6^{th}	30	0.7		1 crutch	10		
	7^{th}		0.75					
	8^{th}		0.8					
Case 3	1^{st}-4^{th}	-			walk with a third person	5	-	
	5^{th}	10	0.2	2	-		Yes	Remote Control
	6^{th}	15		1	-			
	7^{th}	20	0.25		-		No	Manual
	8^{th}				-			
	9^{th}	25	0.3	0	Alone with supervision	5		
	10^{th}		0.4					
	11^{th}		0.5			10		
	12^{th}	-				15		
Case 4	1^{st}	15/30	0.3	0	None	5	Yes	Manual
	2^{nd}	30	0.5		Alone with supervision	10	No	
	3^{rd}		0.6					
	4^{th}		0.75		Alone			
Case 5	1^{st}-4^{th}	-			4wheeled walker	-		
	5^{th}	15	0.3	1		Daily	Yes	Manual
	6^{th}	20	0.4					
	7^{th}	30	0.45	0				
	8^{th}		0.5					
	9^{th}		0.5					
Case 6	1^{st}	5	0.1	1	None	-	Yes	Remote control
	2^{nd}	10	0.15	1			No	
	3^{rd}	15	0.25	0				

capacity to guide the walker. No additional therapy was performed.

- Case 6: Due to the acute state of this case, the patient started to stand and walk with the ASBGo walker and 15 sessions were performed. In addition to the ASBGo walker therapy, she performed conventional physiotherapy but with no gait training. Due to bureaucracy reasons, this patient did not finish her treatment.

6.3.3 Statistical Analysis

For each parameter the mean and standard deviations were calculated. Then, One-way ANOVA was performed for each parameter (spatiotemporal and postural stability parameters) in order to verify if there were significant differences through the progression of the patients with the different devices. Then, to verify if the variability of parameters significantly decreased between the beginning and end of treatment, Levene's test (right tail) will be performed. The level of significance was set to p<0.05.

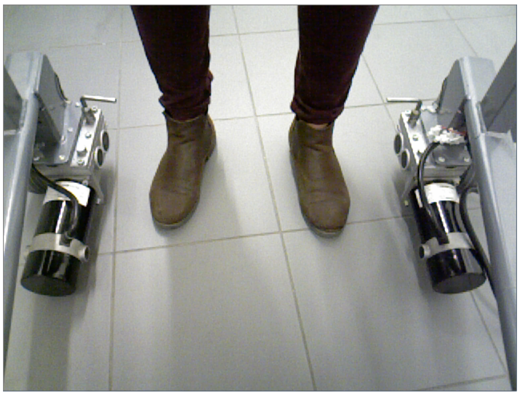


Figure 6.2: Visual Feedback from the legs and feet.

6.4 Results and Discussion

Each patient was evaluated every 5 sessions, which gave us results in terms of qualitative scales, spatiotemporal and postural stability parameters.

6.4.1 Evaluation with the proposed clinical scales

In table 6.2, it can be seen the results of the clinical scales for all patients.

(I) Case 1

In the initial stage (1st evaluation), patient presented a score of 6 points, which means that he had a high risk of falling and was only capable of using a wheelchair to move [182]. At this stage, he needed the assistance of two subjects to stand, to sit and to walk. In one week of training with ASBGo walker, along with tonus training, his score increased to 23 points (2nd evaluation), passing him to the category of medium risk to fall [182]. He started to be capable of climbing stairs with the help of one crutch. At the end of the 3rd week (4th evaluation) he reached 38 points being capable of walking with crutches independently and walk without walking aids when supervised. At this stage, the clinician decided that the patient was capable of leaving the ASBGo walker and continue treatment with two crutches. At the end of his treatment (5th evaluation, 4th week), he presented a BBS score of 42 points, and walked with one or no crutch.

ABC and TUG were not performed by this patient, since, at the time of the experiment, such scales were not part of the evaluation protocol.

Table 6.2: BBS, TUG, ABC and functional gains results for each evaluation. Colored number of evaluation represents the period when the patient used only ASBGo walker for gait training.

Cases	Evaluations	BBS	TUG (s)	ABC(%)	Functional Gains
Case 1	**1st**	6	-	-	-
	2nd	23			Climb the stairs with assistance
	3rd	35			Climb the stairs with 1 crutch
	4th	38			
	5th	42			Climb the stairs alone
Case 2	1st	12	-	52.18	
	2nd	25		-	
	3rd	28			
	4th	31			
	5th	31	76		Confidence to walk with the help of walls; home duties.
	6th	32	60	60	Less tremor to eat.
	7th	22	57	-	-
	8th	30	53		
	9th	30	55/60*		
	10th	34	43*	76	
Case 3	1st, 2nd	4	-	-	-
	3rd,4th	7			
	5th	8			
	6th	10			
	7th	11			Guide the ASBGo walker and correct gait
	8th	27.5			-
	9th	31			
	10th	36			Started to walk alone for some meters
	11th	33			-
	12th	34			
	13th	37			
Case 4	**1st**	30	-	69	-
	2nd	32	33	-	Confident to walk alone
	3rd	43	33		-
	4th	49	12		
	5th	49	11	87	
Case 5	1st	34	24	60	-
	2nd	32	21	-	
	3rd	33	24		
	4th	33	25	55	
	5th	34.5	23	-	Did no experience falls
	6th	41	17		No fall incidents; She was capable of doing home duties with more stability and other new tasks that require some stability.
	7th,8th	41	16		
	9th	44	15	67.3	

*1 Crutch

(II) Case 2

In the initial stage (1st evaluation), the patient presented a score of 12 points, which means that she had a high risk of falling and was only capable of using a wheelchair to move [182]. At this stage, she needed help to stand up from a chair, to sit down and to walk with two crutches. Also at home, no daily tasks were being performed. In one week of training with ASBGo walker, her score increased to 25 points (2nd evaluation, 1st week), passing her to the category of medium risk to fall [182]. Between the 3rd and 4th evaluation, the patient stopped the treatment. However, such interruption did not decrease her score of 31. This results means that the capabilities gained with the ASBGo walker treatment were not lost.

At the 5th evaluation, the patient expressed to be more confident at home, being capable of standing and sitting independently, walking at home and performing home duties. In the next week, she told the physiotherapist that the tremor of her hands decreased, making it easier to eat independently.

From the 6th to the 10th evaluation, it is possible to verify that the BBS stabilized.

In terms of ABC, her confidence increased approximately 18% with the ASBGo walker treatment. With 52% of ABC score, the patient presented a high risk of falling. Since it increased to 76% (>67% [249]) her fall risk decreased significantly. The patient started to gain confidence to walk at home and do some homework that before the treatment she was not capable to do.

The patient only started to perform TUG when she started to be capable of walking with 2 crutches alone and to turn around with them, alone, on the 5th evaluation. Looking at table 6.2, it is possible to observe an evolution in time, where the patient showed improvement in her mobility by performing less time in the TUG task (decreasing from 76s to 43s). At the 9th evaluation, the patient was capable of performing TUG with only one crutch, showing a good evolution from the 9th to the 10th evaluation.

(III) Case 3

In the initial stage, the patient presented a score of 4 points (1st evaluation), which means that she had a high risk of falling and was only capable of using a wheelchair to move [182]. At this stage, she needed help to stand, sit and walk. In the first 20 sessions (5 evaluations) with only conventional physiotherapy, BBS increased to 8 points (5th evaluation). After starting training with ASBGo walker (one week), her score increased to 10 points (6th evaluation). At the 7th evaluation the patient was already capable of guiding independently the walker, and at the 8th evaluation (after 15 sessions with ASBGo walker) the score increased to 27.5, passing her to the category of medium risk to fall [182]. At the 10th evaluation, this patient showed improvements by walking few meters (approximatly 2m) alone. At the end of the ASBGo walker treatment (12^{th} evaluation) she reached 34 points being capable of walking without walking aids and with supervision. At this stage, the clinician decided that the patient was capable of leaving the ASBGo walker (after the 12^{th} evaluation) and she continued treatment walking alone. At the end of her treatment, she presented a BBS score of 37 points (13^{th} evaluation), and was capable of walking alone with low supervision.

Case 3 was not capable of walking without help nor without an assistive device until the last week before being discharged from the Hospital, thus TUG was not performed. As for the ABC, the patient did not want to communicate verbally, so it was not performed.

(IV) Case 4

In the initial stage, the patient presented a score of 30 points (1^{st} evaluation), which means that she had a medium risk of falling [182]. She was capable of passing from a sitting position to a standing position alone, and walk with the help of a third person. Her great progress was observed after 2 weeks of walking with the ASBGo walker, showing at the 3^{rd} evaluation a great increase in BBS score from 32 to 43 points and ending her treatment with 49 points (5^{th} evaluation). At this stage, she was capable of standing alone without supersion and walk alone with no support.

Regarding TUG test, she demonstrated to be more coordinated and stable over time, deacresing her time from 33s to 11s. With ABC, an increase of 18% in condidence was obtained, which is a very positive improvement.

(V) Case 5

In the initial stage, the patient presented a score of 32 points (1^{st} evaluation), which means that she had a medium risk of falling and was capable of using a four-wheeled walker to move [182]. However, this patient experienced daily falls when not using the walker. During the hidrokynesis therapy (from the 1^{st} to the 5^{th} evaluation), the patient did not show any progress on her stability nor on BBS score. After initializing the gait training with the ASBGo walker great improvements were observed scoring 41 points in BBS (6^{th} evaluation). Also, the increase in confidence was remarcable. This can be visualized by the increase in ABC score of 7.3%, passing from high risk of falling (<67% [249]) to low risk of falling. The patient informed us that since she started traning with the ASBGo walker, she did not experience a fall and she was capable of doing more home tasks with confidence and stability. This result is very positive for the ASBGo walker potential, since this patient walks daily with a four-wheeled walker.

In terms of TUG, she decreased from 24s to 15s, which reflects in a better coordination and balance.

(VI) Case 6

During her treatment, she was not capable of standing alone nor supported in order to perform the proposed tests. In BBS she only scored 4 points, corresponding to the “has a straight and independent posture while sitted” question.

6.4.2 Spatiotemporal parameters Results

In each evaluation, the patient had to walk 20 meters while the sensorial system was acquiring the signals for the following calculation of spatiotemporal parameters. The spatiotemporal parameters used in alternative devices were also calculated by using a video-camera and markers on the floor.

(I) Case 1

Throughout the rehabilitation, this patient’s gait has undergone many changes and was subjected to some observations. The gait cycle was evolving very consistently, starting from the time the patient began using the ASBGo walker. Before starting the gait training with the ASBGo walker, the patient had a serious lack of balance, not completing one gait cycle alone. In the stance phase duration (*STAD*) the patient could not get a dorsiflexion of the foot to start the heel contact, since the foot slid with the plant in full contact with the floor. The single foot support was very committed, performing a very little stride length (*STR*) to avoid this support.

When he started gait training with the ASBGo walker his cadence (*CAD*) began to be more consistent, improving significantly (p<0.05), as shown in table 6.3. The patient started to get aware about his gait pattern, because of the feet feedback, and began to make an effort to perform each gait phase more correctly. The ground attack started to be done with more heel. The knee at the phase of "midstance" already with slight bending movement that did not happen without the ASBGo walker. The fact that the patient started to feel safer to walk with the ASBGo walker made it more concentrated on his gait.

Analyzing each spatiotemporal parameter in detail, it was verified that stride length (*STR*) of both legs increased little from week to week making this parameter not significant (p<0.05) because the ASBGo walker influenced its values. Since velocity (*Avspd*) is pre-defined by the physiotherapist (Table 6.3) and the device has dimension limits, the patient is forced to decrease his stride length and maintain it constant. Step length (*STP*) from both legs was initially short and uncoordinated, however, it increased significantly (p<0.05) through time

Table 6.3: Case 1: Velocity and Cadence values for all devices in evaluation: ASBGo walker, standard walker, one crutch and without assistance.

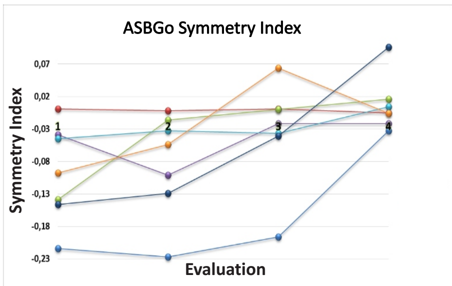
Parameters	**ASBGo walker**				**Standard Walker**		**1 Crutch**		**Without Assistance**		
	1^{st}	2^{nd}	3^{rd}	4^{th}	1^{st}	2^{nd}	3^{rd}	4^{th}	3^{rd}	4^{th}	5^{th}
Avspd (m/s)	0.1	0.2	0.3	0.5	0.15	0.29	0.42	1	0.4	1	1
CAD (step/min)	38	60	60	65	31	44	50	80	50	76	91

for both legs. Gait cycle (*GC*) and Step Time (*STPT*) significantly (p<0.05) decreased since the velocity of gait increased. In terms of step width (*WIDTH*), this parameter increased significantly (p<0.05), which means that the patient increased its base of support, and learned how to walk with a more stable pattern. At the beginning, when the patient walked with the ASBGo walker, he presented a narrow step width (*WIDTH*) and was instructed to extend the width. Thus, the increase in *WIDTH* is a very satisfying result. Regarding gait phases, stance phase duration (*STAD*), swing phase duration (*SWD*) and double support duration (*DS*) the patient improved his pattern by presenting values closed to healthy normal subjects [19], i.e. *STAD* and *SWD* were approximately 60% and 40%, respectively, and DS approximately 20%. The progression of these values was also significant (p<0.05).

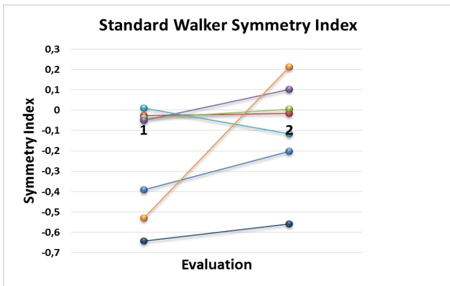
Stride-to-stride variability is an indicator of fall risk and stability of gait [165]. By performing Levene's Test, it was verified that from week to week the variability of all parameters, in three dynamic conditions (ASBGo walker, crutches and walk alone), decreased significantly (p<0.05), meaning that the patient presents an increase in stability and decrease in risk of falling. Other indicator that the patient improved was his gait symmetry. Figure 6.3a presents the gait parameter' results in terms of symmetry index (*SI*) of the evaluations done with the ASBGo walker. As it can be seen, all parameters had a good evolution for the improvement of the patient's gait pattern. Since most parameters present negative *SI* (Figure 6.3a), the left leg is the one responsible for the asymmetric gait. Looking for the evolution of *SI*, one can see that *SI* of all parameters tend to zero week to week.

Meanwhile, the gait training with ASBGo walker allowed the patient to gain enough stability to walk correctly with the standard walker, improving his gait symmetry, in general - *SI* tend to zero - as it can be seen in figure 6.3b. It also allowed him to leave the standard walker and just walk with one crutch. In one week, the patient increased his velocity (*Avspd*) and cadence (*CAD*) with one crutch, started to climb stairs and became more independent, not requiring supervision. His gait symmetry remained almost unchanged, and his spatiotemporal parameters did not change significantly (p<0.05), however his motivation and motor control improved in order to give him enough confidence to try to walk without assistance.

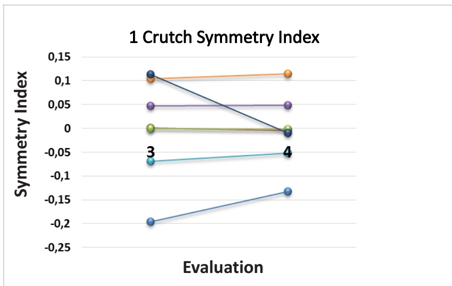
When the patient started to walk alone, he presented a very safe pattern, walking slowly and controlling his movements with good gait symmetry (3^{rd} evaluation, figures 6.3c and 6.3d). In one week, his velocity (*Avspd*) and cadence (*CAD*) increased (Table 6.3), increasing



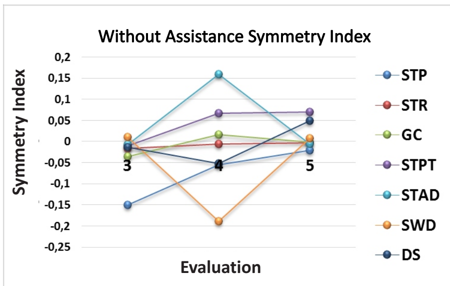
(a)



(b)



(c)



(d)

Figure 6.3: Case 1: Symmetric indices for all devices: a) ASBGo walker, b) standard walker, c) one crutch and d)without assistance.

his asymmetry in gait (4^{th} evaluation, figure 6.3d). However, he learned how to correct his movements, decreasing his *SI* (4^{th} evaluation, figure 6.3d) again, tending to zero.

This quantitative analysis is thus consistent with the aforementioned. observations. In the following link - https://www.youtube.com/watch?v=Txvdjzap12E - improvements in gait pattern can be visualized.

(II) Case 2

When the patient began the gait training with the ASBGo walker, in addition with the hydrokynesis therapy, she was capable of walking with two crutches and with very close supervision. The weight transfer from one limb to the other was compromised since the patient had a tendency to transfer weight the posterior side. At the first assessment (1^{st} evaluation), she exhibited poorly coordinated leg movements, which resulted in abnormal and variable swing foot trajectories and foot placement, increased variability in length and timing of steps (*STP* and *STPT*), slow walking velocity (*Avspd*), profound trunk sway, and an inconsistent base of support (alternating too narrow or too wide). In the stance phase, the ground attack was made with the full support of the foot plant, and the remaining cycle of this phase could not meet the normal percentage of duration (60%) [19]. In the swing phase, the patient presented an over-

Table 6.4: Case 2: Velocity and Cadence values for all conditions.

Parameters	ASBGo walker/Crutch									
	1^{st}	2^{nd}	3^{rd}	4^{th}	5^{th}	6^{th}	7^{th}	8^{th}	9^{th}	10^{th}
Avspd (m/s)	0.1/0.03	0.2/0.03	0.3/0.05	0.4/0.05	0.5/0.08	0.5/0.11	0.6/0.12	0.7/0.1	0.75/0.1	0.8/0.15
CAD (step/min)	22/14	30/16	37/18	37/18	52.5/21	53/18.5	46/19	50/22	50/18	55/24

extension of the knee when decelerating, concluding with a kind of kicking that unbalanced her until the next phase of support.

Thus, in the first week of training, the patient showed marked rigidity in her movements, with evident tremor, few control of movements and narrow base of support (*WIDTH*), when walking with ASBGo walker.

Over the evaluations (1^{st} to the 6^{th} evaluation) with the ASBGo walker gait training, in addition with the hydrokynesis therapy, there was a significant decrease ($p<0.05$) in stride length (*STR*) size, since the patient learned how to position her feet correctly in a controlled way. Step length (*STP*) presented the same improvements (p<0.05). In terms of gait cycle (*GC*) and step time (*STPT*), these parameters presented a good progression with an increase in velocity (*Avspd*), showing that the patient learned how to control her step time (*STPT*). Looking at stance phase duration (*STAD*), swing phase duration (*SWD*) and double support duration (*DS*), until the 4^{th} evaluation the patient presented a high *STAD* with increased *DS*, but tended to the normal percentage (60% of *STAD* and 20% of *DS*). After the 4^{th} evaluation the patient stabilized, presenting values close to healthy normal subjects. Thus, the ASBGo walker gait training started to improve the *SWD* phase. It was visualized that after this evaluation (4^{th} evaluation), the patient started to control the knee extension during swing, having more time and balance to support the heel on the ground in the stance phase, without the need to support the foot plant.

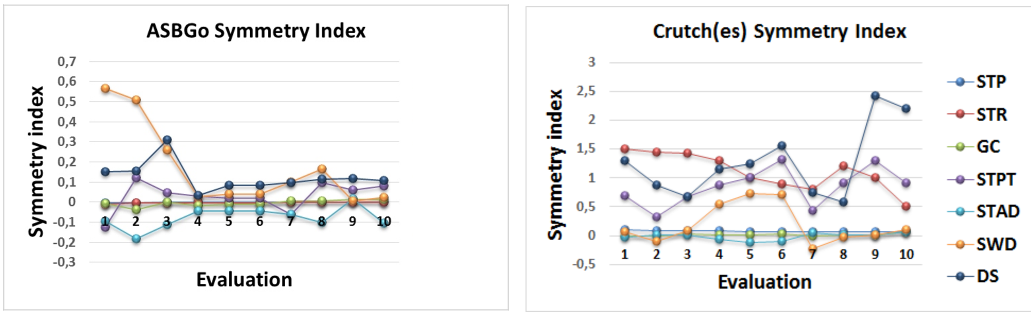
From the 7^{th} to the 10^{th} evaluation, improvements were verified in the aforementioned variables, however it was not significative ($p<0.05$).

Whenever the patient felt comfortable, the velocity of the ASBGo walker was increased. As it can be seen in table 6.4, from evaluation to evaluation the walking velocity (*Avspd*) increased as well as the cadence (*CAD*).

In terms of stride-to-stride variability, Levenes' test indicates that all parameters showed a decrease in variability, in both dynamic conditions (ASBGo walker and crutch(es)) proving that the patient learned how to control her gait pattern. The rigidity of the movement disappeared and the tremor was less evident.

Looking at figure 6.4a, it is obvious that the symmetry (*SI*) tends to zero across the evaluations with ASBGo walker.

These improvements in gait lead to the improvement of her gait pattern when walking with crutches. *SI* was not consistent as it can be seen in figure 6.4b. However, *Avspd* and



(a) (b)

Figure 6.4: Case 2: Symmetric indices for all devices: a) ASBGo walker, b) Crutches (9^{th} and 10^{th} evaluation were done with one crutch).

CAD improved as well as the stride-to-stride variability (*p*<0.05). Such improvements made it possible for the patient to walk only with one crutch from the 8^{th} evaluation.

At the 9^{th} evaluation, the double support (*DS*) *SI* increased because the patient was using one crutch. This increase (Figure 6.4b) is related to the transition from two crutches to one crutch, which led to a greater assymetry.

An important note to be verified is that from the 3^{rd} evaluation to the 4^{th} evaluation, the patient stopped therapy for 3 weeks. Besides this interruption, no significant changes were verified in the gait parameters in consequent evaluations. Showing that the improvements that the patient gained on the first weeks of gait training did not disappeared.

From the 1^{st} to the 6^{th} evaluation the patient performed hydrokynesis therapy and gait training with the ASBGo walker. After that, only ASBGo walker training was performed. The authors observed a big evolution when bothhydrokynesis therapy and gait training was performed. Then, the patient stabilized her gait pattern. However, she was capable of walking faster with good cadence, improve her confidence at home with home duties, and walk with one crutch. Authors think that these two types of therapy make a good complement in the improvement of the state of the patient. It is noteworthy that before using the ASBGo walker, the patient was in a chronic state, not improving her gait performance. After starting the use of ASBGo walker, her state markedly improved, as it was shown.

At the end of the rehabilitation, the patient could walk with less supervision with two crutches and began walking with one crutch. The cadence and velocity continue to be compromised with the crutches, since the patient has to stop to rebalance, but is evident that the gain in confidence improved as well as her gait pattern. In the following link - http://youtu.be/gXRzT1-O4kI - improvements in gait pattern can be visualized.

Table 6.5: Case 3: Velocity and Cadence values for all conditions.

Parameters	ASBGo walker								Without Assistance			
	5^{th}	6^{th}	7^{th}	8^{th}	9^{th}	10^{th}	11^{th}	12^{th}	10^{th}	11^{th}	12^{th}	13^{th}
Avspd (m/s)	0.2			0.25		0.3	0.4	0.5	0.23	0.44	0.41	0.47
CAD (step/min)	81.8	60	52.2	63	60	69	76	84	52	76	70	79

(III) Case 3

In this patient the most difficult task to overcome was the acceptance of weight during swing phase of the contralateral limb. Due to a condition called pushing syndrome, the patient rejected the weight of her right side, presenting a gait with shorter steps and a irregular cadence since the right limb did not accept load during a long time.

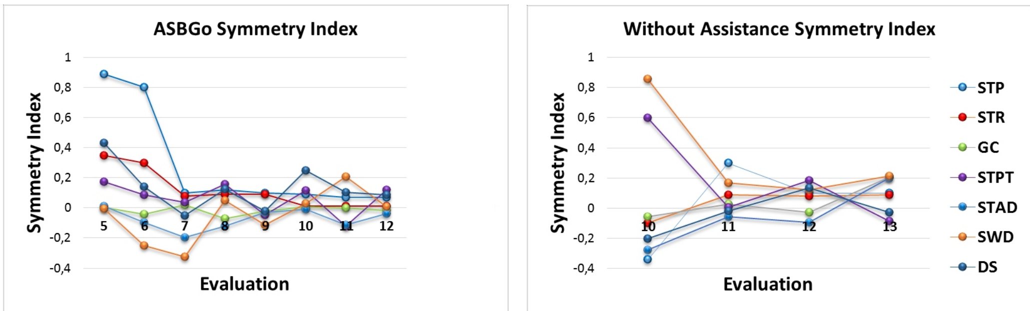
With the ASBGo walker, the patient, initially, performed a narrow base of support (*WIDTH*). She presented a gait with short step length, where the legs met prior to the trunk, because of her support on the elbows in the ASBGo walker and posterior load.

She presented irregular cadence (*CAD*), as seen in table 6.5, causing a high energy expenditure. Her patterns consisted in performing long and short steps (*STP*) with the same low velocity (*Avspd*). As it can be seen in table 6.5, the patient started to walk with low velocity on the ASBGo walker, but her *CAD* was high. Over time, the *CAD* decreased, showing a better coordination of steps for the same *Avspd* (*e.g.* 0.2 m/s). When the patient learned to control her steps, the *Avspd* increased showing a great evolution in her coordination.

In a healthy individual, the stance phase duration (*STAD*) occupies about 60% of the gait cycle, which happened with this patient. However, in terms of time duration, each leg differed from the other. In the left lower limb this phase is higher than in the opposite limb due to the pushing syndrome presented by the patient, which changes the load acceptance on the right leg. This caused a negative symmetry index (*SI*). However, it is noteworthy, that such effect was reduced when walking with the ASBGo walker (figure 6.5a). This can be observed by the "signal" of the *SI*. When walking with ASBGo walker, the signal was positive for *STAD* and negative for swing phase duration (*SWD*), which is the opposite of walking without assistance. Since the patient had an extra support on the elbows and posture correction, the pushing effect was reduced with the ASBGo walker, this can be observed in figure 6.5a. Thus, with the ASBGo walker training, the patient began to be aware of her gait, improving her concentration on the heel contact and the transference between both limbs.

In terms of *SI*, in figure 6.5a, it can be observed that in the first sessions, the patient presented great asymmetry (greater than zero) on her gait, improving over time.

The same happened when walking without assistance (Figure 6.5b). First, she presented a short *STP* with irregular *CAD* and great asymmetry in gait. Over time, she improved her coordination and symmetry.



(a) (b)

Figure 6.5: Case 3: Symmetric indices for all devices: a) ASBGo walker and b) without assistance.

Statistically, all variables are significantly different over time (p<0.05), showing the correct progression. In terms of stride-to-stride variability, all parameters in both dynamic conditions (ASBGo walker and wak alone), except step time (*STPT*) and *CAD*, showed a significant reduction in variability. This proves that the patient improved in terms of gait pattern.

At the end of the ASBGo walker sessions, the patient presented already an independent gait without the need of walking aids. Despite still having an irregular cadence, she was capable of presenting a functional gait, not losing her balance. Such irregularity may be due to the fact that the patient as compromised vision that implies some alterations in her gait pattern.

In the following link - http://youtu.be/V8YL2JoZvUY - improvements in gait pattern can be visualized.

(IV) Case 4

This patient had conventional physical therapy during 1 month and 2 weeks without great evolutions in her balance and gait. She was able to stand alone, with some instability, but presented difficulties to go from sit to stand position. She was not capable and not confident to walk alone, needing a third person to help her walking. Her gait pattern was very unstable, with high cadence (*CAD*), short steps (*STP)*, low velocity (*Avspd*), high step width (*WIDTH*) and crawl feet. Because of her static state (did not show improvements) after the therapy time, her clinician asked to introduce the ASBGo walker on her therapy in order to verify if any evolution was possible on her balance and gait pattern.

She performed gait training with the ASBGo walker daily and significant differences started to appear.

First, her gait with the ASBGo walker was slow (*Avspd*) with high and inconsistent *CAD* (Table 6.6) and short *STP*. Then, over the weeks, the ASBGo walker's velocity was increased, since the *CAD* started to be better controlled, presenting a good gait rythm. She stopped to crawl her feet and started to present a more natural gait. Stance phase duration (*STAD*) and swing phase duration (*SWD*) changed significantly during the first to weeks (p>0.05), chaging from 70%-30% and then these variables started to show a 60%-40% (of gait cycle) behaviour, which indicates that the patient presented a similar behaviour, in terms of temporal parameters, to the healthy subjects' pattern. Double support duration (*DS*) did not change significantly (p>0.05), however its values were always close to normal (20% of gait cycle). Regarding *WIDTH*, the patient had also no atypical behaviour when walking with the ASBGo walker, which reflected on constant values during her rehabilitation. Step time (*STPT*) and gait cycle (*GC*) where very short, even for slow *Avspd*, at the begining. But through the weeks, their values stabilized as well as *CAD* (Table 6.6).

Regarding symmetric index (*SI*) when walking with ASBGo walker, in general, there is an improvement, since almost all *SI* parameters tended to zero. *SWD* was very assymetric, being greater to the rigth side (positive *SI*). This behavior in addition to a greater *STP* for the left side, shows that the patient tend to increase her load to the left side of the walker, showing a greater instability for this side. Since she is more supported for the left side she has more propulsion strenght to move her left leg, performing a higher step length. In terms of *STAD*, there are very few differences (*SI* close to zero), but with a negative tendency, i.e. more supported for the left side.

When walking alone, at first, the patient presented high *CAD*, short *STP*, great *WIDTH* and STAD (>80%). This pattern changed and improved significantly, by increasing her *STP* (p>0.05), deacresing her *WIDTH* (p>0.05) and deacreasing her *STAD* (p<0.05) and consequently increasing her *SWD* (p>0.05). The other parameters did not change significantly. Despite the improvement of such parameters, normal gait pattern was not achieved. Her *Avspd* increased, but *CAD* did not reach desired values for a normal gait, as well as her *WIDTH*.

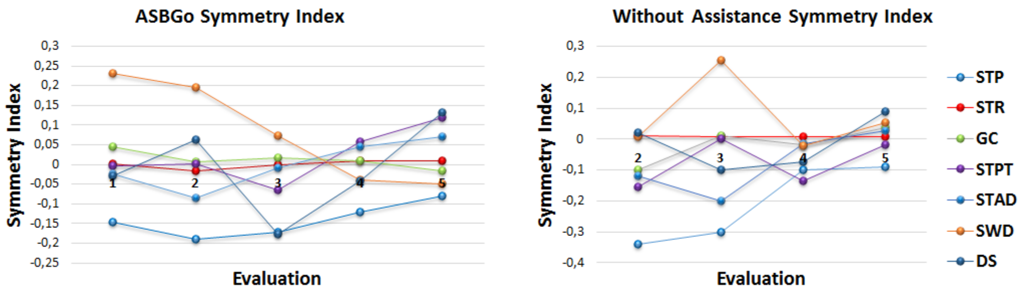
Regarding *SI*, the patient first presented a left assymetric behaviour, but through time, *SI* tended to zero, showing an improvement in coordination to walk alone. It was highlighted through observation that this result may be influenced by other parameters that were not taken into account in this study, such as kinematic parameters (kneen flexion/extension, hip flexion/extension, etc).

After obtatining enough independence and confidence to walk alone without any help, the patient stopped using the ASBGo walker, to continue her physical therapy in order to correct her pattern. However, such correction may not be attainable, because of her ataxia. Some disturbances are not possible to be recovered.

In the following link - https://youtu.be/IRRLEiu1YjI - it is possible to visualize the described evolution.

Table 6.6: Case 4: Velocity and Cadence values for all conditions.

Parameters	ASBGo walker					Without Assistance			
	1^{st}	2^{nd}	3^{rd}	4^{th}	5^{th}	2^{nd}	3^{rd}	4^{th}	5^{th}
Avspd (m/s)	0.3	0.3	0.5	0.6	0.75	0.19	0.2	0.54	0.66
CAD (step/min)	46.6	44.7	40.9	69	78	78	100	95	105



(a) (b)

Figure 6.6: Case 4: Symmetric indices for all devices: a) ASBGo walker and b) without assistance.

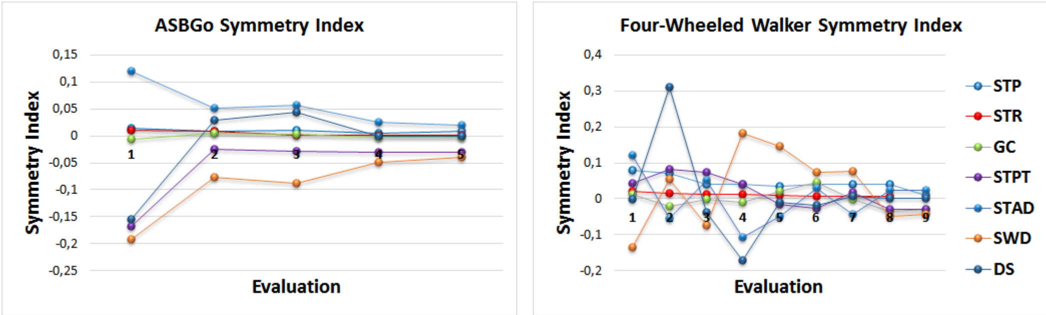
(V) Case 5

This patient was subjected to hidrokynesis during 20 sessions, being evaluated every 5 sessions. After those sessions, 20 sessions only with ASBGo walker were performed. One important detail about this patient is that she is a daily four-wheeled walker user and her goal was to improve her balance and confidence in walking safely. Her gait pattern presented normal velocity (*Avspd*) and cadence (*CAD*) as it can be seen in table 6.7. Velocity of ASBGo walker was low, but it was not safe to increase it, because of the safety of the four-wheeled walker. Because of this, no diferences were observed in *Avspd* and *CAD* when walking with the four-wheeled walker. As the patient was used to walk slowly, she was not prepared to walk faster (*Avspd*), becoming tired quickly. Over time, with the ASBGo walker training, she improved her physical condition, being capable of walking faster during more time and with normal *CAD*.

Regarding the gait pattern, her stride and step length (*STR* and *STP*) did not change significantly (p>0.05) during hidrokynesis nor ASBGo walker training. Also, gait phases and durations (stance phase duration, *STAD,* swing phase duration, *SWD,* double support duration, *DS,* step time, *STPT* and gait cycle, *GC*) did not change significantly (p>0.05). Only step

Table 6.7: Case 5: Velocity and Cadence values for all conditions.

Parameters	ASBGo walker									Without Assistance				
	1^{st}	2^{nd}	3^{rd}	4^{th}	5^{th}	6^{th}	7^{th}	8^{th}	9^{th}	5^{th}	6^{th}	7^{th}	8^{th}	9^{th}
VEL (m/s)	0.3	0.28	0.3	0.3	0.31	0.37	0.28	0.33	0.31	0.3	0.4	0.45	0.5	0.5
CAD (step/min)	58	55	52	50	50	50	51	51	52	43.7	45	46.1	48	49



(a) (b)

Figure 6.7: Case 5: Symmetric indices for all devices: a) ASBGo walker and b) 4-wheeled walker.

width (*WIDTH*) presented a significant increase in its value (p<0.05), during the time of ASBGo walker gait training, which allowed the patient to walk more safely and stable. Another achievement with this increase in *WIDTH* was the fact that the patient stopped crossing her legs and shocking one foot against the other.

In terms of symmetry index (*SI*), the patient presented lower strenght in propelling her motion when using the right leg. This caused a larger assymetry for in terms of *STAD, SWD* and *DS*. Over time, mainly after starting ASBGo walker training, such assymetries tended to zero, as it can be seen in figure 6.7a. Such improvements were also visualized when walking with the four-wheeled walker.

In addition to the improvement of *SI*, variability of gait parameters also deacreased, reflecting on her greater stability.

Since this patient has a degenerative condition, the results achieved with the use of the ASBGo walker are remarkably good. In the following link - https://youtu.be/N40tf1KZ6FQ - it is possible to visualize the described evolution.

(VI) Case 6

During the short time of treatment this patient, despite presenting a great evolution, was still not capable of walking independently with the ASBGo walker nor to stand alone. A harness was necessary to impose a straight posture while walking with the ASBGo walker and remote control was used as ASBGo walker mode (Figure 6.8). At the begining, she was not capable of moving her feet correctly to walk, needing the help of the physiotherapist. After one week, she started to present more independence, by moving her feet alone and presenting a correct cadence.

No data was acquired with this patient because of her acute state.

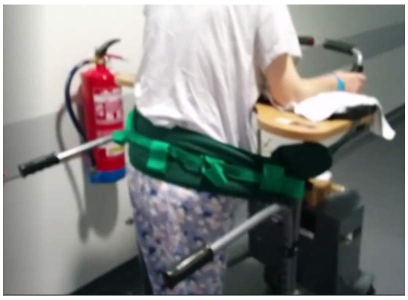


Figure 6.8: Case 6 walking with ASBGo walker with the support of a harness.

6.4.3 Postural Stability Results

Postural stability parameters were calculated during static and dynamic positions (section 6.2.2.5). COM displacement was acquired for all conditions (CS, SSL, SSR and ASBGo walker). In order to have a better visualization of the evolution, in time, of the patient in terms of stability, the COM displacement was approximated to an ellipse. Taking the outside margins of the COM displacement, an ellipse was drawn, as illustrated in figure 6.10, 6.12 and 6.14.

It is noteworthy that no "healthy individual" signal was used as reference. Also, no comparasion was made between cases. The goal of this study was not too achieve a specific balance nor trajectory of COM displacement. The goal was to perform an intra-individual evaluation in order to verify the evolution of the case itself, comparing the different evaluations in time for the same case.

(I) Case 1

In figure 6.9 the studied conditions are illustrated with the patient involved in the study. All mean values of postural parameters presented a significant decrease (p<0.05) through all conditions of evaluation. Also, the variability decreased significantly (p<0.05) for all conditions through the weeks. This result is very satisfying since it means that the patient progressed week to week, gaining more and more stability to walk, decreasing his risk of falling.

In figure 7 it is shown the COM ellipse displacement for all conditions. On all conditions, it can be seen that the patient had a large medial-lateral displacement, meaning that he presented a lateral displacement that could cause his instability while walking. It was observed that, when walking alone, the patient tend to fall to one of the lateral sides. However, this instability was reduced over the weeks, allowing the patient to better control his posture while walking with ASBGo walker and alone. It is noteworthy, that the ML displacement has reduced more than AP, showing greater improvements.

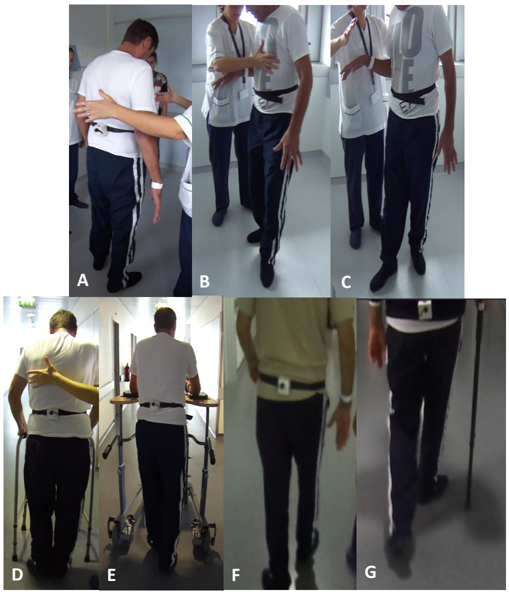


Figure 6.9: Postural stability evaluation tests with the patient of case 1: A- Comfortable stance (CS); B- Left semi-tandem stance (SSL); C- Right semi-tandem stance (SSR); D – Standard Walker; E- ASBGo walker; D- Without Assitance; and G – One Crutch.

In all cases the ellipses decreased their radius. This result comes to reaffirm the gain of stability presented by the patient through this rehabilitation.

Therefore the patient presented a constant evolution of his balance and posture, helping him to improve his gait pattern. He ended the gait training with low risk of fall and more confidence and attention to his gait.

(II) Case 2

In figure 6.11 the studied conditions are illustrated with the patient in the study. All mean values of postural parameters presented a significant decrease (p<0.05) through all conditions of evaluation, over all weeks, except right and left semi-tandem stance. By observation, such conditions improved, however the patient still demonstrates some instability, despite being independently positioned and some supervision is being required. All the other conditions, showed improvement through all weeks with the two types of treatment.

Variability also decreased significantly (p<0.05) for all conditions over the weeks, showing that the patient progressed week to week, gaining more and more stability, decreasing her risk of falling.

In figure 6.12 it is shown the COM ellipse displacement for all conditions performed by the patient. The dashed line presents the results of the patient doing hidrokynesis therapy and ASBGo walker. The continuous line represents the evaluations of the patient using only the

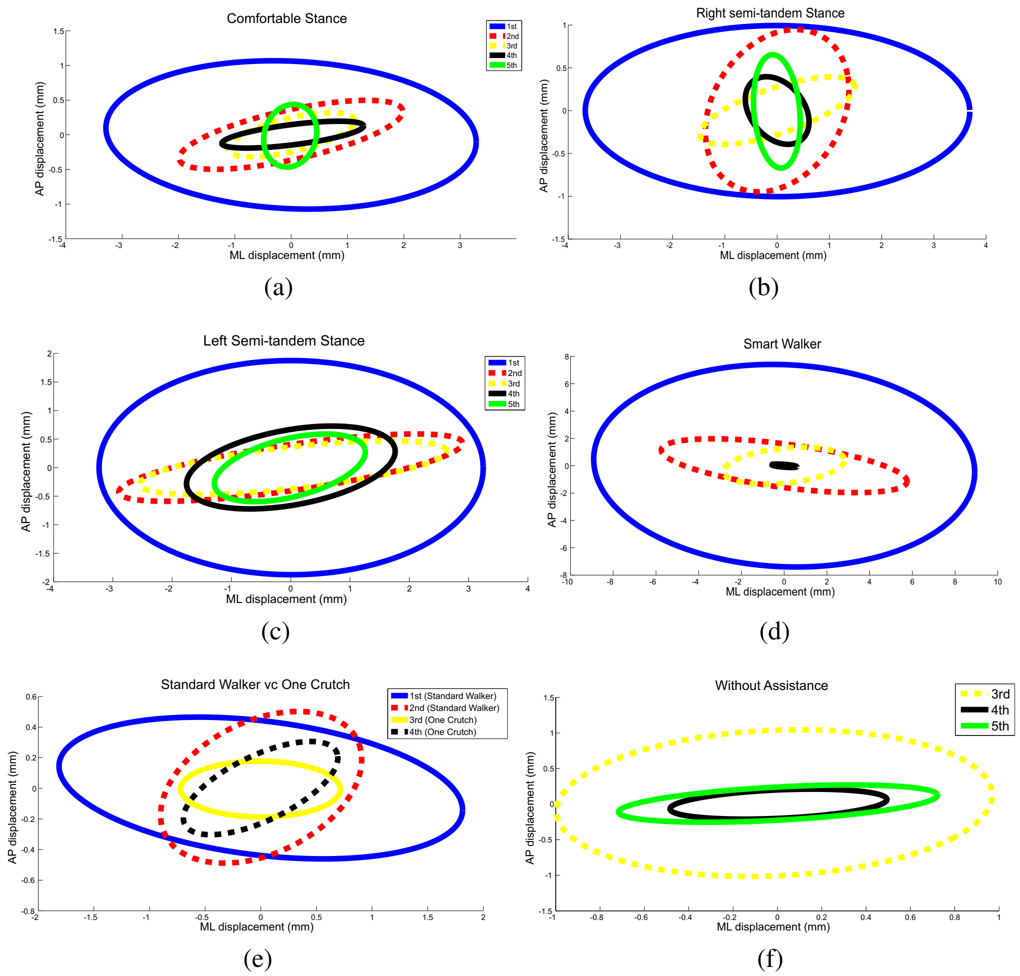


Figure 6.10: Case 1: ML and AP COM displacement in (a) comfortable stance, (b) right and (c) left semi-tandem stance and (d) walking with ASBGo walker ASBGo walker, (e) standard walker, Crutch and (f) without assistance.

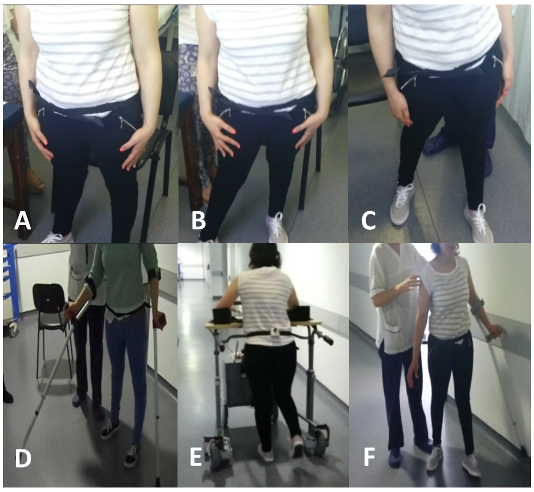


Figure 6.11: Postural stability evaluation tests with the patient of case 2: A- Comfortable stance (CS); B- Right semi-tandem stance (SSR); C- Left semi-tandem stance (SSL); D – Walk with two crutches; E- ASBGo walker; F – One crutch.

ASBGo walker with no extra therapy. It can be seen that during the first therapy combination, there was a significant improvement in COM displacement. Then, only with ASBGo walker, the two static conditions, left and right semi-tandem stance, presented a little reduction, not being so significant.

This patient had less stability to when supported to the left side and in comfortable stance, which can be observed on the SSL and CS graphics by comparison with SSR. Because of this, when walking with one crutch, such device was used on the left side by the patient to give her more stability. In all static conditions, the greater improvement is visualized in the medial-lateral displacement. In the dynamic conditions, she showed a similar improvement in both directions. This resulted in a greater stability and postural control while walking with assistance of walking aids. The fact that the ASBGo walker presents a constant and controlled velocity and supports the patient in a way that she does not unbalance to the posterior side, caused the patient to take conscience that the center of gravity has to be on the anterior side, to walk safely.

(III) Case 3

In figure 6.13 the studied conditions are illustrated with the patient in study. All mean values of postural parameters presented a significant decrease (p<0.05) through all conditions of evaluation, when the ASBGo walker was included in the treatment. Before that, there were not significant changes in the parameters. Variability also decreased significantly (p<0.05) for all conditions over the weeks, only when using the ASBGo walker. This result is very satisfying

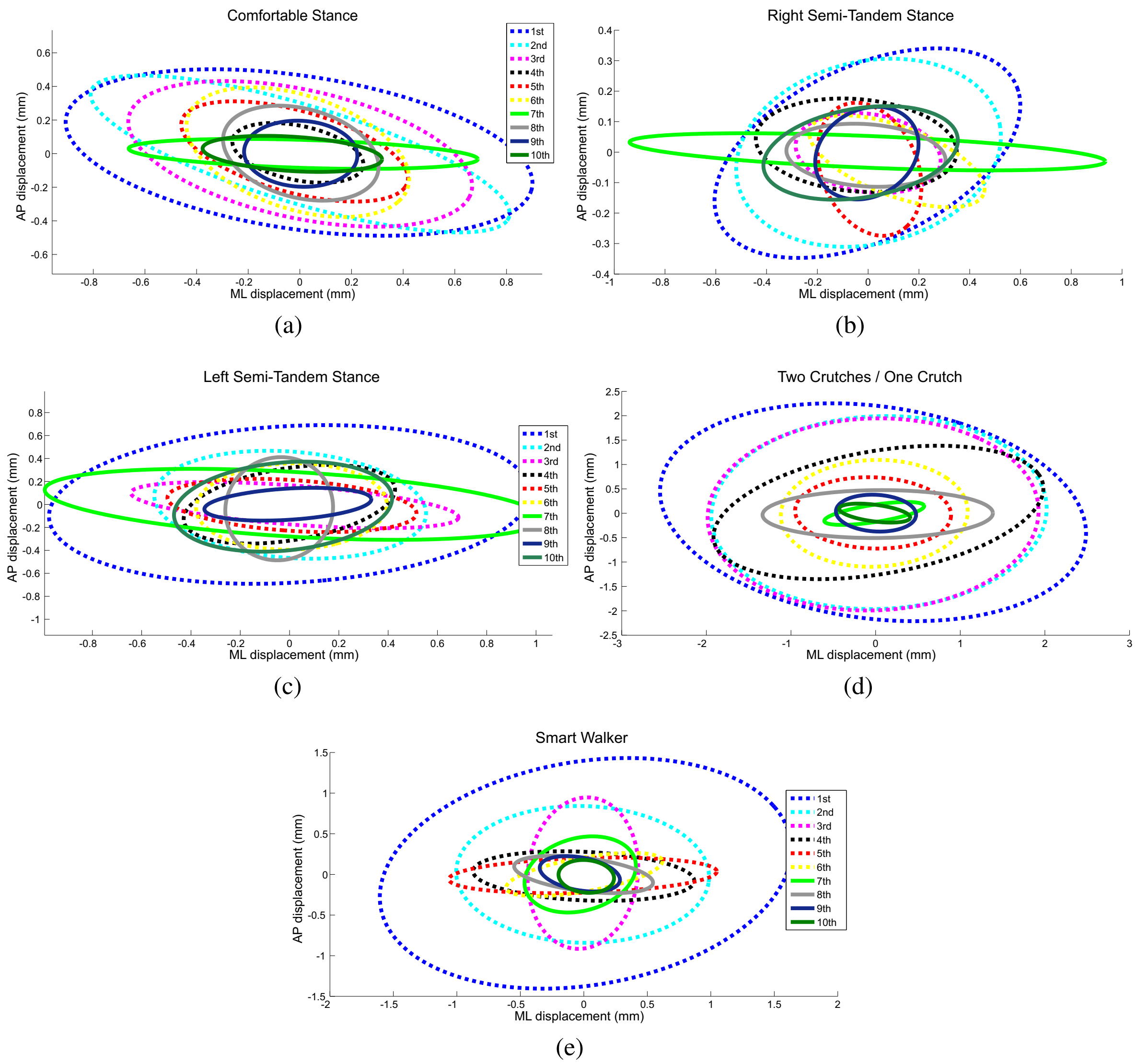


Figure 6.12: Case 2: ML and AP COM displacement in (a) comfortable stance, (b) right and (c) left semi-tandem stance, walk with (d) two/one crutch and with (e) ASBGo walker.

Figure 6.13: Postural stability evaluation tests with the patient of case 3: A- Comfortable stance (CS); B- Left semi-tandem stance (SSL); C- Right semi-tandem stance (SSR); D – Walk with ASBGo walker; E- walk without assistance.

since it means that the patient progressed week to week, gaining more and more stability to walk, decreasing her risk of falling.

In figure 6.14, the COM ellipse displacement is shown for all conditions performed by the patient. The dashed line presents the results when the patient was doing only conventional therapy. The continuous line represents the evaluations while the patient was using the ASBGo walker.

Initially the patient had no balance in static position, always leaning to the left due to the rejection of the weight on the right. It can be seen that during conventional therapy there was not a significant improvement in COM displacement. Reduction is visualized after the treatment with the ASBGo walker began, since it might helped the patient to maintain her balance and concentrate on her gait. Then, when the patient started to stabilize on the COM radius (after the 10^{th} evaluation), in the12^{th} evaluation, the patient left the ASBGo walker treatment, doing only gait training without assistance.

In detail, one can see that the patient showed a similar displacement in all static conditions, presenting a greater instability in the medial-lateral direction. Since the heel support was not apparent at the stance phase (she put her foot plant entirely on the ground) and the hip was rotated externally in every phase of the gait cycle, the anterior-posterior load on the swing phase was compromised, causing the patient to oscillated lateraly (medial-lateral direction). This instability led the patient to be more likely to fall sideways. Fortunately, this trend decreased during therapy, showing a great reduction in the medial-lateral displacement.

(IV) Case 4

In figure 6.15 the studied conditions are illustrated. All mean values of postural parameters presented a significant decrease (p<0.05) with ASBGo walker and CS condition. Walk without assistance, SSL and SSR presented a decrease in the postural paramters, but it was not significant (p>0.05). Variability decreased significantly (p<0.05) for all conditions over the evaluations, showing a gain in stability.

In figure 6.16, it is possible to observe that this patient presented an instable behavior in CS and on dynamic conditions. On the opposite, she was stable in SSR condition, which explains the little deacrease in COM displacement. The SSL condition was the one with lower progression to recovery, despite the low values. Right side stability showed a greater improvement over the left side. This difference in recovery was observed when the patient walked, having the tendency to fall to the left.

In terms of static stability (CS), this patient showed very good improvements which coincides with her BBS score. When walking with ASBGo walker the patient also showed a great improvement. However, when walking alone the same did not happen. Despite the decrease of COM displacement radius, the patient continued to present an instable gait. It is true that she was capable of walking alone, with no support, however her gait was far from presenting a healthy pattern (i.e. she continued to present an “ataxic” pattern).

(V) Case 5

In figure 6.17 the studied conditions are illustrated.

All mean values of postural parameters in medial-lateral (ML) direction presented a significant decrease (p<0.05) in all conditions after begining to train with ASBGo walker. In anterior-posterior (AP) no significant changes were observed (p>0.05). Variability decreased significantly (p<0.05) for all conditions over the evaluations, showing a gain in stability. This decrease was reflected in the deacrease of fall risk of the patient. This patient was used to fall every day and after the ASBGo walker gait traing, such event became rare or inexistent.

The improvement of stability in ML direction can be also visualized in figure 6.18. In all conditions there is a significant reduction in the radius of the COM displacement in ML direction, mainly after begining the ASBGo walker training.

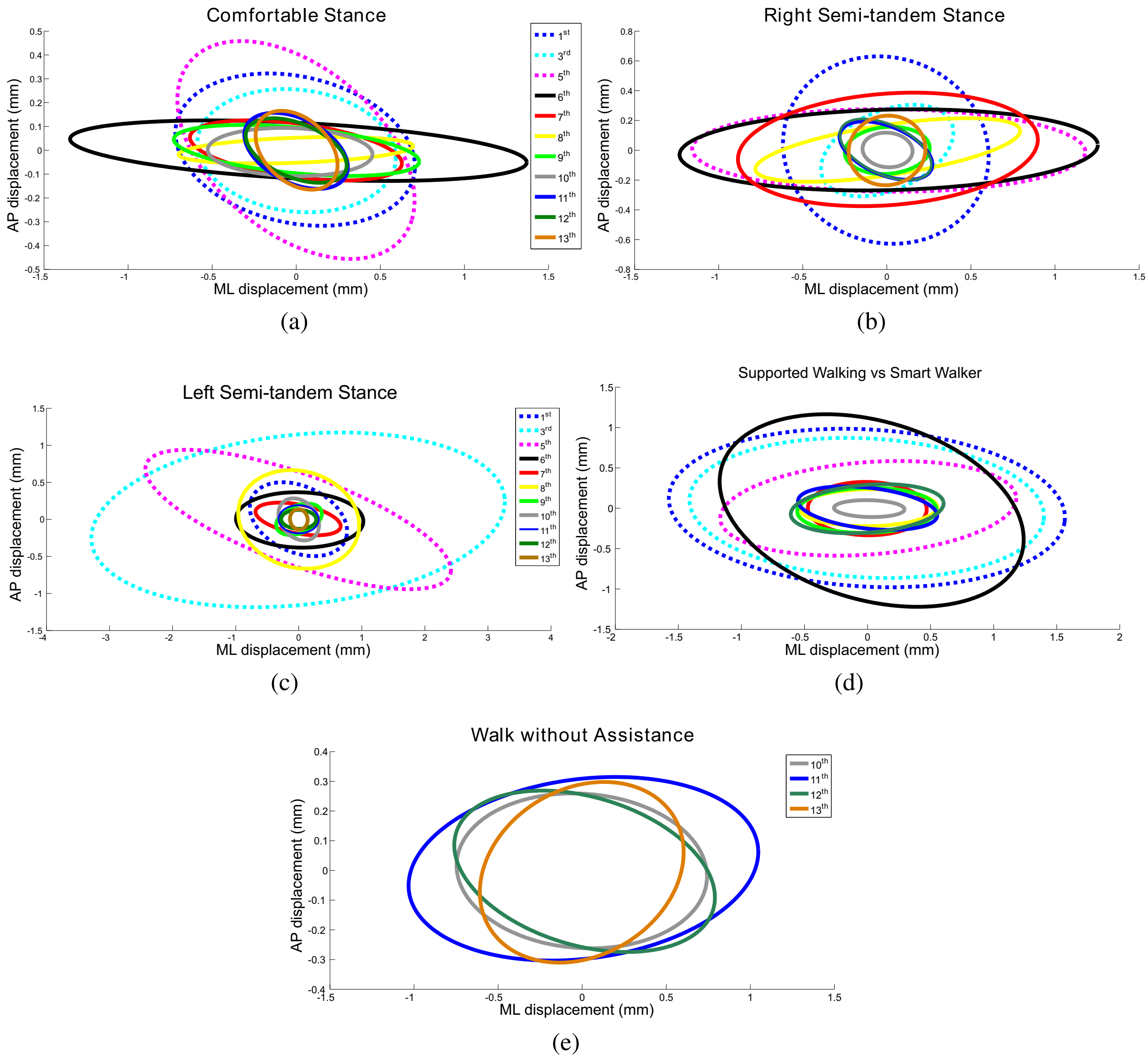


Figure 6.14: Case 3: ML and AP COM displacement in comfortable stance, right and left semi-tandem stance and walking with ASBGo walker/supported walking and without assistance.

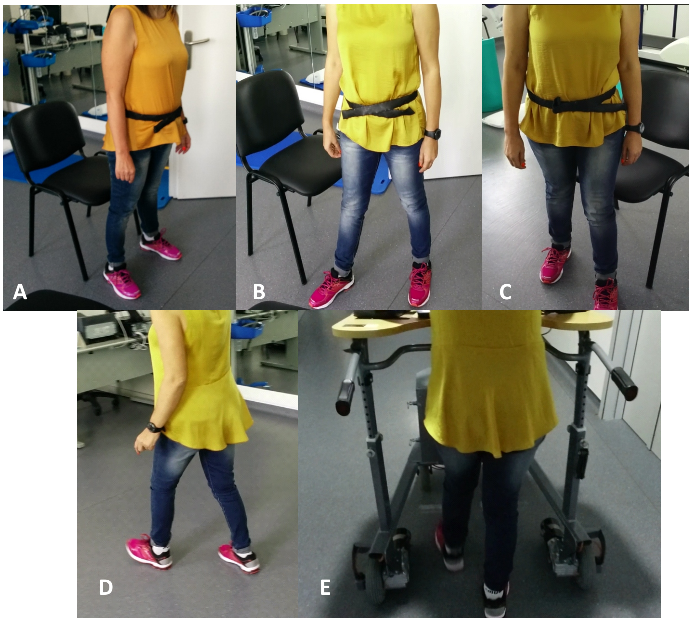


Figure 6.15: Postural stability evaluation tests with the patient of case 4: A- Comfortable stance (CS); B- Right semi-tandem stance (SSR); C- Left semi-tandem stance (SSL); D- walk without assistance; E – Walk with ASBGo walker.

6.4.4 General Discussion

In this study, six different patients with ataxia performed gait training with ASBGo walker. Different improvements, in different recovery times, with different functional gains were achieved by the patients. However, similar measures and protocol were performed. ASBGo walker velocity was predefined by the physiotherapist and it was very important for the patients' gait training. This type of patients tend to have a very inconsistent velocity, presenting many accelerations and decelerations. The constant velocity obliges them to maintain the consistency of their gait. Despite not being the maximum velocity that they were capable of walking, the physiotherapist wanted to force them to control their velocity.

Each patient was subjected to different interventions depending on the presented condition. By this, ASBGo walker was compared with different interventions, to test its potential. Case 1 and case 6 only performed gait training with ASBGo walker, with successful results. Case 2 intervention was divided into hidrokynesis aside with ASBGo walker training and ASBGo walker training alone. On both type of interventions evolution was observed. Case 3 perfomed first conventional therapy and then ASBGo walker training. Before iniating ASBGo walker training, her state was improving, but slowly. After ASBGo walker training, her recovery improved very quickly, showing a great increase in BBS score. Case 4 was in a stagnated state, not improving her condition with conventional therapy. After initiating ASBGo walker

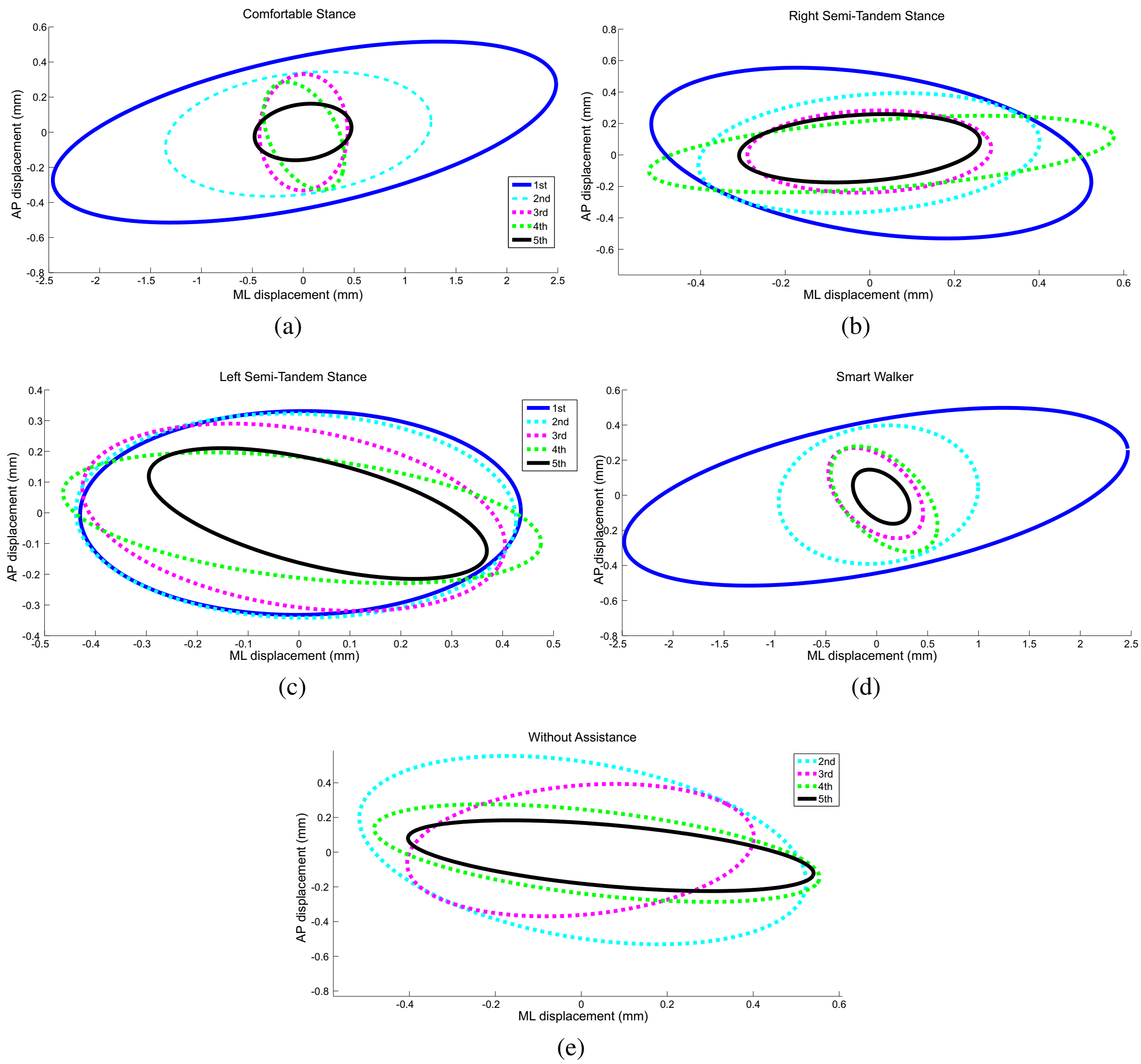


Figure 6.16: Case 4: ML and AP COM displacement in comfortable stance, right and left semi-tandem stance and walking with ASBGo walker and without assistance.

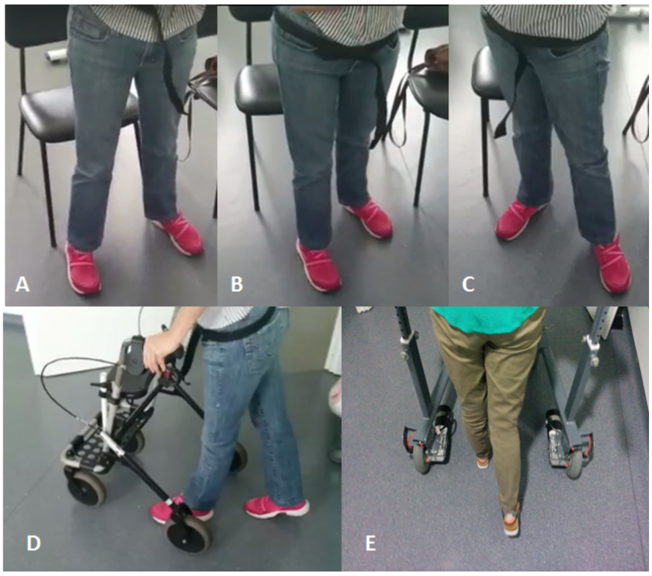


Figure 6.17: Postural stability evaluation tests with the patient of case 5: A- Comfortable stance (CS); B- Right semi-tandem stance (SSR); C- Left semi-tandem stance (SSL); D- walk with 4-wheeled walker; E – Walk with ASBGo walker.

training, her balance and gait pattern started to improve. Case 5 began with hidrokynesis therapy, not showing improvements, and then with ASBGo walker training her balance and confidence increased a lot. Case 6 proved that the ASBGo walker can be used during acute stages of therapy, allowing the possibility for the patient to start gait training earlier.

In terms of postural control, most patients showed a great instability on the medial-lateral direction, having tendency to fall sideways. This was identified by the COM displacement method, which predicted the lateral fall risk. Then, instability was reduced with the use of ASBGo walker, which was also showed by COM displacement method.

It is believed that a number of factors contributed to the outcomes, including the motivation of the patients and the support and dedication of their family.

Symmetry index, stride-to-stride variability and COM displacement were considered the best outcomes to evaluate the evolution of these type of patients, giving quantitative information about their improvements.

Results from these case studies suggest that the ASBGo walker gait training is a promising intervention for improving gait in patients with cerebellar ataxia. Findings from these cases provide possible support for research demonstrating the importance of cerebellar structures in gait adaptation and in practice-dependent motor learning. The correct intensity and duration of gait training using ASBGo walker to achieve functional gains is not known nor standardized. In addition, it is not clear how much additional therapy is necessary to achieve these outcomes. Studies are needed to determine the optimal intensity and duration of gait training

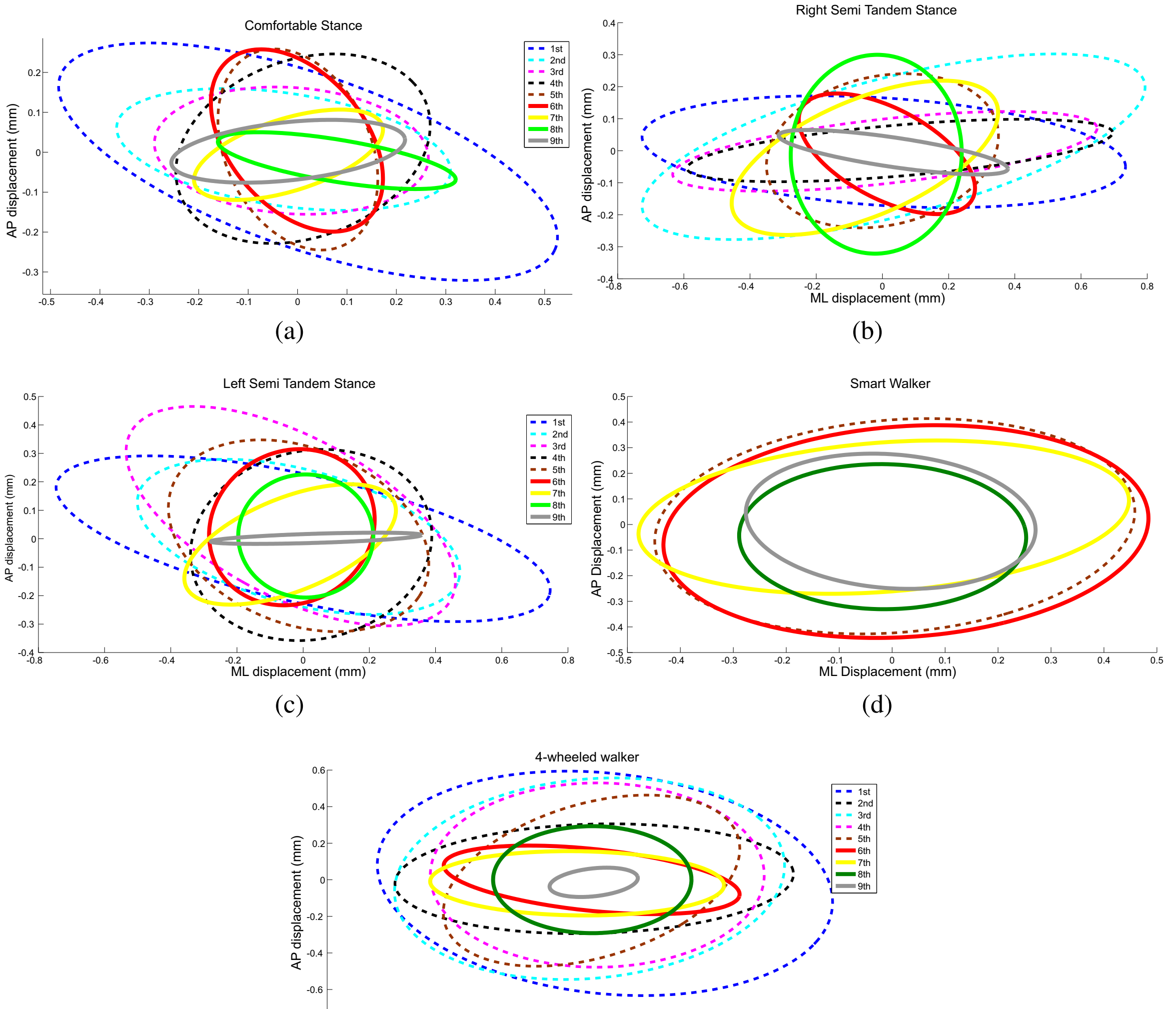


Figure 6.18: Case 5: ML and AP COM displacement in comfortable stance, right and left semi-tandem stance and walking with ASBGo walker and with 4-wheeled walker.

to optimize functional walking outcomes following cerebellar pathology. The team believes that it depends on the patient state and motivation, as it was mentioned before.

6.5 Conclusions

There was a marked improvement of balance and gait pattern objectified by the application of scales and the parameters collected through sensors inserted in the walker and patient. Given this increased stability, the gait speed was gradually augmented. The clinical improvement observed in the evolution of the patients was very satisfying, encouraging the author to extend this study to similar clinical features. Findings of this study show that gait training equipment can be improved and, consequently, better functional gains can be achieved if quantitative methods for evaluating walking performance are developed, such that it can be possible to establish baseline training parameters for each patient and then progress each patient in an optimal recovery. Until now, quantitative assessments were expensive and required large labs to be performed. With this study it is shown that such a fact it is not necessarily true anymore. The instrumentation now available with gait-training robotics makes quantitative methods possible. This study was the first step for including ASBGo walkers in the rehabilitation program of patients that need this kind of help. We hope with this study to incentive investigators to also include their prototypes into the rehabilitation of more patients with ataxia, Parkinson, stroke, etc.

Chapter 7

Conclusions and Future Work

A major problem of those affected to the musculoskeletal and neurological system level, relates to the fact that they do not trust in conventional walker, tending to resort to a wheelchair and/or bed. In this way, they lose all the residual capacity of movement and suffer the long-term disadvantages of these solutions, affecting other physiological systems. These solutions tend also to exclude people from society and isolate them, reducing their quality of life.

There is a strong need for alternatives to conventional walkers in the market that can provide efficient help to a small population that needs extra support to walk, and does not have.

Therefore, the main goal of this thesis was to develop an motorized walker to improve the stability of assisted gait of people with high balance problems in hospitals and clinics, recovering/giving them the necessary motivation for their rehabilitation. This device will enable those people to have a proper and adapted treatment, integrating them into society, improving the independence and motivation by giving them the opportunity to live their life and have access to health care and adequate technical support.

The proposed smart walker prototype will provide a new opening in assistive devices market. This device fills many of the problems presented by conventional walkers, which despite its great rehabilitation potential, have been loosing its users, since many fall while using it.

There are already, as mentioned in the state of art, studies and prototypes of smart walkers, however, there are still a number of problems to be solved for this type of device before it can actually go out to the market and be accepted by the user.

In this work, an “end-user” approah was considered and studied in order to find which are the main requirements of this type of device by the ones that really need them and by the ones who work with it (physicians and physical therapists). Many questions arose in terms of design, adaptation of the device to its user treatment, integration of a gait assessment tool for a better evaluation of the user clinical state as well as the integration of sensors to monitor user’s safety and assisted gait. These issues were discussed with healthcare professionals and handled in detail in this thesis.

This chapter summarizes the contributions of this thesis and future directions of research.

7.1 Main Findings and Contributions

In a first stage, an exaustive survey about conventional and smart walkers was performed in order to highlight the strengths and weaknesses of this area. Many problems were found in terms of defining objective and standard outcomes to assess rehabilitation effects on walker-users. Also, there is no specification about rehabilitation programs that involve smart walkers and their effectiveness. Therefore, there is a strong need of team work involving clinicians and researchers to undertake clinically relevant research ideas so that they can prescribe the correct technology for their patients.

This stage allowed to answer to Goal 1 and to define the next goals of this thesis, strengthening the position that the developement of smart walkers need to be focused in the "end-user" and validated as rehabilitation or functional compensation tools.

Thus, regarding to Goal 2, the design of a new smart walker was proposed taking into consideration end-users's and medical staff's concerns. A device struture with adaptive capabilities was designed, being suitable for a high range of users in terms of their body measures, physical and cognitive capabilities and treatment. Also, a smart interface based on a mechanic handlebar adapted to the user was developed and forearms supports were added to give more stabilization to the user. Different operation modes and functionalities were discussed and defined to be integrated on the proposed smart walker in order to turn it onto a adaptative and versatile device that is capable to answer to the different user's needs of treatment. The first steps were made and presented.

In terms of operation modes, only the concept was defined and further studies were established in order to develop it.

The functionality that was handled in detail, as defined in Goal 3, was the development of a gait assessment system integrated in the smart walker in order to provide an objective tool for the medical staff to evaluate the effects of the rehabilitation treatment on the patient. Different systems were analyzed and new algorithms were developed. A first system, based on an active depth sensor and a laser range finder sensor, was used to capture, in real time, the relative evolution of the lower limbs. The respective algorithms were developed, tested and validated with end-users. Also, a fusion with these sensors was performed in order to reduce measure errors. The lower-limbs motion provided spatiotemporal information of gait, such as speed, stride length, etc, and orientation of the feet to infer motion, such as turning to the left or right, with low measure errors. Another system based on an accelerometer was used to acquired the movement of the trunk. Algorithms for such detection were based on known algorithms in the literature and estimation of the centre of mass displacement in order to infer balance and posture was performed. These two systems provided the necessary information to acquire many important gait and posture parameters defined in the literature and consider relevant to perform gait analysis. Other system was also developed, in order to address Goal 4, where the safety of the user is monitored and infered. A state-machine based on a fuzzy supervisor, enables the smart walker to deal with dangerous situations that may occur during assited gait.

These systems are an important contribution for the smart walkers research since they enable the smart walker with the possibility of being a gait parameters' and posture's acquisition tool, in real-time, in order to be used in a rehabilitation program.

This equipped smart walker enabled new findings to answer the main research questions of this thesis and address the remaing goals. Chapters 2 to 4 created the support to build those findings and then chapters 5 and 6 present such findings.

First, preliminary studies were performed in order to characterize the assisted human gait, as defined in Goal 5, through the processing of the information obtained by the gait assessment systems. It was demonstrated that with the use of multivariate analysis approaches and multi-classification one can assess differences and similarities in gait performance between different assistive devices in order to improve the quality of prescription of such devices.

The first research question (RQ1) was how do walkers with forearm supports influence and modify the walking gait pattern and posture of their users, in comparison with crutches and standard walkers? In chapter 5, patients subject to total knee arthroplasty (TKA) were selected to answer this question. Walkers with forearm supports provide a symmetrical gait, with continuous movement. All spatiotemporal parameters showed a significant improvement when the patient walked with such walker. However, in terms of posture some attention has to be highlighted. Despite this walker stabilizes the position of the trunk, it forces some flexion of the trunk which may be a long-term problem. Because of this, special attention in walker's height has to be considered.

There are differences between the gait parameters of TKA patients depending on the used device. This divergence between the assistive devices may be due to the type of motion pattern presented while using each device. When feeling discomfort on the operated leg, the patient overcompensates its movement, by changing its pattern. In general, it was found that the walker with forearm supports provides a proper support and functional compensation on the studied patients, because even in a debilitated condition, patients were able to walk with the assistance of this device with a natural and faster gait pattern. In contrast, standard walkers tend to induce a slower gait and crutches induce an asymmetric gait. Depending on the results that the physical therapist wants to achieve, patients that are recovering from TKA surgery can use this walker during the recovery in order to present a more natural gait, not putting unnecessary effort on the non-operated leg. From the postural and fall-risk parameters it is possible to conclude that there is a reduction on the fall-risk and an increase in stability with standard walker. Thus, in general, standard walker provides a stable gait. This means that if the patient presents a more debilitated state in terms of stability for walking, the standard walker device should be prescribed.

Looking more specifically to the characteristics that differentiate these devices, it was found symmetry, center of mass range of motion, sway lenght, center of mass displacement and acceleration. Moreover, these parameters may provide complementary information to gait velocity, cadence and clinical scales to assess the functional capacity of patients that passed through TKA, which are the conventional outcomes of the walkers evaluation.

Regarding RQ2, if is there a need to individualize the gait pattern evaluation in order to prescribe a walking aid? Commonly, a device is prescribed according to the general diagnosis that the patient has. However, it is believed that each patient differs from the other, even if he/she presents the same diagnose as other patient. In chapter 5, it was demonstrated that an individual approach and evaluation needs to be done, before prescribing an assistive device. Such understanding needs to be further studied, but a new and promising methodology was proposed in here. These findings might help in physical therapy.

Finally, the proposed smart walker was clinically validated through experiments with patients with ataxia in a rehabilitation program, as defined in Goal 6. This validation answered RQ3, can the smart walker with forearms be prescribed as a rehabilitation tool to correct specific gait disorders (ataxia)? It was verified that the proposed smart walker may be prescribed as a rehabilitation tool to correct ataxia. A strong positive feedback was gathered from this validation, where the quality life of six patient was changed and improved. At the end, all patients wanted to purchase one smart walker for their personal use, since incredible results were obtained.

This validation (chapter 6) also allowed to establish quantitative measurement for the assessment of the progression of ataxic patients' gait (Goal 7).

Regarding RQ4, which parameters are important to evaluate and diagnose the recovery of a patient with balance that is performing gait training with a smart walker? The presented quantitative information objectively indicated the functional motor recovery of the patients, i.e. with spatiotemporal, posture and symmetry parameters an efficient evaluation can be made. Thus, a first step was given to create evidence-based guidelines for a more efficient and effective application of smart walker devices in rehabilitation programas.

Moreover, the results of this validation showed that a smart walker with forearms can be prescribed as a rehabilitation tool to correct specific gait disorders, such as ataxia.

Depiste these findings, more remains to be learned. In the next subsection improvements and future studies will be presented.

7.2 Future Directions

Innovative aspects include the fact that it is still necessary to develop a smart walker that is adaptable to the type of disorder, addressing the effects of different pathologies. Thus, it is needed to investigate adaptation in order to propose an intelligent device that can provide autonomously higher levels of care for the users. If the user needs to go down an inclined path, the walker will provide a pulling force; it will also adapt its height, improve user's posture, identify danger situations like a fall, restrict some progression forces that provoke sudden accelerations and help in the turning movement. The underlying idea is that if the device can understand the users' walking patterns and infer his/her intentions, it can therefore control the mobility assistance system accordingly. Thus, control of the mobility assistance will be provided in terms of control of velocity, height correction, way of turning, time of response and pulling force. In order to do so, it is necessary to verify the intentions subjacent to gestures considering gait pattern, posture, hesitations in velocity, dangerous situations, etc.

This methodology is user-centred since it performs a personalization of the walker to the patient. This control should be guided by smooth motions in order to achieve a confortable assisted motion by this walker. We consider that smoothness can be quantified as a jerk function, which will be minimized, as well as, energy consumption. Based on the assumption that a natural trajectory demands the least global effort from the user, one needs to minimize the measured sum of efforts at the hands and at the feet. This controller system should output the linear and angular velocity that the walker will perform. These data will be sent to a lower-level architecture that will run a PID controller that also receives encoder's information about the actual motion of the walker.

Finally, validation of the control strategies of the guidance of the walker considering selected and representative groups is necessary, adapting the programs to the electronic architecture. Thus, it will be possible to establish what type of help a particular patient with a particular disease requires.

Other point that should be handled in detail is battery power. One main goal is to charge the device so that it has a good range (6 hours minimum), fast loading and simple, not requiring much manoeuvring by the user. Thus, the aspect of energy autonomy of the walker should be explored. Particular attention should be given to the technology to be adopted i.e. what battery chemistry is more suitable, given its volumetric energy density and weight. Apart from this question, the proper loading of batteries is another key aspect to maximize battery life, but also to optimize daily operation times and discharge cycles under normal operation of the device. Thus, it is proposed to conciliate these parameters and implement a charger with contactless energy transfer, which facilitates its use outside controlled environments, as is the case of health care units. With contactless power transfer, the placement of the device in charging mode would be a task without any difficulty, even for people with serious motor limitations, since the stroller itself can optimize its positioning in the loading device, thereby maximizing the efficiency of energy transfer.

Furthermore, at the electronic level is still necessary to incorporate significant technological developments in the state of art, dealing with the integration of electronic and motors and their protection. This concern has been little explored and is an essential factor for the development of such devices. Also, there is the need to expand the device with omnidirectional wheels for a better movement. Thus it will be possible to obtain a user-centered device, making it a more attractive product.

Regarding gait analysis, further studies should collect more data from TKA patients (and other disorders) creating a database with the proposed features on this study, to then create a model that can be used to help the physicians on deciding which assistive device can be more adequate to a certain TKA patient.

Also, better equipment is necessary. By replacing the accelerometer with foot pressures more detail can be obtained in terms of balance and base of support (BOS) of the patient. It is believed that BOS can be an important outcome for ataxic patients, since it infers the lower limbs load on the floor.

References

[1] S. M. Bradley and C. R. Hernandez, “Geriatric assistive devices,” *American family physician*, vol. 84, no. 4, pp. 405–411, 2011.

[2] A. Eakman, M. Haven, S. Ager, R. Buchanan, N. Fee, S. Gollick, M. Michels, L. Olson, K. Satterfield, and K. Stevenson, “Fall prevention in long-term care: An in-house interdisciplinary team approach,” *Topics in Geriatric Rehabilitation*, vol. 17, no. 3, pp. 29–39, 2002.

[3] G. Gresham, D. Alexander, D. Bishop, C. Giuliani, G. Goldberg, A. Holland, M. Kelly-Hayes, and et al., “Rehabilitation,” *Stroke*, vol. 28, pp. 1522–1526, 1997.

[4] M. Martins, C. P. Santos, A. Frizera, and R. Ceres, “Assistive mobility devices focusing on smart walkers: Classification and review,” *Robotics and Autonomous Systems*, vol. 6, pp. 548–562, 2012.

[5] R. Ceres, “Las tecnologias de apoyo para la vida independiente de los mayores,” *Entre Mayores*, pp. 10–11, 2007.

[6] G. Lacey and K. Dawson-Howe, “Evaluation of robot mobility aid for the elderly blind,” in *Proceedings of the Fifth International Symposium on Intelligent Robotic Systems*, Aug 1997, pp. 1–7.

[7] L. Saint-Bauzel and V. Pasqui, “Intuitiveness facilities rehabilitation: Clinical results,” in *2009 Proceedings of IEEE International Conference on Rehabilitation Robotics*. IEEE, June 2009.

[8] H. Bateni and B. Maki, “Assistive devices for balance and mobility: benefits, demands, and adverse consequences,” *Archive Physical Medical Rehabilitation*, vol. 86, no. 1, pp. 134–145, 2005.

[9] M. Martins, C. P. Santos, and A. Frizera, “A review of the functionalities of smart walkers,” *Medical Engineering and Physics*, pp. 1–12, 2015.

[10] M. Martins, “Online control of a mobility assistance smart walker through a built-in monitoring system to infer the user’s navigations intentions,” Master’s thesis, Algoritmi Center, Minho University, 2011.

[11] M. Martins, C. P. Santos, A. Frizera, and R. Ceres, “Review and classification of human gait training and rehabilitation devices,” in *2011 Proceedings of the 11th European Conference for the Advancement of Assistive Technology in Europe*. AAATE, Setp 2011, pp. 774–781.

[12] ——, “Online control of a mobility assistance smart walker,” in *2012 Proceedings of the 2nd Portuguese Meeting in Bioengineering (ENBENG)*. IEEE, Feb 2012, pp. 1–6.

[13] M. Martins, C. P. Santos, A. Frizera, and E. Seabra, “Design, implementation and testing of a new user interface for a smart walker,” in *Proceedings of the 2014 IEEE International Conference on Autonomous Robot Systems and Competitions (ICARSC)*. IEEE, May 2014, pp. 217–222.

[14] M. Martins, C. P. Santos, A. Frizera, and R. Ceres, “Smart walker control through the inference of the user’s command intentions,” in *2012 Proceedings of the 9th International Conference on Informatics in Control*. IEEE, Sept 2012, pp. 458–463.

[15] M. Martins, C. Santos, E. Seabra, L. Basilio, and A. Frizera, “A new integrated device to read user intentions when walking with a smart walker,” in *2013 Proceedings of the 11th IEEE International Conference on Industrial Informatics (INDIN)*. IEEE, July 2013, pp. 299–304.

[16] V. Faria, J. Silva, M. Martins, and C. P. Santos, “Dynamical system approach for obstacle avoidance in a smart walker device,” in *Proceedings of the 2014 IEEE International Conference on Autonomous Robot Systems and Competitions (ICARSC)*. IEEE, May 2014, pp. 261 – 266.

[17] M. Martins, E. Seabra, L. BasÃlio, and C.Santos, “Conceção, projeto e desenvolvimento de um guiador para um andarilho motorizado,” in *International Conference on Engineering UBI*. ICEUBI, Nov 2013, pp. 27–29.

[18] M. Martins, C. P. Santos, A. Frizera, and R. Ceres, “Real time control of the asbgo walker through a physical human-robot interface,” *Measurement*, vol. 48, pp. 77–86, 2014.

[19] M. Martins, C. Cifuentes, A. Elias, V. Schneider, A. Frizera, and C. P. Santos, “Assessment of walker-assisted human interaction from lrf and wearable wireless inertial

sensors," in *1st International Congress on Neurotechnology, Electronics and Informatics*. INSTICC, Sept 2013, pp. 143–151.

[20] S. Page, M. Martins, L. Saint-Bauzel, C. P. Santos, and V. Pasqui, "Fast embedded feet pose estimation based on a depth camera for smart walker," in *Proceedings of the 2015 IEEE International Conference on Robotics and Automation (ICRA)*. IEEE, May 2015, pp. 4224 – 4229.

[21] M. Martins, S. Page, L. Saint-Bauzel, C. P. Santos, V. Pasqui, and A. Mézière, "Real-time gait assessment with an active depth sensor placed in a walker," in *Proceedings of the 2015 IEEE International Conference on Rehabilitation Robotics (ICORR)*. IEEE, Aug 2015.

[22] M. Martins, C. P. Santos, A. Frizera, and R. Ceres, "Legs tracking for walker rehabilitation purposes," in *2014 Proceedings of the 5th IEEE RAS/EMBS International Conference on Biomedical Robotics and Biomechatronics (BioRob2014)*. IEEE, Aug 2014, pp. 387 – 392.

[23] A. Tereso, M. Martins, C. P. Santos, M. Silva, L. Rocha, and L. M. Gonçalves, "Detection of gait events and assessment of fall risk using accelerometers in assisted gait," in *2014 Proceedings of the 11th International Conference on Informatics in Control, Automation and Robotics (ICINCO)*. IEEE, Sept 2014, pp. 788 – 793.

[24] A. Tereso, M. Martins, and C. P. Santos, "Evaluation of gait performance of knee osteoarthritis patients after total knee arthroplasty with different assistive devices," *Research on Biomedical Engineering*, 2015.

[25] M. Martins, L. Costa, C. P. Santos, A. Frizera, and R. Ceres, "Multivariate analysis of walker-assisted ambulation," in *2013 Proceedings of the 3rd Portuguese Meeting in Bioengineering (ENBENG)*. IEEE, Feb 2013, pp. 1–4.

[26] ——, "Gait feature selection in walker-assisted gait using nsga-ii and svm hybrid algorithm," in *2014 Proceedings of the 22nd European Signal Processing Conference (EUSIPCO)*. IEEE, Sept 2014, pp. 1173 – 1177.

[27] M. Martins, A. Elias, C. Cifuentes, M. Alfonso, A. Frizera, C. P. Santos, and R. Ceres, "Assessment of walker-assisted gait based on principal component analysis and wireless inertial sensors," *Research on Biomedical Engineering*, vol. 30, no. 3, pp. 1–12, 2014.

[28] M. Martins, L. Costa, A. Frizera, and C. P. Santos, “Feature reduction and multi-classification of different assistive devices according to the gait pattern,” *Disability and Rehabilitation: Assistive Technology*, 2015.

[29] ——, “Feature reduction with pca/kpca for gait classification with different assistive devices,” *International Journal of Intelligent Computing and Cybernetics*, vol. 8, no. 4, 2015.

[30] M. Martins, L. Costa, C. P. Santos, A. Frizera, and R. Ceres, “Hybridization between multi-objective genetic algorithm and support vector machine for feature selection in walker-assisted gait,” *Computer Methods and Programs in Biomedicine*, vol. 113, no. 3, pp. 736–748, 2014.

[31] M. Martins, T. Pereira, A. Matias, M. Cotter, F. Pereira, A. Frizera, and C. P. Santos, “Smart walker use for ataxia’s rehabilitation: Case study,” in *Proceedings of the 2015 IEEE International Conference on Rehabilitation Robotics (ICORR)*. IEEE, Aug 2015.

[32] T. Pereira, M. Martins, C. P. Santos, and A. Matias, “Avaliação da eficácia de um andarilho motorizado numa paciente com tetraparésia atáxica,” 2015, poster presented at 9th Congresso Nacional de Fisioterapeutas, Fisioterapia é Saúde, July, Cascais.

[33] M. Cotter, M. Martins, T. Pereira, C. P. Santos, A. Matias, and F. Pereira, “Motorized walker gait training: Gait and balance improvement for cerebellar ataxia,” 2015, poster presented at 9th World Congress of the ISPRM. Germany, Berlin.

[34] M. Cotter, M. Martins, T. Pereira, C. P. Santos, and A. Matias, “Motorized walker for gait training in patients with ataxia,” 2015, poster presented at 9th World Congress of the ISPRM. Germany, Berlin.

[35] ——, “Integração de um andarilho motorizado no programa de reabilitação de doentes com ataxia cerebelosa,” 2015, poster presented at XVI Congresso SPMFR.

[36] M. LaPlante and D. Carlson, “Disability in the united states: Prevalence and causes,” U.S. Department of Education, National Institute on Disability and Rehabilitation Research, Washington, DC, Tech. Rep. 7, 1992.

[37] B. Joyce and R. Kirby, “Canes, crutches and walkers,” *Journal American Family Physician*, vol. 2, no. 43, pp. 535–542, 1991.

[38] R. Craven and P. Bruno, “Teach the elderly to prevent falls,” *Journal of Gerontology Nursing*, vol. 12, pp. 27–33, 1986.

[39] J. Wasson, V. Gall, R. McDonald, and M. Liang, "The prescription of assistive devices for the elderly: practical considerations," *Journal of General Internal Medicine*, vol. 5, no. 1, pp. 46–54, 1990.

[40] J. Russell, G. Hendershot, F. LeClere, L. Howie, and M. Adler, "Trends and differential use of assistive technology devices: United states, 1994," National Center for Health Statistics, Hyattsville, Maryland, Tech. Rep. 292, 1997.

[41] E. Cubo, C. Moore, S. Leurgans, and C. Goetz, "Wheeled and standard walkers in parkinson's disease patients with gait freezing," *Parkinsonism and Related Disorders*, vol. 10, no. 1, pp. 9–14, 2003.

[42] B. Grant and L. Capel, "Walking aid for pulmonary emphysema," *The Lancet*, vol. 300, no. 7787, pp. 1125–1127, 1972.

[43] R. Gupta, D. Brooks, Y. Lacasse, and R. Goldstein, "Effect of rollator use on health-related quality of life in individuals with copd," *Chest*, vol. 130, no. 4, pp. 1089–1095, 2006.

[44] P. Honeyman, P. Barr, and D. Stubbing, "Effect of a walking aid on disability, oxygenation, and breathlessness in patients with chronic airflow limitation," *Journal of cardiopulmonary rehabilitation*, vol. 16, no. 1, pp. 63–67, 1996.

[45] D. Wright and T. Kemp, "The dual-task methodology and assessing the attentional demands of ambulation with walking devices," *Journal of the American Physical Therapy Association and Fysiotherapeut*, vol. 72, no. 4, pp. 306–312, 1992.

[46] R. Wellmon, K. Pezzillo, G. Eichhorn, W. Lockhart, and J. Morris, "Changes in dual-task voice reaction time among elders who use assistive devices," *Journal of geriatric physical therapy*, vol. 29, no. 2, pp. 74–80, 2006.

[47] V. Miyasike-daSilva, J. Tung, J. Zabukovec, and W. McIlroy, "Use of mobility aids reduces attentional demand in challenging walking conditions," *Gait and posture*, vol. 37, no. 2, pp. 287–289, 2013.

[48] C. Holder, E. Haskvitz, and A. Weltman, "The effects of assistive devices on the oxygen cost, cardiovascular stress, and perception of nonweight-bearing ambulation," *The Journal of orthopaedic and sports physical therapy*, vol. 18, no. 4, pp. 537–542, 1993.

[49] M. Foley, B. Prax, R. Crowell, and T. Boone, "Effects of assistive devices on cardiorespiratory demands in older adults," *Physical therapy*, vol. 76, no. 12, pp. 1313–1319, 1996.

[50] J. Priebe and R. Kram, “Why is "walker-assisted" gait metabolically expensive?” *Gait and posture*, vol. 34, no. 2, pp. 265–269, 2011.

[51] J. Roomi, A. Yohannes, and M. Connolly, “The effect of walking aids on exercise capacity and oxygenation in elderly patients with chronic obstructive pulmonary disease,” *Age and ageing*, vol. 27, no. 6, pp. 703–706, 1998.

[52] S. Solway, D. B. L. Lau, and R. Goldstein, “The short-term effect of a rollator on functional exercise capacity among individuals with severe copd,” *Chest*, vol. 122, no. 1, pp. 56–65, 2002.

[53] V. Probst, T. Troosters, I. Coosemans, M. Spruit, O. Pitta, M. Decramer, and R. Gosselink, “Mechanisms of improvement in exercise capacity using a rollator in patients with copd,” *Chest*, vol. 126, no. 4, pp. 1102–1107, 2004.

[54] C. Leung and P. Yeh, “A study on the potential risk factors for the elders using walker,” 2004.

[55] J. Stevens, K. Thomas, and L. T. A. Greenspan, “Unintentional fall injuries associated with walkers and canes in older adults treated in u.s. emergency departments,” *Journal of the American Geriatrics Society*, vol. 57, no. 8, pp. 1464–1469, 2009.

[56] D. Andersen, B. Roos, D. Stanziano, N. Gonzalez, and J. Signorile, “Walker use, but not falls, is associated with lower physical functioning and health of residents in an assisted-living environment,” *Clinical interventions in aging*, vol. 2, no. 1, pp. 123–137, 2007.

[57] H. Liu, “Assessment of rolling walkers used by older adults in senior-living communities,” *Geriatrics and gerontology international*, vol. 9, no. 2, pp. 124–130, 2009.

[58] A. Kloos, D. Kegelmeyer, S. White, and S. Kostyk, “The impact of different types of assistive devices on gait measures and safety in huntington’s disease,” *PloS one*, vol. 7, no. 2, p. e30903, 2012.

[59] D. Kegelmeyer, S. Parthasarathy, S. Kostyk, S. White, and A. Kloos, “Assistive devices alter gait patterns in parkinson disease: advantages of the four-wheeled walker,” *Gait and posture*, vol. 38, no. 1, pp. 20–24, 2013.

[60] G. Smidt and M. Mommens, “System of reporting and comparing influence of ambulatory aids on gait,” *Physical therapy*, vol. 60, no. 5, pp. 551–558, 1980.

[61] J. Crosbie, “Comparative kinematics of two walking frame gaits,” *The Journal of orthopaedic and sports physical therapy*, vol. 20, no. 4, pp. 186–192, 1994.

[62] A. Fast, F. Wang, R. Adrezin, M. Cordaro, J. Ramis, and J. Sosner, “The instrumented walker: usage patterns and forces,” *Archives of physical medicine and rehabilitation*, vol. 76, no. 5, pp. 484–491, 1995.

[63] A. Deathe, R. Pardo, D. Winter, K. Hayes, and J. Russell-Smyth, “Stability of walking frames,” *Journal of rehabilitation research and development*, vol. 33, no. 1, pp. 30–35, 1996.

[64] E. Melis, R. Torres-Moreno, H. Barbeau, and E. Lemaire, “Analysis of assisted-gait characteristics in persons with incomplete spinal cord injury,” *Spinal cord*, vol. 37, no. 6, pp. 430–439, 1999.

[65] T. Ishikura, “Biomechanical analysis of weight bearing force and muscle activiation levels in the lower extremities during gait with a walker,” *Acta Medica Okayama*, vol. 55, no. 2, pp. 73–82, 2001.

[66] R. Bachschmidt, G. Harris, and G. Simoneau, “Walker-assisted gait in rehabilitation: a study of biomechanics and instrumentation,” *IEEE transactions on neural systems and rehabilitation engineering*, vol. 9, no. 1, pp. 96–105, 2001.

[67] L. Vogt, K. Lucki, M. Bach, and W. Banzer, “Rollator use and functional outcome of geriatric rehabilitation,” *The Journal of Rehabilitation Research and Development*, vol. 47, no. 2, pp. 151–156, 2010.

[68] M. Schwenk, M. Schmidt, M. Pfisterer, P. Oster, and K. Hauer, “Rollator use adversely impacts on assessment of gait and mobility during geriatric rehabilitation,” *Journal of rehabilitation medicine: official journal of the UEMS European Board of Physical and Rehabilitation Medicine*, vol. 43, no. 5, pp. 424–429, 2011.

[69] J. Youdas, J. B. Kotajarvi, D. J. Padgett, and K. R. Kaufman, “Partial weight-bearing gait using conventional assistive devices,” *Archives of physical medicine and rehabilitation*, vol. 86, no. 3, pp. 394–398, 2005.

[70] T. Alkjaer, P. Larsen, G. Pedersen, L. Nielsen, and E. Simonsen, “Biomechanical analysis of rollator walking,” *Biomedical engineering online*, vol. 5, no. 2, pp. 1–7, 2006.

[71] K. Mcquade, M. Finley, and A. Oliveira, “Upper extremity joint stresses during walker-assisted ambulation in post-surgical patients,” *Revista Brasileira de Fisioterapia*, vol. 15, no. 4, pp. 332–337, 2011.

[72] L. Haubert, D. Gutierrez, C. Newsam, J. Gronley, S. Mulroy, and J. Perry, "A comparison of shoulder joint forces during ambulation with crutches versus a walker in persons with incomplete spinal cord injury," *Archives of physical medicine and rehabilitation*, vol. 87, no. 1, pp. 63–70, 2006.

[73] M. Takanokura, "Optimal handgrip height of four-wheeled walker on various road conditions to reduce muscular load for elderly users with steady walking," *Journal of biomechanics*, vol. 43, no. 5, pp. 843–848, 2010.

[74] J. Tung, W. Gage, P. Poupart, and W. McIlroy, "Upper limb contributions to frontal plane balance control in rollator-assisted walking," *Assistive Technology: The Official Journal of RESNA*, vol. 26, no. 1, pp. 15–21, 2014.

[75] H. Bateni, E. Heung, J. Z. W. McLlroy, and B. Maki, "Can use of walkers or canes impede lateral compensatory stepping movements?" *Gait and posture*, vol. 20, no. 1, pp. 74–83, 2004.

[76] K. Vennila and A. Aruin, "Postural control in response to a perturbation: role of vision and additional support," *Experimental brain research*, vol. 212, no. 3, pp. 385–397, 2011.

[77] H. Liu, M. McGee, W. Wang, and M. Persson, "Comparison of gait characteristics between older rolling walker users and older potential walker users," *Archives of gerontology and geriatrics*, vol. 48, no. 3, pp. 276–280, 2009.

[78] S. Faruqui and T. Jeblon, "Ambulatory assistive devices in orthopaedics: Uses and modifications," *Journal of the American Academy of Orthopaedic Surgeons*, vol. 18, no. 1, pp. 41–50, 2010.

[79] F. Gray, C. Gray, and J. McClanahan, "Assessing the accuracy of partial weight-bearing instruction," *American journal of orthopedics (Belle Mead, N.J.)*, vol. 27, no. 8, pp. 558–560, 1998.

[80] C. Warren and J. Lehmann, "Training procedures and biofeedback methods to achieve controlled partial weight bearing: an assessment," *Archives of physical medicine and rehabilitation*, vol. 56, no. 10, pp. 449–455, 1975.

[81] L. Naing, T. Winn, and B. Rusli, "Practical issues in calculating the sample size for prevalence studies," *Archives of Orofacial Sciences*, no. 1, pp. 9–14, 2006.

[82] G. Lacey and K. Dawson-Howe, “The application of robotics to a mobility aid for the elderly blind,” *Robotics and Autonomous Systems*, vol. 23, no. 4, pp. 245–252, 1998.

[83] M. Spenko, H. Yu, and S. Dubowsky, “Robotic personal aids for mobility and monitoring for the elderly,” *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 14, no. 3, pp. 344–351, 2006.

[84] A. Rentschler, R. Simpson, R. Cooper, and M. Boninger, “Clinical evaluation of guido robotic walker,” *Journal of Rehabilitation Research and Development*, vol. 25, no. 9, pp. 1281–1294, 2008.

[85] J. Miró, V. Osswald, M. Patel, and G. Dissanayake, “Robotic assistance with attitude: a mobility agent for motor function rehabilitation and ambulation support,” in *Proceedings of the 2009 IEEE 11th International Conference on Rehabilitation Robotics*. IEEE, June 2009, pp. 529–534.

[86] N. Nejatbakhsh and K. Kosuge, “Adaptive guidance for the elderly based on user intent and physical impairment,” in *Proceedings of the 15th International Symposium on Robot and Human Interactive Communication*. IEEE, Sept 2006, pp. 510–514.

[87] R. Tan, S.Wang, Y. Jiang, K. Ishida, and M. Nagano, “Adaptive controller for motion of an omni-directional walker,” in *Proceedings of the 2010 IEEE International Conference on Mechatronics and Automation*. IEEE, Aug 2010, pp. 156–161.

[88] P. Griffiths and R. Gillespie, “Shared control between human and machine: haptic display of automation during manual control of vehicle heading,” in *Proceedings of 12th International Symposium on Haptic Interfaces for Virtual Environment and Teleoperator Systems*. IEEE, March 2004, pp. 358–366.

[89] G. Wasson, J. Gunderson, S. Graves, and R. Felder, “Effective shared control in cooperative mobility aids,” in *Proceedings of the Fourteenth international Florida Artificial intelligence Research Society Conference*. AAAI Press, 2001, pp. 509–513.

[90] A. Morris, R. Donamukkala, A. Kapuria, A. Steinfeld, J. Matthews, J. Dunbar-Jacobs, and S. Thrun, “A robotic walker that provides guidance,” in *Proceedings of IEEE International Conference on Robotics and Automation (ICRA)*. IEEE, 2003, pp. 25–30.

[91] C. Barrué, E. X. Martin, F. Campana, R. Annicchiarico, C. Caltagirone, and F. S. Lucia, “Towards an intelligent service to elders mobility using the i-walker,” in *AAAI Fall Symposia AI in Eldercare: New Solutions to Old Problems*. AAAI Press, 2008, pp. 1–7.

[92] C. Huang, G. Wasson, M. Alwan, P. Sheth, and R. Ledoux, "Shared navigational control and user intent detection in an intelligent walker," 2005.

[93] B. Graf, "An adaptive guidance system for robotic walking aids," *Journal of Computing and Information Technology*, vol. 17, pp. 109–120, 2009.

[94] B. Graf and R. Schraft, "Behavior-based path modification for shared control of robotic aids," in *Proceedings of the 10th International Conference on Rehabilitation Robotics*. IEEE, June 2007, pp. 317–322.

[95] K. Song and S. Jiang, "Force-cooperative guidance design of an omnidirectional walking assistive robot," in *2011 IEEE International Conference on Mechatronics and Automation*. IEEE, Aug 2011, pp. 1258–1263.

[96] O. C. y, Y. Hirata, and K. Kosuge, "Active type robotic mobility aid control based on passive behavior," in *Proceedings of the 2007 IEEE/RSJ International Conference on Intelligent Robots and Systems*. IEEE, Oct-Nov 2007, pp. 165–170.

[97] W. Gharieb, "Intelligent robotic walker design," in *2007 Proceedings of the International Conference on Automation, Robotics and Autonomous Systems*. IEEE, 2007.

[98] H. Shim, E. Lee, J. Shim, S. Lee, and S. Hong, "Implementation of an intelligent walking assistant robot for the elderly in outdoor environment," in *2005 Proceedings of the 9th International Conference on Rehabilitation Robotics*. IEEE, June 2005, pp. 452–455.

[99] I. Ulrich and J. Borenstein, "Vfh+: Reliable obstacle avoidance for fast mobile robots," in *Proceedings of the 1998 IEEE International Conference on Robotics and Automation*. IEEE, May 1998, pp. 1572–1577.

[100] M. Chang, W. Mou, C. Liao, and L.Fu, "Design and implementation of an active robotic walker for parkinson's patients," in *SICE Annual Conference*. IEEE, Aug 2012, pp. 2068–2073.

[101] S. McLachlan, J. Arblaster, D. Liu, J. Miro, and L. Chenoweth, "A multi-stage shared control method for an intelligent mobility assistant," in *Proceedings of the 9th International Conference on Rehabilitation Robotics*. IEEE, June 2005, pp. 426–429.

[102] W. Zhou, L. Xu, and J. Yang, "An intent-based control approach for an intelligent mobility aid," in *Proceedings of the 2nd International Asia Conference on Informatics in Control, Automation and Robotics*. IEEE, March 2010, pp. 54–57.

[103] F. Shi, Q. Cao, C. Leng, and H.Tan, “Based on force sensing-controlled human-machine interaction system for walking assistant robot,” in *Proceedings of the 8th World Congress on Intelligent Control and Automation*. IEEE, July 2010, pp. 6528–6533.

[104] A. Frizera, J. Gallego, E. Rocon, J. Pons, and R. Ceres, “Extraction of user’s navigation commands from upper body force interaction in walker assisted gait,” *BioMedical Engineering Online*, vol. 9, no. 37, pp. 1–16, 2010.

[105] C. Cifuentes, C. Rodriguez, A. Frizera-Neto, T. Bastos-Filho, and R. Carelli, “Multimodal human-robot interaction for walker-assisted gait,” *IEEE Systems Journal*, no. 99, pp. 1–11, 2014.

[106] S. Egawa, Y. Nemoto, A. Koseki, S. Hattori, and M. Fujie, “Power-assisted walking support device,” in *Human Friendly Mechatronics*. Elsevier Science, 1999, pp. 117–122.

[107] J. Ye, J. Huang, J. He, C. Tao, and X. Wang, “Development of a width-changeable intelligent walking-aid robot,” in *Proceedings of the 2012 International Symposium on Micro-NanoMechatronics and Human Science*. IEEE, Nov 2012, pp. 358–363.

[108] Y. Kai, T. Tanioka, Y. Inoue, T. Matsuda, K. Sugawara, Y. Takasaka, and I. Nagamine, “A walking support/evaluation machine for patients with parkinsonism,” *The Journal of Medical Investigation*, vol. 51, no. 1, pp. 1–16, 2004.

[109] H. Hashimoto, A. Sasaki, Y. Ohyama, and C. Ishii, “Walker with hand haptic interface for spatial recognition,” in *Proceedings of the ninth IEEE International Workshop on Advanced Motion Control*. IEEE, 2006, pp. 311–316.

[110] Y. Huang, H. Yang, C. Ko, and K. Young, “Human intention recognition for robot walking helper using anfis,” in *Proceedings of the 8th Asian Control Conference*. IEEE, May 2011, pp. 311–316.

[111] A. Frizera, “Interfaz multimodal para modelado, estudio y asistencia a la marcha humana mediante andadores roboticós,” Ph.D. dissertation, Alcalá University, UAH, Spain, 2010.

[112] Y. Hirata, A. Muraki, and K. Kosuge, “Standing up and sitting down support using intelligent walker based on estimation of user states,” in *Proceedings of the 2006 IEEE International Conference on Mechatronics and Automation*. IEEE, June 2006, pp. 13–18.

[113] V. Pasqui, L. Saint-Bauzel, and O. Sigaud, "Characterization of a least effort user-centered trajectory for sit-to-stand assistance," *IUTAM Symposium on Dynamics Modeling and Interaction Control in Virtual and Real Environments*, vol. 30, pp. 197–204, 2011.

[114] Y. Morita, D. Chugo, Y. Sakaida, S. Yokota, and K. Takase, "Standing motion assistance on a robotic walker based on the estimated patient's load," in *Proceedings of 4th IEEE RAS EMBS International Conference on Biomedical Robotics and Biomechatronics (BioRob)*. IEEE, June 2012, pp. 1721–1726.

[115] H. Jun, Y. Chang, B. Dan, B. Jo, B. Min, H. Yang, W. Song, and J. Kim, "Walking and sit-to-stand support system for elderly and disabled," in *2011 Proceedings of IEEE International Conference on Rehabilitation Robotics*. IEEE, June 2011, pp. 1–5.

[116] M. Alwan, A. Ledoux, G. Wasson, P. Sheth, and C. Huang, "Basic walker assisted gait characteristics derived from forces and moments exerted on the walker's handles: Results on normal subjects," *Medical Engineering and Physics*, vol. 29, no. 3, pp. 380–389, 2007.

[117] J. Henry and V. Aharonson, "Gait monitoring for the elderly using a robotic walking aid," in *Proceedings of the 26th Convention of Electrical and Electronics Engineers*. IEEE, Nov 2010, pp. 392–394.

[118] A. Abellanas, A. Frizera, R. Ceres, and J. Gallego, "Estimation of gait parameters by measuring upper limb-walker interaction forces," *Sensors and Actuators A: Physical*, vol. 162, no. 2, pp. 276–283, 2010.

[119] H. Wu, C. Chien, Y. Jheng, C. Chen, H. Chen, and C. Yu, "Development of intelligent walker with dynamic support," in *Proceedings of the 18th IFAC World Congress*. IFAC, Aug 2011, pp. 5980–5984.

[120] A. Frizera, R. Ceres, E. Rocon, and J. Pons, "Empowering and assisting natural human mobility: The simbiosis walker," *International Journal of Advanced Robotic Systems*, vol. 8, no. 3, pp. 34–50, 2011.

[121] G. Lee, E. Jung, T. Ohnuma, N. Chong, and B. Yi, "Jaist robotic walker control based on a two-layered kalman filter," in *Proceedings of the 2011 IEEE International Conference on Robotics and Automation*. IEEE, May 2011, pp. 3682–3687.

[122] T. Pallejà, M. Teixidó, M. Tresanchez, and J. Palacín, "Measuring gait using a ground laser range sensor," *Sensors*, vol. 9, no. 11, pp. 9133–9146, 2009.

[123] G.Lee, T. Ohnuma, and N. Chong, “Design and control of jaist active robotic walker,” *Intel Service Robotics*, vol. 3, no. 3, pp. 125–135, 2010.

[124] R. Hu, A. Hartfiel, J.Tung, A. Fakih, J. Hoey, and P. Poupart, “3d pose tracking of walker users’ lower limb with a structured-light camera on a moving platform,” in *Proceedings of the 2011 IEEE Computer Society Conference on Computer Vision and Pattern Recognition Workshops (CVPRW)*. IEEE, June 2011, pp. 29–36.

[125] C. Joly, C. Dune, P. Gorce, and P. Rives, “Feet and legs tracking using a smart rollator equipped with a Kinect,” in *Workshop on ”Assistance and Service Robotics in a Human Environment” in conjonction with IEEE/RSJ Int. Conf. on Int. Rob. and Sys. (IROS)*. IEEE, Nov 2013, pp. 1–9.

[126] G. Paolini, A. Peruzzi, A. Mirelman, A. Cereatti, S. Gaukrodger, J. Hausdorff, and U. D. Croce, “Validation of a method for real time foot position and orientation tracking with microsoft kinect technology,” *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 22, no. 5, pp. 997–1002, 2014.

[127] M. Sinn and P. Poupart, “Activity recognition for users of rolling walker mobility aids,” in *Proceedings of the 2010 Workshop on Machine Learning for Assistive Technologies*. NIPS, Dec 2010, pp. 1–9.

[128] O. Postolache, M. Ribeiro, P. S. Girão, J. M. D. Pereira, and G. Postolache, “Unobtrusive sensing for gait rehabilitation assessment,” in *Proceedings of the 8th International Conference on Pervasive Computing Technologies for Healthcare*. ICST (Institute for Computer Sciences, Social-Informatics and Telecommunications Engineering), 2014, pp. 386–389.

[129] K. Yu, C. Lam, M. Chang, W. Mou, S. Tseng, and L. Fu, “An interactive robotic walker for assisting elderly mobility in senior care unit,” in *Proceedings of the 2010 IEEE Workshop on Advanced Robotics and its Social Impacts (ARSO)*. IEEE, Oct 2010, pp. 24–29.

[130] M. Martins, C. P. Santos, A. Frizera, and T. B. R. Ceres, “A novel human-machine interface for guiding: The neoasas smart walker,” in *2012 Proceedings of the Biosignals and Biorobotics Conference (BRC)*. ISSNIP, Jan 2012, pp. 1–7.

[131] G. Schöner and M. Dose, “A dynamical systems approach to task-level system integration used to plan and control autonomous vehicle motion,” *Robotics and Autonomous Systems*, vol. 10, no. 4, pp. 253–267, 1992.

[132] J. Silva, C. P. Santos, and J. Sequeira, “Navigation architecture for mobile robots with temporal stabilization of movements,” in *2013 Proceedings of the 9th Workshop on Robot Motion and Control (RoMoCo)*. IEEE, July 2013, pp. 209–214.

[133] C. Santos, “Generating timed trajectories for an autonomous vehicle: A non-linear dynamical systems approach,” in *Proceedings of the 2004 IEEE International Conference onRobotics and Automation*. IEEE, 2004, pp. 3741–3746.

[134] V. Faria, “Robôs móveis autónomos: Estudo e implementação de técnicas de navegacção,” Master’s thesis, Algoritmi Center, Minho University, 2014.

[135] W. Tao, T. Liu, R. Zheng, and H. Feng, “Gait analysis using wearable sensors,” *Sensors (Basel)*, vol. 12, no. 2, pp. 2255–2283, 2012.

[136] S. Kaur and V. Singh, “Quantitative gait analysis of healthy adults using foot print method and win track,” *Journal of Nursing and Health Science (IOSR-JNHS)*, vol. 3, no. 3, pp. 16–21, 2014.

[137] B. Mariani, M. Jimènez, F. Vingerhoets, and K. Aminian, “On-shoe wearable sensors for gait and turning assessment of patients with parkinson’s disease,” *IEEE Transactions on Biomedical Engineering*, vol. 60, no. 1, pp. 155–158, 2013.

[138] M. S. A. Nardone, “The role of instrumental assessment of balance in clinical decision making,” *European Journal of Physical and Rehabilitation Medicine*, vol. 46, no. 2, pp. 221–237, 2010.

[139] C. Kirtleyl, *Clinical Gait Analysis: Theory and Practice*. Elsevier, 2006.

[140] M. Whittle, *Gait analysis:an introduction*, 4th ed. Churchill Livingstone, 2006.

[141] S. Mathias, U. Nayak, and B. Isaacs, “Balance in elderly patients: the ’get up and go’ test,” *Archive Physical Medical Rehabilitation*, vol. 67, no. 1, pp. 387–389, 1986.

[142] S. Simon, “Quantification of human motion:gait analysis-benefits and limitations of its application to clinical problems,” *Journal of Biomechanics*, vol. 37, no. 12, pp. 1869–1880, 2004.

[143] L. Cruz, D. Lucio, and L. Velho, “Kinect and rgbd images: Challenges and applications,” in *2012 25th SIBGRAPI Conference on Graphics, Patterns and Images Tutorials (SIBGRAPI-T)*. IEEE, Aug 2012, pp. 36–49.

[144] D. Winter, *The Biomechanics and Motor Control of Human Gait: Normal, Elderly and Pathological*. Canada: University of Waterloo Press, 1987.

[145] A. Senior, A. Hampapur, Y. Tian, L. Brown, S. Pankanti, and R.Bolle, "Appearance models for occlusion handling," *Image and Vision Computing*, vol. 24, no. 11, pp. 1233–1243, 2006.

[146] V. Lugade, V. Lin, and L. Chou, "Center of mass and base of support interaction during gait," *Gait and Posture*, vol. 33, no. 3, pp. 406–411, 2011.

[147] J. E. Jackson, *A User's Guide to Principal Components*. John Wiley & Sons, 1999.

[148] H. Kheyruri and D. Frey, "Comparison of people detection techniques from 2d laser range data," 2010, accessed in http://ais.informatik.unifreiburg.de/teaching/ws09/robotics2/projects/mr2-p1-paper.pdf.

[149] G. Chalvatzaki, G. Pavlakos, K. Maninis, X. Papageorgiou, V. Pitsikalis, C. Tzafestas, and P. Maragos, "Towards an intelligent robotic walker for assisted living using multimodal sensorial data," in *2014 EAI Proceedings of the 4th International Conference on Wireless Mobile Communication and Healthcare (Mobihealth)*. IEEE, Nov 2014, pp. 156–159.

[150] J. Xavier, M. Pacheco, D. Castro, A. Ruano, and U. Nunes, "Fast line, arc/circle and leg detection from laser scan data in a player driver," in *Proceedings of the 2005 IEEE International Conference on Robotics and Automation*. IEEE, April 2005, pp. 3930–3935.

[151] N. Bellotto and H. Hu, "Multisensor-based human detection and tracking for mobile service robots," *IEEE Transactions on Systems, Man, and Cybernetics, Part B: Cybernetics*, vol. 39, no. 1, pp. 167–181, 2009.

[152] J. M. Hausdorff and N. B. Alexander, *Gait Disorders: Evaluation and Management*. Informa Healthcare, 2005.

[153] C. Vaughan, B. Davis, and J. Connor, *Dynamics of Human Gait*. South Africa: Kiboho Publishers, 1999.

[154] C. Zong, M. Chetouani, and A. Tapus, "Automatic gait characterization for a mobility assistance system," in *Proceedings of the International Conference on Control, Automation, Robotics and Vision (ICARCV 2010)*. IEEE, Dec 2010, pp. 473–478.

[155] A. Frizera, A. Abellanas, R. Ceres, J. Pons, and R. Raya, "Estudio y caracterización de la cinemática de los pies en marcha asistida con andadores," *Revista Iberoamericana de Automatica e Informatica Industrial (RIAI)*, vol. 6, no. 4, pp. 74–84, 2009.

[156] B. Greene, T. Foran, D. McGrath, E. Doheny, A. Burns, and B. Caulfield, "A comparison of algorithms for body-worn sensor-based spatiotemporal gait parameters to the gaitrite electronic walkway," *Journal of applied biomechanics*, vol. 28, no. 3, pp. 349–355, 2012.

[157] M. Tinetti, "Performance oriented assessment of mobility problems in elderly patients," *Journal of the American Geriatrics Society*, vol. 34, no. 1, pp. 118–126, 1986.

[158] A. G. O. Mutambara, *Decentralized Estimation and Control for Multisensor Systems*, 1st ed. Boca Raton, FL, USA: CRC Press, Inc., 1998.

[159] L. Zadech, "Fuzzy sets," *Information and Control*, vol. 8, no. 3, pp. 338–353, 1965.

[160] V. Fernandes, "Desenvolvimento de um sistema para anÃ¡lise de equilibrio baseado em sensores inerciais," Master's thesis, Algoritmi Center, Minho University, 2013.

[161] M. Rigoberto, "Wearable inertial sensors for human movement analysis during the standing postural position and prior to the first step," Master's thesis, Chiba University, 2011.

[162] S. B. S, P. Couturier, F. Favre-Reguillon, and R. Guillemaud, "Evaluation of postural stability by means of a single inertial sensor," in *Proceedings of the Annual International Conference of the IEEE Engineering in Medicine and Biology Society*. IEEE, 2004, pp. 2275–2278.

[163] M. Gietzelt, G. Nemitz, K. Wolf, H. M. Z. Schwabedissen, R. Haux, and M. Marschollek, "A clinical study to assess fall risk using a single waist accelerometer," *Informatics for health & social care*, vol. 34, no. 4, pp. 181–188, 2009.

[164] N. Ishigaki, T. Kimura, Y. Usui, K. Aoki, N. Narita, M. Shimizu, K. Hara, N. Ogihara, K. Nakamura, H. K. H, M. Ohira, Y. Yokokawa, K. Miyoshi, N. Murakami, S. Okada, T. Nakamura, and N. Saito, "Analysis of pelvic movement in the elderly during walking using a posture monitoring system equipped with a triaxial accelerometer and a gyroscope," *Journal of biomechanics*, vol. 44, no. 9, pp. 1788–1792, 2011.

[165] E. Doheny, D. McGrath, B. Greene, L. Walsh, D. McKeown, C. Cunningham, L. Crosby, R. Kenny, and B. Caulfield, "Displacement of centre of mass during quiet

standing assessed using accelerometry in older fallers and non-fallers," in *Proceedings of the International Conference of the IEEE Engineering in Medicine and Biology Society*. IEEE, 2012, pp. 3300–3303.

[166] B. Greene, A. O'Donovan, R. Romero-Ortuno, L. Cogan, C. Scanaill, and R. Kenny, "Quantitative falls risk assessment using the timed up and go test," *IEEE transactions on biomedical engineering*, vol. 57, no. 12, pp. 2918–2926, 2010.

[167] F. Riva, J. Toebesb, M. Pijnappelsb, R. Stagnia, and J. van Dieenb, "Estimating fall risk with inertial sensors using gait stability measures that do not require step detection," *Gait and posture*, vol. 38, no. 2, pp. 170–174, 2010.

[168] J. Kavanagh and H. Menz, "Accelerometry: a technique for quantifying movement patterns during walking," *Gait and posture*, vol. 28, no. 1, pp. 1–15, 2008.

[169] M. O'Sullivan, C. Blake, C. Cunningham, G. Boyle, and C. Finucane, "Correlation of accelerometry with clinical balance tests in older fallers and non-fallers," *Age and ageing*, vol. 38, no. 3, pp. 308–313, 2009.

[170] H. Menz, S. Lord, and R. Fitzpatrick, "Acceleration patterns of the head and pelvis when walking on level and irregular surfaces," *Gait and posture*, vol. 18, no. 1, pp. 35–46, 2003.

[171] J. Huisinga, M. Mancin, R. George, and F. Horak, "Accelerometry reveals differences in gait variability between patients with multiple sclerosis and healthy controls," *Annals of Biomedical Engineering*, vol. 41, no. 8, pp. 1670–1679, 2013.

[172] L. Cao, K. Chua, and W. Chong, "A comparison of pca, kpca and ica for dimensionality reduction in support vector machine," *Neurocomputing*, vol. 55, pp. 321–336, 2003.

[173] S. Dudoit, J. Fridlyand, and T. Speed, "Comparison of discrimination methods for the classification of tumors using gene expression data," *Journal American Statistics Association*, vol. 97, pp. 77–87, 2002.

[174] M. Efroymson, "Multiple regression analysis," in *Mathematical Methods for Digital Computers*, A. Ralston and H. Wilf, Eds. Wiley, 1960.

[175] L. Huang, H. Zhang, Z. Zeng, and P. Bushel, "Improved sparse multi-class svm and its application for gene selection in cancer classification," *Cancer Inform*, vol. 4, no. 12, pp. 143–153, 2013.

[176] C. Cortes and V. Vapnik, “Support-vector networks,” *Machine Learning*, vol. 20, no. 3, pp. 273–297, 2013.

[177] R. L. Braddom, *Physical Medicine & Rehabilitation*, 3rd ed. Elsevier Health Sciences, 2006.

[178] R. Debi, A. Mor, G. Segal, O. Segal, G. Agar, E. Debbi, N. Halperin, A. Haim, and A. Elbaz, “Correlation between single limb support phase and self-evaluation questionnaires in knee osteoarthritis populations,” *Disability and Rehabilitation*, vol. 33, no. 13, pp. 1103–1109, 2011.

[179] N. Shakoor, J. Block, S. Shott, and J. Case, “Nonrandom evolution of end-stage osteoarthritis of the lower limbs,” *Arthritis& Rheumatology*, vol. 46, no. 12, pp. 3185–3189, 2002.

[180] R. Debi, A. Mor, G. Segal, O. Segal, G. Agar, E. Debbi, N. Halperin, A. Haim, and A. Elbaz, “Differences in gait patterns, pain, function and quality of life between males and females with knee osteoarthritis: a clinical trial,” *BMC musculoskeletal disorders*, vol. 10, no. 127, 2009.

[181] A. Elbaz, A. Mor, O. Segal, G. Agar, N. Halperin, A. Haim, E. Debbi, G. Segal, and R. Debi, “Can single limb support objectively assess the functional severity of knee osteoarthritis?” *The Knee*, vol. 19, no. 1, pp. 32–35, 2012.

[182] K. Berg, S. Wood-Dauphinee, J. Williams, and B. Maki, “Measuring balance in the elderly: validation of an instrument,” *Canadian Public Health Association*, vol. 2, no. 4, pp. 7–11, 1992.

[183] A. Martínez-Ramírez, D. Weenk, P. Lecumberri, N. Verdonschot, D. Pakvis, and P. Veltink, “Pre-operative ambulatory measurement of asymmetric lower limb loading during walking in total hip arthroplasty patients,” *Journal of NeuroEngineering and Rehabilitation*, vol. 10, no. 41, pp. 1–10, 2013.

[184] J. Zhu, T. Hastie, S. Rosset, and R. Tibshirani, “L1-norm support vector machines,” *Neural Information Processing systems*, vol. 15, pp. 49–56, 2003.

[185] E. Allwein, R. Schapire, and Y. Singer, “Reducing multi-class to binary: A unifying approach for margin classifiers,” *Journal of Machine Learning Research*, vol. 1, pp. 113–141, 2000.

[186] Y. Lee, Y. Lin, and G. Wahba, “Multicategory support vector machines, theory and application to the classification of microaaray data and satellite radiance data,” *Journal of the American Statistical Association*, vol. 99, no. 465, pp. 67–81, 2004.

[187] K. Crammer and Y. Singer, “On the algorithmic implementation of multiclass kernel-based vector machines,” *Journal of machine learning research*, vol. 2, pp. 265–292, 2001.

[188] R. Begg and J. Kamruzzaman, “A machine learning approach for automated recognition of movement patterns using basic, kinetic and kinematic gait data,” *Journal of Biomechanics*, vol. 38, pp. 401–408, 2001.

[189] G. Jurman and C. Fuelanello, “A unifying view for performance measures in multi-class prediction,” *PLoS ONE*, vol. 7, no. 8, pp. 1–8, 2012.

[190] S. Olney, M. Griffin, and I. McBride, “Multivariate examination of data from gait analysis of persons with stroke,” *Journal of the American Physical Therapy Association*, vol. 78, pp. 814–828, 1998.

[191] M. Karg, K. Kuhnlenz, and M. Buss, “Recognition of affect based on gait patterns,” *IEEE Transactions on systems, man, and cybernetics-part B: cybernetics*, vol. 40, no. 4, pp. 1050–1061, 2010.

[192] I. Jolliffe, *Principal Component Analysis*. New York: Springer, 2002.

[193] E. Sadler, R. Graha, and J. Stevenson, “The personal lift-assist device and lifting technique: a principal component analysis,” *Ergonomics*, vol. 54, no. 4, pp. 391–402, 2011.

[194] M. Turk and A. Pentland, “Eigenfaces for recognition,” *Journal of Cognitive Neuroscience*, vol. 3, pp. 71–86, 1991.

[195] A. Daffertshofer, C. Lamoth, O. Meijer, and P. Beek, “Pca in studying coordination and variability: A tutorial,” *Clinical Biomechanics*, vol. 19, pp. 415–428, 2004.

[196] T. Chau, “A review of analytical techniques for gait data. part 1: Fuzzy, statistical and fractal methods,” *Gait and Posture*, vol. 13, pp. 49–66, 2001.

[197] B.Scholkopf and A. Smola, *Learning with kernels: Support vector machines, regularization, optimization and beyond*. Cambridge, MA: MIT, 2002.

[198] V. Talis, A. Grishin, I. Solopova, T. Oskanyan, V. Belenky, and Y. Ivanenko, “Asymmetric leg loading during sit-to-stand, walking and quiet standing in patients after unilateral total hip replacement surgery,” *Clinical Biomechanics*, vol. 23, pp. 424–433, 2008.

[199] D. Hurwitz, C. Hulet, T. Andriacchi, A. Rosenberg, and J. Galante, “Gait compensations in patients with osteoarthritis of the hip and their relationship to pain and passive hip motion,” *Journal of Orthopaedic Research*, vol. 15, pp. 629–635, 1997.

[200] A. Kocsor and L. Tóth, “Kernel-based feature extraction with a speech technology application,” *IEEE Transactions on Signal Processing*, vol. 52, pp. 2250–2263, 2004.

[201] P. Levinger, D. Lai, R. Begg, K. Webster, and J. Feller, “The application of support vector machines for detecting recovery from knee replacement surgery using spatio-temporal gait parameters,” *Gait and Posture*, vol. 29, no. 1, pp. 91–96, 2009.

[202] Y. Saeys, I. Inza, and P. Larrañaga, “A review of feature selection techniques in bioinformatics,” *Bioinformatics Advance Access*, vol. 23, no. 19, pp. 2507–2517, 2007.

[203] T. Hamdani, J. Won, A. Alimi, and F. Karray, “Multi-objective feature selection with nsga ii,” *Adaptive and Natural Computing Algorithms*, vol. 4431, pp. 240–247, 2007.

[204] C. Igel, *Multi-objective model selection for support vector machines*. Berlin: Springer-Verlag, 2005.

[205] L. Wang and X. Shen, “On l1-norm multi-class support vector machines methodology and theory,” *Journal of the American Statistical Association*, vol. 102, pp. 583–594, 2007.

[206] H. Zhang and W. Lu, “Adaptive-lasso for cox’s proportional hazard model,” *Biometrika*, vol. 94, pp. 691–703, 2007.

[207] O. Yamashita, M. Sato, T. Yoshioka, F. Tong, and Y. Kamitani, “Sparse estimation automatically selects voxels relevant for the decoding of fmri activity patterns,” *Neuroimage*, vol. 42, no. 4, pp. 1414–1429, 2008.

[208] M. V. J. Bussmann, J. Verhaar, L. A. A. Furlan, and M. Reijman, “Recovery of physical functioning after total hip arthroplasty: systematic review and meta-analysis of the literature,” *Physical Therapy*, vol. 91, no. 5, pp. 615–629, 2011.

[209] S. Morton and A. Bastian, “Cerebellar control of balance and locomotion,” *Neuroscientist*, vol. 10, pp. 247–259, 2004.

[210] M. Pandolfo and M. Manto, "Cerebellar and afferent ataxias," *Movement Disorders. Continuum (Minneap Minn)*, vol. 5, no. 19, pp. 1312–1342, 2013.

[211] K. Cernak, V. Stevens, R. Price, and A. Shumway-Cook, "Locomotor training using body-weight support on a treadmill in conjunction with ongoing physical therapy in a child with severe cerebellar ataxia," *Physical Therapy*, vol. 88, no. 1, pp. 88–97, 2008.

[212] S. Perlman, *Evaluation and Management of ataxic disorders-an overview for physicians*. Minneapolis, MN: National Ataxia Foundation, 2007.

[213] W. Ilg, A. Bastian, S. Boesch, R. Burciu, P. Celnik, J. Claaben, K. Feil, R. Kalla, I. Miyai, W. Nachbauer, L. Schöls, M. Strupp, M. Synofzik, J. Teufel, and D. Timmann, "Consensus paper: Management of degenerative cerebellar disorders," *Cerebellum*, vol. 13, no. 2, pp. 248–268, 2014.

[214] K. Armutlu, "Ataxia: Physical therapy and rehabilitation applications for ataxic patients," in *International Encyclopedia of Rehabilitation*, J. Stone and M. Blouin, Eds., 2010.

[215] M. Orsini, V. Bastos, M. Leite, M. de Freitas, C. Bruno, and J. de Souza, "Neurological rehabilitation in patients with spinocerebellar ataxia: it's really effective and permanent?" *Physical Medicine and Rehabilitation*, vol. 1, no. 1, pp. 1–2, 2014.

[216] W. Ilg, D. Brotz, S. Burkard, M. A. Giese, L. Schols, and M. Synofzik, "Long-term effects of coordinative training in degenerative cerebellar disease," *Movement Disorders*, vol. 25, no. 13, pp. 2239–2246, 2010.

[217] C. Gibson-Horn, "Balance-based torso-weighting in a patient with ataxia and multiple sclerosis: A case report," *Journal of Neurologic Physical Therapy*, vol. 32, no. 3, pp. 139–146, 2008.

[218] G. Abbud, C. Janelle, and M. Vocos, "The use of a trained dog as a gait aid for clients with ataxia: a case report," *Physiotherapy Canada*, vol. 66, no. 1, pp. 33–35, 2014.

[219] K. Armutlu, R. Karabudak, and G. Nurlu, "Physiotherapy approaches in the treatment of ataxic multiple sclerosis: a pilot study," *Neurorehabilitation and Neural Repair*, vol. 15, no. 3, pp. 203–211, 2001.

[220] Y. Baram and A. Miller, "Auditory feedback control for improvement of gait in patients with multiple sclerosis," *Journal of the Neurological Sciences*, vol. 254, pp. 90–94, 2007.

[221] T. Brown, J. Mount, B. R. K. Kautz, R. Barnes, and J. Kim, “Body weight-supported treadmill training versus conventional gait training for people with chronic traumatic brain injury,” *Journal Head Trauma and Rehabilitation*, vol. 20, no. 5, pp. 402–415, 2005.

[222] K. Brown, S. W. G. Marchetti, D. Wrisley, and J. Furman, “Physical therapy for central vestibular dysfunction,” *Archives of Physical Medicine and Rehabilitation*, vol. 87, no. 1, pp. 76–81, 2006.

[223] U. Bultmann, D. Pierscianek, E. Gizewski, B. Schoch, N. Fritsche, D. Timmann, M. Maschke, and M. Frings, “Functional recovery and rehabilitation of postural impairment and gait ataxia in patients with acute cerebellar stroke,” *Gait and Posture*, vol. 39, no. 1, pp. 563–569, 2014.

[224] K. C. D. Kaur-Mann, H. Cooper, A. Mansfield, J. Offord, and D. Marple-Horvat, “Rehearsal by eye movement improves visuomotor performance in cerebellar patients,” *Experimental Brain Research*, vol. 146, no. 2, pp. 244–247, 2002.

[225] M. Dias, F. Toti, S. Almeida, and T. Oberg, “Efeito do peso para membros inferiores no equilibrio estático e dinâmico nos portadores de ataxia,” *Acta Fisiatria*, vol. 16, no. 3, pp. 116–120, 2009.

[226] H. Dordal, “Intensive mobility training as a means of late rehabilitation after brain injury,” *Adapted Physical Activity Quarterly*, vol. 6, pp. 176–187, 1989.

[227] T. Folz and M. Sinaki, “A nouveau aid for posture training in degenerative disorders of the central nervous system,” *Journal of Musculoskeletal Pain*, vol. 3, pp. 59–69, 1995.

[228] J. Freund and D. Stetts, “Use of trunk stabilization and locomotor training in an adult with cerebellar ataxia: a single system design,” *Physiotherapy Theory and Practice*, vol. 26, pp. 447–458, 2010.

[229] B. Gialanella, M. Bertolinelli, V. Monguzzi, and R. Santoro, “Walking and disability after rehabilitation in patients with cerebellar stroke,” *Minerva Medica*, vol. 96, pp. 373–378, 2005.

[230] K. Gill-Body, R. Popat, S. Parker, and D. Krebs, “Rehabilitation of balance in two patients with cerebellar dysfunction,” *Physical Therapy*, vol. 77, no. 5, pp. 534–552, 1997.

[231] G. Gillen, "Improving activities of daily living performance in an adult with ataxia," *American Journal Occupational Therapy*, vol. 54, no. 1, pp. 89–96, 2000.

[232] A. H. Y. Nilsagard, A. Forsberg, H. Pepa, E. Skargren, and B. Oberg, "Evaluation of therapeutic riding (sweden)/hippotherapy (united states) replicates in eleven patients with multiple sclerosis," *Physiotherapy Theory and Practice*, vol. 21, no. 1, pp. 51–77, 2005.

[233] M. Harris-Love, K. Siegel, S. Paul, and K. Benson, "Rehabilitation management of friedreich ataxia: lower extremity force-control variability and gait performance," *Neurorehabilitation and Neural Repair*, vol. 18, no. 2, pp. 117–124, 2004.

[234] W. Ilg, C. Schatton, J. Schicks, M. G. MA, L. Schöls, and M. Synofzik, "Video game-based coordinative training improves ataxia in children with degenerative ataxia," *Neurology*, vol. 79, no. 1, pp. 2056–2060, 2012.

[235] W. Ilg, M. Synofzik, D. Brötz, S. Burkard, M. Giese, and L. Schöls, "Intensive coordinative training improves motor performance in degenerative cerebellar disease," *Neurology*, vol. 73, no. 1, pp. 1–8, 2009.

[236] L. Jones, Y. Lewis, J. Harrison, and C. Wiles, "The effectiveness of occupational therapy and physiotherapy in multiple sclerosis patients with ataxia of the upper limb and trunk," *Clinical Rehabilitation*, vol. 10, no. 1, pp. 277–282, 1996.

[237] J. Jorge-Rodríguez, E. Fernández-Martínez, D. Pérez, A. Peralta-Flores, and J. Bergado, "Motor improvement in cerebellar ataxia after integral rehabilitation," *Journal of Neurorestoratology*, vol. 2013, no. 1, pp. 31–36, 2013.

[238] M. Karakaya, N. Köse, S. Otman, and T. Ozgen, "Investigation and comparison of the effects of rehabilitation on balance and coordination problems in patients with posterior fossa and cerebellopontine angle tumours," *Journal Neurosurgical Sciences*, vol. 44, no. 4, pp. 220–225, 2000.

[239] O. Kaut, H. Jacobi, C. Coch, A. Prochnicki, M. Minnerop, T. Klockgether, and U. Wüllner, "A randomized pilot study of stochastic vibration therapy in spinocerebellar ataxia," *The Cerebellum*, vol. 13, no. 2, pp. 237–242, 2014.

[240] J. Keller and A. Bastian, "A home balance exercise program improves walking in people with cerebellar ataxia," *Neurorehabilitation and Neural Repair*, vol. 28, no. 8, pp. 770–778, 2014.

[241] W. Kim, S. Jung, M. Oh, Y. Min, J. Lim, and N. Paik, “Effect of repetitive transcranial magnetic stimulation over the cerebellum on patients with ataxia after posterior circulation stroke: A pilot study,” *Journal of Rehabilitation Medicine*, vol. 46, no. 1, pp. 418–423, 2014.

[242] A. Marquer, G. Barbieri, and D. Pérennou, “The assessment and treatment of postural disorders in cerebellar ataxia: a systematic review,” *Annals of Physical and Rehabilitation Medicine*, vol. 57, pp. 67–78, 2014.

[243] C. Torriani-Pasin, M. Arjona, R. G. Leitão, R. Z. de Lima, and F. N. Cyrillo, “Dual task effect in ataxia gait,” *Science in Health*, vol. 1, no. 2, pp. 101–110, 2010.

[244] B. Zeigelboim, S. D. de Souza, H. Mengelberg, H. A. G. Teive, and P. B. N. Liberalesso, “Vestibular rehabilitation with virtual reality in spinocerebellar ataxia,” *Audiology-Communication Research*, vol. 18, no. 2, pp. 143–147, 2013.

[245] R. Ribeiro and C. Iwabe-Marcese, “Using virtual reality for motor rehabilitation in a child with ataxic cerebral palsy: case report,” *Audiology-Communication Research*, vol. 22, no. 1, pp. 97–102, 2015.

[246] M. Morgan, R. Hewer, and R.Cooper, “Application of an objective method of assessing intention tremor-a further study on the use of weights to reduce intention tremor,” *Journal of Neurology, Neurosurgery & Psychiatry*, vol. 38, pp. 259–267, 1975.

[247] N. Clopton, D. Schultz, C. Boren, J. Porter, and T. Brillbart, “Effects of axial loading on gait for subjects with cerebellar ataxia: preliminary findings,” *Journal of Neurologic Physical Therapy*, vol. 27, pp. 15–21, 2003.

[248] P. S. Branco, “Validation of the portuguese version of the "activities-specific balance confidence scale",” *Revista da Sociedade Portuguesa de Medicina FÃsica e de ReabilitaÃ§Ã£o*, vol. 19, no. 2, 2010.

[249] Y. Lajoie and S. Gallaghe, “Predicting falls within the elderly community: comparison of postural sway, reaction time, the berg balance scale and the activities-specific balance confidence (abc) scale for comparing fallers and non-fallers,” *Archives of Gerontology and Geriatrics*, vol. 38, no. 1, pp. 11–26, 2004.

[250] J. Hidler, “What is next for locomotor based studies?” *Journal of Rehabilitation Research & Development [Guest editorial.]*, vol. 42, no. 1, pp. xi–xiv, 2005.

Appendix A

Conventional Walkers Summary

Table A.1: Summary of the articles investigating more than one walker in their study.

Authors	Participants	Testing Protocol/Instrumentation	Parameters to evaluate	Data processing/Statistical analysis	Purpose of the investigation	Main Findings: Benefits	Main Findings: Risk/Demands
Wright & Kemp (1992)	10 subjects; mean age: 30.8±8.1 (22-49) years; Inclusion criteria: healthy and had never used a walker. Informed consent signed.	- Instrumentation: microphone, timer and alarm (1500Hz tone); - 3 conditions: without AD, SW and 4WW; - 3 phases: 1st only execute the secondary probe; 2nd Execute the 3 walking conditions; 3rd Execute both tasks. - 10 trials for 1ª phase; 12 trials for each subject, for each condition in 3rd phase. - Walk distance: 6.1m. - Explanation of procedure session. (Secondary probe: reproduce a vocal response as rapidly as possible to the presentation of a tone).	Spatiotemporal: Step length, time and number of steps, Cadence; VRT.	- Mean and standard variation; - ANOVA; - Teste Tukey post hoc; - p<0.05	Preliminary examination of the attentional demands of ambulating with a SW and 4WW and introduce a dual-task methodology.	-	Walking aided is motion demanding and results on lower cadence; greater attentional demand and greater VRT with SW;
Holder et al. (1993)	9 subjects; mean age: 29.11 ± 2.62 years; mean weight: 66.5 1±12.10 kg; mean height: 168.94± 6.53 cm; Inclusion criteria: healthy and had no history of chronic or acute cardiovascular or respiratory disorders. Informed consent signed.	- Instrumentation: timer, treadmill, metabolic measurement devices on the subject; - 4 conditions: no AD, axillary crutches, SW and 4WW; randomized order; - Before test: Oxygen consumption test on a treadmill; - Pre-adjusted devices (hand grips' angle – 15 to 30 degrees - and height -5cm below the axilla); - Training session; - Walk test: walk at self-selected velocities for 7 minutes during each condition on a 24.5 m oval path laid out on a level concrete floor; - 2 trials, 1 week apart, each condition; - Rest interval between each condition until heart rate and blood pressure returned to resting levels;	Velocity; metabolic measures: oxygen cost, cardiovascular stress, heart rate, blood pressure, rate pressure product and perception of effort.	- Mean and standard deviation; - ANOVA; - Tukey's test; - Pearson product moment correlations; - t-test; - Test-retest reliability; - p<0.05	Compare metabolic measures between unassisted gait and assisted gait (axillary crutches, SW and 4WW).	Compare metabolic measures between unassisted gait and assisted gait (axillary crutches, SW and 4WW).	SW: Increased energy cost; 4WW: Greatest oxygen consumption per meter, thus increasing the heartbeat; SW and 4WW: Reduced velocity and more effort to move the walker;
Foley et al. (1996)	10 subjects (3 male, 7 female); mean age: 60.3±8.4 (50-74) years; Inclusion criteria: healthy with no functional limitations secondary to medical problems and not dependent on AD for ambulation. Informed consent signed.	- Instrumentation: timer, metabolic measurement devices on the subject; - 4 conditions: no AD, SW, 4WW, and single-point cane; randomized order; - Explanation of procedure session; - Training session (5 min); - Walk test: walk for 2 minutes at a self-selected speed through a flat, smooth concrete indoor surface with 45.7 m long and 7.6 m wide. - Rest interval between each condition until heart rate and blood pressure returned to resting levels.	Metabolic measures: Oxygen consumption, hear rate, MET (metabolic equivalent of resting metabolism), blood pressure, rate-pressure product (RPP), rating of perceived exertion (RPE), systolic pressure (SBP), diastolic pressure (DBP),ventilation per minute, tidal volume and frequency of breaths; velocity.	- Mean and variance; - ANOVA; -Newman-Keuls post hoc analysis; - p<0.05	Quantify and compare cardiorespiratory demands imposed during unassisted ambulation and ambulation with various ADs in older adults.	No difference was detected for physiologic demands between unassisted ambulation and ambulation with a cane and 2WW walker.	SW: Require > oxygen per meter than unassisted ambulation; > oxygen per meter than ambulation with a 4WW; > heart rate per meter as compared with unassisted ambulation and with 4WW, respectively. Walker is not recommended for myocardia infarct patients.

Table A.2: Summary of the articles investigating more than one walker in their study.(continuation)

Authors	Participants	Testing Protocol/Instrumentation	Parameters to evaluate	Data processing/Statistical analysis	Purpose of the investigation	Main Findings	
						Benefits	Risk/Demands
Roomi et al. (1998)	27 subjects (15 male, 12 females); mean age: 75 (70-82) years; Inclusion criteria: out-patients, clinically stable with no change in medication; Exclusion criteria: COPD exacerbation or oral steroid use within 6 weeks, confusional state, participation in a pulmonary rehabilitation programme and exercise limitation by other diseases. Informed consent signed.	- Instrumentation: timer, pulse oximeter and spirometer; - 4 conditions: no AD, Zimmer, 4WW and gutter frames; randomized order; - Walk test: 6MW tests for each condition; - 4 trials in 4 days, each condition; - Rest 6-min.	Distance walked in 6 min; metabolic measure: Oxygen saturation.	- Paired t-tests; - p<0.05	Evaluate the effect of three different walkers with the 6MW test and oxygen saturation during exercise in elders COPD patients.	Gutter frames improve exercise capacity and oxygen saturation during exercise in elderly COPD patients; increase the walked distance and velocity. Use of a 4WW did not significantly affect the unaided ambulation values. 4WW improves exercise capacity in elderly patients with COPD.	Zimmer frame reduced the mean distance, low velocity, increased metabolic costs, mobility reduction.
Cubo et al. (2003)	19 subjects; Inclusion criteria: 5 PD out-patients and 14 non-demented.	- Instrumentation: laser beam and video-camera. - 5 conditions; no AD, SW, 4WW, 4WW with a laser beam turned on throughout the trial, and with a laser beam activated by the patient when freezing occurred; randomized order; - Training session; - Test: walked up through a standard course (rising from a chair, walking through a doorway, straightway walking, pivoting, and return); - 3 trials each condition. - UPDRS, MMSE.	Temporal: Walking time and freezing time; number of freezes.	- Mixed models and Friedman's test (α=0.05); - Signed Rank tests; - Bonferroni correction for multiple tests; - p<0.017 - Software: SAS.	Compare the efficacy of two walking assistance devices (4WW and SW) to unassisted walking for patients with PD and gait freezing.	Neither walker reduced any index of freezing, nor the laser attachment offered any advantage to the 4WW.	Use of either type of device significantly slowed walking compared to unassisted walking. The SW increased freezing, and the 4WW had no effect on freezing.
Leung & Yeh (2008)	19 subjects (11 male, 8 female); mean age: 73.21±8.48 (64- 88) years; Inclusion criteria: age > 65 years, no acute medical illness in the past 3months, such as coronary heart disease, heart failure, or pulmonary infection, no orthopedic diagnoses, using walker or has used walker, and no neurological disease.	- Instrumentation: video-camera; - A questionnaire consists of 37 items and is divided into four parts: background of participants, walker using state, activities in using walker and the opinion of user. The opinions of user were 5-point Likert Scale. - Walk with the AD.	Questionnaire answers; video data.	- Mean and standard variation: - One-mean t-test; - One-way ANOVA with blocking design; - If the results are significance, use LSD (Least significant difference) post-hoc test; - p<0.05 - Software: SPSS.	Investigate whether the potential risk factors causing the injury exist or not while elders are using walkers and the needs to redesign a new walker.	Stable equilibrium, holding gesture, and prevented slipping are recognized as significant important function of walkers. Most elders use SW; therefore, they believed that the function of brake is the least important.	Pain in wrists, shoulders, back and waist, mal-function of brake, and bumping into feet or legs while using walker are recognized as significant annoyance cause by using walker. Some improvement to redesign walker have been suggested.

Table A.3: Summary of the articles investigating more than one walker in their study.(continuation)

Authors	Participants	Testing Protocol/Instrumentation	Parameters to evaluate	Data processing/Statistical analysis	Purpose of the investigation	Main Findings	
						Benefits	Risk/Demands
Liu (2009)	158 subjects; Inclusion criteria: walker users for 24 months and age > 65 years.	- Instrumentation: video camera; - Questionnaire about walkers (2WW and 4WW); - MMSE; subjects' demographic data; - 7 months of study.	Walker height; maintenance of the walker; posture during standing and walking, gait pattern while walking in a straight line, and during making turns; Number of subjects who have fallen.	- Test 'chi-square'; - ANOVA; - $p<0.05$ - Software: SPSS.	Provide information about wheeled walker use that can aid clinical professionals in understanding how to educate walker users.	Two and four wheels increase of the base support; improve balance and mobility; Reposition the body forward;	Two and four wheels cause the loss of arm swing; Incorrect posture; increase the risk to fall.
Stevens et al (2009)	47,312 subjects; Inclusion criteria: age > 65 years and older, treated in emergency departments of hospitals for a nonfatal and unintentional fall injury in a selected period of time; Exclusion criteria: cases whose falls were not associated with their own AD.	- Surveillance data of injuries treated in hospital emergency departments (EDs), January 1, 2001, to December 31, 2006; - The National Electronic Injury Surveillance System All Injury Program, which collects data from a nationally representative stratified probability sample of 66 U.S. hospital EDs.	Sex, age, whether the fall involved a cane or walker, primary diagnosis, part of the body injured, disposition, and location and circumstances of the fall.	- Direct variance estimation procedure that accounted for the sample weights and complex sampling design; - $p < 0.05$.	To characterize nonfatal, unintentional, fall-related injuries associated with walkers and canes in older adults.	-	Walker-related injuries were 7 times higher than the incidence of cane-related incidents.
Priebe & Kram (2011)	10 subjects (5 males, 5 females); mean age: 25.8±4.4 years; mean body mass: 70.9±12.2 kg; Inclusion criteria: young and healthy. Informed consent signed.	- Instrumentation: treadmill, gas analysis system and video camera; - 4 conditions: no AD, 2WW, 2WW and SWs; randomized order; - Two gait patterns: bipedal and step-to. - Device height adjusted with elbow flexion angle 20o-30o; - Training session (mimic elders); - First test: walk at a preferred speed on a 30-m walkway on a treadmill; - Walk test: Walk at a 0.30 m/s speed in 7 min; - 2 trials each condition; - Rest 3-min.	Spatiotemporal: preferred walking speed, cadence, stride length; metabolic measures: average rates of oxygen consumption and carbon dioxide production and the gross cost of transport.	- ANOVA - Tukey post hoc; - $p < 0.05$.	Comparison of metabolic consumption for different walkers and determination of temporal-spatial parameters.	4WW requires less oxygen consumption than 2WW walker.	The metabolic cost per distance walked was greater with SW. The high cost of SW is due to: the slow walking speed, the step-to gait pattern and the repeated lifting of the walker. 4WWs do not have the inherent static stability of a SW.

Table A.4: Summary of the articles investigating more than one walker in their study.(continuation)

Authors	Participants	Testing Protocol/Instrumentation	Parameters to evaluate	Data processing/Statistical analysis	Purpose of the investigation	Main Findings	
						Benefits	Risk/Demands
Kloos et al. (2012)	21 subjects; Inclusion criteria: HD patients; ability to comprehend complex instructions as documented by ability to appropriately follow instructions needed to perform the standard UHDRS (Unified Huntington's Disease Rating Scale) neuropsychiatric cognitive tests; ability to walk a minimum of 10 meters without an AD or physical assistance; absence of any additional central nervous system disorders; and absence of orthopedic and peripheral neurological disorders affecting the LEs. Informed consent signed.	- Instrumentation: GaitRite® ; - 7 conditions: no AD, a cane, a weighted cane, a SW, and 2WW, 3WW, 4WW; randomized order; - Training session; - Walk test: walk 4.88 m at a normal and comfortable pace and walk around two obstacles in a figure-of-eight pattern. - The 1st trial under each condition was a practice trial. - 4 trials each condition and task.	Spatiotemporal: step time, stride length, swing time and double support time, number of stumbles, base of support width.	- Coefficient of variation; - One-way repeated-measures ANOVA; - Tukey post hoc; - $p < 0.05$. - Software: SAS.	Examine the effects of assistance devices use on gait in individuals with HD.	3WW and 4WW produce higher velocity and higher gait stride length compared to other walkers; 4WW produced less variability in gait; fewer falls and stumbles; less effort to maneuver the walker; more stability; best device to use in paths with obstacles;	SW increased time in double support; More varied gait; 2WW and SW produced significant reduction of gait velocity and stride length compared unassisted gait. 2WW, 3WW, 4WW produced a narrow base of support compared to unassisted gait; 3WW resulted in more time in double support than 4WW or unassisted gait; 4WW causes fear of falling.
Kegelmeyer et al. (2013)	27 subjects (22 male, 15 female); mean age: 69.7±1.3 (55–83) years; Inclusion criteria: PD patients, no regular use of an AD, gait and balance deficits on the UPDRS (Unified Parkinson's Disease Rating Scale); ability to walk a minimum of 10 m without an AD, absence of any orthopedic and peripheral neurological disorders affecting the LEs.	Same as Kloos el al (2012).	Spatiotemporal: number of stumbles, number of freezing episodes, velocity, stride length, swing time, stance time and base of support width.	- Coefficient of variation; - One-way repeated-measures ANOVA; - Tukey post hoc; - $p < 0.05$. - Software: SAS.	Examine the effects of assistance devices use on gait in individuals with PD.	4WW produces a more secure gait; less varied gait.	SW produces a more variable and slow gait compared to all other devices; demand for greater attention; SW and 2WW decreased gait velocity and stride length compared to other devices; 2WW produces a greater number of freezing curves.

Table A.5: Summary of the articles investigating the influence of the standard walker.

Authors	Participants	Testing Protocol/Instrumentation	Parameters to evaluate	Data processing/Statistical analysis	Purpose of the investigation	Main Findings	
						Benefits	Risk/Demands
Smidt & Mommens (1980)	25 subjects (12 male, 13 female); mean age: 22 years; mean height; 168 cm; mean weight: 64.6 kg; Inclusion criteria: Normal and healthy.	- Instrumentation: automated gait analysis system with triaxial accelerometers (placed posterior to the sacrum), pressure-sensitive foot switches (attached to the heel and forefoot of each shoe), signal-amplification unit, and a laboratory computer; Graduated strips of tape on the walkway; - 10 conditions: (1) no AD and (2) canes, crutches and SW with nine different types of assisted gait patterns; randomized order; - Training session; - (1) 4 categories: a self-selected velocity, moderate velocity (71-110 cm/sec), slow velocity (31-70 cm/sec), and very slow velocity (30 cm/sec or less); 2 trials; - (2) 2 trials each condition.	Spatiotemporal: velocity, step length; step time, gait cycle, Swing/time Ratio, Stance/time Ratio, Step-distance ratio, Step-time, Double stance time; kinematics: body acceleration.	- Mean and standard deviation; - p <0.05.	Present a standardized approach for describing gait when ADs are used; report reference data for unassisted and assisted gait patterns for normal adults; and discuss clinical implications for selected variables of gait.	Swing/time Ratio and Stance/time were symmetrical.	Subjects walked slower with SW than without them. Vertical accelerations were disproportionately elevated. Asymmetry of step length. Double-stance time and step times were asymmetrical.
Crosbie (1994)	10 subjects (3 males, 7 females); mean age: 58.3 (45-67) years; Inclusion criteria: Normal and healthy.	- Instrumentation: 3 cameras (two sagittal sides and end of the walkway – front side)[25Hz], body markers on the selected anatomical and SW locations; - Walker height adjusted at the handgrips to the same level as the radial styloid; elbows flexed 30o; - 2 gait patterns: gait D (the SW was advanced followed by the right foot and then the left) and gait S (the walker and right foot moved simultaneously forward, followed by the left foot); randomized order; - Training session; - Walk test: walk at a self-selected speed over the 8-m walkway; - 5 trials each gait pattern;	Spatiotemporal displacements: walker travel distance, walker travel time, right step length, left step length, right step time, left step time, cycle duration, cadence e average velocity of walker/subject; Linear and angular displacements of trunk, thigh, shank and walker.	- The middle of two-cycles of each walk was used for analysis; - Temporal-Spatial displacements were divided by the subject's total height; - Data was subjected to Fourier analysis and filtered with a fourth-order, zero-lag Butterworth filter with a cutoff value of 6 Hz; - Mean and standard deviation; - ANOVA; - Schéffé's multiple range post hoc analyses; - p<0.05 - Software: Minitab.	Compare the two gaits with respect to patterns of joint motion and spatiotemporal parameters.	Gait D is slower but causes less perturbation of balance; may offer added security and stability to the user.	Discrete gait causes no benefit to forward linear momentum of the body; Gait S imposes less of a flexed posture on the protected hip joint during the period of weight transfer forward onto the frame and protected foot than gait D.
Fast et al. (1995)	12 subjects (7 female, 5 male); age range: 24-90 years; Inclusion criteria: various diagnoses leading to gait dysfunction.	- Instrumentation: Strain gauges were mounted on each leg at a height of approximately 15 inches from the ground; - Calibration session; - Walker's height adjusted to elbows flexed at 20 to 30 ° of flexion; - A tester were standing next to the patient who was ambulating; - Walk test: walk 30 seconds each at self-selected speed; - 2 trials.	Kinetics: Axial load, anterior-posterior bending and medial-lateral bending.	- The electric signals were converted from millivolts to pounds.	Develop an instrumented walker, to evaluate the forces that are transferred through its frame during ambulation, and to observe the walker's usage pattern by several patients with gait problems.	Reduction of the load/body weight on the LEs; support a large percentage of the body weight.	Requirement of walker specification for each patient.

Table A.6: Summary of the articles investigating the influence of the standard walker.(continuation)

Authors	Participants	Testing Protocol/Instrumentation	Parameters to evaluate	Data processing/Statistical analysis	Purpose of the investigation	Main Findings	
						Benefits	Risk/Demands
Deathe et al. (1996)	11 subjects (7 male, 5 female); mean age: 64.5±15.8 years; mean weight: 69.9±11.5 kg; amputation level: 2 above knee, 9 below knee; Inclusion criteria: Unilateral amputation and walker dependents yet able to ambulate at least 5 m without needing to rest. Informed consent was signed.	- Instrumentation: Telemetry system, Strain gauges in each walker leg (3D ground force data); Foot Switches; Frequency: 125 Hz; - 3 height conditions: level of the patient's distal ulnar styloid, 3cm above and 3cm below; - Training session; - Walk test: walk 30 s; - 3 trials each condition;	COP; spatiotemporal: stance and swing time, % of full stride, full stride duration, prosthetic leg swing, mid-stance, intact leg swing, walker unload, walker swing, walker-Two leg stance and Walker-One leg stance.	- Force data: 2nd order critically damped low-pass digital filter (cut-off 10Hz); Used to calculate COP; - Parameters calculated for each phase and full walker stance; - Mean and standard variation; - Within-subject analysis of variance; - p<0.05	Examine the relationship between stability and walking frame height during ambulation.	Walker height does not influence the duration of walker strides; lower height alleviates the load on the prosthetic leg; patient with UE limitations needs a higher walker.	Walker height and increased horizontal forces applied to the walker may generate some instability; Lower height increased the demand on UEs.
Melis et al. (1999)	10 subjects (3 female, 7 male); mean age: 41±24 (24-72) years; Inclusion criteria: all subjects had sustained an incomplete SCI more than 1 year prior to the study, had completed their formal rehabilitation and could walk at least 20 m with their AD.	- Instrumentation: 16 strain gauges at the tip of each device leg [300Hz]; 3 reflexive markers on the devices; 7 reflective markers on the subjects weaker side; Video camera (60Hz); - 3 conditions: SW; elbow crutches and cane; - Walker height adjusted; - Walk test: walk over a 10 m flat pathway at self-preferred speed and walking style; - 5 trials each condition; - Rest periods;	Kinetics: Axial compressive force along the device shaft, the antero/posterior bending force, the medio/lateral bending force; device orientation; Spatio-temporal: gait speed, cadence, step length and stance/swing ration, kinematic: trunk, hip, knee and ankle angles.	- Data filtered at 10 Hz; - Angles were normalized to 100% of the gait cycle; - Mean and standard deviation; - Forces filtered at 25 Hz and normalized to user's body weight, lever arm and 100% of contact time; - Software: Ariel Performance Analysis System.	Determine the influence of walkers, crutches and canes on assisted-gait following SCI.	More stable; high force can be supported by walker (vertical support).	Slower cadence with walker; higher stance with walker; decreased hip excursion and reduce step length; flexed trunk posture.
Bachschmidt et al. (2001)	7 subjects (4 males, 3 females); mean age: 27.9±7.5 years; mean weight: 74.8±24.4; mean height: 174±64 cm; Inclusion criteria: no prior LE surgery, no history of stroke or other musculoskeletal pathology, no cardiovascular limitation, no prior use of a walker, right-handedness. Informed consent signed.	- Instrumentation: 6-camera (60 Hz) with reflexive markers on arms, legs, neck and pelvis; Strain-gauges on each walker leg and six sets on each handle; audio feedback system; walker dynamometer. - Calibration session of sensors; - 3 load conditions: 0%, 10% and 50% of body weight; - Walker height adjusted at the level of distal ulnar styloid; - Training session; - Gait pattern: three count, delayed, five-point-gait; - Walk test: walk 10 m at self-preferred speed; - 3 trials each condition;	Spatiotemporal: cadence, speed, stride length and stance-to-swing ratios; kinematic: joint angular positions, velocity and acceleration; Kinetic: walker loads, internal joint forces and moments.	- Force data: 15 Hz anti-aliasing filter; - Data were time normalized as a % of the walker stride cycle. - Mean and standard variation; - Differences of the maximum and minimums were analyzed with: Mann-Whitney U test and Wilcoxon Matched Pairs test; - Nonparametric statistics; - p<0.05	Study changes in UE kinetics that occurs with the use of a SW.	The arms of the walker take on the role of legs, supporting the body against ground reaction loads.	Demands on the elbow and shoulder joints.

Table A.7: Summary of the articles investigating the influence of the standard walker.(continuation)

Authors	Participants	Testing Protocol/Instrumentation	Parameters to evaluate	Data processing/Statistical analysis	Purpose of the investigation	Main Findings	
						Benefits	Risk/Demands
Ishikura (2001)	30 subjects (15 females, 15 males), mean age: 22.3±3.3 years; mean weight: 58.3±9.1 kg; mean height: 164.7±6.9 cm; Inclusion criteria: healthy subjects, who did not have any past history of disease causing functional disorders of the UEs and LEs or trunk. Informed consent signed.	Experiment 1: force plates (100Hz). - 2 conditions: with and without the walker with the hip joint at various flexion angles; - Gait pattern: Knee joints extended; UEs flexed to 90º at the elbow joints; - The walker height was adjusted; - Walk test: walk 10 m at self-preferred speed; - 10 trials; Experiment 2: EMG (500Hz) - 2 conditions: without walker and with walker with the hip joint at various flexion angles; - Measure muscle strength 10 times for 10 sec.	Kinematics: Hip flexion angle; Kinetics: vertical ground force, left-right ground reaction force, anterior-posterior ground reaction force; speed; muscle activation levels: left rectus and biceps femoris muscles.	- Vertical forces normalized; Hip joint flexion angle normalized regarding body weight and the duration of the standing period; the maximum values were obtained; - Electromyography data high-pass filtered at 20Hz; - 'Chi-square test'; - Spearman's correlation coefficient ; - One-way analysis of variance; - Fisher PLSD method; - p<0.05	Investigate the weight bearing force and its variations during the walker gait biomechanically; explore the possibility of applying the walker gait to the partial weight bearing gait; and the relationship between the hip joint flexion angle and muscle activation level during the walker gait.	Reduction of the load/weight of the LEs; walker supports great percentage of the body weight; high muscle activation level that maintains/increases muscle strength; the walker can be used for partial weight bearing gait.	Exaggerated hip flexion, yielding: reduced movement of the hip (without extension) and alteration of the tissues around the hip.
Bateni et al. (2004)	10 subjects; mean age: 23 (22-27) years; mean body mass: 63 (51-86) kg; mean height: 168 (158-181) cm; Inclusion criteria: right-handed, right leg dominant; no medical conditions or medication use affecting balance or limb movement; physically active and healthy. Informed consent signed.	- Instrumentation: Movable platform for postural reactions and provoke perturbations; subject wore a harness; video-camera (500Hz); load cells on the walker; - 5 conditions: walker, walker-top, cane, cane-top, without device; - Test: Subject with the device on the platform and push down the device with constant force; Perturbations were activated when the loading was within 5% of body weight of the desired level; - 11 trials with various perturbations; - Seated rest breaks.	Kinetics: walker axial forces; Spatial: stride length.	- Step length expressed as a % of body heigth; - ANOVA - Tukey's Studentized Range test; - p<0.05.	Study the potential of walkers to interfere with, or constrain, lateral movement of the feet and thereby impede execution of compensatory stepping reactions during lateral loss of balance.	Biomechanical stabilization (reaction forces generated by the user's hands); These forces prevent instability and recovering balance, in the event of a disturbance.	It can prevent lateral movement of the legs and consequently disabling the implementation of the 'stepping compensatory reactions' in situations of loss of balance.
Andersen et al. (2007)	40 subjects; mean age: 86.8±6.0 years; Inclusion criteria: subjects had not received any form of exercise intervention or physical therapy within the past year of the study, not presence of cardiovascular problems, free of COPD, renal problems and anemia. Informed consent signed.	- Two subjects' classifications: fallers/nonfallers and walker users and nonusers; - Subjects were administered a number of physical tests and questionnaires prior and after training: Short Form-36 and a simple fall inquire.	Total SF-36 score.	- Covariance analysis; - p<0.05; - Software: SPSS;	Study of the relationship between perceived health and walker use.	Affects positively both mobility and fitness levels.	Weakening of physical functioning and general health; These limitations can cause psychological and physical consequences.

Table A.8: Summary of the articles investigating the influence of the standard walker.(continuation)

Authors	Participants	Testing Protocol/Instrumentation	Parameters to evaluate	Data processing/Statistical analysis	Purpose of the investigation	Main Findings	
						Benefits	Risk/Demands
Vennila & Aruin (2011)	8 subjects (5 male, 3 female); mean age: 25±4 years; mean body mass 69 ± 11 kg; Inclusion criteria: normal vision, right handed and right-legged based on their self-report on using preferential hand and leg during daily activities, no neurological or musculoskeletal disorders. Informed consent signed.	- Instrumentation: Force platform; Accelerometer taped on the subject's left clavicle; EMG on muscles legs; strain-gauges; frequency sample: 1000Hz; - 4 conditions: without walker and with vision; with walker and with vision; without walker and without vision; with walker and without vision; - Walker height adjusted; - training session; - Test: maintain a vertical position; pendulum impact to cause perturbation; - 5 trials for each condition; - 3-min rest interval.	Kinetics: Ground forces and moments; EMG; kinematics: accelerometer signals.	- EMG signals: low-pass filter, 2nd order zero-lag Butterworth (10-500Hz), rectified and amplified (gain 2000); - Ground forces and moments were low-ass filtered (20Hz) 2nd order zero-lag Butterworth; - Mean and standard deviation; - Two-way repeated measures analysis of multivariance (MANOVA); - Two-way repeated measures ANOVA; - Bonferroni correction; - $p<0.05$; - Software: SPSS;	Investigate the availability of vision and additional support on anticipatory and compensatory postural adjustments and their interaction.	Improvement in postural control when no visual information is committed	-
Mcquade et al. (2011)	20 subjects (9 female, 11 male); mean age: 67 (49-85) years; mean height: 170 (150-190) cm; mean weight: 90 (56-110) kg; mean post-operative day: 9 (2-34) days; Inclusion criteria: hip or knee joint replacement surgery; good health, functional UEs range of motion, normal UEs' function and intact UEs sensation. All of them had their postoperative pain level relieved and controlled. Informed consent signed.	- Instrumentation: 3D gait analysis with three active infra-red cameras (100Hz); 2 strain-gauges transducers integrated on both handles (200Hz); - Walker height was adjusted at the level of ulnar styloid; elbow flexed 20-25 degrees; - Training session; - Gait pattern: Three-one-point; - Walk test: walk 10 steps at self-preferred speed; - 3 trials;	Joint force and moments: elbow, wrist and shoulder; 3D Euler angles.	- Collect 3 complete strides per trial and averaged the 9 trials; - The strides were normalized with regards to the step cycle; - Inverse dynamics; - Forces and moments were normalized regarding the body weight; - Filtered on 20Hz with a recursive 4th order Butterworth; - Paired t-test; - Ensemble average; - Spearman Rank Correlation; - Software: SPSS;	Describe the joint kinematics, forces and moments of the wrist, elbow and shoulder.	-	Compressive forces about 20% of the weight applied to each joint of the UEs; Increased load on the elbow, which leads to a higher requirement of the extensor muscles of the elbow and shoulder.

Table A.9: Summary of the articles investigating the influence of the two-wheeled walker.

Authors	Participants	Testing Protocol/Instrumentation	Parameters to evaluate	Data processing/Statistical analysis	Purpose of the investigation	Main Findings	
						Benefits	Risk/Demands
Youdas et al. (2005)	10 subjects (5 male, 5 female); mean age: 24.3±6.0 year; mean body mass: 70.5±9.7kg; mean height: 169.4±10.1cm; Inclusion criteria: no history of LEs surgery and minimal experience using assisted AD, normal and good muscle performance of the UEs. Informed consent signed.	- Instrumentation: 10 video cameras; 4 force plates (600Hz); - 4 conditions: unassisted; 2WW; auxillary crutches and forearm crutches; randomized order; - Gait pattern: 3 point partial weight bearing; - Training session to offload the right leg by 50%; - Walk test: walk 25m; - 5 trials for each condition;	Spatiotemporal: Stride and step length, speed, cadence, stance time and step width; kinetics: vertical ground reaction force.	- Average of 5 trials for each parameter and condition; - The vertical ground reaction force normalized by body weight; - Repeated-measures ANOVA; $p<0.05$; - Newman-Keuls post hoc comparisons; $p<0.01$ - Paired t tests with Bonferroni adjustment; $p<0.01$	Determine if subjects can offload the right LE to a targeted amount of weight bearing using ADs.	Greater base of support (comparing with crutches) and gives more support to unload the weight from the LEs than crutches.	Speed, cadence, step width and stride length are reduced.
Haubert et al. (2006)	14 subjects (11 male, 3 female); 5 with tetraplegia, 9 with paraplegia; mean age: 37 years; average time since SCI: 7 years; Inclusion criteria: able to walk a minimum of 15 m with walker and crutches; Exclusion criteria: UEs pain requiring medical intervention. Informed consent signed.	- Instruments: Telemetry system (2500 Hz); Footswitch (2500 Hz); 6-component load cells under the handles of the device (2500 Hz); 6 video-based infrared cameras (50 Hz); Reflexive markers taped on the devices and over specific bony landmarks on the UEs and trunk; - 2 conditions: 2WW and arm crutches; randomized order; - Devices height adjusted with elbows flexed 15º; - Walk test: 10-m walkway at a self-preferred speed; - 2 trials for each condition; - 3-min rest interval; - American Spinal Injury Association motor score.	Spatio-temporal: speed, cadence and stride length; kinetics: 3D forces applied on the device, shoulder joint forces, peak force, rate of loading, and force-time integral; kinematics: displacement of trunk, UEs, walker and crutches.	- Average of 2 trials for each parameter and condition; - An ensemble average of force data was calculated for each 1% of Gait Cycle (GT) from approximately 5 strides for each AD condition; - Displacements' data filtered : 4 Hz low-pass 2nd order recursive Butterworth digital filter; - Shoulder joint forces by inverse dynamics; - t-test; - Coefficient of multiple correlation; - Test nonparametric Wilcoxon signed-rank; - $p<0.05$; - Software: SPSS;	Compare 3-dimensional shoulder joint reaction forces and stride characteristics during bilateral forearm crutches and 2WW ambulation in persons with incomplete SCI.	Load on the shoulder by walker is lower than with crutches; increased cadence.	Increased load on the shoulder; decreased stride length.

Table A.10: Summary of the articles investigating the influence of the three-wheeled walker.

Authors	Participants	Testing Protocol/Instrumentation	Parameters to evaluate	Data processing/Statistical analysis	Purpose of the investigation	Main Findings	
						Benefits	Risk/Demands
Grant & Capel (1972)	5 subjects (male); age range 55-65 years; Inclusion criteria: pulmonary emphysema; walk less than 60 m and no evidence of heart disease.	- Instrumentation: Bag of oxygen and a respirometer place on the walker; - Walker height adjusted; - 4 conditions: unassisted and assisted air respiratory with and without 3WW; - Walk test: walk as far as possible at a self-preferred speed; - At the end of the exercise the subjects were asked why they stopped; - Rest break;	Spatiotemporal: speed and walking distance; metabolic measure: minute ventilation.	- Average and standard deviation of the parameters.	Evaluation of exercise capacity with a 3WW in people with pulmonary emphysema.	Double the walking distance with no respiratory aid.	-

Table A.11: Summary of the articles investigating the influence of the four-wheeled walker.

Authors	Participants	Testing Protocol/Instrumentation	Parameters to evaluate	Data processing/Statistical analysis	Purpose of the investigation	Main Findings	
						Benefits	Risk/Demands
Honeyman et al. (1996)	11 subjects; Inclusion criteria: diagnosis with COPD; distance covered in 6MW test less than 300m; unaccustomed of using a 4WW; Exclusion criteria: ambulation is limited by angina, claudication or arthritis. Informed consent signed.	- Instrumentation: Manual records, pulse oximeter; - 2 conditions: unaided and with the assistance of a 4WW; - Randomized crossover design; - Quiet and silence room; - Training session; - Test: perform 6MW test; - Walk test: walk in a 50 m long hospital corridor; - 2 trials each condition on each of 2 study days; - 1 hour rest in between tests.	Spatial: distance walked in 6 minutes: metabolic measures: change in oxyhemoglobin saturation during the walk, and breathlessness using a modified Borg Scale.	- Mean values of two walks for each individual; - Paired t-test; - p<0.05.	Evaluation of the effect of walker with wheels on disability, oxygenation, and breathlessness in patients with severe disability secondary to chronic irreversible airflow limitation.	Increase in 6-minute walking distance; significant reduction in hypoxemia with walking and a significant reduction in breathlessness during the walk test; improve quality of life in individuals with severe impairment in lung function.	-
Solway et al. (2002)	40 subjects; age range 55-85 years; Inclusion criteria: COPD diagnosis, clinically stable, unaccustomed to the use of a walking aid; Exclusion criteria: medical conditions that limited exercise tolerance or an inability to communicate in English.	- Instrumentation: respiratory inductance plethysmograph, volume spirometer, distance sensor, pulse oximeter, upper extremity strain gauges, and stride counter (100Hz); - Randomized crossover design; - 2 conditions: unaided and with the assistance of a 4WW; - Walker height adjusted at the ulnar styloid level; - Test: Two 6MW test; - 1 hour rest in between tests; - Walk tests: walk in a 60 m long corridor; Quiet and silence; - Subjects accompanied by the tester.	Metabolic measures: breathlessness using a modified Borg Scale, cardiorespiratory function and gait, respiratory rate, minute volume, rib cage and abdominal excursions; Spatio-temporal: distance walked in 6 minutes, stride length, cadence and speed; Kinetics: weight bearing by UEs.	- Mean and standard deviations for all parameters; - Two-tailed test, erro of 0.05 and power of 90%; - Multiple linear regression; - ANOVA; - Univariate regression; - Stepwise multiple regression; - p<0.05 - Software: LabVIEW;	Study the effects of using the 4WW on functional exercise capacity in patients with COPD; Characterization of patients who benefit most from the use of the 4WW.	Reduction in dyspnea, increase in walking ability and high sense of security; arms stabilization reduces the dyspnea.	-
Probst et al. (2004)	15 subjects; Inclusion criteria: diagnosed with COPD, no neurological or limitation in locomotion, not need for supplemental oxygen during the 6MW test, no practice in the use of AD. Informed consent signed.	- Instrumentation: telemetry system (metabolic data); pulse oximeter; breathing mask; - 6MW tests in two conditions: with and without 4WW; - Walker height adjusted at the ulnar styloid level and elbows flexed 30º; - Training session; - Walk tests: walk in a 53m corridor; Quiet and silence; - 30-min rest intervals;	Metabolic measures: breathlessness using a modified Borg Scale, calculation of minute ventilation, tidal volume, volume of oxygen, produced carbon dioxide, respiratory rate, level of carbon dioxide released at the end of expiration and the ratio between inspiratory time and total respiratory cycle; heart rate; Walking efficiency: ratio of volume of oxygen in the last minute of the test and the distance walked in 6 min; forced expiratory volume and forced vital capacity; speed.	- Non-parametric tests; - Wilcoxon signed-rank test; - Stepwise multiple regression analysis; - p<0.05 - Software: data analyzed by Metasoft and statistics in SAS.	Study the effect of using the 4WW on walking distance and physiological variables in patients with COPD.	Improves functional exercise capacity: less dyspnea, increased ventilation and increased distance; improvement in distance walked in 6MW test; reduction of hypoxia and shortness of breath; Increase walking speed; feeling of security and stability - this allows patients to walk faster with the same volume of oxygen.	-

Table A.12: Summary of the articles investigating the influence of the four-wheeled walker (continuation).

Authors	Participants	Testing Protocol/Instrumentation	Parameters to evaluate	Data processing/Statistical analysis	Purpose of the investigation	Main Findings	
						Benefits	Risk/Demands
Wellmon et al. (2006)	105 subjects; 3 groups: 35, mean age 79.9±4,5 years who did not use AD; 35, mean age 84.1±5.6 years who use cane; 35, mean age 87.8±5.5 years who use 4WW; Each group with 10 male, 15 female; Inclusion criteria: elder and capable of walk independently with and without the walker and intact hearing. Informed consent signed.	- Instrumentation: digital stop clock; wireless FM microphone; auditory stimulus of 1500 Hz buzzer; 2 infrared photocell switches and digital timer; - 2 conditions: with and without the 4WW; - Quiet and silence room; - 3 tasks: 1º measure VRT (Voice Reaction Time) stimulus in standing with 15 trials (single task); 2º Single-task walking speed: ambulate along a 6.10 m walkway without stimulus with 5 trials; 3º Dual-Task: Measurement of VRT while walking with 10 trials;	VRT; speed.	- ANOVA; - Mean; - Post hoc pairwise; - $p<0.05$ - Software: SPSS.	Study of attention required to walk with a device in the elderly. Execution of two tasks.	Easy to use.	Greater delay in response to the 2nd task with the 4WW than without it. Increase the attentional requirements.
Alkjaer et al. (2006)	7 subjects (female); mean age: 34.7 years; mean height: 170 cm; mean weight: 64.7 kg; Inclusion criteria: any history of injuries or muscle skeletal dysfunction in their LEs. Informed consent signed.	- Instrumentation: 15 reflexive markers (model by Vaughan, Davis, & Connor, 1999); 5 video cameras (50Hz); 2 force platforms (1000Hz); Velocity controlled by photocells; - Walker height adjusted, with the handles on a level with ulnar styloid; - Training session; - 2 conditions: with and without the walker; - Walk test: walk at a 4.5km/h speed on the force plates.	Kinematics: Internal flexor and extensor joint moments (ankle, knee,hip joints), hip adductor/abductor moment, angular impulse and their peak values (plantar flexors, knee extensors, hip extensors, flexors and abductors), angular position (ankle, knee, hip joints) and range of motion of joint angles.	- Position data low-pass filtered: 4th order Butterworth filter, cut-off 6Hz; - Only data from left leg during stance was analyzed; - 6 gait cycles normalized by interpolation and averaged for each condition and subject; - Joint moments normalized to body mass; - Ensemble averages; - Student's t-test; - $p<0.05$; - Software: MatLab;	Study of the biomechanical effects of walking with and without 4WW on a walking pattern of a healthy subject.	Reduction in knee moment; support the weight of the trunk; knee extensor energy absorption is reduced; joints' range of motion is reduce;	Increased angular impulse of the hip extensors and duration of the hip extensor moment; hip joint more flexed during stance, due to trunk forward flexion.
Gupta et al. (2006)	31 subjects; Inclusion criteria: post rehabilitation patients with COPD, clinically stable, had not had previous 4WW use an unassisted 6MW distance <375m; Exclusion criteria: medical conditions that limited exercise tolerance and inability to communicate. Informed consent signed.	- Randomized controlled study design; - Subjects were randomized to: 4WW (18 subjects) or non4WW (13 subjects) groups for 8 weeks; - 4WW group had to integrate walker in their daily life; At the end of the study period indicate how the 4WW had impact on the activities; - Chronic Respiratory Questionnaire (CRQ) at baseline, 4 weeks and 8 weeks; - Test: Two 6MW test for each subject and group and Sickness Impact Profile).	Distance in 6 minutes; Chronic Respiratory Questionnaire answers.	- Variance analysis for repeated measures; - Mean and standard variation.	Study of the effect of the use of the 4WW in the quality of life in patients with COPD.	Reduction in dyspnea, increase in walking ability and high sense of security.	-

Table A.13: Summary of the articles investigating the influence of the four-wheeled walker (continuation).

Authors	Participants	Testing Protocol/Instrumentation	Parameters to evaluate	Data processing/Statistical analysis	Purpose of the investigation	Main Findings	
						Benefits	Risk/Demands
Liu et al. (2009)	33 subjects (18 TUR and 15 PUR); Inclusion criteria: age > 65 years, residing in an assisted living facility, no medical history of impairments, Score MMSE>24, able to walk 9.15m indoors with 4WW, no previous diagnosis of stroke or neurological condition such as PD, history of at least one fall in the last year (PUR). Informed consent signed.	- Instrumentation: GAITRite® (5.18 m x 0.88 m); - Questionnaire about the 4WW; - Walker height adjusted at the level of the wrist crease line; - 2 conditions: with and without walker; - Gait pattern: upright posture and maintain the body between the two posterior legs of the walker; - Training session; - Walk test: walk over the carpet and wore safety belts; - Quiet and silence room; - PUR: 3 trials with walker; - TUR: 3 trials with and 3 trials without the walker; - 3-min rest interval; - Eliminate acceleration and deceleration effects, by starting and finishing 2 m after and before the walkway.	Spatiotemporal: cadence, speed, swing and stance time, double support time, step and stride length.	- Mean and standard variation; - ANOVA; - Paired samples t-test; - p<0.05 - Software: GAITRite GOLD and SPSS.	Study of differences in gait parameters while walking among RUs and potential users.	Reduction of the load on the LEs.	Decreased cadence, walking speed, swing time, step length and stride length and increased stance and double support times, because of the afraid to fall; posture changes.
Vogt et al. (2010)	90 subjects; Inclusion criteria: age > 65 years, independent mobile, able to understand and follow study instructions, able to perform TUG (time up-go), FTSST (five times sit to stand test) and POMA-B (performance-oriented mobility assessment - balance); Exclusion criteria: presence of acute neurological impairments or other severe physical or psychiatric illness.	- Quase-experimental pre- and postdesign; - 3 groups: first-time users (30), long-term users (30) and control subjects (30); Selection process in detail (Vogt et al. 2010); - MMSE exam; - 12 sessions, 30 min each of physiotherapy; 6 sessions, 30 min each of ergotherapy; 13 sessions, 20 min each of ergometer exercises; - Test: TUG, FTSST and POMA-B tests.	TUG, FTSST and POMA-B results; difference between the admission and final score.	- Mean and standard variation; - Non-parametric Krustal-Wallis test; - p<0.05	Examine the effect of 4WW use on functional rehabilitation outcome in geriatric patients.	Maintains or improves motor skills at the highest level possible; Increased confidence and safety for the user; Does not interfere with rehabilitation; Improvement in balance and mobility.	-
Takanokura (2010)	600 subjects; age >70 years; mean height: 1.609 m (male) and 1.472 m (female¬); mean weight: 59.3 kg (male) and 49.5 kg (female).	- Mechanical model (Takanokura 2010); - Instrumentation: Push-pull digital force gauge; - Walk test: walk on dry asphalt and dry gravel roads with the walker; - Posture: stretched UEs and elbows not bended.	Model parameters: Friction coefficient; walker mass; subjects' weight and height; handgrip height; trunk flexion angle; normal force at handgrip; muscular force at elbows, shoulders, chest and to support user's body; extension and flexion force of upper body muscles; objective function.	- Regression analysis; - Mechanical model results for optimization.	Optimization of the height of the 4WW (handgrip) by a 2D mechanical model to reduce the muscular loads in the UEs and LEs with various road conditions during a steady displacement.	Higher walker and push it in the perpendicular direction by leaning their upper body on the walker helps to maintain upright posture to maintain gait pattern characteristics and relieve muscular loads in the upper and lower body.	Requirement of a correct adjustment of the height of the 4WW and body posture.

Table A.14: Summary of the articles investigating the influence of the four-wheeled walker (continuation).

Authors	Participants	Testing Protocol/Instrumentation	Parameters to evaluate	Data processing/Statistical analysis	Purpose of the investigation	Main Findings	
						Benefits	Risk/Demands
Schwenk et al. (2011)	109 subjects; mean age: 83.1 years; Inclusion criteria: no severe cognitive impairment (MMSE>17), allowed full weight-bearing post-operatively, ability to walk at least 4 m without a walking aid, use of a 4WW at the beginning of rehabilitation, no uncontrolled neurological, cardiovascular, or metabolic disorders, no severe visual defects. Informed consent signed.	- Instrumentation: GAITRite® system (4.9 m); - Prospective, longitudinal study; assessment at the beginning and prior to discharge of a 3-week rehabilitation period; - 2 conditions: with and without the walker; - Tests: POMA and TUG; - 1 trial for each condition and test.	Spatiotemporal: speed, cadence, stride time, stride length, base of support and double support/stride time; POMA and TUG results-	- t-test or Man-Whitney U test; - paired t-test or Wilcoxon test; - Standardized response mean; Cohen's criteria; - $p<0.05$ - Software: SPSS.	Evaluate the influence of using the 4WW in the assessment of gait and mobility;	Compensate defects in gait and mobility.	-
Miyasike-daSilva et al. (2013)	14 subjects:12 subjects (6 female, 6 male), age range 20-39 years and 2 subjects (1 female, 1 male) with ages of 71 and 72 years; Inclusion criteria: no gait impairments.	- Instrumentation: digital voice recorder; speakers; head-mounted microphone; Digital camera; - Single task: walk; - Double task: walk and repeat sounds; - 4 conditions: with and without walker on a narrow level ground and on a wooden beam; - 2 experiments: 4 conditions with single task and double task; - Walk test: walk 6 m; - 6 trials each condition and experiment.	Speed; voice reaction time; number of missteps to recover balance.	- Three-way ANOVA; - Two-way ANOVA; - $p<0.05$	Study the effect of using the 4WW in gait and attentional demand in conditions of possible imbalance.	Reduction of attentional demand and increased stability of gait in conditions where the equilibrium is 'challenged' (narrow surface); Normal gait speed.	-
Tung et al. (2014)	1st experiment: 11 subjects (6 female, 5 male); age range: 20-39 years; Inclusion criteria: no gait impairments. 2nd experiment: 20 subjects; mean age: 89.1 years (10 RU) and 83.1 years (10 non RU); Inclusion criteria RU: use of 4WW to perform daily mobility activities, arthritis affecting the UEs if pain not experienced when using the walker, and ability to follow two step commands in English; Exclusion criteria RU: palliative care, had uncorrected vision, experienced significant pain in standing or moving for brief periods, had diagnosed pathology which severely affected physical function, or were identified as a frequent faller (>2 falls in 1 month).	1º experiment: - Instrumentation: Optical beam switches; Video-camera; Strain gauges on each walker leg; - 2 condition: with and without walker; - 2 tasks: normal walking (NORM) and challenged balance (BEAM); - BEAM: walk along a narrow wooden beam; - Walker height adjusted at the level of radial styloid; - Walk test: walk 6 m walkway at self-preferred speed; - 4 trials per condition and task; 2nd experiment: - Instrumentation: GaitRite® (4 m); - Berg Balance Score; - 2 condition: with and without walker; - Training session; - Walk test: walk 4 m walkway at self-preferred speed; - 2 trials each condition; 15 steps; - Walker height adjusted; - The testers walked alongside subjects;	1º experiment: Spatio-temporal: speed, number of missteps to recover balance and cadence; kinetics: moments of UEs, axial force on the walker; COP. 2º experiment: kinetics: COP, axial force; Spatio-temporal: speed, cadence, step length, step width and step width variability; Berg score.	1º experiment: - Strain gauges signals filtered: low pass filter 2nd order Butterworth (cut off: 10Hz); - COP calculated through force data on the walker; vertical force equal to the sum of the 4 forces; - Mean; - Paired t-test; - ANOVA; - Tukey's least test; - $p<0.05$. 2º experiment: - COP and vertical force equal to the 1st experiment; - Unpaired t-tests; - Mean and standard variation; - One-way ANCOVAs; - $p<0.05$.	Characterize the manner in which the UEs may be used for balance control during walking with a 4WW, and investigate the consequences of using the UEs for balance control on walking with a 4WW performance.	During assisted gait UEs play a key role in the control of balance when the frontal plane is affected by compensating for the limitations of the LEs.	Increased use of UEs when walking under challenge conditions (narrow surface).

Appendix B

Hospital Ethical Committee Approval

CONSENTIMENTO INFORMADO, LIVRE E ESCLARECIDO PARA PARTICIPAÇÃO EM INVESTIGAÇÃO

de acordo com a Declaração de Helsínquia e a Convenção de Oviedo

Por favor, leia com atenção a seguinte informação. Se achar que algo está incorrecto ou que não está claro, não hesite em solicitar mais informações. Se concorda com a proposta que lhe foi feita, queira assinar este documento.

Título do estudo: Estudo e desenvolvimento de dispositivos externos de apoio à marcha com aquisição de sinais biomédicos.

Enquadramento: Hospital de Braga em conjunto com a Professora Cristina Santos e a Engenheira Maria Martins do Departamento de Electrónica Industrial da Universidade do Minho.

Explicação do estudo: Os participantes abrangidos pelos critérios de seleção previamente definidos serão recrutados no Hospital de Braga e classificados pela médica fisiatra (Dra. Catarina Matias). O grupo de selecionado irá realizar o treino de marcha com o andarilho motorizado. O protocolo experimental será detalhadamente esclarecido ao participante, que, optando pela participação, deverá concordar e assinar o Termo de Consentimento Informado, Livre e Esclarecido para participação em investigação. Todos os participantes realizarão um programa de reabilitação prescrito pelas médicas fisiatras, ajustado à sua condição clínica atual, iniciando o treino de marcha assim que apresentarem capacidade para tal. Os doentes serão avaliados periodicamente pelas médicas fisiatras, sendo também avaliado o equilíbrio em ortostatismo com recurso ao andarilho monitorizado e aos sensores integrados, pelas engenheiras, com apoio da fisioterapeuta. Os doentes serão também submetidos a avaliação clínica periódica, utilizando as escalas de Berg e ABC e serão também avaliados com o andarilho motorizado, avaliando equilíbrio estático e dinâmico, base de suporte, tamanho do passo, distância percorrida, simetria, harmonia e coordenação durante o treino de marcha com o andarilho. O processo de aquisição de dados será transparente ao utilizador. Durante a realização das avaliações haverá sempre uma pessoa próxima para os devidos esclarecimentos, acompanhando toda a pesquisa. Os participantes serão fotografados e filmados. As fotografias e videos serão divulgadas para fins académicos e meios científicos e para isso o rosto será tapado para garantir o sigilo. Os resultados obtidos durante a pesquisa serão colocados em forma de gráficos e imagens.

Condições e financiamento: Este é um estudo de carácter voluntário da participação e não existem quaisquer prejuízos, caso não queira participar. Este estudo mereceu um Parecer favorável da Comissão de Ética do Hospital de Braga.

Confidencialidade e anonimato: As informações obtidas serão mantidas em sigilo e não poderão ser consultadas por pessoas leigas sem a prévia autorização por escrito do participante. As informações, assim obtidas poderão ser usadas somente para fins estatísticos ou científicos, sempre resguardando a privacidade do participante.

Assinatura/s: … … … … … … … … … ... … … … …... … … … … … … …
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Hospital
Braga

N/ Referência: 73/CE-2013

Data: 16/08/2013

V/ Referência: 004/CCA-2013

C/C:

Exma. Senhora
Professora Cristina Santos
Departamento de Eletrónica Industrial
Escola de Engenharia
Universidade do Minho
Campus de Azurém
4800-058 Guimarães

ASSUNTO: **Autorização para a realização do Estudo denominado: Estudo e desenvolvimento de Dispositivos Externos de apoio à marcha com aquisição de sinais biomédicos.**

Exma. Senhora,

O Hospital de Braga- Escala Braga- Sociedade Gestora do Estabelecimento, S.A., vem pelo presente comunicar que autoriza a realização do Estudo denominado " Estudo e desenvolvimento de Dispositivos Externos de apoio à marcha com aquisição de sinais biomédicos", nas suas instalações, sitas no Hospital de Braga, Lugar de Sete Fontes, São Victor, 4710-243, Braga.

Com os melhores cumprimentos,

O Diretor Clínico

Fernando Pardal

Fernando Pardal
Diretor Clínico

Hospital de Braga – Escala Braga – Sociedade Gestora do Estabelecimento, S. A.
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Figure B.1: Ethical Approval.

Appendix C

Physical Therapy of Patients with Ataxia - Table Review

Author/yr	Study Design	Purpose	Participants	Intervention					Outcomes	
				Evaluation	Rehabilitation Session	Duration	Frequency	Comparison	Measures	Main Findings
Balliet et al. (1987)	Case study	This study presents an ambulation retraining approach designed to facilitate coordination and balance while minimizing upper extremity weight-bearing.	5 Adults with Central Nervous System (CNS) damage, including clinical signs of cerebellar dysfunction (4 traumatic brain injury; 1 cerebellar dysfunction)	Pre-intervention and immediately post-intervention (3 months-2 years)	Standardized activities commenced with isolated leg movements in sitting, progressing to static and dynamic standing balance activities and assisted walking with graded external support.	1h	2 sessions per week for 3 months; 3 of the 5 participants continued therapy for 1 session every 2 months until independently walking (up to 21 months).	N/A	Type of gait aid used; amount of upper limb weight bearing on gait aid; level of assistance required for walking; distance traversed without rest.	Improvements in all categories; progressed to a single assistive device (e.g. stick), standby level of assistance to walk or better, walk at least 300m.
Dordal (1986)	Case Study	The effects of an intensive mobility training in the phase of late rehabilitation are reported in two case studies.	2 severe ataxia. Patient with ataxia unable to stand or walk without support, difficulty using wheelchair.	Every day.	Training: dynamic graded exercise, stretching, indoor sports, 'special' walking exercises, mobility training, swimming, static bike, trike outdoors, task specific practice.	N/A	Individually tailored 1: 7/12, 225 hrs of training ~ 8 hrs a week, 2: 10/52193 hrs of training ~ 20 hours a week.	N/A	Video analysis, photographic anthropometric assessment of posture, bicycle ergonometry (heart rate), light track measure of simple movements, descriptions of functional tasks, quality of movement and posture.	Improvements in standing without support, functional walk (3m), 20m independent walk, floor-stand-walk, improved swimming and diving, improved quality and speed of movement.
Folz and Sinaki (1995)	Clinical Trial	The study was conducted to facilitate erect posturing in patients with neurodegenerative disorders through use of the Posture Training Support (PTS).	19 Adults with CNS pathology, including 5 with cerebellar dysfunction.	Immediately pre- and post- application of the device and 3-month follow-up.	One-off physiotherapy session for prescription of a postural training support device. PTS was applied posteriorly in subjects with a tendency to lean forward and anteriorly in patients with a tendency to fall backward.	N/A	1 session	Between-groups comparison determined whether the effectiveness of the device varied according to diagnosis: amyotrophic lateral sclerosis (n=7); parkinsonism (n=7); cerebellar dysfunction (n=5).	Subjective assessment of standing posture; self-reported independence in gait.	All of the patients, except for one, reported an improvement in gait.

Author/yr	Study Design	Purpose	Participants	Intervention					Outcomes	
				Evaluation	Rehabilitation Session	Duration	Frequency	Comparison	Measures	Main Findings
Jones et al. (1996)	Wait list con-trolled compari-son study	The aim of the study was to evaluate the effectiveness of therapy intervention in reducing impairment and disability due to upper limb and trunk ataxia in chronic multiple sclerosis (MS).	37 Adults with MS, presenting with clinical signs of cerebellar dysfunction.	Baseline (one-off, pre-intervention) and immediately post-intervention (2 weeks).	Customized inpatient physiotherapy to improve balance and independence.	30 min	8 sessions, 2 weeks.	Control group received no physiotherapy intervention.	Assessments used included the Jebsen Test of Hand Function (JTHF), the Kurtzke Functional Systems Scale (FSS) and Expanded Disability Status Scale (EDSS), the Northwick Park Index of activities of daily living (NPI), and patient and assessor visual analogue scales.	The findings support the clinical impression that therapy used to improve dynamic posture and methods of performing functional tasks can result in improvement of functional ability in patients with MS where spontaneous improvement would not otherwise be expected.
Gill-Body et al. (1997)	Case Study	This case report describes two individualized treatment programs and documents functional improvements in two patients with different etiologies, durations, and clinical presentations of cerebellar dysfunction.	Tumor, Xanthomatosis: (1) Stay 30s, 7.62 cm. TUG = 14.34s. Standing on the dominant leg: 30s. Increased postural sway: posterior direction. (2) Turns slowly; forward bent posture; Stay 30s, 7.62 cm; standing on the dominant leg: 15s; TUG=18s.	Baseline (one-off, pre-intervention) and immediately post-intervention (6 weeks).	Customized physiotherapy. Treatment included 3 phased with increased demands: velocity of head movements; Eye movements; static stance; base width; gait speed; types of trajectory; arms position; type of base of support; stretching exercises.	30-45 min.	5 times a week during 6 weeks. Daily home exercise program.	N/A	Gait parameters (velocity, base of support (BS), double limb support time (DS), and COG–COP moment arm) Timed Up and Go (TUG), Dynamic Gait Index, Self-reported independence in functional activities.	(1) Decrease postural sway; no change TUG; Increase 20%speed; 36% BS decreased; 28% DS decrease. (2) Standing on the dominant leg: 3s; 10% faster; 24% BS decreased; 28% DS decrease; TUG=14s. People with long term cerebellar dysfunction can improve postural stability.
Gillen (2000)	Case Study	This case report describes occupational therapy interventions focused on improving the activities of daily living (ADL) performance.	1 Ataxia syndrome (motor control deficits, visual dysfunction, gait disturbance) secondary to multiple sclerosis. Unable to care for himself, severe upper limb tremor	Baseline (one-off, pre-intervention) and immediately postintervention (5 weeks).	Treatments focused on task-specific training in basic ADL incorporating the occupational therapy modalities of orthotics, environmental adaptation, adaptive equipment prescription, and movement retraining. The basis for all interventions was decreasing the degrees of freedom required to participate in each task while simultaneously decreasing postural requirements.	90 min	Daily occupational therapy plus physical therapy. 5 weeks.	N/A	FIM items and personal goals; Feeding, grooming, bathing, upper body dressing, lower body dressing, bladder and bowel management, sexual activity, instrumental activities of daily living.	Feeding, grooming, bathing, bladder management improved from 1-2 to 6 (FIM) indicative of less assistance, sexual activity and ADL goals were achieved.

Author/yr	Study Design	Purpose	Participants	Intervention					Outcomes	
				Evaluation	Rehabilitation Session	Duration	Frequency	Comparison	Measures	Main Findings
Karakaya et al. (2000)	Before and after compari-son study	This study was planned to investigate and compare the effects of acute rehabilitation program on balance and coordination problems in patients with posterior fossa and cerebellopontine angle tumours.	40 Adults with cerebellar dysfunction due to cerebellar or brainstem tumor.	At admission to and discharge from rehabilitation.	Customized inpatient physiotherapy including: Frenkel exercises and balance training.	N/A	Daily treatment, 5 days per week until discharge (minimum. 2 weeks).	Between-groups comparison investigated whether rehabilitation outcome differed for participants with posterior fossa (n=20) versus cerebropontine angle (n=20) tumour.	Customized evaluation of gait ataxia and limb ataxia. Static balance tests (rated using Mokken's ordinal scale).	Both groups improved overall balance and standing balance. Those with cerebellopontine angle tumors made more improvement than the posterior fossa tumor group. Provides some evidence that balance and co-ordination improved in this population with rehabilitation. Those with more aggressive tumours (posterior fossa) had a less favourable rehabilitation prognosis.
Armutlu et al. (200)	RCT	This study was planned to investigate the efficacy of neuromuscular rehabilitation and Johnstone Pressure Splints (JPS) in the patients who had ataxic multiple sclerosis.	MS, predominantly ataxic, slight muscle weakness, walk unassisted.	Before starting. All assessments were repeated after 4 weeks of treatment.	The principles during the treatment were as follows: patients received treatment in the morning; the activities were selected from easier to harder; and frequent intervals were given to avoid fatigue. The control was given neuromuscular rehabilitation, Frenkel coordination exercises, approximation antagonists/agonists, static & dynamic balance training, training semi-tandem and tandem, single limb stance on balance board, Cawthorne-Cooksey, walking on uneven ground. Progression from eyes open to eyes closed. The study group was treated with JPS in addition.	>20 min	3 days a week for 4 weeks. All patients also were given a home exercise program consisting of active exercises.	Control n =13; Intervention n = 13.	Sensory assessment, step width, walking velocity (3m), Ambulation Index, anterior balance (Lovet- Reynold method), and other balance scales.	JPS demonstrated no additional benefit compared to 'conventional therapy'.
Crowdy et al. (2002)	Before and after proof of principle study	Verify if eye movement improves visuomotor performance in cerebellar patients.	N=2, cerebellar degeneration.	Baseline recordings of eye movements and locomotor performance (5 walks to 18 footfall targets along a walk way).	Verbal instruction to concentrate on making accurate eye movements to the footfall target rather than accurate steps plus eye movement rehearsal to first 6 targets. Followed by 3 further test walks.	N/A	N/A	N/A	Step phase duration (stance time, double support time), saccadic eye movement, % of double, triple or quadruple saccades.	Marked improvement in oculomotor and locomotors performance; increased regularity and accuracy of stepping and increased proportion of single saccades i.e. reduced saccadic dysmetria.

Author/yr	Study Design	Purpose	Participants	Intervention					Outcomes	
				Evaluation	Rehabilitation Session	Duration	Frequency	Comparison	Measures	Main Findings
Permutter and Gre-gory (2003)	Case Study	Comprehensive inpatient rehabilitation on a woman with paraneoplastic cerebellar degeneration.	1 woman with paraneoplastic cerebellar degeneration.	At admission to and discharge from rehabilitation (3 weeks).	Inpatient physiotherapy including rhythmic stabilisations (PNF), weighted vest and relaxation techniques.	N/A	Daily treatment for 3 weeks.	N/A	FIM scores: transfers and bed mobility.	FIM improved from 2 to 4 for transfers and from 1 to 4 for standing, from 1 to 5 for bed mobility and sitting balance. This case demonstrates the functional improvements of one patient with paraneoplastic cerebellar degeneration after intensive rehabilitation.
Harris-Love et al. (2004)	Case Study	Description of the rehabilitation management during a 12-month period of a 14-year-old female with Friedreich ataxia.	1 Friedreich Ataxia, assistance of 1 to stand, used powered wheelchair, walking frame; Sit independently; reach moderately.	Start of intervention and 12 months later.	UE bimanual task oriented training, LE stretching, functional strengthening (hip and trunk), gait training. Replaced rollator wheeled walking with U-step walking stabilizer (USWS).	60 min	1x month plus 1x per quarter. 1year.	N/A	Manual muscle testing, passive ROM, NHPT, SLST, gradation of force, gait speed, DLST, step length asymmetry, step time asymmetry.	NHPT, SLST, manual muscle testing showed minimal changes. Gait speed decreased by 69.4% concomitant with a 43.7% increase in force variability. Provision of the USWS improved gait performance (speed and reduced falls).
Gialanella et al. (2005)	Retrospective case series	Verify if the presence of extra-cerebellar stroke lesions negatively affects walking and functional capacities recovery of patients with cerebellar stroke.	48 Adults admitted to rehabilitation with cerebellar dysfunction secondary to stroke.	At admission to and discharge from rehabilitation.	Customized Physiotherapy based on the Bobath concept (neurofacilitatory approach).	1h	1 session per day, 5 days per week until discharged from rehabilitation.	Between-groups comparison investigated whether rehabilitation outcome for participants with isolated cerebellar stroke (n=20) was affected by the presence of extra-cerebellar lesions (n=23).	Trunk Control Test Lindmark Scale for Walking Rankin Scale.	Functional recovery present in both groups, those with isolated cerebellar stroke improved more than those with cerebellar and extra-cerebellar involvement.

Author/yr	Study Design	Purpose	Participants	Intervention					Outcomes	
				Evaluation	Rehabilitation Session	Duration	Frequency	Comparison	Measures	Main Findings
Hammer et al. (2005)	Single-subject experi-mental design (type A-B-A)	To investigate whether therapeutic riding (TR) hippotherapy (HT) may affect balance, gait, spasticity, functional strength, coordination, pain, self-rated level of muscle tension (SRLMT), activities of daily living (ADL), and health-related quality of life.	11 MS	13 times during program. 3 phases, namely, baseline phase A1 of 35 weeks (the pretreatment period), phase B of 1011 weeks' intervention, and finally, a second baseline phase A2, of 34 weeks.	TR and HT	30 min	10 weekly sessions- 16-18 weeks.	N/A	Berg balance scale (BBS), walking a figure of eight, TUG, 10m walking, the modified Ashworth scale, the Index of Muscle Function, the Birgitta Lindmark motor assessment, part B, and individual measurements. Self-rated measures were: the Visual Analog Scale for pain, a scale for SRLMT, the Patient-Specific Functional Scale for ADL, and the SF-36.	Balance and Role-Emotional were the variables most often improved, but TR/HT appeared to benefit the subjects differently. BBS increased from 27 to 44.
Schuhfried et al. (2005)	Double-blind, random-ized con-trolled trial.	To examine whether a whole-body vibration (mechanical oscillations) in comparison to a placebo administration leads to better postural control, mobility and balance in patients with multiple sclerosis.	12 MS patients with moderate disability (Kurtzke's Expanded Disability Status Scale 2.5/5) were	15 min, one week and two weeks after the application.	In the intervention group a whole-body vibration at low frequency. In the placebo group a Burst-transcutaneous electrical nerve stimulation (TENS) application on the non-dominant forearm.	5 min	five series of 1 min each with a 1-min break between the series.	MS allocated either to the intervention group or to the placebo group.	Posturographic assessment using the Sensory Organization Test, TUG and the Functional Reach Test immediately preceding the application.	The results of this pilot study indicated that whole-body vibration may positively influence the postural control and mobility in MS patients.
Stoykov et al. (2005)	Case Study	To examine the effectiveness of postural training on upper extremity performance in an ataxic individual.	3 years post left midbrain hemorrhage, dependent for all ADL. Right hemiplegia, severely ataxic. Barthel ADL index = 0.	Before-after intervention.	Course of postural training. Progressive tailored programme included; training for sitting balance, passive ROM, rolling, dynamic balance activities in sitting, scapular strengthening exercises caregiver training.	1h	3x a week, during 4 weeks.	N/A	Fugl-Meyer Upper Extremity Motor Scale (FMUEMS) Postural Assessment Scale for Stroke Patients (PASS).	FMUEMS improved from 35/66 to 53/66 PASS improved from 2/36 to 7/36 Barthel remained at 0. Increased time of comfortable sitting sufficient to support participation in chosen activities. The patient demonstrated an increase in function of the ataxic limb, as evidenced by appreciable increases in the Fugl-Meyer score and modest increases in PASS score.

Author/yr	Study Design	Purpose	Participants	Intervention					Outcomes	
				Evaluation	Rehabilitation Session	Duration	Frequency	Comparison	Measures	Main Findings
Brown et al. (2005)	RCT	To compare body weight support treadmill training (BWSTT) to conventional overground gait training (COGT).	Twenty subjects with chronic traumatic brain injury (TBI).	Baseline and after intervention.	Control: conventional over ground training. Intervention: BWSTT, body weight support gradually reduced from 30% to 10%, +/- physical assistance from up to 3 physiotherapists, gait speed increased as tolerated. No over ground practice.	15+30 min	2 days per week, for 3 months.	10 control, 10 intervention	Functional Ambulation Category (FAC), Functional Reach (FR), TUG; gait velocity, step width (BOS) and step length differential.	Trends towards significant improvement favoured the control group.
Oliveira and Freitas (2006)	Case Study	An SCA patient was submitted to a learning-based specific motor training in view of improving balance and functional independence in daily activities.	1 Spinocerebellar ataxia	Base line and after the 16-week program.	Functional training directed to the task, using simulations of the activities performed in day-to-day, static and dynamic balance training, positions of transfer of training, gait training in various environment, range of training and fine movements.	50 min	3x per week/ 48 sessions (16 weeks).	N/A	Barthel index and BBS.	Increase in balance (37.8%) and in the Barthel index (35%) (65 to 90). Motor control gains were transferred to daily activities, thus showing the proposed training is a valid therapeutic option in the treatment of SCA. BBS 23 to 44.
Smedal et al. (2006)	Single-subject experi-mental study design with ABAA phases.	Evaluate physiotherapy based on the Bobath concept, applied to MS patients.	2 Relapsing–remitting MS in stable phase.	12 times, three at each phase: A (at baseline); B (during treatment); A (immediately after treatment); and A (after two months).	Individually tailored treatment: 1: postural stability and orientation, dynamic activities designed to experience movement through postural adaptation. 2: mobilisation of LE in preparation for weight bearing and facilitation of trunk control and activity.	1h	5x a week, 2 months	N/A	BBS, gait parameters; velocity, SL, double stance phase as a % of gait cycle (at 3 speeds), TUG, RVGA, self report (VAS) perceived gait problem, Borg exertion scale, RMI, PGIC, CGIC.	Both improved on TUG (22 to 18.8; 11 to 8.9), BERG (42 to 45, 38 to 46) and RVGA and Reported improvements in balance and gait. Speed (0.5 to 0.7 and 1.2 to 1.3). DS – 44% to 47% and 47 to 25%.

Author/yr	Study Design	Purpose	Participants	Intervention					Outcomes	
				Evaluation	Rehabilitation Session	Duration	Frequency	Comparison	Measures	Main Findings
Brown et al. (2006)	Retrospective case series.	To determine if vestibular physical therapy (PT) leads to improved functional outcomes in people with central vestibular dysfunction.	48 Adults with vestibular disorders due to CNS pathology. 22 patients used an assistive device.	Baseline (one-off, pre-intervention) and immediately postintervention.	Customized Physiotherapy comprising: balance and gait training, general strengthening and stretching exercises, and vestibular habituation training.	N/A	The duration of physiotherapy treatment was individually tailored, (2–12 sessions). Cerebellar participants attended an average of 4.5 treatment sessions.	Between-groups comparison: stroke (n=10); cerebellar dysfunction (n=11); traumatic brain injury (n=5) and idiopathic vestibulopathy (n=22).	Dynamic Gait Index, TUG, Five times sit-to-stand test, Activities-specific Balance Confidence Scale DHI.	Participants remained at risk of falling and had reduced confidence in their balance abilities.
Baram and Miller, 2007	Prospective study	To study the use of auditory feedback for gait management and rehabilitation in patients with Multiple Sclerosis (MS).	14 MS/ 11 control group.	The ambulation parameters, baseline walking speed and stride length were measured first, without the device. Walking speed and stride length were measured four times with the device. Parameters were again evaluated without the device.	Auditory feedback during gait.	N/A	1 session	Control group and intervention	walking speed (meters/second) and stride length (meters).	For the patient group, the on-line walking speed improved, on average, 12.84% (standard deviation 18.74%). On-line average improvement in stride length was 8.30% (standard deviation 11.87%). Audio-based assistive technology may have special importance in reinforcement of functions in patients who suffer from visual impairment.
Cernak et al.(2007)	Case Study	This case report describes the effects of locomotor training using body-weight support (BWS) on a treadmill and during overground walking on mobility in a child with severe cerebellar ataxia who was nonambulatory.	Nonambulatory girl with severe cerebellar ataxia following a brain haemorrhage 16 months previously. The patient required a tracheotomy and feeding tube, dependent in all activities of daily living, and used a wheelchair for in-home and community mobility.	Before the beginning of clinic training, immediately after completion of clinic training, 1 month after completion of clinic training, and after the completion of 4 months of home training.	Partial BWS treadmill training (30%-10%), with over ground practice, followed by PWSTT daily practice at home. Progressively increased speed 0.18-0.8m/s and reduced assistance.	40 min / 90 min (home)	5 days/week for four weeks and 2x a week at home for 4 months.	N/A	Gillette Functional Walking Scale, Paediatric FIM transfers and mobility subscale, number of unassisted steps.	Gillette improved from some stepping with assistance to walking for household distances. Transfers improved from moderate assistance to modified independence. Walking improved from maximum assistance to supervision. No. of unassisted steps improved from 0-200-no assistance. Six months after initiation of the intervention, all steps on the treadmill were unassisted.

Author/yr	Study Design	Purpose	Participants	Intervention					Outcomes	
				Evaluation	Rehabilitation Session	Duration	Frequency	Comparison	Measures	Main Findings
Gibson-Horn(2008)	Case Study	This case report describes the effect of torso-weighting to counteract directional balance loss in a woman with relapsing/remitting multiple sclerosis.	MS. Clinical examination of a 40-year-old woman after multiple sclerosis exacerbation revealed loss of balance in the posterior direction during quiet standing as well as loss of dynamic balance in the posterior and lateral directions. The patient's standing posture was with her trunk posterior to her pelvis. She exhibited decreased strength in both extremities and trunk, diminished sensation in the right lower extremity and palms, and an unstable ataxic gait. Difficulty with walking and severe fatigue and dizziness were also reported.	6 times. Three conditions were chosen to determine the continued effects of BBTW: baseline (BL), non-weighted vest (NW), and her weighted vest (BBTW).	Based on balance examination results, the patient was fitted with a 0.5-lb vest containing 1.5-lb of additional weight placed anteriorly on the torso at the level of the umbilicus. Progressive balance, gait, and functional activities were repeated both with and without weighting the torso over six weeks.	30 min	2x per day, 6 weeks	N/A	Gait speed; TUG, Romberg test; Sharpened Romberg eyes open, eyes close, and single-leg stance.	The patient was able to accomplish more challenging activities with better balance while weighted. Functional improvement in walking and improved control during balance activities were demonstrated in later treatment sessions without weighting. Placing small amounts of weight asymmetrically on the torso, based on directional loss of balance and alignment, seemed to assist this patient in maintaining balance during static and dynamic activities.
Vaz et al.(2008)	Case study (ABA)	To investigate changes in gait quality, balance and mobility associated with treadmill training for ataxic individuals.	n=2, chronic ataxia following TBI, able to walk at least 10m independently; (1) 25yrs, ICARS (ataxia severity) 22/100, gait speed 0.51m/s; (2) 53 yrs, ICARS 60/100, gait speed 0.33m/s.	Baseline phases (A) lasted three weeks and consisted of clinical assessments only. Treadmill training phase (B) lasted four weeks. During all phases, assessments were performed three times a week always by the same examiner.	Treadmill training. Progressive increases in velocity and step length and reduced upper limb support.	20 min	3x per week, 10 weeks.	N/A	Gait parameters (speed, cadence, step length), TUG, customized balance assessment, Rivermead Visual Gait Assessment.	Comparing the first and last assessments, subject 2 had improved his performance 133% for step length, 75% for Timed Up and Go and 69% for balance, and subject 1 improved her performance 62% for step length, 27% Timed Up and Go, and 31% for balance.
Dias et al 2009	RCT	To evaluate the effect of ataxia sufferers using weights on the lower members while walking.	21 individuals with progressive or acquired Ataxia. BBS= 41.4/43.7.	Before (first evaluation), after treatment (second evaluation), and after 30 days (third evaluation).	GP were submitted to standard physiotherapy with 500g weight on the shin of each leg. SP the same thing without the weight.	30 min	20 sessions	Patients were randomly divided into two groups: with weights (GP n=10) and without weights (SP n=11).	BBS, Dynamic Gait Index, Equiscale, International Cooperative Ataxia Rating Scale, Functional Independence Measure.	The addition of weight in the distal region of the legs has improved static balance, and anticipatory reactive coordination gait ataxia in patients over time, as well as tremor and functional independence. BBS = 43.5 - 50.1.

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Ilg et al.(2009)	Clinical Trial	To investigare the benefit of physiotherapeutic training for patients with cerebellar degeneration.	16 patients with progressive ataxia due to cerebellar degeneration (n=10) or degeneration of afferent pathways (n=6).	Patients were examined 4 times: 8 weeks before intervention (E1), immediately before the first coordinative training (E2), immediately after the last training (E3), and after 8 weeks for follow-up assessment (E4).	Exercises included the following categories: 1) static balance, e.g., standing on 1 leg; 2) dynamic balance, e.g., sidesteps, climbing stairs; 3) whole-body movements to train trunk-limb coordination; 4) steps to prevent falling and falling strategies; 5) movements to treat or prevent contracture. After the 4-week intervention period, all patients received an individual written training schedule. They were asked to perform exercises by themselves at home for 1 hour each day. After assessment E4, they assessed by interview which patients regularly performed training at home.	1h	4-week course of intensive training with 3 sessions per week.	N/A	SARA, ICARS, BBS, goal attainment score (GAS). Velocity, step length, step width, and lateral body sway and temporal variability of intra-limb coordination.	Intensive coordination training improves gait in terms of velocity, lateral sway, and variability of intra-limb coordination pattern.
Widener et al.(2009)	RCT	To determine whether balance-based torso weighting (BBTW) has immediate effects on upright mobility in people with multiple sclerosis.	36 MS.	Baseline and again with weights placed according to group membership..	In phase 1, the control group had no weights placed. In phase 2, the alternate treatment group received standard weight placement of 1.5% body weight.	N/A	1 session	2-phase randomized clinical trial. In phase 1, 36 participants were randomly assigned to experimental and control groups. In phase 2, the control group was subsequently randomized into 2 groups with alternate weight-placement.	TUG, sharpened Romberg, 360-degree turns, 25-foot walk, and computerized platform posturography.	People with BBTW showed a significant improvement in the 25-foot walk (P = .01) over those with no weight, and the TUG (P = .01) over those with standard weight placement. BBTW participants received an average of 0.5 kg, less than 1.5% of any participant's body weight.
Freund et al 2010	Case Study (A-B-A)	The purpose of this study is to describe the effects of trunk stabilization training and locomotor training (LT) using body-weight support on a treadmill (BWST) and overground walking on balance and gait in an adult with severe ataxia secondary to brain injury.	The subject was a 23-year-old male who had a traumatic brain injury 13 months prior. BBS=5.	The pre-intervention baseline period was 6 weeks, the intervention period was 10 weeks, and the post-intervention period was 6 weeks.	This study had an A-B-A withdrawal single-system design with trunk stabilization exercises and locomotor training using body-weight support on a treadmill and overground ambulation during the intervention period and not during the pre/post intervention periods.	60-90 min	28 sessions, two or three times a week for 10 weeks; 18 ninety-minute sessions for the first 6 weeks and 10 sixty-minute sessions for the next 4 weeks.	N/A	BBS, timed unsupported stance, Functional Ambulation Category (FAC), 10-meter walk test (10-MWT), Outpatient Physical Therapy Improvement in Movement Assessment Log (OPTIMAL), transverse abdominis (TrA) thickness, and isometric trunk endurance tests.	Using BWST and overground walking, and trunk stabilization training may be effective in improving balance, gait, function, and trunk performance in individuals with severe ataxia. BBS=12 and decreased to 8, post-intervention.

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Ilg et al.(2010)	Clinical Trial	To evaluate long-term benefits and translation to real world function.	14 patients suffering from degenerative cerebellar disease. All patients were able to walk a distance of 10 m with or without walking aid.	Before training (BT), after 4-week training (AT), and at long-term assessment (LT) after 1 year.	Exercises included the following categories: (1) static balance e.g. standing on one leg; (2) dynamic balance e.g. sidesteps, climbing stairs; (3) whole-body movements to train trunk-limb coordination; (4) steps to prevent falling and falling strategies; (5) movements to treat or prevent contracture. After the 4-week intervention period, all patients received an individual training schedule and were asked to perform exercises at home.	1h	4-week intensive training with three sessions of per week. At home every day.	N/A	SARA, BBS, GAS, gait parameters, temporal variability of intra-limb coordination. Dynamic balance task: subjects stood in an upright position with both legs on a treadmill. Subjects were were instructed to compensate the perturbation by anteriorly directed steps.	Reduction of ataxia symptoms measured by the clinical scale SARA for the cerebellar group, which persisted after 1 year. Importantly, long-term outcome seems to be influenced by training intensity at home. Thus, continuous training of whole body coordination exercises seems crucial for stabilizing improvements in patients with ataxia.
Stephan et al.(2011)	Clinical Trial	To investigate whether long-term climbing training improves motor function in patients with cerebellar ataxia.	4 patients suffering from limb and gait ataxia (cerebellar). The patients were diverse in terms of pathology, duration of illness, and age, but in all of them the brain damage included the cerebellum. BBS=21.5,30.5,54,56.	6 times at intervals of 2 weeks: before and after baseline, three times during training, and after the training period (follow-up). The questionnaire was completed 6 times, at the end of each training week.	The various exercises challenged body balance; movement accuracy in pointing and grasping afterwards; movement smoothness; movement velocity; the planning of single movements and movement sequences; the integration of somatosensory information.	40-60 min	6 weeks	N/A	Unrestricted 3-dimensional arm and leg pointing movements, BBS and manual dexterity tests and a questionnaire on self-perception of symptoms.	BBS=33,37,54,56 . The long-term coordinative training improves motor performance and reduces ataxia symptoms in patients with cerebellar ataxia. Furthermore, the fact that improvements occurred in upper as well as in lower limbs goes in line with our suggestion that climbing training is a suitable method to train the whole motor system.
Ilg et al.(2012)	Intra-individual control design	Investigate the effects of whole-body controlled video game technology.	10 children with progressive spinocerebellar ataxia.	2 weeks before laboratory-based training, immediately prior to and after the laboratory-based training period, as well as after home training.	Training was based on 3 Microsoft Xbox Kinect video games particularly suitable to exercise whole-body coordination and dynamic balance.	N/A	Training was started with a laboratory-based 2-week training phase and followed by 6 weeks training in children's home environment.8-week.	N/A	SARA, dynamic gait index, Activity-specific Balance Confidence Scale, movement analysis.	Ataxia symptoms were significantly reduced and balance capacities improved after intervention. Quantitative movement analysis revealed improvements in gait and in goal-directed leg placement.

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Miyai et al.(2012)	RCT	To investigate short- and long-term effects of intensive rehabilitation on ataxia, gait, and activities of daily living (ADLs) in patients with degenerative cerebellar disease.	42 patients with pure cerebellar degeneration.	Short-term outcome was compared between the immediate and control groups. Long-term evaluation was done in both groups at 4, 12, and 24 weeks after the intervention.	The immediate group received inpatient physical (PT) and occupational therapy (OT), focusing on coordination, balance, and ADLs. The control group received the same intervention after a 4-week delay. After discharge, patients were instructed to maintain a similar level of activity in their daily lives as they did before. Patients receiving home-based rehabilitation at baseline—usually 20 to 40 minutes of physical therapy per week— were advised to continue the program after discharge at the same frequency as they did before admission.	2h (on week-ends 1h of PT or OT).	7 days per week, 4 weeks.	Imidiatte, Control and Inter-vention groups	SARA, Functional Independence Measure, gait speed, cadence, functional ambulation category, and number of falls.	Although the improvement of ataxia was prominent in the trunk, limb ataxia also showed a significant gain. This implies that the effect is not explained by promotion of physical fitness alone. Long-term follow-up revealed that the gains gradually attenuated. Although functional gains attenuated to the baseline level at 24 weeks after the intervention, studies regarding natural history of degenerative cerebellar diseases indicated that ataxic symptoms were progressive even in a half year. It is suggested that intensive and focused rehabilitation can at least partially overcome impaired motor learning in patients with degenerative cerebellar diseases.
Jorge-Rodriguez et al.(2013)	Clinical Trial	To confirm the results of a pilot study in a small group of patients with ataxia that suggested a positive impact of a restorative program on motor control.	20 patients with acquired or degenerative ataxia undergoing rehabilitation.	All patients were evaluated before and 28 and 42 days after treatment.	Neurorestorative treatment following the steps defined in the program, i.e., general physical condition, specific physical condition, prefunctional, training and functional training.	The daily schedule includes 5 hours of PT, one hour of OT, and one hour of speech therapy, compris-ing 7 hours of guided activity per day.	42 days	N/A	ICARS and measurement of maximal strength.	The beneficial effect of the program is reflected in a significant reduction of the global ICARS score. The intensive rehabilitation strategy induced significant changes after 28 days of treatment, but extending it for 15 additional days did not result in further significant improvement.
Abbud et al.(2014)	Case Study	To illustrate the use of a trained dog as a therapeutic tool to optimize physical and psychosocial adaptation of clients with ataxia.	2 cerebellar ataxia using different gait aids.	Before and after intervention.	The use of a trained intervention dog and an assistance dog, were compared.	N/A	Twice a week for 6 months	N/A	gait pattern and gait speed	Trained dogs may represent an innovative and positive alternative for mobility for people with ataxia, improving both physical and psychosocial parameters.

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Bultman et al.(2014)	RCT	Assess motor deficits in the acute phase after isolated cerebellar stroke focusing on postural impairment and gait ataxia and outlines the role of lesion site on motor outcome, the course of recovery and the effect of treadmill training.	11 patients with acute and isolated cerebellar infarction.	Acute phase and in a follow up after 2 weeks and 3 months.	Treadmill training during the first 2 weeks after enrollment in the study. This training took place additionally to a classical physiotherapy after stroke according to the Bobath concept. Velocity during the training was varied and progressively increased according to the patients' ability.	30 min	Daily, 2 weeks	N/A	ICARS, dynamic posturography, MR imaging.	After 3 months a mild ataxia in lower limbs and gait, especially in gait speed persisted. Because postural impairment had fully recovered, remaining gait ataxia was likely related to incoordination of lower limbs. Treadmill training did not show significant effects.
Kaut et al.(2014)	Single-center double-blind sham-controlled study(AAABBB)	Examined the effect of stochastic Whole body vibration (WBV) on ataxia in spinocerebellar ataxia.	32 ataxia in spinocerebellar ataxia types 1, 2, 3, and 6 (SCA 1, 2, 3 and 6)	At baseline and after the last treatment	Stochastic WBV was applied, each treatment consisting of five stimulus trains at a frequency of 6.5Hz. Patients allocated to the sham group received the same treatment with 1 Hz.	60s	4 days	N/A	SARA, SCAFI, and INAS	Significant improvements of gait, posture, and speed of speech in the verum group while limb kinetics and ataxia of speech did not respond. Stochastic WBV might act on proprioceptive mechanisms and could also stimulate non-cerebellar/compensatory mechanisms.
Keller et al.(2014)	Clinical Trial	To determine if a home balance exercise program is feasible for improving locomotor and balance abilities in these individuals	14 patients with cerebellar ataxia	5 testing sessions (2 pre-training, 1 mid-training, 1 post-training, and 1 one-month follow-up visit).	Home-based balance exercise Program.	20 min	4 to 6 days a week. 6 weeks	N/A	ICARS, 1) Dynamic Gait Index (DGI) (2) Timed Up and Go (TUG)(3) Functional Reach Test (FR) and (4) Activities Specific Balance Confidence Scale (ABC), postural sway, walking speed, stride length, double limb support.	Walking speed improved across visits, as did stride length, percentage double-limb support time, TUG, and Dynamic Gait Index. Significant rehabilitative improvements occurred over the 6-week training period, and all but TUG gains were retained 1 month later. Improvements in walking speed were affected by the level of balance challenge but not by age, ataxia severity, proprioception, or duration of exercise. Improvement in locomotor performance was observable after a 6-week home balance exercise program.
Kim et al.(2014)	Randomized, double-blind, sham-controlled pilot study.	To investigate the safety, feasibility and preliminary efficacy of low-frequency repetitive transcranial magnetic stimulation (rTMS) over the cerebellum in ataxic patients with acute posterior circulation stroke.	32 ataxic patients with posterior circulation stroke.	Before rTMS, immediately and 1 month after the last rTMS session.	1 Hz cerebellar rTMS	15min	5 sessions, 5 consecutive days	randomized to real (n = 22) and sham (n = 10) rTMS groups.	10-m walk test (10MWT) and BBS.	BBS of real rTMS group improved significantly (24 to 38). This study demonstrated that 1 Hz rTMS over the cerebellum is safe, feasible and may have a beneficial effect in ataxic patients with posterior circulation stroke.