Modelling of bipedal locomotion for the development of a compliant pelvic interface between human and a balance assistant robot

Tiseo, Carlo

2018

Tiseo, C. (2018). Modelling of bipedal locomotion for the development of a compliant pelvic interface between human and a balance assistant robot. Doctoral thesis, Nanyang Technological University, Singapore.

<http://hdl.handle.net/10356/73249>

<https://doi.org/10.32657/10356/73249>

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*A thesis submitted to the Nanyang Technological
University in partial fulfilment of the requirement for
the degree of Doctor of Philosophy*

2018

Acknowledgement

I owe a debt of gratitude to a myriad of people who made it possible for me to reach this achievement.

Firstly, I would like to thank my supervisor Professor Ang Wei Tech, who has always supported me in the pursuit of my investigation. Even when it seemed to be a dive into the dark to the most, he has always encouraged me to look forward to the light at the end of the tunnel.

I would also like to thank both Professor Lorenzo Masia and Prof Kalyana C. Veluvolu for the invaluable support provided to my research activities, Professor Pham Quang Cuong for his advice, and Professor Erdal Kayacan for giving me access to his motion capture equipment.

My gratitude also goes to my lab mates and the RRC staff who supported me and assisted in identifying a solution to technical and administrative problems I have encountered along the way. Among them are Agnes, Cheng Yap, Zenan, Mr Lim, Mr You, Yan, Zhen YI, Koon, Lukpla, Ming Jeat, Jie Kay, William, Khanh, Gia Hoang, Efe and Aishah. The merit of this achievement also has to be shared with all the amazing people I have met along this journey, and they are a second family to me —Paolo, Paola, Cristobal, Margherita, Astrini, Chris, Masoud, Leonardo, Sara, Fabio, Asif, Michele, Hiroko, Qiuyin, Sonia, Helena, Simone, JP, JC, Jesl, Rash, Jean-Pierre, Aamani, Marco, Ajay, Novia, Meng Zhe, Yuan Hsin, Ray, Amir,

Claudia, Julien, Anil, Mohammad, Enea, Francesca and Julien. Foremost, my girlfriend, Chadette, with whom I have had the honour of sharing the latest part of my life.

I am also thankful for all the amazing people who shaped me into who I am today. Among them are my teachers, and more importantly friends, Carla, Massimiliano, Donatella, Roberto, Andrea, Dino and Dario. No less important are the friends I had the honour to meet. Some of them are Marco Cap., Marco Cal., Ivan, Francesco S., Dea, Francesco Da, Francesco G., Andrea, Yuri, Giulia G., Maria Vittoria, Elena, Annarita, Mara, Silvia, Gabriella, Luciana and Eleonora. Lastly, the greatest merit goes to my family which includes Luca and Daniele —two childhood friends who are like brothers to me. I could not have the opportunities and the strength to push forward and face all adversities without my parents who built the environment that I call home, which is always open to everyone. They exposed me to the world at a young age allowing me to develop an independent open mind and leading an exemplary life made of love, empathy and sharing. I express my apology to my only brother for not being there physically to see him growing in the amazing adult he is becoming. I also express my gratitude to my ‘six siblings’—my elder cousins —with who I grew up with. Despite their highs and lows in life, they gifted me with unconditional love and eight amazing nephews. No less important are my grandparents, aunts and uncles who made it possible to build such fantastic environment around us.

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Abstract

With the rapidly ageing of the world's population, the WHO predicted that, by 2050, there will be about one billion people who are 65 years or older suffering from mobility-related problems. Robotic rehabilitation has been proven to provide effective therapies for people with motor deficiencies. Although multiple solutions have been developed for both balance and gait rehabilitation, both exoskeletons and robotic walkers have been shown to not have any benefit when compared with current therapies. A possible justification is a significant alteration of the activity dynamics (i.e. balance and locomotion) that prevents the skill translatability in daily living. The current knowledge about bipedal dynamics was reviewed to identify a viable solution for the improvement of both locomotion and balance therapy. The review revealed a gap in the current knowledge in the human balance and locomotion motor control, which does not provide a satisfactory explanation of how humans control and optimize their locomotion.

The following hypotheses have been developed to address the limitations of bipedal models: first, the brain accounts for fixed fulcrum for the legs' inverted pendulum model, and second, the legs are not simply synchronised but are deployed as coupled oscillators. These hypotheses have enabled the formulation of an algebraic model for human locomotion that has been tested against data from multiple experiments. The results show that the proposed

model can produce outputs that are fully contained within the variance of human data in both the centre of mass transverse plane trajectories and the antero-posterior swing trajectories of the feet. On the contrary, the estimation of the centre of mass of the vertical trajectories has not given definitive results, but it provided proof that most of the energy misestimation can be associated with the ankle strategies.

The deeper understanding of human locomotion strategies has led to identify in the high stiffness interfaces used in most of current devices a probable cause of the altered locomotion strategies. A 5 degrees of freedom selective compliance mechanism has been designed to improve the interface between the mobile robotic base and the patient. The proposed design introduces a buffer area to minimise the interaction forces and to increase manoeuvrability. Two prototypes of 5 degrees of freedom interface have been developed. Although they share the same mechanism design, they have different joint range of motion and inertias. The first prototype has a larger motion range and weighs about 12 kg, while the second prototype has a lower inertia and a smaller range of motion. The two prototypes have been mounted on two manual carriages to study the alteration of the gait strategies generated in healthy subjects. The first prototype has been tested with 5 female subjects. The data show an alteration of their speed, step length, step width and step frequency by -23%, -22%, -7% and -2.8%, respectively. The second prototype, on the other hand, has been tested with 12 healthy subjects (7 males and 5 females) on a highly manoeuvrable carriage that allows the regulation of the interface height. The data report a reduction in the walking speed and step length of 7.6% and 9.7%, respectively; they instead increase the step width and the step frequency of 9.2% and 2.55%, respectively. Moreover, the data also suggest that the alterations generated by the proposed interface may be

due to the selection of a more conscious movement strategies made from the subjects rather than their response to an external perturbation.

In conclusion, the knowledge gathered about both bipedal equilibrium and human motor-control of locomotion served as a guide to build a better interface for the pelvis that is able to provide harness support and which was the main scope of this work. Furthermore, this knowledge has beneficial outcomes not only in the development of rehabilitative technologies but also in improving bipedal robots' controllers.

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Chapter 1

Introduction

1.1 Background

The mobility impairment affects all the activities of daily living (ADL) significantly compromising the independence of people. It may also introduce psychological problems, due to the need of relying on others for the accomplishment of basic tasks such as personal hygiene. Furthermore, the mobility deficits are not only connected with pathological conditions but also with physiological ageing process [1]. The scenario becomes even more critical with the elderly population expected to reach 15% of the total world population by 2050, making of it one of the priorities for the World Health Organization [2, 3].

Robot-aided rehabilitation has been proven to be an affordable, effective and efficient solution that will lead to a sustainable health service for motor-impaired people. In the last two decades, different robotic technologies have been developed to provide therapies for both upper and lower limbs. Upper limb technologies have achieved outstanding results and are, currently, the treatments suggested by the American Heart Association. However, lower

limb technologies have not proven to be better than traditional therapies [4].

A possible explanation for this performance discrepancy may be found in the intrinsic diversity between upper limbs and balance movements [5]. Being the reaching movements used for the upper limb rehabilitation a task with kinematic predominance, while the maintenance of balance is by definition a diversified dynamical task composed of various strategies chosen under different situations [6]. The control of the balance is associated with 3 types of actions [6]:

- The maintenance of a posture;
- Voluntary changes of postures;
- Counteracting external perturbations.

The control mechanism is achieved through the synergy of different types of control distributed within our body and synchronised by the brain [5, 7]. However, some of these actions are implemented without a direct intervention of the brain to respond fast enough to stabilise the body against high-frequency perturbations [8, 9]. Indeed, this complexity renders it impossible to have an effective balance training through selective training of the different abilities, such as walking on the treadmill or keeping balance in a static position. In fact, it has been proven that it is more effective to train various abilities at the same time [10]. The approach to robot-aided balance rehabilitation has targeted either gait and standing balance therapies. In gait rehabilitation, patients are supported by robotic devices in the completing gait trajectory similar to the physiological motions. Some examples of these devices are the Lokomat (Hocoma AG, Switzerland), LOPES (Lower-extremity Powered ExoSkeleton) [11]. On the other hand, standing balance rehabilitation therapies are carried out through devices such as the Pro-BalancerTM

(Lab Rehab Pte Ltd, Singapore) [12] and Nintendo Balance-Board (Nintendo Co. Ltd., Japan) [13]. What these technologies have in common is the ability to train either the gait or postural control, but not both at the same time. Therefore, their biggest limitation is that they target specific sub-problems without addressing issues on locomotion in its entirety.

Contrarily, the KineAssistTM (Kinea Design LLC, USA), to the best of our knowledge, was the first device able to provide therapy targeting multiple components of balance at the same time and tries to bring lower limb rehabilitation into daily living tasks [11, 14, 15]. Recently, other devices have been following such approach; the Robotic Walker developed at the National University of Singapore [16] and the Andago commercialised by Hocoma AG in February 2016. The NUS Robotic Walker still shows the interference with the gait strategies already reported for the KineAssist [14]. To the best of our knowledge, Andago is the first device, to use compliant interface between the patient and robot; however, we have not found any data assessing its performances, which is probably related to the recent commercialisation of this product.

The limited performances of current rehabilitation therapies to locomotion and balance [17] have raised the following query:

- *Are the limitations of the current therapeutic approaches connected to an incomplete understanding bipedal balance and locomotion?*

A background research in human motor control revealed that although available models can provide some answers, there is still a significant gap in the understanding the locomotion motor-control [5, 18, 19]. Furthermore, an analysis of the current model of bipedal locomotion revealed that the absence of a model capable of accurately track bipedal dynamics [20, 21]. Therefore, a

better understanding of human locomotion and balance is required for the design of technology to be able to provide effective rehabilitation therapies.

1.2 Objective

The objective of the dissertation is to study the underlying mechanisms behind human balance and locomotion to identify a viable approach to the design of better robots for both motor rehabilitation and assistance. The investigation mainly focuses on the identification of the design principles to develop safe transparent interfaces with the human pelvis, which provide harness support without significantly affecting the dynamics of the movements. Moreover, it will also look into the identification of kinematics parameters for balance assessment to evaluate the patient equilibrium without the knowledge of the ground reaction forces, which are difficult to measure in unstructured environments.

1.3 Scope

The investigation presented in this dissertation addresses the modelling of bipedal dynamics and the design of a transparent pelvic interface with healthy subjects. Furthermore, the validations are limited to standing and straight walking because both tasks provide clear gait parameters for the evaluation of human strategies. The analysis of more complex movements, the development of robotic follow-me base, and the investigation of the interface performance with patients will be considered in the future. Accordingly, the scope of this thesis can be summarised as follows:

1. *Analysis of human locomotion and balance strategies during standing*

and straight walking in healthy subjects.

2. *Design and implementation of a human robot interface at the pelvis to test the concept of providing balance assistance.*
3. *Experiments with healthy human subjects to validate the proposed balance modelling framework and to evaluate the performance of the proposed mechanism.*

1.4 Organization

Chapter 1 defines the problem and objectives of this report. The Chapter 2 reviews state of the art for rehabilitation devices and practices, motor control and balance. Chapter 3 explains the proposed balance model and the experiments conducted for validation. Chapter 4 introduces the system as it is envisioned at first, explaining the link between the chosen design and the findings presented in Chapter 3. Secondly, it also describes the 5-DoF compliant pelvic interface prototypes and the preliminary experimental validations. The last Chapter concludes this research and presents the future development of the Mobile Robotic Assistive Balance Trainer. Appendix A describes the derivation of the model parameters from the motion capture data. Lastly, Appendix B contains the parameters used for the generation trajectories of the toe-off and heel-strike angles by the proposed planner.

Chapter 2

Literature Review

The ageing of the world population carries with it the increase in age-related pathologies among the population. The mobility impairments are the most common because they are related to both physiological ageing and pathologies (i.e. Parkinson's disease) [3].

For example, stroke is one of the most common age-related pathologies, with an incidence between 0.001 and 0.004, depending on the country. The average age of the first attack lies between the late fifties and seventies. Moreover, worth considering are the 30-Day Case-Fatality rate of about 25%, the hospitalisation rate of about 70% and the survivors' neurological deficits or impairment; leading to high cost of both economic and human capital for the entire world [22, 23].

The 2009 Global Health Risk reports the correlation of stroke with unhealthy lifestyles and indicates major risk factors such as smoking, alcohol, drugs abuse, sedentary lifestyles and bad eating habits [2]. The factors mentioned above show how the investment in healthcare policy and education is going to be the pillar of keeping healthcare and welfare affordable for our societies. The problem is so complex that governments shall actuate multi-

comprehensive policies investing on:

1. Prevention and increasing awareness in the population: implementing policies that stimulate the community to follow a healthier lifestyle and increase screening to reduce, as much as, possible the risk of age-related pathologies.
2. Improve the devices, procedures and techniques used in both screening and intervention.
3. Improve the rehabilitation techniques, follow-up protocols and assistive devices. This increases the life quality of the patient and minimises the risk of both the recurrent traumas and the insurgency of concurrent pathologies.

This chapter analyses state-of-the-art about human equilibrium, postural control, motor learning and available technologies for gait and balance rehabilitation.

2.1 Nervous System and Neuroplasticity

The Nervous System (NS) is composed of the Central Nervous System (CNS) and the Peripheral Nervous System (PNS). The CNS consists of the brain and the spinal cord while the PNS comprises the neurones and the ganglion cells that are outside the CNS. The neurones and the ganglion cells are grouped together to form the nerves [24, 25]. There are two types of cells in the nervous system, the neurones and the neuroglia cells. These cells are organised in networks, elaborating the information generated by our body, proprioceptive, and received from the external environment, exteroceptive, and generate the appropriate responsive behaviours [24–26].

The neural networks are not static structures but are dynamic. They can learn and adapt to new scenarios as well as recover the functions that have been damaged. This property is known as neuroplasticity. For example, it has been observed in normal people that learning to juggle produces a macroscopic change of the grey matter in the visual area responsible for object detection and movement prediction. It has less impact on the motor area, which is in charge of planning the movements [27–29].

The neurorehabilitation therapies need to be designed to stimulate the neuroplasticity of the affected areas in order to appropriately maximise the recovery of these areas. This will be possible only by taking into account all the aspects influencing neuroplasticity, like the neurogenesis and the reconfiguration of the brain synapsis [4, 10, 23, 25, 30, 31].

The patients' psychological state is one aspect of neuroplasticity. Unfortunately, this is not considered by engineers while designing rehabilitation devices. The medical literature widely reports how depression and stress have a negative effect on neuroplasticity in both the hippocampus (important for episodic learning) and the prefrontal cortex (used for high cognitive functions such as reasoning and long-term planning). Moreover, both stress and depression also induce a reduction of glial cells that regulate the glutamate, which is one of the neurotransmitters driving neuroplasticity. They also have a negative effect on the Ventral Striatum, the reward system of our brain. On the other hand, stress and depression enhance the neuroplasticity of the Amygdala, which is the responsible emotional learning process, principally driven by fear [25, 32, 33].

To conclude rehabilitation devices have to take into account all factors to provide adequate support for the therapies. For example, balance is a highly unstable task where the fear of falling drives our choices and learning

the process. In this case, an over-restrictive system that makes the patient feel safe and limits his movements preventing him from experiencing the instability, thereby reducing the effectiveness of the therapy.

2.2 Balance

The balance in humans can be defined as the process of keeping stability and postural control [6]. It is a complex dynamical process implemented through the actuation of different strategies depending on the situation. Its control is associated with 3 kinds of actions [6]:

- The maintenance of a posture;
- Voluntary changes of postures;
- External;

Balance is the ability of the humans to maintain their Center of Mass (CoM) within the Base of Support (BoS) that it is not much different compared to the balancing of a Segway [6, 20, 34, 35]. However, the task become extremely challenging due to the high latency introduced by the neuronal transmission, which has been overcome by nature through a semi-autonomous hierarchical architecture distributed in the body and synchronised by the brain [5, 7, 9, 36, 37]. This allows the body to take actions without the direct involvement of the brain, known as reflexes, characterised by low response time that stabilises our body from sudden perturbations that are faster than the brain response time [5, 6, 9, 37]. The performance of this distributed control architecture degrades with the ageing reducing the stability in the elderly, increasing the risk of fall and the occurrence of mobility-related pathologies [9].

2.2.1 Motor Control Strategies Involved in the Balance

The CNS intensely relies on the synergistic fusion of different sensorial feedback information that can be classified into three broad categories: proprioceptive, vestibular and exteroceptive information. Although a large number of studies have been conducted to study how human balance works, the exact mechanisms used by the CNS to merge and subsequently react to this information remain unclear [5,6,9,28,37,38]. The literature reveals that sensory feedback information is integrated at the brainstem; this knowledge is provided to the motor cortex which issues corrective movement commands, refining the change subject's intentions and the acquired knowledge. This is a slow feedback that is processed by the brain and requires at least 200ms to be completed [8,9,37–41]. It is supported by a faster feedback mechanism, which quickly reacts to the somatosensory updates thereby preventing falls and trips. This control is implemented in the spinal cord otherwise known as spinal reflexes. Meanwhile, the cerebellum decodes such signals providing the brain with an estimation of what is happening in the body [8,38,40,42–45].

The planning and control of the voluntary movements are implemented by the motor cortex, which controls the execution of pre-defined movement strategy driven by the subject's will and inputs from other parts of the brain [25,36,38]. The cerebellum supports the control of automatic movements acquired with practice such as in walking and bicycling [27,36,38,40,46]. Whereas, the cerebral cortex contains the information about the different tasks that enables the implementation of various control strategies for the same task and can be regarded as feed-forward control [38,47,48].

2.2.2 Balance Models

The need to better understand the balance and its control mechanism has led to the development of several different models. These models either use the inverted pendulum or the six determinates of the gait model [8, 9, 20, 34, 35, 49–54]. The inverted pendulum models are more suited to estimate the kinematic trajectory of the CoM. However, they often misinterpret the energetic cost of walking. In contrast, the six determinates of the gait allow a better estimation of the CoM energy, but do not produce an accurate kinematics of the motion [16, 20, 21, 31, 42, 54–63]. Thus, the available balance models use a hybrid inverted pendulum model that is supported by a dynamic model of the step-to-step transition, which requires knowledge of the ground reaction forces [16, 20, 21, 31, 42, 54, 56–63]. These derived models are computationally expensive. It is also difficult to identify neurophysiological parallels and justification under these derived models. They are likewise intrinsically incompatible with feedback delay introduced by the nervous system [5, 16, 18, 20, 21, 31, 42, 54, 56–62, 64].

Existing Models for Balance

The aforementioned limitation in the description of balance resulted in the development of models that target particular aspects of the balance necessary to the application in specific cases. This section is intended to provide the reader with the general knowledge, while a more in-depth analysis is presented in Chapter 3, where the model developed for this project is presented.

Regardless of the approach, all balance models share the concepts of CoM (Centre of Mass), Centre of Pressure (CoP) and the BoS (Base of Support). While the CoM concept is straightforward and precisely defined, this cannot be said for both BoS and CoP that present multiple definitions

within the same model [4, 7, 9, 35, 37, 49, 53, 65, 66, 66–74]. The CoP is defined as the point of application of the ground reaction. This implies that the constrained in the foot supporting the body during single support posture, but lies somewhere in between the two feet for double stance [9, 20, 35, 49]. The actual position can be determined from the ground reaction forces provided by the feet [9, 20, 35, 49]. Similarly, the BoS is the area where the body can counteract the gravitational field and thus maintain a stable vertical posture [9, 20, 35, 49]. Therefore, the BoS in single stance is limited to the region under the foot, while the BoS in double stance comprises both the area under the feet and the space separating the two [9, 20, 35, 49]. However, this area is posture-dependent; different researchers have provided a multitude of shapes depending on their application [6, 7, 32, 35, 49, 50, 74–76]. The principal models are:

- *The Inverted Pendulum and Compass Gait*: These formulations rely on the analogy between the legs and an inverted pendulum. Being a good and simple analogy, it is accurate to describe the system dynamic during single support [20, 21, 62, 77–84]. The basic formulations of these models assume a rigid system and a rigid isometric environment. Extended formulations are also available for generalizing the model to more complex scenarios [60, 77–79, 85, 86]:
 - *The Zero Moment Point (ZMP)*: The ZMP is a bioinspired model that allows generating trajectory which mimics human behaviour by controlling the angular momentum during the gait [77, 87–92]. The model provides the equations for deriving the foot placement that nullifies the angular momentum for the current trajectory [89, 93, 94]. While the direct problem is computationally simple, the association of the desired trajectory from the foot placement

is extremely elaborative [77, 87–92]. Due to these characteristics, it has been mainly employed for the generation of walking pattern in humanoid robots [21, 42, 85–92, 95].

- *The Extrapolated Centre of Mass (XCoM)*: The XCoM is a model that can estimate the foot position for walking, based on the limit cycle stability of the pendulum [35, 52, 53]. The model is computationally inexpensive and has been applied mainly in analysing human response to external perturbations [35, 52, 53, 56, 71, 74–76, 96].
- *The Six Determinants of the Gait*: This model has been accepted for a long time. It is based on the concept that human walk retains both the vertical and lateral movement of the CoM to minimise the energetic expenditure [20, 55]. The model provided is based on the following parameters: pelvic rotation, pelvic tilt, knee flexion, foot mechanics, knee mechanisms and lateral displacement of the pelvis [20, 55]. The model indeed provides analysis of the important features in the gait assessment in relation to the energetic cost of walking although originally meant as an approach to gait and balance [55]. When this method was employed for trajectory generation, the results are ineffective and led to inaccurate and inefficient trajectories [20].

2.3 Lower Limbs and Balance Rehabilitation Technologies

While balance impairments are very effective in reducing the patients' independence, a general physiotherapy that can help them regain the lost abilities is yet to be developed [6]. Moreover, to the best of our knowledge,

there is no study that proves the effectiveness of alternative approaches such as sensory stimulation, functional electrical stimulation, electromyographic feedback, and force feedback. Lastly, the only approach that seems to have benefits in balance rehabilitation is the static balance training, which is based on standing and sitting tasks. Contrarily, other therapy approaches based on gait training involve both static and dynamic postural control of the body [6], body weight supported treadmill training [11] and robot-assisted gait training [97]. They have shown a positive effect on general balance control in stroke patients albeit insignificant. The current state of the art in robotic devices for balance rehabilitation are introduced below, but none of these systems can independently assist the patients during the execution of different ADLs [98].

However, as reported by Miller *et al.* in [4] and Winstein *et al.* [17], there is no definitive proof that these technologies have major benefits when compared to traditional therapies. In contrast, the upper limbs rehabilitation robots are the recommended therapies for upper limbs neuromotor rehabilitation.

2.3.1 Treadmill Gait Trainers

The treadmill gait trainers are designed to aid the therapists in partial Body-Weight Support Treadmill Training (PBWSTT). Traditional therapy is executed by three therapists with the backing of a harness system. These robots replicate this therapy by substituting the therapist with an active exoskeletal orthosis with at least 2 degrees of freedom (DoF) per leg to support patients to execute a planar gait [11, 95, 98–101]. Although some of these devices are commercially available (i.e. the Lokomat, the LokoHelp and the ReoAmbulator), the clinical data do not show any significant benefit compared to

the traditional therapeutic approach [17]. Nonetheless, they reduce the work load for therapists by allowing them to service more patients [98].

Lastly, there are two devices under development that target the secondary gait deviations in post-stroke patients while walking on a treadmill. They are the String Man and the Robotic Gait Rehabilitation (RGR) Trainer. These devices apply force on the pelvis or at the trunk of the patients to support the patient to recover their motor abilities [11].

2.3.2 Foot-Plate-Based Gait Trainers

The foot plate based gait trainers are devices where the patient's feet are located in two different foot plates that guide their foot into the gait patterns regulated based on the individual abilities of the patient. A harness system guarantees patient safety. The Reha-Stim Gangtrainer GT I is the only device available in the market although many devices are under development [11]. The Haptic Walker, which allows simulating walk on rough surfaces, sliding and high-order dynamics movements, is a major GT I upgrade. This utilised in evaluating spinal cord injury and stroke patients [102]. Furthermore, the University of Tsukuba has been developing another system called The GaitMaster5 where the patient's feet are located on pads which can move forward/backwards and up/down simulating both walking and climbing. Lastly, The Lower-Limb Rehabilitation Robot (LLRR) assists patients to exercise footsteps and leg muscles. This system has structures that allow the control of posture and support the user's weight. The system has a total of 6 DoF allowing training patients to try different walking conditions [11].

2.3.3 Over-ground Gait Trainers

The over-ground gait trainers differ from the other devices as they allow patients to experience real environments such as pavements with an uneven surface, change of slope and physical qualities while having full control of their movements without being driven according to predetermined movement patterns. The Re-Walk is an exoskeleton for lower limbs that support patients to execute gait while detecting movements in the upper part of the body. Meanwhile, HAL (Hybrid Assistive Limb), a wearable robot for the lower limbs was developed in five different versions. Its version varies from an all body exoskeleton to a single leg orthosis that allows HAL devices to be used in a broad range of applications, either in rehabilitation or substantial works support [11, 98, 103].

Another configuration of this category of devices are the ambulators which support the patients pelvis to avoid falling. The KineAssist (HDT Global inc.) is a system for gait and balance training supporting patients for different tasks like standing, walking, sitting to standing and other ADL related activities [14]. Experiments with the human subjects have shown that this technology is unable to be transparent to the users altering the movement strategies. The Walk Trainer, Swortec SA, is composed of a body weight support, a two-leg orthosis and a real-time controlled electro-stimulator. This allows the patients to safely ambulate with gait support, while simultaneously receiving the electrical stimulation of the muscles [14]. Recently, other research groups have followed the approach introduced with the KineAssist [16, 104]. The most relevant is the overground Robotic Walker for pelvic motion support developed by NUS. It has a lighter and simpler structure of the KineAssist, but has the same limitation as that of the KineAssist [16]. The NaTure-Gaits developed at Nanyang Technological University aims to

help the patient to execute a natural gait while providing a body weight support. It is a deambulator with an assistance mechanism for pelvic movements and a 6 DoF orthosis for the lower limbs [105]. The structure is very intrusive and heavy compromising the interface with the user. Lastly, Hocoma AG has recently commercialised Andago, a body weight support for ADL-based rehabilitation. This system utilises a compliant interface with the user. The unavailability of technical data on the device prevents an accurate analysis of the system; however, the compliant interface will probably make of it the best system on the market.

2.3.4 Ankle and Knee Rehabilitation Systems

The ankle and knee rehabilitation systems have been developed to restore the physiological motion of the ankle and the knee in post-stroke patients. The motion of the ankle is very complex because it is composed of four main motions, namely, dorsiflexion/plantarflexion, inversion/eversion, abduction/adduction, and pronation/supination. This category of robots can be subdivided into two categories —stationary and active foot orthoses.

Stationary Systems

The stationary systems are designed to help the patient exercise ankle and knee joints without walking. The first device of this kind that has been developed is the Rutgers Ankle which has been proven to help patients regain strength and endurance with its 6 DoF and virtual reality-based exercises [11]. Another device in this category is the High-Performance Ankle Rehabilitation Robot developed by the Istituto Italiano di Tecnologia (IIT) which trains plantar/dorsiflexion and inversion/eversion using a redundant platform, allowing a wider operational space and a higher dexterity of its

structure [106]. Recently, the Active Knee Rehabilitation Orthotic Devices (AKROD) has been developed to facilitate the recovery in neurological patients, providing a variable damping to the knee joints [11].

Active Foot Orthoses

In contrast to the stationary systems, active orthoses support patients walking on a treadmill or over-ground. They are an evolution of the passive lower limb orthoses. The first robots developed in this category are the Powered Gait Orthosis (PGO) and the Pneumatic Active Gait Orthosis (PAGO) [11]. Although clinical trials have been completed, they are yet to be commercialised. The INMOTIONANKLE of the Bionik Laboratories Corp. is the only system which has 3 DoF that can be used for both overground and on a treadmill. Clinical test in two pilot study on the device developed by the Massachusetts Institute of Technologies (MIT), showed good performance in improving the walking distance covered by patients and time [5]. Other devices of this kind are the ALOUD of the University of Delaware, the Knee-Ankle-Foot-Orthosis (KAFO), the Ankle- Foot-Orthosis (AFO), the Robotic Gait Trainer (RGT), the Yonsei-AAFO and the SUkorpion AR of the Sabanci University [11].

2.3.5 Balance Boards

Balance Boards are stationary devices which are either active or passive, helping the patients train to maintain the balance in the static standing task with and without external perturbations. The ProBalance™ commercialised by Rehab Pte Ltd is an active haptic device that trains the balance with virtual reality exercises, targeting patients affected by both neural damages and impairments of the lower limbs [12]. Another approach for the same

problem is to employ passive devices like the Nintendo® Balance Board system. The easy Balance Virtual Rehabilitation (eBaViR) system uses the Nintendo® Balance Board with Nintendo® Wii games to provide therapies to the patients [13].

Chapter 3

Human Balance & Motor Control

Human locomotion is a fundamental activity of human life. Even a partial impairment in locomotion has a great impact on the independence of the people and consequentially on their perceived disability. It has been previously explained that the equilibrium degradation is part of the physiological ageing of our body and it affects about half of the elderly population [1, 3, 107, 108]. Such centrality of balance and ambulation in human lives has led to a multitude of studies aiming to understand, describe, model and lastly, replicate human balance [5, 7, 9, 21, 35, 49, 55, 66, 76, 81, 82, 89, 109, 110, 110–113].

However, despite all efforts exerted, the mechanisms behind human balance control are largely unknown, and the devices developed for rehabilitation are unable to provide a reasonable improvement to the therapies [5, 9, 17, 21, 66]. By contrast, the positive results obtained in the last three decades for upper limbs rehabilitation leads to the question [17, 114]: what made this difference in the upper limbs?

There is no easy answer to such question, but upper limbs rehabilitation

robots are evolved from devices that were originally designed to study the motor control for reaching movements in the mid 1980s [115, 115–117]. This provides an understanding of human motion strategies for reaching movements used for the design of adequate robot-aided therapies [5]. On the contrary, the understanding of balance motor-control and human force interaction, in general, are still in their infancy [5, 18, 19, 64].

The limited understanding of human stability control is reflected by the fragmentation of currently available models for the locomotion [9, 20, 21, 35, 49, 56, 89, 109, 118]. In addition, the complexity of the phenomenon and even simple passive walkers manifests variegated dynamics when deployed on rough terrains [78, 79]. This questions the source of variety in such behaviours and the limiting factor for the current models to replicate the bipedal dynamics.

A possible approach to this problem was suggested by the ability that human beings to estimate the stability of others by the observation of their kinematics. Hof *et al* have already introduced an initial step towards this approach [35]. Their model has been mainly applied to study human stability assessment and their reaction to perturbations [35, 52, 53, 56, 71, 74–76, 88, 96]. Its main limitation is that the model relies on the inverted pendulum model and thus, on its hypothesis of isotropic behaviour and invariance of the pendulum parameters. These hypotheses introduce a simplification of both the human body and the environment that leads to misestimation of the energy associated with the trajectories. Therefore, the development of an alternative model is required to describe the double stance phase using the ground reaction forces, while keeping the inverted pendulum model for the single stance phase [20].

3.1 Models of Bipedal Locomotion

Bipedal locomotion is modelled using the inverted pendulum analogy due to both its computational inexpensiveness and the good representation of the single stance phase. Nonetheless, the inverted pendulum model is unable to track the system dynamics during double stance and requires the introduction of additional models to enable its extension. Figure 3.1 compares the inverted pendulum model and the human trajectories for bipedal walking strategies.

Over the years formulation of multiple models to extend the inverted pendulum dynamics to the double stance phase have been proposed. These models identify movement constraints that allow relating the dynamics of anthropomorphic bipeds to the inverted pendulum by enabling to represent a step forward as a shift of π [rad] in the inverted pendulum state. The presence of numerical models can be regarded as the main limitation in all the current models because it does not allow to identify a generalised approach to the step transition but only *ad hoc* solutions once either a CoM trajectory or a step placement have been identified. Bipedal models can be classified into two categories depending on the target application. Therefore, the model developed for bipedal robots and human locomotion will be introduced in separate sections at first and, subsequently, compared before proceeding to the formulation of our solution.

3.1.1 The ZMP Model and Anthropomorphic Bipedal Robots

The development of bipeds has increased within the last three decades steadily due to the high flexibility and efficiency of bipedal gait [20, 77]. Notwithstanding the steady improvements of the robots over this time span, their

Figure 3.1: The schematic on the left shows the inverted pendulum model analogy with both legs laying on the same sagittal plane, while the human walking strategy is portrayed on the right. Although the two strategies seem to be identical according to the lateral view, the top view shows a different story. The Human trajectory is three-dimensional while the inverted pendulum model generates a planar trajectory. Thus, the model cannot track the pendulum angle and, as a consequence, the representation of human walking dynamics is inaccurate.

performances are still far behind human beings [119–123]. For example, a robot requires a few seconds to plan the first step in a novel environment, which is characterised by flat surfaces and the absence of obstacles. The subsequent steps are characterised by a planning time of few hundreds of milliseconds [90].

The limitation of current planners derives from the Zero Moment Point (ZMP) model, which is currently the gold standard for step planning [90, 119, 121]. The ZMP model provides a method for the calculation of foot landing

if both the support foot and the desired CoM position at the moment of the landing are known [93,121]. Where the ZMP is defined as the foot placement that allows the landing leg to stabilise the CoM through the application of a force along the equivalent inverted pendulum rod [93]. Therefore, it implies that the force acting in the tangent planes of the two pendula are equal and opposite.

The discrepancy of performances in the planning time mentioned above can be related to the ZMP model, which requires the knowledge of the desired CoM trajectory during the transition for computing the associated foot placement [90,121]. This implies that the first time the robot controller plans in the new environment has to run an optimisation algorithm to identify the optimal CoM trajectory. However, once the trajectory has been planned, it can be re-utilised as long as the environmental condition does not change [90]. Another difference between the human and robots walking is that the CoM trajectory of the latter is constrained on a single transverse plane, while human CoM has a vertical component [119].

3.1.2 XCoM Model and Human Gait Analysis

The eXtrapolated-CoM (XCoM) model has been developed to extend the inverted pendulum model to analyse the human walking locomotion strategies about 10 years ago [35, 52, 53]. Over the last decade, it became one of the most used models in the analysis of human locomotion and its stability, proving that humans generally adopt locomotor strategies characterised by the asymptotic dynamics of the inverted pendulum model [35, 52, 53, 56, 71, 74–76, 88, 96, 124, 125].

The formulation of the XCoM model states that if humans deploy the two legs as inverted pendula, then both systems have to be in a stable state

during the transition. This implies that the distance between the two fulcra is determined by the hyperbolic behaviours of the inverted pendulum model [35, 53]. In other words, the model imposes a synchronisation between the two pendula by imposing the desired landing state to the swinging leg [53]. Therefore, the model provides a computationally inexpensive method for the instantaneous evaluation of the system stability, which leads to its widespread application in human studies [56, 71, 74–76, 88, 96, 124]. However, the XCoM also requires a recursive approach for the generation of trajectories as the ZMP model [53].

3.2 Extension of the Inverted Pendulum

The analysis of both the ZMP and the XCoM models revealed that both can identify viable solutions for the step-to-step transition by the imposition of constraints. In so far this work is concerned, this aims to identify a model that characterises the anthropomorphic dynamics and to prove that it can be used for generating human-like behaviour.

The identification of a more convenient representation of bipeds started from the analysis of the inverted pendulum phase diagram as shown in Figure 3.2. The extension of the inverted pendulum and the removal of the hypothesis of collinearity in the feet along the locomotion direction are possible by considering both legs working as coupled oscillators at the same time. However, the analysis of the system dynamics based on the saddle phase diagram will be possible only after the identification of a model for the potential energy of the pendulum as a function of the state coordinate x , which has a posture-dependent relationship.

Figure 3.2: The phase diagram on the left is the classical approach to the stability analysis of the inverted pendulum. This model is centred on the support foot. Consequently, it does not allow tracking both feet simultaneously but requires the synchronization of the legs [53]. On the right side, the proposed approach accounts for both legs at the same time and to describe human walking as a closed-loop around the saddle fixed point. Current models account for the legs as oscillators working at the natural frequency (ω_n) which is the limit of stability [53]. In contrast, human walking is performed at a lower frequency equal to half of the step cadence [53, 63]. This leads to the hypothesis that the walking frequency ω_{walking} provides a better representation of the walking strategy.

3.2.1 The Definition of the Saddle Space for Modelling the System Potential Energy

Figure 3.3 shows the frontal and top views of the ‘dome-shaped’ energy (paraboloid surfaces) generated by two inverted pendula with their fulcrum

separated from a distance D_f . The intersection of the two paraboloids forms a saddle surface characterised by a removable singularity along the line where the two paraboloids intersect, which is one of the two principal direction of the surface. The other principal direction is aligned with line connecting the fulcrum of the two pendula, which is by the definition orthogonal to the first principal direction. Therefore, they can be used for the definition of a Saddle reference system. The ordinate (y_{Saddle}) has been chosen to be aligned with the principal direction between the two fulcra. Meanwhile, x_{Saddle} is collinear with the other principal direction which is defined by the intersection of the two paraboloids.

The application of the model to the human body requires the identification static reference points within the human body for the two fulcra and the Centre of Mass (CoM), as shown in Figure 3.4. The CoM position has been chosen in accordance with anthropometric data to be on the longitudinal axis of the human body at 57% of the body height in the geometrical center of the pelvis [9, 119, 126]. The two fulcra have been defined based on the geometry of the foot sole without considering the toes, which coincides with the Base of Support (BoS) produced by the foot according to the results from *Hof et al* [35]. Although this latest choice differs from the normal definition for the CoP, the hypothesis is consistent with the recently published model from *Carpentier et al* [119]. The analytical derivation of both the CoM and CoP from the anatomical landmarks of the pelvis and foot [127] are described in Appendix A.

The Saddle Space Transformation

The modelling of the system potential energy in the saddle space requires the formulation of the geometrical transformation between the saddle space

Figure 3.3: Shown above are the frontal and top views of the potential energy surface determined by the coupling of two inverted pendula. The resultant Saddle Surface can be characterised by their lengths (h_{RF} and h_{LF}) and the distance between the two fulcrum (D_f).

and the task-space.

The first step of the process is the placement of the saddle point. Its position depends on h_{RF} , h_{LF} and the distance D_f , and can be calculated using the following system of equations:

Figure 3.4: The CoM reference frame is oriented according to the human anatomical planes. Its centre is placed at height h_{CoM} which is posture-dependent. $\mathbf{x}_{CoPl}$ and $\mathbf{x}_{CoPr}$ are the Centres of Pressure, which are considered fixed in the foot within this work, while h_{RF} and h_{LF} are the length of the two pendula, and, having a maximum value of $0.57h_{body}$ in accordance with human anthropometry [126].

$$\begin{cases} h_{RF}^2 - h_{LF}^2 = D_{SR}^2 - D_{SL}^2 \\ D_f = D_{SR} + D_{SL} \end{cases} \quad (3.1)$$

where D_{SR} and D_{SL} are the variables representing the distances of the saddle point from the right and left CoP projections on the ground, respectively, which can be calculated as follows:

$$D_{SR} = \frac{D_f^2 + h_{RF}^2 - h_{LF}^2}{2D_f}$$

$$D_{SL} = \frac{D_f^2 + h_{LF}^2 - h_{RF}^2}{2D_f}$$
(3.2)

Equation 3.2 describes the distances of the CoPs from the origin of the Saddle Frame (Figure 3.5), which allows writing the equations of the Saddle Space axis in the Task-Space, as follows:

$$\begin{cases} y = \frac{x_{CoPl} - x_{CoPr}}{y_{CoPr} - y_{CoPl}} x + \frac{h_{LF}^2 - h_{RF}^2 + x_{CoPr}^2 + y_{CoPr}^2 - x_{CoPl}^2 - y_{CoPl}^2}{y_{CoPl} - y_{CoPr}} = \\ \quad = m_{S\perp} x + C_{S\perp} \\ y = \frac{y_{CoPl} - y_{CoPr}}{x_{CoPl} - x_{CoPr}} x + \frac{x_{CoPl} y_{CoPr} - x_{CoPr} y_{CoPl}}{x_{CoPl} - x_{CoPr}} = m_{S//} x + C_{S//} \end{cases}$$
(3.3)

where $m_{//}$ is the slope of the line passing between two CoPs ground projection, and $m_{\perp}$ is the perpendicular slope. Therefore, the rotation angle between the two frames is $\lambda = \arctan(m_{S\perp})$ and the saddle coordinates $\vec{x}_{saddle_0} = [x_{saddle_0}, y_{saddle_0}]$ are:

$$x_{Saddle0} = \frac{m_{S\perp}(C_{S//} - C_{S\perp})}{m_{S\perp}^2 + 1}$$

$$y_{Saddle0} = \frac{(C_{S//} - C_{S\perp})}{m_{S\perp}^2 + 1} + C_{S//}$$
(3.4)

Equation 3.4 together with λ provides the information required from the following equation to represent the transformation between the two spaces:

$$\begin{bmatrix} x_{TS} \\ y_{TS} \end{bmatrix} = R(\lambda) \vec{x}_{saddle} + \vec{x}_{saddle_0} =$$

$$= \begin{bmatrix} \cos(\lambda) & -\sin(\lambda) \\ \sin(\lambda) & \cos(\lambda) \end{bmatrix} \begin{bmatrix} x_{saddle} \\ y_{saddle} \end{bmatrix} + \begin{bmatrix} x_{saddle_0} \\ y_{saddle_0} \end{bmatrix}$$
(3.5)

3.2.2 Potential Field and Margin of Stability

Equation 3.5 correlates the TS coordinates with the Saddle Space. Thus, the system potential energy in the TS can be defined as follows:

$$U(x_{TS}, y_{TS}) = \begin{cases} M_{CoM}gh_{CoML}, & y_{saddle} \geq 0 \\ M_{CoM}gh_{CoMR}, & y_{saddle} \leq 0 \end{cases}$$

Being :

(3.6)

$$h_{CoML} = \sqrt{h_{LF}^2 - (x - x_{CoPl})^2 - (y - y_{CoPl})^2}$$

$$h_{CoMR} = \sqrt{h_{RF}^2 - (x - x_{CoPr})^2 - (y - y_{CoPr})^2}$$

Figure 3.5: (a) The saddle and the CoM spaces are aligned when both feet lay on the frontal plane. (b) A posture different from the configuration presented in (a) generates a transformation between the two spaces with the subsequent deformation of both Margins of Stability (MoS) and the Base of Support (BoS).

Furthermore, Equation 3.6 is differentiable, enabling the calculation of the gravitational force field with the gradient operator.

$$\vec{F} = -\nabla U(x_{TS}, y_{TS}) \quad (3.7)$$

Base of Support (BoS) and Margins of Stability (MoS)

The Base of Support (BoS) is defined as the region of the space where the body is able to maintain stability by counteracting the gravitational forces acting on the CoM, as shown in Figure 3.5. Its borders are the Margins of Stability (MoS) as shown in the saddle space in Figure 3.6 [6, 7, 27, 35, 49, 50, 56, 66, 74–76, 128]. Regardless of the consistency of their general definition of BoS and MoS, there is no unique model for them. Their shape changes are different among various studies, but it is task-dependent in most cases. The most common definition considers the BoS to be the area under the foot in single support, while it is enlarged to include both feet and the area between them during double support [35, 49, 56, 74–76].

In accordance with the hypothesis of legs synchronism on which this model is based, the BoS and MoS are designed to always account the position of both feet in the evaluation of the system stability. Figure 3.5.(a) shows that the saddle and the CoM reference frames are aligned when both CoPs lay on the body sagittal plane. Thus, the measure of the BoS conducted in such posture allows identifying the BoS in the saddle space. Once such shape has been identified, the BoS for a generic posture can be obtained using Equation 3.5.

The analysis of previous studies of the BoS dimension in the aforementioned posture leads to the following Equation [35, 56]:

$$\begin{aligned}
MoSs = & \left\{ \begin{array}{ll} y_{saddle}^2 + x_{saddle}^2 = CoP_{L_{saddle}}^2, & \text{if } y_{saddle} \geq 0 \\ & \& x_{saddle} \leq +d_{CoP} \\ y_{saddle}^2 + x_{saddle}^2 = CoP_{R_{saddle}}^2, & \text{if } y_{saddle} \leq 0 \\ & \& x_{saddle} \geq -d_{CoP} \\ otherwise & \end{array} \right. \quad (3.8) \\
BoS \leq & MoSs
\end{aligned}$$

where $d_{CoP} = 10$ cm describes the Anterior-Posterior (AP) dimension of the BoS reported by Hof *et al.* [35].

3.2.3 Foot Swing

The AP-trajectory of the swinging foot is defined based on the hypothesis presented earlier in this chapter that the CoM lays on y_{Saddle} during the walking, as shown in Figure 3.7.

$$x(t)_{Swing} = \left\{ \begin{array}{ll} \frac{d_{SW}}{m_{S//}(t)} + x_{CoPr}, & \text{Right Stance} \\ \frac{d_{SW}}{m_{S//}(t)} + x_{CoPl}, & \text{Left Stance} \end{array} \right. \quad (3.9)$$

where d_{SW} is the step width, x_{CoPl} and x_{CoPr} are the abscissas of the foot providing the support.

Figure 3.6: *The two-coupled pendula structure generate a potential energy with the shape of a saddle surface. This is useful in defining a reference frame centred on the saddle point with the two axes aligned with the principal direction of the saddle. Thus, a rigid transformation between the potential energy space and the Cartesian space can be defined by measuring the foot position in the space, which enables the mapping in the Cartesian space of the Margins of Stability (MoS) defined in the saddle space.*

Figure 3.7: *The hypothesis that human CoM trajectory is always laying on the segment connecting the two CoPs enables the analogy between the phase diagram (Figure 3.2) and human walking. In fact, the aforementioned bound makes y_{Saddle} a valid state variable of the system.*

3.2.4 Heel-Strike Model

The ankle behaviour has been proved to be crucial in the walking energetics, which has driven scientists to focus on the development of devices dedicated to the ankle for human enhancement, assistance and rehabilitation [5, 21, 60, 112, 113, 129, 130]. Currently, the most successful application of these devices is in human enhancement which can significantly reduce the metabolic cost of walking [21, 112, 113]. This suggest a relation between that the energetic misestimation of the inverted pendulum model and its inability to describe the ankle where majority of the energy is exchanged.

The proposed model is formulated based on the hypothesis that human beings account for a fixed CoP, while the other models use the point of interaction with the ground that moves along the foot length during the stance

Figure 3.8: Usually, the HS-TO phase is modelled as a trajectory of the CoP on the ground. This method relies on the measure of the contact point via force sensors. In contrast, the proposed model tracks the extension to the pendulum's length introduced by the ankle posture, which allows modelling the HS effect on the system dynamics by measuring the foot kinematic. The height variation has been modelled as the length of the chord of a circumference with radius (d_h), which is associated with an angle $\theta_{zFoot}(t)$.

phase. Although this hypothesis allows monitoring the rotation fulcrum without tracking the ground reaction forces, it also requires a model of the foot kinetics to account the motion of the foot during the stance phase. The chosen model is schematised in Figure 3.8, where the foot motion is represented as a change in length of the pendulum. This variation is equal to the cord of a circumference with a radius (d_h) equal to the distance between the heel and the CoP, resulting in the following equation:

$$\Delta h_{iF} = 2d_h \sin(\theta_{zFoot}(t)/2) \quad (3.10)$$

3.2.5 Formulation of the CoM Trajectory

The CoM trajectory while walking straight at constant speed is a linear function of the time in the AP direction, while it has a cosinusoid function in the Medio-Lateral direction (ML) [9, 20, 89, 94]. Moreover, humans are reported to minimise their angular momentum during the transition from one foot to another, which is the foundation for the ZMP-models [21, 42, 77, 85–89, 89–94, 118, 121, 131]. In other words, the AP trajectory is tangent to y_{saddle} during the transition from one foot to the other. In fact, moving along y_{saddle} guarantees that the angular momentum is consequentially nullified because the gravitational forces acting on the system are perfectly aligned with the motion direction. Such condition determines that the amplitude of the AP-cosinusiod is as follows:

$$A_{yI} = \frac{m_S //_{step}}{2\omega} = \frac{d_{SW}}{2\pi\omega_0 d_{SL}} \quad (3.11)$$

$$\omega_0 = \frac{v_{des}}{d_{SL}}$$

where ω_0 is the step frequency in Herz, v_{des} is the walking speed and d_{SL} is the step length. Equation 3.11 allows defining the Cartesian trajectory of the CoM as in the following equation:

$$\left\{ \begin{array}{l} x_{des}(t) = v_{des}t \\ y_{des}(t) = A_{yI} \cos(\pi\omega_0 t + \phi) \\ z_{des}(t) = \begin{cases} h_{CoML}(t), & \text{if } Left \text{ Support} \\ h_{CoMR}(t), & \text{if } Right \text{ Support} \end{cases} \end{array} \right. \quad (3.12)$$

where h_{CoML} and h_{CoMR} are the CoM height calculated with respect to the supporting foot, in accordance with the following equation:

$$\begin{cases} h_{CoML}(t) = \sqrt{h_{LF}(t)^2 - (x_{des}(t) - x_{CoPl})^2 - (y_{des}(t) - y_{CoPl})^2} \\ h_{CoMR}(t) = \sqrt{h_{RF}(t)^2 - (x_{des}(t) - x_{CoPr})^2 - (y_{des}(t) - y_{CoPr})^2} \end{cases} \quad (3.13)$$

3.2.6 Validation

The validation of the model has been conducted testing our model with both data reported in literature and motion capture data from the KIT whole-body motion database (KITDB, [127]). The KITDB is a scientific repository of human movement, which contains straight walking trajectories collected from 6 subjects that are used to validate the proposed model.

The method used for validation is the following:

1. Equations 3.9, 3.11 and 3.12 are tested with the data reported by Orendurff *et al.* [63].
2. The ability of the model in reproducing trajectory similar to the motion capture data is tested. In order to conduct such test, the subject height, step length, the step width, walking velocity and step frequency are extracted by the motion capture data from the KITDB [127]. The heel strike angle trajectories ($\theta_{zFoot}(t)$ in Equation 3.13) are approximated to a linear trajectory with a maximum HS angle of 9 degrees (Figure 3.9) with a duration of 240 ms. The results have been evaluated by comparing the model-generated CoM trajectories with the respective motion capture trajectories. The foot swing trajectories comparison has been performed by evaluating the difference between the slope from Equation 3.9 with the angular coefficient of the swinging foot

AP-trajectories, calculated as follows:

$$m_{swinging} = \frac{x_{foot}(t_{MidSwing} + .04) - x_{foot}(t_{MidSwing} - .04)}{0.8} \quad (3.14)$$

where $t_{MidSwing}$ is the time where the trajectory reaches the half way point.

3. An experiment with 12 healthy subjects is conducted to test if the transformation in Equation 3.5 can reproduce the shapes of the BoS in different postures starting from the BoS geometry defined in the saddle space in Equation 3.8.

Figure 3.9: *The Data from KITDB reveal a high variability in the ankle strategy during Heel Strike (HS). There is a maximum angle of 9 ± 7 degrees, with the highest recorded value of almost 40 degrees.*

Results for the Test with Orendurff' s Data [63]

Orendurff' s *et al* analysed the walking strategies in 10 healthy subject at 4 speed [63]. Their work reports the amplitudes of the CoM trajectories in both the vertical and lateral direction, the step length, step width and step frequencies. Therefore, it allows to testing the validity of Equations 3.11, 3.12 and 3.13.

The test results show that the model is able to estimate stepping frequency and the ML-CoM oscillation amplitude (A_y) quite accurately, while there is a larger overestimation for the CoM vertical movement (Δz_{CoM}). To further elaborate, the stepping cadence is estimated with an error of 2.38 ± 2.00 6 steps per minute, which is consistent with the reported variability (± 3.40 [steps/minutes]). Similarly, A_y is overestimated with an error of 1 ± 1 cm that is also consistent with the measure variability [63]. However, this is not the case for the error in estimating Δz_{CoM} , where there is an error of 3.46 ± 1.35 cm. Nonetheless, this error have been expected since it is the limitation of the inverted pendulum models because it does not account for the change in the pendulum length.

Results for the Test with KITDB Motion Capture Data

The KITDB provides the motion capture data in straight walking for 6 subjects (4 males and 2 females). The age, height, and weight of the subjects were 25 ± 1.7 , 1.79 ± 0.096 m and 63.3 ± 10.3 kg respectively.

Figure 3.10 shows that the results for A_y confirms the error of 0.70 ± 0.49 cm. Furthermore, the data for Δz_{CoM} also show a reduction in the estimation error, which is 1.43 ± 1.15 cm with heel strike and 2.18 ± 1.15 cm without heel strike (Figure 3.11).

The data report that the heel strike reduces the mean error by the 35%,

Figure 3.10: The shaded areas are the 90% prediction bounds of the fitting. (a) The data obtained for A_y report an estimation error within the human variability. Thus, it can be concluded that the model is accurate in estimating the CoM ML-trajectory. It is also able to identify the divergence of 2 subjects which walk with a smaller step width from the normal behaviour. (b) The analysis of the CoM mediolateral trajectories shows how the proposed model output is fully included within human variability.

Figure 3.11: *The shaded area is the 90% prediction bounds of the fitting. Introducing of the proposed model for the HS strategy shows a significant improvement in the amplitude of the vertical CoM trajectories. However, it also indicates that the current model is not able to completely account for the overestimation of human trajectories, especially at high speed.*

but there is still a significant discrepancy between the model and the human behaviour along the vertical direction. This can be explained as a combination of multiple factors, such as high variability of the heel strike strategy (Figure 3.9) and the necessity of accounting for the toe off phase. This latest factor is supported by Figure 3.12, which shows how the HS strategy is able to adjust the ascending part of the CoM trajectory, while retaining a significant misestimation in the first part of the trajectory.

Figure 3.12: *The comparison of the test results from both models with the vertical CoM trajectory of the motion capture data shows that even a simplified heel strike model can explain the divergence between the inverted pendulum model and the human behaviour. Nonetheless, there is still a divergence of behaviour connected to the toe off strategy that is not yet considered by the proposed model.*

Lastly, the AP swing trajectory simulation has a difference of 8.41 ± 3.75 degrees in the slope compared to the motion capture trajectories. Figure 3.13 shows the difference in the last part of the swing when there is the transition of the supporting foot, indicating a modification of the trajectory strategy during the execution of the toe-off and heel-strike.

Figure 3.13: *The comparison between the motion capture AP swing trajectory and the output of the model shows the models' ability to adequately track most of the trajectory. In fact, the divergence is localised about 100ms before the HS at the beginning of the Toe-Off period, when the body is pushed forward by the rising of the support foot.*

Validation of BoS and MoS

To quantify the model accuracy of Equations 3.5 and 3.8, 12 healthy subjects without any medical history of neural or cardiovascular system diseases have been recruited (IRB-2016-09-015). The subjects were asked to explore their BoS in 4 different feet postures as presented in Figure 3.14, while their movements were recorded via a motion capture system.

- *Subjects' Demographics:* The recruited healthy people are 5 females and 7 males. The age is 28.4 ± 3.4 years old; the height is 171 ± 10.7 cm and, the weight is 71.2 ± 12.6 kg.
- *Experimental Set-Up:* the subjects' movements have been recorded via an 8 Camera Vicon Motion Capture System and the data elaborated

Figure 3.14: *The four feet postures recorded during the experiment.*

with the included Nexus Software (Vicon Motion Systems Ltd, UK). The passive markers have been placed in the following anatomical reference points:

- *Marker-1*: Above the Right Spina Iliaca Anterior Superior.
- *Marker-2*: Above the Left Spina Iliaca Anterior Superior.
- *Marker-3*: Above the Right Spina Iliaca Posterior Superior.
- *Marker-4*: Above the Left Spina Iliaca Posterior Superior.
- *Marker-5*: Right Heel.
- *Marker-6*: Base Knuckle of the Right first toe (1^{ST} Metatarsal Bone).
- *Marker-7*: Base Knuckle of the Right fifth toe (5^{TH} Metatarsal Bone).
- *Marker-8*: Left Heel.
- *Marker-9*: Base Knuckle of the Left fifth toe (1^{TH} Metatarsal Bone).

– *Marker-10*: Base Knuckle of the Left fifth toe (5^{TH} Metatarsal Bone).

- *The instruction given to the subjects*: the participants were instructed to explore their stability by moving their pelvis for about 30 seconds for each posture. The feet position should have been maintained, but the lift of either the heel or the metatarsus were allowed as long as the foot orientation were maintained.

Results and Discussion:

The 10 markers placed on the human body allow deriving the trajectories of CoM and the two CoPs, using the methodology described in Appendix A.

Figure 3.15 reports the comparison between the measured CoM trajectory and the MoS calculated using Equations 3.5 and 3.8 in the 4 postures, showing a good degree of accuracy in tracking the transformation of the BoS. Furthermore, Figure 3.16 confirms that the proposed model allows tracking the transformation occurring to the mean BoS recorded for the different postures. This validates the geometrical transformation between the Saddle and Task Spaces (Equation 3.5), where the mean BoS is the confidence ellipse associated with the covariance analysis of the matrix composed by CoM trajectories of the 12 subjects in the transverse plane.

Subsequently, the accuracy of the BoS geometry has been analysed using the following two parameters:

1. Percentage of Inclusion: $PoI = 100 \text{ (Number of Sample in the BoS) / (Number of Samples) .}$
2. Percentage of Inclusion 360: $PoI_{360} = 100(N_S)/(360)$, where N_S are the number of samples included in the BoS when considering only the

Figure 3.15: *The comparison between the model output and subject's data show that the model has a good level of accuracy in all 6 postures, although the subject is able to exceed the boundaries identified by equation 3.5 that is coincident with the BoS portrayed in Posture 1.*

CoM sample at the maximum distance from the Saddle point in every direction.

The analysis of the two parameters shows that, in both cases, the mean percentage of inclusion are above 87.9%, as shown in Figures 3.17 and 3.18.

The data also shows the presence of two outliers highlighting the main limitation of this postural model of the BoS; it considers only the pelvic and foot relative positions. This is an intrinsic limitation of having the MoS based on the geometry of the BoS because this boundary can be exceeded

Figure 3.16: *The confidence ellipse of the CoM covariance analysis (Covariance Ellipses) in the transversal plane shows that Equation 3.5 is able to transform the geometry of the BoS to track the effect of a change in the foot posture. The MoS used in this analysis has been calculated according to Equation 3.5, using the mean positions of the feet (CoP_r and CoP_l) used from the 12 subjects as input for the two CoPs.*

without compromising the stability when it is possible to counteract the external forces acting on the system [6]. Such limitation can be overcome by accounting for the CoM kinematics, which allows evaluating if the forces are being somehow compensated (e.g. change in the posture or using the hand to provide additional support), or the dynamic state of the body is heading toward instability (i.e. fall).

Figure 3.17: *The statistical analysis of the PoI shows a level of inclusion proportional with the step length, which may be related to the reduction in available range motion in the different postures. The PoI is 87,9% at its minimum where it also reaches its higher variability.*

In conclusion, the proposed model is accurate in tracking the BoS geometrical shape in the different posture. However, the information is not sufficient in evaluating patient stability, which requires the analysis of the CoM kinematics. Furthermore, this experiment proves that it is possible to track the BoS by measuring the heel position and the orientation of the foot when the anthropometry of the subject's feet is known. Therefore, it provides the information required to choose the sensors needed on the follow-me wheelchair to assess the patient equilibrium state.

Figure 3.18: *The PoI360 considers only the farthest point from the Saddle Point reached by the CoM trajectories in every direction. Although the data show a similar trend to the PoI data shown in Figure 3.17, they confirm the presence of the two outliers.*

3.3 Bipedal Stability and Energy Analysis

The proposed model shows that it is possible to achieve a better estimation of the CoM energy from the body kinematics by including the toe off and heel strike strategies. Therefore, it allows a better evaluation of the walking stability in both the AP and ML directions by kinematic measures. However, the execution of this evaluation requires defying stability criteria and, formulating a walking planner that is able to generate a reference trajectory for performing such evaluation.

3.3.1 CoM trajectory Planning

The Bipedal System (Figure 3.4) has 3 fixed points as shown in the phase diagram portrayed in Figure 3.2. They are the two fulcra of the two pendula and the saddle point. These points allow the subdivision of the CoM trajectory (Equation 3.12) in four recurring phases characterised from four via points:

1. *Left Via Point* ($\vec{x}_{Lvp}$): The CoM moved from the saddle point towards the fixed point of the left foot (1st quadrat of the phase diagram in Figure 3.19).
2. *Left Saddle Via Point* ($\vec{x}_{LSvp}$): The CoM moved from the left foot fixed point towards right foot fixed point (4th quadrat of the phase diagram in Figure 3.19).
3. *Right Via Point* ($\vec{x}_{Rvp}$): The CoM moved from the saddle point towards the fixed point of the right foot (3rd quadrat of the phase diagram in Figure 3.19).

4. *Right Saddle Via Point* ($\vec{x}_{RSvp}$): The CoM moved from the right foot fixed point towards left foot fixed point, (2nd quadrat of the phase diagram in Figure 3.19).

This results in the following formulation:

$$\begin{aligned}
 \vec{x}_{Lvp} &= \begin{cases} x_{lvp} = (\frac{d_{SW}}{2} - A_y)\cos(\alpha_{//}) + x_{CoPl} \\ y_{lvp} = (\frac{d_{SW}}{2} - A_y)\sin(\alpha_{//}) + y_{CoPl} \end{cases} \\
 \vec{x}_{LSvp} &= \begin{cases} x_{svp} = (\frac{d_{SW}}{2})\cos(\alpha_{//}) + x_{CoPl} \\ y_{svp} = (\frac{d_{SW}}{2})\sin(\alpha_{//}) + y_{CoPl} \end{cases} \\
 \vec{x}_{Rvp} &= \begin{cases} x_{rvp} = (\frac{d_{SW}}{2} - A_y)\cos(\alpha_{//}) + x_{CoPr} \\ y_{rvp} = (\frac{d_{SW}}{2} - A_y)\sin(\alpha_{//}) + y_{CoPr} \end{cases} \\
 \vec{x}_{RSvp} &= \begin{cases} x_{svp} = (\frac{d_{SW}}{2})\cos(\alpha_{//}) + x_{CoPr} \\ y_{svp} = (\frac{d_{SW}}{2})\sin(\alpha_{//}) + y_{CoPr} \end{cases}
 \end{aligned} \tag{3.15}$$

where $a_{//}$ is the angle between y_{saddle} and the world reference frame when both feet are laying on the body sagittal plane as in Figure 3.5.(a). This formulation will allow the generation of a CoM trajectory while walking at a generic velocity (v_{CoMP}) as shown in Figure 3.4, if the step length (d_{SL}), the step width (d_{SW}) and the step frequency (ω_0) are known.

3.3.2 Relationship between Walking Velocity, Step Length and Step Frequency

Equation 3.11 identifies a linear relationship between the walking velocity and step frequency. Using the data from the KITDB, it is possible to identify a linear fitting for the data that results in the following relationship:

Figure 3.19: The via points of the CoM trajectory constitute the intersection between the close loop and the axis of the phase portrait.

$$d_{IdSL/2}(v_{CoMP}) = \frac{d_{IdSL}(v_{CoMP})}{2} = 0.1802v_{CoMP} + 0.1351[m] \quad (3.16)$$

the fitting goodness is: SSE= 0.0015, R-square= 0.98, adjusted R-square= 0.98 and RMSE= 0.005.

Furthermore, it is possible to identify the two borders of stability for the walking strategy. The first is based on the XCoM condition of stability, which identifies the limit cycle to be [35]:

$$\omega_{step} = 2\omega_n = 2\frac{\sqrt{g}}{\sqrt{0.57h_{Body}}} \quad (3.17)$$

where g is the gravitational acceleration, h is the body height and ω_n is the natural frequency of the pendulum. The second is identified by the maximum position that can be reached without jumping with fixed point

Figure 3.20: The schematic shows the concept of having a sequential optimal planning of the step at the desired speed. The bipedal system has 3 fixed points (2 absolute maxima and the saddle point), suggesting that an optimal trajectory has to pass through the via points that are connected with the position of the fixed points. Furthermore, an optimal trajectory should be, as much as, possible along y_{saddle} , allowing to nullify the forces acting normally along x_{saddle} . This is consistent with the proposed implementation for the swinging trajectory as in Equation 3.9.

(either Foot or Saddle). The combination of Equation 3.16 with the two stability conditions is graphically represented in Figure 3.21. Moreover, they allow defining the following equation that allows the quantification of the stability during walking.

$$\begin{cases} S_{SL_{Pend}}(t, v_{CoM_p}) = \frac{M_{d_{SL/2}}(t) - d_{SL/2_{min}}(v_{CoM_p})}{d_{SL/2}(v_{CoM_p}) - d_{SL/2_{min}}(v_{CoM_p})} \\ S_{SL_{Jump}}(t, v_{CoM_p}) = \frac{M_{d_{SL/2}}(t) - d_{SL/2_{Max}}(v_{CoM_p})}{d_{SL/2}(v_{CoM_p}) - d_{SL/2_{Max}}(v_{CoM_p})} \end{cases} \quad (3.18)$$

where $M_{d_{SL/2}}$ is either the chosen or measured step length. The first condition

in Equation 3.18 is the limit introduced by the inverted pendulum limit cycle, while the second evaluates the reach limitation.

Figure 3.21: The regression of half of the step length in function of the walking velocity confirms the linear relationship hypothesised with Equation 3.11. The data from Ordendurff *et al.* [63] also confirms such relationship; however, there is a slight discrepancy of behaviour that is consistent with different walking strategies used by the two groups of subjects. However, both data sets are within the stable walking area delimited by the XCoM-based stability condition on the right, and the border between walking and running on the top ($d_{SL/2_{MAX}}$).

3.3.3 Relationship between Walking Velocity and Step Width

The relationship of the stepping width and the velocity is more complex. The high variability of the data (Figure 3.22) indicates the presence of other factors influencing the regulation of the step width. However, it was possible to define a model for selecting the step width in base of the velocity. The identified relationship presents a double saturation as defined in the following equation and as shown in Figure 3.22.

$$d_{IdSW}(v_{CoMP}) = \begin{cases} 0.22, & v_{CoMP} < 0.6[m/s] \\ -0.2128v_{CoMP} + 0.3456, & v_{CoMP} \in [0.6, 1.1] \\ 0.10, & v_{CoMP} > 1.1[m/s] \end{cases} \quad (3.19)$$

where d_{IdSW} unit is meters. The fitting goodness is: SSE= 0.0080, R-square= 0.157, adjusted R-square= 0.154 and RMSE= 0.038. The low values of both R-Squares are consistent with the high data variability as can be observed in Figure 3.22. The medio-lateral stability condition can be defined by imposing that the CoM lateral position is constrained within the two feet as defined by the MoS in Equation 3.8, due to the limited amount of force that can be generated outside these boundaries [75]. This allows defining the following stability condition:

$$S_{SW}(t) = \begin{cases} 1 - \frac{y_{CoM_{saddle}}(t)}{y_{CoPl_{saddle}}(t)}, & \text{if } y_{CoM_{saddle}}(t) > 0 \\ 1 - \frac{y_{CoM_{saddle}}(t)}{y_{CoPr_{saddle}}(t)}, & \text{if } y_{CoM_{saddle}}(t) < 0 \end{cases} \quad (3.20)$$

where all the quantities are expressed with respect to the saddle space frame.

Figure 3.22: The regression for the step width also allowed the identification of a linear relationship, but there is evidence of saturation for both high and low velocities. However, the step width data from the KITDB show higher variability, indicating that it is not strictly regulated as the step length from an optimal forward policy. On the contrary, it seems to be optimised for every step.

3.3.4 Importance of the Toe-Off and the Heel-Strike Strategies for the Walking Energy and Stability

The relationship between the ankle strategies, equilibrium and walking energetics is vastly reported in literature [5, 21, 67, 113, 130, 132–134]. Although, there have been many attempts to redirect the inverted pendulum model to adequately address the walking energetics, they are mainly focused on estimating of the metabolic cost rather than making a balance assessment [57].

Currently, the equilibrium analysis still relies on the measure of the ground reaction forces to correct the divergence of the inverted pendulum model for the energetics of walking [20, 75, 135].

However, Figure 3.2.6 shows how introducing feet postures into the inverted pendulum model may correct its energetics by accounting for both the change of length due to the legs' postures and the virtual elongation due to the ankle posture. To better evaluate the contribution of the ankle strategies to the walking energetics, the model in Equation 3.10 is extended to include the Toe-Off phase as follows:

$$h_{CoM}(t) = \begin{cases} h_{CoM_{r/l}}(t), & \begin{aligned} & \text{if } d_{AP}(t) < d_{SL} \\ & -d_h - v_{CoM_p} t_{TO} \\ & = d_{AP}(t_{TO_{start}}) \end{aligned} \\ \\ h_{CoM_{r/l}}(t) + 2d_h \sin(\theta_{TO}(t)), & \begin{aligned} & \text{else if} \\ & d_{AP}(t) < d_{SL} - d_h = \\ & d_{AP}(t_{TO_{start}} - t_{HS_{delay}}) \end{aligned} \\ \\ h_{CoM_{l/r}}(t) + 2d_h \sin(A_{Max}), & \begin{aligned} & \text{else if} \\ & sgn(y_{CoM_{saddle}}(t)) = \\ & sgn(y_{CoM_{saddle}}(t - 1)) \end{aligned} \\ \\ h_{CoM_{l/r}}(t) + 2d_h \sin(\theta_{HS}(t)), & \begin{aligned} & \text{else if} \\ & \theta_{HS}(t) > 0 \end{aligned} \\ \\ h_{CoM_{l/r}}(t), & \text{else} \end{cases} \quad (3.21)$$

where

$$\begin{cases} \theta_{TO}(t) = \frac{A_{Max}}{m_{TO}}(t - t_{TO_{start}}) \\ \theta_{HS}(t) = \frac{A_{Max}}{m_{HS}}(t - t_{TO_{start}} - t_{HS_{delay}}) \end{cases}$$

t_{TO} being the time length of the Toe-Off strategy, A_{Max} the maximum foot angle, d_h based on anthropometric data to be $0.076 h_{body}$ [42], d_{AP} the distance between the feet on the Antero-Posterior direction and, m_{TO} and m_{HS} constants for adjusting the slope of the TO and HS angles trajectories. Although this model is a simplified approximation of the two strategies, this allows estimating the impact of this strategy on the CoM energy.

3.3.5 Validation

Firstly, the Equations 3.6-3.12, 3.15, 3.16, 3.19 and 3.21 have been used to implement a planner based on the architecture described in Figure 3.23. This provides a tool for simulating human walking and studying the influence of the TO-HS strategies on the walking energetics. Figures 3.24 and 3.25 shows two trajectories at the same speed. The first figure is obtained with the data generated by the planner, while the second is the trajectory from the initial step of movement from the KITDB. The two figures show similar behaviours by passing through the saddle point even though they start from different initial conditions. This indicates the presence of a control policy driving the CoM to converge toward a trajectory congruent with the one generated by our model. Thus, it proves that the human motor control uses a strategy congruent with the 4 via point segmentation proposed by our model. Sequentially, the influence of the TO-HS strategies on the CoM energetics and stability is evaluated.

Figure 3.23: *The optimal forward planning architecture implemented takes the body height, the mass and the desired velocity as inputs. Then, using the phase segmentation introduced by the via point architecture generates the full characterization of the CoM trajectory. This provides not only the kinematic parameters at output but also the expected gravitational forces and the BoS along the entire trajectory. However, the current planner does not automatically select the heel strike strategies which are manually selected and tuned before the simulation.*

Influence of TO-HS Strategies on the System Energetics

The investigation of the ankle strategies influence on the system energetics begun with setting the parameters required from Equation 3.21. The analysis considered maximum angles of 5, 10 and 15 degrees in agreement with the mean and standard deviation in Figure 3.9. The other parameters used in

Figure 3.24: *The simulated trajectory shows the ability of the proposed planner architecture to produce an output consistent with the expectations. It passes through the identified via points having a trajectory consistent with human movement, as observed when comparing these results with the human trajectory in Figure 3.25.*

the simulation can be found in Appendix B. An example of the trajectories produced by the different angles is shown in Figure 3.26, which shows how the change of ankle strategy can greatly modify the vertical motion of the CoM.

The importance of the ankle strategies on the walking energetics is confirmed by both Figures 3.27 and 3.28. Notably, the latest figure describes how the ankle strategies can be used to regulate the gravitational forces acting on the sagittal direction, while it has a contained effect on the Medio-Lateral

Figure 3.25: *Human behaviour during the acceleration phase is consistent with a reference trajectory going from the foot right via point to the saddle via point at the desired final walking speed. The presence of points very close to and outside the IBoS is noticeable in the initial part of the trajectory. This indicates a de-synchronization of the swinging foot from the CoM trajectory. This phenomenon is commonly observed during acceleration and deceleration allowing the use of gravitational forces for supporting the acceleration and the deceleration. Furthermore, these behaviours prove that despite being locally unstable, the global stability is not compromised if the body is able to bring the CoM inside the BoS with a change of its posture.*

force component. Thus, it allows to adjust the inverted pendulum model to the experimental observation on the ankle contribution in controlling bipedal

locomotion [28, 35, 43, 55, 60, 67, 71, 113, 128, 129, 136–140].

Figure 3.26: *The Toe Off (TO) and Heel Strike (HS) strategies deeply affect the vertical trajectory of the CoM. This makes them an invaluable tool in the equilibrium management by enabling the body to counteract sudden perturbations and to optimise the energetic cost of walking.*

Trajectory Generation and Stability Conditions Accuracy

The comparison between Figures 3.24 and 3.25 shows the ability of the proposed architecture to mimic human walking trajectories. However, this comparison provides only a qualitative analysis of the planner performance. A quantitative evaluation can be obtained by analysing the discrepancy between the desired behaviour that is provided as input to the model from Equation 3.16 and Equation 3.19 and the model output. The analysis of the step length can be done via a direct measure of the step length of the feet

Figure 3.27: *Ankle strategies are important in analysing system dynamics because they can partially account for the potential energy misestimation. However, it should be observed that the angle range has been chosen in accordance with the KITDB data, which has a maximum velocity about 1.2 m/s. This makes it impossible to determine if the increased discrepancy at 1.6 m/s is due to a bigger angle or the intervention of other strategies.*

trajectory generated for the feet, which is reported in Figure 3.29. The validation of the amplitude ML-trajectory of the CoM A_y can be validated by verifying that the minimum value of the ML stability (Equation 3.20) produced by the planner is consistent with human behaviour. The data obtained from this latest analysis are shown in Figure 3.22. Given the consistency of

Figure 3.28: *Aside from their significant impact on the system strategies, ankle postures also introduce a huge reduction in the AP component of the gravitational force. On the other hand, their effect is less evident along the ML direction where they induce at most a 10% reduction in the gravitational force, underscoring the significant role played by the ankle in keeping the balance.*

both the output with the respective reference data, it can be concluded that the Equations 3.16 and 3.19 accurately model the human behaviours.

Lastly, it should be observed that for the current analysis, both the Equations 3.18 and 3.20 are considered as adequate metrics of the stability because they are both implementation of pre-existing metrics that are currently used in human balance studies [35, 52, 53, 56, 71, 74–76, 88, 96, 141].

Figure 3.29: The $d_{SL/2}$ measured from the output of the proposed planner show how this architecture is able to reproduce the desired behaviour provided at the input.

3.4 Relationship between the Dynamic and Static Stability of Bipedal Systems

The Base of Support (BoS) is defined as the region of the space where the body is able to counteract the gravitational forces [6, 49, 56, 81, 107], implying that the BoS is determined by the stable point of the bipedal system. Although this condition guarantees the static stability of the system, it is only a sufficient condition for dynamic stability.

Furthermore, the bipeds are reconfigurable mechanical systems. Their MoS is posture dependant while attaining the dynamical stability requires

Figure 3.30: *The amplitude of the CoM ML-trajectory (A_y) cannot be directly compared with an exact desired output behaviour. However, using the ML stability condition, it was possible to verify that the produced output is consistent with the behaviour of the data sets used as reference for the planner implementation.*

adequate movement of the legs to allow the system stabilization. This leads to the definition of the following two types of BoS [141]:

- *Instantaneous-BoS (IBoS)*: This BoS is associated with the current posture. It allows evaluating if the static stability can be reached in the current posture or there is a need for a postural adjustment. Therefore, the IBoS makes it possible to verify the stability in static conditions that do not require a modification of the feet posture on the transversal directions.

- *Ballistic or Expected-BoS (BBoS)*: this BoS is associated with the expected future posture of a movement, allowing the prediction of the stability conditions, thus, an important tool in evaluating dynamical stability.

The two BoS provide the tools needed in evaluating and predicting system stability condition at a given posture by allowing the identification of the MoS. Once the MoS are identified, then the conservation of energy principle can be used to study the system energetics. The energy of a mechanical system can be written as:

$$U + K + W + E_{external} = 0 \quad (3.22)$$

where U is potential energy, K is kinetic energy, W is the energy contribution, and $E_{external}$ is the energy introduced by external forces. The potential and kinetic energies can be calculated with the Equations 3.6 and 3.12. This implies that given the system state and the expected configuration, the active work required by the body for maintaining the stability in the desired configuration is as follows:

$$W = U(MoS) - U(Com) - K - E_{external} \quad (3.23)$$

where $U(MoS)$ is the potential energy of the margin of stability in the direction of the motion. However, when gravity is the sole exogenous energy, then Equation 3.22 defines the active work to be dependent only from the body kinematics and posture during unperturbed walking. For the extension to a general case, it requires the introduction of either a measure or an estimation of the external perturbations.

3.4.1 BoS as a Region of Attraction

The regions of attraction are defined as a subset of the space around an equilibrium point where a system can be stabilised, congruent with the definition of BoS [82, 118, 142]. Particularly interesting in this control architecture is that it can be extended to time-variant systems using the region of finite time invariance; where the points within the region can be considered to be time invariant [142].

Majumdar *et al.* [142] demonstrated that a robust on-line planner can be implemented if it is possible to define a region where the system can be considered invariant for small periods of time. This definition of region of finite time invariance matches the concept of IBoS, which describes the dynamical conditions that currently surrender CoM. Meanwhile, the BBoS can be used as region of finite attraction to identify the stability condition associated with a future action. Thus, it allows the identification of the global constraints for the on-line planner implemented by the IBoS. However, both systems require a Lyapunov-based stability methodology. Equation 3.22 is a Lyapunov candidate for both controllers. This allows to generate a control architecture where the BBoS and IBoS are integrated together in a single control architecture as shown in Figure 3.31. The only component of the controller that has not been considered in this study is the modelling of the active energy generated by the bipedal structure (W). However, the existence of a Lyapunov function for such structures have already been demonstrated in multiple studies and is commonly used in balance stability analysis [76, 77, 82, 142–148].

3.4.2 Human Balance Motor Control Parallel

One of the features of the finite time invariance control architectures is that complex behaviours of the system are obtained by combining of basic motion strategies organised in libraries [142]. It enables this architecture to be an ideal candidate for analysing how human motor control combines motor primitives and synergies for implementing of complex actions in non-linear and time-variant tasks. In order to validate the hypothesis that humans may rely on such architectures, both the current knowledge regarding human motor control and the walking strategies observed during walking will be analysed.

The first step is identifying possible candidates for implementing the different components depicted in Figure 3.31. The main components comprising such architecture are the following:

- *Global Planning in the Cartesian space and global stability* is responsible for action planning and deciding the policy of interaction, modelling the voluntary actions and the limit to the stability imposed by the desired task. The BBoS forward controller in the Cartesian space allows the identification of the global constraints to be imposed to the lower controllers for the achievement of global stability.
- *Local Planning in the Joint Space* is a centralised controller tasked to generate the desired joint trajectory in accordance with the directives from the forward controller. The controller relies on the IBoS as the region of finite invariance for the identification of the condition for the local stability. One possible implementation for the joint planning is the model proposed by Tommasino *et al.* in [149]. Their method uses non-linear inverse optimization to implement human-like movement and is

currently implemented only for pointing strategies. It is particularly suited to be included in the proposed architecture, because it allows selecting a sequence of consecutive postures by analysing a cost-based function called discomfort function [149, 150]. This is an energy-based function used for selecting the fittest posture among a postural library based on motor synergies and primitives [149, 150]. The idea is that the Global Planner and the local constraints from the IBoS will be used for reducing the space of solution available for the joint planning. This will then choose the subsequent posture deployed in the movement based on the discomfort function. One of the most important characteristics of this latest function is that it can be regulated to provide different solutions. This allows the implementation of a variable motor policy based on the task requirements by enabling the tuning of the controller in the selection criteria [149]. For example, if a task requires precise space-positioning, the discomfort function will promote the posture with better precision over the one with less energy cost. When the task does not impose strict constraints in the operational space, the discomfort function will promote energy-efficient postures.

- *Decentralised Feedback Control* is composed of lower level feedback controllers that control the joints during the execution of the directives from higher level controllers. Furthermore, the gain of these controllers are not static but are regulated by the policies chosen by the higher control for the task. This implies, that their gain should be increased when the task requires a higher reluctance to external perturbation and *vice versa*.

Autonomy of the Periphery in Counteracting Unpredictable Perturbation

The proposed architecture is constituted by the combination planners in a cascade that identifies the desired joint behaviour. It also relies on a decentralised feedback control architecture for executing and stabilising the planned joints' trajectories. Thus, the proposed controller has a semi-autonomous architecture, similar to the one hypothesised for the lower-limbs motor control by Ahn and Hogan [5]. This allows overcoming the limitation introduced from neural delay, which represents one of the main challenges of a feedback-based architecture of the motor control in dynamic conditions [28, 37, 45, 54, 151]. Furthermore, this can also explain the human flexibility in interacting with the exogenous phenomena by modifying the body compliance via a regulation of the feedback controller gains. These behaviours have been observed in multiple studies about the oscillatory motions of the legs. They show how these movements can be generated and stabilised by a decentralised controller that rely only on the brain for directives and synchronization [42, 81, 88, 109]. Especially important among them are the results obtained by Ivanenko *et al.* [109] and by Jo *et al.* [42]. The first shows how a variegated set of control policy can be obtained by controlling the pattern of activation in the spinal cord. The second proves that this architecture is able not only to generate an adequate pattern but also to counteract external perturbations.

This confirms the initial hypothesis made in the formulation of the control architecture that the higher controllers do not directly intervene in the immediate response to perturbation. Instead, they act as supervisors and evaluate whether the energy introduced in the system by an external perturbation is compatible with the current task or there is a need to change the

movement strategy to overcome the perturbation.

It can be concluded that the best candidates for implementing the feedback controllers are the spinal cord and the inter-muscular feedback that modulates the input from the α motor neurons [36].

Involuntary Supervision and Replanning

The IBoS based planner provides the joint planning based on the directive provided by the voluntary planner, and by both the proprioceptive and exteroceptive informations. That is usually described in motor control models as a slow feedback controller [36, 38].

The architecture proposed in the first part of this section is based on the concept of planner within the region of finite time invariance. It differs from the traditional architecture based on slow feedback control because it does not rely on the complete on-line knowledge of the sensorial information in generating the desired behaviour and allows the direct integration of the visual and the proprioceptive information.

This is possible because the region of finite time invariance that guarantees the models of the entities involved in the planning are time invariant within its borders. Thus, it allows having accurate predictions regarding the system evolution. However, the final stabilisation requires lower hierarchy semi-autonomous controllers that counteract the perturbations occurring before the planner can issue the adequate countermeasures.

The proposed architecture may be implemented by combining the pre-motor and motor cortexes. They are also supported by the parietal cortex and the cerebellum that provide state estimation and stability supervision needed for planning. Furthermore, they revive input from voluntary control issuing directives in terms of Cost to Go Function, implemented in the pro-

posed model by the discomfort and in the brain from the basal ganglia via the regulation of the brain reward circuit [36, 38].

The evidence in support of this architecture is found in multiple studies analysing the response of the nervous system to external perturbations. Maki *et al.* observed that the early balance intervenes in about 80 ms, implying that it is impossible for the motor-cortex to intervene as it receives the information with a delay between 100 and 200 ms from the perturbation occurrence [37]. It effectively excludes any possibility of brain involvement in an immediate response to external perturbation affecting local stability. Especially, considering that even if the brain issues a corrective response, by the time the action is implemented by the musculoskeletal system, it may be counterproductive to the stability.

On the contrary, if the brain acts with a strategy of planning and surveillance, it simply has to evaluate if the energetic input from the external perturbation can be absorbed during the execution of the ongoing action and re-plan otherwise. Such behaviour is consistent with a multitude of behaviours observed in humans during blindfolded walking and their reaction to external perturbations [71, 75, 96, 152–154]. These studies report that when the subjects are alerted of the possibility of tripping, they increase the clearance of the foot from the ground. However, they do not change the gait strategy in terms of temporal parameters such as walking velocity, step length and cadence [153]. This is consistent with our results which show a constrained linear relationship among these three parameters during walking, which impose loose limitations to the implementation on step width and joint control allowing the optimization of the locomotion [152, 154]. Furthermore, these studies also report that the reaction to external perturbation are always implemented in subsequent phases, with a first local reaction consistent with a

feedback. However, the later actions are the result of a replanning because they alter the global strategy of the movement by relocating the foot in agreement with the Extrapolated Centre of Mass model predictions [71, 75, 96, 154]. Another evidence in support of the brain acting as planner based controller is based on the findings of Hak *et al.* [76] and the effect of the pelvic fixation [155], that lead to a modification of the stepping frequency, step length and width in order to maintain a stability consistent with unperturbed walking. This type of phenomena has also been observed in other tasks such as the wrist-pointing task, wherein they found that the strategies converge to a single behaviour when the patients are coupled with a wrist robot without force feedback [156–158]. On the other hand, they observed that the variability of strategy emerges again when the force feedback was introduced in the robot control loop [157, 158]. This provides evidence that human planning accounts for the dynamical environmental conditions and also in predominantly kinematic tasks.

Voluntary Planning and Global Supervision

The voluntary planner is the forward planning arising from the decision of acting, which generates the desired global trajectory and identifies the global condition for stability. This is the only part that is currently implemented by our model, with the architecture described in Figure 3.23. The results presented in the previous section show that with an adequate knowledge of the body configuration and the environmental interaction, it is possible to generate human-like walking strategies and to identify the global condition for stability. However, this implementation also underscores that the voluntary planning needs to adequately account for both the possible environmental conditions and the body postures. This implies a strict interconnection with

the lower controllers which needs to provide the most accurate estimation of the expected condition. These properties are currently considered to be allocated in the frontal cortex which is usually believed to be the locus of high cognitive functions. It also has the required interconnection with the lower controllers via the pre-motor cortex [36, 38, 159].

It does not have an equivalent structure in current motor control models that do not account for a precise architecture of the voluntary choice. Nonetheless, they indirectly refer to it in the implementation of the cost function of the basal ganglia which accounts for the desired behaviour coming from the voluntary decision-making [36, 38]. The latter is represented in the proposed architecture from discomfort function which drives the involuntary planning and supervision. This part of the controller is a hybrid between voluntary and involuntary global supervision. In fact, the voluntary decision is represented by the input of the desired walking speed and direction. The supervision of the global condition of stability and its eventual replanning are meant to be autonomous responses that take into account the imposition of the voluntary planning. The anatomic analogous is hypothesised to be an internal model allocated in the cerebellum. This supervises the coherence of the incoming sensor information with the strategies implemented by the lower controllers. However, this hypothesis is coherent with both the current beliefs regarding its role in providing internal models and its anatomical placement allowing to adequately interconnect with all the entities involved in the generation of the motor planning [36, 38, 40].

Evidence in support of these theses can be identified analysing the finding from multiple studies. The aforementioned findings of Hak *et al.* and Veneman *et al.* show how constant perturbation are accounted in the global planning [76, 155]. Furthermore, it is also supported by the outcomes from stud-

ies comparing motion strategies of able-bodied subject *vis-a-vis* pathological patients, wherein they identify that people with compromised motor ability modify their motion strategies to compensate for the reduction of their motor abilities [52,124,160]. For example, multiple step balance recovery is still consistent with the Extrapolated Centre of Mass, which provides one of the global stability conditions in the proposed architecture [53,56,71,75,161,162]. This proves that the global supervision is constrained by the voluntary planning, but its reaction does not require a conscious driven intervention; thus, supporting the proposed thesis implemented in the cerebellum.

3.5 Summary

The model presented in this Chapter has been proven to have the ability to provide an adequate metric of the human balance in term of kinematics parameters which is the scope of this work. Nonetheless, the investigation leads to the definition of a novel architecture for studying human motor control. However, the theory is still in its early stage and crucial components of the architecture are yet to be developed. Nonetheless, the preliminary analysis made by comparing the implication of such architecture with the current knowledge of motor control and experimental observation provide sufficient ground for further exploring this theory.

Lastly, it shall be noted that the proposed architecture does not contrast with any of the available feedback based motor control theories. Current models are mainly designed based on experiments in kinematic dominated tasks, where the dynamics of the task is consistent to model and allow the identification of optimal behaviours. These models start to fail when the task dynamics become more complex. Identifying an optimal solution depends on

the local conditions, where the time-dependent dynamic systems continuously change, and a feedback based controller is not any more sustainable, especially, with the time delay introduced by the neural transmission. On the other hand, the proposed architecture has been shown to be theoretically able to account these limitations by providing the aforementioned hybrid architecture. Where the first stage provides a global optimised planning, the second stage adapts to the directives considering local constraints and energetics. The final local stabilisation needs to deal with unpredictable changes of conditions is left to the decentralised feedback controller. This controller has access to the information fast enough to implement a feedback reaction. One consequence of this type of architecture is that when the task does not occur in a challenging dynamic environment, the output converges to the initially planned optimal trajectory; thus, mimicking the feedback controller implemented by current models. Furthermore, the architecture described in Figure 3.31 has strong parallels with recent neuroscientific theories about the interaction between the human body and the environment. In particular, they report that the nervous systems allocate to different neural circuits the analysis of the environment and body dynamics which are subsequently combined for the movement generation [18, 19, 163, 164].

Figure 3.31: The planning architecture proposed in this model can be compared to the voluntary decision and consequent planning, that in the brain is implemented by the frontal cortex with the support of the pre-motor cortex. The latter handles the planned information to the motor cortex, which is tasked to issue the motor planning in the joint space to the lower feedback based controllers. The low-level controllers then handle the implementation of the appointed tasks, which also includes counteracting all the fast perturbations that bypass the direct brain intervention due to the high delay of the neural transmission. The interaction with the environment is then sensed by the sensorial system and elaborated in the parietal cortex with the support of internal models stored in the cerebellum. However, the latter also implement the task supervision through a particular kind of internal model which verifies that the task is executed within the margins of stability and trigger an eventual reaction from the two planners. These latter models derive the margins based on the planned action and are able to tune the reward centre allocated in the basal ganglia. A substantial theoretical evidence supporting such architecture is that its behaviour converges to one of optimal feedback controllers for kinematics dominated tasks like the one that has been used in the validating of current theories.

Chapter 4

Interface Design

The lower limbs and balance rehabilitation have been well studied in the last decade, but the research has particularly focused on realising tools that support the therapists to provide the traditional therapies. This leads to the development of devices that focused mainly on gait training either on a treadmill or off the ground exercises. The principal limitation of these approaches is the consistency of the walking surface (i.e. even ground and absence of obstacles), which fails to trigger and train the reflexes to counteract the perturbation introduced by the ground irregularities that are encountered in daily living environments. This resulted in the ineffectiveness of such technologies and the development of devices dedicated to the training of the reflexes at the ankle level [4, 5, 8]. However, this separates the two tasks making the therapy less effective rather than an approach that will target both [10].

A possible way of achieving a better balance rehabilitation is improving the interaction between the robots and the patient. In fact, a robot that is not transparent to the user will significantly alter the environment of the task, thus providing in a less effective therapy [31, 165]. The KineAssist

is the first attempt to move toward ADL based therapies while minimising the interference of the robot with the tasks. However, the experimental data shows a relevant interference with the walking strategies [14, 15]. The Robotic Walker from NUS shows the same interference as reported in by Mun *et al* [16].

The alteration of the gait strategies produced by both the Robotic Walker and the Kine Assist are consistent with the pelvic fixation with a significant reduction of the step width, velocity and step frequency [15, 16, 67, 155]. These phenomena are consistent with an alteration of the dynamic environment surrounding the pelvis, which has been proved to be pivotal in the planning of the locomotor strategies and balance in Chapter 3.

The structural design and of the interfaces also affects the manoeuvrability of these devices in unstructured environments. In fact, all these systems are controlled with a minimisation of the interaction force using force/torque sensing at the interface, but without having any environmental awareness [14, 16, 104]. This makes them suited for operating in controlled environments such as the hospitals, clinics and elderly centres.

4.1 Proposed Approach and System Design

The analysis of human locomotion dynamics and motor control presented in Chapter 3 has revealed that humans take advantage of the gravitational forces acting on their centre of mass during locomotion. Specifically, human strategy is comprised the following actions:

- The mediolateral CoM trajectory oscillates following the hyperbolic trajectories of an inverted pendulum oscillating at a frequency ω_{Step} .
- The vertical CoM trajectory is used to control the locomotion velocity

and stability and is controlled mainly via the ankle strategies during heel-strike and toe-off phases.

- The swing foot trajectories are controlled to have the foot always aligned with the segment between support foot fulcrum of rotation and the CoM which enable the CoM trajectory to occur along y_{Saddle} (Figure 3.3).

The aforementioned devices use either a rigid or high-stiffness interface to connect the robot to the patient pelvis that relies on the dynamic behaviour of the robot base to follow the patient and his movements. The unsatisfactory results mentioned above prove that the current technologies are probably related to their alteration of the body dynamics, which has been proven to be strictly entangled with both the planning and control of locomotion.

This work proposes an alternative to rigid or actively actuated interfaces by studying the feasibility of a low impedance interface able to minimise the perturbations induced to the body dynamics. Therefore, enabling the patient to experience a dynamic environment consistent with free walking during rehabilitation therapies and leading to the identification of the following characteristics for the interface:

- 5 Degrees of Freedom (DoFs) to accommodate the CoM 3-dimensional trajectories.
- The interface should include a compliant vertical degree of freedom to accommodate the CoM trajectory along this direction during locomotion.
- It should be able to allow a range of motion that guarantee the manoeuvrability of mobile robotic base in wheelchair friendly environments. A concept design of the envisioned robot is shown in Figure 4.1.

The design of the interface is presented in details in Section 4.2.

Figure 4.1: *The envisioned final device is supposed to follow the patient in the daily environment and to avoid the fall. The aim of this work is not to realise the entire device, but to prove its feasibility. The focus is on the development of a compliant interface and a balance model for identifying the patient's equilibrium from its kinematics.*

4.2 5-DoFs Selective Compliant Interface

The perturbation of the balance strategies due to the pelvic fixation has been identified as the main problem of the robotic walkers [14–16, 155]. The results obtained by Mun *et al.* in [16] show that even with a lighter structure and a performant holonomic base do not carry any substantial improvement in the transparency.

The aforementioned technologies all share a rigid interface that has proven itself inadequate to obtain a transparent interface. This phenomenon is mainly connected to a high intrinsic impedance of these systems that cannot be made transparent to the patient. However, the interface limitations are relevant particularly to task initialization, but they are counterbalanced by the benefit introduced for a patient requiring full body weight support [14, 15]. However, the limitation during gait initialization was relevant enough to drive HDT Global inc to transform the KineAssist pelvis from an overground trainer to a treadmill-based device. On the other hand, the only available technology is Andago from Hocoma AG that uses a compliant interface between the mobile base and the patients but at the best of my knowledge, there is no information regarding its performance.

Two sub-systems compose the interface proposed in this study: 3-DoFs implemented by the planar pentagonal chain and 2-DoFs introduced to interface the chain to the belt through a vertical parallel pulley system.

4.2.1 The 3-DoFs Pentagonal Chain

The development of the Pentagonal Chain started with the modelling of the kinematics and dynamics. Subsequently, a first prototype was built and used for the experimental validation of our models. The data collected in

Figure 4.2: *The kinematic schematic of the Pentagonal Chain in kinematics [166].*

the experiment were used to improve the design and develop a first control strategy for the mobile base. The design chosen for the 3-DoFs interface is the pentagonal close chain as in Figure 4.2. It has one additional degree of freedom of the 5-bar linkage due to its connection to the frame via two independent revolute joints [166,167]. It also introduces a circular symmetry around the frame that is only limited by mechanical constraints present in the frame design Figure 4.3. The link dimension has been chosen to be 250 mm each that allows a task-space radius of 400 mm from the chain reference frame as identified in the Figure 4.2.

Figure 4.3: (a) *The task-space of a chain with 250 mm linkages is compared with the dimensions of a standard wheel chair to provide an overview of the manoeuvrability introduced by the chain.* (b) *Three configurations on the margins of the chain task-space in relation to the Mobile Assistive Balance Trainer concept design. They show how it allows for manoeuvring even in small spaces to reach almost 90-degree angle between the patient and the base.*

Kinematics

The kinematics analysis of the mechanism allows studying how the joint design and the link length affect the shape of the task space as well as the motion kinematics [166, 168]. Therefore, the evaluation of the suitability of the design to the task becomes possible.

The Pentagonal Chain has been modelled as two three-link-arms that are joined together in the back of the patient [166, 168]. The frame has been placed in accordance with Denavit-Hartenberg methodology as shown

in Figure 4.2 [166, 168].

$$\left\{ \begin{array}{l} \mathbf{X}_{e1} = \begin{bmatrix} x_{e1} \\ y_{e1} \\ \phi_{e1} \end{bmatrix} = \begin{bmatrix} l(\cos(\theta_1) + \cos(\theta_1 + \theta_2) + \cos(\theta_1 + \theta_2 + \theta_3)) \\ l(\sin(\theta_1) + \sin(\theta_1 + \theta_2) + \sin(\theta_1 + \theta_2 + \theta_3)) \\ \theta_1 + \theta_2 + \theta_3 \end{bmatrix} \\ \\ \mathbf{X}_{e2} = \begin{bmatrix} x_{e2} \\ y_{e2} \\ \phi_{e2} \end{bmatrix} = \begin{bmatrix} l(\cos(\theta_6) + \cos(\theta_6 + \theta_5) + \cos(\theta_6 + \theta_5 + \theta_4)) \\ l(\sin(\theta_6) + \sin(\theta_6 + \theta_5) + \sin(\theta_6 + \theta_5 + \theta_4)) \\ \theta_6 + \theta_5 + \theta_4 \end{bmatrix} \end{array} \right. \quad (4.1)$$

where $\phi_{e2} = \theta_6 + \theta_5 + \theta_4 = \phi_{e1} + \pi = \theta_1 + \theta_2 + \theta_3 + \pi$. This allows defining the following relationships between the joints angles and the pentagonal geometry:

$$\begin{aligned} l_{13} &= 2l \cos\left(\frac{\theta_2}{2}\right) \\ l_{14} &= 2l \cos\left(\frac{\theta_5}{2}\right) \\ \alpha &= \arccos\left(\frac{-l^2 + l_{13}^2 + l_{14}^2}{2l_{13}l_{14}}\right) \\ \beta &= \arccos\left(\frac{-l_{14}^2 + l_{13}^2 + l^2}{2l_{13}l}\right) \\ \lambda &= \arccos\left(\frac{-l_{13}^2 + l_{14}^2 + l^2}{2l_{14}l}\right) \\ \theta_6 &= \theta_1 - |\alpha| - |\theta_5| - |\theta_2| \\ \theta_3 &= -|\pi - |\beta|| - \left|\frac{\theta_2}{2}\right| \\ \theta_4 &= -|\pi - |\lambda|| - \left|\frac{\theta_5}{2}\right| \end{aligned} \quad (4.2)$$

Inverse Kinematics: The 3-link representation also allows defining an arithmetic model for the inverse kinematics using the following equations and equation (4.2.1):

$$\begin{aligned} l_{13} &= \sqrt{x_3^2 + y_3^2} = \sqrt{(x_e - \frac{l}{2} \cos(\phi_e))^2 + (y_e + \frac{l}{2} \sin(\phi_e))^2} \\ l_{14} &= \sqrt{x_4^2 + y_4^2} = \sqrt{(x_e + \frac{l}{2} \cos(\phi_e))^2 + (y_e - \frac{l}{2} \sin(\phi_e))^2} \end{aligned} \quad (4.3)$$

Dynamics

The dynamics can also be formulated starting from the 3-link arm dynamics as in eq.(4.4), using the superimposition of forces.

$$B_n \ddot{\mathbf{q}}_n + (C_n - B_n \dot{J}_{An} + F_{dn}) \dot{\mathbf{q}}_n + F_{sn} \text{sign}(\dot{\mathbf{q}}_n) = J_{An}^T \mathbf{h}_{rn}, \quad n = 1, 2 \quad (4.4)$$

where B_n , C_n , J_{An} , F_{dn} and F_{sn} are the inertia, the Coriolis, the Jacobian, the stiction, and the static friction matrices of the 3-link arm, respectively [167, 168].

Where F_{dn} and F_{sn} are given by:

$$\begin{aligned} F_{dn} &= \begin{bmatrix} \alpha & 0 & 0 \\ 0 & \alpha & 0 \\ 0 & 0 & \alpha \end{bmatrix} \\ F_{sn} &= \begin{bmatrix} \beta & 0 & 0 \\ 0 & \beta & 0 \\ 0 & 0 & \beta \end{bmatrix} \end{aligned} \quad n = 1, 2 \quad (4.5)$$

The friction coefficients in the aforementioned matrices (α and β) are equal for all the joints due to their identical design.

Direct Dynamics: The direct dynamics of the 3-DoFs Pentagonal Chain is obtained by pre-multiplying both sides of eq.(4.4) for J_{An}^{-T} that leads to the task-space dynamics [167, 168]. Thus, it allows using the principle of superposition of forces for obtaining the chain dynamics as follows:

$$\begin{aligned} \mathbf{h}_r = \mathbf{h}_{r1} + \mathbf{h}_{r2} = & (B_{A1} + B_{A2})\ddot{\mathbf{x}}_{e1} + (C_{A1} + C_{A2})\dot{\mathbf{x}}_{e1} + \\ & + (F_{dA1} + F_{dA2})\dot{\mathbf{x}}_{e1} + F_S \end{aligned} \quad (4.6)$$

where:

$$\begin{aligned} B_{A1} &= J_{A1}^{-T} B_1 J_{A1}^{-1} \\ B_{A2} &= J_{A2}^{-T} B_2 J_{A2}^{-1} \\ C_{A1} &= J_{A1}^{-T} C_1 J_{A1}^{-1} - J_{A1}^{-T} \dot{J}_{A1} B_1 J_{A1}^{-1} \\ C_{A2} &= J_{A2}^{-T} C_2 J_{A2}^{-1} - J_{A2}^{-T} \dot{J}_{A2} B_2 J_{A2}^{-1} \\ F_{dA1} &= J_{A1}^{-T} F_{d1} J_{A1}^{-1} \\ F_{dA2} &= J_{A2}^{-T} F_{d2} J_{A2}^{-1} \\ F_S &= J_{A1}^{-T} F_{s1} \text{sign}(J_{A1}^{-1} \dot{\mathbf{x}}_{e1}) + J_{A2}^{-T} F_{s2} \text{sign}(J_{A12}^{-1} \dot{\mathbf{x}}_{e2}) \end{aligned}$$

Inverse Dynamics: Equation 4.6 allows calculating the task space inverse dynamics for the chain as [167]:

$$\begin{cases} \ddot{\mathbf{x}}_{e1} = B_{Chain}^{-1}(\mathbf{h}_r - (C_{Chain} + F_{dChain})\dot{\mathbf{x}}_{e1} - F_S) \\ B_{Chain} = B_{A1} + B_{A2} \\ C_{Chain} = C_{A1} + C_{A2} \\ F_{dChain} = F_{dA1} + F_{dA2} \end{cases} \quad (4.7)$$

Distribution of the forces between the two arms: The derivation of the input force distribution within the pentagonal chain can also be derived using the analogy with two welded two three-link arms [169]. In fact, given

the mechanical impedance of the two arms the force are distributed among them as follows:

$$\left\{ \begin{array}{l} \mathbf{H}_{r1} = Z_1 \dot{\mathbf{X}}_e \\ \mathbf{H}_{r2} = Z_2 \dot{\mathbf{X}}_e \\ \text{Which implies:} \\ \mathbf{H}_{r1} = Z_1 Z_2^{-1} \mathbf{H}_{r2} \end{array} \right. \quad (4.8)$$

where Z_1 , Z_2 are the mechanical impedances of the two arms. The relationship defined in eq: 4.8 leads to the following distribution of the input force:

$$\left\{ \begin{array}{l} \mathbf{H}_{r1} = (I + Z_2 Z_1^{-1})^{-1} \mathbf{H}_r \\ \mathbf{H}_{r2} = (I + Z_1 Z_2^{-1})^{-1} \mathbf{H}_r \end{array} \right. \quad (4.9)$$

where I is the identity matrix, Z_1 and Z_2 are the dynamics in the task-space of the two arms, $\mathbf{H}_r$ is the input force, while $\mathbf{H}_{r1}$ and $\mathbf{H}_{r2}$ are the forces distributed to the two arms. The presented formulation allows mapping the intrinsic impedance of chain as a function of the configuration joints.

Pentagonal Chain Design

The design validation has undergone different phases. Firstly, the kinematics and dynamics behaviours of the design have been studied to verify the suitability with the application requirements. Sequentially, a first prototype was developed to validate the models and their results.

The first prototype (Figure 4.4) is a passive system that weighs about 12 kg, made of stainless steel and aluminium. It was designed to serve as a test bench for validating the kinematics and dynamics model and to study both the linkage and the joint designs. Sequentially, the development was

Figure 4.4: *The rendering of the first prototype of the 5-DoFs interface.*

focused on minimising of the chain inertia and resolving some design problems identified with the first prototype.

Preliminary Analysis of the Kinematics and Dynamics: The study of the kinematics started with the identification of the link length. It has been chosen as an equilateral pentagonal chain, having 3 DoF on the transverse plane and, to maximise the chain workspace [166]. The links' lengths have been determined to be 250 mm, allowing the generation of a workspace that enables a good manoeuvrability when integrated with a wheelchair, as shown in Figure 4.3 [166, 167].

Subsequently, the intrinsic chain impedance was evaluated for sinusoidal movements along the sagittal and lateral directions under the assumption that joint friction is negligible [167]. The simulations account for the following movements' amplitudes —small (5 cm), medium (10 cm) and wide (Amplitude=15 cm) at different frequencies. The chain impedance provided by the dynamic model was then compared with the inertia of a 60 kg mass

moved along the same trajectory. Thus, allowing to quantify the percentage of energy absorbed by the chain during the execution of these trajectories as follows:

$$\left\{ \begin{array}{l} \mathbf{I}_H(t) = -\mathbf{M}_H \ddot{\mathbf{x}}_e(t) \\ \mathbf{I}_{tot}(t) = -\mathbf{I}_H(t) + \mathbf{h}_r(t) \\ \left[\begin{array}{c} I_{Xratio} \\ I_{Yratio} \end{array} \right] = 100 \left[\begin{array}{c} \sum \frac{|\mathbf{I}_{Xtot}(t)|}{|\mathbf{I}_{XH}(t)|} \\ \sum \frac{|\mathbf{I}_{Ytot}(t)|}{|\mathbf{I}_{YH}(t)|} \end{array} \right] \end{array} \right. \quad (4.10)$$

where $\mathbf{M}_H$ is the human body mass. The results reported in Table 4.2.1 show that the chain can be considered transparent during human walking, where it increases the inertial resistance by less than 5% [167].

A better understanding of the chain kinetics and dynamics can be obtained by the analysis of both the manipulability and force ellipsis [166, 168]. In fact, the geometrical relationships between the chain angles allow the redirection of both the manipulability and force ellipsis by tuning the impedance of the two three-links arms, as presented in Figure 4.5 where three possible actuation strategies have been considered. A more detailed analysis of this property will be required during the actuation design in order to exploit the intrinsic properties of the mechanism to maximise the performance in active balance support, which is beyond the scope of this work.

The Second Design: The first prototype has shown some weaknesses in the initial design. Firstly, the joints' design does not allow to properly tighten them, that introduces plays at the joint level that alter the system dynamics. Secondly, the frame is not stiff enough and the application of even a small load induces a significant deformation in the C-shaped support (Figure 4.4).

Amplitude [cm]	Frequency [Hz]	$I_{x_{ratio}}$ [%]	$I_{y_{ratio}}$ [%]
5	1	3.61	3.53
5	2	3.62	3.53
10	1	5.93	5.96
10	2	6.41	10.81
15	1	4.15	13.07
15	2	5.92	25.83

Table 4.1: *The intrinsic chain impedance introduced a small effect on the Antero-Posterior (AP, x -axis) direction regardless of the movement amplitude, which is fundamental to reduce the interference in the gait initialization. On the contrary, in the Medio-Lateral (ML, y -axis) direction, the chain impedance is constantly low for small amplitudes regardless of the frequency, but it increases for higher amplitudes. This allows the interface to intrinsically counteract excessive lateral movement while maintaining its transparency for walking ML movements that are supposed to be smaller than 5 cm [63].*

These limitations lead to a revision of the joint's design as shown in Figure 4.6. In contrast to the earlier joints that have a roller and a trust bearing, the new design uses two roller bearings (Figure 4.7). This configuration solves the joint play problem and brings the two arms to the same plane at the same time. However, the new coplanar configuration requires a redesign of the link geometry to maintain the range of motion of the first structure. Figure 4.8 shows how using an L-shaped link makes possible to obtain a broad range of motion of the joints in a coplanar configuration.

The weight reduction of about 2 kg for each of link was possible by re-

Figure 4.5: *The force ellipsis shows how different actuation strategies at the joints lead to many chain behaviours. When the chain is balanced, the ellipsis eccentricity is reduced. An unbalanced actuation leads to a rotation of its axis and an increase in the eccentricity. Such characteristics can be used to adapt the chain behaviour to the performed tasks.*

placing the 50x50 mm Aluminium links with 30x30 mm carbon fibre profiles and Al-7075-T6 shaft insert (Figure 4.9). However, the new link can support only up to 60 kg while the link in the previous design can support up to 100 kg. This has been compensated by the modular design that enables using two pentagonal chains in parallel, as in Figure 4.10, matching the performance of the first design while having a chain weight of about 6 kg.

The preliminary structural analysis of the new design executed with SolidWorks 2016 simulation tool (Dassault Systmes SolidWorks Corp., France) from the 3D CAD design revealed a safety factor of 1.7 when a 100 kg load is applied to a fully extended chain. However, being a preliminary analysis

Figure 4.6: *The render of the last prototype of the 5-DoFs interface. The picture includes six encoders to show that the chain has been designed to accommodate up to 6 actuators.*

the performance of the carbon fibre aluminium interface needs to be experimentally tested.

4.2.2 The 2-DoFs Parallel Pulleys Vertical Interface

The three-dimensionality of the centre of mass trajectory requires the addition of other 3 degrees of freedom in the system. However, the selection of the pelvis as an anchor point for our system allows considering the rotation about the mediolateral axis to be negligible in our application. Thus, the 2-DoFs interface was designed to allow a constrained vertical translation and the pelvic roll.

Initially, linear ball bearing guides have been considered for providing the vertical movement, but the application loads required using large diameters for the shaft to avoid grooving. This resulted in the guide to exceed both the dimensional and weight constraints. Subsequently, linear dry bearing

Figure 4.7: *The design of the joint connecting the pentagonal chain to the frame allows the connection of sensors and actuators to both links 1 and 5. In fact, the shaft of the first link can be accessed from the top, while the fifth link can be accessed from the bottom of the joint. Furthermore, by relying on four rollers bearing, it solves the play issue of the first design, and at the same time, it allows for assembly and disassembly of the joints.*

guides have been examined. Although they meet both the aforementioned constraints, their driving force can reach up to two times the load applied in the current application (Figure 4.11), making them unsuitable for the realising the vertical interface.

The proposed solution to this problem has been inspired by the traction drum arrangement lift, where the car and the counterweight are connected in a closed loop and the cable is routed through a pulley in the basement. This configuration is particularly suited to limit the driving force because when the belt and the counterweight are balanced, its resistance is generated only by the transmission losses.

Furthermore, this mechanism allows the pelvis roll which meets the harness system safety requirements by using two parallel independent transmis-

Figure 4.8: *The co-planarity of links in the new design would have introduced an excessive reduction of the joints' range of motions. To solve this problem, the geometry of the link was changed to an L-shape. Hence, the robotic arm is the virtual segment connecting the centre of the two shafts and, thus equal, to the hypotenuse of the right angle defined by the L-link.*

sion lines as shown in Figure 4.12.

A first prototype has been realised according to the design in Figure 4.12. Although the belt anchors and the counterweight are not balanced in the current set-up, it is possible to derive the intrinsic resistance using the following relationship:

$$\left\{ \begin{array}{l} F_R = F_{Out} - (M_{Belt} - \frac{M_{CW}}{\eta})g \\ \text{where:} \\ \eta = \frac{2M_{CW}g}{|F_{Out}^+ - F_{Out}^-| - 2M_{Belt}g} \end{array} \right. \quad (4.11)$$

where g is the gravitational acceleration, M_{Belt} the mass of the belt, M_{CW} is the counterweight mass, η the pulley efficiency, F_{Out}^+ and F_{Out}^- are the resistive force measure going upward and downward, respectively. The intrinsic resistance is 16.39 N, while the maximum force recorded during the experiments is about 22 N as reported in Figure 4.13. Furthermore, the evaluation

Figure 4.9: *The main issue when connecting the carbon fibre with other materials is the identification of the appropriate connection between the two. Usually, the inserts are directly glued to the profiles using an epoxydic glue, but for this application, the inner connectors were added to reduce the shear stress generated on the glue from a pulling force on the profile axial direction.*

of *eta* allows defining the balance condition between the belt weight and the counterweight to be $M_{CW} = \eta M_{Belt}$.

Lastly, the current design revealed that while the 2-DoFs can be considered to be a rigid interface for the planar forces, this is not true for the yaw where the ropes torsion allow misalignment of the belt and the end-effector. To address this shortcoming, a guiding system (Figure 4.14) has been designed to reduce the belt anchors torsional movement.

Figure 4.10: *The deployment of two parallel chains provides a full-body weight support for people up to 100 kg. The total weight of the interface in this configuration is about 5.5 kg without considering the electronics and sensors. The figure also shows the introduction of a cover in the 2-DoFs vertical interface. It also serves as the safety feature to avoid access to the counterweights from the front. Moreover, it solves the problem of a relative movement between the belt anchors and the third link introduced by the torsion of the ropes.*

Dynamics and Kinematics of the 2-DoFs Vertical Interface

Currently, the models are formulated under the ideal case that the 2-DoFs are ideally rigid on the transverse plane and its interference in the vertical direction negligible (less than 3% of a healthy adult body weight), as reported in Figure 4.13. Under the above assumptions, both models can be updated

Figure 4.11: The data is obtained with the Drylin[®] Expert 2.0 (IGUS[®], Germany) available at <http://www.igus.co.uk>. The linear guide has a driving force proportional to both the load and the acceleration applied. For example, the driving force required for our application is 130% of the load at 3 m/s² and reaches 200% at 9.8 m/s².

by considering the vertical interface as a part of the pentagonal chain third link and moving the end-effector to be in the belt anchors. The vertical position and roll angle are only acquired for measuring the human posture.

However, in the case where the system is not rigid, both the kinematics and the dynamics models can be updated considering each of the anchor to be associated with one arm of the pentagonal mechanism, where the end-effectors are moved from the middle of the third link to between the two

Figure 4.12: *The third link main body is composed of the 2-DoFs interface, with total weight of about 1.5 kg. The two parallel independent pulley mechanisms are decoupled for the design of the two anchors which can freely rotate about the rope connectors through the trust and flanged dry bearings. The current design is 25 cm height. It guarantees a minimum vertical movement range of 8 cm and a roll angle of about 30 degrees.*

anchors. Thus, enabling the definition of the Denavit-Hartenberg link between the two end-effector as follows.

The D-H matrix for the left arm (Joints 1-2-3 and left pulley) is:

$$T_{2DoFs_L} = \begin{bmatrix} c(\phi_A) & -s(\phi_A)c(\theta_{roll}) & s(\phi_A)s(\theta_{roll}) & d_{AC}(\phi_A) \\ s(\phi_A) & c(\phi_A)c(\theta_{roll}) & -c(\phi_A)s(\theta_{roll}) & d_{AS}(\phi_A) \\ 0 & s(\theta_{roll}) & c(\theta_{roll}) & d_{zA} \\ 0 & 0 & 0 & 1 \end{bmatrix} \quad (4.12)$$

Figure 4.13: The 2-DoFs Vertical Interface resistive forces to vertical movements have been conducted with optical force sensors (Optoforce OMD-30-FE-450N), connected to a PC via the included DAQ-USB interface and software. The DAQ sampling rate has been set to 100Hz with a low-pass filter at 15 Hz. The set-up has been used for collecting 80 vertical movements along the entire range of motion of the interface. Subsequently, they have been analysed with Matlab (Mathworks inc, USA). The recorded forces are 13.95 ± 3.76 N for the upward movements and 18.83 ± 2.31 N for the downward movements. Meanwhile, the intrinsic resistance calculated with Equation 4.11 is 16.39 N.

while the matrix for the right arm (Joints 6-5-4 and right pulley) is:

$$T_{2DoFs_R} = \begin{bmatrix} -c(\phi_A) & -s(\phi_A)c(\theta_{roll}) & s(\phi_A)s(\theta_{roll}) & -d_Ac(\phi_A) \\ s(\phi_A) & -c(\phi_A)c(\theta_{roll}) & c(\phi_A)s(\theta_{roll}) & d_As(\phi_A) \\ 0 & s(\theta_{roll}) & c(\theta_{roll}) & d_{zA} \\ 0 & 0 & 0 & 1 \end{bmatrix} \quad (4.13)$$

Figure 4.14: A low-friction rail was introduced to minimise the undesired yaw due to the cable torsional movement. Furthermore, the clearance left both on the back, and the lateral directions allow the cable connector to float in the rail and to enter into contact with the rail only in case of rotation. Lastly, the railing system panel was introduced to intercede the access to the counterweights from the front of the 5-DoFs interface (Figure 4.10).

where ϕ_A is the yaw angle between the 3-DoF interface end-effector and the anchors, θ_{roll} is the roll angle of the belt, d_A is the length of the virtual link connecting the anchor to the pentagonal mechanism end-effector, and d_{zA} is displacement of the belt from the middle of the pentagonal mechanism

The dynamic parameters will be required to be experimentally determined and continuously monitored due to their dependence on the rope tension.

Figure 4.15: *The current prototype before being integrated with the sensors.*

4.2.3 Sensors

The 5-DoFs interface (Figure 4.15) can be used to measure the dynamic interaction between the robot and the user by measuring the mechanism kinematics. The data needed to perform such measure are :

1. *3-DoF Interface Joint Kinematics*: It requires to place 3 encoders in the joints. Then, the other joints kinematics can be derived using the geometrical relationship in Equation 4.2. Additionally, it requires an accelerometer to be placed at the end effector to evaluate gravitational forces projected on the chain plane.
2. *2-DoF Interface*: It requires measuring the displacement and the roll angle in order to model the interaction dynamics. This can be achieved using distance sensors for measuring the position of the counterweights and integrating the data with the information collected by an IMU sensor placed on the harness belt.

The chosen sensors architecture is shown in Figure 4.16. The information were all routed to a LattePanda Board (RAM:4GB/Storage:64GB), which takes charge of the data elaboration. However, this measure chain is a preliminary stage because it is yet to be tested in the operational scenario.

Figure 4.16: *The chosen IMU is an LSM9DS0 9DOF Edison shield board from SparkFun Electronics (USA). The accelerometer and gyroscope are integrated on the Genuino 101 board that also acquires the counterweights' position from two SparkFun ToF range Finder Sensor-VL6180X. Lastly, the encoders are 3 AMS-AS5047P 16-bit connected to the Latte Panda with Arduino micro-controllers that handle the SPI serial bus.*

Some tests have been conducted on the ToF sensors to verify their compatibility with the application. The sensor characterisation has been performed in two phases, as follows:

1. *Linear Accuracy:* Based on the data sheet report, with bad lighting and reflectivity conditions, the sensor has an accuracy of 2 mm. The set up shown in Figure 4.17 was designed to evaluate the linear accuracy of

the sensor. The motor was used to drive 64 trajectories of 15 cm long, while the encoder was used as reference distance measure. The results show that the measure of 1.8 mm agrees with the sensor data sheet.

2. *Angular Accuracy:* The roll angle can be measured by evaluating the distance between the two counterweights (d_h). Where $\theta_{roll} = \text{atan2}(d_h, d_p)$, where $d_p = 54$ mm is the distance between the pulley centres. The accuracy of the roll angle is 3.8 degrees and has been evaluated in comparison with the roll angle collected with two accelerometer sensors over 80 trajectories. The reference roll angle is calculated as the angular difference between the gravitational acceleration measured by the accelerometer in the Genuino 101 board placed on the third link of the pentagonal mechanism and the IMU sensor placed on one of the anchors (Figure 4.18).

Figure 4.17: *The experimental set up used in evaluating the accuracy of the VL6180X ToF sensors. The encoder data were recorded using Quanser QPID DAQ set with a sampling rate of 50 Hz synchronised with the Genuino 101 via a digital input signal.*

Figure 4.18: *The experimental set up for the angular accuracy evaluation.*

Joint	Angle [degrees]
1	$[-90, 140]$
2	$[-10, -170]$
3	$[60, -140]$
4	$[-60, 140]$
5	$[10, 170]$
6	$[-140, 90]$

Table 4.2: *The range of motion are expressed in accordance with the reference frame in 4.2*

4.3 Interface Experimental Validation

Chapters 3 and Section 4.2 have presented a new framework bipedal equilibrium and a passive interface based on its core principles respectively. This section focuses on the experimental validations conducted for the two prototypes.

4.3.1 Validation of the First Prototype of the 5-DoFs Interface

The first prototype has been mounted on a carriage (Figure 4.19) to evaluate the 5-DoFs interface with the human natural walking strategy. The subjects are all female working in the Robotic Research Centre. The selection of the subjects was imposed by the current system limitation that does not allow the height regulation. Thus it can be used only with people about 1.60 m tall.

Figure 4.19: *The chain was mounted on a carriage manually pushed by one of the researchers.*

The 5 healthy subjects' data are as follows:

- *Height*: 1.602 ± 0.028 m
- *Weight*: 54 ± 2.55 kg
- *Age*: 27 ± 1.9 years old

The experimental protocol required each subject to walk for eight times with the system and eight times without the system for a distance of about 4 meters. The trajectory was recorded with 8 OptiTrack Prime-13 cameras and analysed using Motive Software Suite (NaturalPoint, Inc. DBA OptiTrack). The subjects were made to wear four passive markers on the following body parts:

- *Marker-1*: Above the Right Spina Iliaca Anterior Superior
- *Marker-2*: Above the Left Spina Iliaca Anterior Superior
- *Marker-3*: Right Heel
- *Marker-4*: Left Heel

Since the pelvic markers cannot be placed directly on the body while using the system, they have been placed on the harness belt instead. The gait velocity, step width, step length and stride lengths, which are required in analysing the walking performances were measured from the marker data. The measures have been executed with the following methodology:

- *Gait Velocity*: Pelvic marker time trajectory slope in the walking direction as in Figure 4.20.
- *Stride Length*: Distance between two subsequent plateaus in the walking direction time trajectory of the heel markers as in Figure 4.21.

Figure 4.20: *The velocity was calculated as the change in one of the pelvic marker position in one second.*

- *Step Length*: Distance between the intersection with the opposite foot trajectory and the precedent or subsequent landing position, which is also shown in Figure 4.21.
- *Step Width*: Heel markers distance along the perpendicular to the walking direction during the double support phase, as shown in Figure 4.22.

Figure 4.21: *The stride length is the distance between two subsequent stance positions of the same foot, while the step length is the distance between two feet in the double stance phase.*

Results and Discussion

The data show that although the subject slows down about 23% when using the chain (Figure 4.23), it seems connected to the presence of learning phase. In fact, the mean deceleration decreases to 18% when considering only the last two trajectories of every subject. The high transparency of the interface is also supported by the 2.8% reduction of the step frequency (ω_{Step}). As can be observed, the latter is inversely proportional to step frequency presented

Figure 4.22: *The step width is defined as the distance between the heels perpendicular to the walking direction.*

in Figure 4.24, which has been proved to increase when a perturbation is applied at the subject pelvis [76,155]. In contrast, the normalised step width is increased by 26% with the system and has an increased variability as in Figure 4.25. Such results may be an indication of an interference of the chain. It is, however, not probably connected with a pelvic fixation effect which induces the reduction of the step width [16,155].

This validates the initial hypothesis that introducing a compliant interface between a mobile base and the subject allows minimising the alteration of the gravitational field and, as a direct consequence, generate a safe environ-

Figure 4.23: *The subjects reduce their speed while walking with the system. However, they present a constant increase in the walking speed with the trials.*

Figure 4.24: *The step frequency has been shown in Chapter 3 to be kept constant during unperturbed walking independently from the speed. The experimental results report a 2.8% decrease in the step frequency that is within the margins of the human variability and, thus, the interface can be considered transparent [63].*

ment for impaired subjects to train their balance in unperturbed dynamical conditions.

Figure 4.25: *The normalised step width is increased 26% when the subject is attached to the system, but the data show a greater increase in its variability. This behaviour is not consistent with the effect of the pelvic fixation. Hence, further investigation is required to understand its causes in detail.*

4.3.2 Validation of the Second Prototype of the 5-DoFs Interface

The second prototype of the interface has been validated by mounting it on a mobile carriage with height regulation (Figure 4.26); therefore removing the constraints on the participants' height imposed by the previous version. Furthermore, the analysed walking trajectories have been increased from 8 to 12; the walked distance has been extended to about 6 meters; a more comprehensive data analysis has been conducted. Meanwhile, the camera system, the markers set-up and the gait parameters analysis have been left unaltered.

Twelve naive healthy subjects (7 males and 5 females) with no medical history of neural and cardiovascular system have been recruited for the experiment (IRB-2017-02-010). Their age is 27.3 ± 2.6 years old, their height is 171 ± 10.5 cm, and the weight is 65.3 ± 14.8 kg. The increased number of subjects and trials enabled the analysis based on the following methodology:

Figure 4.26: *On the left is shown a subject during the baseline collections. On the right shows the subject connected to a manual carriage via the proposed interface.*

- A baseline dataset has been prepared using 12 trajectories without wearing the device.
- The velocity, normalised step width and step frequency of the two datasets have been compared as in Section 4.3.1. Differing from the previous experiment, the variables distributions have been compared using Ansari-Bradley test with a 5% significance level.
- The Ansari-Bradley test has also been used for the intra-trials analysis of the subjects' walking strategies designed to identify an alteration of the variance in subsequent trials.
- Lastly, the Kruskal-Wallis test has been used to compare the walking strategies of the single subjects with and without the chain.

Results and Discussion

The results show a reduced impact on the gait parameters with respect to the results in the previous version. Figures 4.27-4.29 report 7.6% decrease in velocities, an increase of the step frequency of the 2.55%, and an increase in the normalised step width of the 15%. Furthermore, the Andari-Bradley test revealed that there is no significant modification of the data variance between the two datasets.

Figure 4.27: *The velocity comparison shows a reduction of the mean velocities of 7.6%, which is significantly better than the 23% measured with the first prototype. Furthermore, the variance analysis of the two datasets (Ansari-Bradley) does not report any significant difference between the two datasets, which suggests the absence of external factors interfering with the regulations of the velocities when the subjects are attached to the interface.*

The study of the evolution of the gait parameters between trial 1 and 12 shows that they oscillate around the mean value from the second trial onwards (Figure 4.30-4.32). The mean value and the inter-trial variance are significantly different for the first trial, suggesting that it may be related to the uncertainty of walking in an unknown novel environment.

Figure 4.28: The mean step frequency (ω_{Step}) is increased of 2.55%, which is similar in magnitude but opposite in sign of the results recorded during the previous experiment. The Ansari-Bradley test also revealed that there is no significant difference between the two distributions, supporting the hypothesis that it is rather connected to a regulation of the gait strategy from the humans rather than a significant perturbation of the movement dynamics.

Figure 4.29: The normalised step mean value increments of the 15% when the subjects are connected to the interface, which is the 11% lower of the perturbation introduced by the first prototype. Furthermore, the two distributions are characterised by the same variance in agreement that confirms the results obtained for the velocity and the step frequency.

Figure 4.30: *The mean velocities oscillate about the mean value except for the first trial, which is characterised with a mean velocity 16.3 % lower than the baseline, and the Ansari-Bradley test does not report any inter-trials variance change.*

The data considered for each subjects are reported in Figure 4.33. They show that 11 out of the 12 subjects significantly reduced their speed. Five of them also have a significant reduction of their step frequencies, two of which had a substantial increase in either the normalised step width or the step width.

The results prove that the interface does not produce alterations of the gait strategies that are consistent with a pelvic fixation [15, 16, 155]. Par-

Figure 4.31: *The step frequency (ω_{Step}) oscillates about the mean value in all the trials. Although there are two cases where the inter-trials distribution changes, the first trial coincides with a divergence of another gait determinants.*

ticularly, the results reported for overground walkers show that the presence of a pelvic fixation significantly reduces the gait velocity, step length, step width and frequencies [15, 16]. In contrast, the perturbations observed in the experiment are consistent with a lateral disturbance of the equilibrium, which induced a reduction of the step length, an increase in frequency and a widening of the step width [76, 124]. However, the data also show only one of the subjects experienced significant concurrent alterations of the three parameters, as shown in Figure 4.33, suggesting that the observed alteration

Figure 4.32: *The normalised step width oscillates about the mean value. There is no significant difference in variance between 11 of the trials. The first trial constitutes an exception where we observe both wider mean normalised step width and a variance different from the subsequent trial.*

may be due to the implementation of cautionary gait strategies rather than responses to external perturbations. This thesis also finds support in the analysis of the relationship between the step lengths and the velocities collected with both prototypes, which is reported in Figure 4.34. The figure shows how datasets have a similar 9-10% offset of the step length, which is congruent with the human variability reported in observed in data from other studies [63, 127], as shown in Figure 4.34.

4.4 Summary

The proposed design can provide the range of motion required by the system. The pentagonal chain operational space lies between the circles of radius 100 mm and 500 mm, which is solely limited by the range of motion of the links 1 and 5 as shown in Figure 4.3. The latest design (Figure 4.6) has the following range of motion as shown in Table 4.2.3. The 2-DoFs Parallel Pulleys Vertical Interface allows an 8 cm translation and a roll angle that ranges between ± 30 degrees. Lastly, the preliminary evaluation of the sensors confirms that they provide good accuracy in measuring the 5-DoFs Interface end effector pose. The accuracy can be further improved with a sensor fusion algorithm. The experiments with healthy subjects have validated that both prototypes have a limited effect on the subjects' gait strategies. Furthermore, these changes are not consistent with a pelvic fixation, but they seem closely related to the gait regulations observed in healthy subjects when a lateral perturbation is applied [76, 124].

In conclusion, although the available data do not allow to determine whether these alterations are due to a cautionary strategy or a response to a dynamics perturbation, they confirmed that the proposed mechanism solves the pelvic fixation issue observed in other systems. Moreover, the gap of performances between the two prototypes shows how the manoeuvrability of the mobile robotic platform will have a central role in the performance of final system.

Figure 4.33: The analysis of the gait parameters for the 12 subjects shows that 11 of them have significantly different velocities between their locomotion, with and without wearing the interface. However, the step frequency is significantly different only in 5 cases that are the main indicators of change in the strategy in the step length. Two of these subjects also presents a modification of either the step width or the normalised step width, implying a significant modification of the planning strategy.

Figure 4.34: *The motion capture data show that there is an offset of about 10% between the baseline and the data recorded wearing the two interfaces. However, the comparison with other datasets shows that this level of divergence can be related to a physiological regulation of the walking strategy. Lastly, it shall be noticed that a suspected outlier subject has been excluded from the regression of both the Old Baseline and Old Interface data reported in this Figure.*

Chapter 5

Conclusion and Future Work

5.1 Conclusion

The main goals for this work were to acquire of a better understanding of human locomotion by studying balance during and straight walking, to introduce a design of a transparent pelvic interface able to provide body weight support and to validate of the proposed solutions.

The formulation and validation of a novel analytical model for bipedal dynamics are presented in Chapter 3. The proposed model has successfully reproduced human data from multiple experiments, proving that humans move their legs as couple oscillators and take advantage of the dynamic environment generated by the legs. The investigation also revealed how the ankle strategies are used to control both the energy efficiency and the stability of locomotion by significantly altering the potential energy of the system. Furthermore, this dependence of human strategy upon the environment allows defining an analytical relationship between the system dynamics and body kinematics; thus, enabling the design of a bioinspired Task-Space planner of the CoM and the CoPs for straight locomotion and balance supervision.

Moreover, some of the findings can also be applied for the improvement of bipedal robots controllers.

The development of the two prototypes of the transparent pelvic interface presented in Chapter 4 shows that it is possible to build a device with the following characteristics: 5-DoFs to accommodate the CoM during locomotion, a work-space engineered to increase the manoeuvrability of the robotic walker in environments of daily living and guarantees harness-like safety. The main difference between the two prototypes is that the second prototype has a lighter modular structure with a slightly smaller workspace. Furthermore, the experiments with healthy subjects in Section 4.3 prove that both versions do not cause alteration consistent with a lateral perturbation of the gait rather than the pelvic fixation, which has been identified as the cause of alteration in other pelvic interfaces [16]. Furthermore, the statistical analysis of the variability of the gait parameters shows that there is a consistency of the dataset variability between the data collected with and without the interface. This behaviour suggests that the alteration is most likely connected to anticipatory adjustment of the strategy rather than a response to external perturbation, which is also supported by the findings of Hak *et al.* that show how human modify their locomotor strategy when they are expecting a perturbation [76]. Therefore, the data confirmed the hypothesis that the proposed mechanism does not disrupt the movement dynamics and can be considered transparent.

To summarise, this work has made the following fundamental contributions:

- i The bipedal model presented and validated in Chapter 3 provides a deeper understanding of the bipedal locomotion and the methods to track the patient stability from the kinematics.

- ii Chapter 4 shows how the knowledge of the system dynamics can be deployed into the design process of the proposed interface.
- iii The experiments in Section 4.3 validate that the proposed design can minimise the perturbation to the locomotion.

5.2 Future Work

The outcomes of this work proved the feasibility of the Mobile Robotic Assistive Balance Trainer. However, there are still multiple components to be implemented before the system can be deployed into a clinical environment. The process can be divided in the following points:

1. Extending of the models to include more complex movements required for the development of an architecture as shown in Figure 5.1, where selecting an adequate control policy requires an accurate balance evaluation in the task of daily living, which are not limited to standing and straight walking.
2. Develop the automated wheelchair to be used with the proposed interface. The device should be agile enough to follow human movement without interfering with their locomotor strategies and should provide both harness and sit-to-stand support. Furthermore, the wheelchair should be sensorised to collect all the information required by the controller to estimate the balance from the kinematics parameters.
3. Once the robot is completed, perform a study of the system performances in assisting frail elderlies in activities of daily living should be done before testing the system ability to support gait and balance rehabilitation therapies.

Figure 5.1: *The control of the Mobile Robotic Assistive Balance Trainer will require measuring the feet kinematics to integrate with the pelvic information collected via the chain to estimate the user equilibrium and motor intention. The diagram above provides the information required by selecting the system control policy. Currently, three different policies are being considered. The first is to follow with minimisation of the interference. The second is configuring the system for reacting to a balance loss. The last is an active support of the patient balance. The first two control policies rely on a passive interface used as a sensor for measuring the system/user interaction. In contrast, the chain acts as selective compliant interface modulating the interaction between the mobile base and the user and requires the development of an actuation system.*

Lastly, the results obtained in Chapter 3 allow to hypothesise a new control architecture for the human motor control. Therefore, a deeper investigation of the proposed theories should be considered independently from

the development of the Mobile Robotic Assistive Balance Trainer.

Appendix A

CoM and CoP Derivation

The CoM position and orientation on the transverse plane were calculated by relying on the four pelvic markers —two placed on the Anterior Superior Iliac Spine and two placed on the Posterior Superior Iliac Spine. The position is derived as follows:

$$\left\{ \begin{array}{l} x_{CoM}(k) = \frac{x_{LASIS}(k)+x_{RASIS}(k)+x_{LPSIS}(k)+x_{RPSIS}(k)}{4} \\ y_{CoM}(k) = \frac{y_{LASIS}(k)+y_{RASIS}(k)+y_{LPSIS}(k)+y_{RPSIS}(k)}{4} \\ \Delta x_{CoM}(k) = \frac{x_{LASIS}(k)+x_{RASIS}(k)}{2} - \frac{x_{LPSIS}(k)+x_{RPSIS}(k)}{2} \\ \Delta y_{CoM}(k) = \frac{y_{LASIS}(k)+y_{RASIS}(k)}{2} - \frac{y_{LPSIS}(k)+y_{RPSIS}(k)}{2} \\ \theta_{CoMxy}(k) = atan2(\frac{\Delta y_{CoM}(k)}{\Delta x_{CoM}(k)}) \end{array} \right. \quad (A.1)$$

The foregoing equation allows to compute the CoPs using the Heel marker, the 1st and 5th Metatarsal Bones.

$$\left\{ \begin{array}{l} x_{Foot}(k) = \frac{2x_{Heel}(k) + x_{MT1}(k) + x_{MT5}(k)}{4} \\ y_{Foot}(k) = \frac{2y_{Heel}(k) + y_{MT1}(k) + y_{MT5}(k)}{4} \\ z_{Foot}(k) = \frac{2z_{Heel}(k) + z_{MT1}(k) + z_{MT5}(k)}{4} - z_{CoPg} \\ \\ Where : \\ \\ z_{CoPg_{ground}} = \frac{2z_{Heel_{ground}} + z_{MT1_{ground}} + z_{MT5_{ground}}}{4} \end{array} \right. \quad (A.2)$$

While the foot orientations are derived as follows:

$$\left\{ \begin{array}{l} \Delta x_{Foot}(k) = \frac{x_{MT1}(k) + x_{MT5}(k)}{2} - x_{Heel}(k) \\ \Delta y_{Foot}(k) = \frac{y_{MT1}(k) + y_{MT5}(k)}{2} - y_{Heel}(k) \\ \theta_{Footxy} = atan2\left(\frac{\Delta y_{Foot}(k)}{\Delta x_{Foot}(k)}\right) \\ \\ D_{xy}(k) = \sqrt{((\Delta x_{Foot}(k))^2 + (\Delta y_{Foot}(k))^2)} \\ \\ \Delta z_{Foot}(k) = \frac{z_{MT1}(k) + z_{MT5}(k)}{2} - z_{Heel}(k) \\ \\ \theta_z(k) = atan2\left(\frac{\Delta z_{Foot}(k)}{D_{xy}(k)}\right) \end{array} \right. \quad (A.3)$$

Appendix B

Toe-Off and Heel-Strike Parameters

1. $v_{CoM_p} = 0.7$:

- **05 degrees:** $t_{TO} = 0.3$ s, $m_{TO} = 1.5$ s, $m_{HS} = 0.175$ s and $t_{HS_{delay}} = 1.7m_{HS}$.
- **10 degrees:** $t_{TO} = 0.3$ s, $m_{TO} = 2$ s, $m_{HS} = 0.175$ s and $t_{HS_{delay}} = 1.7m_{HS}$.
- **15 degrees:** $t_{TO} = 0.3$ s, $m_{TO} = 0.6$ s, $m_{HS} = 0.178$ s and $t_{HS_{delay}} = 1.7m_{HS}$.

2. $v_{CoM_p} = 1.0$:

- **05 degrees:** $t_{TO} = 0.48$ s, $m_{TO} = 2$ s, $m_{HS} = 0.145$ s and $t_{HS_{delay}} = 2m_{HS}$.
- **10 degrees:** $t_{TO} = 0.48$ s, $m_{TO} = 1.55$ s, $m_{HS} = 0.15$ s and $t_{HS_{delay}} = 2m_{HS}$.

- **15 degrees:** $t_{TO} = 0.48$ s, $m_{TO} = 0.48$ s, $m_{HS} = 0.153$ s and $t_{HS_{delay}} = 2m_{HS}$.

3. $v_{CoM_p} = 1.2$:

- **05 degrees:** $t_{TO} = 0.38$ s, $m_{TO} = 2$ s, $m_{HS} = 0.145$ s and $t_{HS_{delay}} = 2m_{HS}$.
- **10 degrees:** $t_{TO} = 0.38$ s, $m_{TO} = 1$ s, $m_{HS} = 0.15$ s and $t_{HS_{delay}} = 2m_{HS}$.
- **15 degrees:** $t_{TO} = 0.38$ s, $m_{TO} = 0.35$ s, $m_{HS} = 0.13$ s and $t_{HS_{delay}} = 2m_{HS}$.

4. $v_{CoM_p} = 1.6$:

- **05 degrees:** $t_{TO} = 0.40$ s, $m_{TO} = 2$ s, $m_{HS} = 0.145$ s and $t_{HS_{delay}} = 2m_{HS}$.
- **10 degrees:** $t_{TO} = 0.40$ s, $m_{TO} = 1$ s, $m_{HS} = 0.15$ s and $t_{HS_{delay}} = 2m_{HS}$.
- **15 degrees:** $t_{TO} = 0.40$ s, $m_{TO} = 0.48$ s, $m_{HS} = 0.153$ s and $t_{HS_{delay}} = 2m_{HS}$.

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