Biomechanical Analysis of Slip-like Perturbations for Target Specifications Definition Towards a Fall Preventing Wearable Device

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Universidade do Minho
Escola de Engenharia

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STATEMENT OF INTEGRITY

I hereby declare having conducted this academic work with integrity. I confirm that I have not used plagiarism or any form of undue use of information or falsification of results along the process leading to its elaboration.

I further declare that I have fully acknowledged the Code of Ethical Conduct of the University of Minho.

Resumo

De acordo com a OMS, um total de 684 000 quedas fatais ocorrem todos os anos, fazendo deste evento a segunda causa mais frequente de morte, por lesão. De entre as possíveis causas de uma queda, as quedas por escorregamento evidenciam-se. De forma a limitar a sua incidência, surge a necessidade do desenvolvimento de soluções tecnológicas orientadas especificamente para esta problemática. Estas devem ser capazes de prever, detetar e responder a uma situação iminente de queda.

A presente dissertação procurou identificar as especificações alvo para o desenvolvimento de um dispositivo robótico *wearable* através da análise de dados experimentais, onde os indivíduos foram induzidos a escorregar, em laboratório. A análise dos dados recolhidos permitiu uma compreensão da globalidade da resposta biomecânica humana da perna perturbada bem como da perna não perturbada. Neste estudo foram consideradas diversas variáveis externas nomeadamente a velocidade da marcha, o pé (esquerdo ou direito) que foi perturbado, a inclinação da superfície e a intensidade da perturbação. Variáveis cinemáticas, cinéticas, espaço temporais e EMG foram analisadas de forma a proceder-se a uma investigação completa da resposta biomecânica. A partir desta análise foram definidos torques, RoM, rpm e tempos de atuação e deteção para o dispositivo a desenvolver.

A resposta biomecânica da anca, joelho e tornozelos da perna perturbada foi caracterizada, respetivamente, por momentos extensores, fletores e de flexão plantar. Por sua vez, a resposta biomecânica da perna não perturbada foi caracterizada por flexão da anca, extensão do joelho e flexão plantar do tornozelo. Estas respostas foram influenciadas pela velocidade da marcha, inclinação da superfície e intensidade da perturbação. As especificações alvo determinadas relativas ao RoM foram de 85.90°, 106.34°, e 95.23°, respetivamente, para a anca, joelho e tornozelo. Relativamente aos rpm, foram definidas as gamas de [17.85 - 51.10] para a anca, [21.73 - 63.80] para o joelho e [17.52 - 57.14] para o tornozelo. Por fim, os valores de torque definidos foram de [-3.05 a 3.22], [-1.70 a 2.34] e [-2.21 a 0.90] em Nm/kg, respetivamente para as articulações da anca, joelho e tornozelo.

Palavras-chave: Prevenção de quedas, Resposta Biomecânica, Dispositivos robóticos *wearable*, Perturbações por escorregamento, Especificações alvo

ABSTRACT

According to WHO, 684 000 fatal falls occur every year, making them the second leading cause of death by injury. Slip is the most common cause of a fall among the many possible reasons. As a result, because it is critical to limit the occurrence of these incidents, there is a need to focus on the development of purpose-oriented technological solutions capable of predicting, detecting, and responding to an imminent fall situation.

The current dissertation sought to identify the target specifications for the development of a wearable robotic system for slip-like fall prevention based on the analysis of experimental data in which individuals were asked to handle induced slip-like perturbations. The statistical analysis of experimental data allowed a full understanding of the natural human biomechanical response to slip perturbations of both perturbed and unperturbed legs. The research considered a wide variety of external characteristics such as gait speed, perturbed foot, surface inclination, and perturbation intensity. Kinematic, kinetic, spatiotemporal, and EMG data were analysed to complete the investigation of this natural reaction. Based on the experimental data analysis, torques, RoM, rpm, detection and actuation times were established.

The biomechanical response to slip-like perturbations by the hips, knees, and ankles' of the perturbed limb were defined by extension, flexion, and plantarflexion moments, respectively. Biomechanical response in the unperturbed limb was characterised by hip flexion, knee extension, and ankle plantarflexion. These responses were also affected by gait speed, surface inclination, and perturbation intensity. The RoM goal parameters for hips, knees, and ankle joints were 85.19°, 106.34°, and 95.23°, respectively. For the hip, knee, and ankle joints, rpm values of [17.85 - 51.10], [21.73 - 63.80], and [17.52 - 57.14] were found, respectively. Finally, for the hip, knee, and ankle joints, flexion/extension torque values of [-3.05 to 3.22], [-1.70 to 2.34], and [-2.21 to 0.90] in Nm/kg were estimated.

Keywords: Fall prevention, Biomechanical response, Wearable robotic device, Slip-like perturbations, Target specifications

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List of Acronyms and Abbreviations

AP	Antero-Posterior
APO	Active Pelvis Orthosis
ANOVA	Analysis of Variance
BF	Biceps Femoris
BIRDLAB	Biomedical Robotic Devices Laboratory
BoS	Base of Support
CMEMS	Centre for Micro Electro-Mechanical Systems
CNS	Central Nervous System
DV	Dependent Variable
CoF	Coefficient of Friction
CoM	Centre of Mass
CoP	Centre of Pressure
CoF	Coefficient of Friction
CusToM	Customizable Toolbox for Musculoskeletal simulation
ECG	Electrocardiography
EEG	Electroencephalography
EMG	Electromyography
FSM	Feature Selection Methods
GC	Gait Cycle
GL	Gastrocnemius Lateralis
GRF	Ground Reaction Forces
HC	Heel Contact
HS	Heel-Strike
IMU	Inertial Measurement Unit
IV	Independent Variable
KPI	Key Performance Indicators

LOB	Loss of Balance
MG	Medial Gastrocnemius
MH	Medial Hamstrings
ML	Medio-Lateral
MoS	Margin of Stability
MVC	Maximum Voluntary Contraction
PC	Principal Component
PCL	Planar Covariation Law
PLA	Polylactide acid
RF	Rectus Femoris
RoL	Rate of Loading
RoM	Range of Motion
RPM	Revolutions per Minute
RQ	Research Questions
SAM	Segment Angular Momentum
SDI	Slip Distance I
SDII	Slip Distance II
SEA	Series Elastic Actuator
TA	Tibialis Anterior
V	Vertical
VL	Vastus Lateralis
VI	Vastus Intermedius
WBAM	Whole Body Angular Momentum
WHO	World Health Organization

1 Introduction

This dissertation presents the work developed during the 5th year of the Integrated master's in Biomedical Engineering, more specifically in the branch of Biomaterials, Rehabilitation and Biomechanics during the academic year of 2021/2022.

The investigation work addressed in this dissertation was performed in the Biomedical Robotic Devices Laboratory (BiRDLab) of the Centre for Microelectromechanical Systems (CMEMS), at the University of Minho, in Portugal. The study of the human biomechanical response to slip-like perturbations is the main goal of the developed biomedical research. From experimental data, the research work conducted in this dissertation, seeks to address of the preliminary stages of the design of a wearable device for slip-like falls prevention through the determination of the target specifications.

1.1 Motivation

Society's living conditions' improvement is certainly a constant challenge for which science and technology must join synergies. This challenge takes even greater importance if we consider the recent World Health Organization (WHO) forecasts for the current decade (2020-2030). According to the same organization, the number of people in the world aged 60 or over will be 34% higher than in 2019, with a predicted increase from 1 to 1.4 billion people. Considering the future year of 2050, this value will increase, being predicted the existence of 2.1 billion people aged 60 or over. The same organization adds that 2020 was the first year in the history of humanity in which the number of people aged 60 or over exceeded the number of children aged 5 or under, and this conclusion tends to be accentuated in upcoming years [1].

Population ageing in result of the increase in average life expectancy naturally increases the incidence of distinct chronic diseases that result in mobility impairments. In addition to several economic and social barriers, the emergence of the diseases associated with an ageing world population leads to another public health problem worldwide highlighted by the WHO: falls occurrence [2].

According to the WHO, an estimated 684 000 reported fatal falls occur every year, being these events the second largest cause of death by injury. In addition to a public health problem, the increasing number of people who suffer falls each year also results in an economic problem for health services worldwide [2]. For example, a comprehensive study performed in Netherlands estimated an average medical and social care cost of 9370 € per fall. If this figure is applied to the whole EU, at least 25 billion euros are estimated to be spent, every year, treating fall related injuries [3]. In the less severe falls, 9 days of hospital stay are reported while in the most serious cases this value can exceed the 25 days in intensive care units [4]. In addition, people who have experienced a fall, even one that does not cause an injury, develop fear of falling, which is a serious issue that negatively impacts their mental and physical health. Because of this apprehension, a person's everyday activities are reduced, which raises the possibility of another fall due to muscular weakening [5].

Slipping is the primary contributor to falls among the wide range of factors that might cause them [6]. Slip-related falls often cause more severe consequences in addition to their frequency. In fact, they are typically held accountable for a large percentage of hip fractures in elderly people [7].

For this reason, researchers have been examining slips' biomechanics to better understand human response to these events and mitigate their negative consequences by developing fall prevention strategies [8]–[19]. Slips occur when there is an insufficient Coefficient of Friction (CoF) between the subject's foot and the floor, and environmental conditions can greatly enhance the risk of slipping [20]. These events occur mostly when the foot contacts or leaves the floor, and they resemble crucial body weight transfer circumstances between the lower limbs, particularly when the heel strikes the floor [20]. Slips started at the Heel-Strike (HS) cause a backward Loss of Balance (LOB) by deviating the subject's Centre of Mass (CoM) relative to the Base of Support (BoS). When human sensory systems identify this deviation, information is conveyed to the motor control areas of the Central Nervous System (CNS) through afferent nerves. The CNS interprets the information and produces efferent signals to specific skeletal muscles to compensate for the LoB by correctly contracting to keep the body position inside the BoS. The coactivity of recruited lower limb muscles counteracts foot displacement and facilitates slip recovery [20].

For all the reasons listed here, it becomes vital to seek for fresh approaches to the issue of the falls incidence. In addition to enabling a knowledge of how people naturally react to such situations, the research of falls' biomechanical response is crucial to the project of wearable technology that may mimic this reaction, to be developed in future. Thus, this devices' development, is expected to reduce the impact and frequency of falls, their negative effects in elderly subjects and their economic and social costs.

1.2 Problem statement and scope

Considering the high incidence and catastrophic increasing consequences of falls, due to the population ageing, highlighted by the WHO, the development of real-time wearable assistive technologies appears to be a solution for fall prevention [1], [21], [22]. Additionally, the feeling of safety conveyed to the user by these devices allows reducing the fear of falling again, thus avoiding the reduction of the person's daily activities. In this regard, the reduced need of post-fall rehabilitation can also be interpreted as a benefit in terms of costs and human resources reduction.

The study of the biomechanical response to slip events is a fundamental preliminary step in the development of these devices. This biomechanics analysis is crucial for understanding the function and importance of each muscle and joint engaged in this biomechanical process, as well as for defining the device specifications. As a result, a vast set of crucial knowledge may be gathered for the design and development of these devices. As a first step, data has to be acquired considering well delineated protocols with varying crucial conditions, that will enable the biomechanical analysis and interpretation.

Regarding the biomechanical response to slip perturbations, the literature usually studies mainly hip, knee and ankle's joint sagittal kinematic movements to prevent a fall after such perturbation [23], [24]. Other research focuses on changes in spatiotemporal variables from the beginning of the slip through the completion of the recovery process [25], [26]. Variables obtained from electromyographic (EMG) data such as muscular synergies, latency periods and muscular peak magnitudes are the most usually analysed [27]–[29]. This type of information allows to understand the muscles involved in slip recoveries thus to deeply understand the lower limb's characteristic movements during slip recoveries. The influence of different independent variables (IV's) such as intensity and gait speed are also subject of study in some literature allowing to induce variability in slip and thus in the respective biomechanical response [25], [30]. Despite the existence of different studies in this field there is still a need for an analysis that includes and complements all these types of data by also simulating other environment conditions, namely surface inclination [6].

In turn, literature presents some developed devices capable of preventing falls by acting on the hip or knee joints [22], [31], [32]. The primary purpose of these devices is to detect a LoB situation and produce an actuation capable to restore a suitable biomechanical position to allow the subject to resume a steady walking gait following a slip-like perturbation. Despite promising initial results, there is still a lack in clearly and quantitatively identifying which of the lower limb joint may produce a more effective response.

Thus, an in-depth analysis of the biomechanical response to slip perturbations considering different walking conditions was performed with the objective to understand the role of each lower limb joint and muscle involved in this response and to quantitatively identify the joint where and when robotic actuation strategy can be applied in a more effective way. The biomechanical analysis previously mentioned also allowed to define the target specifications of a fall preventing wearable robotic device.

1.3 Goals and research questions

The main goal of this dissertation is to develop a comprehensive study of the human biomechanical response to slip-induced perturbations and to conclude about the role and importance of lower limb muscles and joints in response to slip-like perturbations in different conditions. Additionally, this analysis intends to help defining the target specifications for the design of a fall preventing wearable robotic device to be developed in the future. Finally, this analysis will also help to clarify quantitatively which lower limb joints actively intervene in the recovery process after a slip situation.

Thereby, within the scope of this dissertation, it is necessary to achieve the following goals:

- **Goal 1:** To gather knowledge about the human biomechanical response to slip perturbations considering multivariate data (kinetic, kinematic, spatiotemporal and EMG data. Understanding the most effective and significant biomechanical techniques that people naturally employ when confronted with a slip-like perturbation is made possible by this study that looks into the scientific literature. Also, it will be fundamental for the analysis of experimental data about the same topic. This topic will be addressed in Chapter 3. Key Performance Indicators (KPIs): i) biomechanical strategies evidenced in the analysed literature; ii) events' chain biomechanical responses to slips; iii) torques and joint angles involved in this response; iv) EMG variables (namely latency, peak amplitude, and time to peak) evidenced in literature; v) review article publication.

- **Goal 2:** To gather information about the wearable robotic devices developed for slip-like falls prevention. This objective will provide a general overview of the mechanical design and actuation strategy of these devices, allowing an understanding of their main components and respective function. These devices requirements' analysis will be also useful to understand the requirements to be considered during these devices' project and design. This topic will be addressed in Chapter 2. KPIs: i) main specifications, functionalities, and mechanical components of the already developed wearable robotic devices for fall prevention against slips.

- **Goal 3:** To analyse slip-induced perturbations considering experimental kinetic, kinematic, spatiotemporal and EMG data. This study allows to understand the role of each lower limb joint and muscle during a slip recovery. In this analysis dependent variables (DV's) as gait speed, surface inclination, perturbed foot and perturbation intensity will be also addressed to understand their influence on slip recoveries. This analysis will allow the definition of target specifications to consider during the development of a wearable robotic device capable of mimicking the studied response. This topic will be addressed in Chapter 5. KPIs: i) p-value < 0.05 to detect statistically significant differences between normal gait, perturbation, and biomechanical response situations, per DV, using kinetic, kinematic, spatiotemporal and EMG data.

- **Goal 4:** To rank lower limb's joint according to their importance during slip recovery. This goal allows to gather a set of information to facilitate a quantitative based decision about the joint or joints to produce actuation. This topic will be addressed in Chapter 5. KPIs: i) discriminative quantitative ranking of the importance of each DV in the biomechanical response to a slip-like perturbation.

- **Goal 5:** To define the target specifications to consider when developing a wearable robotic device to prevent slip related falls. In future, this goal will allow to select the mechanical components capable of satisfying the device requirements allowing to mimic the kinetic, kinematic, and spatiotemporal variables involved in the human recovery process after slip perturbations. This topic will be addressed in Chapter 6. KPIs: i) torque, Range of Motion (RoM), detection and actuation times, revolutions per minute (rpm), weight, and other general requirements to considered in the development a wearable robotic device to prevent slip-related falls.

The following Research Questions (RQ) are expected to be answered in the present work:

- **RQ1:** What are the biomechanical responses to slip-induced perturbations highlighted in the scientific literature? The answer in included in Chapter 3.

- **RQ2:** What are the biomechanical strategies to avoid falls, after slip perturbations, obtained from experimental data analysis? The answer in included in Chapter 5.

- **RQ3:** What are the target specifications in the project of slip-related falls prevention wearable robotic devices? The answer in included in Chapter 6.

1.4 Contribution to knowledge

The main contributions to knowledge of this dissertation are:

- A systematic literature review of biomechanical strategies to avoid falls during a slip event considering kinetic, kinematic, spatiotemporal and electromyographic data.

- Labelling improvement of a previously existent dataset with extensive and relevant kinematic and biosignal information collected during normal and perturbed treadmill walking, that allows to study the changes to the human motion induced by slip-like perturbations.

- Lower limb's quantitatively ranked joints by its influence in slip recovery process.

- Target specifications to the development of wearable robotic fall prevention devices depending on the joint where actuation is to be produced.

1.5 Publications

The literature review work performed in Chapter 3, enable the submission of a review article entitled:

- **J. C. Nunes**, N.F. Ribeiro, O. N. Carvalho, C. P. Santos, "Biomechanical strategies to avoid falls during a slip event considering multivariate data - A Review ", Experimental Gerontology. [under review].

Moreover, the work performed in the Chapter 5 will allow the publication of the following article:

- **J. C. Nunes**, N.F. Ribeiro, O. N. Carvalho, C. P. Santos, "Biomechanical Analysis of Slip-like Perturbations for Target Specifications Definition Towards a Fall Preventing Wearable Device ", Experimental Gerontology. [to be submitted].

1.6 Thesis outline

This dissertation is organized as follows. **Chapter 2** presents a biomechanical analysis of the human gait where the main parameters, phases and associated events will be highlighted, among other aspects. The joints and muscles that most contribute to human locomotion as well as their biomechanical role will also be addressed. Also in this chapter, the current existing wearable robotic devices for fall prevention will be presented as well as their mechanical components and specifications.

In **Chapter 3** is included a state-of-the-art review about the biomechanical strategies to avoid falls after a slip event. This analysis will include multivariate data namely kinematic, kinetic, spatiotemporal and EMG data.

Chapter 4 introduces the project conceptual design of the dissertation. In this Chapter, the dissertation's main phases will be presented and discussed considering the respective contribution to achieve the dissertation goals.

Chapter 5 presents the experimental protocols previously developed to collect slip-like perturbations data. In this chapter, will be also presented the steps related to the data pre-processing, the statistical analysis performed, and the ranking procedures used to rank the DV's. The main results of the statistical analysis performed will be presented and discussed.

In turn **Chapter 6** addresses the methodology used to obtain the torques involved in the trials developed at Birdlab*.* Based on the statistical analysis performed in Chapter 5 the target specifications to the project of wearable devices to slip-related falls prevention will be also defined.

Finally, **Chapter 7** will present the main conclusions of the work developed as well as the answers to the RQs pointed out in this chapter. Finally, future work to be developed to complement this dissertation will be also addressed.

2 Human Gait and Fall Preventing Robotic Devices

In this chapter two different topics will be addressed. Firstly, in section 2.1 a biomechanical analysis of healthy human gait data will be carried out, detailing its different phases and highlighting some of the most important events and parameters as well as the main muscles and joints involved. In turn, in section 2.2, the literature related to the development of wearable robotic devices for slip-related falls prevention will be presented. In this topic, will be carried out a general overview of the devices present in scientific literature. The actuation mode and the main technical specifications will also be discussed.

2.1 Human Gait

This subsection aims to address in a generic way the main anatomical and biomechanical concepts related to the human gait. Generally, this method of locomotion can be divided in different phases and events which will be presented. Kinetic, kinematic, and spatiotemporal parameters characteristics of the human gait will also be explained. Finally, regarding to anatomy, the muscles and joints that actively intervene in human locomotion will be briefly addressed.

2.1.1 General Overview

Walking can be generically defined as ''a method of locomotion involving the use of the two legs, alternately, to provide both support and propulsion'' [33]. During a person's lifetime, this is unquestionably essential and crucial. Although it is a typical activity, walking is the consequence of a complex series of events that include the neurological system, several elements of the musculoskeletal system, and the cardiorespiratory system [34]. Additionally, it is naturally influenced by several factors, among which we can highlight socio-cultural factors, age, personality, diseases and even mood [34].

Running, skipping, ascending and descending stairs are just a few of the gait variations that allow humans to adapt to a wide range of daily conditions [35] although sharing certain similar characteristics.

When muscles are activated, they develop tension, which results in forces and moments in the joints responsible for human locomotion. These forces and moments cause movement in the rigid segments of the human skeleton, causing their movement and the exertion of forces on the external environment [35]. In a generic way, the set of events that allow human locomotion can be presented in the following summary form also illustrated in Figure 2.1 [35].

- Registration and Activation of the gait command in the CNS
- Transmission of the gait signals to the Peripherical Nervous System
- Contraction of the muscles and tension development
- Generation of forces at, and moments across, synovial joints
- Regulation of the joint forces and moments by the rigid skeletal segments based on their anthropometry
- Movement of the segments in a manner that is recognized as functional gait
- Generation of ground reaction forces.

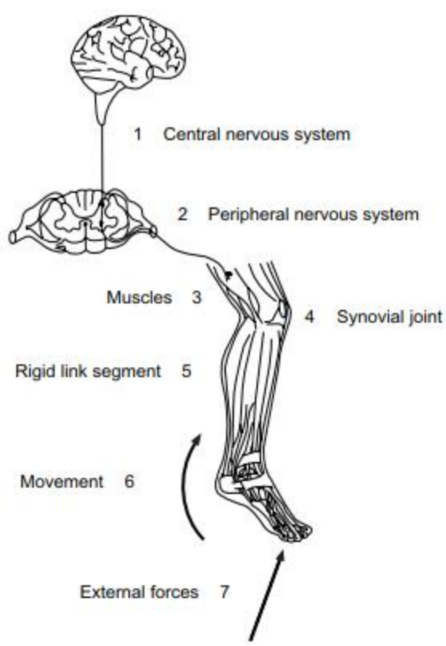


Figure 2.1 Sequence of events that generate human gait. Taken from [35].

As a complex event involving several systems of the human organism, human gait analysis is a biomechanical research area that could be interesting for several purposes. Firstly, and strictly related with the objective of the present dissertation, the analysis of the human gait is, in many cases, a fundamental preliminary step in the development of assistive robotic devices. Another example is related to the relationship between some pathologies and gait disturbances. In this case, the biomechanical study of the gait is a tool to discover or corroborate the existence or absence of neurological, cardiorespiratory, and other pathologies [34].

2.1.2 Human Gait Biomechanics

As a preliminary point for the biomechanical analysis of the human gait that will be carried out during this dissertation, it is important to make some previous considerations related to the human movement. Firstly, the anatomical reference position, shown in Figure 2.2, is the reference posture used to describe the relative position and movement between the anatomical segments of the human body. This corresponds to a posture in which the human body appears straight with the feet slightly apart and the arms suspended laterally with the hand palms directed forward [36].

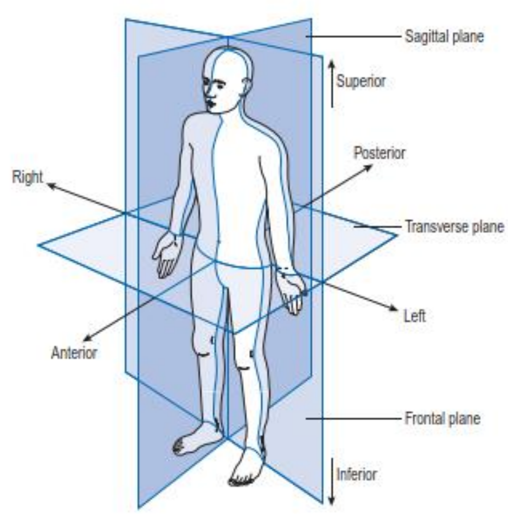


Figure 2.2 Anatomical reference position, with the 3 reference plans and 6 fundamental directions. Taken from [35].

The movement of the upper and/or lower limbs in relation to the anatomical reference position previously discussed as well as the specific terminological definition of each type of movement are described through 3 planes: the so-called anatomical reference planes. These are orthogonal and each plane divides the human body in two halves of equal mass, being the common point of intersection the CoM of the body. There are three planes: sagittal, frontal, and transverse. In addition, these planes are also important to define 6 fundamental directions used to define spatial locations related to them. The sagittal plane divides the body vertically into its left and right halves. In turn, the transverse plane divides the body into its upper and lower halves, and finally the frontal plane divides the body into its posterior and anterior halves. Therefore, we can define inferior and superior directions related with the transverse plane, anterior and posterior related with the frontal plane and finally right and left directions related with the sagittal plane as shown in Figure 2.2 [36].

Specifically, in the case of human gait, sagittal plane is the most important one since mostly movements occur in this plane. Nevertheless, it is important to consider the other planes because many pathologies or perturbations are expressed, in terms of gait parameters alterations, in the transversal or frontal planes [37].

Finally, during human gait, it is possible to consider the existence of two subdivisions of the human body: passenger, and locomotor, considering their functionality. Despite the existence of movement in both, the intensity with which it occurs is markedly different when comparing both subdivisions [37]. Regarding the locomotor subdivision, in this subdivision we can find the body segments responsible for the locomotor function, among which are the lower limbs' articulations, the pelvis and a set of 57 muscles that intervene in human locomotion. Also, the bone segments are part of this group as levers. Although the main function of this subdivision is human locomotion, we can highlight the existence of 4 essential sub-functions for it: propulsive force and forward motion, upright stability maintenance, minimization of the shock of floor impact and finally, energy conservation to achieve a reduced muscular effort [37].

In turn, the passenger subdivision includes head, neck and arms and is so named since these segments do not contribute directly to the act of walking. In turn, their main function is to ensure a neutral vertebral alignment to minimize posture changes during human gait [37].

2.1.3 Human Gait Phases and Events

Human gait is segmented into phases. Among them we can highlight important events associated to each one. This segmentation into phases and events allows a more detailed and direct gait analysis since each phase is related to specific action of specific joints and muscles. In this way, by specifically analyzing a certain phase, it is possible to correlate it with the action of certain joints and muscles [34].

Gait cycle (GC) is one of the most important parameters when studying human gait. It can be defined as the time interval between two successive occurrences of one of the repetitive events that occur during human gait. The act of walking concerns a repetitive sequence of events, and the GC is therefore the elementary unit of the entire act of walking [33][35]. Each GC can be divided into two main phases: stance and swing phases. Stance phase concerns the period during which the foot is in contact with the ground. This phase is dominant in a walking cycle and takes up 60% of the total GC time. On the other hand, the swing phase, which occupies the remaining 40% of a GC, is related to the time during which the foot is suspended, thus allowing the body to move forward [35], [37]. Stance and swing phases representation are shown in Figure 2.3.

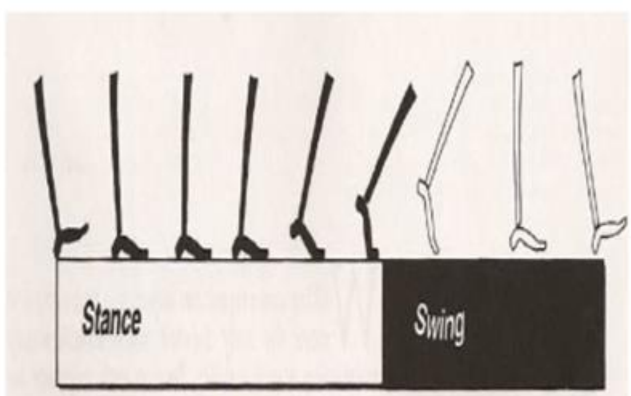


Figure 2.3 Stance and swing phases representation. Taken from [35].

Depending on their behaviour in a certain phase of the GC, both lower limbs can have different designations. Trailing leg refers to the leg which is in swing phase during a given phase of the GC. On the other hand, leading leg is the leg that contacts with the ground [35].

Apart from the general division of a GC into the two large phases mentioned above, it can also be divided into subphases that will be presented in this chapter. Each one of these subphases contributes to the accomplishment of 3 tasks that are essential during the human gait: limb advancement, weight acceptance and single limb support are also important [33]. Below, will be briefly presented the different walking tasks as well as the phases and sub-phases that contribute to accomplish them:

I) **Weight Acceptance:** This is the most demanding task in GC as the body weight is transferred onto a limb that has recently finished the swing phase and therefore has an unstable alignment. In this task 3 different functional patterns are needed: shock absorption, initial limb stability and the preservation of progression [33].

- **Heel contact - HC (0%):** In this phase the hip is flexed, the knee extended, and the ankle dorsiflexed. The heel of the leading leg touches the ground while trailing leg is at terminal stance [33].

- **Loading Response (0%-10%):** This is the initial double stance period, and it begins with the initial floor contact and continues until the other foot is lifted for swing. The body weight is transferred to the leading leg and the knee is flexed to absorb shock while the ankle is plantarflexed. The leading leg is in its pre-swing phase [33].

II) **Single Limb Support:** When the trailing leg is in its swing phase all the body weight is supported by the leading leg both in sagittal and frontal planes. Two phases can be distinguished to accomplish this task [36].

- **Mid Stance (10%-30%):** Mid Stance is the first half of the single limb support. It starts as the trailing leg's foot is lifted and continues until the body weight is aligned over the forefoot. The leading leg advances over the stationary foot by ankle dorsiflexion while the knee and hip are extended [33], [35].

- **Terminal Stance (30%-50%):** This phase completes single limb support, and it starts with the heel lift continuing until the other foot strikes the floor. The body weight still be supported by the forefoot. In this phase occurs an increasing of the extension of the knee and hip to put the limb in a favourable trailing position [33], [35].

III) Limb advancement: To proceed to the limb advance, preparatory posturing begins in stance phase. The limb swings through three postures as it lifts itself, advances and prepares for the next stance period. Four gait phases are involved in this task [33].

- **Pre-Swing (50%-60%):** This subphase begins with the initial touch of the opposite limb in the ground ending with the ipsilateral toe-off. As the leg is preparing itself to start the swing phase an increased plantarflexion, a greater knee extension and a reduced hip extension are typical in this subphase while the opposite limb is in loading response [33], [35].

- **Initial Swing (60%-70%):** Initial Swing begins with the lift-off of the foot from the floor and ends when the swing foot is opposite to the stance foot. While the stance limb is in mid-stance phase the foot of the swing leg advances by hip and knee flexion [33], [35].

- **Mid Swing (70%-85%):** Characterized by hip flexion and knee extension helping the advancement of the limb. Mid swing starts when the swinging limb is opposite to the stance limb, and it ends when the swing limb is forward and the tibia vertical. The other limb is in mid stance phase [33], [35].

- **Terminal Swing (85%-100%):** It's the final subphase of the whole GC. It begins with a vertical tibia ending when the foot strikes the floor. Limb advancement in result of knee flexion is complete as the leg moves ahead of the thigh [33], [35].

Figure 2.4 aims shows the different phases and subphases previously mentioned

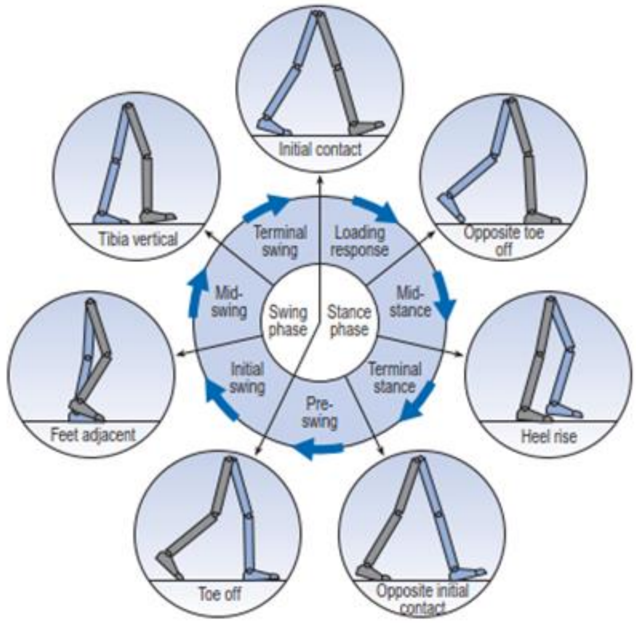


Figure 2.4 Phases, subphases and events during a single GC by the right (grey) leg. Taken from [35].

As previously mentioned, and shown in Figure 2.4, besides the relevance of the different phases and subphases of the human gait, there are events that deserve to be highlighted as they often mark the end of a subphase and the beginning of the next one. The nomenclature of these events is representative of what they consist of and is related to the foot's movement. The most important events in the GC will be presented below and grouped according to the GC phase they belong to:

I) Stance Phase:

- **Heel Strike or Heel Contact (HC):** This event initiates the GC and represents the first contact between heel and floor. At this point the body's center of gravity is at its lowest position [33], [35].

- **Foot Flat:** The plantar surface of the foot touches the ground while occurs the opposite foot toe-off [33], [35].

- **Mid Stance:** Occurs when the swinging foot passes the stance foot. In this position the body center of gravity is at its highest position [33], [35].

- **Heel rise:** occurs when the heel loses contact with the ground [33], [35].

- **Toe-off (TO):** Occurs after the opposite initial contact. This event terminates the stance phase as the entire foot leaves the ground [33], [35].

II) Swing Phase:

- **Acceleration:** begins when the foot leaves the ground, and the subject activates the hip flexor muscles to accelerate the leg forward [33], [35].

- **Mid-Swing:** The foot passes the body coincidental with midstance for the other foot [33], [35].

- **Deceleration:** The muscles slow down the leg and stabilize the foot in preparation for the next heel strike [33], [35].

2.1.4 Human Gait Biomechanical Parameters

In addition to the phases and events of the human gait, there are some parameters that also allow a complete analysis of the human gait. In general, these parameters can be divided into 4 large groups: kinetic, kinematic, spatiotemporal, and physiological [35].

Concerning kinetic parameters, these refer to the study of forces, masses, moments, and accelerations related to human walking. Usually, these parameters are studied, for instance, using force platforms. One of the main kinetic parameters are the Ground Reaction Forces (GRF) [35][38]. Human body propulsion during movement results in the appearance of forces on the ground – GRF in Antero-

Posterior (AP), Medio-Lateral (ML) and vertical directions. These forces allow the determination of other interesting parameters regarding the human gait, such as the Rate of Loading (RoL) and the breaking and acceleration impulses. Regarding RoL, this parameter is exclusively related to the maximum value of the vertical component of the GRF during the HC, being often used to measure the severity of the contact between foot and ground. This parameter is therefore related to the breaking impulse, since it also refers to the GRF components during the HC, which are higher when the subject desired to stop walking. In turn, acceleration impulse is related to the GRF generated by plantarflexion at the end of the stance phase and this acceleration is responsible for the propulsion of the body for the remaining GC [38].

Kinematic parameters, on the other hand, are used to measure joint movement profiles independently of the internal and/or external forces that cause the movement. These parameters are often measured using motion capture systems. One of the most studied kinematic parameters are the joints angles (hip, knee, and ankle) during a GC in sagittal plane. Figure 2.5, related to a healthy human gait, shows a typical graphic of these parameters [35].

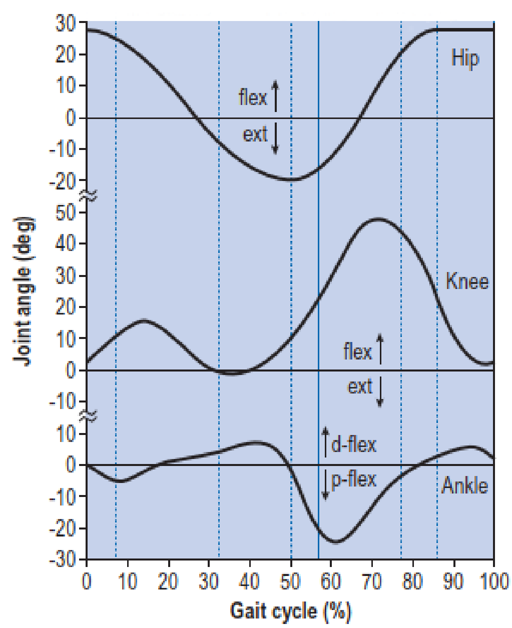


Figure 2.5 Hip, knee, and ankle joints in sagittal plane during a healthy human gait. Taken from [35].

In turn, spatiotemporal parameters, are related, to the movement along time and space during a GC. Addressing some of the most common spatiotemporal parameters, stride length is the measure of distance between two points of contact made with the same foot. On the other hand, step length also refers to a distance, but between two successive contact points made with different feet. To these two parameters, stride duration and step duration are, respectively, associated. Another parameter related to stride duration is cadence (in steps per second), defined as half-strides per 60 seconds or full strides per 120 seconds. Naturally, walking speed is also an important spatiotemporal parameter which refers to the distance covered in a period [33][39].

Another important spatiotemporal parameter is the BoS, also called walking base, defined as the side-to-side distance between the line of the two feet, usually measured at the midpoint of the back of the heel or sometimes below the centre of the ankle joint. Finally, toe-off angle is defined as the angle, in degrees, between the direction of progression and a reference line on the sole of the foot, usually the midfoot line. These parameters are shown in Figure 2.6 presented below [33]. Finally, the physiological parameters are related with energy costs and metabolism as is the case of brain (using Electroencephalography - EEG), muscle (EMG) and cardiovascular (Electrocardiography - ECG) activity [35].

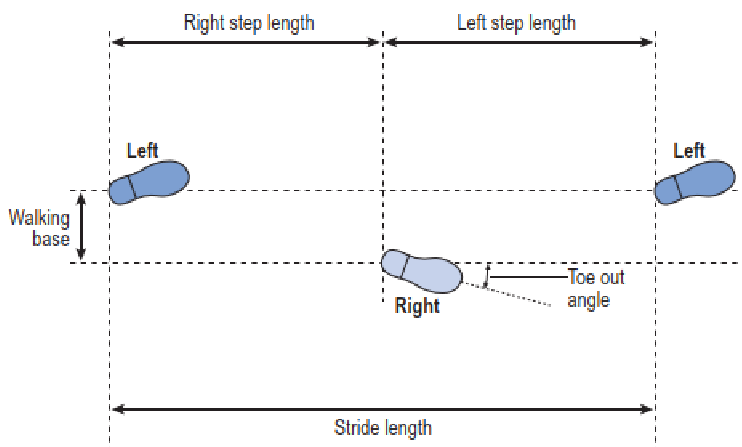


Figure 2.6 Spatiotemporal parameters related with foot displacement. Taken from [35].

2.1.5 Lower Limb's Anatomy

As previously mentioned, the activation of certain muscles is a fundamental aspect of locomotion. Thus, and since in this dissertation analysis of EMG data will be performed, the muscles that intervene actively will be briefly presented in this section together with the joints that contribute to the human locomotion namely hip, knee, and ankle joints [40].

Among the various types of existing human joints, the main joints responsible for human locomotion belong to the class of synovial joints, also called diarthroses, as these joints allow a wide range of movement. In a synovial joint the bony endings are covered by hyaline cartilage and the joint is surrounded by a synovial capsule that secretes a lubricating fluid - the synovial fluid. Many of these joints are stabilized by ligaments which are bands of relatively inelastic fibrous tissue that provide a connection between two bony structures [40].

Regarding the hip joint it joint is the only true “ball-and-socket” joint in the human body. These joints are also called enarthrosis and are made up of a spherical bone segment that fits inside a cavity with the same shape and therefore can rotate in all directions. In the case of the hip joint the “ball” is the head of the femur and the “socket” is the acetabulum of the pelvis [40].

During stance phase gravity pushes the body weight and acetabulum against the femoral head. More significantly, each femoral head is pushed up against its socket when the foot contacts the ground and then pushed against the ground. During this phase the forces on the hip joint can reach 4 times the body weight. These forces are mainly caused by the GRF and hip abductor muscles as they intervene in the deceleration process of the rotations of the pelvis. In contrast, during the swing phase as the limb is not in contact with the ground the compression force on the hip is about 50% of the body weight [40].

In turn, knee joint is the largest synovial joint in the human body. It consists of the contact between the medial and lateral condyles of the femur at the top and the corresponding condyles of the tibia at the bottom. The gap between these two bony segments that is filled both sides by the meniscus, a cartilage whose main functions are load distribution and pressure reduction at the point of contact. As a trochlearthrosis, the main movements allowed by this joint are flexion and extension in the sagittal plane. There are also movements with less amplitude, such as medial and lateral rotation [40].

During the swing phase ending, when the swinging leg prepares to contact the ground, there is an extension of this joint so that the HC may occur. In relation to the stance phase, the knee minimizes the contact of the flat foot with the ground, since this situation is associated to a lower energy dispersion and, consequently, to higher impact forces. The main forces involved related to this joint during the act of walking are compressive and frictional forces. To support the former, the meniscus plays a fundamental role by increasing the contact area and thus reducing the pressure. Frictional forces, on the other hand, appear because of the combination of vertical compressive forces and horizontal shear forces. The synovial fluid ensures that both bony ends never touch with each other, thus allowing a movement without friction and wear [40].

Finally, regarding the ankle joint, although it is composed of 3 different joints, the talocrural joint is the one that most actively participates in the walking process. It is cylindrical and formed superiorly by the tibia and inferiorly by the talus. As the knee joint, this is approximately a trochlearthrosis. Although uniaxial its axis of rotation is dynamic depending on the degree of knee flexion: When the knee is straight the range of dorsiflexion is 10 degrees while for the flexed knee it increases to 30 degrees. On the other hand, the range of plantarflexion is about 30-40 degrees [40]. Table 2.1 shows the muscles that cause the main movements of each one of the articulations as well at its range.

Table 2.1 Joints RoM and muscles that actively intervene in human gait [41]

Joint	Movement	Muscles
Hip	Flexion (100°)	Psoas major, iliacus, and rectus femoris assisted by tensor fasciae and sartorius.
	Extension (30°)	Gluteus maximus, biceps femoris, semi-tendinosus, semi-membranosus and adductor magnus
Knee	Flexion (150°)	Biceps femoris, semitendinosus and semimembranosus, assisted by gracilis, sartorius and popliteus. Gastrocnemius and plantaris also assist when the foot is on the ground
	Extension (0°)	Quadriceps femoris assisted by tensor fasciae latae.
Ankle	Dorsiflexion (between 10° and 30°)	Tibialis anterior assisted by digitorum longus and hallucis longus and fibularis tertius
	Plantar Flexion (between 30° to 40°)	Gastrocnemius

2.2 Wearable Robotic Devices for Slip-related Falls Prevention

As previously mentioned, there are not many works in this concrete area of robotic assistive devices for slip-related falls prevention. Mioskowska et al [31] developed a compressed gas actuated knee assistive exoskeleton for slip induced fall prevention during human walking. This cable driven device is very similar to a knee orthopaedic brace to be easily used, lightweight and comfortable during walking. It has as its main function the extension of the trailing leg knee to interrupt early the swing phase, being this one of the main strategies further discussed, in Chapters 3 and 5, to increase the stability of the body. These authors chose to place the actuator and the electronic parts in a backpack to minimize the interference with the unperturbed gait. This device is shown in figure 2.7 [31].

Regarding to the mechanical design, the orthopaedic knee brace is 3D printed from Polylactic Acid (PLA) plastic with an integrated commercial aluminium hinge which includes a mechanical hard stop to prevent knee overextension for safety reasons. As shown in Figure 2.7 this device is attached on subjects' thigh and shank with Velcro straps [31].

This device has a pneumatic actuator, i.e., a cylinder with a 12g CO_2 cartridge. The piston of the air cylinder is connected to the brace through a Bowden cable. When pressure is released to the cylinder the piston pulls the cable creating a torque for knee extension. The assistive torque exerted by the device on the knee is based on cylinder pressure, according to the following equation:

$$\mathbf{T\ assist} = \mathbf{P} \times \mathbf{A} \times \mathbf{r}, \tag{2.1}$$

where P is the applied pressure on the cylinder, A the surface area of the piston and r the hub's radius. The maximum torque value for this device is 25 Nm. Naturally, for safety reasons this device has pressure regulators, a solenoid, and a microcontroller to guarantee an adequate pressure in the cylinder. All these components are powered from a common power bank [31].

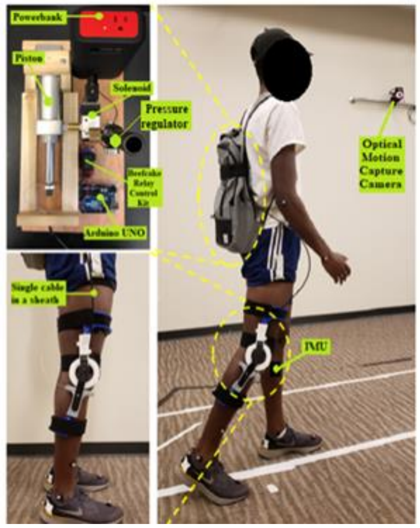


Figure 2.7 Robotic Assistive Device and its main components developed by Mioskowska et al. Taken from [31].

After sets of tests with subjects, the authors corroborated that the action of the device helps in the early contact of the foot, being this the main objective of its use. The actuation is triggered at 66% of the gait, ending with the foot on the ground at 78% of the gait, in mid-swing subphase (Figure 2.8 a).

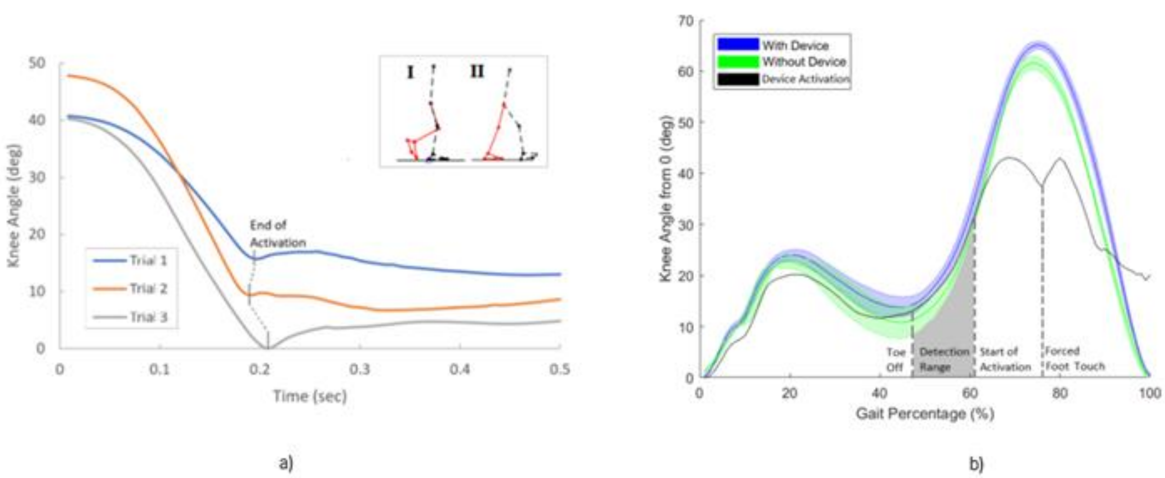
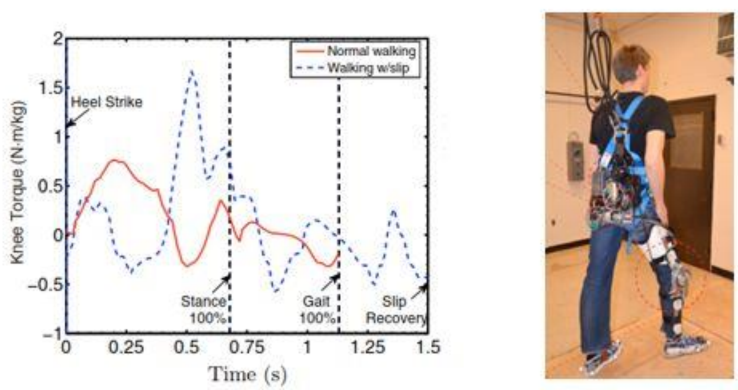


Figure 2.8 a) Knee angle over time for stationary tests. b) Knee angle over time during test with three different conditions. Adapted from [31].

Additionally, the tests performed allowed the authors to demonstrate that the device developed is light enough, thus having a residual alteration of the healthy gait, with no great difference between the presence or absence of it, as shown in Figure 2.8 b). Regarding the aspects to be improved, the authors highlight the power losses due friction with this value reaching 20% when testing knee extension with subjects seated, and the use of Inertial Measurement Units (IMUs) to detect the slip event [31].

Trkov et al [32] also developed a robotic knee-actuating device to prevent slip related falls. Similarly, to the previous device, these authors have chosen to place the electronic components on the back to mitigate the existence of disturbances in the gait as shown in Figure 2.9 b). In this case, the knee actuation torque is produced by an electric motor, followed by a Series Elastic Actuator (SEA) unit. These authors found considerable differences between the knee joint's torque during normal walking and slip situations (Figure 2.9 a)). When this happens, the device produces a torque with half of the magnitude of the peak torques in both situations to restore natural movement [32].



a) b)

Figure 2.9 a) Differences between knee torques between normal gait and slip situation. b) Robotic Assistive Device developed by Trkov et al. Adapted from [32].

The device developed by these authors also integrates a clutch mechanism in its design. The use of this type of mechanism in robotic assistive devices when needed is of great importance, since this mechanism enables the device to be either constantly engaged or unactuated during normal walking. This mechanism offers advantages in terms of power consumption being also useful for safety reasons [32].

In the case of this device also the motor is placed at the back. The motion of this is forwarded to a bearing-supported shaft. In this shaft spiral mitre gears are used to transfer the motion 90 degrees to the main shaft. A harmonic drive is used to reduce the motor gear ratio to 80:1 with its output connected to a torsional spring with a stiffness of 150 Nm/rad. Two potentiometers are used to measure the angle between the thigh and the shank and the deformation of the torsional spring. The two brakes on the thigh brace limit the knee flexion to 120 degrees for safety reasons [32].

After testing subjects in steady walking situations, the authors determined that the device does not interfere with the natural dynamics of the human gait for medium and fast walking speeds. Additionally, in slip recovery situations, the device allows the control of the adequate positions of the joints to guarantee the slip response [32].

Monaco et al [22] also developed a slip recovery strategy in an active pelvis orthosis (APO) with the objective of acting when needed for elderly and amputees. The device used in this study consists of two joints (one in each hip) assisting extension and flexion movements. In this device, the movement in the hip is produced by a DC electric motor and transmitted to the subject through a SEA [22]. SEA unit is also composed of an encoder, a harmonic drive, and a transmission bar mechanism. These components allow, respectively to measure the absolute hip angle, reduce the motor's gear ratio (80:1) and transmit the movement to the user's hip. As shown in Figure 2.10, the authors have chosen to build a SEA unit consisting of two axes in order to place the bulkier components in a less inconvenient area to users [22][42].

SEA actuators are very useful in robotic where is a constant human-robot interaction as they are characterized by a low impedance control (better compliance). These components allow the discomfort prevention due excessive interaction with the device especially during high frequency movements. The stiffness of the custom torsional spring has a value of 100Nm/rad comparable to the average hip joint during walking [22], [32].

Additionally, this device is supported by a C-ring and three shells, two on the legs and one on the back. The existence of various adjustment mechanisms makes it possible to adapt the device to different users. On the other hand, the inclusion of a passive degree of freedom (hip abduction and adduction) results in greater comfort [43].

Regarding the materials, the orthosis structure is made of lightweight materials, i.e., carbon fibre and aluminium, while the robot-user interfaces are made of orthopaedic material, i.e., a thermoplastic polyethylene [43]. Both, device's structure, and SEA unit are shown in Figure 2.10.

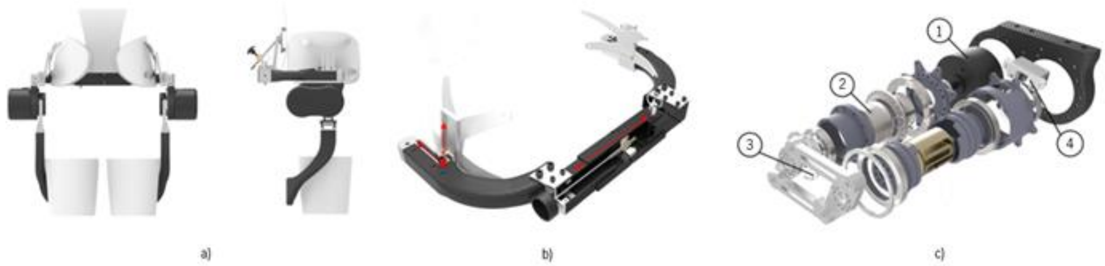


Figure 2.10 a) Robotic device general overview; b) C-shape bar and c) SEA united developed by Monaco in [43].

Based on previously collected data, the control algorithm employed by these authors allowed the definition of thresholds for the CoM stability related to the BoS. When the threshold is exceeded, it triggers the action of the APO, thus re-establishing an adequate position for the continuation of the human gait, thus avoiding a fall. This algorithm has two different modes: Z-mode and A-mode. The former is characterized by an absence of assistance provided to the users, whereas in A-mode is activated in potential fall situations to supply assistive torques at hip joints when detected postural transitions. In this case the device enables the fall mitigation strategy (red and yellow start) and the CoM is confined to the green zone presented in Figure 2.11 – stability region [22].

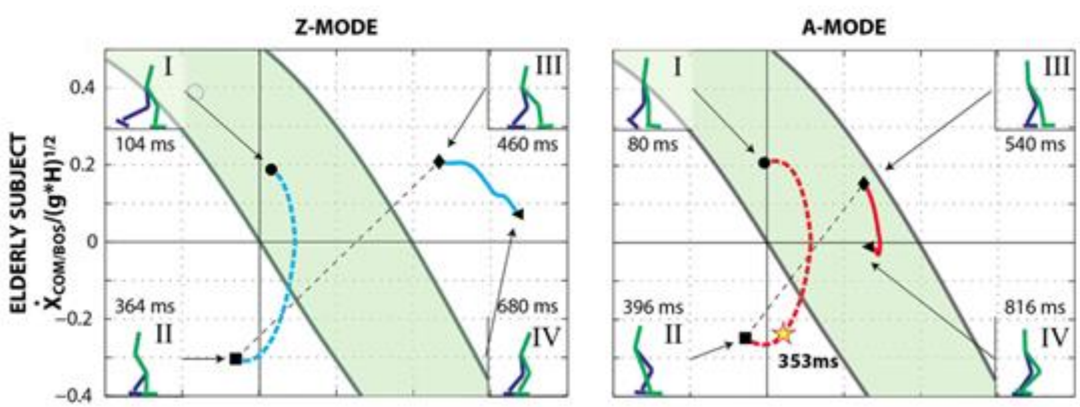


Figure 2.11 Z-Mode and A-Mode actuation modes. Adapted from [22].

With a different purpose, Kumar et al [44] when developing a new control policy for slip-like falls prevention, have concluded about the existence of a relationship between the mechanical design of these devices (hip orthosis in this concrete case) and stability region [44]. These authors when, theoretically, comparing different design possibilities stated that 3D actuation, both in sagittal and frontal planes increases the stability region in all directions comparatively to 2D actuation, as shown in Figure 2.12.

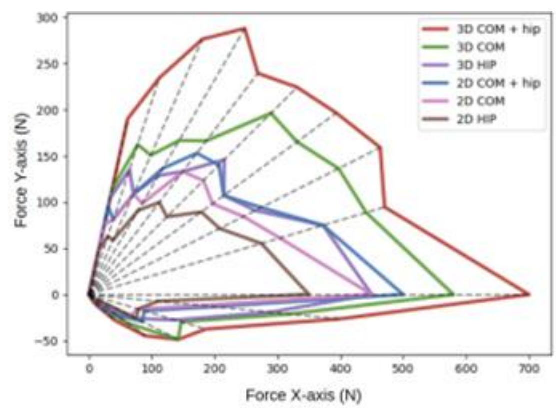


Figure 2.12 Stability region in 6 different mechanical designs studied in [44].

Table 2.2 summarizes the main specifications of the devices discussed in this section. In the case of last study, since this is more directed to a control algorithm, there is lack of information about the mechanical specifications of the device.

Table 2.2 Main specifications of the analyzed fall prevention robotic devices where N\A = Not Available

Study / **Specifications**	**Mioskowska** [31]	**Trkov** [32]	**Monaco** [22]	**Kumar** [44]
Maximum Torque	20 Nm (due friction losses, in theory this value would reach up to 25)	40 Nm (35 Nm in sit-to-stand situations)	0,2 Nm/ kg (orthosis weight inclusive)	30 Nm
Movement Range	0-90° was the maximum range assessed in the study	0-90° was the maximum range assessed in the study	-20 to 90°	"Similarly to anatomical"
Response Time	Depends on cylinder pressure: 82 ms when with 100 psi from 90° to 0° and 72 ms from 60° to 0° also with 100 psi	Less than 200 ms	350 ms	N\A
Detection Time	100 ms	30 - 90 ms	300 ms	N\A
Weight	2,66 kg (0,49 kg on the leg and 2,17 kg on the back)	6,7 kg (2,5 kg on the leg and 4,2 kg on the back)	4,2 kg without considerer the control unit (expected to reach 5-6 kg)	N\A
Angular velocity	N\A	1,2 rps	Between 4 rad/s and 5 rad/s	N\A

3 Biomechanical Strategies to avoid Falls during a Slip Event Review

The analysis of the biomechanical response to slip events emerges as an essential study in the field of slip related falls prevention. Thus, the present chapter presents a state-of-the-art review of the existing literature related to the biomechanical response to slip events. In this review various types of data were included namely kinetic, kinematic, spatiotemporal and EMG data.

Considering the prevention approaches of the high falls' incidence worldwide, existent literature shows the relevance and capacity of repetitive training in preparing subjects for real world perturbations, resulting in proactive and adaptive responses in potential slip situations, thus reducing the incidence of falls. Briefly, in this type of procedure, the subjects repetitively face simulated slip situations [11], [45], [46]. Also, in the last few years the literature associated to the development of medical robotic devices to detect and prevent falls occurrence caused by slip-like perturbations have increased as an alternative approach to reduce slip falls' incidence. In addition to slip detection, the main goal of this type of devices is to restore an appropriate biomechanical position to allow the subject to regain a steady walking pattern after a slip perturbation [22], [31], [32].

The study of the biomechanical response to slip perturbations cannot be dissociated from both approaches. While in the case of repetitive training, the study of this response allows quantifying and analyzing the evolution of the strategies developed by individuals as a result of this training, in the case of the development of devices that act on this problem, the study of the biomechanical response emerges as an important step in the definition of requirements and specifications of the device during the initial phase of the project [31], [32], [47].

The analysis of the biomechanical response to slip perturbations, in most situations, is carried out using motion capture systems that allow the analysis of kinematic variables, namely joint angles [23]. Frequently, this type of analysis is also complemented with the analysis of kinetic variables as GRF and EMG activity [27], [28].

Regarding the existent reviews related to this topic, a preview study conducted by McCrum et al addressed only the different perturbation methods used to provoke slip perturbations not focusing on the biomechanical strategies developed by the subjects [17]. Also, Tokur et al [48] have performed a review to evaluate the balance recovery in response to external perturbations. Although they consider distinct perturbations (slips, obstacles, or ground level changes for example), the biomechanical responses to slip disturbances are not analysed in detail either [48]. Finally, Grönqvist's review related with slip perturbations addressed different strategies to measure surface slipperiness. Despite considering joint angles and EMG measurements as variables that can be considered to measure floor slipperiness, this review does not provide a complete analysis of these parameters focused on the biomechanical response evidencing only conclusions related with foot-floor contact [49].

The main objective of this state-of-art analysis is to study the biomechanical response to slip perturbations, considering multivariate data, during the period between slip onset, which happens some milliseconds after the HS, and steady walking regaining, thus, provide a complete kinematic analysis of the biomechanical response to slip perturbations complemented with EMG and spatiotemporal metrics. This analysis will also be useful as a preliminary point for the study of experimental data carried out in the Chapter 5. Hence, the following RQ's were addressed in this state-of-the-art review: i) "What are the biomechanical responses of the lower limbs during a slip event?"; ii) "What roles do the trailing and slipping leg distinctly play?"; iii)" Which muscles of the lower limbs are involved in the recovery process?"; iv)" What are their activation times?"; v) "How does age influence this response?"; vi) "What happens to the relative motion between BoS and COM during slip perturbations and the respective response ?" and vii) "Which are the repetitive training effects in the variables previously discussed?".

3.1 Review strategy

To study the biomechanical response after slip perturbations, a state-of-the-art review of the literature associated with this type of perturbation was performed in SCOPUS, PubMed, Web of Science and CINAHL (EBSCO) databases. This search was carried out until 28 February 2022 using the following search key: (gait OR walking OR walk OR locomotion) AND slip* AND (training OR exercise OR adaptation OR rehabilitation OR adaptive OR repeated OR repetition OR task OR response OR adjustments OR biomechanical). The search with these keywords was restricted to the articles' title and abstract.. The selected key was not adapted from any of the existent reviews due to the lack of biomechanical response analysis already enunciated. It was selected with the aim of ensuring to cover as many articles as possible where the biomechanical response was likely to be analysed. Thus, some keywords such as "repetition" or "adaptive" were included due previously acknowledge of the existence of a high number of studies in repetitive training to prevent slip perturbations. The time window considered includes all articles from the beginning of the year 2000 to the research date to encompass a large number of papers. Also, during the search the emergence of more articles related to this topic after 2000 was confirmed, so this was the year considered.

A total of 1990 articles were collected from this search. After eliminating duplicates using Mendeley Software ®, a total of 794 were left for further screening. Firstly, a title-based screening was conducted. This process allowed the inclusion of articles that meet the following inclusion criteria: i) articles written in English; ii) the paper was not a review; iii) articles where the slip perturbation was applied only in healthy subjects to facilitate the comparison between different articles; iv) perturbation applied during normal gait (sit-to-stand and other routine tasks were not considered) and finally v) perturbations induced only with the subject's hands free (backpacks carrying a percentage of the subject´s body mass were excluded) and finally vi) include lower limbs' biomechanical analysis. Although age was not considered an exclusion criterion, its influence on the biomechanical response to this type of perturbation is an aspect to be analysed during this review. A total of 203 articles resulted from this screening procedure. Articles where was not clear to evaluate the inclusion of the criteria selected were selected for an abstract based screening process. Apart from the criteria used for title analysis were added the following exclusion criteria: analysis of shoe-floor parameters (e.g. CoF). After this stage remained 115 articles. Through the reading of the abstract, it was not possible to screen some of these articles, so they were read in full to ensure compliance with all the criteria mentioned above. Although articles whose approach is related to repetitive training were included, the present review is dedicated to a deeper analysis of the strategies developed after a slip and not the variation of parameters as a result of a period of repetitive training. Naturally, kinematic, kinetic, spatiotemporal and EMG metrics in articles referring to repetitive training will also be analysed, however, the metrics more specifically related to repetitive training results (e.g., retention period, number of repetitions) will not be the subject of a more detailed analysis. Also, articles related to fatigue effects were included for allowing to conclude about the importance of some articulations and muscles in slip responses situations. Following the completion of the screening procedure, a total of 41 articles were included in the analysis. Figure 3.1 shows the Preferred Reporting Items for Systematic Review and Meta-Analysis (PRISMA).

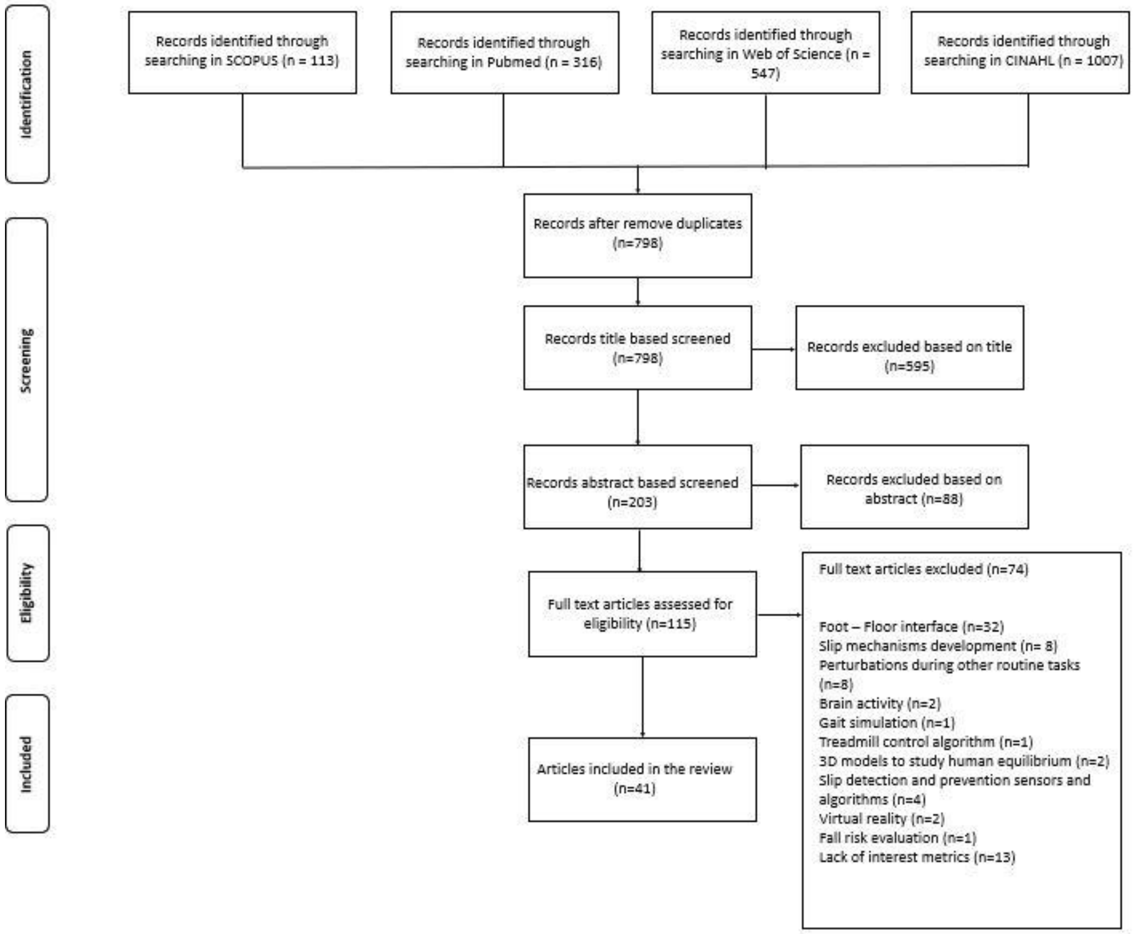


Figure 3.1 PRISMA flow diagram.

3.2 Reviewed literature

Considering all the 41 articles included in this state-of-art analysis, it was possible to generally group them according to each of the following approaches: i) kinetic, kinematic and electromyographic metrics analysis (detailed further below) after a first unexpected slips; ii) biomechanical computational models and finally ii) repetitive training. Table 3.1 addresses the subjects' information, perturbation mechanism, and independent variables considered in each of the included studies while Table 3.2 shows the articles grouped according to the procedure previously described and the variables considered in each study.

Table 3.1 Subjects characteristics, perturbation mechanism, and independent variables defined in the literature screened. N\A= Not Available

Articles	Subjects Number	Age (Years)	Weight (Kg)	Height (Means)	Perturbation Mechanism	Subject's Gait Speed	Iv's Studied
[50]	19 (10 males and 9 females)	24 ± 4	N\A	N\A	Treadmill acceleration	1 m/s	Gait asymmetries effect namely variable step lengths
[12]	58 (29 middle-aged and 29 young)	middle-aged 60.6 ± 3.9 young: 23.5 ± 3.2	69.1 ± 13.5	173 ± 8.6 cm	Glycerol and water mixture	N\A	Steady walking vs slip perturbations
[45]	45 (19 males and 26 females)	71.4 ±3.6	56.5 ± 10.5	154.6 ± 7.9 cm	Treadmill acceleration	2 km/h	Repetitive training effects
[51]	10	N\A	64.77 ± 15.5	169.4 ± 7.0 cm	Platform release	Self-selected	Steady walking vs slip perturbations
[52]	14 young, 14 middle aged and 14 older (7 males and 7 females for each age group)	22.6 ± 2.1; 46.9 ± 13.6; 75.5 ± 6.8	68.7 ± 9.6; 75.5±16.1 76.8 ± 13.3	169.7 ± 6.1 cm; 173.5±6.3 cm; 170.2 ± 6.4 cm	Treadmill acceleration	Self-selected	Fatigue and age effects
[26]	10 young males	21.0 ± 1.0	63.1 ± 6.2	1.71 ± 0.08 m	Treadmill acceleration	80 and 140 bpm	Steady walking vs slip perturbations and Gait speed effect
[9]	16 (10 males and 6 females)	24.66 ± 3.58	65.86 + 10.93	1.75 + 0.07m	Water and jelly mixture	Self-selected	Steady walking vs slip perturbations and Muscular Fatigue effect
[53]	69 (35 males and 34 females)	25.87 ± 4.5	63.87 ± 11.7	168.47 ± 8.4 cm	Platform release	N\A	Steady walking vs slip perturbations

[10]	22 (10 females and 12 males)	Females: 24.1 Males: 25,.5	Females: 49.7 Males: 69	Females: 159.7 cm Males: 171 cm	Soap patch	120 step/min and 90 step/min	Steady walking vs slip perturbations
[54]	16 (8 males and 8 females)	23 ± 4	68.7 ± 6.8	N\A	Water and oil mixture	Self-selected	Steady walking vs slip perturbations
[55]	24 young and 24 middle-age	Young: 23.75 ± 2.83 Middle-Age: 57.13 ± 2.83	Young: 68.92 ± 10.09 Middle-Age: 81.81 ± 14.22	Young: 1.73 ± 0.08 m Middle-Age: 1.69 ± 0.08 m	Glycerol and water mixture	Self-selected	Steady walking vs slip perturbations and slip severity
[56]	14 (7 males and 7 females)	27.29 ± 4.27	N\A	N\A	Platform release	Self-selected slow, normal, and fast	Steady walking vs slip perturbations
[27]	25 (14 males and 11 females)	24.0 ± 1.9	56.8 ± 9.9 kg	168.8 ± 9.0 cm	Soap Patch	Self-selected	Steady walking vs slip perturbations and successfully recoveries vs falls
[57]	8 young (4 males and 4 females) and 8 adults (5 males and 3 females)	24 ± 2.7 and 65 ± 4.8	63 ± 13 and 67 ± 14	1.69 ± 0.09 and 1.69 ± 0.10 m	Treadmill acceleration	1 m/s	Steady walking vs slip perturbations and age effects
[58]	29 young (14 female and 15 male) and 28 older (15 female and 13 male)	Young female: 25 ± 4 Young male: 23 ± 2 Older female: 55 ± 3 Older male: 58 ± 6	Young female: 63 ± 12 Young male: 75 ± 11 Older female: 82 ± 18 Older male: 88 ± 13	Young female: 166 ± 5 cm Young male: 178 ± 7 cm Older female: 164 ± 5 cm Older male: 177 ± 6 cm	Glycerol and water mixture	Self-selected	Steady walking vs slip perturbations and knee muscular strength effect
[59]	10 (8 males and 2	25.4 ± 3.4	80.7 ± 14.5	1.75 ± 0.07 m	CoF alteration	N\A	Steady walking vs slip

	females)						perturbations and slip onset timing
[60]	17 young (8 males and 9 females) and 17 older (3 males and 14 females)	25.2 ± 3.7 and 62.4 ± 6.6	71.8 and 10.1 kg 66.5 and 11.3 kg	176.1 ± 8.1 cm 161.8 ± 7.2 cm	Treadmill acceleration	1.2 m/s	Repetitive training effects
[61]	6 young adults (4 male and 2 female) and 6 older subjects (2 male and 4 female)	24.0 ± 1.7 and 66.7 ± 5.4	65.2 ± 8.8 and 64.5 ± 12.3	169.5 ± 8.14 and 162.5 ± 6.16	Treadmill acceleration	Normalised speed	Steady walking vs slip perturbations and age and intensity effects
[62]	10 young adults (6 female and 4 male)	27.3 ± 4.1	68.5 ± 10.3	169 ± 10 cm	Treadmill acceleration	Self-selected	Steady walking vs slip perturbations
[63]	24 (12 in control group and 12 in training group)	Control: 74.18 ± 5.82 Training: 71.24 ± 6.82	Control: 69.63 ± 9.45 Training: 68.24 ± 8.04	Control: 169.41 ± 9.16 cm Training: 167.45 ± 11.52 cm	Treadmill acceleration	self-selected	Repetitive training effects
[15]	16 (8 males and 8 females)	24.66 ± 3.58	65.86 ± 10.9	174.86 ±7.67 cm	Slippery floor	self-selected	Steady walking vs Perturbation and fatigue effect
[25]	5 (3 males and 2 females)	25.4 ± 3.1	63.2 ± 11	1.7 ± 0.1 m	Platform release	Normalised speed	Steady walking vs Perturbation and perturbation intensity effect
[64]	75 (57 females and 18 males) divided in 3 different	74.4 ± 5.8 73.0 ± 5.9 72.2 ± 6.3	N\A	N\A	Platform release	self-selected	Repetitive training effects

	groups						
[65]	20 (10 males and 10 females)	23.3 ± 3.3	67.6 ± 12.2	173.2 ± 7.6 cm	Treadmill acceleration	Comfortable speed chosen by participants	Steady walking vs perturbation
[8]	20 (10 younger and 10 older)	Young: 24.4 ± 2.5 Older: 66.3 ± 5.1	Young: 63.1 ± 9.1 Older: 66.9 ± 10.8	Young: 1.69 ± 0.07 Older: 1.66 ± 0.08	Treadmill acceleration	Normalised speed	Steady walking vs perturbation and age effect
[29]	18 (9 males and 9 females)	23.06 ± 3.42	66.87 ± 12.36	172.96 ± 7.96 cm	Treadmill acceleration	Comfortable speed chosen by participants	Steady walking vs Perturbation
[16]	20 (9 females and 11 males) divided in mild and severe slips	Mid: 24.17 ± 2.79 Severe: 22.75 ± 1.48	Mid: 68.41 ± 11.89 Severe: 70.00 ± 11.37	Mid: 171.75 ± 8.59 Severe: 175.19 ± 7.57	Glycerol and water mixture	self-selected	Steady walking vs Perturbation and slip severity
[23]	8	Between 19 and 27	N\A	N\A	Soap and water mixture	self-selected	Steady walking vs Perturbation and dominance effect
[28]	17 (8 males and 9 females)	25.2 ± 3.7	71.8 ± 10.1	176.1 ± 8.1 cm	Glycerol and water mixture	self-selected	Steady walking vs perturbation and age effect
[11]	38 (18 males and 20 females) divided in 2 groups	73.42 ± 5.42 70.13 ± 4.75	74.59 ± 12.57 77.26 ± 12.76	1.69 ± 0.10 m 1.71 ± 0.11 m	Platform release	self-selected	Repetitive training effects
[66]	25 (16 females and 9 males)	fall: 73 ± 4.9 recovery: 74 ± 4.1	67.1 ± 12.8 73.3 ± 13.8	1.66 ±0.082 m 1.67±0.129 m	Platform release	self-selected	Steady walking vs Perturbation
[67]	40 (17 females and 13 males)	No Fatigue: 23.1 ± 1.7 Fatigue 24.2 ± 3.0	No Fatigue: 58.1 ± 10.8 Fatigue 59.9 ± 6.9	No Fatigue: 167.7 ± 7.2 Fatigue 167.9 ± 6.7	Water and detergent mixture	self-selected	Steady walking vs Perturbation and muscular fatigue
[68]	34 (13 females and	26.5 ± 5	N\A	N\A	Platform release	self-selected	Repetitive training effects

<table>
<tr><td></td><td>21 males)</td><td></td><td></td><td></td><td></td><td></td><td></td></tr>
<tr><td>[14]</td><td>15 young (9 females and 6 males) and 13 older (5 females and 8 males)</td><td>23.5 ± 3.3 and 61.1 ± 3.7</td><td>66.8 ± 10.4 And 76.5 ± 11.8</td><td>170.2 ± 8.3 cm and 165.8 ± 7.7 cm</td><td>Glycerol and water mixture</td><td>self-selected</td><td>Steady walking vs Perturbation</td></tr>
<tr><td>[18]</td><td>N\A</td><td>N\A</td><td>N\A</td><td>N\A</td><td>Platform release</td><td>N\A</td><td>Steady walking vs Perturbation and gait speed</td></tr>
<tr><td>[19]</td><td>12 (8 female and 4 male)</td><td>20.67 ± 1.23</td><td>67.01 ± 6.43</td><td>173.75 ± 9.45 cm</td><td>Rollers lock and unlock</td><td>self-selected</td><td>Repetitive training effects</td></tr>
<tr><td>[47]</td><td>10 young adults (5 females and 5 males)</td><td>24.4 ± 2.9</td><td>64.7 ± 15.5</td><td>1.69 ± 0.07 m</td><td>Platform release</td><td>self-selected</td><td>Steady walking vs perturbation</td></tr>
<tr><td>[69]</td><td rowspan="2">15 (10 males and 5 females)</td><td rowspan="2">26.1 ± 1.3</td><td rowspan="2">68.8 ± 12.3 kg</td><td rowspan="2">1.78 ± 0.06 m</td><td rowspan="2">Treadmill acceleration</td><td rowspan="2">Normalised speed</td><td>Steady walking vs perturbation and subject's dominance</td></tr>
<tr><td>[24]</td><td>Steady Walking vs Perturbation and perturbation side and direction</td></tr>
<tr><td>[13]</td><td>67 in the first set and 60 in the second set</td><td>Set 1: 26 ± 5
Set 2: 25 ± 5</td><td>Set 1: 63 ± 13
Set 2: 67 ± 14</td><td>Set 1: 1.69 ± 0.09 m
Set 2: 1.69 ± 0.1 m</td><td>Platform release</td><td>N\A</td><td>Steady Walking vs Perturbation</td></tr>
<tr><td>[70]</td><td>45 (34 females and 11 males) divided in 2 groups</td><td>74.5 ± 6.9
75.0 ± 5.4</td><td>68.1 ± 12
70.3 ± 9.3</td><td>1.6 ± 0.1 m
1.7 ± 0.1 m</td><td>Platform release</td><td>self-selected</td><td>Repetitive training effects</td></tr>
</table>

Table 3.2 General approach and dependent variables analysed in the included articles

Type of approach	***Variables Analysed***	***Number of studies***	***Articles***
Unexpected first slip	Kinematic	2	[23], [24]
	Kinetic	1	[29]
	EMG	3	[27], [28], [55]
	Muscular synergies	1	[16]
	Kinetic and kinematic	1	[54]
	Kinematic, kinetic and EMG	1	[12]
	Spatiotemporal and kinetic	1	[10]
	Spatiotemporal and kinematic	2	[25], [26]
	EMG, Spatiotemporal and kinematic	1	[65]
	Muscular synergies and kinematic	1	[66]
	Slip intensity and kinematic	1	[67]
	Slip intensity and kinetic	3	[9], [15], [58]
	Slip intensity, kinetic and kinematic	1	[14]
	Slip intensity, EMG and spatiotemporal	1	[19]
	Slip intensity and spatiotemporal	4	[52], [57], [59], [61]
	Limb stability and spatiotemporal	2	[13], [53]
	Whole-body angular momentum (WBAM) and segment angular momentum (SAM)	2	[50], [69]
	Planar covariation law (PCL), spatiotemporal and kinematic	1	[8]
Biomechanical Computational models	Kinetic	2	[47], [51]
	Spatiotemporal	1	[18]
Repetitive training	EMG	1	[45]
	Spatiotemporal and kinematic	3	[56], [60], [64]
	Spatiotemporal	1	[62]
	Slip intensity, EMG, kinetic and kinematic	1	[63]
	Spatiotemporal and kinetic	1	[11]
	Limb stability and spatiotemporal	1	[70]
	EMG and kinematic	1	[68]
Total		41	

3.3 Unexpected first slip – Multivariate data analysis

In most situations, kinetic, kinematic, spatiotemporal and EMG metrics of interest are compared between induced perturbations and steady walking situations to understand those that become evident when comparing both situations [71]. This comparative analysis has the objective to understand the parameters that allow a successfully recovery after a slip-like perturbation allowing to gather a set of information and a general understanding of this biomechanical response with the objective to help in slip-related falls reduction.

3.3.1 Kinematic, kinetic, and spatiotemporal parameters' behaviour during the biomechanical response

Hip, knee, and ankle joint movements are frequently studied in the bibliography included in this state-of-the-art review. These metrics analysis allows, in general, to understand the role of a given joint during slip response. In most of the analysed bibliography, the study of these joints movements in the sagittal plane is the most relevant, thus allowing the analysis of the extension-flexion movements after an unexpected slip [9], [10], [12]–[15], [23], [26], [29], [45], [53], [54], [58], [60], [63].

Beschorner et al [12] studied the variables with higher deviations when comparing a slip response with normal human gait. In terms of joints movements, the responses of the slipping leg´s knee (116 ms after slip onset) were extensor dominant with the angular velocity of this joint reaching its maximum value after 111 ms of the slip onset. These variables present greater deviations values. The sequence of biomechanical responses is then characterized by changes of the ipsilateral hip velocity (149 ms); plantarflexion movement of the leading ankle angle (156 ms) and finally the hip response, where in the case of the leading leg its flexion movement occurs between 170 and 200 ms. In turn, an extension movement 200 ms after slip onset happens in the trailing leg. Between both last responses appears the biomechanical response of the contralateral knee (approximately 170 ms).

In spite of considering all these variables the authors referred that their response times appear during the muscle latency period (between 170 and 200 ms). Thus, it is referred that the biomechanical deviations aforementioned are caused by the perturbation itself and not by the biomechanical response. Despite this assumption these responses are also important as these movements are responsible for the human body's postural response triggering. The fact that leading foot somatosensorial and knee velocity were the first variables to undergo changes, these changes may be critical in triggering the initial postural response [12]. Regarding joint movements evident 200 ms after the slip onset this study highlights trailing leg responses which should be considered the first postural response after a slip. Trailing leg response is a well-known biomechanical response to slip perturbations resulting in an earlier swing phase interruption addressed in most of the literature included in this review [12], [14], [23], [26], [29], [56], [62], [65].

You Li-Chou et al [10] studied joints moments by application of Newton Euler equations after data collection. In slip situations, after the perturbation, subjects have a greater hip extension moment, knee flexion moment and plantarflexion moment on the slipping leg. In turn, trailing leg response was mainly characterized by hip flexion, smaller knee flexion moment and ankle plantarflexion moment. Looking for the spatiotemporal variables namely CoM and BoS relative movement, after slipping leg's HS the forward velocity of CoM in respect to BoS increases being reduced after the contralateral leg toe-off highlighting the importance of the biomechanical response of this leg as this movement to restore an adequate position of the CoM in relation to the BoS. In fact, when comparing recovery situations with subjects that fell there was a greater speed of the BoS in relation to the CoM in falls situations. In turn, when a successfully slip recovery happened, CoM and BoS velocities were approximately the same, so the CoM was brought forward which does not happen in falls situations.

Hirata et al [26] also studied the relative velocity between CoM and BoS. These authors found that for higher walking velocities the subjects have had no need to step back the trailing foot. In these situations, it was placed in a position very close to the leading foot ("narrow strategy") or, in other cases subjects were able to continue walking ("get over strategy"). In turn for lower walking velocities besides narrow strategies subjects needed to step back the trailing foot to prevent the fall. These findings are justified by the fact that when subjects walk at a faster speed, their movement velocity is equal or greater than the maximum slip velocity (defined as 1.6 m/s) being easier to overcome the perturbation [26]. Additionally, this study concludes that for higher walking velocities the joints that present significant differences in comparison with steady gait are the ankle joints. In turn, when the perturbation is given at lower velocities authors highlight both trailing and leading limb hip's roles. Regarding the unperturbed limb's hip, its extension movement appears evidenced as this movement is fundamental to perform the step back strategy. Leading hip movement is also significantly different comparing with normal gait: increased hip flexion results in a forward landing of the leading leg constructing a more stable front part of the BoS. For these reasons the authors highlight the importance of both hips movement in counteracting the anterior BoS displacement in slip situations.

Aprigliano et al [25] also studied spatiotemporal parameters during slip responses. Regarding compensatory step, defined as the time elapsed between perturbation onset and heel strike of the unperturbed foot, it increases with the increase of perturbation intensity (Table 3.3). Also Margin of stability (MoS), defined as the difference between BoS and CoM position, decreased with the intensity of the perturbation showing low balance stability during the strong perturbations. Martelli et al [57] also confirmed these findings when studying compensatory step length and its flight time [57]. Age effects on these variables were also analysed in this study. Greater perturbations intensities were associated with a reduced MoS during slip response confirming a more destabilizing effect. Thus, for both situations older adults' response is left effective comparing with younger adults as it is associated with lower MoS lengths. The first group compensatory step was also longer (Young:410 ± 0.01 s, Elderly: 0.480 ± 0.02 s) [57]. Tropea et al [61] also confirmed these spatiotemporal parameters variation during slip responses considering two different age groups (Table 3.3) [61]. Slipping time, defined as the time elapsed between the start of the perturbation (perturbed foot HS) and the following ipsilateral foot heel strike, was higher for more intense perturbations and for older adults regardless of perturbation intensity. Compensatory step time was also founded to be greater for higher intensity perturbations and for older adults considering all perturbation intensities. Another variable named Motor Control Test (MCT), defined as the time to produce a motor response after a slip was addressed by Lockhart et al [52]. These authors conclude that in older participants MCT is longer comparing with younger participants.

In turn, Cham et al [54] studied only slipping leg moments by using GRF, inertial properties of body segments and Plagenhoef's equations. These authors segmented the period from the beginning of the perturbation to the end of the stance phase of the perturbed foot by considering periods corresponding to multiple percentages of 10%. Two hip and knee different compensatory responses can be distinguished: between 25% and 45% and between 45% and 55% of the stance phase. In the first period hip and knee response is characterized by extension and flexion, respectively, while in the second period hip flexion and knee extension responses are emphasized. Regarding the ankle joint, authors refer that this is a passive joint slip perturbations' biomechanical response when comparing its role with hip and knee joints. These authors also addressed two important points related with slip perturbations: firstly, consistent to You Li-Chou study, the biomechanical response starts, approximately, after 190 ms of the slip onset and, in second, the slip itself only starts in the 60-80 ms after HS. Although authors only have analysed the response of the perturbed leg, the importance of the trailing leg is not discarded being addressing in the future works section of this article.

In contrast to the authors previously cited, Rasmussen et al [59] considered other perturbations directions apart from AP and, also, analysed perturbations in different moments of the stance phase with different severities [59]. Early, mid, and late stance were, respectively, defined as 0-33 %, 34-67% and 68-100% of the stance phase. These authors study mostly the ankle joints' response and concluded that this response differ depending on the stance phase perturbation timing: when the perturbation is given in a later stance phase the perturbation is less severe including less risks, less variable and increases the chance of a more effective biomechanical response as the contralateral limb is in an advantageous position to perform the toe-off strategy. In result contralateral steps tend to be shorter and slower in these situations [23][59]. In addition, later perturbations are characterized by lower slip distances, peak velocities, and upper body angular moment. Also, lateral, and AP displacement of the foot is reduced. Despite having less evident movements, lateral displacements of the foot also may be considered as a biomechanical response [59].

Kima et al [23] also highlighted the contralateral toe-off strategy to increase the BoS thus restoring an adequate BoS – CoM relationship. The reactive control of stability of the unperturbed (and non-dominant) limb had an impact on recovery of the whole body because this preferentially provides support to the dominant limb propulsion thus being essential to stability and balance maintenance. Regarding the slipping limb hip extension, knee extension and a flatter foot position are the strategies highlighted by the authors. Also, in response to a slip perturbation both right and left thighs appear to show increased adduction.

Dongyual et al studied the trailing leg's importance during compensatory slip response. Similarly to [14] and [47] these authors also highlight trailing leg hip extension and knee flexion during this compensatory to allow foot-floor contact and prevent a backward trunk movement [29]. Also, ankle plantarflexion and hip flexion movements are referred as significant movements characteristic of the trailing leg response.

Regarding the hip flexion movement, the authors refer that this, in some situations, may be the result of a more pronounced extension movement [29].

The study of angular moments of the whole body (WBAM) and its segments (SAM) is also an approach found in the analysed literature. Chang Liu et al [50] studied these moments, during a slip response where asymmetries (step length changes for example) were also introduced during subjects' gait as a way to understand the variation of the CNS response considering different inputs. Using Principal Component (PC's) Analysis these authors found that contributions from the lower limbs were typically dominant in the first PC's during slip responses. Within the lower limbs the segments that are furthest from the CoM are those that contribute the most as they may be subject to higher accelerations thus contributing more vehemently. Regarding the effect of step length asymmetry on intersegmental coordination patterns these authors found that the increase of asymmetry's magnitude resulted in an increased deviation of intersegmental coordination patterns from symmetrical walking while there was no change in integrated WBAM momentum compared to symmetrical walking. These findings confirm that healthy people use a flexible combination of intersegmental coordination patterns rather than invariant reactions to maintain WBAM during perturbation responses when walking with asymmetric gait patterns. Also, WBAM is a task-level variable that is stabilized by the nervous system during perturbation recovery [50].

3.3.2 Biomechanical Response study through computational models

In some of the included literature, biomechanical response analysis to slip perturbation is carried using forward dynamics biomechanical models. Hip height normalised by body height, measured as the vertical distance of the bilateral hip midpoint to the surfaces is also a variable analysed in some articles. This variable is highly correlated with the magnitude of the vertical impulse generated by the stance limb, and, for this reason, is used to characterize subject's vertical limb support against gravity. This variable becomes more important when studying slip perturbations situations as when instability combines with poor limb support at the instant prior to the recovery step, a subsequent fall incidence becomes nearly inevitable (88.9%) [13], [53]. Besides hip height, while developing a slip prediction model, Yang et al [53] also studied hip vertical velocity and limb support quotient (hip vertical velocity/hip height). Firstly, these authors conclude that the most critical period after a slip perturbation is between the slip onset and the instant prior to the trailing limb touch down where there is a deterioration of the analysed parameters. Considering this whole period, the instant before the trailing foot touch down is the most decisive in what may be a fall. Because it is the most decisive instant, the analysis of biomechanical parameters in this phase takes on greater importance in the determination of the slip outcome (faller or non-faller) with higher accuracy. Comparing with CoM stability, limb support quotient presents higher sensitivity to predict falls occurrence. Higher hip height (48% to 50% of the body height) values and lower descending velocities during critical period are associated with a higher probability of successfully recoveries [53][13]. In [13] the same authors also conclude about the importance of trailing foot touchdown strategy: this strategy should be as fast as possible and result in the trailing foot placement posterior to the slipping foot.

Yang et al [51] after-collect slip induced perturbation data from pre-existing databases analysed the role of lower limb joints in response to a slip their moments impact in CoM and BoS related variables. These authors concluded that, in slipping limb, the moments that result in greater stability of the CoM are the knee flexor and hip extensor moments. These moments are also the ones that have the greatest impact on reducing the speed of the slip foot. Authors also refer that swing interruption is expected to be critical to prevent a fall potentially overriding preceding advantage of the stance limb during this period [51].

Yang et al [18] have also studied the threshold velocity of the COM that must be exceeded at the LO of the trailing foot relative to the BoS to prevent a backward fall during recovery from the slip in the single stance. Using a 7-link model and simulating slip conditions, the authors varied the position of the CoM relative to the BoS and concluded that a more posterior position of the CoM requires a higher initial velocity to bring the CoM to the BoS zone, with these values being even higher for slip situations. In situations where this was not possible, the study subjects had to end the swing phase early by placing the trailing leg slightly posterior to the stance leg to increase support.

Using the same computational model these authors additionally studied the reactive movements in each stance leg joint (hip, knee, and ankle) and their RoM limits during slip recoveries. These authors only analysed the period 160 ms after the slip onset as this is the period corresponding to the biomechanical response starting and it is also considered as a critical period for a successfully recovery since corresponds to the single stance phase. Regarding to the slipping limb's biomechanical response this is characterized by hip extensor and knee flexor moments allowing velocity reducing of the slipping limb. Concerning the ankle joint its plantar flexor moment is inhibited comparing with normal walking. Finally, when studying the operation limits of each joint during a slip response, knee joint stands out since, comparing with hip and ankle joints, its RoM is considerably less limited. Comparing swing and stance limb, former limb joints can tolerate at least twice as much alteration than stance limb [47].

3.3.3 Interlimb and Intralimb coordination

Martelli et al [69] also studied WBAM and SAM in response to multidirectional perturbations in order to conclude about the CNS response to different inputs and dominance effect during perturbation responses [69]. Regarding dominance influence, these authors found that the WBAM generated during biomechanical response does not depend on the subject dominance. Conversely, overall motor outcome was obtained by differently coupling body segments with respect to the perturbation side evidencing asymmetric interlimb coordination behaviour. The fact that both legs (dominant and non-dominant) appear to have different functions during steady walking, i.e., dominant limb is mainly responsible to propel the body forward whereas the main role of the non-dominant limb is to provide body support, can explain these findings [69].

These authors concluded that the perturbation direction significantly affected metrics related do compensatory step's kinematics and dynamics with diagonal disturbances being those that result in the most destabilizing effects [24][69]. Concerning the intralimb coordination, the perturbation effect results in WBAM changes revealing that the contribution of all body segments covaried more consistently during steady locomotion. WBAM modulation appears as a mechanism that leads balance regulation by properly organizing the covariation of elemental variables [69]. In another study the same authors also concluded about the existence of different compensatory's steps responses concerning the perturbation direction determining a greater reduction of the stride when diagonal perturbations were induced in the left side [24]. Joint angles and ROM also showed significant differences when considering perturbation directions. The perturbed leg hip joint ROM decreased significantly (31.1° - 48.6° vs 11.1° – 52.0°) being forward direction perturbations characterized by lower values while hip ROM of the unperturbed leg did not change significantly after perturbation despite being affected by perturbation direction and side (ROM increased with perturbation delivered toward the south direction). Also, perturbed, and unperturbed knee joints were significantly affected by perturbation side and direction (perturbed: 60.9 ± 5.0° to 37.7 ± 20.1°, unperturbed to 48.2 ± 8.9°). Moreover, perturbations delivered toward the north direction were characterized by lower values while the perturbation side only affected the perturbed limb's ROM whereby perturbations delivered toward the right side involved higher ROM. Both perturbed and unperturbed ankle RoM were affected by perturbation direction being these values higher after the perturbation (perturbed 27.0 ± 4.7° to 38.8 ± 27.6° unperturbed: to 32.8 ± 8.1°) [24].

Finally, some authors, besides evaluate and differentiate trailing and slipping limb responses also evaluated intra and interlimb coordination between both perturbed and unperturbed limbs. Moyer [14] et al characterized the trailing leg's biomechanical response to slips and studied the intralimb and interlimb coordination of both legs with simulated slip conditions. These authors classified the existence of 4 different strategies in response to a slip based mainly on the distance and duration of the swing and the orientation of the HC after initiation of the slip. All the different strategies were characterized by the early interruption of the swing phase. The authors concluded that, similarly to the leading leg, the response in trailing leg is also characterized by hip extension to potentiate the contact with the ground and knee flexion to decelerate the movement of this leg allowing energy absorption. Naturally, the response to more intense perturbations also involves higher joints torques. In addition to determining that healthy gait is decisive for the recovery strategy that the subjects naturally use, this study also allowed the authors to confirm the existence of interlimb and intralimb coordination. Regarding intralimb coordination, the authors have determined the existence of a strong correlation between knee flexion and hip extension, showing the capacity of the CNS to modulate the joint response of different joints on the same leg. In relation to interlimb coordination, the authors determined a relationship between the knee joint of the leading leg and the trailing leg strategy, which is also influenced by the severity of the perturbation. The swing phase of the trailing leg is interrupted to prevent body collapse when the flexor moment of the knee in the leading leg is not sufficient to accept the body weight's transfer [14].

Intralimb and aging modifications in this coordination were also discussed by Aprigliano et al [8] through Planar Covariation Law (PCL) analysis. PCL strategy is an intralimb coordination strategy characterized by the covariation of the elevation angles of lower limb segments during walking-related motor tasks. In further detail, when the elevation angles of thigh, shank, and foot are plotted one *versus* the others in a 3D space, they describe regular trajectory loops. If these loops are constrained close to a plane, it is possible to state that thigh, shank, and foot elevation angles do not evolve independently of each other, but they covary along an attractor plane [8][24]. Perturbation intensities were also addressed in this study.

Concerning this topic after the perturbation young subjects showed the greatest reduction of the stride, meaning a quicker response, while stance reduction was founded in older subjects. Step length and width also decreased after perturbation for both groups. Segments RoM also suffered changes in result of perturbations. Shank, thigh, and foot segments presented significantly reduced RoM. In detail the young group showed the greatest reduction in the RoM at the shank and the thigh after the perturbation onset while the shank of the unperturbed limb in the older people was significantly greater than that of the younger participants. Conversely, perturbation intensity did not affect the RoM of elevation angles [8]. Finally, the PCL of elevation angles was observed only for the UL, confirming intralimb coordination in this leg for all subjects and across all experimental conditions, while perturbed limb's segments did not lay close to a plan thus involving an altered intralimb coordination strategy being this alteration more evident for north direction perturbations in result of limb joint stiffening to prevent knee hyperextension [8], [24]. On the other hand, the remaining perturbations allow the limb joints to freely span their whole RoM, eliciting a coordination strategy comparable to that observed during unperturbed walking. While Table 3.3 and Figure 3.2, present a summary of the spatiotemporal data, Table 3.4 and Figure 3.3 address kinetic and kinematic data of the articles considered in this section.

Table 3.3 Overview of the spatiotemporal parameters during slips biomechanical response. N\A = Not Available

Study	***Unperturbed foot react time***	***Compensatory length***
[9]	**Non-Fatigue:** 159.70 ± 44.99 ms **Fatigue:** 256.30 ± 58.45 ms	N\A
[57]	**Response Time** Younger: 0.32 ± 0.01 s Older: 0.36 ± 0.01 s	N\A
	Backward Time: Intensity_1: 0.096 ± 0.010 s Intensity_2: 0.089 ± 0.013 s Intensity_3: 0.122 ± 0.015 s	N\A
	Compensatory time: Younger: 0.41 ±0.01 s Older: 0.48 ± 0.02 s Intensity_1: 0.443 ± 0.010 s	N\A

	Intensity_2: 0.430 ± 0.011 s Intensity_3: 0.462 ± 0.015 s	
	N\A	**Forward Swing Length** Young: 0.365 ± 0.010 m Older: 0.420 ± 0.011 m
	N\A	**Backward Swing Length** Intensity_1: 0.096 ± 0.010 m, Intensity_2: 0.089 ± 0.013 m, Intensity_3: 0.122 ± 0.015 m
[59]	N\A	0.34 m contralateral steps lengthen for later stance phase perturbations
[61]	**Intensity_1:** Young: 423 ± 48.8 Older: 486 ± 18.2	N\A
	Intensity_2: Young: 398 ± 21.4 Older:406 ± 69.5	N\A
	Intensity_3: Young:410 ± 26.1 Older: 460 ± 20.0	N\A
	Intensity_4: Young: 393 ± 28.7 Older: 437 ± 36.7	N\A
[25]	**Soft:** 0.35 ± 0.004 s	N\A
	Strong: 0.39 ± 0.02 s	N\A

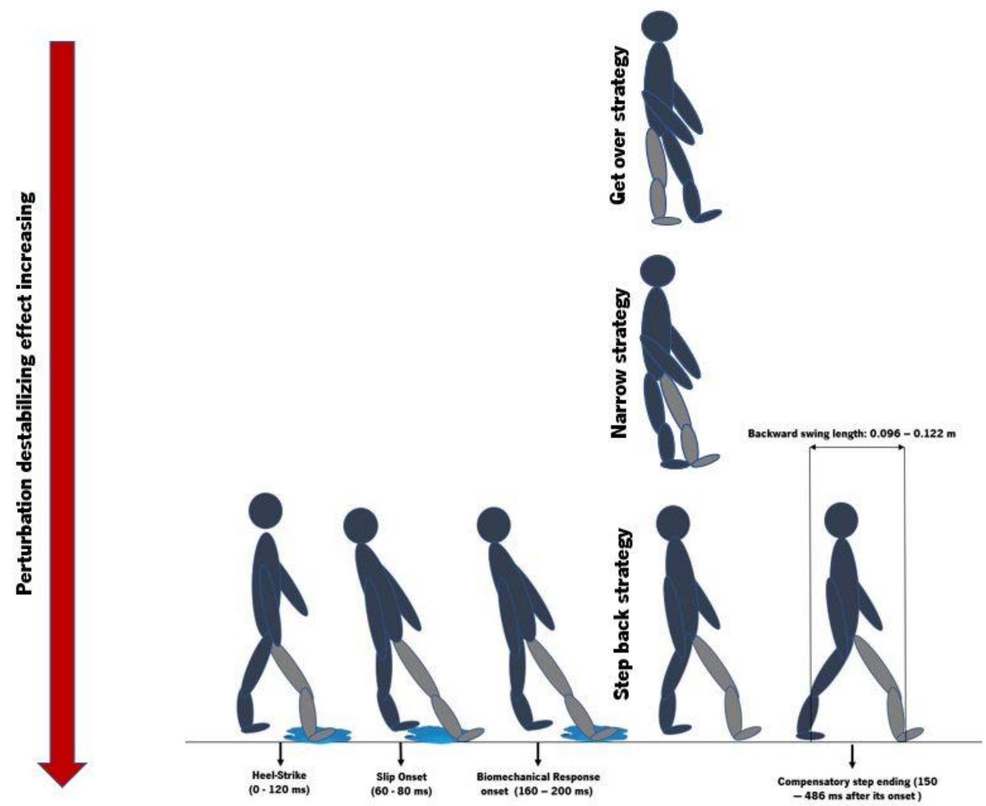


Figure 3.2 Overview of the spatiotemporal parameters during different biomechanical strategies, obtained through the state-of-the-art review. Compensatory step timings presented are related to the step back strategy. Slipping leg is presented in grey.

Table 3.4 Overview of the kinetic and kinematic parameters during slips biomechanical response. N\A = Not Available

Study	***Considered joints***	***Study Condition***	***Recovery movements***	***Time after slip (ms)***
[12]	Slipping Knee	N\A	6.7 °Extension at 87◦/s	116 ms
	Slipping Ankle	N\A	4.4 ° Plantarflexion at 86◦/s	
	Slipping Hip	N\A	18.9°/s Flexion	between 170 and 200 ms
	Trailing Knee	N\A	N\A	approx. 170 ms
	Trailing Hip	N\A	Extension	after 200 ms
[10]	Slipping Hip (+= flexion) *	HS at 120 steps/min	0.512 ± 0.202 Nm/kg	N\A
		HS at 90 steps/min	0.356 ± 0.126 Nm/kg	N\A
		TO at 120 steps/min	1.088 ± 0.423 Nm/kg	N\A
		TO at 90 steps/min	1.016 ± 0.249 Nm/kg	N\A
	Slipping Knee (+= flexion) *	HS at 120 steps/min	– 0.286 ± 0.058 Nm/kg	N\A
		HS at 90 steps/min	– 0.206 ± 0.071 Nm/kg	N\A
		TO at 120 steps/min	– 0.671 ± 0.395 Nm/kg	N\A
		TO at 90 steps/min	– 0.617 ± 0.319 Nm/kg	N\A

	Slipping Ankle (+= plantarflexion) *	HS at 120 steps/min	0.016 ± 0.013 Nm/kg	N\A
		HS at 90 steps/min	0.007 ± 0.015 Nm/kg	N\A
		TO at 120 steps/min	0.436 ± 0.157 Nm/kg	N\A
		TO at 90 steps/min	0.394 ± 0.212 Nm/kg	N\A
	Trailing Hip (+= flexion) *	HS at 120 steps/min	– 0.314 ± 0.584 Nm/kg	N\A
		HS at 90 steps/min	– 0.265 ± 0.376 Nm/kg	N\A
		TO at 120 steps/min	– 0.548 ± 0.169 Nm/kg	N\A
		TO at 90 steps/min	– 0.501 ± 0.133 Nm/kg	N\A
	Trailing Knee (+= flexion) *	HS at 120 steps/min	– 0.574 ± 0.368 Nm/kg	N\A
		HS at 90 steps/min	– 0.521 ± 0.24 Nm/kg	N\A
		TO at 120 steps/min	– 0.214 ± 0.098 Nm/kg	N\A
		TO at 90 steps/min	0.191 ± 0.076 Nm/kg	N\A
	Trailing Ankle (+= plantarflexion) *	HS at 120 steps/min	1.824 ± 0.278 Nm/kg	N\A
		HS at 90 steps/min	1.683 ± 0.188 Nm/kg	N\A
		TO at 120 steps/min	0.071 ± 0.083 Nm/kg	N\A
		TO at 90 steps/min	0.028 ± 0.039 Nm/kg	N\A
[26]	Slipping Hip Flexion	Slow velocity	Wide/back step: 18.60 ± 9.73° Narrow: 27.68 ± 4.62°	N\A
		Fast velocity	Narrow: 25.79 ± 6.67° Get over: 27.72 ± 5.58°	N\A
	Slipping Knee Flexion	Slow velocity	Wide/back step: 8.29 ± 4.86° Narrow: 6.91 ± 2.78°	N\A
		Fast velocity	Narrow: 9.25 ± 2.81° Get over: 9.38 ± 6.82°	N\A
	Slipping Ankle Flexion	Slow velocity	Wide/back step: 1.30 ± 3.69° Narrow: 0.11 ± 7.80°	N\A
		Fast velocity	Narrow: 2.86 ±4.92° Get over:2.51 ± 4.48°	N\A
	Trailing Hip Extension	Slow velocity	Wide/back step: 8.68 ± 10.08° Narrow: 14.16 ± 3.84°	N\A
		Fast velocity	Narrow: 16.32 ± 8.08° Get over: 14.90 ± 4.91°	N\A
	Trailing Knee Extension	Slow velocity	Wide/back step: 22.38 ± 20.70° Narrow: 10.62 ± 4.69°	N\A
		Fast velocity	Narrow: 11.18 ± 5.41° Get over: 17.34 ± 9.31°	N\A
	Trailing Ankle Plantarflexion	Slow velocity	Wide/back step: 6.50 ± 9.19° Narrow: 8.44 ± 2.52°	N\A
		Fast velocity	Narrow:8.84 ± 5.18° Get over:4.55 ± 5.85°	N\A
[54]	Slipping Hip (+ = flexion) *	Recovery	0.00 ± 0.34 Nm/kg – Flexion	30% stance
		Fall	-0.06 ± 0.21 Nm/kg – Extension	30% stance
		Recovery	0.51 ± 0.28 Nm/kg – Flexion	50% stance
		Fall	0.78 ± 0.01 Nm/kg - Flexion	50% stance
	Slipping knee (+ = extension) *	Recovery	160.66 ± 9.43° 0.36 ± 0.37 Nm/kg	30% stance
		Fall	165.88 ± 4.37° 0.16 ± 0.19 Nm/kg	30% stance
		Recovery	163.77 ± 6.46°	50% stance

			0.20 ± 0.24 Nm/kg	
		Fall	146.42 ± 7.75° 1.05 ± 0.15 Nm/kg	50% stance
	Slipping ankle (+ = dorsiflexion) *	Recovery	83.80 ± 3.42° 0.16 ± 0.14 Nm/kg	30% stance
		Fall	90.85 ± 2.19° 0.10 ± 0.09 Nm/kg	30% stance
		Recovery	76.18 ± 2.25° 0.57 ± 0.13 Nm/kg	50% stance
		Fall	74.82 ± 4.34° 0.06 ± 0.06 Nm/kg	50% stance
[23]	Slipping Shank (+ = flexion) *	N\A	105.77 ± 3.42° 103.97 ± 4.44°	30 to 50% (Slip Start - Slip End)
	Slipping Foot (+ = flexion) *	N\A	174.33 ± 5.57° 172.25 ± 5.92° 172.49 ± 6.66°	30 to 70% (Slip Start - Slip End)
	Trailing Thigh (+ = flexion) *	N\A	79.54 ± 4.51° 84.70 ± 7.36° 90.13 ± 10.13°	50 to 90% (Slip Start - Slip End)
	Trailing Shank (+ = flexion) *	N\A	54.75 ± 5.48° 48.30 ± 7.93° 42.87 ± 9.33° 39.95 ± 8.99°	30 to 90% (Slip Start - Slip End)
	Trailing Foot (+ = flexion) *	N\A	135.80 ± 12.82° 121.87 ± 19.31° 108.56 ± 24.22° 105.04 ± 20.90°	30 to 90 % (Slip Start - Slip End)
[29]	Trailing hip Extension	Normal Walking	≈ -0.8 Nm	N\A
		Slip	≈ -2.0 Nm	N\A
	Trailing hip Flexion	Normal Walking	≈ 0.5 Nm	N\A
		Slip	≈ 1.20 Nm	N\A
	Trailing knee Flexion	Normal Walking	≈ 0.8 Nm	N\A
		Slip	≈ 1.20 Nm	N\A
	Trailing ankle Plantarflexion	Normal Walking	≈ -0.15 Nm	N\A
		Slip	≈ -0.10 Nm	N\A

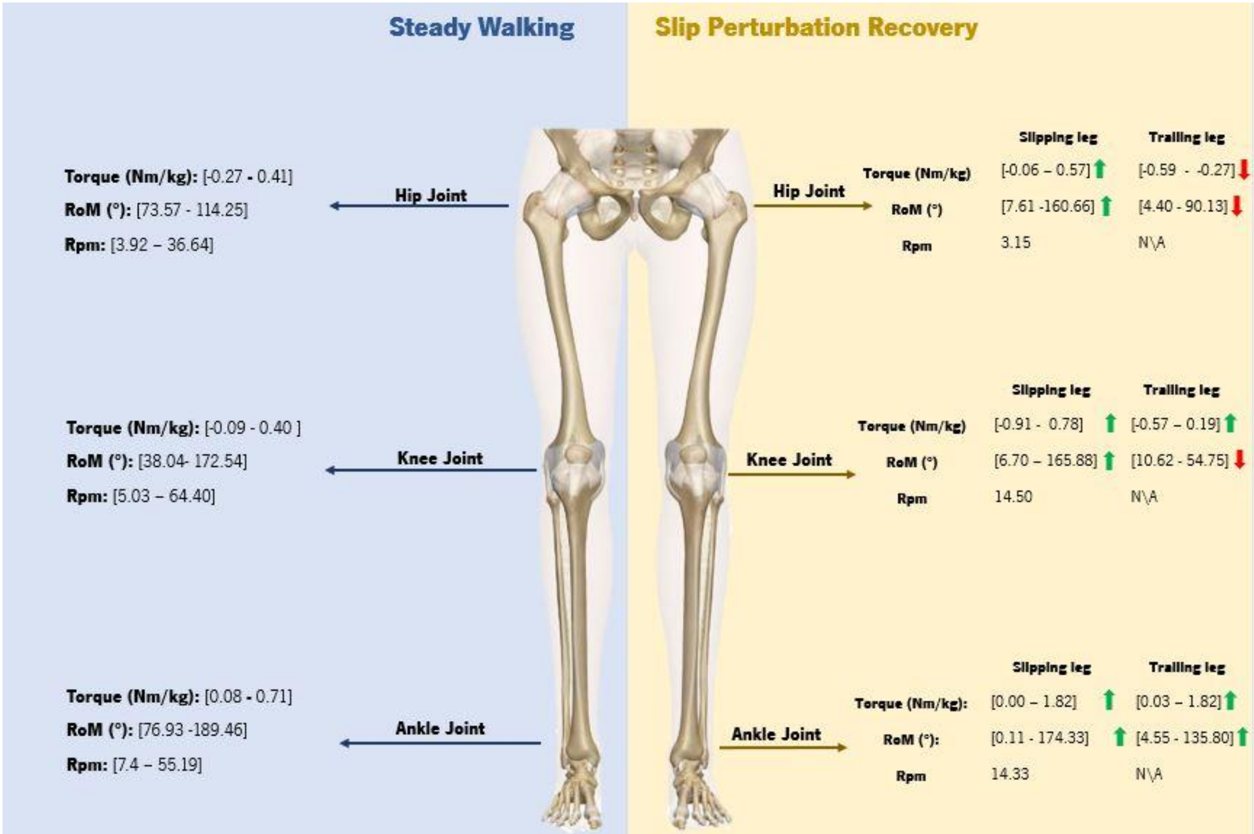


Figure 3.3 Overview of the kinetic and kinematic parameters during slips biomechanical response, obtained through the state-of-the-art review. N\A = Not Available.

3.3.4 EMG data

Another approach to study biomechanical response to slip perturbations frequently presented in literature is the study of EMG data. The study of muscle activity often concerns the study of some parameters as latency periods, activation peak and peak amplitude among others. Some authors, when analyzing muscle activity in response to a slip focus on muscle synergies analysis while others are dedicated to the comparative study of antagonists and agonists. This section will be dedicated to the analysis of the literature that uses this type of approaches to study the biomechanical response to a slip. During a slip response, Tibialis Anterior (TA), Medial Gastrocnemius (MG), Rectus Femoris (RF), Medial Hamstrings (MH), Biceps Femoris (BF) and Vastus Lateralis (VL) are the muscles mainly analysed in the literature included as they appear to play an important role in this biomechanical response [9], [16], [27], [28], [55], [65].

Xingda Qu et al [27] studied muscular activation latency, muscular peak amplitude, time-to-peak and co-contraction index of tibialis anterior (TA), *gastrocnemius medialis* (MG), rectus femoris (RF) and medial hamstring (MH) muscles of the perturbed and unperturbed limb making a comparison between subjects that fell with subjects who obtained a successful recovery. Considering these two groups the authors conclude that failed balance recovery was associated with larger muscular peak amplitude in RF of the perturbed leg and smaller muscular peak amplitude in MH of the unperturbed leg. However successful recoveries were associated to a faster response in the RF (172.8 ± 128.3 ms) of the perturbed leg. For this reason authors conclude that RF's temporal response after muscular activity onset is more important than its amplitude. The function of MH of the unperturbed leg is to place the foot on the ground posterior to the CoM, to increase the BoS. Thus, higher amplitudes of this muscle activity (4.7 ± 2.9 vs 2.4 ± 1.3) results in increasing the chances to regaining balance. Finally greater co-contraction index of TA and MG was found to increase the chances of a successfully recovery. As aforementioned these muscles are responsible for controlling the ankle movement. Due the higher degree of ankle, muscle co-contraction results in improved ankle stability.

In respect of the muscle's activation period, MH of the perturbed appear as the one with the fastest response (118 ms) explaining knee flexion and hip extension of the perturbed leg which happens immediately after the perturbation. When comparing TA and MG with RF and MH, the first group appeared to be less active after the occurrence of slips. Accordingly to Cham, et al [27] the knee and hip play a more important role to recover from slips compared to the ankle.

Chambers et al [28] also studied the activation patterns of TA, MG, MH, and Vastus Lateralis (VL) muscles in a response to a slip, only in the slipping leg. Also, these authors included in the study two groups of different ages to analyse age effect on the muscle's response to a slip. Proactive strategies considering muscle responses were also addressed as in some trials subjects were previously informed about the floor conditions. In general, these authors concluded that hazardous, i.e., slips that resulted in fall, were characterized by longer durations and reactive power compared with the non-hazardous ones, and young adults showed longer durations compared to older adults during the reactive response. Similarly, to Xingda Qu *et al* [27] the first muscle response of the perturbed leg was the MH activation (175 ms). The remaining muscle responses were TA (189 ms), MG (219 ms) and VL (239 ms). Similar findings were highlighted by Prakriti et al [63] related to the activation sequence of the same muscles: MH (≈160 ms);

MG (≈180 ms); TA (≈188 ms) and VL (≈240 ms). Thus, knee flexors were activated significantly sooner than knee extensors. A delayed activation of TA was also founded to be correlated with hazardous slips as the achievement of foot-flat which allows gait continuation is also delayed. Also, VL (involved in knee extension) delayed activation times were associated with hazardous slips as the COM is later placed over the BOS. Knee buckling is also associated with decreased extension movements and consequently with insufficient biomechanical responses. In another study, Marigold et al determined the following muscles activation sequence after an induced slip: TA (145.97 ± 13.8 ms) and BF (153.54 ± 30.2 ms), followed by RF (174.65 ± 32.7 ms) and MG (199.23 ± 95.9 ms) [19].

Comparing both younger and older groups young adults demonstrated a longer, more powerful response. Lockhart et al also found delayed knee muscular responses of older participants delaying and hindering the biomechanical response [52].

Regarding to proactive strategies, an earlier MG activation and an increased activation power of MG, VL and MH were reported. While knee muscle contraction is still uncertain as a proactive reaction, earlier activation of the MG during heel striking provides for better foot posture. This co-contraction, however, can stiffen the knee joint and slow down speed motion.

Nazifi et al [16] also studied MH, TA, VL and MG activation in response to an unexpected slip. In comparison with the literature previously mentioned, the approach of these authors was different as muscle synergies were analysed in different periods following the slip thus dividing the slip in four different physical sub-tasks. Four muscles synergies were defined based on this strategy. Simultaneously, slips with different intensities were induced to study severity influence on the analysed variables.

First muscle synergy characterized by slipping limb hip extensors, knee flexors and dorsiflexors activation (TA, VL and MH) named swing limb deceleration synergy happens in the first 100 ms after slip onset. For this reason and considering the assumption that the biomechanical response to a slip starts between 120 and 170 ms after slip onset, this synergy it is not a reactive strategy to a slip, being assumed as a normal gait process whose function is the deceleration of the swing limb for the end of this phase.

Concerning the second synergy, mainly characterized by slipping limb VL activation resulting in hip flexion, knee extension and plantarflexion, its peak of activation happens 200 ms post HS starting about 70-50 ms before. The main function of this synergy is the weight support on the slipping limb, and it also can be considered as an attempt to continue gait and the forward weight progression on the slipping limb preventing knee buckling of the unperturbed limb.

In turn, the third muscular synergy, assumed to generate the primary slip response, is mainly characterized by hip extension, knee flexion and plantarflexion of the slipping limb and knee extension of the non-slipping knee. In this response TA and MH activity increase becoming more evident between 150-200 ms after slip allowing the placement of the COM closer to the BoS. In the fourth synergy, the unperturbed limb muscles namely TA, MH and VL have greater activity. This response arises from 200 ms after slip onset and through hip extension and knee flexion of the unperturbed limb allows the placement of the trailing foot in the ground (toe-touch strategy) increasing the BoS.

Regarding to slip severity, although the biomechanical response is very similar, the subjects who suffer the less severe slips activated their muscle synergies faster than the severe slips group. Namely synergies two and four are activated earlier in the first situation.

Sawers et al [66] also analysed the muscle response considering the muscle synergies involved in this response [66]. In general, the number of muscle synergies recruited during slip trials was significantly smaller among participants who fell compared with those who recovered evidencing decreased motor performance. Also, muscle activity complexity was lower in participants who fell [66]. These participants presented also an all-on muscle synergy characterized by extensive coactivity across all bilateral muscles. Concerning the participants who recovered successfully, presented synergies that control the knee suggesting the importance of both knees control during slip response to allow the BOS restoration and generating the knee extension torque required to maintain body weight support. Participants who fell were uncapable to initiate sufficient knee flexion to return the leading foot underneath their CoM.

O'Connell et al [55] when studying slip severity effects on trailing limb's muscle activation also concluded that in more severe slips trailing limb´s VL time-to-peak is slower comparing with less severe slips. However higher VL magnitudes are associated with more severe slips. Regarding to MH, more severe slips were associated to faster activation times, but peak magnitude and time-to-peak of these muscles were not modulated by slip severity. Finally, similarly to [23] and [19], O'Connell also found that trailing limb hamstrings and VL muscles activate sooner with increasing slip severity (MH: 162 ± 29 ms and VL: 164 ± 25 ms). Compared to some previous literature (namely [27] and [28], the fact that trailing limb reactive muscle responses appear earlier than the slipping limb reactive responses highlights the importance of trailing limb touchdown strategy which is caused by MG and VL activation. In turn, earlier activation of the trailing limb MH occurring with increasing slipping severity may contribute to one toe clearance of the trailing limb or eccentric deceleration of the forward slipping motion of the stance limb while VL contributes to swing limb knee deceleration and limb extension to step preparing [55]. Also, these authors found no significant differences between two ages groups in the variables analysed during slip responses.

3.3.5 Fatigue influence in slip perturbations' biomechanical response

As aforementioned some literature addresses muscular fatigue effects on slip responses. During these studies, a fatigue session was firstly induced using isokinetic exertions. Parijat et al [9] studied the effects of hamstrings (RF, VL and Vastus Intermedius (VI)) fatigue in slip responses. These muscles are mainly responsible for extension and flexion movements of the knee. During slip trials, the results indicated that participants exhibited a later slip stop compared with non-fatigue slip trial. The period from trailing foot onset to foot down was slower in fatigue slip trials causing a delayed reaction to increase the BoS, thus delaying the whole biomechanical response. For this reason, slip distances in fatigue trials were greater and consequently the slips were more severe. In turn, more severe slips are associated with higher knee joint moments and higher knee power. Increased slip distances can also be explained by the increased time taken to reach the peak joint moment in the slipping leg (60% of the stance phase in fatigue trials vs 40% in non-fatigue trials). Although it takes longer the joint moments are higher in fatigue trials. This finding can be explained by the recruitment of other muscles. During slip recovery knee moments are extensor dominant [9][15]. Ling Lew et al [67] also addressed fatigue effects on biomechanical response to slip perturbations. In this study greater ankle plantarflexion and Slip distances I (Anterior-posterior distance travelled from slip-start to mid-slip) and Slip distances II (Anterior-posterior distance travelled from mid-slip to slip end) were variables associated to the fatigue condition. Increased ankle plantarflexion, which could be associated to TA fatigue, results in decreased foot-floor angle, being a postural attempt to reduce the likelihood of slip initiation. Increased SDI could be explained by proprioceptive degradation in result of muscular fatigue while increased slip distance II (more evident in lower limbs fatigue comparing with upper limbs fatigue) can be explained by an insufficient reactive recovery response and further lead to a hazardous slip, while upper-limb fatigue would not adversely affect reactive recovery response after a slip [67]. Increased slip distances were also associated to older subjects' response to a slip. In addition to fatigue related effects, Lockhart et al [52] found longer SDI and SDII for older subjects comparing with a younger group underlining the increase severity in elderly people due delayed corporal response. In this study HC velocity and accelerations during HS were also found to be higher in the older group contributing to a more severe slip requiring a more effective biomechanical response.

3.3.6 Muscular strength influence in slip perturbations' biomechanical response

Knee strength is also a variable related to muscular response during slip perturbations that appears in Sarah's et al study [58]. These authors studied the relationship between knee flexion/extension muscular strength and slip severity. Also, knee kinematic variables analysis was an outcome of this study. Concerning the kinematic biomechanical response these authors refer that, after slip onset, knee response can be divided in two distinct periods with different movements. Firstly, in the first 130 ms after slip onset knee response is mainly characterized by knee flexion. Knee flexion is associated with slipping foot deceleration reducing fall risk. Knee extension is prevalent between 130 ms and 180 ms which is responsible to prevent knee buckling, limb loading, trunk support and allows normal gait resumption. These findings are in accordance with [43] and [51]. Regarding knee muscular strength influence in slip severity subjects with lower extension/flexion rate torque development and knee extension peak torque values experienced more severe slips (higher Peak slip velocity values). These authors suggest that, according to their findings, knee extension peak torque plays a more critical role comparing with knee flexion and appears to require more strength comparing with the first biomechanical response [58]. Reduced knee flexion strength was also associated to severe slips in older people in [52]. Table 3.5 and Figure 3.4 summarize the EMG variables analysed in all the articles considered.

Table 3.5 Overview of the EMG variables analyzed during slips biomechanical response. (N\A = Not Available)

Study	***Considered muscles***	***Latency (ms)***		***Time to peak activation (ms)***		***Peak Amplitude ****	
		Hazard	**Non-hazard**	**Hazard**	**Non-hazard**	**Hazard**	**Non-hazard**
[27]	Slipping Limb TA	260.3	258.8	211.7	218.5	2.6	2.4
	Slipping Limb MG	294.7	273.8	319.2	196.5	1.8	1.6
	Slipping Limb RF	247.4	172.8	389.2	216.3	6.6	4.1
	Slipping Limb MH	152.2	118.8	321.3	297.5	4.0	2.7
	Trailing Limb TA	358.4	315.1	237.9	156.6	2.52	3.1
	Trailing Limb MG	233.2	160.5	169.4	233.7	2.0	1.8
	Trailing Limb RF	351.3	268.1	203.4	188.1	6.8	7.2
	Trailing Limb MH	321.3	314.3	286.8	165.0	2.4	4.7
[28]	Slipping Limb VL	Young:216 Older:326	Young:157 Older:204	N\A	N\A	N\A	N\A
	Slipping Limb MH	Young:139 Older:221	Young:146 Older:167	N\A	N\A	N\A	N\A
	Slipping Limb TA	Y:151 O:228	Y:156 O:209	N\A	N\A	N\A	N\A
	Slipping Limb MG	Y:151 O:280	Y:165 O: 235	N\A	N\A	N\A	N\A
[55]	Trailing Limb VL	N\A	164 ± 25	N\A	N\A	N\A	N\A
	Trailing Limb MH	N\A	162 ± 29	N\A	55.6 ± 21.6	N\A	0.88 ± 0.60
[19]	Slipping Limb RF	N\A	174.65 ± 32.7	N\A	N\A	N\A	———
	Slipping Limb BF	N\A	153.54 ± 30.2	N\A	N\A	N\A	5.87 (prior experience)
	Slipping Limb TA	N\A	145.97 ± 13.8	N\A	N\A	N\A	5.90 (prior experience)
	Slipping Limb MG	N\A	199.23 ± 95.9	N\A	N\A	N\A	7.35 (prior experience)
[16]	Slipping limb TA, VL and MH	N\A	N\A	N\A	First 100	N\A	N\A
	Slipping limb VL	N\A	N\A	N\A	Between 130 and 170	N\A	N\A
	Slipping limb TA and MH	N\A	N\A	N\A	150-200	N\A	N\A
	Trailing limb	N\A	N\A	N\A	After 200	N\A	N\A

	TA, MH and VL						
[63]	Trailing Limb MG	N\A	188 ± 33.66	N\A	335 ± 25.50	N\A	N\A
	Trailing Limb TA	N\A	197 ± 22.23	N\A	312 ± 33.96	N\A	N\A
	Trailing Limb MH	N\A	155 ± 11.76	N\A	250 ± 13.96	N\A	N\A
	Trailing Limb VL	N\A	238 ± 23.54	N\A	365 ± 25.35	N\A	N\A
[45]	Slipping Limb BF	N\A	167 ± 83.7 191.4 ± 114.8	N\A	N\A	N\A	1.40 ± 0.50 1.35 ± 0.45
	Slipping Limb VM	N\A	87.3 ± 27.9 88.0 ± 29.4	N\A	N\A	N\A	1.73 ± 0.46 1.65 ± 0.45
	Slipping Limb MG	N\A	336.7 ± 148.1 347.8 ± 141.1	N\A	N\A	N\A	0.98 ± 0.28 0.93 ± 0.25
	Slipping Limb TA	N\A	139.1 ± 45.0 136.5 ± 33.5	N\A	N\A	N\A	3.13 ± 1.58 3.09 ± 1.35

*Normalised values

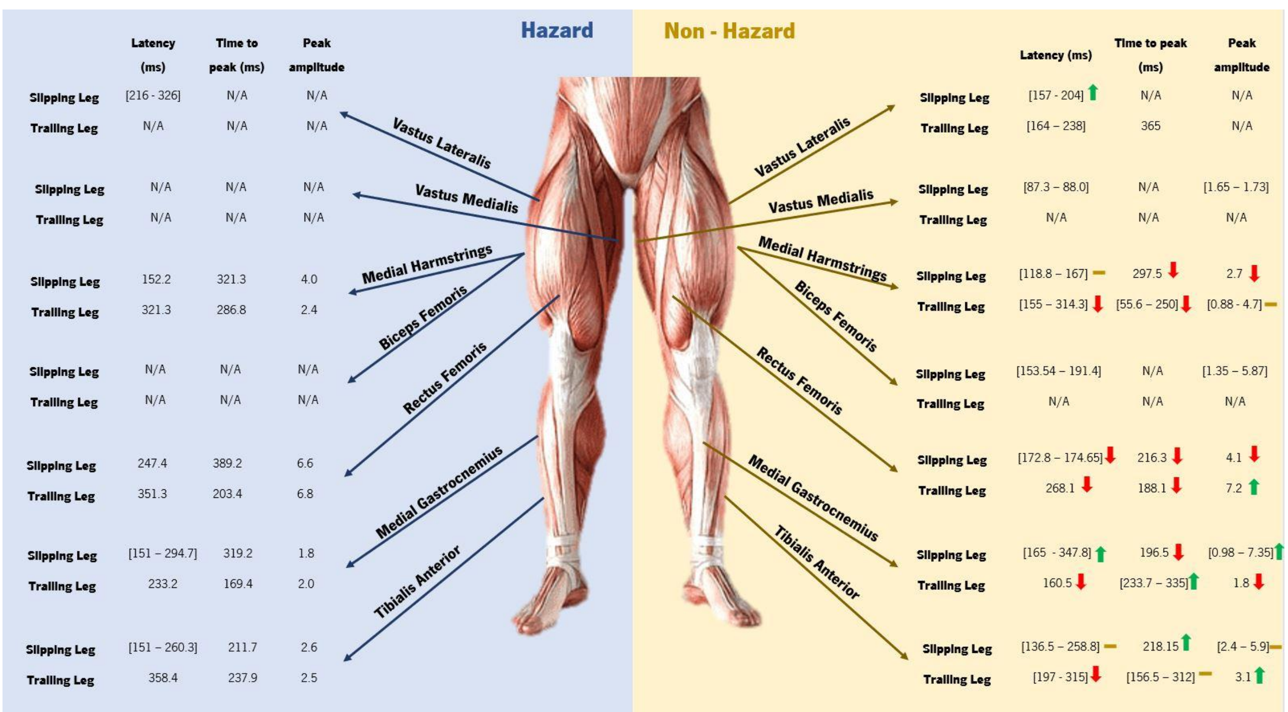


Figure 3.4 Overview of the EMG variables analysed, in the scientific literature included, during slips biomechanical response.
N\A = Not Available.

3.3.7 Repetitive training

Repetitive training is frequently presented in a literature related with slips as a strategy with promising outcomes to slip prevention [22][47][58][59][62]-[64]. Although, in some cases, the metrics are not directly related to the biomechanical response to induced slip perturbations, it is possible to extract information about this topic. This section will include the conclusions of interest from the articles related to repetitive training that relate to the biomechanical response.

Sakai et al [45] analysed mainly change of EMG parameters due slip adaptation. Also, CoM acceleration was analysed: as the trials progressed, the performance of the subjects improved as the forward and backward accelerations decreased indicating a smaller body's sway caused by the perturbation. With regard to the EMG magnitude ratio values (IEMG Ratio: perturbed EMG values divided by normal EMG values) TA had the greatest reaction followed by VM, as presented in Table 3.5. The lowest reaction then pertains to GM, which is below one and shows that the perturbed IEMG was smaller than the baseline IEMG. In the second part of the trials, perturbed IEMGs of VM and GM were dramatically reduced, whereas normal IEMGs were not. Regarding the TA, IEMG ratio was not significant between the two halves but both normal and perturbed EMG magnitudes were higher in the first phase. These findings suggest that, rather than the latency the muscles activations amplitudes affect the motor adaptation for postural control. Based on these results, the authors refer that the first slip response is characterized by a hyper-reaction of VM and GM. This hyper-reaction resulted in stiffer joint moments hindered quick reactions thus resulting in a greater body sway. Also, the reduced GM response in the second phase allowed a more effectively TA and consequently ankle response [45].

Adaptation relevance of the relative movement between CoM and BoS was also addressed by other authors as an important proactive strategy to slip avoidance [56][60]. Forward trunk inclination reduced braking impulse and shorter steps were adaptations that result in a more beneficial CoM/BoS relationship to slip recovery. Foot-flat strategy was also associated to post slip adaptations which results in reduced BoS displacement and velocity [56]. Concerning the step lengths, Debelle et al [60] concluded that keeping the step length close to normal levels was an important component of balance recovery as it can compensate larger CoM displacements.

A flat foot landing, improved knee flexion, and decreased propelling power in the contralateral limb might have all contributed to significant reductions in the sliding limb braking impulse, resulting in minimal to no BoS displacement. [56], [68] Knee flexion was also highlighted as a proactive strategy developed by the participants in [60], [68]. Skateover and walkover strategies were also referred as two of the slip outcomes. Regarding their characteristics, in the walkover strategy the pre-slip propulsive impulse under the contralateral limb, coupled with a flat foot landing and increased knee flexion, could have led to greater decreases in the slipping limb braking impulse allowing very minimal to no BoS displacement. In turn, skateover strategy was associated to greater pre-slip propulsive impulse until contralateral toe-off which could have carried the CoM through, catching up with the sliding BoS without causing backward balance loss [56].

Prakriti et al [63] also studied repetitive training effect in older adults. SDI and SDII were found to be lower comparing training and control group. Also, Peak Slip Heel Velocity showed greater reduction for training group. All these variables demonstrate slip severity reduction after repetitive training. Regarding kinematic variables, successful recoveries relied on increased peak ankle plantarflexion, knee flexion and hip flexion. Peak knee angular velocity also decreased in the training group. The time to reach trunk and hip peak angular velocities was also reduced in the training group. Comparing with control group EMG, MH and TA response was quicker in the training group. An early activation of MH may help in stabilizing the knee joint and assist in slip recovery process. Peak knee and ankle coactivity were also reduced in the training group increasing net joint torque and decreasing energy expenditure in these joints. Concerning the compensatory step it was also reduced in the training group (110 ± 19.9 ms vs 150 ± 29.8 ms) showing a quicker recovery process initiation. Finally, the transitional acceleration of the whole-body CoM increased more in the training group compared to control [63]. The training group was able to quickly reverse trunk extension as compared to the control. Reducing forward trunk rotations are believed to have a significant effect in bringing the CoM of the body within stability limits. CoM position and velocity relative to the BoS was also addressed by Xuan Liu et al [64] as an improved outcome after slip training [64]. Reduced recovery step length and greater knee flexion led to changes in the CoM position and BoS velocity, which in turn raised the CoM velocity that was related to it [64]. While studying repetitive training retention in older adults, Bhatt et al [64] also found increased CoM stability relative to the BoS as a strategy to fall prevention [64], [68]. Also increased hip height (limb support) enhanced the success of recovery from slip by providing adequate limb support even under low-friction conditions, enabling subjects to maintain natural progression in stepping with their contralateral limb [11], [70].

Finally, Marigold et al [19] besides studied adaptive responses also studied the prior knowledge effects on slips biomechanical response. Considering the trials where the subjects had no prior knowledge of the conditions, recovery adaptation was found after repeated exposures. These adaptations were characterized by an attenuated excitatory response of BF and TA allowing knee and ankle flexion resulting in body lowering toward the BoS. while the MG response became inhibitory, breaking impulse decreasing, acceleration impulse gradual increasing, reduced foot angle and CoM elevation. Also, swing limb interruption strategy which was seen in the first slip response vanished after some slips exposure. Concerning the proactive strategies in situations with prior knowledge of the surface, these strategies are mainly characterized breaking impulse and RoL reducing with the latter allowing and approximation of the Centre of Pressure (CoP) from the contralateral limb's BoS resulting in breaking impulse reducing. Also, foot angle reducing was associated to these trials indicating a flatter foot strategy to facilitate the breaking impulse reducing and increasing the contact area. In prior knowledge trials, a higher CoM position that is likewise closer to the contralateral limb BoS was found to have a larger MoS

This study highlights the importance of an adequate CoM position in response to slips and evidence the role of the accelerations involved in the foot-floor contact. The adaptive and proactive responses allow to conclude that subjects were conscious that a large propulsive force on a slippery surface would further increase the risk of fall. Subjects in this study also adopted the skateover strategy also addressed in [56] when slippery surface conditions were informed [19]. Table 3.6 present a summary of the spatiotemporal data and Table 3.7, address kinetic, kinematic, and spatiotemporal data of the articles considered the repetitive training studies.

Table 3.6 Overview of the spatiotemporal parameters during slips biomechanical response after repetitive training. N\A = Not Available

Study	*Unperturbed foot react time*	*Compensatory length*
[63]	**Control:** 150 ± 29.8 ms	N\A
	Training: 110 ± 19.9 ms	N\A

Table 3.7 Overview of the kinetic and kinematic parameters during slips biomechanical response after repetitive training where N\A = Not Available

Study	*Considered joints*	*Study Condition*	*Recovery movements*	*Time after slip (ms)*
[60]	Slipping knee	Extension	between 0.4 and 0.8 Nm/kg	23 to 79 % stance
	Slipping ankle	Plantarflexion	between 1.4 and 1.6 Nm/kg	37 to 90% stance phase
[63]	Slipping hip (+ = flexion) *	Slip 1	12.44 ± 3.96°	N\A
		Slip 2	7.61 ± 2.45°	N\A
	Slipping knee (+ = flexion) *	Slip 1	25.63 ± 5.50°	N\A
		Slip 2	18.04 ± 3.68°	N\A
	Slipping ankle (+= plantarflexion) *	Slip 1	108.60 ± 5.34°	N\A
		Slip 2	103.38 ± 4.23°	N\A
[15]	Slipping knee Extension	N\A	approx. 0.4 rad between 2 and 4 Nm/kg	N\A
[14]	Slipping Hip (+ = extension) *	MIN**	-30.4 ± 2.6° 0.55 ± 0.19 Nm/kg	N\A
		FF**	-29.3 ± 5.0° 0.57 ± 0.18 Nm/kg	N\A
		MID**	-31.4 ± 6.2° 0.56 ± 0.17 Nm/kg	N\A
		TD**	-29.3 ± 4.5° 0.13 ± 0.0 Nm/kg	N\A
	Slipping knee - (+ = flexion) *	MIN**	25.8 ± 4.8° 0.78 ± 0.23 Nm/kg	N\A
		FF**	19.6 ± 2.9° -0.74 ± 0.17 Nm/kg	N\A
		MID**	26.0 ± 4.8° -0.90 ± 0.23 Nm/kg	N\A
		TD**	23.9 ± 2.5° -0.91 ± 0.07 Nm/kg	N\A
	Slipping ankle - (+ = dorsiflexion) *	MIN**	-3.9 ± 2.3° 0.00 ± 0.06 Nm/kg	N\A
		FF**	- 7.5 ± 3.1° 0.22 ± 0.11 Nm/kg	N\A
		MID**	-5.4 ± 2.8° 0.17 ± 0.10 Nm/kg	N\A
		TD**	-5.1 ± 3.8° 0.08 ± 0.08 Nm/kg	N\A

Trailing hip - (+ = extension) *	MIN**	-5.4 ± 4.3° -0.55 ± 0.13 Nm/kg	30% stance phase
	FF**	-5.2 ± 7.5° -0.55 ± 0.15 Nm/kg	30% stance phase
	MID**	-4.4 ± 6.0° -0.51 ± 0.15 Nm/kg	30% stance phase
	TD**	-4.9 ± 1.0° -0.44 ± 0.12 Nm/kg	30% stance phase
Trailing knee - (+ = flexion)	MIN**	53.4 ± 4.2° -0.21 ± 0.08 Nm/kg	N\A
	FF**	53.1 ± 5.4° -0.16 ± 0.11 Nm/kg	N\A
	MID**	55.6 ± 4.1° -0.18 ± 0.07 Nm/kg	N\A
	TD**	56.5 ± 4.5° - 0.16 ± 0.06 Nm/kg	N\A
Trailing ankle - (+ = dorsiflexion)	MIN**	-13.6 ± 4.6° 0.08 ± 0.04 Nm/kg	N\A
	FF**	-11.7 ± 7.8° 0.08 ± 0.08 Nm/kg	N\A
	MID**	-15.4 ±6.1° 0.08 ± 0.04 Nm/kg	N\A
	TD**	-14.2 ± 7.3° 0.08 ± 0.05 Nm/kg	N\A

*Joint moments signal convention used by authors of the corresponding study. ** Recovery strategies addressed in [14] based on the distance and duration of the swing and the orientation of the HC after initiation of the slip.

3.4 Research Questions Discussion

3.4.1 What are the biomechanical responses of the lower limbs during a slip event? What roles do the trailing and slipping leg distinctly play?

The analysis of kinematic and kinetic variables, allowed to conclude about hip, knee, and ankle movements in response to a slip. In the analysed literature, the movements in the sagittal plane are the most studied, so conclusions about the extension and flexion movements of the three previously mentioned joints are obtained. Additionally, the analysis of this data also distinguishes the different roles of the slipping and trailing leg in response to a slip event. The analysis of these variables should consider the muscle latency period, i.e., between 150 and 190 ms after slip onset in order to include only the movements that are effectively related to the biomechanical response [54].

The role of the trailing leg is highlighted in most of the literature included in this state-of-the-art review. Hip extension and knee flexion movement interrupts the swing phase earlier, allowing this leg to be placed in contact with the ground, thus increasing the BoS and also contributing to the deceleration of its movement allowing energy absorption - trailing leg response strategy [12], [14], [23], [26], [29], [56], [62], [65]. Only two of the articles in the literature reviewed referred to hip flexion as a trailing leg response [10], [23]. Despite this fact the trailing leg touch-down is highlighted in [10]. The literature study also demonstrates that this reaction is not always present during slip occurrences since the body has the capacity to modify it based on the perturbation's intensity and the speed of the gait at the time the perturbation occurs. The individuals have occasionally not needed to step back the trailing foot when walking at faster speeds (140 bpm defined with a metronome) defined in [26]. In these situations, trailing foot was placed in a position very close to the leading foot i.e. the narrow strategy, or, in other cases subjects were able to continue walking [26]. These findings are justified by the fact that when subjects walk at a faster speed, their movement velocity is equal or greater than the maximum slip velocity (defined as 1.6 m/s) being easier to overcome the perturbation [26].

Perturbation intensity also influences time and length of the compensatory step (Table 3.3). More severe perturbations are associated with longer compensatory steps [6][61]. The age is also a factor that influences the compensatory step (subsection 3.4.3) [25]. Additionally, the instant in which the perturbation is given also influences this trailing limb's response: when the perturbation is given in a later stance phase, i.e., after the first third of this phase, the perturbation is less severe including less risks, is less variable and increases the chance of a more effective biomechanical response as the contralateral limb is in an advantageous position to perform the toe touch strategy.

Regarding the behaviour of the slipping leg during slip responses, foot and knee velocity changes are a consequence of the perturbation and are critical in triggering the initial postural response [12]. In kinematic terms, the response of this leg can be divided into 2 distinct responses: between 25% and 45% and between 45% and 55% of the stance phase. In the first period the hip and knee responses are characterized by extension and flexion respectively, while in the second period hip flexion and knee extension are emphasized [54]. Hip extension, knee flexion, to allow deceleration of this leg, and flat-foot are also strategies highlighted in [23], [47], [51], [52].

Despite presenting different movements and playing different roles in the slip response, there is coordination between both legs during this period: The swing phase of the trailing leg is interrupted to prevent body collapse when the flexor moment of the knee in the leading leg is not sufficient to accept the body weight's transfer [14]. Still regarding intersegmental coordination, WBAM is stabilized by the CNS regardless of gait patterns. Also the WBAM generated during the biomechanical response does not depend on the subject's dominance but, conversely, the global motor outcome was obtained by different coupling body segments in relation to the slipping side evidencing an asymmetric interlimb coordination behaviour. [58] [50], [69]

Finally, limb support measured through hip height is also a variable that appears with some frequency in the literature considered: higher hip height values (48% to 50% of the body height) and lower descending velocities during critical period, i.e., between the slip onset and the instant prior to the trailing limb touch down, are associated with a higher probability of successfully recoveries [11], [13], [53], [70].

3.4.2 Which muscles of the lower limbs are involved in the recovery process? What are their activation times?

Concerning the articles considered, the electromyographic activity of TA, MG, RF, MH, BF, VL and VI was analysed. The analysis of certain parameters like the latency period, time to peak activation or peak amplitude allows to draw conclusions about the relevance of the considered muscles in the biomechanical response to a slip.

In general, hazardous slips were characterized by longer durations and reactive power compared with the non-hazardous ones [16]. Also, the number of muscle synergies recruited during slip trials was significantly smaller among participants who fell compared with those who recovered [28], [66].

Considering the articles included in this review, the MH appear to intervene quite actively during slip recoveries. Its activation in the perturbed limb explains hip extension and knee flexion, while its activation in the trailing limb allows the trailing foot touch down, strategy already discussed [16], [27], [28], [52], [55], [63]. For more severe slipping situations MH activation time is shorter [55].

Although in some of the literature analysed it was considered as a passive joint or of less importance in the recovery process, some authors also highlight the muscular activity of the muscles that control the ankle joint [27], [52], [63]. The co-contraction of TA and MG is associated to successful recoveries, as these muscles increase the stability of this joint, allowing the reduction of the ankle angle (flat foot) and helping the contact with the ground [27], [63].

Regarding the VL, whose function is related to the knee extension movement and swing limb deceleration, this muscle assists the knee joint secondary response allowing an adequate relative position between COM and BOS avoiding also knee buckling [63]. The activation time of this muscle, similar to the MH, is shorter for more severe slips [55]. As previously mentioned, knee secondary response is mainly characterized by extension and it was also confirmed with EMG data in [15], [26], [58].

Analyzing the temporal responses (Table 3.5) the MH of the slipping limb stands out as the first body response to a slip event in all analysed articles (with a latency from ≈ 120 to ≈ 160 ms). After the activation of this muscle, the activation of the MG (after 294.7 and 233.2, respectively for slipping and trailing leg) and TA (after 258.8 and 315, respectively for slipping and trailing leg) allows ankle joint stabilizing during foot floor contact. In addition to these muscles, the activation of the RF is also highlighted in some articles as one of the first responses of the slipping limb. Also, VL activation, usually associated to longer latency periods, confirms that this muscle intervenes in the secondary response of the knee characterized by its extension. Regarding the unperturbed limb, the response of the MH is also highlighted to allow the trailing foot to contact the ground, thus increasing the BoS. This is also one of the first decisive responses of both limbs in slip situations. Once again, the ankle control by the TA and MG also participates in this interruption [16], [19], [27], [28], [45], [55], [63].

With regard to articles where the effect of fatigue and muscle strength on the response to a slip is analysed, muscle fatigue or muscle strength degradation are variables that result in more severe slips as a result of a delayed and insufficient response [9], [15], [52], [58], [59], [67].

3.4.3 How does age influence this response?

As these events have a higher incidence in elderly people, the study of the biomechanical response to a slip event considering the age factor is also deeply approached in the analysed literature [8], [28], [52], [57], [58], [61]. In these references previously mentioned, differences were found regarding the biomechanical response between young and elderly subjects. In [57] it was concluded that the older adults' response is less effective when compared with younger subjects being associated, for all perturbation intensities studied, to a reduced MOS indicating a more destabilizing effect. Additionally, the compensatory step time in adults is longer, thus indicating a slower response and a more severe slip, associated with higher SDI and SDII [52], [57], [61]. The time to produce a motor response after a slip was also analysed in [52] confirming that the response in older adults is longer. Despite these differences reported in the literature, in [26] the authors concluded that age is not a factor that influences PCL.

Considering the EMG data, younger adults' response is longer compared to older adults, thus presenting a powerful response. Reduced muscle strength in the muscles controlling the knee joint as a result of ageing is associated with more severe slips in older adults, again highlighting a reduced effectiveness of the biomechanical response in this age group [58].

3.4.4 What happens to the relative motion between BoS and COM during slip perturbations and the respective response? – Spatiotemporal data

The relative movement between BoS and CoM is also addressed quite frequently. The relative position between these two variables allows the determination of unbalanced situations and is therefore often addressed. After slipping leg's HS, the forward velocity of CoM in respect to BOS increases being reduced after the contralateral leg toe-off. When a successfully slip recovery happened, CoM and BoS velocities are approximately the same, so the CoM was brought forward which does not happen in falls situations [10]. These spatiotemporal variables are influenced by perturbation intensities as well. A lower MoS during the slip response was related to higher perturbation intensities [25].

3.4.5 Which are the repetitive training effects in the variables previously discussed?

The bibliography related to the repetitive training allows to take conclusions about the importance of modelling the previously discussed variables (EMG, kinematics, kinetics and spatiotemporal) after successive perturbations.

Regarding the EMG data, the response to a first slip, compared to the following responses, seems to be characterized by a hyper-reaction of some of the muscles (namely GM and VL). The subsequent responses with lower magnitudes facilitate the action of other muscles, namely the TA, associated to ankle control [45]. Earlier MH activations and reduced peak knee and ankle coactivity after repetitive training increased net joint torque with a decreased energy expenditure in these joints and are thus associated with a more effective response [63]. In relation to the spatiotemporal variables, forward trunk inclination reduced braking impulse and shorter steps were adaptations that resulted in a more beneficial CoM/BoS relationship to slip recovery [56], [60]. Reduced propulsive force in the contralateral limb, flat foot landing and increased knee flexion, are other proactive strategies that allow very minimal to no BOS displacement by reducing the breaking impulse [56], [68].

3.5 Chapter Conclusions

This chapter analysed the current state-of-the-art related to the biomechanical response to slip perturbations. The analysis performed considered multivariate data including kinetic, kinematic, EMG and spatiotemporal data. The analysis of all these types of data allowed a comprehensive characterization of the response to a slip perturbation. In terms of joint movements, the role of the slipping limb in the deceleration of that limb and the role of the trailing foot in the early interruption of the swing phase were highlighted. The analysis of the EMG data allowed corroboration of the function performed by both legs and also allowed concluding which muscle groups actively intervene in this process. Finally, the analysis of the spatiotemporal parameters allowed highlighting the importance of the contralateral step in the restoration of an adequate relationship between CoM and BoS thus contributing for a successfully recovery.

As it was possible to determine throughout the present literature analysis, the conjugation of EMG, kinematic, kinetic, and spatiotemporal data provides different complementary information so that future work to be developed in this dissertation should include the analysis of all these parameters. Also, the ranking of the various parameters analysed to understand those that most influence the slip outcome should also be considered in the upcoming tasks of this dissertation. Finally, the dominance effect during slip responses should be further studied.

4 Project Conceptual Design

This chapter describes the conceptual design of this dissertation. In addition to gathering a set of crucial theoretical knowledge for the practical chapters that will follow, the previous analysis of the state-of-the-art made it possible to identify some gaps regarding the analysis of the biomechanical response to slip perturbations and its significance in the design and target specifications definition for robotic devices to mimic this human response. Thus, the research developed in this dissertation is divided into two main phases that will be further explored in this Chapter.

4.1 Introductory Insight

WHO data highlight the high incidence of falls worldwide and simultaneously warn that this incidence may increase in the upcoming years because of the world's aging population. In addition to socio-economic problems and injuries among the elderly people, falls cause a long-term sense of fear that negatively affects the elders from performing daily tasks, resulting in autonomy and physical capacity reduction, and thus increasing the risk of future falls. Hence, there is a need to develop alternative technological solutions to reduce this problem, while improving the quality of life. Given the previous problem, this dissertation aims to contribute to the study of the biomechanical response to slip disturbances through the analysis of kinematic, kinetic, spatiotemporal and electromyographic experimental data, previously collected in *Birdlab* – University of Minho [72]. Along with providing a thorough understanding of this biomechanical response, the multivariate experimental data analysis used in this dissertation also made it possible to define the target specifications in a quantitative and objective manner for use in the development of technological solutions that address this issue. Obtaining quantitative and objective data, as a result of the analysis of slip induced perturbations, capable of being used as inputs in the process of choosing the action strategy will further enhance the promising results associated with these robotic devices designed for fall prevention. Additionally, the work developed has not the objective to select an actuation strategy or articulation for falls prevention but rather to gather a vast set of information of interest for the design of fall prevention devices regardless of where the actuation is desired. Thus, scientific outputs not only for the development of this dissertation but also for other works under development in the same field can be produced. In addition to this, the goal is to rank the lower limb joints according to how well they function and how crucial they are during slip recoveries.

4.2 Project Phases

To address the previously issues in the field of elderly falls prevention, this dissertation is divided into two main points: i) Biomechanical analysis of slip induced falls previously collected at Birdlab – University of Minho; and ii) Target specifications definition based on quantitatively data obtained in phase i). Figure 4.1 presents the phases of this dissertation, schematically.

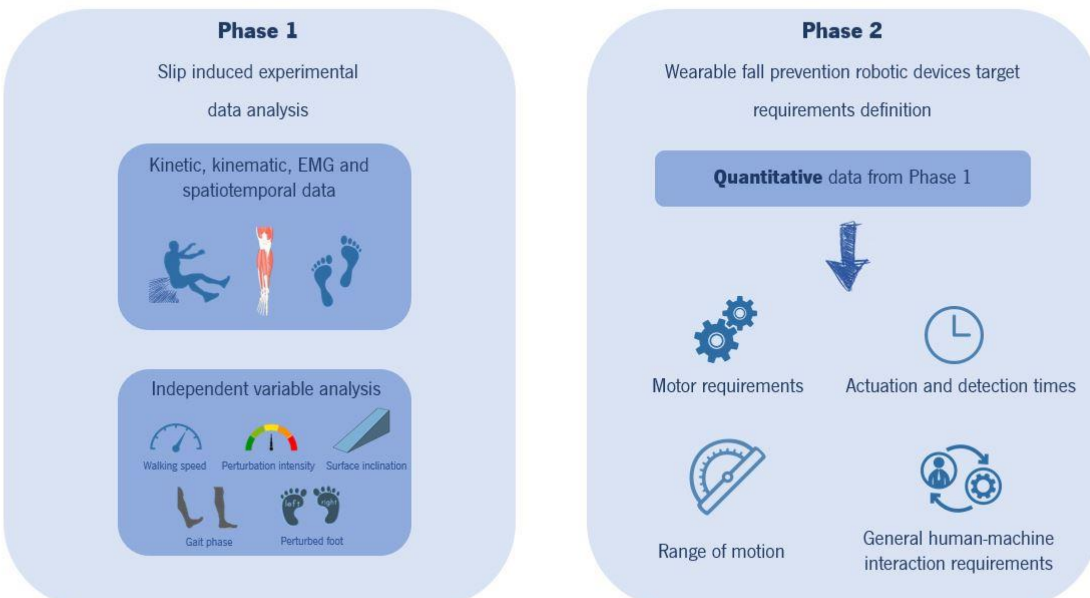


Figure 4.1 Schematic of the Project Phases. Phase 1: Slip induced perturbations experimental data analysis. Phase 2: Target specifications definition for a wearable robotic device for fall prevention.

Thus, in terms of the various stages of the product development process, this dissertation aims to fulfil the following initial steps highlighted in Ulrich et al [73]: i) Opportunity/necessity identification; ii) identifying costumer needs; and iii) product specifications.

Regarding the first point, the identification of the opportunity was done through the review of the existing scientific literature related to this topic. This necessity was then complemented with data from the WHO [7] and with the lack of quantitative data to provide information about the articulation that should be actuated upon after analyzing the existing fall prevention wearable robotic devices.

Considering the step of identifying costumer needs, this was done in a broad way, considering only the need to avoid a slip related fall and other transversal aspects to the design and development of wearable medical devices. Both phase 1 and phase 2, will be addressed with more detail in the subsequent subsections.

4.2.1 Slip-induced experimental data analysis

The collection of slip-induced falls experimental data prior to the start of this dissertation was critical for Phase 1. From the data collection protocol it was possible to analyse several types of variables related to the biomechanical response including kinematic, kinetic, spatiotemporal and electromyographic. Additionally, the influence of several external conditions was also included in the analysis. Firstly, the study of the various DV's allowed to obtain a complete analysis of the biomechanical response to this type of perturbation. In turn, the analysis of IV's such as inclination, gait speed, perturbed foot, and perturbation's intensity allowed to conclude about their influence on the recovery process. It was feasible to comprehend the biomechanical response to unsafe scenarios with changeable circumstances using this research, making it easier to discover solutions for fall prevention. Obtaining quantitative data as an outcome of this study was another goal to consider during the analysis in order to aid target specifications definition, which is addressed in 4.2.2. Biomechanical analysis to slip-induced perturbations will be presented in Chapter 5.

4.2.2 Definition of target specifications for wearable fall prevention robotic devices

Since the development of robotic devices is one of the currently growing approaches to fall prevention, the second major goal of this dissertation is to define target specifications for a future robotic device for fall prevention purposes. Based on the quantitative outputs from the biomechanical analysis. Further, these specifications should consider the possibility of actuating different joints of the robotic devices, so the definition of the target specifications was made for the three joints of the lower limbs (hip, knee, and ankle). The target specifications definition based on quantitative experimental data allow proper design and customization of the device considering the specific needs for each joint, thus avoiding over-dimensioning and the use of standard quantitative data which can be detached from the device purpose. Robotic devices' target specifications to prevent falls due to slip-induced perturbations will be presented in Chapter 6. Firstly, general requirements for the design of wearable robotic devices will be introduced. Then, some quantitative specifications namely torques, RoM and rpm for each lower limb's joints will be addressed as well as the LoB detection and actuation times.

4.3 Research Hypothesis

The investigation activities that were developed within the scope of this dissertation are based on the following hypotheses:

- That statistical analysis of the user's kinetic, kinematic, spatiotemporal and EMG parameters can conclude about the most relevant variables during slip recoveries process [74]–[76]. (Goal 3; Chapter 5)
- That FSM are able to rank joints according to their importance/effectiveness during slip-related perturbations' biomechanical response using kinetic, kinematic, spatiotemporal and EMG data [77], [78], [78]. (Goal 4; Chapter 5)
- That GRF and toques can be estimated using inertial data [79], [80]. (Goal 5; Chapter 6)
- That wearable robotic devices have the potential to prevent slip related falls [22], [31], [32], [44]. (Goals 2 and 5; Chapter 2 and 6).

4.4 Outcomes

The aging of the world's population results into an increased prevalence of neurological diseases such as Dementia, Parkinson, or cerebrovascular accidents. In addition to other consequences, the aforementioned diseases result in reduced mobility in people, thus increasing the incidence of falls. Besides the economic consequences for the world's health care systems, these events also result in post-fall injuries. For all these reasons and considering the need to develop alternative strategies for fall prevention, this dissertation intends, based on a biomechanical analysis, to gather a set of quantitative information for the design of robotic devices for fall prevention with the purpose of mimicking the natural human response to these events. The biomechanical analysis performed, considering multivariate data, (kinetic, kinematic, spatiotemporal and EMG) will also allow a global understanding of the biomechanical response to slips and, therefore, is also an outcome of this dissertation.

This state-of-the-art review, besides allowing to understand the human fall prevention responses and the involved variables of interest to be analysed during the study of experimental data from slip perturbations, it also allowed to find some gaps in the literature, which, as far as possible, will be filled during the analysis of experimental data collected at BirdLab. This outcome was achieved in Chapter 3.

Regarding the experimental data analysis' outcome this step will be a fundamental step in defining the target requirements of the project. Additionally, the multivariate experimental data analysis performed will provide a more comprehensive understanding of the biomechanical response to slip perturbations considering several conditions of the environment and gait, thus filling the gaps found in the review of the existing literature - another outcome of this dissertation.

Finally, the definition of the quantitative target requirements for the design of robotic devices for fall prevention was also a gap determined after the analysis of the existing literature. By presenting the general requirements for the creation of these devices and determining quantitative targe specifications, namely torques, RoM, rpm and detection and actuation times the present dissertation intends to overcome this literature gap.

4.5 Discussion

This dissertation aims to, based on a comprehensive and multivariate study of the biomechanical response to slip perturbations, gather a set of information useful to the development of wearable robotic devices to fall prevention. Two major phases were selected and introduced in this Chapter: slip-induced experimental multivariate data analysis and target requirements definition for fall prevention robotic devices. In this Chapter research hypothesis that were considered to support the investigation described were also addressed as well as the outcomes expected to achieve in this dissertation. Subsequent chapters will consider the stages of conceptual project design defined in this chapter. Phase 1 will be developed in Chapter 5 while phase 2 will be addressed in Chapter 6. The research work presented in this dissertation, seeks to address of the preliminary stages of the design of a wearable device for slip-like falls prevention allowing the determination of the target specifications from experimental data. This method allows to achieve a personalized, purpose-oriented, and efficient selection of the components to be included in the mechanical design of the device, thus overcoming this gap founded in scientific literature.

5 Slip-induced falls multivariate data analysis

The purpose of this chapter is to present a comprehensive analysis of the biomechanical response to slip-like perturbations considering kinetic, kinematic, spatiotemporal and EMG information during different walking conditions on a treadmill (i.e. speed and inclination) and considering variable slip intensities. This chapter will also address the experimental protocol executed before this dissertation that provides support information. Data processing will also be scrutinized in this chapter. The main outcome of this chapter is the analysis of quantitative data obtained via a biomechanical study about slip reactions, thus providing a comprehensive study of the biomechanical response to slip-like perturbations.

5.1 Introductory Insight

As an initial step in the analysis of the natural biomechanical response of humans to slip events, it is crucial to gather information capable of determining the characteristics of actuation capable of mimicking this response. Therefore, two experimental protocols were designed and conducted by BirdLab team work, to study variables of interest in slip responses analysis. Thus, it will be firstly presented both experimental protocols and the data processing steps. The dependent and IV's choice will also be addressed as well as all the steps of the statistical analysis performed.

In this analysis, the previously collected data were firstly processed using Matlab ® software (2021b, The Mathworks, MA, USA) to obtain the desired metrics to be analysed in SPSS ® software. The statistical analysis performed allowed to understand the interaction effect between the IV's considered. After the statistical procedures, two distinct methods were conducted to rank the DV's accordingly to their contribution to perform an effective biomechanical response. Eta partial squared, obtained from the statistical analysis and Feature Selection Methods (FSM) were conducted to obtain a DV ranking. Figure 5.1 presents schematically the whole process performed, from data collection to statistical analysis as well as the expected outcomes.

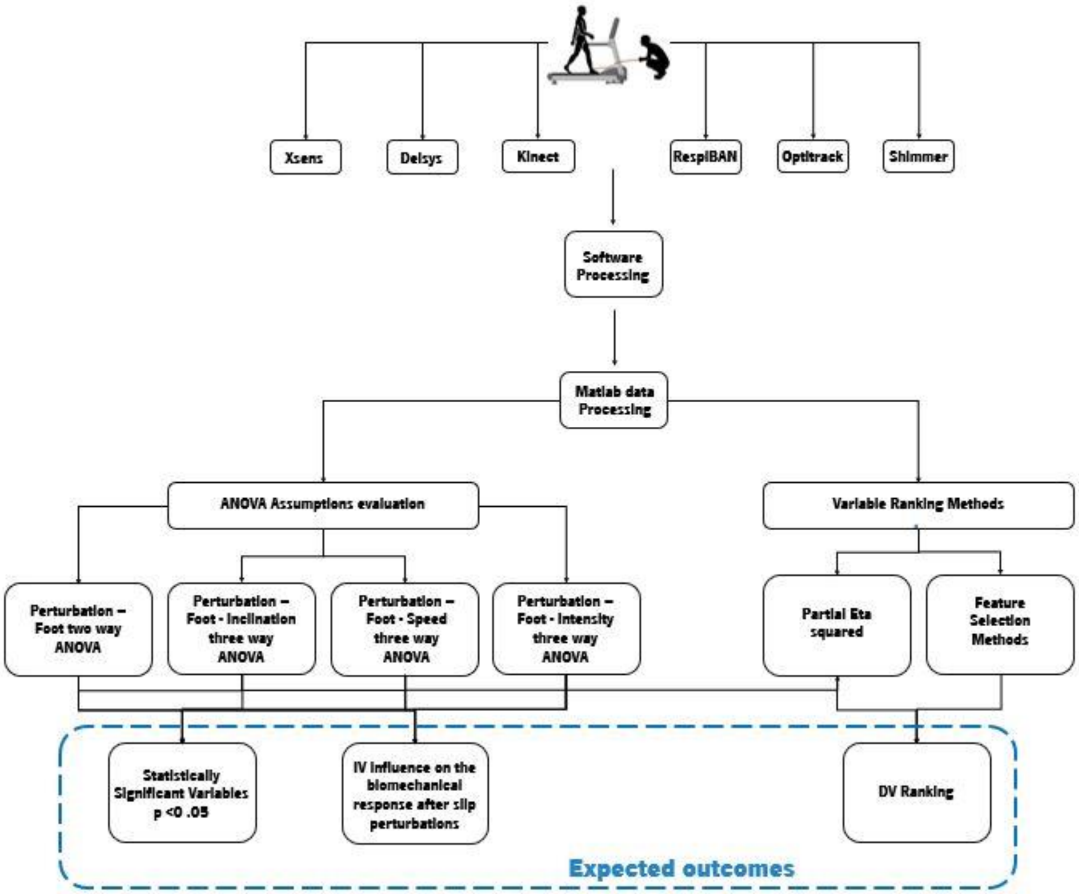


Figure 5.1 Slip-induced falls multivariate data processing and analysis.

5.2 Methods

5.2.1 Experimental Protocol

In order to study the biomechanical response to slip perturbations two experimental protocols were conducted at Birdlab. As further discussed, the second protocol emerged in order to complement the data analysis obtained by the first protocol. Regarding the first one [72] , eleven healthy young participants (age: 24.55 ± 2.15; height: 1.70 ± 0.09 m; weight: 63.25 ± 7.11 kg; males = 6; females = 5) were selected for the experience. In the second protocol 4 subjects were selected (age: 24.55 ± 2.15; height: 1.76 ± 0.05 m;

weight: 72.00 ± 5.00 kg; males = 4; females = 0). All the subjects that participated in these protocols presented right dominance. In both protocols, subjects were enrolled if they presented: i) healthy locomotion; ii) total posture balance; iii) more than 18 years; and iv) body mass lower than 135 kg. Subjects were excluded if they: i) presented a disease or deficit that affects locomotion; and ii) were recently subjected to surgical procedures that affect mobility. All participants provided written informed consent and voluntarily accepted to participate in the experimental trials. Each participant performed the qualitative assessment of the preferred foot by completing the Waterloo Footedness Questionnaire [81].

To provide multivariate data to better understand the biomechanical response due to slip perturbations, a wide range of sensors were used in the first protocol. *Xsens MVN Awinda* ® (Enschede, The Netherlands) and Optitrack V120 Trio ® (Corvallis, OR, USA) systems provide information about any potential changes in motion kinematic variables during both normal walking and slip perturbations. In turn, *Delsys Trigno* ® (Natick, MA, USA) provided muscles' electrical activity data, *RespiBAN* (Lisbon, Portugal) collected subject's respiration data and *Shimmer* ® (Dublin, Ireland) Galvanic Skin Response provided information from subject's galvanic skin response and heart frequency rate. Furthermore, *Kinect v2.0* ® camera (Redmond, WA, USA) offered video support to the labelling of events in the data samples. In the first protocol, participants were equipped with the Xsens MVN Awinda which is composed by 17 IMUs placed in the following body landmarks: i) head; ii) sternum; iii) pelvis; iv) right and left shoulders; v) right and left upper arms; vi) right and left forearms; vii) right and left hands; viii) right and left upper legs; ix) right and left lower legs; and x) right and left feet. These data were collected at 60 Hz and after the sensor placement, participants underwent the N-Pose calibration of the system. Afterwards, reflexive markers were placed in the following body landmarks: i) head; ii) sternum; iii) midtrunk; iv) right and left shoulders; v) right and left elbows; vi) right and left wrists; vii) right and left hips; viii) right and left knees; ix) right and left heels; and x) right and left feet. These markers were tracked at 120 Hz by *Optitrack* V120 ® Trio camera bar. Any existing shiny surface from subjects' clothes was removed to reduce *Optitrack* ® cameras noise. *Delsys Trigno* wearable sensors, which collected EMG data at approximately 1111 Hz were placed in some lower body muscles namely the RF, BF, TA and *gastrocnemius lateralis* (GL) from both legs. Three trials of Maximum Voluntary Contraction (MVC) were performed for each muscle to allow further normalization of EMG envelope. *RespiBAN* ® system was worn on the upper trunk, between the sternum and Xiphoid process, and the Shimmer GSR ® device was placed on the dominant forearm with the electrodes placed on the middle fingers and index. These devices collected data at 1000 Hz and 100.21 Hz, respectively.

Finally, *Kinect ® camera* was used to provide video recordings at 30 frames per second. The reflexive marker and IMU placements are presented in the following figure [72].

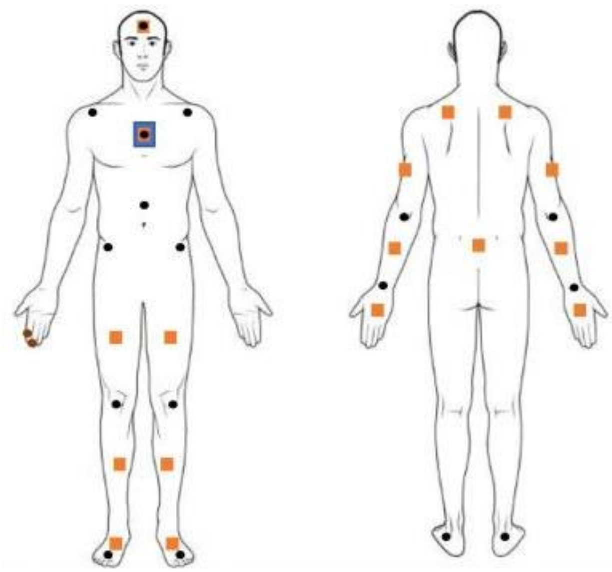


Figure 5.2 IMU (orange squares), Reflexive marker (black dots), RespiBAN device (blue square) and Shimmer electrodes (brown dots) placement. Taken from [72].

During the trials, subjects also worn a safety harness system to prevent falls in case of an irreversible LoB. This system consisted in a vest that was attached to a structure in the ceiling using a rope. The length of the rope was adjusted to register a minimum of 15 cm between the knees and the treadmill belt when the subject was suspended. This step was performed by asking subjects to raise their feet, which led to the application of all body weight into the harness system.

To achieve synchronous data acquisition from all the sensors used, Sync Lab Desktop Application, developed by colleagues at *Birdlab*, for *Windows OS* was used to synchronously start and stop data collection from all sensors previously mentioned. The trigger signals sent by the Desktop application are electronic or wireless pulses. The former ones are either sent via: i) *Syncbox* – a previously team-developed hardware interface that connects to the Xsens ® and Delsys ® systems; or ii) by direct USB communication. Direct wireless connectivity is established with the RespiBAN ® and Shimmer GSR ® systems. Figure 5.3 summarizes the experimental trial used for the data collection. *Optitrack* cameras were tilted to capture all the reflexive markers placed on the subject's body.

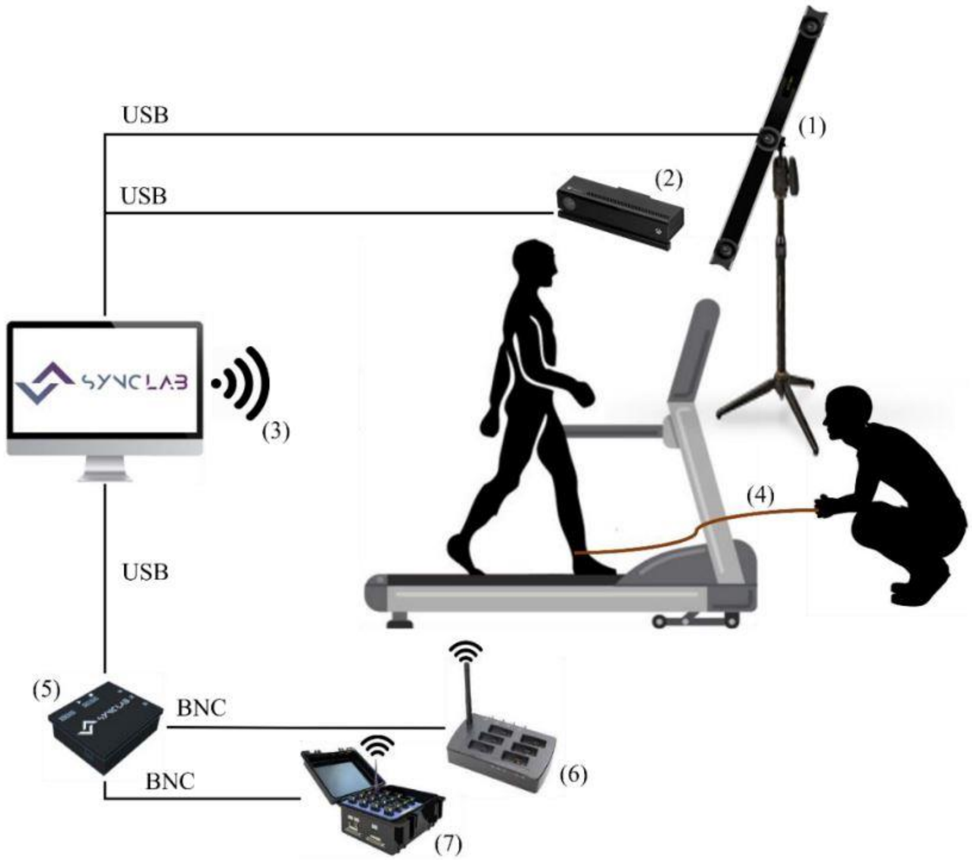


Figure 5.3 Experimental setup used for slip-like perturbations data collection. 1) Optitrack V120 Trio cameras. 2) Kinect v2.0 camera. 3)wireless communication between the computer running the RespiBAN app and Shimmer systems. 4) Rope attached to the participant's ankle, which is pulled by the operator to induce the perturbation. 5) Sync Box. 6) Xsens Awinda station, which establishes wireless communication with the Xsens IMUs 7) Delsys Trigno Workstation, which establishes wireless communication with the Delsys sensors. The safety harness system connected do the subject was not included for simplification. Taken from [72].

In contrast, in the second protocol only Xsens and Delsys sensors were used. Regarding the Xsens IMU'S, participants were equipped only in the lower limbs in the following body landmarks: i) right and left upper legs; ii) right and left lower legs; and iii) right and left feet. Additionally in this protocol, a Xsens IMU was placed in the rope to allow a more precise perturbation detection and to allow the study of the perturbation intensity's influence by collecting the rope accelerations. Similarly to the first protocol these data were collected at 60 Hz and following the sensor placement, participants underwent the N-Pose calibration of the system. In turn, *Delsys Trigno* ®wearable sensors, were placed in the same muscles of the first protocol.

During both protocols, subjects were asked to manage unexpected slip-like perturbations while walking in the treadmill. Subjects were not informed about the protocol to not cause any prior bias on their biomechanical response. A familiarization trial was also performed by the subjects walking in the treadmill without induced slip-like perturbations using the entire sensor setup. To simulate a real-world slip perturbation a trained operator pulled a hidden rope tied up to the subjects' ankle at some heel strike or toe-off event. As the rope was always attached to one of the subject's feet during all the trials, the participants did not know if there was going to happen a perturbation or not.

Regarding the first protocol each participant underwent 8 trials, being thus exposed to all the combinations between perturbed gait event (HS or toe-off); perturbed leg (right or left leg) and treadmill inclination (0 and 10%). In the second protocol perturbations were only induced at the HS and with 0% of inclination in both legs and at the same speeds defined in the first protocol. Table 5.1 shows each trial's order and characteristics for both protocols.

Table 5.1 *Trials order during the experimental protocols for data acquisition*

Protocol	***Trial Number***	***Perturbed leg***	***Perturbed gait event***	***Treadmill inclination (%)***
1	1	Right	HS	0
	2	Right	HS	10
	3	Right	TO	0
	4	Right	TO	10
	5	Left	HS	0
	6	Left	HS	10
	7	Left	TO	0
	8	Left	TO	10
2	1	Right	HS	0
	2	Right	HS	0
	3	Right	HS	0
	4	Left	HS	0

In each trial, 6 sub-trials were performed as subjects walked at 3 different speeds (1.8 km/h, 5.4 km/h and a normalised speed calculated depending to the subject's leg length – equation 5.1). For each velocity a perturbation and a non-perturbation sub-trial were performed. Slow and fast gait speed were defined according to the literature [30], [82], [83]. The normalised speed (v) for each subject was calculated accordingly with the dynamic similarity principle which is expressed by the equation 5.1 [84].

$$v\ (m/s) = \sqrt{F_r gl} \tag{5.1}$$

where Fr is the Froude number (0.15); g is the gravity acceleration ($9.81 m/s^2$), and L is the leg length measured from the prominence of the greater trochanter external surface to the lateral malleolus.

Table 5.2 demonstrates the characteristics from each of the 6 sub-trials. These sub-trials were conducted randomly to make perturbations unpredictable. During the trials where perturbations were delivered, the operator applied 3 perturbations in random moments of the trial. Non-perturbation trials had a mean duration of 30 seconds. In turn, perturbation trials had a variable duration generally between 30 seconds and 1 minute.

Table 5.2 Characteristics of the 6 sub-trials performed in each trial

Velocity	Perturbation?
1.8 km/h	Yes
1.8 km/h	No
Normalised velocity	Yes
Normalised velocity	No
5.4 km/h	Yes
5.4 km/h	No

5.2.2 Data Pre-Processing

After data collection, data were processed using *Matlab* ® software to convert all sensors' data into Matlab ® table format. Regarding the EMG data collected using Delsys ® software, were previously normalised with the respective MVC information using EMG Analysis software. In turn, *Optitrack* ® reflexive markers were labelled using Motive software, however some markers were excluded as they were hidden by some obstacle during the trial. Therefore the labelled markers were: i) head; ii) sternum; iii) midtrunk; iv) right and left shoulders; and v) right and left hips. The frames obtained from the Kinect camera ® were aligned together using Adobe Premiere Software ® to produce a video for each trial.

After these steps, data were processed in Matlab ® by adjusting the sampling frequency to the lowest value, excluding the Kinect v2.0 ®. As the system with the lower sampling frequency was Xsens with 60 Hz, all data were downsampled to 60 Hz. Afterwards, data from each sensor were organized into mat tables with each column corresponding to a feature extracted from the sensors. For each trial, the number of Xsens data samples served as reference and the excess data samples acquired from the other sensors were excluded and empty samples were added if there was a lack of data samples. Data samples from the different systems were temporally aligned according to the timestamps of start and stop data recording provided by the Sync Lab Desktop App. All the data collected from all the sensors were aligned and had the same number of samples for each trial. Then, sensor data tables were concatenated to generate a single data for each trial.

After this process, all the data were labelled considering the following events of interest: i) start of a sub-trial: considered in the frame of the first HS at the foot being perturbed since the subject achieved steady walking during the sub-trial; ii) end of a sub-trial: marked in the frame of the last heel strike of the foot being perturbed in steady walking during the sub-trial; iii) perturbation onset: marked in the frame where the operator starts to pull the rope to perturb the participant's gait; iv) end of a perturbation: marked in the first frame where the rope is curved after its maximum extension caused by the pull rope; v) start of the biomechanical response: the frame immediately after the frame considered as the end of the perturbation; vi) end of the biomechanical response: marked in the frame of the first heel strike of the perturbed foot after the subject has recovered from the perturbation and regained steady walking. During this labelling process, some perturbations were not capable to clearly destabilize the subject. In this case, perturbations were not considered together with the remaining. Instead, they were labelled differently to guarantee interference absence in the biomechanical response analysis.

Labelling process of the first protocol's data was performed in *Djv* ® software as this software allowed to identify Kinect frame numbers. Using the timestamps provided by Sync Lab Desktop App table data timestamps were correlated to the table timestamps to mark the events. Then, the frames of interest were introduced in Matlab ® script to label the data samples from each trial table with the respective event. Regarding the second protocol data, the end of the perturbation was determined by the minimum value of the AP (movement direction) GRF, after the perturbation onset, which was calculated with Customizable Toolbox for Musculoskeletal simulation (*CusTOM)* toolbox presented in Chapter 6. Using the protocol 1 data labelling (using Djv), it was possible to confirm that the minimum of the AP GRF corresponds to the end of the perturbation. Also, and as previously mentioned, the IMU placed on the rope allowed to determine the onset perturbation frame. In this case it was selected the SUM VM of the rope accelerometer signal inversion just before the peak. Figure 5.4 and figure 5.5. show, respectively, the signals used for perturbation onset and ending frames determination.

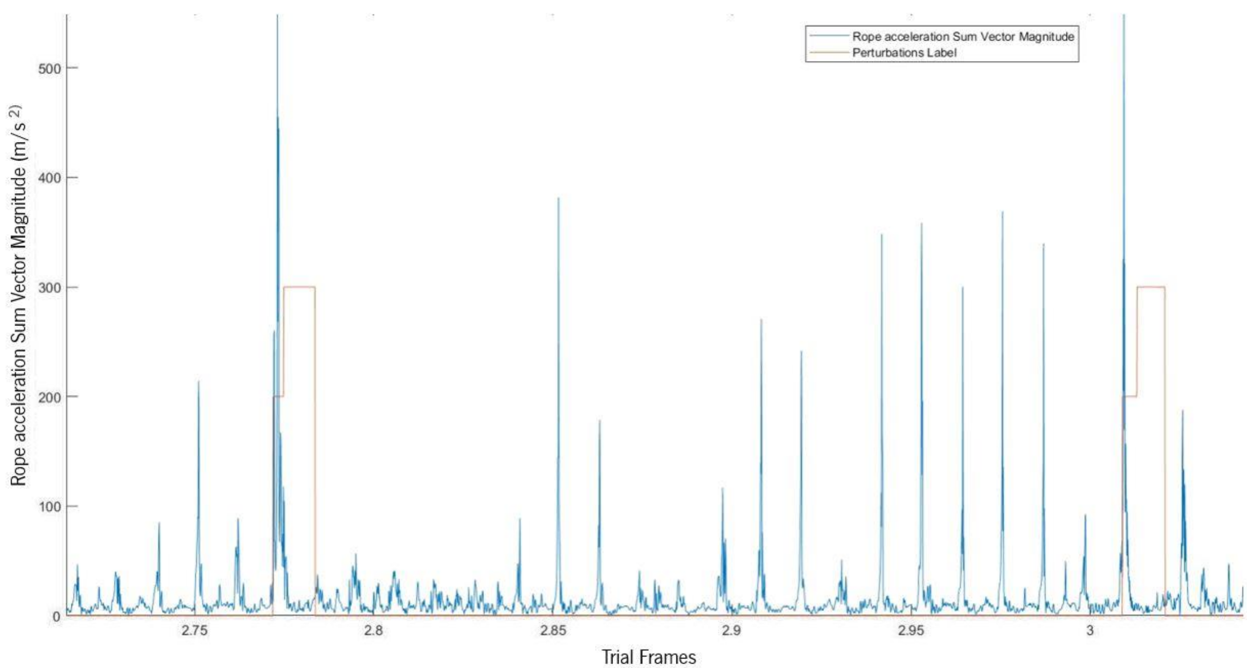


Figure 5.4 Perturbation onset labelling based on the Sum VM of the rope accelerometer. Blue signal is related to the rope accelerometer Sum VM while orange label indicates the three gait situations accordingly with the following labels: 0: steady walking; 200: rope pull and 300: biomechanical response.

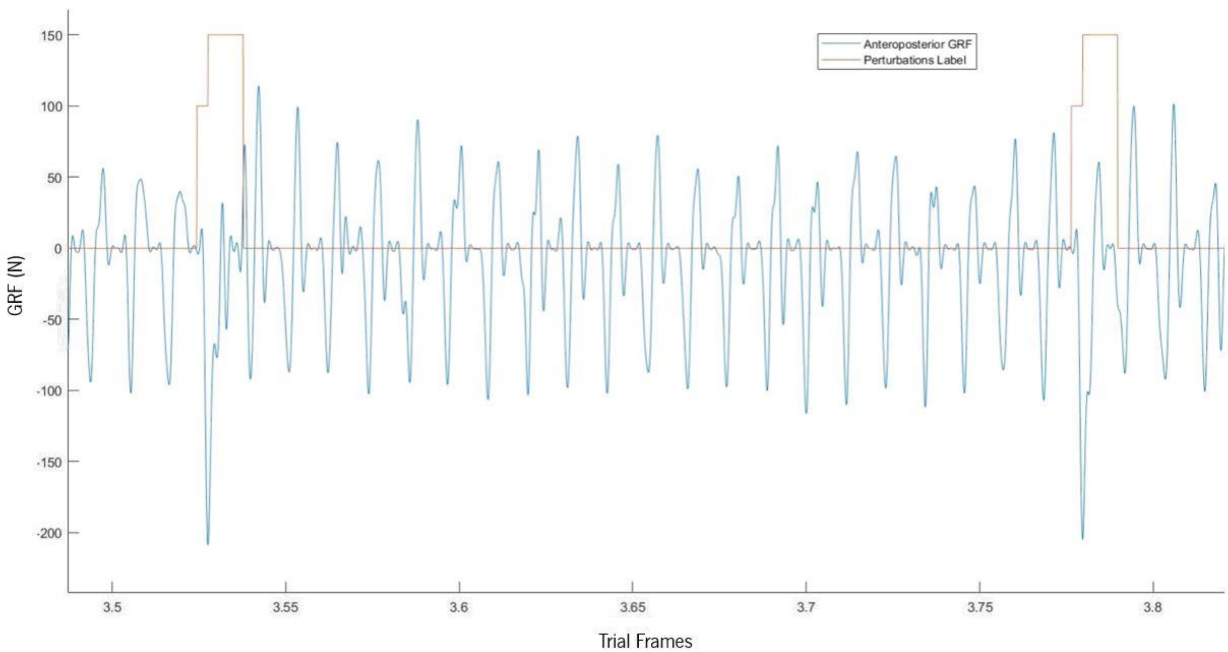


Figure 5.5 Perturbation ending labelling using AP GRF. Blue signal is related to the AP GRF while orange label indicates the three gait situations accordingly with the following labels: 0: steady walking; 100: rope pull and 150: biomechanical response.

5.2.3 Statistical Analysis – Preliminary steps

Prior to the statistical analysis, the variables of interest were chosen based on the articles reviewed in Chapter 3. Kinetic, kinematic, EMG and spatiotemporal variables were selected to obtain a comprehensive analysis of the biomechanical response to slip perturbations. The state-of-the-art review of the biomechanical responses to slip-like perturbations allowed to select the most relevant variables for this analysis. Table 5.3 shows the variables that were chosen for investigation based on their group.

Table 5.3 Kinematic, spatiotemporal and EMG variables included in the analysis

Kinematic variables (Angles mean values)	***Spatiotemporal variables (Mean values)***	***EMG Variables***
✓ Right Hip in frontal plane ✓ Left Hip in frontal plane ✓ Right Hip in sagittal plane ✓ Left Hip in sagittal plane ✓ Right knee in sagittal plane ✓ Left knee in sagittal plane ✓ Right ankle in sagittal plane ✓ Left ankle in sagittal plane	✓ CoM velocity in anteroposterior direction (CoM_x) ✓ COM velocity in medio-lateral direction (CoM_y) ✓ COM velocity in vertical direction distance (CoM_z) ✓ between both feet in x and y axes ✓ Distance between both feet in x, y and z axes ✓ Distance between CoM and Right Foot ✓ Distance between CoM and Left Foot	✓ Latency ✓ Excitatory power response ✓ Inhibitory power response

The kinematic variables, hip, knee, and ankle sagittal angles are analysed in much of the literature on this topic. Additionally, hip movements in the frontal plane (adduction/abduction angles) were also addressed in [32] as a movement with an important role in the biomechanical response after a slip-like perturbation. For this reason, these variables were also included in this analysis.

CoM related parameters allow to study the subject balance behaviour during steady walking, perturbation, and biomechanical response. [85] Thus, CoM velocities in AP, ML and V directions were included in the analysis. Distance between CoM and both feet were also included as well as the distance between right and left feet considering two (x and y) and three dimensions (x, y, and z).

Finally, in order to complement the kinetic, kinematic, and spatiotemporal analysis, EMG related variables were also studied, considering perturbation, speed, and perturbed foot IV's. Latency periods, and EMG inhibitory and excitatory responses were the variables selected for this complementary study. Similarly to Marigold [86] and Xingda Qu [27], activations threshold for excitatory and inhibitory responses were distinguished. Regarding the first one, excitatory response is related to the muscular activity above the steady walking mean plus two standard deviations. In contrast, inhibitory responses were related to the muscular activity under the steady walking mean less two standard deviations. As previously referred, the mean and standard deviations values were calculated only during steady walking situations to guarantee its independence from the number of perturbations induced in each sub-trial. Also, it was calculated for each speed and subject to avoid the interference of both IV's. Figure 5.6 shows both, excitatory and inhibitory thresholds defined for the study of the EMG variables.

Concerning the muscular activation latency it was determined considering the interval between the HC and muscle activity onset. HC was defined using the contact points provided by *Xsens software* ®. In turn, muscle activity onset was defined as the time when the EMG signal first deviated above or beyond the activation threshold for at least 30 ms [86]. Finally, EMG power was obtained by the integral of the EMG signal above or under the activation threshold for, respectively, excitatory, and inhibitory responses. All these variables were extracted for each HS detected and the data were posteriorly labelled in perturbation and non-perturbation situations.

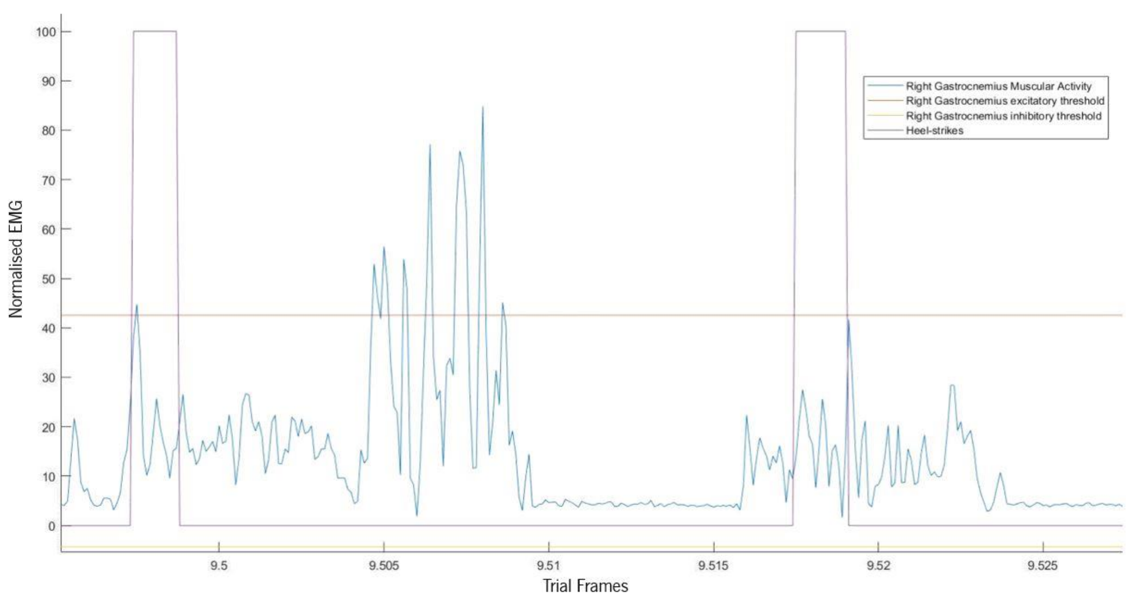


Figure 5.6 Metrics included in the statistical analysis. Blue signal presents the muscular activity of Gastrocnemius. Orange and yellow lines indicate, respectively, the excitatory and inhibitory thresholds. Heel-strikes appear in purple.

Average values of the kinematic and spatiotemporal variables were calculated using *Matlab* ® software and then imported to SPSS ® software. To calculate the average values, data were firstly sequentially segmented accordingly to the label of each frame (steady walking, slip-like perturbation, or biomechanical response). Afterwards, the means for each period previously segmented were calculated.

Before performing the statistical test, the Analysis of Variance (ANOVA) assumptions were analysed. Multicollinearity analysis, sample independence and multivariate normality were checked for all the ANOVA's models performed to guarantee an increased power of the statistical test and to reduce type I error [87]. Considering the observations number of the datasets obtained by the two protocols performed (n= 659 for the first dataset and n=96 in the second dataset) data homogeneity, linear relationship between IV and DV assumptions were not checked as for group sizes higher than thirty, ANOVA is robust against violations of homogeneity and variance-covariance of the IV and DV assumptions [75]. Considering the dataset's size, outliers were also not excluded. Regarding the multivariate normality assumption it can be tested using different methods. Kolmogorov-Smirnov and Shapiro-Wilk tests, are commonly used to test this assumption [76]. Skewness and kurtosis can also be used in normality tests when analysing datasets with n>300 samples. Literature reports that in large samples, normality checking methods rely on the Central Limit Theorem i.e., the true population mean is approximately normally distributed around the average of a large number of IV's. For this reason skewness and kurtosis values were used to test normality in the first dataset while Kolmogorov-Smirnov test was used to test the second protocol dataset's normality. Regarding the first method, data is assumed to be normally distributed when kurtosis value is less than 4 or skewness value is less than 2. Regarding the second method, the null hypothesis states that when p> 0.05 the null hypothesis is accepted, and data is normally distributed. To study this ANOVA assumption, data were first split accordingly the data labelling of the IV's included in the analysis. Finally, to analyse the multicollinearity between the DV included in this study, a Pearson correlation test was performed. An absolute value of the Pearson coefficient higher than 0.9 indicates that the variables are strongly correlated. In these cases, one of the correlated variables was selected being the other one discarded as present approximately the same information. All the assumptions were conducted in the statistical software SPSS ®.

When analyzing ANOVA results, a significant p-value (< 0.05) demonstrates that population means differ from one another considering the interaction of the IV's studied. After the ANOVA analysis, Tukey B post hoc was performed to determine more specifically which means differ. Tukey-B was used as this post hoc test allows to make pairwise comparison between groups with different sizes [74]. Additionally, the effect size was also included in the statistical analysis in order to obtain a quantitative difference between the groups analysed. In comparison to the p-value which only indicates if there is or not statistically significance between two groups, the effect size indicates quantitatively how large the difference is [88]. Further it will be presented the results of each ANOVA's model performed.

As aforementioned, perturbation, speed, inclination, perturbed foot, and perturbation intensity were the variables included in this analysis. In order to study the slip-like perturbation effect in all situations, this IV was included in all ANOVA's tests performed. Also, in all the ANOVA's tests performed the perturbed foot IV was included in order to distinguish the role of the perturbed and unperturbed limb during slip-like responses. A two-way ANOVA and 3 three-way ANOVA analyses were performed combining perturbation and perturbed foot with the 3 remaining IV's (speed, inclination, perturbed foot, and intensity). ANOVA's tests were posteriorly complemented with Tukey-B post hoc. The statistical procedure conducted is shown in Figure 5.7. Appendix I provides Tukey-B post-hoc results for the IV's interactions studied.

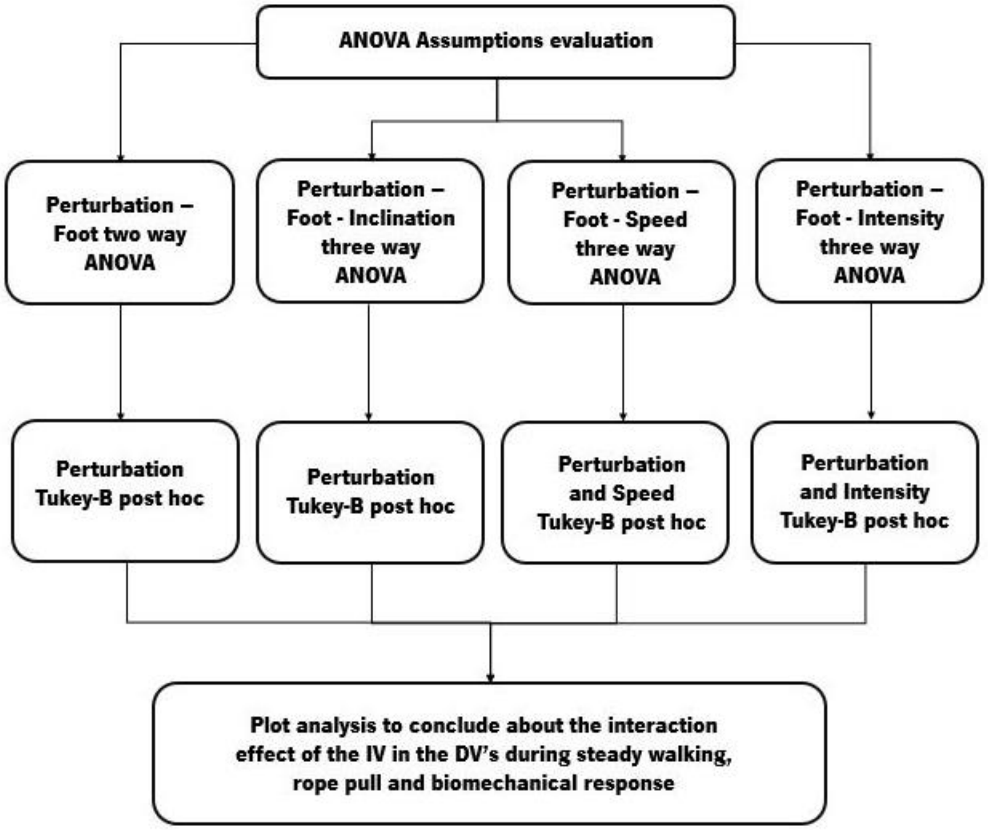


Figure 5.7 Statistical analysis methods

Prior to the analysis of the interaction of the IV's perturbation and intensity, perturbations were clustered accordingly to its intensity. As aforementioned, the intensity was measured by the Sum VM of the accelerations measured by the IMU placed in the rope. Using k-means algorithm of Matlab ®, these intensities were clustered in three different groups hereafter referred to as low, intermediate, and severe intensity slip-like perturbations. The distance selected for this clustering procedure was the squared Euclidean distance. As a result of perturbations being manually provoked, some occurred during the swing phase. Therefore, trials collected during protocol 2 in which perturbations occurred 400 ms or more before the closest heel strike were not considered for analysis. 400 ms was the selected period in order to obtain enough data to study the interaction effect of all the IV's. Figure 5.8 and Table 5.4 show the three perturbations clustering as well as the respective means and extreme values.

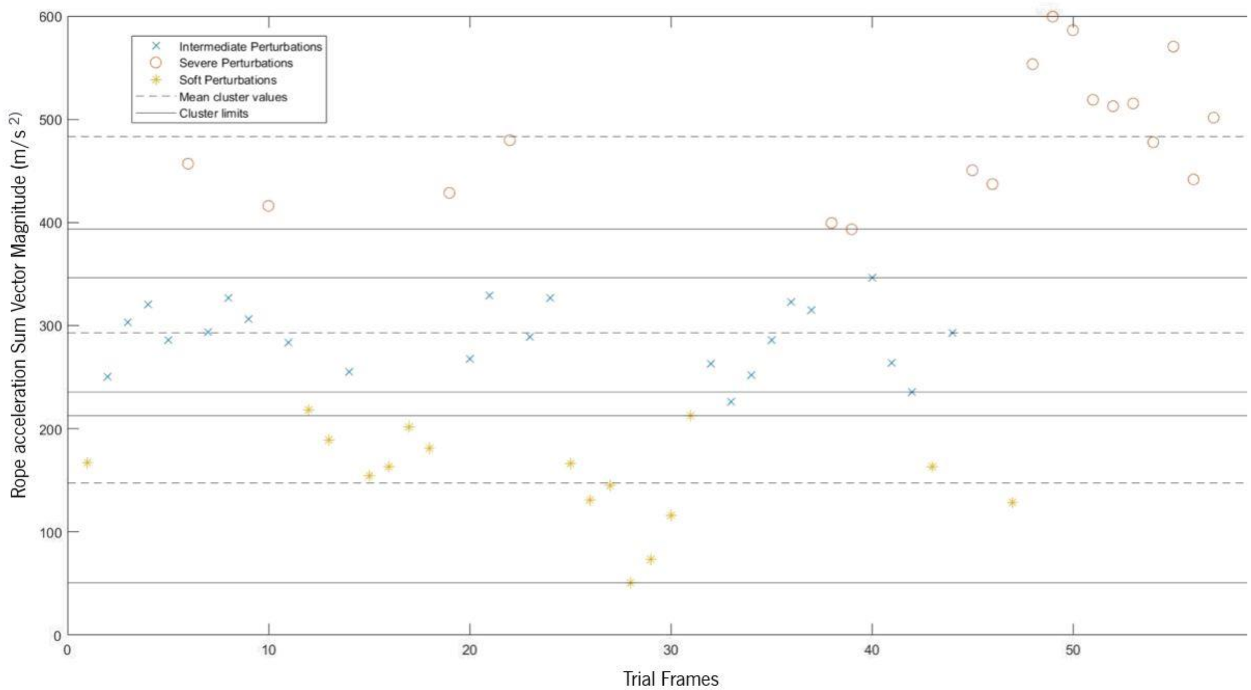


Figure 5.8 Intensity perturbations clustering using Matlab ® k-means. Soft, intermediate, and severe perturbations are respectively presented in yellow, blue, and red symbols.

Table 5.4 Perturbations' intensity clustering properties obtained using Matlab ® k-means algorithm

Cluster	***Maximum Value (m/s^2)***	***Minimum Value (m/s^2)***	***Number of Perturbations***
Soft	212.83	50.99	19
Intermediate	346.67	235.32	16
Severe	599.40	393.35	12

5.2.4 Variable Ranking Procedures

As previously mentioned, one of the main objectives of this dissertation was the ranking of the various DV's to understand their influence, in a quantitative way, on the biomechanical response to slip perturbations. With this objective, two distinct approaches were followed. Firstly, the effect size of the

ANOVA's models using the value of partial eta was included in the statistical analysis to quantify the influence of a given variable in causing significant difference between groups. In theory, higher partial eta squared values are associated with lower p values. Eta values were determined for all ANOVA models analysed. It should be noted that, naturally, this value changes when the ANOVA model is also changed. Therefore, the ranking of the variables according to this value allows conclusions to be drawn only for a specific ANOVA's model. Partial eta squared < 0.01 are associated with variables whose variation in means was low because of the interaction of the DV's under analysis. In turn, values between 0.01 and 0.06 and values > 0.14 refer, respectively, to situations of medium and large effect size [88].

Regarding the second approach, FSM were also used to allow variable ranking. These methods are commonly used in Machine Learning methods [78]. The outcome of these methods is a variable ranking that indicates which variables show greater differences when comparing two distinct situations (classes). The FSM performed are shortly introduced in Table 5.5. This table also presents if the variable ranking depends on similarity or dissimilarity.

Table 5.5 Brief explanation of the FSM used. N\A = Not Available

Feature Selection Method	***Brief Method Description***	***Ranking based on***
ILFS	Algorithm that performs the ranking step by considering all the possible subsets of features exploiting the convergence properties of power series of matrices [78].	Similarity
InfFS	The Inf-FS is a graph-based method which exploits the convergence properties of the power series of matrices to evaluate the importance of a feature with respect to all the other ones taken together [77].	Similarity
ECFS	Ensemble methods construct a set of many individual classifiers (and combine them to classify new data points by taking a weighted or unweighted vote of their predictions [89].	Similarity
MRMR	MRMR is an algorithm that ranks features based on their importance in predicting the target variable, where importance has a relevance and redundancy component. Relevance indicates how well the feature is correlated to the target while redundancy is related to how well a feature is related to the features selected in previous iterations [90].	Similarity
Relieff	Is a popular multivariate filter based on nearest neighbours. It works by randomly selecting samples and searching for nearest neighbours from the same class [91].	Similarity
MCFS	Method that is able to select the set of features that can cover all the possible clustering in the data. In MCFS, spectra analysis is used to measure the correlation between different features without label information needed [92].	Similarity
UDFS	Method that aims to select the most discriminative features for data representation. The algorithm optimizes the features and provides an output with feature ranking and weights [78].	Dissimilarity
LLCFS	The LLC algorithm searches for a solution that ensures the cluster labels in the neighbourhood of each point are as pure as possible [93].	Similarity
CFS	In this method a good feature will always be highly correlated to the class and not redundant to any other relevant features [78].	Similarity
FSASL	In this method the global structure learning and feature selection are integrated within the framework of sparse representation; the local structure learning and feature selection are incorporated into the probabilistic neighbourhood relationship learning framework [94].	N\A

After performing the FSM presented in the previous table, the ranking obtained for each variable was normalised between 0 and 1. Then normalised ranks were summed for each variable which allowed to rank all the DV's. In the case of FSM based on similarity, variables with higher values after sum were placed first. In opposite, in dissimilarity methods, DV were ranking upwardly.

5.3 Results

This section presents the results of each of the ANOVA models tested. In addition to the statistically significant different variables which are presented in Table 5.6, these results will allow to conclude about the biomechanical response of the subjects enrolled in this protocol. The analysis carried out did not determine any muscles responses for the muscles marked with *.

Table 5.6 Two and three-way ANOVAs results for kinetic, kinematic, spatiotemporal and EMG DV's. statistically significant variables are presented in green shaded while the non-significant variables are presented in orange. N\A= Not Available.

Dependent Variable	***Perturbation vs Foot***	***Perturbation vs Foot vs Inclination***	***Perturbation vs Foot vs Speed***	***Perturbation vs Foot vs Intensity***
	p-value	p-value	p-value	p-value
Right Hip Frontal AVG	0.000	0.225	0.001	0.746
Left Hip Frontal AVG	0.000	0.194	0.025	0.919
Right Hip Sagittal AVG	0.000	0.207	0.017	0.956
Left Hip Sagittal AVG	0.000	0.104	0.156	0.706
Right Knee Sagittal AVG	0.000	0.009	0.000	0.368
Left Knee Sagittal AVG	0.000	0.000	0.000	0.084
Right Ankle Sagittal AVG	0.000	0.452	0.000	0.413
Left Ankle Sagittal AVG	0.000	0.113	0.000	0.492
CoM_x velocity	0.595	0.992	0.944	0.999
CoM_y velocity	0.633	0.884	0.990	0.999
CoM_z velocity	0.819	0.889	0.903	0.999
3D Foot Distance	0.321	0.278	0.002	0.744
Distance CoM - Right Foot	0.989	0.99	0.843	0.040
Distance CoM - Left Foot	0.851	0.999	0.673	0.091
Right BF Latency	0.366	—	—	—
Left BF Latency	0.305	—	—	—
Right RF Latency	0.650	—	—	—
Left RF Latency	0.210	—	—	—
Right GL Latency	0.541	—	—	—
Left GL Latency	0.083	—	—	—
Right TA Latency	0.661	—	—	—
Left TA Latency	*	—	—	—
Right BF Excitatory Response	0.000	—	—	—
Right BF Inhibitory Response	0.088	—	—	—
Left BF Excitatory Response	0.001	—	—	—
Left BF Inhibitory Response	0.939	—	—	—
Right RF Excitatory Response	0.008	—	—	—
Right RF Inhibitory Response	0.000	—	—	—
Left RF Excitatory Response	0.928	—	—	—
Left RF Inhibitory Response	0.011	—	—	—
Right GL Excitatory Response	0.002	—	—	—
Right GL Inhibitory Response	0.008	—	—	—
Left GL Excitatory Response	0.002	—	—	—
Left GL Inhibitory Response	0.744	—	—	—
Right TA Excitatory Response	0.321	—	—	—
Right TA Inhibitory Response	*	—	—	—
Left TA Excitatory Response	0.813	—	—	—
Left TA Inhibitory Response	*	—	—	—

5.3.1 ANOVA Perturbation vs Foot

As shown in Table 5.6, all variables related to joint movement in the sagittal and frontal planes, in the hip's case, appeared as statistically significant, thus indicating that there is a considerable difference in the average values of these variables when considering the influence interaction of perturbation and perturbed foot. In this subsection, the evolution graphs of the averages of these variables will be demonstrated. Post hoc analysis will be also addressed, in order to concretely understand where the differences in the averages lie. The analysis of the interaction between perturbation and perturbed foot IV's allows to take conclusions about the distinct roles of both perturbed and unperturbed limb.

Both left and right hips present a similar behaviour. The hip of the perturbed limb presents an increase in the means during the rope pull (flexion movement) when comparing with steady walking. Then, in the biomechanical response period the mean decreases indicating that this response is mainly characterized by hip extension of the perturbed leg. In turn, when analysing the hip considering the perturbation in the contralateral side, a reflexed behaviour is observed in the plots. Firstly, when the rope is pulled there is a decrease in the mean indicating a dominant extensor moment and then, there is an increase in the means showing a flexion movement of this joint. In both cases, the means during steady walking situations are very similar to the means obtained after the biomechanical response. Tukey-B post hoc determined significant differences between the three gait labels in the Right Hip averages and between rope pull and biomechanical response in the Left Hip's situation.

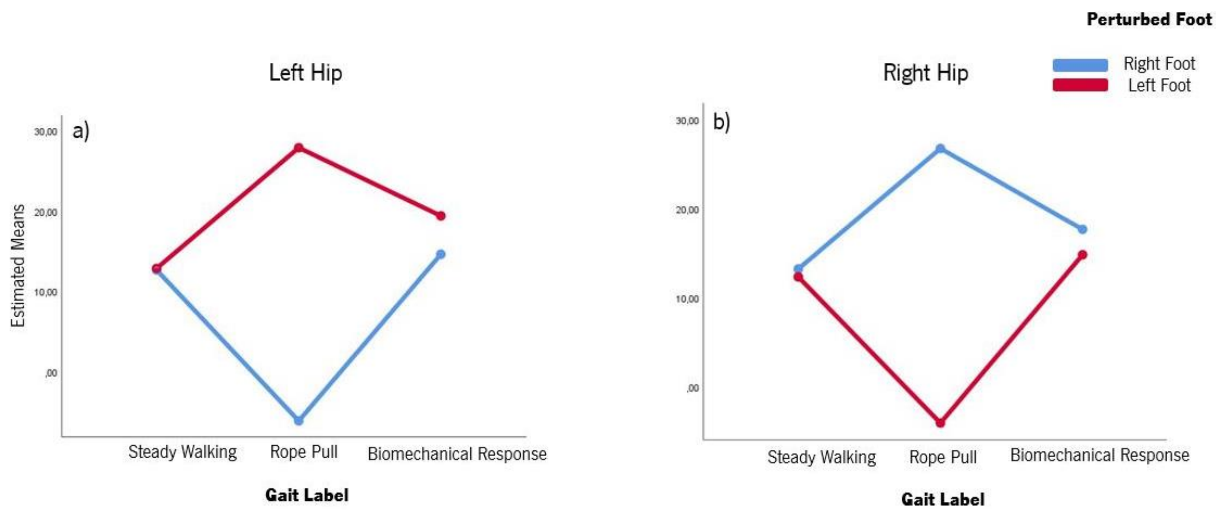


Figure 5.9 a) Left and b) Right Hip means, in sagittal plane, estimated during steady walking, rope pull and biomechanical response considering the interaction effect between perturbation and foot. Angle increasing indicates a flexion movement while decreasing indicates an extension movement.

Regarding the knees response, similar results were obtained: steady walking and biomechanical response means are very similar indicating an effective biomechanical response capable to prevent a fall. Ipsilateral knee movement during the rope is characterized by knee extension being knee flexion movement dominant during the biomechanical response. Regarding the contralateral knee during the perturbation this knee is mainly flexed. After the biomechanical response, this knee extends. In this joint, Tukey-B post hoc differences were situated between rope pull and biomechanical response labels in the Right Knee and between the three gait labels in the case of the left knee.

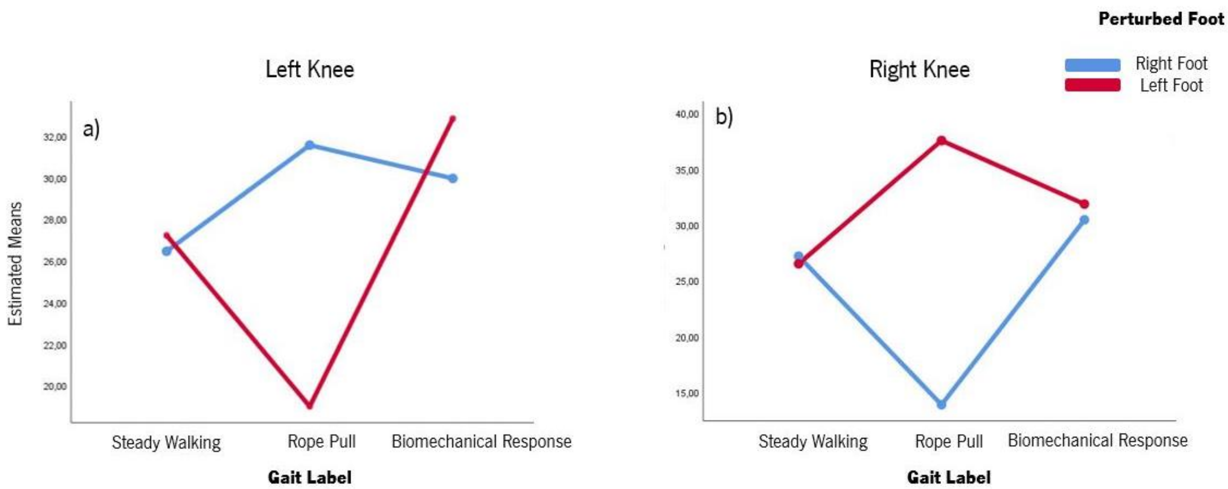


Figure 5.10 a) Left and b) Right Knee means, in sagittal plane, estimated during steady walking, rope pull and biomechanical response considering the interaction effect between perturbation and foot. Angle increasing indicates a flexion movement while decreasing indicates an extension movement.

Both ankles movement were also statistically significant in this ANOVA model. Analysing the perturbed ankle behaviour, the rope pull causes a dorsiflexion movement in this joint. In turn, plantarflexion is dominant during the biomechanical response. When in the contralateral position, both right and left ankles presented different movements when looking to the graph evolution between steady walking and rope pull labels. However, when considering the transition between rope pull and biomechanical response, both ankles performed a dominant plantarflexion movement. Tukey-B post hoc determined significant differences between the three gait labels for both sides ankles.

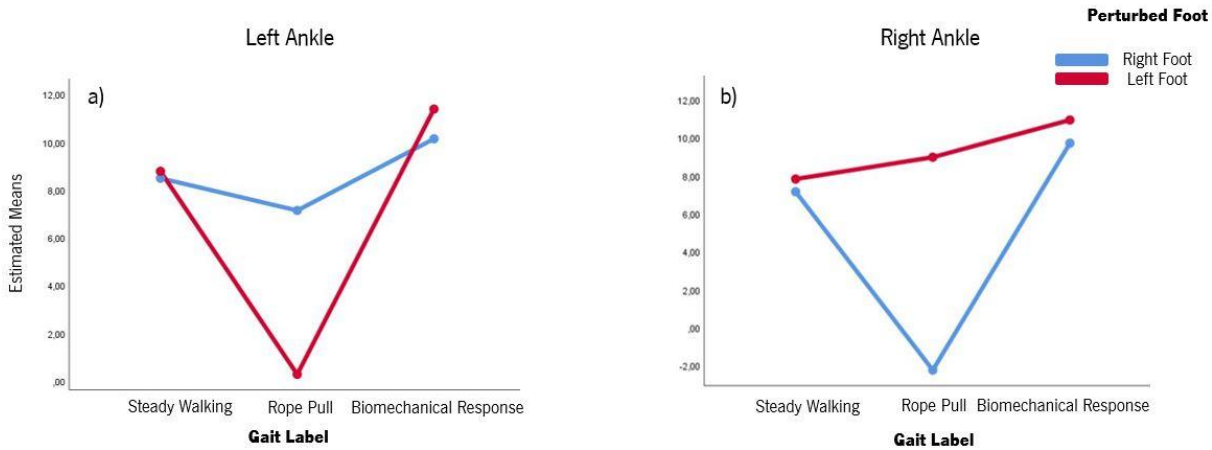


Figure 5.11 a) Left and b) Right Ankle means, in sagittal plane, estimated during steady walking, rope pull and biomechanical response considering the interaction effect between perturbation and foot. Angle increasing indicates a plantarflexion movement while decreasing indicates a dorsiflexion movement.

Regarding the movements in the frontal plane, both hips' means presented statistical significance. Concerning the contralateral hip, we can highlight the strategy to approximate both feet by hip adduction. Regarding the behaviour of the ipsilateral hip during the biomechanical response, both hips presented different movements. Firstly, right hip does not present a statistically significant means difference, while the left hip demonstrated an abduction movement. Post hoc tests resulted in significant differences between the steady walking and rope pull for the Left Hip means in frontal plane while in the means of the Right Hip in the same plane, differences were mainly located in the rope pull label when comparing with both, steady walking, and biomechanical response. Finally, distance between both feet (Figure 5.13) increases when comparing steady walking with rope pull situations then decreasing during the biomechanical response. Post hoc test determined differences considering the three gait labels defined.

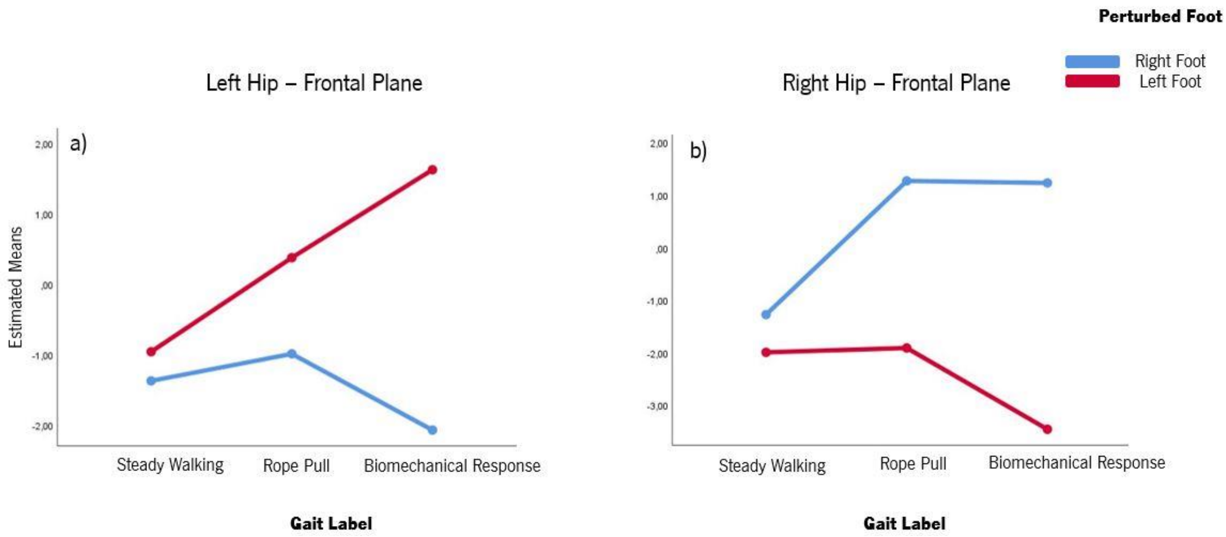


Figure 5.12 a) Left and b) Right Hip means, in frontal plane, estimated during steady walking, rope pull and biomechanical response considering the interaction effect between perturbation and foot. Angle increasing indicates an adduction movement while decreasing indicates an abduction movement.

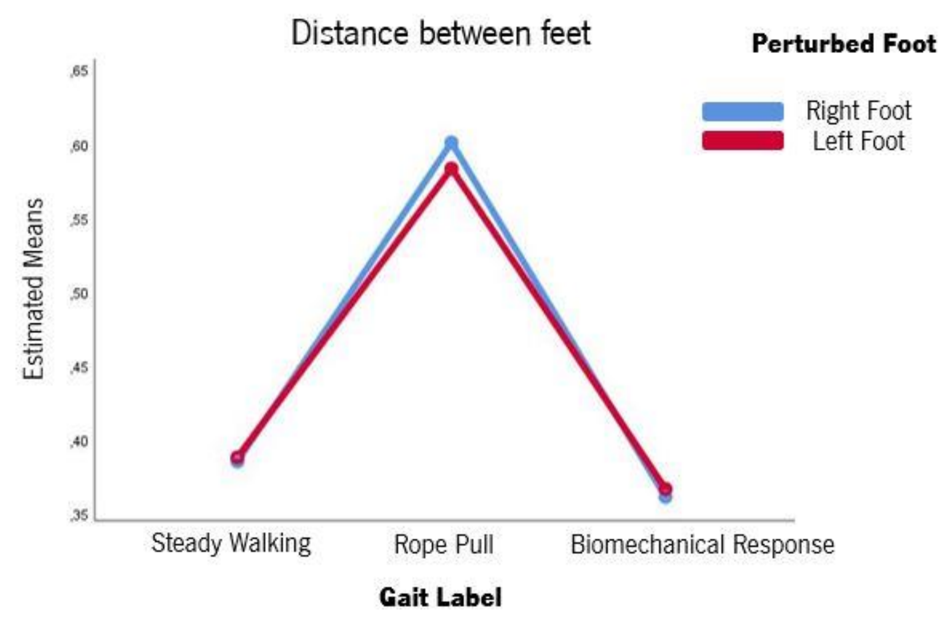


Figure 5.13 Foot distance means estimated during steady walking, rope pull and biomechanical response considering the interaction effect between perturbation and perturbed foot.

5.3.2 ANOVA Perturbation vs Inclination

When considering the interaction effect between perturbation, perturbed foot, and inclination only both knees presented statistical significance. Considering the Tukey-B post hoc results, differences in the means appear with greater evidence in the transition between the rope pull and the biomechanical response for the right knee and between the three gait phases in the left knee. Here in the case of the ipsilateral knee there is no significant differences in the plots evolution when comparing both joints, however in the case of 10° inclination the means are greater. When comparing the contralateral side of both knees for both inclinations, inclination appears to influence the knee's response. While in the 0 degrees of inclination trials, knee is flexed during the rope pull and then is extended, in the case of 10° of inclination, the evolution is different when comparing both right and left knees. Considering the right knee on the contralateral side, when the perturbation is delivered it is slightly flexed being slightly extended during the biomechanical response. Conversely, left knee shows a reflexed behaviour comparing with this knee.

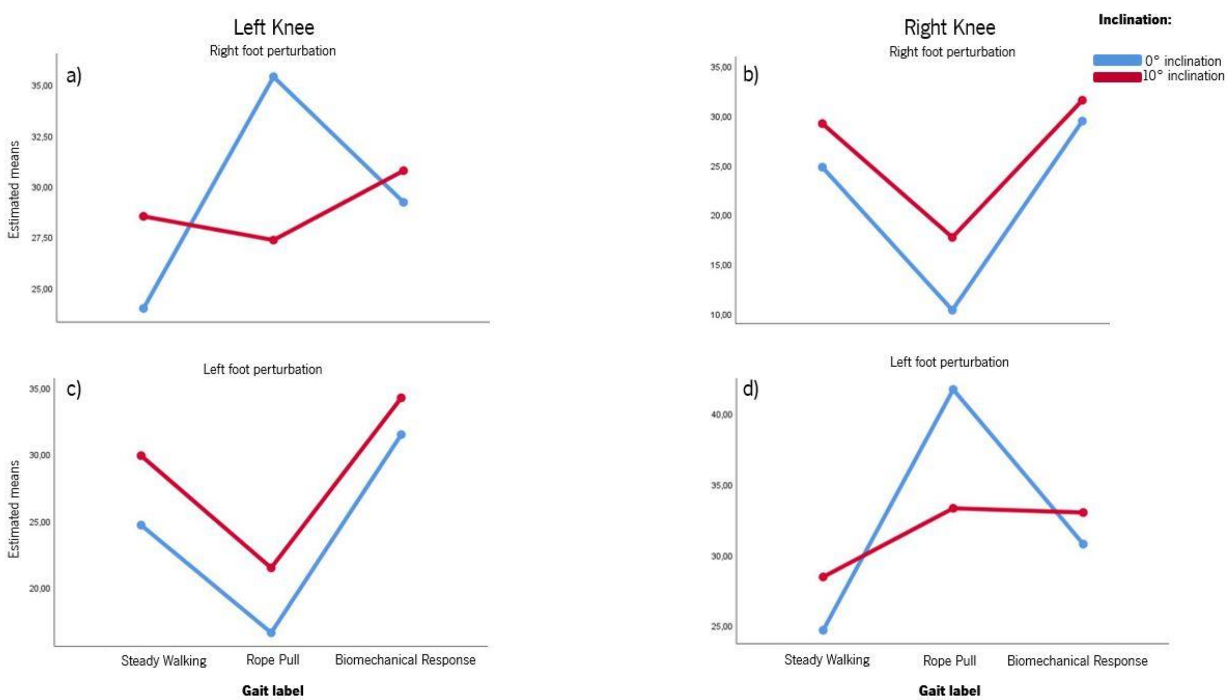


Figure 5.14 a) and c) Left and b) and d) Right Knee means, in sagittal plane, estimated during steady walking, rope pull and biomechanical response considering the interaction effect between perturbation, foot and inclination. Angle increasing indicates a flexion movement while decreasing indicates an extension movement.

5.3.3 ANOVA Perturbation vs Speed

Regarding the interaction effect studied between perturbation, perturbed foot, and speed, all the joints angles were statistically significant except left hip means in the sagittal plane, as previously introduced in Table 5.6. Also the distance between both feet presented statistical significance. Despite not being a significant variable Tukey-B post hoc tests determined significant differences for left hip means in sagittal plane considering the three speed labels (1.8 km/h, 5.4 km/h and self-selected speed). Observing both hips graphs, we can note that speed does not influence the extension and flexion movements in this joint for both, ipsilateral and contralateral situations. Increased speed results in higher means regardless of whether the hip is ipsilateral or contralateral or of the gait labelling.

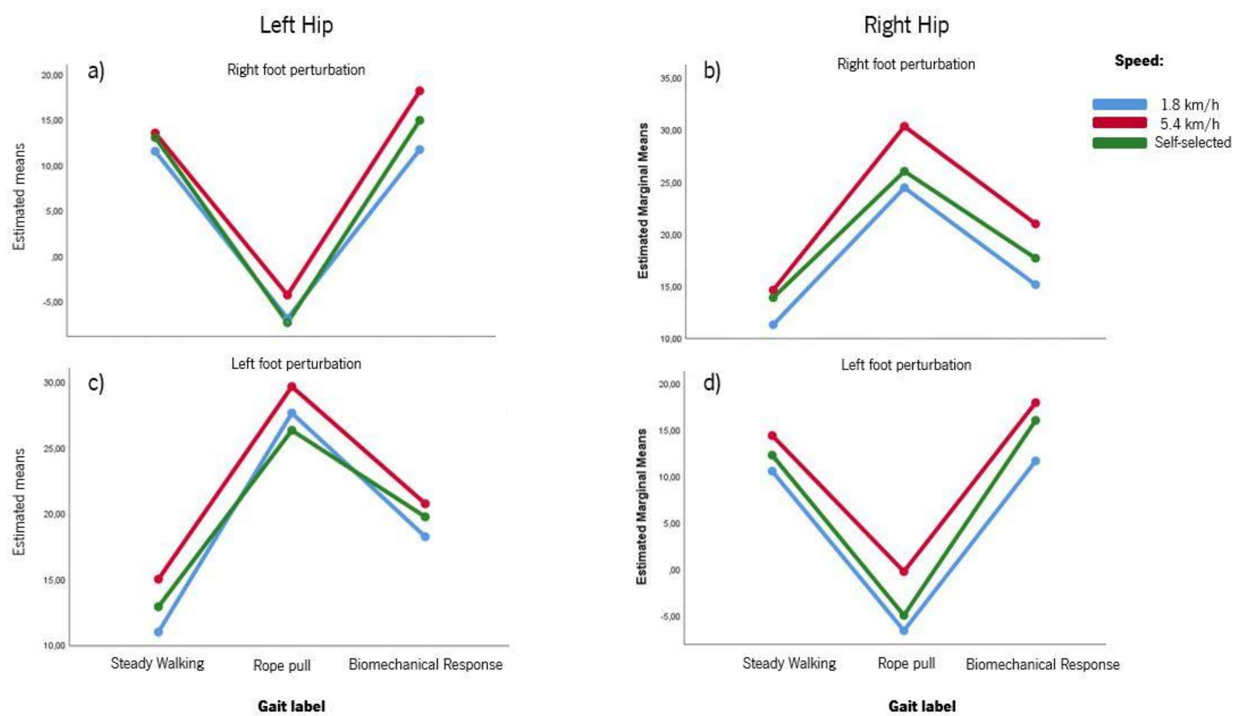


Figure 5.15 a) and c) Left and b) and d) Right Hip means, in sagittal plane, estimated during steady walking, rope pull and biomechanical response considering the interaction effect between perturbation, foot and speed. Angle increasing indicates a flexion movement while decreasing indicates an extension movement.

Concerning the knees' angles, Tukey-B post hoc speed analysis indicated that for both knees, differences appear between all the labels (steady walking, rope pull and biomechanical response). When analysing both knees plots, firstly these graphs, generally, support the main results obtained in the previous described models regarding the dominant movements presented in steady walking, rope pull and biomechanical response situations. However contralateral knee presents a reflexed behaviour for lower velocities. When considering the speed effect in the evolution of this graph, can be observed that for higher velocities, the means appear to be higher regardless of whether the knee is ipsilateral or contralateral or of the gait labelling.

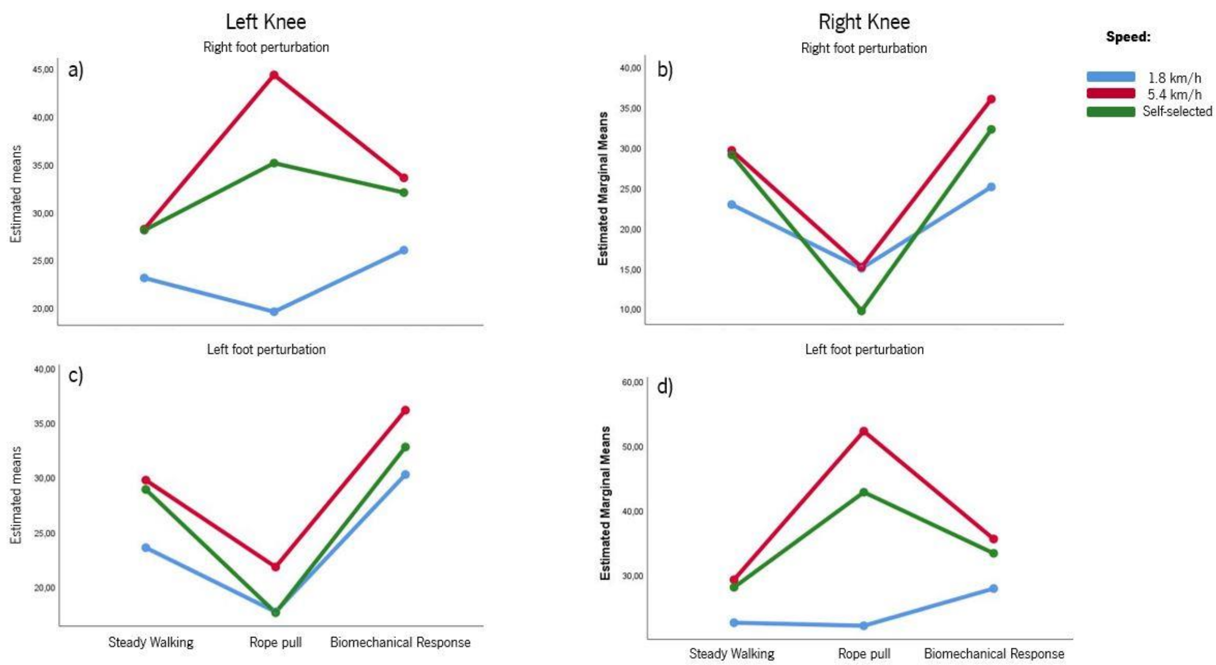


Figure 5.16 a) and c) Left and b) and d) Right Knee means, in sagittal plane, estimated during steady walking, rope pull and biomechanical response considering the interaction effect between perturbation, foot and speed. Angle increasing indicates a flexion movement while decreasing indicates an extension movement.

Regarding the speed influence in the graphs evolution in ankles' joints, it is not possible to find a consistent relationship between speed and mean values when analyzing the ipsilateral ankle. In opposition, when observing the ankle graphs in the contralateral situations, consistent results are finding for both left and right ankles. In this case, the speed, besides influence the way that the subject is influenced by the perturbation also the affects the subjects biomechanical response. Considering self-selected and 5.4 km/h speeds, the biomechanical response of the contralateral ankle is the same obtained from the model addressed in 5.4.1. However, for lower velocities (1.8 km/h), the response of the ankle in the contralateral side presents a reflexed response: when the perturbation is delivered this joint is plantarflexed and, during the biomechanical response, the movement of the contralateral ankle is mainly dorsiflexion. Tukey-B post hoc analysis considering the speed effect highlighted significant differences between the three gait labels for both, right and left ankles.

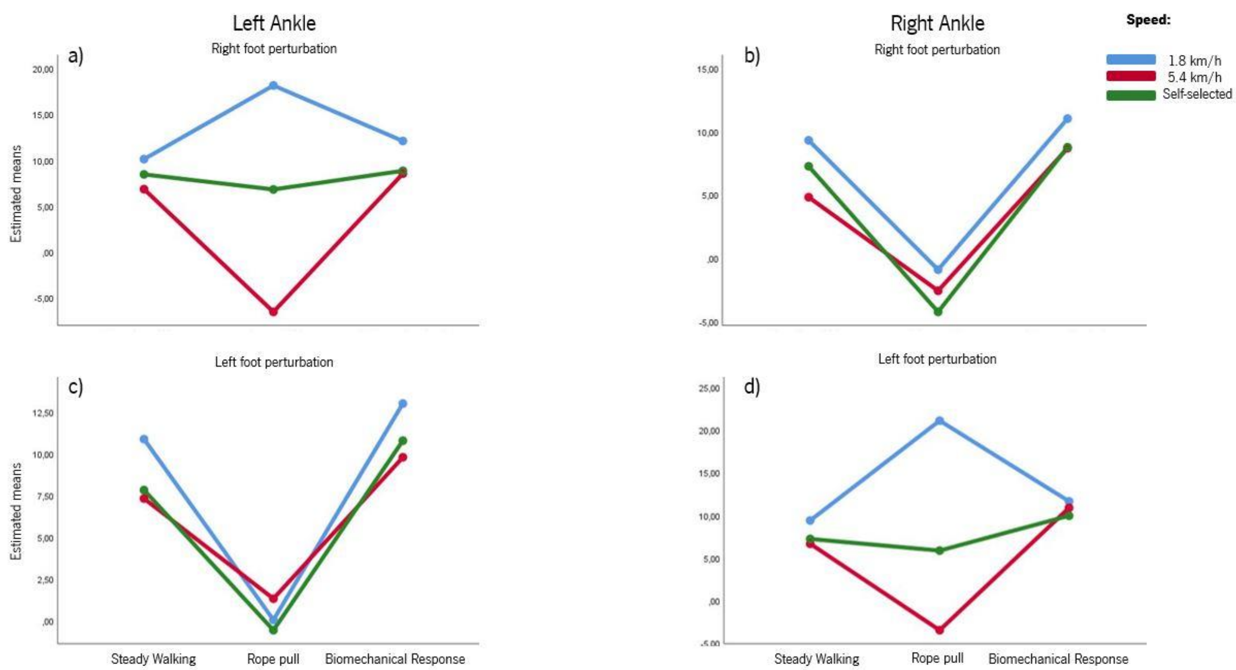


Figure 5.17 a) and c) Left and b) and d) Right Ankle means, in sagittal plane, estimated during steady walking, rope pull and biomechanical response considering the interaction effect between perturbation, foot and speed. Angle increasing indicates a plantarflexion movement while decreasing indicates a dorsiflexion movement.

Tukey-B post hoc tests determined that, considering both hips frontal movements, this difference is mainly caused by the perturbation variable than the interaction of both, perturbation, and speed. Regarding the post hoc speed, this test determined only statistically significant differences, in the case of the right hip in the situations with greater speed. Looking for the plots related to the hips' angles in the frontal plane, ipsilateral hips does not present consistent evolution when comparing both feet. In turn, in the contralateral hips, the graphs obtained corroborate the conclusions previously addressed for higher velocities while, for 1.8 km/h the contralateral hip presents a reflexed behaviour in the frontal plane.

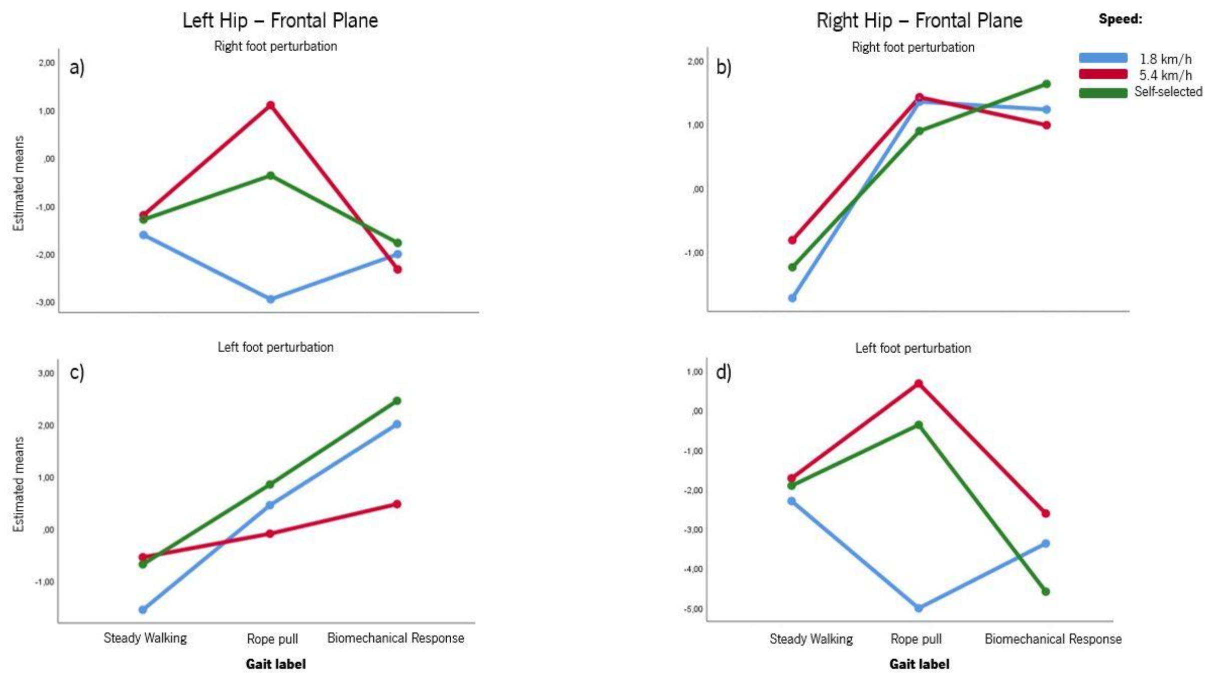


Figure 5.18 a) and c) Left and b) and d) Right Hip means, in frontal plane, estimated during steady walking, rope pull and biomechanical response considering the interaction effect between perturbation, foot and speed.

Concerning the distance between both feet, Tukey-B post hoc tests demonstrated statistical significance for both IV's, perturbation, and speed. As previously mentioned, the rope pull induced a greater distance between both feet, which was reduced with the biomechanical response. Higher speeds induced greater distance between both feet comparing with lower velocities. Also, and despite being not statistically significant considering the interaction between perturbation and speed, as expected, the COM velocity in the AP direction presented higher values for higher velocities, regardless the considered foot.

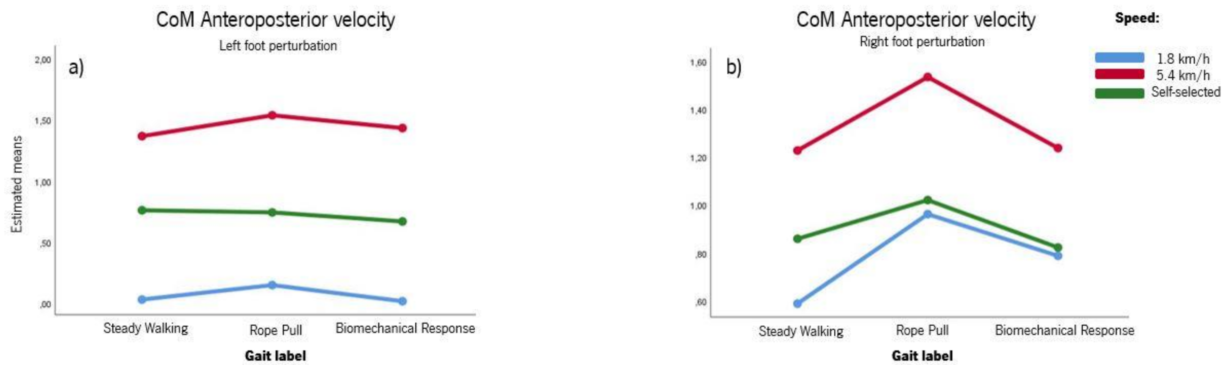


Figure 5.19 CoM velocity means, in AP direction, estimated during steady walking, rope pull and biomechanical response considering the interaction effect between perturbation, foot and speed. Angle increasing indicates an adduction movement while decreasing indicates an abduction movement.

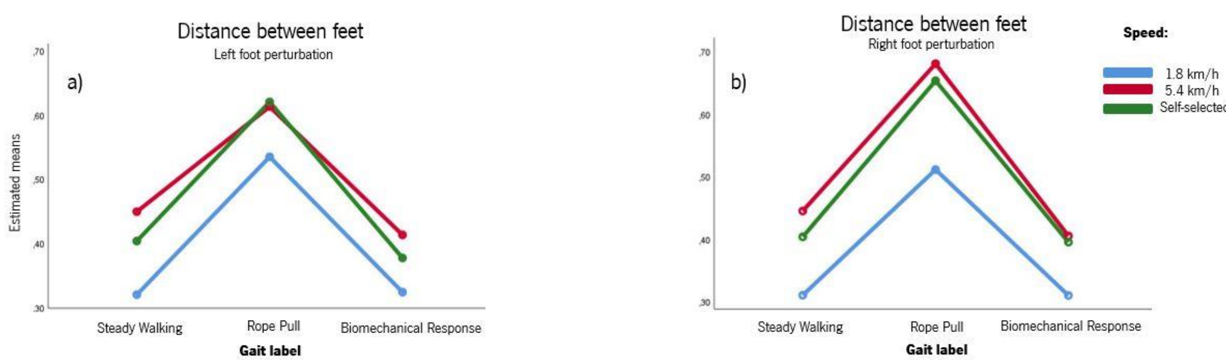


Figure 5.20 Distance between both feet for a) left and b) right perturbations.

5.3.4 ANOVA Perturbation vs Intensity

Regarding the interaction effect between perturbation, perturbed foot, and intensity studied with the data collected with the second protocol only the distance between CoM and left foot was statistically significant with a p-value less than 0.05. Although some DV's were not determined as significant variables, Tukey-B post hoc determined significant differences in some DV's depending on the intensity of the perturbation. These variables behaviour during perturbations at different intensities will be discussed in this subchapter. Regarding the hips movements in the frontal plane, Tukey-B post hoc determined that for severe perturbations the right hip present statistically significant differences in soft and severe perturbations, while left hip does not present statistically significant differences depending on the perturbation intensity. Considering the ipsilateral hip means in frontal plane, for both joints there is an increase in the mean during the rope pull and this value is then reduced during the biomechanical response indicating hip adduction. This result is in concordance with the result obtained in 5.4.1. For more intense perturbations the increase in the means during the rope pull is more evident while, in these cases, the right hip's biomechanical response presented higher RoM. In turn, regarding the contralateral frontal hip behaviour, the results obtained in this ANOVA model are consistent with the previous obtained in 5.4.1.

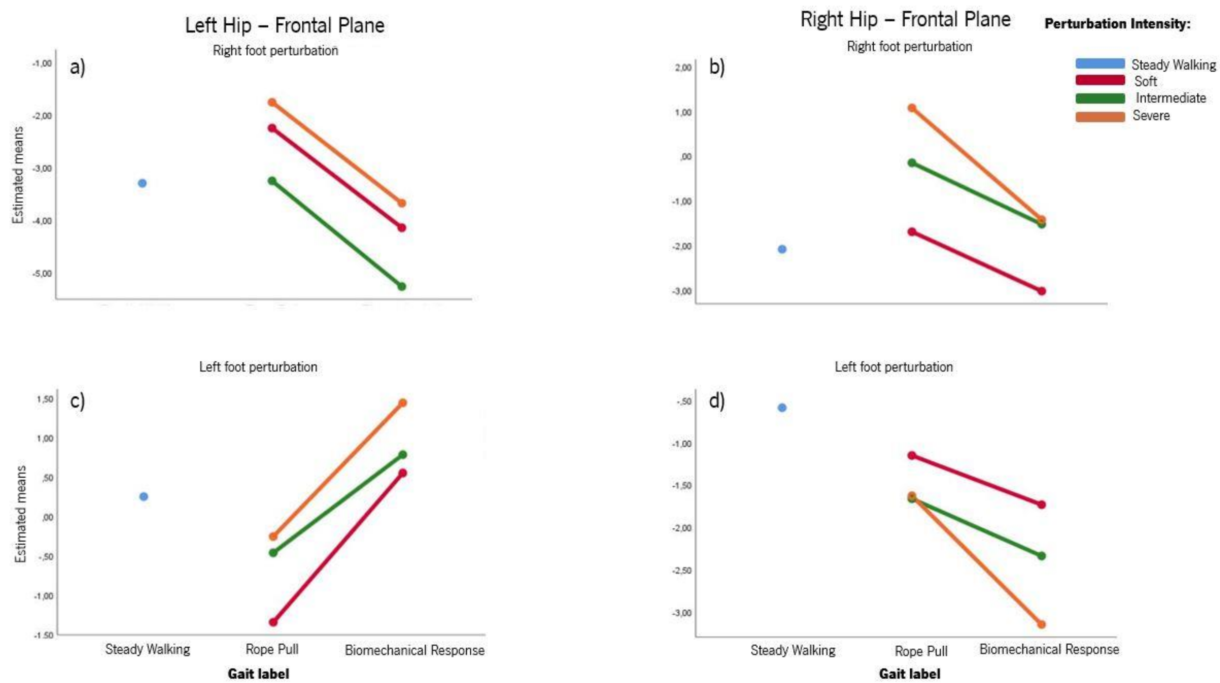


Figure 5.21 a) and c) Left and b) and d) Right Hip means, in frontal plane, estimated during steady walking, rope pull and biomechanical response considering the interaction effect between perturbation, foot and perturbation intensity. Angle increasing indicates an adduction movement while decreasing indicates an abduction movement.

Regarding the same joint in the sagittal plane, Tukey-B post hoc determined significant differences in left hip means during severe perturbations. When looking for the graphs, it is possible to conclude that the biomechanical response of both hips in the contralateral side is not modulated by perturbation intensity.

Conversely, right, and left ipsilateral hips presented higher flexion movements, during the rope pull, for more intense perturbations. In the right hip, the biomechanical response to more intense perturbations also presented a greater RoM.

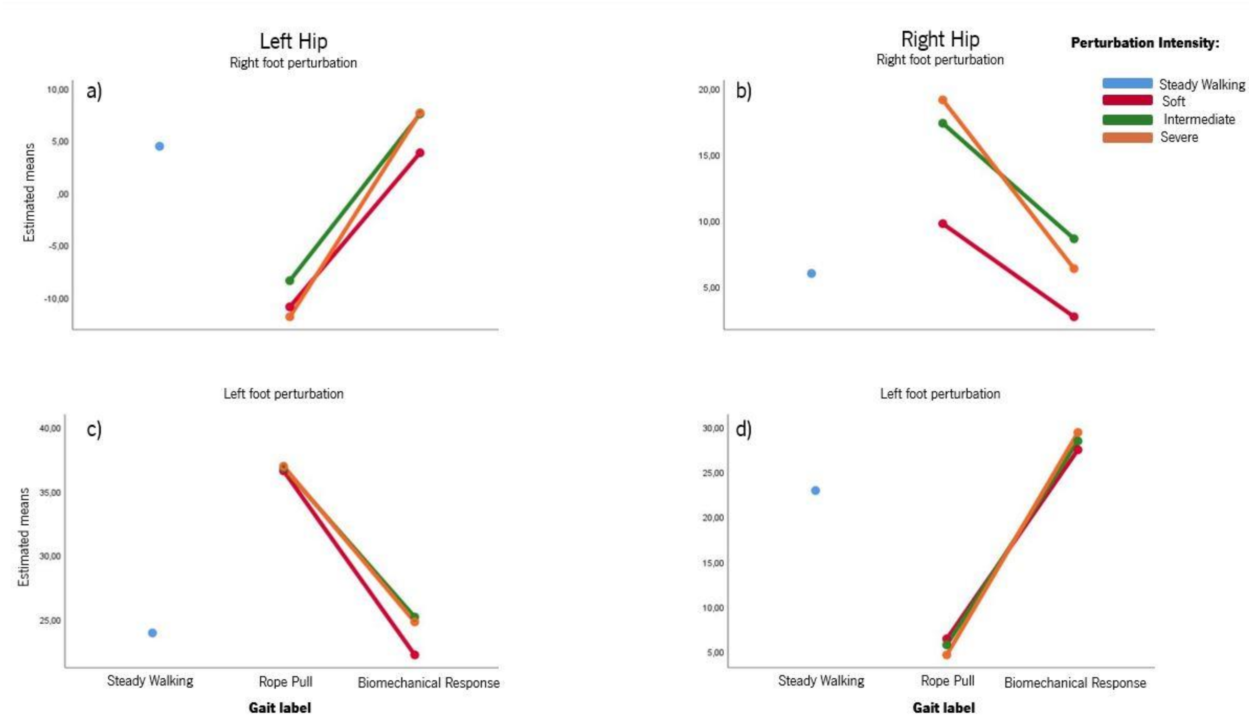


Figure 5.22 a) and c) Left and b) and d) Right Hip means, in sagittal plane, estimated during steady walking, rope pull and biomechanical response considering the interaction effect between perturbation, foot and perturbation intensity. Angle increasing indicates a flexion movement while decreasing indicates an extension movement.

Concerning the knee's means behaviour during perturbations, regarding the right knee, the three perturbations intensity did not cause statistical significance revealed by Tukey-B post hoc, while the left knee presented statistical difference for both, severe and soft perturbations. Graphics analysis of this joint behaviour show that both knees only had small difference depending on perturbation intensity when placed ipsilaterally to the perturbation. In turn, when in the ipsilateral side this joint biomechanical response appears to be intensity dependent in the right knee graph: the more intense the perturbation the more evident the mean value reduction and the biomechanical response presents higher RoM.

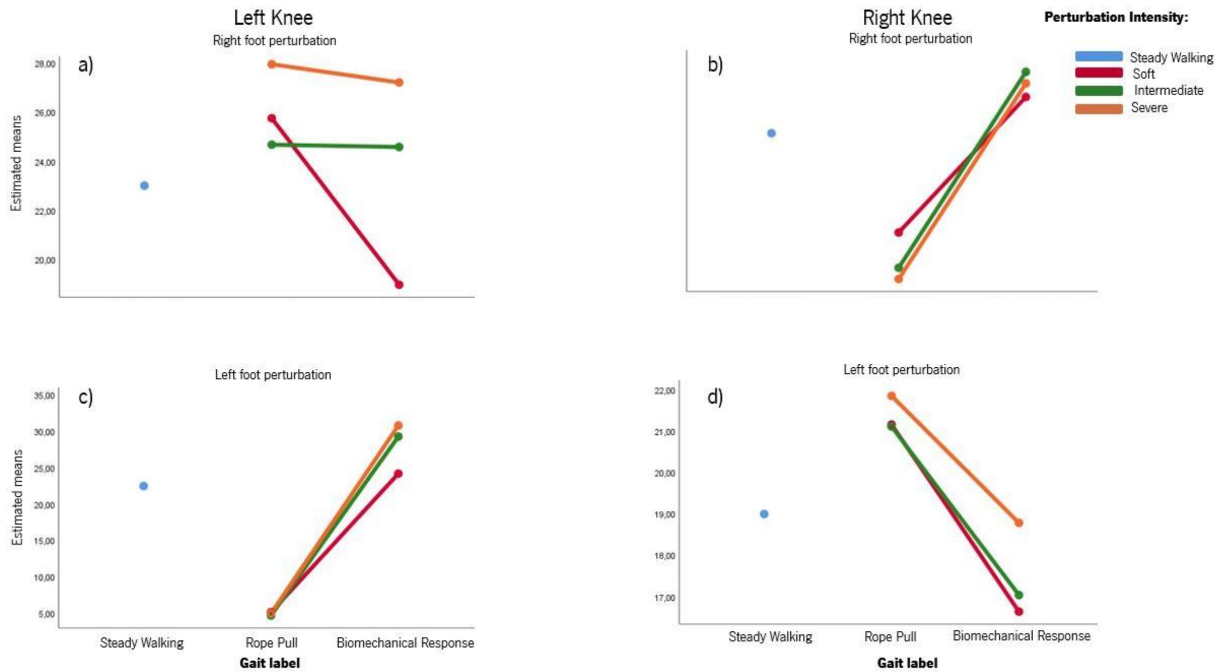


Figure 5.23 a) and c) Left and b) and d) Right Knee means, in sagittal plane, estimated during steady walking, rope pull and biomechanical response considering the interaction effect between perturbation, foot and perturbation intensity. Angle increasing indicates a flexion movement while decreasing indicates an extension movement.

Regarding left and right ankles behaviour, when the perturbation is given ipsilaterally to this joint, there are no major differences in their evolution, being characterised by a plantarflexion movement. In turn, when the perturbation is given in the contralateral side, the results obtained do not present a defined pattern.

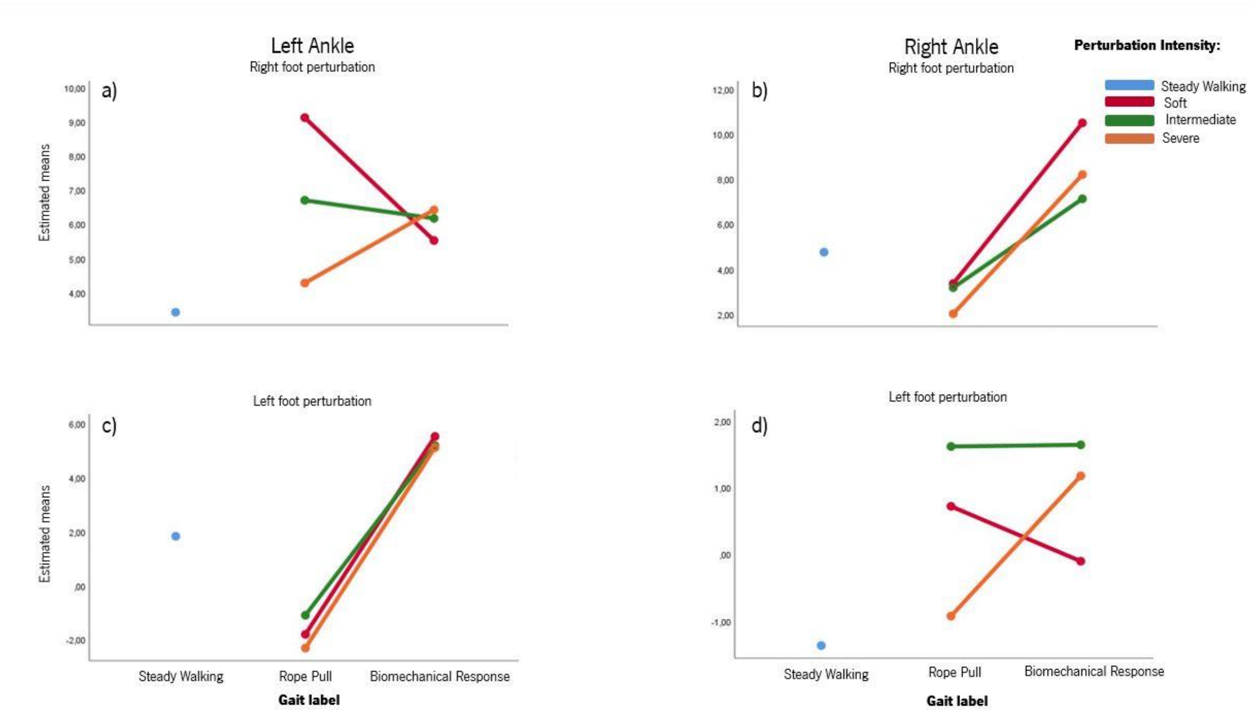


Figure 5.24 a) and c) Left and b) and d) Right Ankle means, in sagittal plane, estimated during steady walking, rope pull and biomechanical response considering the interaction effect between perturbation, foot and perturbation intensity. Angle increasing indicates a flexion movement while decreasing indicates an extension movement.

Finally, concerning the foot distance, Tukey-B post hoc determined statistically significant differences for intermediate and severe perturbations. As expected, for both left and right feet, more intense perturbations induced a greater distance between both feet.

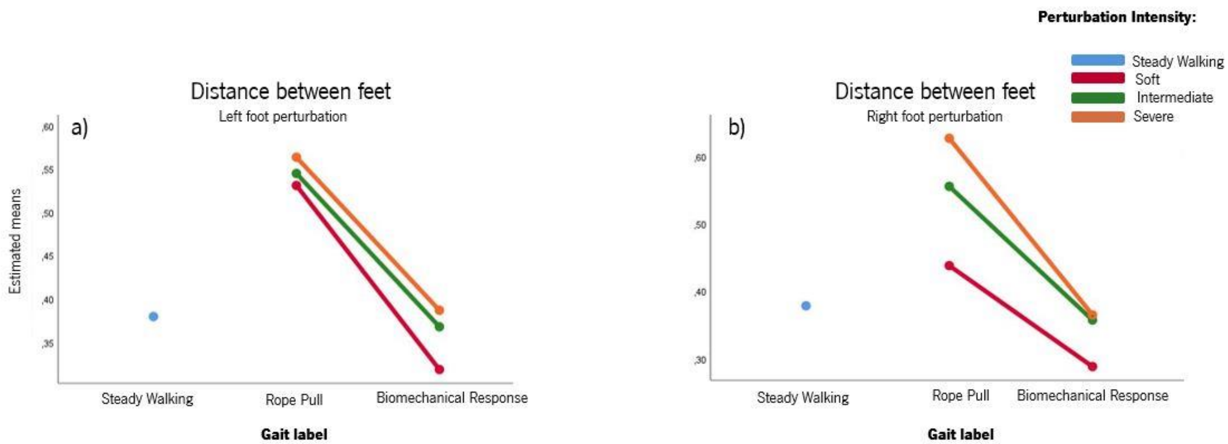


Figure 5.25 Foot distance means estimated during steady walking, rope pull and biomechanical response considering the interaction effect between perturbation, perturbed foot and perturbation intensity. Angle increasing indicates a flexion movement while decreasing indicates an extension movement.

5.3.5 ANOVA Perturbation EMG

Concerning the interaction effect between perturbation and perturbed foot in the EMG variables, p values determined statistical significance for: i) Right and left BF excitatory responses; ii) Right and left RF inhibitory responses; iii) Right RF excitatory response; iv) Right and left GL excitatory responses and v) Right and left GL inhibitory responses. Consistent results were obtained for BF as both, right and left BF, excitatory responses presented statistically significance while, in opposite, both inhibitory responses were determined as statistical insignificant. Despite the increase of both, right and left BF excitatory responses for right and left-side perturbations, the former perturbations resulted in greater increase in the excitatory response of right and left BF.

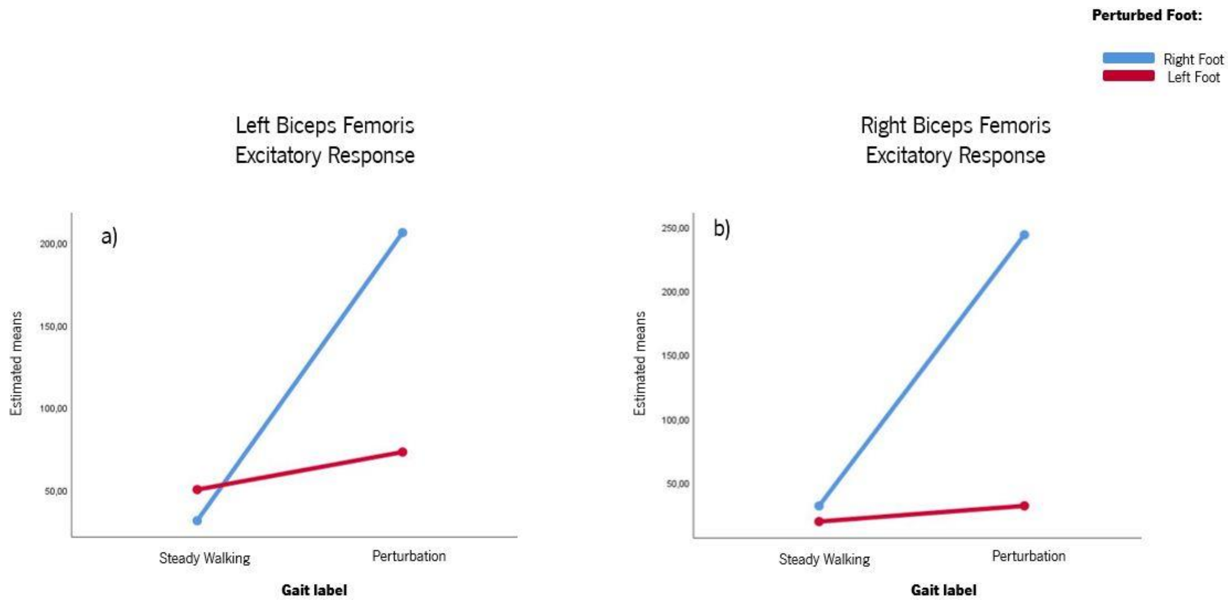


Figure 5.26 a) Left and b) Right Biceps Femoris excitatory response means estimated during steady walking and perturbation situations.

In turn, both RF inhibitory means variations resulted in statistically significant differences while only the right RF excitatory response presented statistically significance when performing ANOVA analysis. Concerning the inhibitory responses, perturbations delivered to the left foot resulted in more powerful inhibitory responses in both, right and left RF. Conversely, right foot perturbations resulted in evident excitatory responses in right RF.

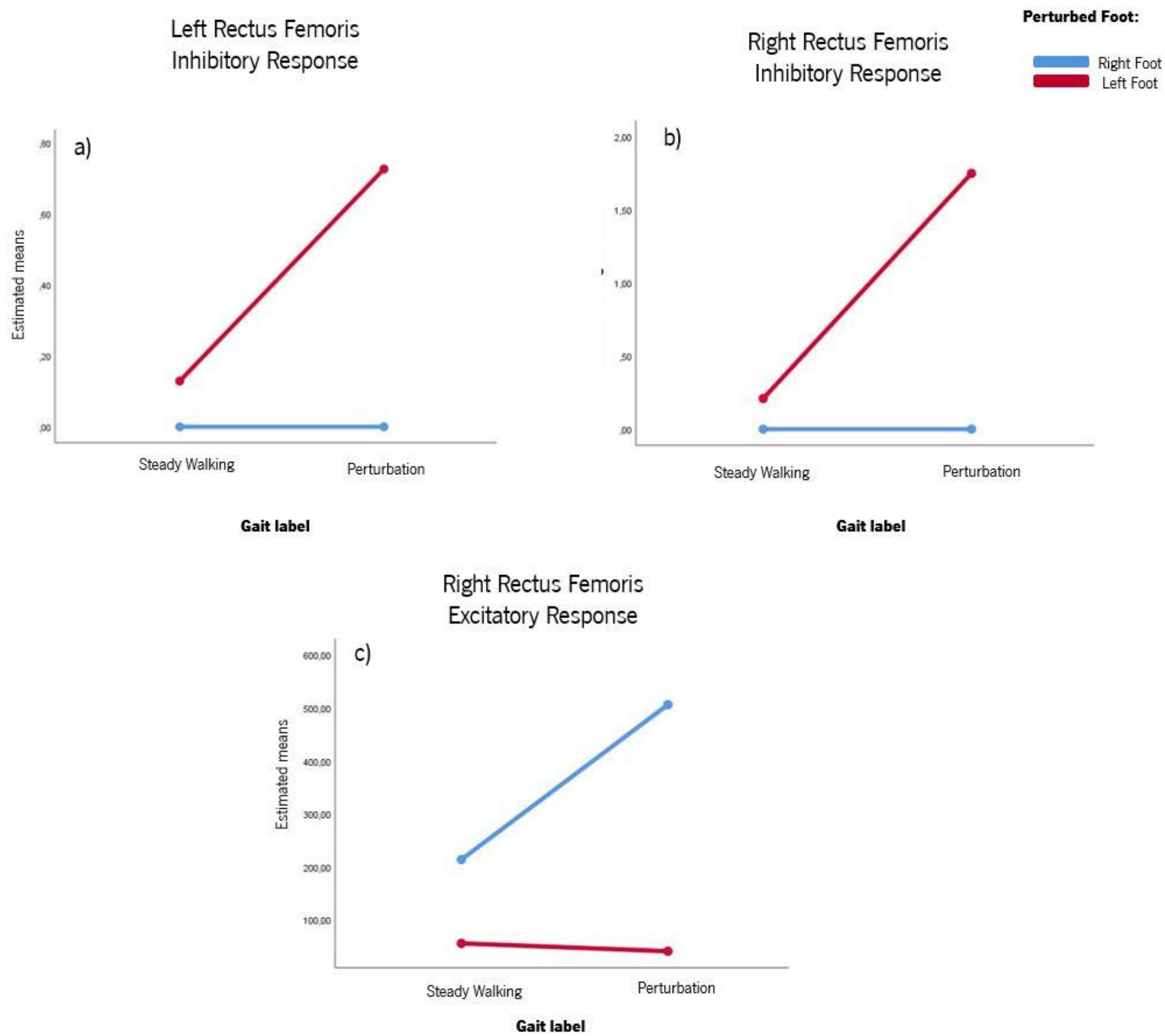


Figure 5.27 a) Right and b) Left Rectus Femoris inhibitory responses and c) Right excitatory responses means estimated during steady walking and perturbation situations.

Concerning the GL excitatory and inhibitory responses, both were determined statistically significant for right and left GL. Similarly to the results obtained for BF, right foot perturbations induced higher increase in the GL muscle power means. Also, both inhibitory responses were more powerful during perturbations delivered in the right limb. Despite the same tendency, Right GL excitatory and inhibitory responses were associated with higher means comparing with left GL. Finally, TA excitatory responses did not result in statistically significant differences while there were no inhibitory responses from this muscle.

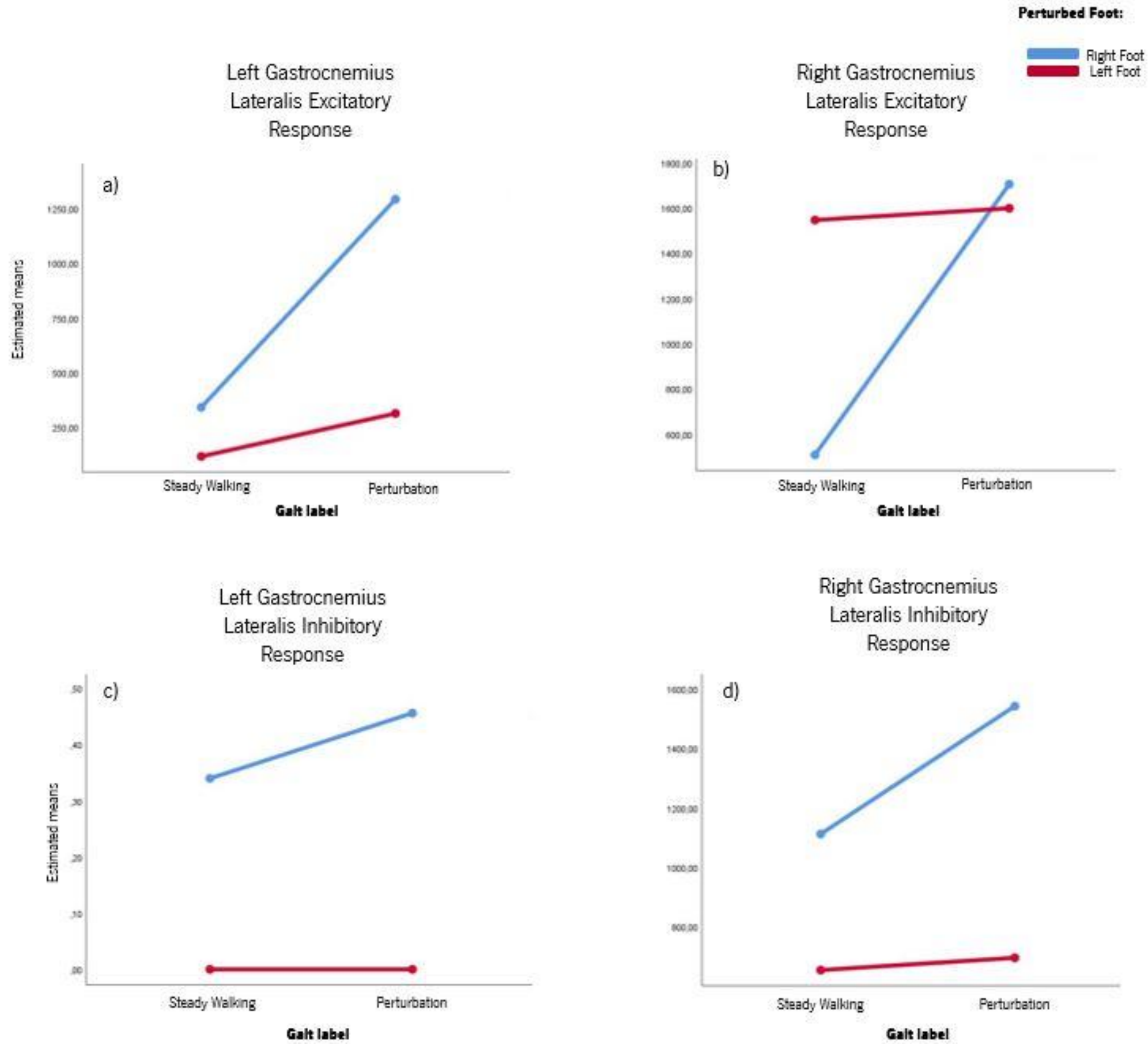


Figure 5.28 a) Right and b) Left Gastrocnemius Lateralis excitatory and c) and d) inhibitory responses means estimated during steady walking and perturbation situations.

Concerning the latency periods analysed for the four muscles included in this study, ANOVA did not find statistically significant differences considering the interaction effect between perturbation label and perturbed foot. Latency periods with the scientific literature addressed in Chapter 3 comparison are shown in Figure 5.29. Only non-hazard were included in this comparison, as during the experimental trials no subjects fell.

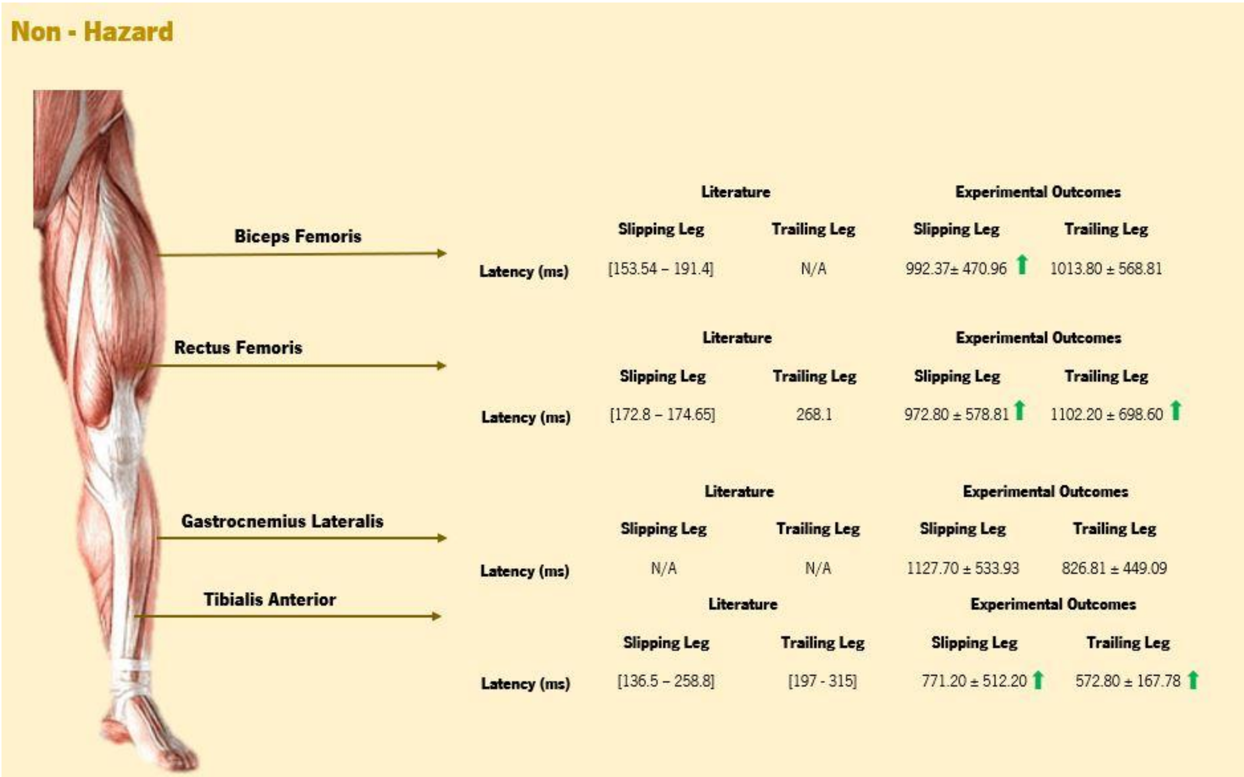


Figure 5.29 Muscle latency periods comparison between experimental data and scientific literature.

5.3.6 Variable Ranking

As mentioned before, one of the aims of the analysis covered in this chapter involves obtaining quantitative values in order to understand which variables related to joint means stand out the most in the response to a slip-like perturbation. Two different methods were followed, and the results are introduced in this subsection. In Table 5.7 the results obtained concerning to the eta partial are shown according to the ANOVA model performed.

Table 5.7 Partial Eta squared values per ANOVA model for all the dependent variables. Small, medium, and large effect sizes are, respectively, shading at orange, yellow and green

Dependent Variable	***Perturbation vs Foot***	***Perturbations vs Perturbed foot vs Inclination***	***Perturbations vs Perturbed foot vs Speed***	***Perturbations vs Perturbed foot vs Intensity***
	Partial Eta squared	Partial Eta squared	Partial Eta squared	Partial Eta squared
Right Hip Frontal AVG	0.0765	0.0046	0.0284	0.040
Left Hip Frontal AVG	0.0409	0.0051	0.0172	0.0011
Right Hip Sagittal AVG	0.6081	0.0049	0.0186	0.0006
Left Hip Sagittal AVG	0.6123	0.0070	0.0103	0.0048
Right Knee Sagittal AVG	0.2773	0.0144	0.0629	0.0136
Left Knee Sagittal AVG	0.1115	0.0256	0.0451	0.0333
Right Ankle Sagittal AVG	0.0595	0.0025	0.0602	0.0120
Left Ankle Sagittal AVG	0.0441	0.0067	0.0522	0.0097
CoM x velocity	0.0016	0.0000	0.0012	0.0000
CoM y velocity	0.0014	0.0004	0.0005	0.0000
CoM z velocity	0.0006	0.0004	0.0016	0.0000
3D Foot Distance	0.0035	0.0040	0.0261	0.0040
Distance CoM - Right Foot	0.0000	0.0000	0.0022	0.0433
Distance CoM - Left Foot	0.0005	0.0000	0.0037	0.0322

Analyzing the results obtained by the study of the partial eta squared values it is possible to observe that in the case of the two-way ANOVA Perturbation *versus* Perturbed Foot the generality of the values obtained are higher comparing to the other ANOVA models, allowing to perceive that the interaction of these IV's is strongly responsible for the significant variation of the means of the DV's investigated. Mean joint angle values in the sagittal plane are the variables most influenced by the interaction effect of perturbation and perturbed foot. Within these variables, the hip joint means stand out as the variables whose change in mean value was most evident, followed by the knee and ankle joints, respectively. Concerning the remaining models, the model where the interaction effect between perturbation and gait speed was analysed, the mean values of both knee and ankle joints in the sagittal plane, which presented higher values in comparison to the others, should be highlighted. The variable ranking was further obtained through the FSM presented in Table 5.8.

Table 5.8 Feature Selection Methods results

Method	*Foot Distance*	*Left Ankle*	*Distance CoM – Right Foot*	*Right Hip Frontal*	*Right Hip*	*Right Knee*	*COM velocity z direction*	*Distance CoM – Left Foot*	*Left Knee*	*Right Ankle*	*COM velocity y direction*	*Left Hip*	*Left Hip Frontal*	*COM velocity x direction*
ILFS	1.00	0.71	0.58	0.21	0.79	0.04	0.00	0.13	0.33	0.63	0.75	0.29	0.08	0.46
InfFS	0.71	0.58	0.96	0.04	0.25	0.21	0.88	1.00	0.33	0.29	0.08	0.17	0.13	0.00
ECFS	0.71	0.58	0.96	0.08	0.29	0.17	0.88	1.00	0.33	0.25	0.00	0.21	0.13	0.04
MRMR	0.96	0.38	0.04	0.67	1.00	0.46	0.88	0.00	0.29	0.25	0.71	0.79	0.54	0.63
Relieff	0.58	0.50	0.33	0.88	0.96	0.79	0.63	0.00	0.83	0.71	0.67	1.00	0.92	0.75
MCFS	0.96	0.46	0.58	0.71	0.29	0.54	0.00	0.63	0.33	0.67	0.79	0.08	0.75	0.83
UDFS	0.92	0.75	0.04	0.71	0.17	0.79	0.63	0.08	0.67	0.54	0.83	0.38	0.58	0.50
LLCFS	0.00	0.83	0.96	0.33	0.46	0.50	0.25	1.00	0.67	0.92	0.17	0.75	0.29	0.21
CFS	0.75	1.00	0.25	0.79	0.67	0.58	0.54	0.42	0.29	0.50	0.92	0.71	0.04	0.08
FSASL	0.08	0.79	0.92	0.96	0.29	0.88	0.58	0.83	0.67	0.17	0.21	0.25	0.63	1.00
USFOL	0.88	0.83	0.38	0.79	0.58	0.75	0.25	0.00	0.63	0.71	0.13	0..17	0.42	0.67
Lasso	1.00	0.17	0.42	0.54	0.83	0.75	0.38	0.46	0.71	0.13	0.88	0.79	0.92	0.08
PCA	0.00	0.46	0.96	0.25	0.38	0.42	0.92	1.00	0.33	0.54	0.17	0.50	0.29	0.21
Total	8.54	8.04	7.38	6.96	6.96	6.88	6.79	6.54	6.42	6.29	6.29	6.08	5.71	5.46
Mean	0.66	0.62	0.57	0.54	0.54	0.53	0.52	0.50	0.49	0.48	0.48	0.47	0.44	0.42
Std	0.37	0.22	0.34	0.30	0.28	0.26	0.31	0.41	0.19	0.24	0.34	0.29	0.30	0.32

Regarding the results of the FSM, of all the DV's considered, the distance between feet was the one that presented the highest sum value, thus being the one that could better distinguish between steady walking, rope pull and biomechanical response situations. Concerning the DV's related with the joint angles' means, the left ankle appears in first place. The DV's related to the right hip also appear in the first positions, namely in the fourth and fifth positions. Making a distinction between the left leg joints and the right leg joints, in the case of the right leg the descending order of the sum value of these variables is hip, knee and ankle. On the other hand, for the left leg, the descending order is inverted, with the ankle coming first, followed by the knee and the hip, respectively.

5.4 Discussion

The analysis presented in this chapter, allowed to understand the biomechanical response to slip-like perturbations of the subjects enrolled in both experimental protocols. The biomechanical response to this type of perturbations was analysed under different gait conditions, namely for different speeds and inclinations on a treadmill. This response was also studied for different perturbation intensities and depending on which leg was perturbed. Concerning the main results of this analysis, the distinct analysis between the leading leg and the trailing leg movement, in 5.4.1, allowed to understand the different movements in both legs when a perturbation occurs. During the rope pull, the ipsilateral leg's movements of the hip, knee and ankle are flexor, extensor and dorsiflexor dominant respectively. All these movements were expected as a result of the movement induced by the pulling of the foot by the rope. Thus, it is possible to conclude that the rope pull was reflected in a change of movement in all joints compared to steady walking situations. In turn, regarding the behaviour of the unperturbed leg during the rope pull, the movements of the hip, knee and ankle joints were respectively extensor, flexor and dorsiflexor dominant. These findings are explained by the fact that during the heel strike of the perturbed leg, the contralateral leg is starting its swing phase. Regarding the biomechanical response of the hip joint, after the rope pull ending, the hip movements are dominantly flexor and extensor in the unperturbed and perturbed leg, respectively. These findings are concordant with the scientific literature [25], [34], [35], [37], [46] and [56] [25]. The extension of the perturbed leg's hip allows counteracting the destabilizing effect caused by the perturbation by allowing the perturbed foot to be placed in a position closer to the CoM. On the other hand, the flexion movement of the unperturbed leg's hip allows the two feet to be brought closer together, thus increasing stability and making it possible for the individual to resume steady walking as referred in [10]. Although in agreement with [10], most of the literature highlights unperturbed hip extension as a movement to proceed a step back strategy in order to compensate for the posterior displacement of the CoM. The fact that, in the analysis carried out, there was no need of the subjects to perform this strategy allows to infer that accordingly to the perturbations' intensity, narrow strategy, addressed in [26], was capable of avoiding a fall. Concerning the knees biomechanical response, the flexor movement of the knee of the perturbed leg, besides allowing the perturbed foot to be placed in contact with the floor in a position closer to the CoM also decelerates the movement of this leg allowing energy absorption. Similar conclusions were presented in [34], [35], [37], [46] and [56]. Conversely, knee extension of the unperturbed leg also contributed to the placement of the unperturbed foot closer to the one that had been disturbed, thus placing the BoS near the behind the CoM. [32] [34] [37] The ankle joint naturally contributed actively to promote the contact of the feet with the ground. The dominant movement in the ankle of the perturbed leg was a plantarflexion movement similarly to [34] and [37]. Conversely, the increased means of the unperturbed leg can be explained by the strategy already discussed, to place both feet together. As aforementioned, during the perturbation the contralateral ankle is in an initial stage of its swing phase being this joint slightly flexed. After the perturbation, increased ankle plantarflexion allows subjects to place the foot in contact with the ground in a position closer to the perturbed foot to increase the BoS to increase stability [10], [27]. Finally, regarding the movements in the frontal plane, the contralateral hip, a strategy to approximate both feet by hip adduction can be highlighted. Regarding the behaviour of the ipsilateral hip during the biomechanical response left hip presented an abduction movement that can be justified by a strategy developed to increase the BoS.

Concerning the speed's effect during the biomechanical response to slip perturbations, this IV results in higher means in hip and knees joints regardless of whether the joints are ipsilateral or contralateral or of the gait labelling, thus indicating that for higher speed trials, hips and knees tend to be more flexed. When studying the speed influence in ankles joints, two different responses can be distinguished depending if the ankle is on the ipsilateral or in the contralateral side of the perturbation. Considering the ipsilateral ankle, the same response as concluded in 5.4.1 is presented. In opposition, when observing the ankle graphs in the contralateral situations, the speed, besides influencing the way that the subjects are perturbed also affects their biomechanical response. For higher velocities, the biomechanical response of the contralateral ankle is the same obtained from the model addressed in 5.4.1. However, for lower velocities, the response of the ankle in the contralateral side presents a reflexed response: when the perturbation is delivered, this joint is plantarflexed and, during the biomechanical response, the movement of the contralateral ankle is mainly dorsiflexion. This difference can be explained, by the fact that, for lower velocity trials, the HS of the perturbed limb coincides with an earlier stage of the propulsive phase of the contralateral limb: since the subjects were conditioned by the speed of the treadmill, a lower speed caused a longer flat-foot phase in this limb explaining higher means in the plots. For this reason, when the perturbation was delivered, subjects took advantage of the fact that the contralateral foot had a greater contact area with the ground and the strategy was to increase this area through dorsiflexion. Differential contralateral knee responses were also determined for 1.8 km/h: in contrast with the knee response for higher velocities, lower velocities were associated to an extensor dominant movement during the perturbation. The extensor dominant movement for the contralateral knee confirms the conclusions previously addressed: slip perturbations at lower velocities coincide with an earlier stage of the contralateral foot's stance phase. The graphs plots related to the hips' angles in the frontal plane, do not present consistency when comparing both feet. This inconsistency can be explained by the fact that all the subjects enrolled in the performed trials were right dominant.

Higher speeds induced greater distance between both feet comparing with lower velocities. As aforementioned, at higher velocities subjects presented shorter periods in double stance making easier to the operator to induce more evident perturbations by pulling the rope causing greater feet distance. When considering the inclination effect, in the case of 10° inclination, the knees' means are greater indicating that during these trials knee appeared more flexed in the three gait labels situations. The response of the contralateral knee appears to be modified for 10° of inclination, presenting a biomechanical response with less averages variation. Concerning the perturbation intensity effect, as expected, more intense perturbations induced a greater distance between feet. During the perturbation, also the flexion movements of both knees were more pronounced confirming the previous result. Literature also confirms a greater distance between CoM and BoS for more intense perturbations [25]. Regarding the biomechanical response, right hip and left knee movements during these periods were associated to higher RoM values. These findings can be explained by the fact that, stronger perturbations cause a more destabilizing effect and thus, the biomechanical response, present higher RoM to compensate the perturbation effect.

Concerning the interaction effect between perturbation and perturbed foot in the EMG variables, ANOVA determined statistically significant different for both, right and left BF excitatory responses. This muscle is responsible for hip extension and knee flexion movements [40]. Increased excitatory response means during perturbations can be confirmed with the extension strategy performed by the perturbed limb's hip joint and the flexion strategy performed by both knees, as previously referred. Unexpectedly, right perturbations were associated with more powerful responses in both, right and left, BF thus highlighting the need to include left-handed subjects in the coming protocols to conclude about the dominance effect.

In turn, GL inhibitory and excitatory responses for both limbs were determined as statistically significant. These muscles intervenes in ankle plantarflexion and knee flexion [40], which are the main kinematic strategies performed by unperturbed limb. As referred in [14] and [25], during biomechanical response to slip perturbations, this muscle helps stabilizing the ankle joint to promote the contact between foot and floor, which is a crucial strategy to increase the BoS. The results obtained in the analysis performed in this Chapter confirm the relevance of this muscle, in controlling ankle's movement, during the biomechanical response after a slip-like perturbation.

Regarding the RF, right and left inhibitory responses were found statistically significant. This muscle is mainly responsible for knee extension [40], thus ANOVA results allow to conclude that knee extension is inhibited during the biomechanical response to slip perturbations. This finding is concordant with the kinematic conclusions previously addressed which highlight a dominant knee flexion movement in slip biomechanical responses of the perturbed limb. In turn, knee extension, is the dominant movement induced by the rope pull and RF appears to actively inhibit this movement, also allowing BF to flex the knee during the recovery process. Right RF excitatory responses was also significant in the analysis conducted. This finding can be explained by two motives, that should be further studied: i) Right RF excitatory response can be influenced by subject dominance and ii) knee extension is the primary response of the unperturbed limb and the secondary strategy of the perturbed limb to allow steady walking resumption after a slip-like perturbation evidenced in literature [28].

Finally, during the EMG data analysis performed in this Chapter, ANOVA did not find statistically significant differences considering the interaction effect between perturbation label and perturbed foot, in muscular latency periods. Also, the latency periods determined in the experimental analysis conducted were greater to the ones presented in scientific literature as shown in Figure 5.29. Literature, related to this topic, highlights significantly different latency periods when comparing steady walking with perturbation situations [16], [19], [28], [55], [67] . One of the possible causes for this lack of concordance between scientific literature and the analysis performed was the time gap between the HS and the induced perturbation onset, as these events were not simultaneous in most of the situations. As the latency periods were obtained considering the HS instant, these may not correspond to the real latency periods. Another possibility can be associated to the use of the rope to provoke the perturbations while in literature, these are provoked by controlled platforms. For this reason, the muscular response may be altered since the application of an external force may result in muscular responses to counteract it.

The analysis presented in this chapter was also complemented with two different methods for variable ranking, namely eta partial squared and FSM. Thus, these two main points addressed in this chapter allowed to understand the natural response of the human slip-like perturbations, and to collect a set of important information for the future development of devices for the prevention of slip related falls.

Regarding the partial eta squared, this statistical parameter allowed to conclude that the interaction of IV's responsible for the greatest variation of the DV's was the interaction between perturbation and perturbed foot. Table 5.7 allows to confirm that the additional interaction of these two IV's with one of the remaining variables is not responsible for a considerable variation of the DV's. Concerning the partial eta squared values obtained for the DV's, joint angles averages in the sagittal plane stand out as the variables with the greatest variations. The results obtained are also consistent considering that there is an order in the values obtained: i) first, both hip joints; ii) second, both knees and iii) both ankle joints. Right-sided knees and ankle joint showed greater changes compared to the left-sided joint. In turn, in the case of the hip, the values are quite similar. The analysis of these values allows to conclude that, among all joints, the hip is the joint that suffers more alteration in its average values when subjected to a perturbation, thus allowing to infer that, among all joints, this is the one that intervenes more actively in the biomechanical response to a slip-like perturbation. In contrast, ankle joint appears to be less relevant during the biomechanical response to a slip-like perturbation being this finding also highlighted in some literature [27], [54].

Finally, in relation to the FSM, these allowed to conclude that the variable with the greatest capacity to distinguish steady walking, rope pull, and biomechanical situations was the distance between both feet. Considering the statistical analysis previously carried out, this result is expected, as in all the interactions of IV's studied, rope pull induced an increase in the mean of this variable, which was then reduced during the biomechanical response. Concerning the joins ranking, obtained by FSM, excluding the left ankle, the joints on the right side appeared in lower positions compared to the same joints on the left side. Additionally, on the left side, the ranking resulted in an order of joints from the lowest to the highest, and this order was reversed on the right leg. These results may be explained by two situations: firstly, all the subjects enrolled in the protocol were right-handed, which may explain the appearance of right leg joints in lower positions. On the other hand, these methods were performed without considering the perturbation side, and this analysis may also be influenced by that reason.

6 Orthosis requirements and motor selection

The goal of this chapter is to find quantitative target specifications for the future development of a robotic wearable device to prevent slip-related falls, considering all conceivable joint in which actuation can be created. The target specifications defined in this chapter were obtained from the slip-induced experimental data addressed in Chapter 5. For this reason, the target specifications are primarily intended to mimic the human response to slip-like perturbations.

6.1 Introductory Insight

Quantitative definition of target specifications based on previously collected fall data allows to customise the device's development for its specific purpose. The formulation of these specifications for fall slip-like perturbations enables purpose-oriented development of the device from its early development phases, ensuring accurate component selection. This allows, firstly to avoid excess of weight or size and consequent restrictions in the natural gait of the user in wearable devices, and second to guarantee the correct operation of the device in its function. This chapter is divided into two major tasks. Firstly, the method used to determine the torques involved in the biomechanical response of the collected data will be explained. Secondly the target specifications for the development of a robotic device capable of preventing slip-related falls will be presented. Besides the more global specifications inherent to the development of all medical devices, quantitative data obtained through the treatment of the collected data will also be presented for each lower limb's joint.

6.2 Methods

6.2.1 CusToM toolbox

In the following subsections, will be presented the methods employed to obtain the target specifications for the future development of a wearable robotic device for slip-related falls prevention. As shown in Figure

6.1 the first tasks were related to the GRF and torque estimation, using CusToM toolbox, and the validation of this tool. Thereafter, the target specifications for the device to be developed were defined.

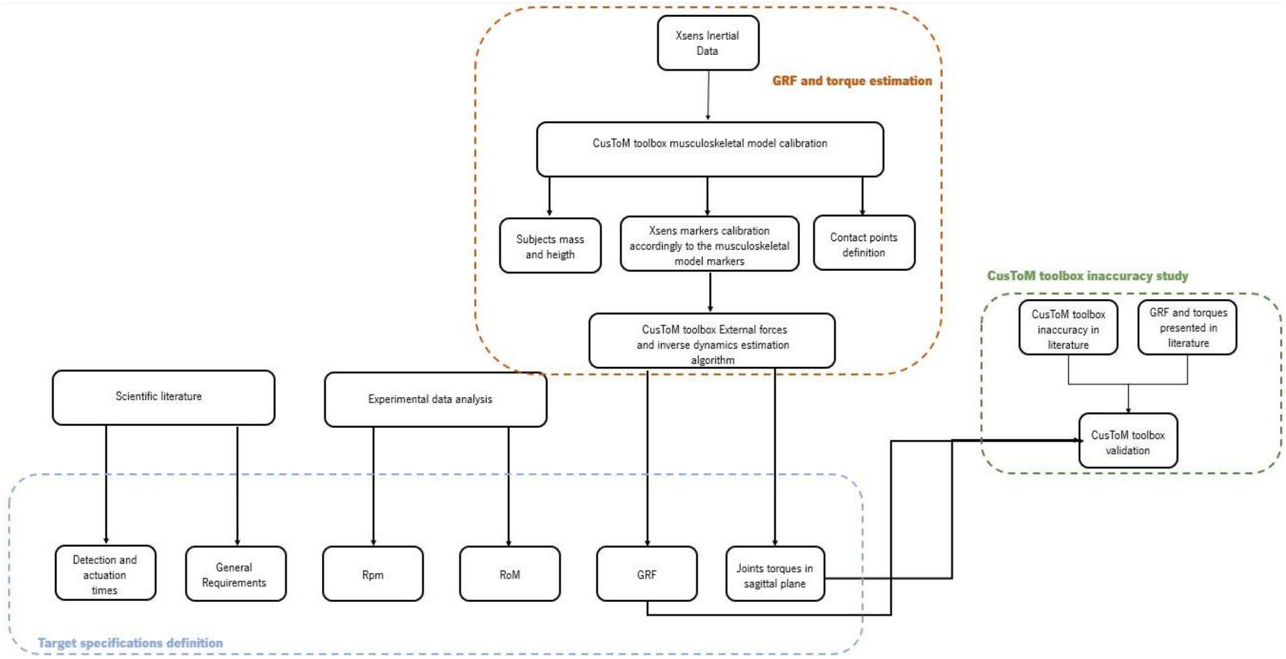


Figure 6.1 Chapter 6 methods overview.

6.2.1.1 GRF and torque estimation from Inertial Data

CusToM Matlab toolbox [95] was used to estimate the GRF as well as the torques involved in each of the joints, from the inertial data obtained by Xsens, during the previously addressed protocols [96]. Generally, CusToM toolbox allows to conduct inverse-dynamics-based musculoskeletal analysis using inertial data previously collected [96]. For this purpose, this tool includes a musculoskeletal model consisting of body segments and a set of markers and muscles combined together. Before calculating the external forces and joint torques by inverse dynamics, this model must be calibrated according to the subject used during the trial.

Motor torque defined as the rotational equivalent of a linear force, represents the capability of a force to produce change in the rotational motion of the body [97]. The determination of the motor torque involved in each joint of a wearable device is one of the most important specifications during the development phase. The inability to include force platforms in the protocols previously addressed made it impossible to directly determine the torque values involved during the trials. For this reason, the research work developed in this dissertation also involved finding a method capable of estimating the GRF from Xsens inertial data to allow the torque's determination. Figure 6.2 presents the general pipeline of the CusToM Matlab toolbox.

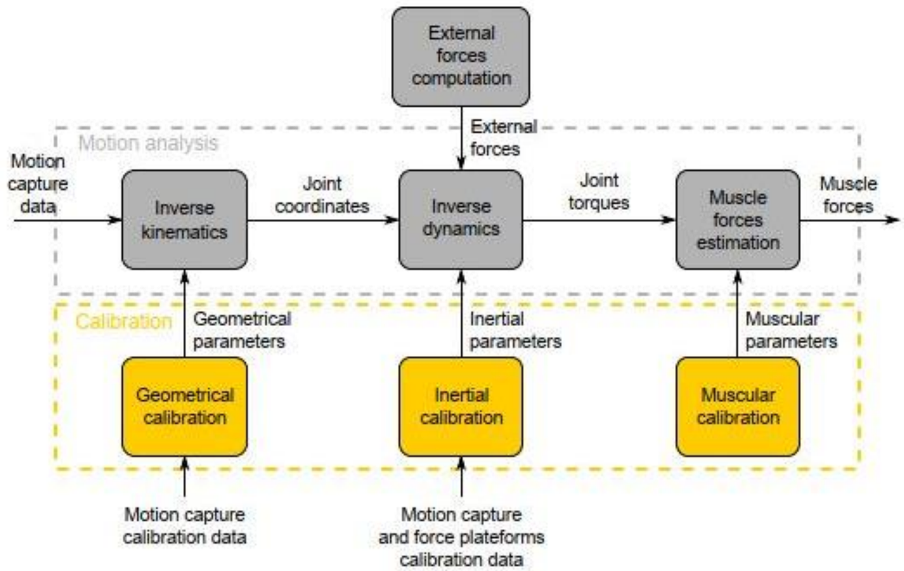


Figure 6.2 CusToM toolbox pipeline. Taken from [87].

6.2.1.2 CusToM toolbox validation

The validation of the CusToM toolbox with scientific literature was also one of the steps included in the torques' estimation. Concerning inaccuracy associated with the CusToM toolbox, a literature analysis of the articles that use this tool was conducted to compare the error obtained between the estimated forces and moments with directly measured values for different daily tasks. Additionally, in order to validate the signals obtained, both from the GRF and the torque values, they were compared with signals found in the literature. After obtaining torques values, outliers were removed through Matlab ® function *rmoutliers* which removes values above or behind three standard deviations from the median [98].

6.2.2 RPM Estimation, RoM and detection and actuation times definition

After the torque estimation using the CusToM toolbox, further requirements for the future development of wearable device, for slip related falls' prevention were determined. Firstly, a literature search was performed to define some general considerations regarding the project of medical devices and, more specifically, wearable devices were addressed. The definition of the RoM required for each of the lower limbs' joints was also studied in this chapter using the angles values collected by Xsens. Maximum and minimum values were primarily extracted, for each of the three gait labels, namely, steady walking, perturbation, and biomechanical response considering both perturbed and unperturbed limb. Additionally, the RoM value, which was obtained by subtracting the maximum value by the minimum value, will be presented. Similarly to the procedure carried out with torque values, RoM outliers were removed. Regarding detection and actuation times, the scientific literature related to the development of wearable devices for slip-related falls prevention as well as the literature included in the state-of-the-art review addressed in Chapter 3 allowed the definition of the timings involved from the start of the slip until the steady walking restoration. Detection and actuation times, together with RoM values allowed to define the number of rpm's necessary as an output of the orthosis motors, depending on the joint where the actuation is desired, accordingly with the following equation.

$$rpm = \frac{RoM_{max} \, . \, 60}{Actuation\ time} \tag{6.1}$$

6.3 Results

6.3.1 CusToM toolbox

In the present subsection the results obtained by the methods referred above will be introduced, initially the torques obtained using the toolbox as well as its validation will be presented. Afterwards, some general thoughts on the creation of wearable fall prevention devices, as well as the corresponding quantitative requirements, will be presented. Finally, the results obtained will be compared to the scientific literature outcomes from Chapter 3.

6.3.1.1 Torque estimation

Minimum and maximum values of the torques for each lower limb joint, estimated with CusToM toolbox, were extracted from the dataset. Table 6.1 summarizes the minimum and maximum values for each joint in the three gait labels: steady walking, rope pull and biomechanical response for slipping and trailing leg.

Table 6.1 Joints torques estimated using CusToM toolbox

Joint	Gait Label	Sagittal Plane Torques			
		Slipping Leg		Trailing Leg	
		Flexion (Nm/kg)	Extension (Nm/kg)	Flexion (Nm/kg)	Extension (Nm/kg)
Right Hip	Steady Walking	2.10	-2.74	2.20	-2.75
	Rope Pull	0.84	-2.20	3.22	-1.54
	Biomechanical Response	1.47	-2.19	2.68	-2.88
Left Hip	Steady Walking	2.16	-2.65	1.91	-2.47
	Rope Pull	1.05	-3.05	2.43	-1.74
	Biomechanical Response	1.39	-2.74	2.22	-2.07
Right Knee	Steady Walking	1.28	-1.29	1.82	-1.70
	Rope Pull	0.44	-1.35	1.89	-1.50
	Biomechanical Response	1.51	-1.32	2.34	-1.29
Left Knee	Steady Walking	1.32	-1.65	1.04	-1.41
	Rope Pull	0.83	-1.50	1.66	-1.39
	Biomechanical Response	1.44	-1.37	2.14	-1.36
Right Ankle	Steady Walking	0.25	-2.00	0.36	-2.21
	Rope Pull	0.20	-0.86	0.20	-1.80
	Biomechanical Response	0.58	-1.92	0.33	-1.87
Left Ankle	Steady Walking	0.40	-1.98	0.36	-2.15
	Rope Pull	0.34	-1.22	0.90	-1.85
	Biomechanical Response	0.55	-2.01	0.45	-2.10

Observing the results in Table 6.1, it is feasible to conclude that maximum and minimum torques relate to rope pull and biomechanical response labels in certain circumstances. As a consequence, it corroborates the requirement to build a slip-induced procedure and consider its outcomes in order to achieve a purpose-oriented project that takes into account the specificities of slip-like disturbance scenarios.

6.3.1.2 CusToM toolbox validation

Concerning the CusToM toolbox validation, Table 6.2 shows the error associated with the toolbox, in all directions, determined in each article, as well as the task that subjects were performing during the inertial data collection. As aforementioned, in order to validate the estimated GRF and torque signals, these signals were compared with signals found in the literature presented in the Toolbox repository at GitHub [80].

Table 6.2 CusToM related error presented in scientific literature

Article	Predicted Variables and tasks performed	Direction	Error (%)
Motion-based prediction of external forces and moments and back loading during manual material handling tasks [99]	GRF during handling tasks	Vertical GRF	7.1 ± 2.2
		Anteroposterior GRF	4.7 ± 1.3
		Medio Lateral GRF	5.3 ± 1.9
		Sagittal Moments	2.0 ± 0.8
		Frontal Moments	3.4 ± 1.7
		Transverse Moments	7.2 ± 2.4
Ground Reaction Forces and Moments Prediction of Challenging Motions: Fencing Lunges [100]	Forces and Moments during lunges movement	Vertical GRF	Left Foot:7.0 ± 6.0 Right Foot: 6.8 ± 4.8
		Anteroposterior GRF	Left Foot:17.2 ± 3.4 Right Foot: 15.5 ± 2.6
		Medio Lateral GRF	Left Foot:21.8 ± 10.8 Right Foot: 20.0 ± 7.7
		Sagittal Moments	Left Foot:5.7 ± 2.8 Right Foot: 7.4 ± 5.9
		Frontal Moments	Left Foot:7.3 ± 2.7 Right Foot: 6.4 ± 2.1
		Transverse Moments	Left Foot:32.2 ± 13.5 Right Foot: 26.3 ± 20.7
Motion-based prediction of hands and feet contact efforts during asymmetric handling Tasks [101]	Forces and Moments during handling tasks	Vertical GRF	Right Foot: 2.7 ± 2.7 Left Foot: 3.1 ± 3.0
		Anteroposterior GRF	Right Foot: 15.8 ± 13.7 Left Foot: 17.8 ± 14.6
		Medio Lateral GRF	Right Foot: 34.8 ± 27.6 Left Foot: 40.7 ± 30.8
		Sagittal Moments	Right Foot: 14.1 ± 10.3 Left Foot: 12.3 ± 8.7
		Frontal Moments	Right Foot: 17.3 ± 18.2 Left Foot: 16.4 ± 12.3
		Transverse Moments	Right Foot: 40.6 ± 36.4 Left Foot: 73.8 ± 46.7

Analysing the percentage of error presented in the literature in Table 6.2, it is perceptible that the forces and moments in the medio lateral plane are those which present a greater inaccuracy in comparison with real measured values. In contrast, the GRF and moments in the sagittal plane, where most critical gait motions occur, as discussed in [13], have a smaller percentage of inaccuracy. The motions in this plane were investigated further in this dissertation. Following a study of inaccuracy errors, the waveform and magnitude of GRF and torques were compared to scientific literature. Figure 6.3 compares the GRF, and torque signals acquired with the same signals discovered in this investigation.

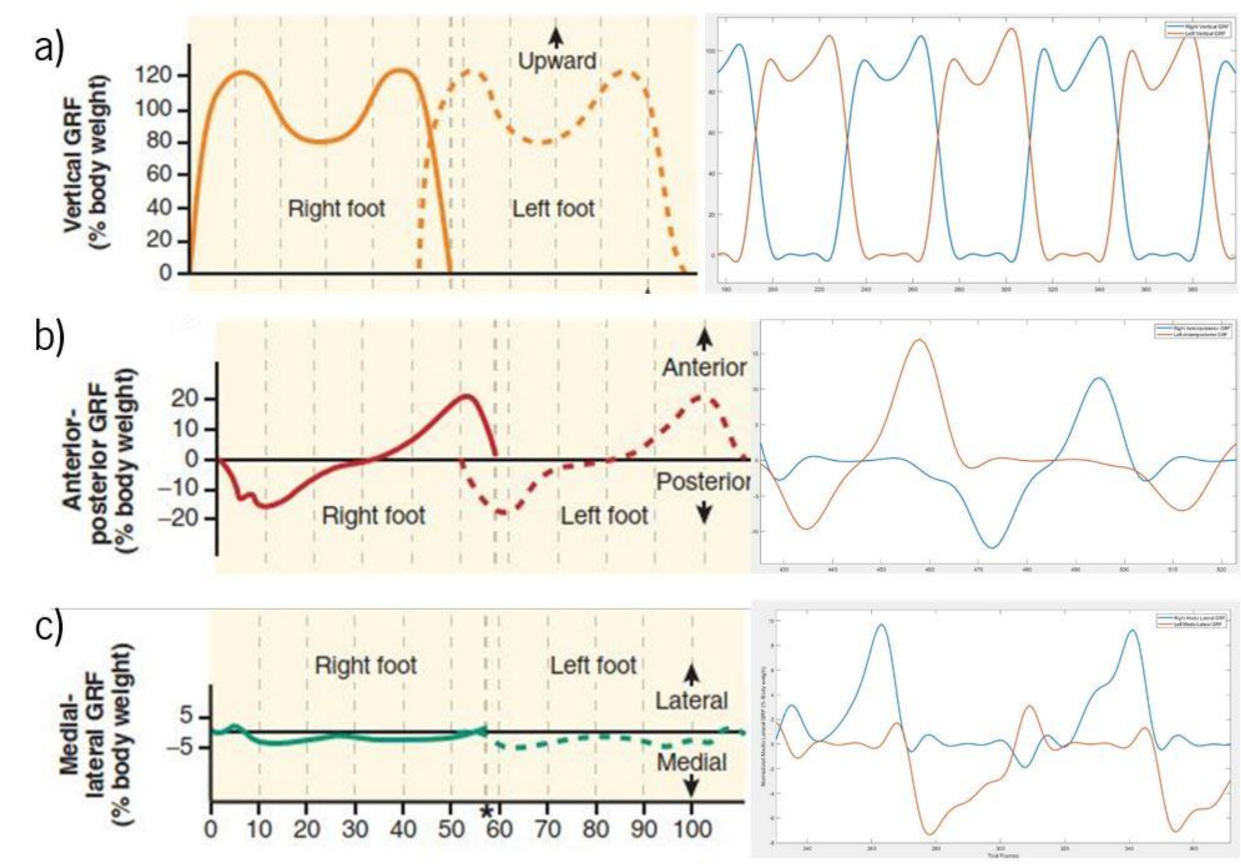


Figure 6.3 a) Vertical; b) Anteroposterior and c) Medio Lateral GRF comparison between literature (left side) and GRF obtained by CusToM toolbox (right side). Left side figures taken from [102].

Comparing the graphs obtained with those found in the scientific literature, considering magnitude and waveform, it is possible to confirm that the toolbox presents very similar results, namely in the case of vertical and AP GRF. In turn, the mediolateral GRF presents some dissimilarities both in waveform and peak magnitude, thus corroborating the values presented in table 6.2 which show a higher percentage of error for mediolateral GRF estimation. The same comparison was made for the torque values obtained for the lower limb joints, using scientific literature where perturbations were induced as shown in Figure 6.4.

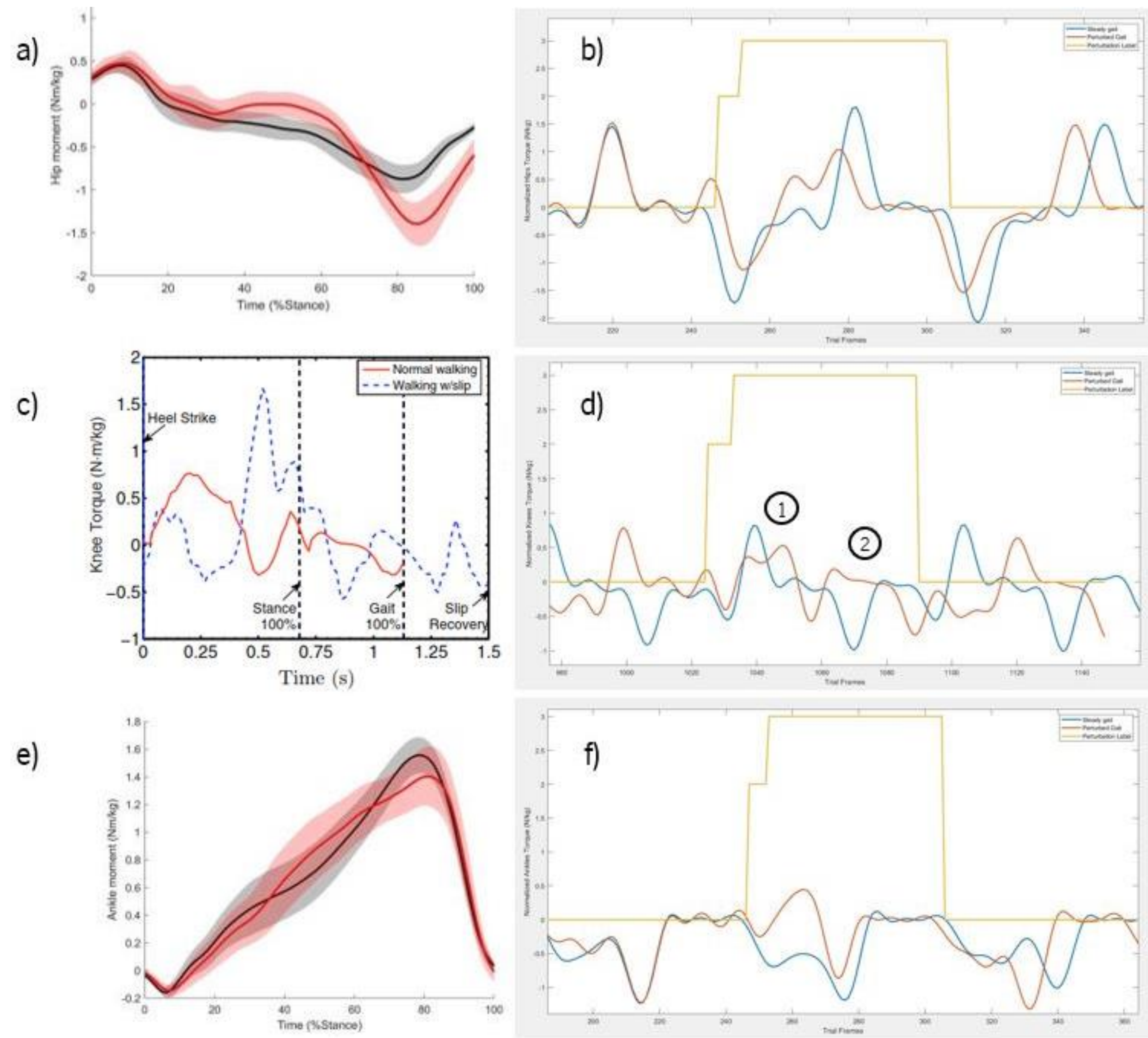


Figure 6.4 Hip (a) and (b); knee (c) and (d) and ankle (e) and (f) joints torques comparison between literature (left side) and torques obtained by CusToM toolbox (right side) for both steady walking and perturbation situations. Left side pictures taken from [29] and [44]. 1 and 2 point out two typical knee responses after slip perturbations discussed below.

In this case, the probability of dissimilarities between estimated and literature signals is higher since the perturbations induced in the literature were provoked quite differently from those provoked by the protocols performed. Still, considering the shape and magnitude of the wave, the torques involved at the hip, knee and ankle are consistent with those obtained in the scientific literature.

Concerning hip joint, when comparing the estimated torques with scientific literature it is possible to observe, that, in both, the signal corresponding to a slip situation appears increased, comparing with steady walking situations. In the graph presented on the left side, red line presents higher values compared with the grey one. On the right the same happens with the orange and blue line, respectively, after perturbation onset. Concerning the knee joint, comparing steady walking with perturbation situations, when a slip-like perturbation occurs, initially the torque values are lower comparing with steady walking situations, as denoted in the time period marked with 1. Then, the minimum valley corresponding to the steady walking signal is substituted by a local maximum value in slip situations as marked with 2. Both findings, agree with the signals indicated in literature. Finally, in the ankle joint, the values obtained presented some dissimilarities compared with literature: as shown in Figure 6.4 estimated torques' magnitude is lower compared with scientific literature.

6.3.2 Orthosis specifications

Considering the results obtained in the previous section, the project target specifications were defined using the CusToM toolbox to proceed to torque estimation. Firstly, common requirements to the development of any medical device and general considerations will be introduced. RoM for each lower limb's joint and the motor specifications depending on the joint where it is desired to produce actuation will be further presented. Finally, some considerations in terms of detection and actuation times will also be addressed based on the information collected in the reviewed literature.

6.3.3 General Considerations

Firstly, the device to be developed will have to present some more global requirements capable of promoting a long, comfortable, ergonomic, safety, effective and reliable utilization [103]. Safety and comfort are the most important requirements for wearable devices. Going deeper, the device to be developed, being a when needed assistance device it must ensure that its passive mode of operation does not interfere with the normal gait of the user, thus avoiding an unbalanced gait. To this end, the definition of maximum size and weight must be considered in the selection of the materials to be used [103]. Regarding the size, the device should be considered anthropomorphic, i.e., the constitution elements are sized accordingly to the proportions among the corresponding segments in the lower limb assisted [104]. The thickness of the device should also be minimized as possible [105]. In turn, concerning to the weight the device should be as lightweight as possible, present a low inertial and strong materials to guarantee durability. Weight distribution is also essential as the loads should be balanced and closer to the body's CoM [105]–[107] and the design of this device should ensure an adequate interaction between user and device. The scientific literature shows the appearance of pressure ulcers after long-term use in some currently developed wearable devices [108]–[113]. Thus, the selection of the orthopaedic cuffs' materials where the physical interaction between user and orthosis will take place must ensure the absence of injuries in the latter. Also, these materials should ensure power transfer in the sagittal plane allowing passive movements in frontal and transverse planes [114]. Additionally, the prevention of misalignment between actuators and joints should be considered in order to maximize device performance and ensure injury prevention [115]–[117]. To this end, the device needs to have features to allow the physical adjustment of the orthosis [107]. As this is a device whose target audience is the elderly population, it should also present an easy and quick donning and doffing processes, without requiring great physical effort from the user [106], [118], [119]. Additionally, the orthosis mechanical design should be adaptable to different users of different body shapes and characteristics. Portability is also a requirement to be considered so that the device can be easily carried by therapists and users [106], [115], [117], [119]. Finally, the literature reports that the use of elastic series actuators presents positive results when used in the design of these devices, as they allow the adaptation of their stiffness to a value close to the human joints stiffness increasing the feeling of comfort reported by the user [104][114].

6.3.4 Joints ROM

Following all these broad considerations, some quantitative specifications derived from experimental data analysis will be provided. First, the lowest and maximum angles of each lower limb joint, as well as the RoM, will be provided. Table 6.3 will list all of these values for each of the three gait labels: steady walking, rope pull, and biomechanical response.

Table 6.3 Lower limb's joints RoM, during steady walking, rope pull, and biomechanical response obtained by experimental data analysis

		Sagittal Plane					
Joint	**Gait Label**	**Slipping Leg**			**Trailing Leg**		
		Max (°)	Min (°)	RoM (°)	Max (°)	Min (°)	RoM (°)
Right Hip	Steady Walking	52.34	-22.02	74.37	55.76	-20.06	75.82
	Rope Pull	45.45	9.38	36.07	19.80	-19.67	39.47
	Biomechanical Response	54.33	-14.50	68.83	65.54	-18.79	84.32
Left Hip	Steady Walking	58.18	-20.59	78.77	58.84	-22.01	80.84
	Rope Pull	45.88	11.85	34.03	8.13	-17.14	25.27
	Biomechanical Response	67.88	-17.31	85.19	64.98	-16.75	81.74
Right Knee	Steady Walking	82.14	-8.41	90.55	85.31	-21.03	106.34
	Rope Pull	44.59	-8.41	53.00	82.09	0.12	81.97
	Biomechanical Response	82.14	-4.60	86.74	91.82	-4.25	96.07
Left Knee	Steady Walking	89.26	-8.89	98.15	83.59	-7.96	91.55
	Rope Pull	46.34	-5.25	51.59	73.16	-0.28	73.44
	Biomechanical Response	89.59	-6.92	96.51	85.65	-1.96	87.62
Right Ankle	Steady	32.37	-40.64	73.01	41.01	-54.23	95.23

	Walking						
	Rope Pull	20.48	-20.50	40.98	37.58	-38.66	76.25
	Biomechanical Response	32.37	-34.66	67.03	41.01	-54.23	95.23
Left Ankle	Steady Walking	39.60	-38.09	77.70	35.56	-49.27	84.84
	Rope Pull	20.86	-20.31	41.17	34.34	-47.36	81.70
	Biomechanical Response	39.60	-33.41	73.02	35.56	-51.73	87.29
Joint	**Gait Label**	**Frontal Plane**					
		Slipping Leg			**Trailing Leg**		
		Max (°)	Min (°)	RoM (°)	Max (°)	Min (°)	RoM (°)
Right Hip	Steady Walking	14.21	-13.00	27.22	15.99	-15.30	31.29
	Rope Pull	14.86	-7.90	22.77	11.77	-13.03	24.80
	Biomechanical Response	16.61	-11.05	27.65	12.74	-17.86	30.60
Left Hip	Steady Walking	20.13	-15.52	35.65	16.77	-13.71	30.48
	Rope Pull	18.24	-9.03	27.27	13.20	-8.25	21.45
	Biomechanical Response	21.97	-12.70	34.67	16.77	-15.54	32.31

Similarly to the results obtained in the torques' analysis in section 6.3.1 extreme RoM values are, generally, associated with rope pull and biomechanical response situations confirming, once again, the destabilization effect of slip-like perturbations and the need to develop a biomechanical response whose movement is more challenging compared to steady walking.

6.3.5 Detection and Response Times

The timings from the slip onset to the end of the biomechanical response are addressed in the literature review in Chapter 3 as well as in the literature related to the development of fall prevention robotic devices. The timings presented in the literature result in successful recoveries, so wearable devices for slip-related fall prevention should be able to at least detect and promote an actuation strategy in these time intervals.

Some authors highlight that the slip does not occurs immediately after the HS. While Lockhart et al [120] refer that most dangerous slips occur between 70 and 120 ms after the HS, Chou et al [10] assume that the perturbations happen between 60 and 80 ms after the HS. After this event, the onset of the human biomechanical response naturally depends on many factors including muscle fatigue or age, as discussed in Chapter 3. Despite some variability in response depending on these and other factors, most authors consider the muscle latency period to be between 160 and 200 ms after the slip onset. After this period, the biomechanical response begins [28][33][34][12][10][18]. Concerning the compensatory step time, considered since the start of the response to the resumption of steady walking, its period also differs. Parijata et al [63] refer that the compensatory step time lasts around 150 ms while Martelli et al [57] refer that the time interval between the slip events and the end of the biomechanical response is 410 ms and 480 ms respectively for elderly and young people.

Regarding the literature concerning the development of robotic devices for slip-related falls prevention, detection and actuation times also differ between the analysed literature. Monaco et al [22] developed a strategy which includes a detection time of 300 ms and an actuation time of 350 ms. In turn, Mioskowska [31] proposed a detection time of 100 ms while Trkov [32] defined a detection period between 30 and 90 ms after the slip. In this case the device actuation ends 200 ms later. In order to calculate the rpm values to mimic the biomechanical response presented in the experimental data, both minimum and maximum actuation times previously mentioned will be considered to obtain a rpm range of values. Thus, the minimum value for the device actuation will be the value of 100 ms referred by Mioskowska [31] while the maximum value will be the value of 350 ms defined by Monaco et al [22].

6.3.6 RPM

Next, the range of rpm values, obtained considering the previously referred actuation times and the equation in 6.1, are shown in Table 6.4. Knees and ankle joints presented higher rpm values compared with both hips joint, as this joints. In turn, Figure 6.5 shows a comparison between the torque, RoM and rpm that resulted from the experimental data, with the same metrics included in scientific literature.

Table 6.4 Range of rpm values obtained considered the previously presented actuation times

Joint	*Minimum rpm value*		*Maximum rpm value*	
	Slipping Leg	**Trailing Leg**	**Slipping Leg**	**Trailing Leg**
Right Hip	17.85	20.24	44.62	50.59
Left Hip	20.45	19.62	51.10	49.04
Right Knee	21.73	25.52	54.33	63.80
Left Knee	23.56	21.97	58.89	54.93
Right Ankle	17.52	22.86	43.81	57.14
Left Ankle	18.65	20.95	46.62	52.37

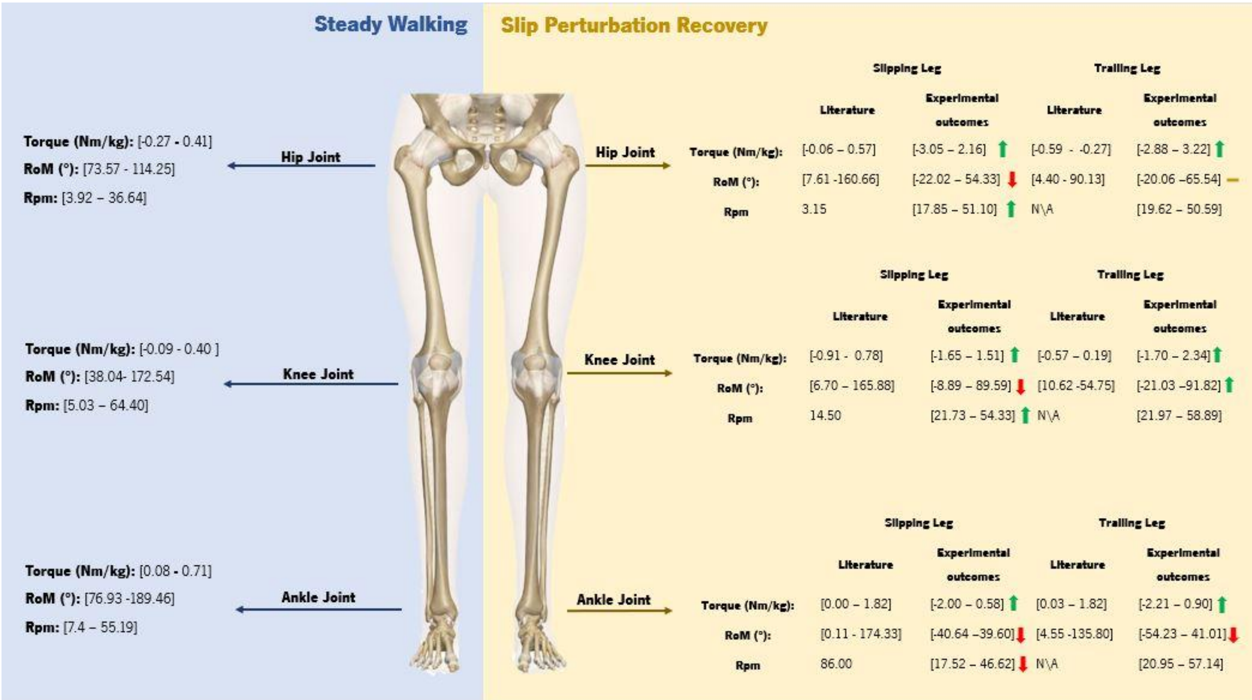


Figure 6.5 Torque, RoM and rpm comparison between literature results and experimental data outcomes.

6.4 Discussion

The research work conducted in this chapter allowed to define the general user needs and to gather a first set of quantitative target specifications for the future development of a wearable robotic device to prevent slip-related falls. Firstly, the method performed to overcome the impossibility to obtain GRF measured directly during the trials was introduced as well as the error associated to the toolbox used. Then, a scientific literature search allowed to conclude about the general user-centred requirements to consider during the development of wearable devices. Actuation and detection times range were also addressed through the analysis of the literature included in Chapter 2, related to fall prevention devices and in Chapter 3, related to biomechanical response to slip perturbations review. Concerning RoM, these were directly extracted from the inertial dataset allowing to determine the motor output, in rpm, to mimic the biomechanical response to a slip-like perturbation.

Concerning the CusToM Matlab toolbox and its validation, literature reports that sagittal and AP GRF and torques, estimated using this tool present a lower value (average error less than 10% in most cases) comparing with the same variables in medio lateral plane. This finding was then confirmed after the GRF and torque estimation and subsequent visual comparison with literature signals. Vertical and AP GRF's graphs were similar to the graphs provided by scientific literature while medio lateral GRF presented considerable differences in terms of waveform and magnitude. Regarding the joint torques, hip and knee torques in both, steady walking, and slip-like perturbation situations, were also similar compared with the graphs shown in literature. Conversely, ankle torques presented some dissimilarities special in terms of signal magnitude. This finding can be explained by the fact, due the induced perturbations, the IMU placed on the perturbed foot suffered undesirable displacements. Also, is expected that the CusToM toolbox error increases in more demanding movements situations as is the case of slip-like perturbations and subsequent biomechanical response. In [100], where subjects were instructed to perform lunge movements, the toolbox error increased compared with [95] and [97] where CusToM toolbox was used to estimate GRF during simpler tasks.

Regarding the RoM and estimated torque values, in most of the considered joints, rope pull, and biomechanical response labels were associated to higher values comparing with steady walking situations. These observations, in addition to confirming the destabilising effect of slip-like perturbations and the challenging subsequent biomechanical response, highlight the importance of considering quantitative experimental data as a preliminary step in the design of devices that mimic human responses in order to properly design and customise these devices to meet their specific needs. As a result, over-dimensioning and the use of conventional quantitative data that is disconnected from the device reality are avoided.

Comparing obtained RoM values with scientific literature, hip and knee values are in agreement with scientific literature, addressed in Chapter 2, which considers flexion/extension movement RoM of approximately 140° and 150°, respectively, for hip and knee joint [41], [121]. In turn, ankle angles appear higher than the reported RoM values in scientific literature for the three gait labels. However these values are smaller to the ones related in [23]. This values naturally influenced the rpm values indicated in Table 6.4. Greater values than the ones reported in scientific literature can be explained by the justification previously enunciated: the perturbation provoked some displacements in the IMU's used. In turn, when comparing the obtained torques with literature where wearable robotic devices for fall prevention are developed, the values estimated using CusToM toolbox were greater comparing to Monaco [22], Mioskowska [31] and Trkov [32] studies. Monaco et al [22] defined a 0.2 Nm/ kg torque to hip actuation while the research work developed in this chapter determined a maximum absolute torque value of 3.0 Nm/kg in the same joint. In turn, concerning the knee joint, Mioskowska [31] and Trkov [32] referred respectively 20 Nm and 40 Nm. Although these values are not normalised by subjects' weight, these might be lower than the 1.9 Nm/kg obtained in the present dissertation. These values discrepancy can be explained by the error associated with the toolbox, which is shown in Table 6.2. Additionally, compared to the protocols of the present dissertation, all the protocols performed in the studied literature were distinct in the way the perturbation was induced.

Finally, it is important to highlight that the target specifications previously determined should be reconsidered in further phases of the project. The values presented should be guaranteed to allow an effective biomechanical response to slip-like perturbation so the motor efficiency should be considered to reach the desirable torque value. Same happens with the maximum user weight that the orthosis can support as a well as the orthosis weight which should be also further considered during the motor selection process.

7 Conclusions

Falls, defined as an incident that causes a subject to unintentionally rest on the ground, floor or another level are the main cause of injury deaths worldwide. Additionally, even if a fall does not result in an injury, individuals develop a fear of falling, which is a major problem that has a detrimental effect on both their mental and physical health. A person's daily activities are diminished as a result of this apprehension, which increases the risk of another fall due muscular deterioration. Among the several reasons that could result in falls, slipping frequently appears to be the main contributor. Slip-related falls frequently result in worse consequences being held accountable for significant proportion of hip fractures in elderly people emphasizing the urgency to address this issue [7].

To intervene in this problem, literature demonstrates the relevance and effectiveness of repetitive training in preparing subjects for real-world perturbations, resulting in anticipatory and adaptive responses in potential slip situations. The development of wearable robotic devices for slip-like falls prevention also emerges as a solution for this problematic. The primary purpose of these devices is to detect a LoB situation and produce an actuation capable to restore a suitable biomechanical position to allow the subject to resume a steady walking gait following a slip-like perturbation. The study of the biomechanical response to slip perturbations appears to be a common preliminary step of both previously addressed approaches. In the case of repetitive training, the study of this response allows quantifying and analysing the evolution of the strategies developed by individuals as a result of this training. This basic phase in the development of wearable robotic systems for fall prevention allows for the selection of the device target. Literature on this topic highlights the common use of motion capture systems to analyse the kinematic variables involved in this response. Kinetic, spatiotemporal and EMG variables are also common is literature related to the study of the biomechanical response after slip perturbations.

Considering the relevance of the wearable robotic devices's approach to act on slip related falls problematic, firstly a state-of- the art study related to wearable robotic devices to slip related falls prevention was conducted. Three devices were identified, confirming a variety of designs, structures and the joint where the actuation is produced. Monaco et al [22] developed a slip recovery strategy in an APO. In this device both hips are assisted by a DC electric motor whose movement is transmitted to the subject through an SEA. In turn, Mioskowska et al [31] developed a compressed gas actuated knee assistive exoskeleton for fall prevention, while Trkov et al [32] also presented a knee actuation device assisted by an electric motor. The main limitation found in these works is that it is not clear why the authors have chosen a particular joint to produce actuation over the other possibilities.

A systematic state-of-the-art review related to the biomechanical response to slip perturbations was also conducted in order to identify, the strategies naturally developed by individuals to avoid a fall in slip situations. Kinematic, kinetic, spatiotemporal and EMG data were included in the analysis in order to obtain a comprehensive overview of the whole biomechanical process. To the best of the author's knowledge, no study has previously addressed this issue considering multivariate data. The main limitations found in the scientific literature related with the study of the biomechanical response to slip perturbations is that, in most situations, authors do not provide a comprehensive study including kinematic, kinetic, spatiotemporal and EMG parameter for different gait and perturbation conditions.

Considering the previous referred limitations found in the scientific literature, the present dissertation seeks to, firstly, present a comprehensive analysis of the biomechanical response to slip-induced perturbations previously collected at BirdLab*.* Besides including kinematic, kinetic, spatiotemporal and EMG this analysis also included a wide variety of IV's namely gait speed, perturbed foot, perturbation intensity and inclination in order to make this analysis more complete and able to study several variables that may condition the slip-perturbation in real-world situations. Through this analysis it was possible to conclude about the differential biomechanical response in each limb depending if this is on the ipsilateral or contralateral leg relative to the perturbation. When the perturbation is delivered, it results in flexor, extensor and dorsiflexor dominant moments in the perturbed leg, mostly because of the rope pull. After the perturbation, both perturbed and unperturbed limbs, presented different behaviours. Regarding the perturbed limb, the hip, knee, and ankle´s response were respectively characterized by extension, flexion, and plantarflexion moments. Conversely, in the unperturbed limb, biomechanical response was characterized by hip flexion, knee extension and ankle plantarflexion. In summary, this coordination between the perturbed and unperturbed limb allows to place both feet closer increasing the subject stability after the perturbation. Concerning the influence of the IV's included in the analysis, speed, surface inclination and perturbation intensity affected the joints behaviour in the three gait labels considered. The biomechanical analysis performed in Chapter 5 also had the objective to rank the joints accordingly to its importance in the biomechanical response. Considering the partial eta squared values analysis, this method allowed to conclude that hip is the joint that intervenes more actively in the biomechanical response to a slip-like perturbation. Conversely, ankle joints appeared to be less active during the biomechanical response. Finally, FSM determined that the variable with the greatest capacity to distinguish steady walking, rope pull, and biomechanical situations was the distance between both feet. Finally, Chapter 6 allowed the definition of some target specifications for the future design of a device capable to mimic the biomechanical response after a slip-like perturbation. In this Chapter some general considerations about the development of wearable devices were presented as well as some quantitative specifications namely torques, RoM, detection and actuation times, and also rpm.

The investigation work developed in the scope of this dissertation allowed to answer the RQs presented in Chapter 1.

- **RQ1:** What are the biomechanical responses to slip-induced perturbations highlighted in the scientific literature?

Concerning the literature search results, different biomechanical responses were distinguished for trailing and slipping leg, as shown in Chapter 3. Regarding the trailing leg, hip extension and knee flexion movement are responsible to an earlier interruption of the swing phase earlier, allowing this leg to be placed in the floor to increase the BoS. Conversely, the leading leg response can be divided in 2 two distinct moments: in the first period the hip and knee responses were mainly characterised by extension and flexion respectively, while in the second period hip flexion and knee extension were emphasised. Scientific literature also highlighted intra and interlimb coordination during biomechanical response to slip-perturbations. EMG data were also included in the analysis. TA, MG, RF, MH, BF, VL and VI were the muscles included in literature studied. Firstly, MH muscles were found to intervene actively during slip recoveries. Its activation in the perturbed limb explains hip extension and knee flexion, while its activation in the trailing limb allows the trailing foot touch down. In turn, the co-contraction of TA and MG was associated to successful recoveries, as these muscles increase the stability of this joint, allowing the reduction of the ankle angle (flat foot) and helping the contact with the ground. Regarding the VL, this muscle assists the knee joint in the secondary joint response allowing an adequate relative position between COM and BoS avoiding also knee buckling. Age was found as a determinant factor to the effectivity of the biomechanical response: older adults' response is less effective when compared with younger subjects being associated to a more destabilizing effect. Additionally, the compensatory step time in adults was found to be longer, thus indicating a slower response.

- **RQ2:** What are the biomechanical strategies to avoid falls after slip perturbations, obtained from experimental data analysis?

The biomechanical strategies to avoid falls determined by the experimental data analysis were presented in Chapter 5. These strategies were distinguished between the perturbed and unperturbed limb. In the perturbed limb, the hip's, knee's, and ankle´s response was respectively characterized by extension, flexion, and plantarflexion moments. Conversely, in the unperturbed limb, biomechanical response was characterized by hip flexion, knee extension and ankle plantarflexion. In summary, both limb's response allows to approximate both feet in the ground restoring and adequate relation between CoM and BoS increasing the subject stability after the perturbation. Regarding hip frontal movements the contralateral hip presented a strategy to approximate both feet by hip adduction. In turn, the ipsilateral hip during the biomechanical response presented an abduction movement that can be justified by a strategy developed to increase the BoS. Perturbed leg hip and knee´s moments also allowed the deceleration of this leg's movement allowing energy absorption.

Speed, inclination, and perturbation intensity also resulted in significant differences in the three considered labels. Firstly, speed resulted in more evident flexion movements for hips and knees. For higher speeds, the distance between both feet during the rope pull increased as at higher velocities subject presented shorter periods in double stance making easier to the operator to induce more evident perturbations by pulling the rope. Speed also influenced the ankle's biomechanical response since for lower velocities the contralateral foot presents a higher flat-foot period so this joint presented higher stability during the rope pulls. Regarding the inclination effect, during 10° inclination trials, knee joint appeared more flexed in the three gait labels situations. Concerning the perturbation intensity effect, more intense perturbations induced a greater distance between feet. Generally, the results obtained agree with the scientific literature studied, concerning the biomechanical response of both perturbed and unperturbed limb in different gait conditions. The only exception is related to the unperturbed hip movement: most of the literature highlights unperturbed hip extension as a movement to proceed a step back strategy, however in the analysis carried out, there was no need of the subjects to perform this strategy. In turn, subjects performed a narrow strategy. Concerning EMG data, the analysis conducted corroborated both, the kinematic findings, and the scientific literature on the topic regarding BF, GL, and RF muscles' excitatory responses. BF excitatory response allowed the hip extension and knee flexion during the biomechanical response while GL responses promoted the contact between foot and floor by controlling the ankle joint. In turn, RF inhibitory responses were responsible to allow knee joint flexion and produce a counter response to the knee extension provoked by the rope pull. Finally, in relation to the results obtained by both variable ranking methods eta partial squared, this method determined that the hip joint intervenes more actively in the biomechanical response to a slip-like perturbation comparing with knees and ankle joints. In contrast, ankle joint appears to be the less relevant joint during the biomechanical response. In relation to the FSM, these allowed to conclude that the variable with the greatest capacity to distinguish steady walking, rope pull, and biomechanical situations was the distance between both feet.

- **RQ3:** What are the target specifications in the project of slip-related falls prevention wearable robotic devices? The answer in included in Chapter 6.

This RQ is addressed in Chapter 6. Firstly, concerning the general requirements for wearable robotic devices, the device should be comfortable, ergonomic, safety, effective and reliable. In order to ensure these objectives the device to be future developed should be: i) as lightweight and present low volume as possible; ii) present comfortable materials where the physical interaction between subject and device is promoted; iii) prevent misalignments between actuators and lower limb joints; iv) allow the physical adjustment of the orthosis and finally v) promote an easy and quick donning and doffing processes, without requiring great physical effort from the user. Concerning quantitative specifications, RoM, torques and rpm were presented for the three lower limb´s joints. Detection and actuation times were also presented accordingly to the scientific literature studied. Firstly, detection time should be as low as possible being the lowest time interval presented in literature of 30 ms. In turn, were considered the extreme values of actuation times presented in literature in order to calculate a range of rpm's. Thus, minimum actuation time was defined as 100 ms and the maximum actuation time of 350 ms. This values allowed to obtain a rpm range of values of [17.85 – 51.10], [21.73 – 63.80] and [17.52 – 57.14] respectively for hip, knee, and ankle joints. 85.19°, 106.34° and 95.23 ° were defined as the maximum RoM target specifications respectively for hips, knees, and ankle joints while a set of flexion/extension torque values of [-3.05 to 3.22], [-1.70 to 2.34] and [-2.21 to 0.90] in Nm/kg were determined for hip, knee, and ankle joint, respectively.

7.1 Future work

Several future work can be developed within the scope of this dissertation. Firstly, regarding to the data collected in the first protocol developed at Birdlab, slip perturbations were also collected during toe-off. Further, perturbations in this gait phase should be also analysed to both enhance the knowledge about the biomechanical response after slip-perturbations and to equipping the orthosis, to be future developed, with information to be capable to detect different slip perturbations.

Then, the repetition of experimental protocol 1 with the use of force plates should be considered in order to allow a direct measured values of GRF and torques. If possible, this protocol should also include subjects of different dominance and age groups in order to understand the effect of these variables during the biomechanical response to slip perturbations. An automatic perturbation triggering mechanism would also allow a more detailed study of the biomechanical response, with complementary information about the perturbation intensity and time after the HS.

Finally, considering the preliminary specifications presented in Chapter 6, mechanical development of the slip-related falls prevention device should be undertaken. The selection of the components should also take into account the qualitative and quantitative specifications already defined. The joint where the actuation be produced will be one of the main challenges in this future choice.

8 References

[1] WHO, “Decade of Healthy Ageing,” *World Heal. Organ.*, pp. 1–24, 2020, [Online]. Available: https://www.who.int/initiatives/decade-of-healthy-ageing%0Ahttps://www.who.int/initiatives/decade-of-healthy-ageing/connection-series%0Ahttps://www.who.int/ageing/decade-of-healthy-ageing.

[2] “Falls.” https://www.who.int/news-room/fact-sheets/detail/falls (accessed Jul. 19, 2022).

[3] S. Turner, R. Kisser, and W. Rogmans, “Falls among older adults in the EU-28 : Key facts from the available statistics,” *Swansea Univ. EuroSafe*, pp. 1–5, 2015, [Online]. Available: https://eupha.org/repository/sections/ipsp/Factsheet_falls_in_older_adults_in_EU.pdf.

[4] C. S. Rau *et al.*, “Geriatric hospitalizations in fall-related injuries,” *Scand. J. Trauma. Resusc. Emerg. Med.*, vol. 22, no. 1, pp. 1–8, 2014, doi: 10.1186/s13049-014-0063-1.

[5] S. H. Park, “Tools for assessing fall risk in the elderly: a systematic review and meta-analysis,” *Aging Clin. Exp. Res.*, vol. 30, no. 1, p. 0, 2018, doi: 10.1007/s40520-017-0749-0.

[6] W. R. Chang, S. Leclercq, T. E. Lockhart, and R. Haslam, “State of science: occupational slips, trips and falls on the same level*,” *Ergonomics*, vol. 59, no. 7, pp. 861–883, 2016, doi: 10.1080/00140139.2016.1157214.

[7] “Falls.” https://www.who.int/news-room/fact-sheets/detail/falls (accessed Oct. 18, 2022).

[8] F. Aprigliano, D. Martelli, P. Tropea, G. Pasquini, S. Micera, and V. Monaco, “Aging does not affect the intralimb coordination elicited by slip-like perturbation of different intensities,” *J. Neurophysiol.*, vol. 118, no. 3, pp. 1739–1748, 2017, doi: 10.1152/jn.00844.2016.

[9] P. Parijat and T. E. Lockhart, “Effects of lower extremity muscle fatigue on the outcomes of slip-induced falls,” *Ergonomics*, vol. 51, no. 12, pp. 1873–1884, 2008, doi: 10.1080/00140130802567087.

[10] Y. L. Chou, J. Y. You, F. C. Su, C. J. Lin, and P. H. Chou, “The dynamics of balance control during slipping,” *Chinese J. Mech. Ser. A (English Ed.*, vol. 17, no. 2, pp. 69–78, 2001, doi: 10.1017/s1727719100003142.

[11] T. Bhatt, F. Yang, and Y.-C. Pai, “Learning to Resist Gait-Slip Falls: Long-Term Retention in Community-Dwelling Older Adults,” *Arch. Phys. Med. Rehabil.*, vol. 93, no. 4, pp. 557–564, 2012, doi: 10.1016/j.apmr.2011.10.027.

[12] K. Beschorner, M. Redfern, and R. Cham, “Earliest Gait Deviations During Slips: Implications For Recovery,” *IIE Trans. Occup. Ergon. Hum. Factors*, vol. 1, no. 1, pp. 31–37, Jan. 2013, [Online]. Available:

https://search.ebscohost.com/login.aspx?direct=true&AuthType=ip,shib&db=edo&AN=ejs29346560&site=eds-live&scope=site.

[13] F. Yang, T. Bhatt, and Y.-C. C. Pai, "Role of stability and limb support in recovery against a fall following a novel slip induced in different daily activities," *J. Biomech.*, vol. 42, no. 12, pp. 1903–1908, Aug. 2009, doi: 10.1016/j.jbiomech.2009.05.009.

[14] B. E. Moyer, M. S. Redfern, and R. Cham, "Biomechanics of trailing leg response to slipping - Evidence of interlimb and intralimb coordination," *Gait Posture*, vol. 29, no. 4, pp. 565–570, 2009, doi: 10.1016/j.gaitpost.2008.12.012.

[15] P. Parijat, T. E. Lockhart, and J. Liu, "Knee Joint Kinetics during Reactive Recovery: Effects of Localized Muscle Fatigue," *Proc. Hum. Factors Ergon. Soc. Annu. Meet.*, vol. 52, no. 15, pp. 1078–1082, Sep. 2008, [Online]. Available: https://search.ebscohost.com/login.aspx?direct=true&AuthType=ip,shib&db=edo&AN=ejs26745169&site=eds-live&scope=site.

[16] M. M. Nazifi, K. E. Beschorner, and P. Hur, "Association between slip severity and muscle synergies of slipping," *Front. Hum. Neurosci.*, vol. 11, Nov. 2017, doi: 10.3389/FNHUM.2017.00536.

[17] C. McCrum, M. H. G. Gerards, K. Karamanidis, W. Zijlstra, and K. Meijer, "A systematic review of gait perturbation paradigms for improving reactive stepping responses and falls risk among healthy older adults," *Eur. Rev. Aging Phys. Act.*, vol. 14, no. 1, pp. 1–11, 2017, doi: 10.1186/s11556-017-0173-7.

[18] F. Yang, F. C. Anderson, and Y. C. Pai, "Predicted threshold against backward balance loss following a slip in gait," *J. Biomech.*, vol. 41, no. 9, pp. 1823–1831, 2008, doi: 10.1016/j.jbiomech.2008.04.005.

[19] D. S. Marigold and A. E. Patla, "Strategies for dynamic stability during locomotion on a slippery surface: Effects of prior experience and knowledge," *J. Neurophysiol.*, vol. 88, no. 1, pp. 339–353, 2002, doi: 10.1152/jn.00691.2001.

[20] T. E. Lockhart, *Biomechanics of Human Gait - Slip and Fall Analysis*, 2nd ed. Elsevier Ltd., 2013.

[21] M. K. Karlsson *et al.*, "Prevention of falls in the elderly: A review," *Scand. J. Public Health*, vol. 41, no. 5, pp. 442–454, Jul. 2013, doi: 10.1177/1403494813483215.

[22] V. Monaco *et al.*, "An ecologically-controlled exoskeleton can improve balance recovery after slippage," *Sci. Rep.*, vol. 7, no. December 2016, pp. 1–10, 2017, doi: 10.1038/srep46721.

[23] S. Kim, K. S. Joo, J. Liu, and J. H. Sohn, "Lower extremity kinematics during forward heel-slip," *Technol. Heal. Care*, vol. 27, no. S1, pp. S345–S356, 2019, doi: 10.3233/THC-199032.

[24] F. Aprigliano, D. Martelli, S. Micera, and V. Monaco, "Intersegmental coordination elicited by

unexpected multidirectional slippinglike perturbations resembles that adopted during steady locomotion," *J. Neurophysiol.*, vol. 115, no. 2, pp. 728–740, 2016, doi: 10.1152/jn.00327.2015.

[25] F. Aprigliano, D. Martelli, P. Tropea, S. Micera, and V. Monaco, "Effects of slipping-like perturbation intensity on the dynamical stability," in *2015 37TH ANNUAL INTERNATIONAL CONFERENCE OF THE IEEE ENGINEERING IN MEDICINE AND BIOLOGY SOCIETY (EMBC)*, Nov. 2015, pp. 5295–5298, doi: 10.1109/EMBC.2015.7319586.

[26] K. Hirata *et al.*, "Relationship Between the Walking Velocity Relative to the Slip Velocity and the Corrective Response," *J. Med. Biol. Eng.*, vol. 41, no. 1, pp. 25–33, 2021, doi: 10.1007/s40846-020-00527-6.

[27] X. Qu, X. Hu, and F. L. Lew, "Differences in lower extremity muscular responses between successful and failed balance recovery after slips," *Int. J. Ind. Ergon.*, vol. 42, no. 5, pp. 499–504, 2012, doi: 10.1016/j.ergon.2012.08.003.

[28] A. J. Chambers and R. Cham, "Slip-related muscle activation patterns in the stance leg during walking," *GAIT \& POSTURE*, vol. 25, no. 4, pp. 565–572, 2007, doi: 10.1016/j.gaitpost.2006.06.007.

[29] D. Yoo, K.-H. H. Seo, and B.-C. C. Lee, "The effect of the most common gait perturbations on the compensatory limb's ankle, knee, and hip moments during the first stepping response," *GAIT \& POSTURE*, vol. 71, pp. 98–104, Jun. 2019, doi: 10.1016/j.gaitpost.2019.04.013.

[30] R. Klemetti, P. Moilanen, J. Avela, and J. Timonen, "Effects of gait speed on stability of walking revealed by simulated response to tripping perturbation," *Gait Posture*, vol. 39, no. 1, pp. 534–539, 2014, doi: 10.1016/j.gaitpost.2013.09.006.

[31] M. Mioskowska, D. Stevenson, M. Onu, and M. Trkov, "Compressed gas actuated knee assistive exoskeleton for slip-induced fall prevention during human walking," *IEEE/ASME Int. Conf. Adv. Intell. Mechatronics, AIM*, vol. 2020-July, pp. 735–740, 2020, doi: 10.1109/AIM43001.2020.9158992.

[32] M. Trkov, S. Wu, K. Chen, J. Yi, T. Liu, and Q. Zhao, "Design of a Robotic Knee Assistive Device (ROKAD) for Slip-Induced Fall Prevention during Walking," *IFAC-PapersOnLine*, vol. 50, no. 1, pp. 9802–9807, 2017, doi: 10.1016/j.ifacol.2017.08.887.

[33] A. Kharb, V. Saini, Y. Jain, and S. Dhiman, "A review of gait cycle and its parameters," *IJCEM Int J Comput Eng Manag*, vol. 13, no. July, pp. 78–83, 2011.

[34] W. Pirker and R. Katzenschlager, "Gait disorders in adults and the elderly: A clinical guide," *Wien. Klin. Wochenschr.*, vol. 129, no. 3–4, pp. 81–95, 2017, doi: 10.1007/s00508-016-1096-4.

[35] M. Whittle, Gait Analysis: an Introduction, 4° ed., British: Butterworth Heinemann Elsiver, 2007.

[36] "Capitulo 4 - Caracterização do Movimento Humano.pdf." .

[37] C. . Vaughan, B. . Davis, and J. . O'Connor, *D Ynamics of*. 1992.

[38] V. Abolins, K. Nesenbergs, and E. Bernans, "Reliability of Loading Rate in Gait Analysis," *IOP Conf. Ser. Mater. Sci. Eng.*, vol. 575, no. 1, 2019, doi: 10.1088/1757-899X/575/1/012002.

[39] R. Baker, "Temporal-spatial-data,-the-gait-cycle-and-gait-graphs-Introductory," *Univ. Salford An Introd. to Clin. Gait Anal.*, pp. 1–8.

[40] H.Gray, "Gray's Anatomy," 2016, 41° ed., Elsevier,2016

[41] K. Luttgens and N. P. Hamilton, *Kinesiology: Scientific Basis of Human Motion*. Brown \& Benchmark, 1997.

[42] G. A. Pratt and M. M. Williamson, "Series elastic actuators," *IEEE Int. Conf. Intell. Robot. Syst.*, vol. 1, pp. 399–406, 1995, doi: 10.1109/iros.1995.525827.

[43] F. Giovacchini *et al.*, "A light-weight active orthosis for hip movement assistance," *Rob. Auton. Syst.*, vol. 73, pp. 123–134, 2015, doi: 10.1016/j.robot.2014.08.015.

[44] V. C. V. Kumar, S. Ha, G. Sawicki, and C. K. Liu, "Learning a Control Policy for Fall Prevention on an Assistive Walking Device," *Proc. - IEEE Int. Conf. Robot. Autom.*, pp. 4833–4840, 2020, doi: 10.1109/ICRA40945.2020.9196798.

[45] M. Sakai, Y. Shiba, H. Sato, and N. Takahira, "Motor adaptation during slip-perturbed gait in older adults," *J. Phys. Ther. Sci.*, vol. 20, no. 2, pp. 109–115, 2008, doi: 10.1589/jpts.20.109.

[46] H. Debelle, C. Harkness-Armstrong, K. Hadwin, C. N. Maganaris, and T. D. O'Brien, "Recovery From a Forward Falling Slip: Measurement of Dynamic Stability and Strength Requirements Using a Split-Belt Instrumented Treadmill," *Front. Sport. Act. LIVING*, vol. 2, Jul. 2020, doi: 10.3389/fspor.2020.00082.

[47] F. Yang and Y. C. Pai, "Reactive control and its operation limits in responding to a novel slip in gait," *Ann. Biomed. Eng.*, vol. 38, no. 10, pp. 3246–3256, 2010, doi: 10.1007/s10439-010-0082-7.

[48] D. Tokur, M. Grimmer, and A. Seyfarth, "Review of balance recovery in response to external perturbations during daily activities," *Hum. Mov. Sci.*, vol. 69, no. October 2019, 2020, doi: 10.1016/j.humov.2019.102546.

[49] R. Grönqvist *et al.*, "Human-centred approaches in slipperiness measurement," *Ergonomics*, vol. 44, no. 13, pp. 1167–1199, 2001, doi: 10.1080/00140130110085556.

[50] C. Liu and J. M. Finley, "Asymmetric gait patterns alter the reactive control of intersegmental coordination patterns in the sagittal plane during walking," *PLoS One*, vol. 15, no. 5, pp. 1–17, 2020, doi: 10.1371/journal.pone.0224187.

[51] F. Yang and Y.-C. C. Pai, "Role of individual lower limb joints in reactive stability control following a

novel slip in gait," *J. Biomech.*, vol. 43, no. 3, pp. 397–404, Feb. 2010, doi: 10.1016/j.jbiomech.2009.10.003.

[52] T. E. Lockhart, J. L. Smith, and J. C. Woldstad, "Effects of aging on the biomechanics of slips and falls," *Hum. Factors*, vol. 47, no. 4, pp. 708–729, 2005, doi: 10.1518/001872005775571014.

[53] F. Yang, T. Bhatt, and Y. C. Pai, "Limits of recovery against slip-induced falls while walking," *J. Biomech.*, vol. 44, no. 15, pp. 2607–2613, Oct. 2011.

[54] R. Cham and M. S. Redfern, "Lower extremity corrective reactions to slip events," *J. Biomech.*, vol. 34, no. 11, pp. 1439–1445, Nov. 2001, doi: 10.1016/S0021-9290(01)00116-6.

[55] C. O'Connell, A. Chambers, A. Mahboobin, and R. Cham, "Effects of slip severity on muscle activation of the trailing leg during an unexpected slip," *J. Electromyogr. Kinesiol.*, vol. 28, pp. 61–66, 2016, doi: 10.1016/j.jelekin.2016.02.007.

[56] T. Bhatt, J. D. Wening, and Y. C. Pai, "Adaptive control of gait stability in reducing slip-related backward loss of balance," *Exp. BRAIN Res.*, vol. 170, no. 1, pp. 61–73, Mar. 2006, doi: 10.1007/s00221-005-0189-5.

[57] D. Martelli, F. Aprigliano, P. Tropea, G. Pasquini, S. Micera, and V. Monaco, "Stability against backward balance loss: Age-related modifications following slip-like perturbations of multiple amplitudes," *Gait Posture*, vol. 53, pp. 207–214, 2017, doi: 10.1016/j.gaitpost.2017.02.002.

[58] S. A. Wyszomierski, A. J. Chambers, and R. Cham, "Knee Strength Capabilities and Slip Severity," *J. Appl. Biomech.*, vol. 25, no. 2, pp. 140–148, May 2009, doi: 10.1123/jab.25.2.140.

[59] C. M. Rasmussen and N. H. Hunt, "Unconstrained slip mechanics and stepping reactions depend on slip onset timing," *J. Biomech.*, vol. 125, 2021, doi: 10.1016/j.jbiomech.2021.110572.

[60] H. Debelle, C. N. Maganaris, and T. D. O'Brien, "Biomechanical Mechanisms of Improved Balance Recovery to Repeated Backward Slips Simulated by Treadmill Belt Accelerations in Young and Older Adults," *Front. Sport. Act. LIVING*, vol. 3, 2021, doi: 10.3389/fspor.2021.708929.

[61] P. Tropea, D. Martelli, F. Aprigliano, S. Micera, and V. Monaco, "Effects of aging and perturbation intensities on temporal parameters during slipping-like perturbations," *Proc. Annu. Int. Conf. IEEE Eng. Med. Biol. Soc. EMBS*, vol. 2015-Novem, pp. 5291–5294, Nov. 2015, doi: 10.1109/EMBC.2015.7319585.

[62] A. H. Huntley, R. Rajachandrakumar, A. Schinkel-Ivy, and A. Mansfield, "Characterizing slip-like responses during gait using an entire support surface perturbation: Comparisons to previously established slip methods," *GAIT \& POSTURE*, vol. 69, pp. 130–135, Mar. 2019, doi: 10.1016/j.gaitpost.2019.01.033.

[63] P. Parijat and T. E. Lockhart, "Effects of moveable platform training in preventing slip-induced falls

in older adults," *Ann. Biomed. Eng.*, vol. 40, no. 5, pp. 1111–1121, 2012, doi: 10.1007/s10439-011-0477-0.

[64] X. Liu, T. Bhatt, S. Wang, F. Yang, and Y.-C. Pai, "Retention of the 'first-trial effect' in gait-slip among community-living older adults," *GeroScience*, vol. 39, no. 1, p. 93, Feb. 2017, doi: 10.1007/s11357-017-9963-0.

[65] B.-C. Lee, C.-S. Kim, and K.-H. Seo, "The Body's Compensatory Responses to Unpredictable Trip and Slip Perturbations Induced by a Programmable Split-Belt Treadmill," *IEEE Trans. NEURAL Syst. Rehabil. Eng.*, vol. 27, no. 7, pp. 1389–1396, Jul. 2019, doi: 10.1109/TNSRE.2019.2921710.

[66] A. Sawers and T. Bhatt, "Neuromuscular determinants of slip-induced falls and recoveries in older adults," *J. Neurophysiol.*, vol. 120, no. 4, pp. 1534–1546, 2018, doi: 10.1152/jn.00286.2018.

[67] F. L. Lew and X. Qu, "Effects of multi-joint muscular fatigue on biomechanics of slips," *J. Biomech.*, vol. 47, no. 1, pp. 59–64, Jan. 2014, doi: 10.1016/J.JBIOMECH.2013.10.010.

[68] T. Bhatt and Y. C. Pai, "Can observational training substitute motor training in preventing backward balance loss after an unexpected slip during walking?," *J. Neurophysiol.*, vol. 99, no. 2, pp. 843–852, 2008, doi: 10.1152/jn.00720.2007.

[69] D. Martelli, L. B. Luciani, S. Micera, and S. Member, "While Walking," *IEEE Trans. Biomed. Eng.*, vol. 60, no. 7, pp. 1785–1795, 2013.

[70] A. Lee, T. Bhatt, X. Liu, Y. Wang, S. Wang, and Y. C. C. Pai, "Can treadmill slip-perturbation training reduce longer-term fall risk upon overground slip exposure?," *J. Appl. Biomech.*, vol. 36, no. 5, pp. 298–306, 2020, doi: 10.1123/JAB.2019-0211.

[71] "Neurological disorders affect millions globally: WHO report." https://www.who.int/news/item/27-02-2007-neurological-disorders-affect-millions-globally-who-report (accessed Oct. 18, 2022).

[72] R. N. Ferreira, "Fall Prevention Strategy for an Active Orthotic System," 2022.

[73] W. J. Aston, *Product design and development*, vol. 7, no. 2. 1992.

[74] "Analysis of variance (ANOVA) - IBM Documentation." https://www.ibm.com/docs/en/cognos-analytics/11.1.0?topic=tests-analysis-variance-anova (accessed Oct. 13, 2022).

[75] P. J. Allen and K. Bennett, *SPSS for the Health and Behavioural Sciences*. Thomson Learning, 2007.

[76] P. Mishra, C. M. Pandey, U. Singh, A. Gupta, C. Sahu, and A. Keshri, "Descriptive statistics and normality tests for statistical data," *Ann. Card. Anaesth.*, vol. 22, no. 1, pp. 67–72, 2019, doi: 10.4103/aca.ACA_157_18.

[77] G. Roffo, S. Melzi, and M. Cristani, "Infinite feature selection," *Proc. IEEE Int. Conf. Comput. Vis.*,

vol. 2015 Inter, pp. 4202–4210, 2015, doi: 10.1109/ICCV.2015.478.

[78] G. Roffo, S. Melzi, U. Castellani, and A. Vinciarelli, "Infinite Latent Feature Selection: A Probabilistic Latent Graph-Based Ranking Approach," *Proc. IEEE Int. Conf. Comput. Vis.*, vol. 2017-Octob, pp. 1407–1415, 2017, doi: 10.1109/ICCV.2017.156.

[79] A. Muller, "musculo-squelettique de l ' humain dans l ' objectif d ' un compromis précision performance Antoine Muller Contributions méthodologiques à l ' analyse musculo- squelettique de l ' humain dans l ' objectif d ' un compromis précision performance," 2017.

[80] "GitHub - anmuller/CusToM." https://github.com/anmuller/CusToM (accessed Oct. 17, 2022).

[81] L. J. Elias, M. P. Bryden, and M. B. Bulman-Fleming, "Footedness is a better predictor than is handedness of emotional lateralization," *Neuropsychologia*, vol. 36, no. 1, pp. 37–43, 1998, doi: 10.1016/S0028-3932(97)00107-3.

[82] I. L. Kyrdalen, P. Thingstad, L. Sandvik, and H. Ormstad, "Associations between gait speed and well-known fall risk factors among community-dwelling older adults," *Physiother. Res. Int.*, vol. 24, no. 1, pp. 1–6, 2019, doi: 10.1002/pri.1743.

[83] L. Quach *et al.*, "The nonlinear relationship between gait speed and falls: The maintenance of balance, independent living, intellect, and zest in the elderly of boston study," *J. Am. Geriatr. Soc.*, vol. 59, no. 6, pp. 1069–1073, 2011, doi: 10.1111/j.1532-5415.2011.03408.x.

[84] D. Martelli, V. Monaco, L. Bassi Luciani, and S. Micera, "Angular momentum during unexpected multidirectional perturbations delivered while walking," *IEEE Trans. Biomed. Eng.*, vol. 60, no. 7, pp. 1785–1795, 2013, doi: 10.1109/TBME.2013.2241434.

[85] M. Vlutters, E. H. F. Van Asseldonk, and H. Van Der Kooij, "Center of mass velocity-based predictions in balance recovery following pelvis perturbations during human walking," *J. Exp. Biol.*, vol. 219, no. 10, pp. 1514–1523, 2016, doi: 10.1242/jeb.129338.

[86] D. S. Marigold, A. J. Bethune, and A. E. Patla, "Role of the unperturbed limb and arms in the reactive recovery response to an unexpected slip during locomotion," *J. Neurophysiol.*, vol. 89, no. 4, pp. 1727–1737, 2003, doi: 10.1152/jn.00683.2002.

[87] H. Finch, "Comparison of the performance of nonparametric and parametric MANOVA test statistics when assumptions are violated," *Methodology*, vol. 1, no. 1, pp. 27–38, 2005, doi: 10.1027/1614-1881.1.1.27.

[88] J. T. E. Richardson, "Eta squared and partial eta squared as measures of effect size in educational research," *Educ. Res. Rev.*, vol. 6, no. 2, pp. 135–147, 2011, doi: 10.1016/j.edurev.2010.12.001.

[89] R. Saini and S. K. Ghosh, "Ensemble classifiers in remote sensing: A review," *Proceeding - IEEE Int. Conf. Comput. Commun. Autom. ICCCA 2017*, vol. 2017-Janua, pp. 1148–1152, 2017, doi:

10.1109/CCAA.2017.8229969.

[90] "Another feature selection algorithm: MRMR | by Elliot | Medium." https://elliot-weissberg.medium.com/another-feature-selection-algorithm-mrmr-3827b6b19e33 (accessed Oct. 13, 2022).

[91] B. Remeseiro and V. Bolon-Canedo, "A review of feature selection methods in medical applications," *Comput. Biol. Med.*, vol. 112, p. 103375, 2019, doi: 10.1016/j.compbiomed.2019.103375.

[92] S. Alelyani, J. Tang, and H. Liu, "Feature Selection for Clustering: A Review," *Data Clust.*, pp. 29–60, 2019, doi: 10.1201/9781315373515-2.

[93] Z. Hong and C. Yiu-ming, "Feature Selection for Local Learning Based Clustering," *Lect. Notes Comput. Sci. (including Subser. Lect. Notes Artif. Intell. Lect. Notes Bioinformatics)*, vol. 5476 LNAI, pp. 414–425, 2009, doi: 10.1007/978-3-642-01307-2_38.

[94] L. Du and Y. D. Shen, "Unsupervised feature selection with adaptive structure learning," *Proc. ACM SIGKDD Int. Conf. Knowl. Discov. Data Min.*, vol. 2015-Augus, pp. 209–218, 2015, doi: 10.1145/2783258.2783345.

[95] A. Muller, C. Pontonnier, P. Puchaud, and G. Dumont, "Documentation for CusToM : Customizable Toolbox for Musculoskeletal simulation."

[96] R. D. Crowninshield, "Use of optimization techniques to predict muscle forces," *J. Biomech. Eng.*, vol. 100, no. 2, pp. 88–92, 1978, doi: 10.1115/1.3426197.

[97] R. A. Serway and J. W. Jewett, *Physics by serway*. 2004.

[98] "Detect and remove outliers in data - MATLAB rmoutliers." https://www.mathworks.com/help/matlab/ref/rmoutliers.html (accessed Oct. 18, 2022).

[99] A. Muller, C. Pontonnier, X. Robert-Lachaine, G. Dumont, and A. Plamondon, "Motion-based prediction of external forces and moments and back loading during manual material handling tasks," *Appl. Ergon.*, vol. 82, 2020, doi: 10.1016/j.apergo.2019.102935.

[100] C. Pontonnier, C. Livet, A. Muller, A. Sorel, G. Dumont, and N. Bideau, "Ground reaction forces and moments prediction of challenging motions: fencing lunges," *Comput. Methods Biomech. Biomed. Engin.*, vol. 22, no. sup1, pp. S523–S525, 2019, doi: 10.1080/10255842.2020.1715005.

[101] A. Muller, C. Pontonnier, and G. Dumont, "Motion-Based Prediction of Hands and Feet Contact Efforts during Asymmetric Handling Tasks," *IEEE Trans. Biomed. Eng.*, vol. 67, no. 2, pp. 344–352, 2020, doi: 10.1109/TBME.2019.2913308.

[102] D. D. Neumann, *Kinesiology To Life and Provide the*. 2010.

[103] E. Rodriguez-Villegas, S. Iranmanesh, and S. A. Imtiaz, "Wearable Medical Devices: High-Level

System Design Considerations and Tradeoffs," *IEEE Solid-State Circuits Mag.*, vol. 10, no. 4, pp. 43–52, 2018, doi: 10.1109/MSSC.2018.2867247.

[104] M. Cenciarini and A. M. Dollar, "Biomechanical considerations in the design of lower limb exoskeletons," *IEEE Int. Conf. Rehabil. Robot.*, pp. 10–14, 2011, doi: 10.1109/ICORR.2011.5975366.

[105] D. I. Fotiadis, C. Glaros, and A. Likas, "Wearable Medical Devices," *Wiley Encycl. Biomed. Eng.*, no. 3, 2006, doi: 10.1002/9780471740360.ebs1326.

[106] C. Lagoda, J. C. Moreno, and J. L. Pons, "Human-robot interfaces in exoskeletons for gait training after stroke: State of the art and challenges," *Appl. Bionics Biomech.*, vol. 9, no. 2, pp. 193–203, 2012, doi: 10.3233/ABB-2012-0056.

[107] Y. W. Hong *et al.*, "Lower Extremity Exoskeleton : Review and Challenges Surrounding the Technology and its Role in Rehabilitation of Lower Limbs," *Aust. J. basic Appl. Sci.*, vol. 7, no. 7, pp. 520–524, 2013.

[108] E. A. Kruger, M. Pires, Y. Ngann, M. Sterling, and S. Rubayi, "Comprehensive management of pressure ulcers in spinal cord injury: Current concepts and future trends," *J. Spinal Cord Med.*, vol. 36, no. 6, pp. 572–585, 2013, doi: 10.1179/2045772313Y.0000000093.

[109] J. H. M. Verschueren, M. W. M. Post, S. De Groot, L. H. V. Van Der Woude, F. W. A. Van Asbeck, and M. Rol, "Occurrence and predictors of pressure ulcers during primary in-patient spinal cord injury rehabilitation," *Spinal Cord*, vol. 49, no. 1, pp. 106–112, 2011, doi: 10.1038/sc.2010.66.

[110] A. Gélis, A. Dupeyron, P. Legros, C. Benam, J. Pelissier, and C. Fattal, "Pressure ulcer risk factors in persons with spinal cord injury Part 2: The chronic stage," *Spinal Cord*, vol. 47, no. 9, pp. 651–661, 2009, doi: 10.1038/sc.2009.32.

[111] I. Benson, K. Hart, D. Tussler, and J. J. Van Middendorp, "Lower-limb exoskeletons for individuals with chronic spinal cord injury: Findings from a feasibility study," *Clin. Rehabil.*, vol. 30, no. 1, pp. 73–84, 2016, doi: 10.1177/0269215515575166.

[112] N. L. Dudek, M. B. Marks, S. C. Marshall, and J. P. Chardon, "Dermatologic conditions associated with use of a lower-extremity prosthesis," *Arch. Phys. Med. Rehabil.*, vol. 86, no. 4, pp. 659–663, 2005, doi: 10.1016/j.apmr.2004.09.003.

[113] A. S. Gorgey, "Robotic exoskeletons: The current pros and cons," *World J. Orthop.*, vol. 9, no. 9, pp. 112–119, 2018, doi: 10.5312/wjo.v9.i9.112.

[114] M. D. C. Sanchez-Villamañan, J. Gonzalez-Vargas, D. Torricelli, J. C. Moreno, and J. L. Pons, "Compliant lower limb exoskeletons: A comprehensive review on mechanical design principles," *J. Neuroeng. Rehabil.*, vol. 16, no. 1, pp. 1–16, 2019, doi: 10.1186/s12984-019-0517-9.

[115] K. Langlois, M. Moltedo, T. Bacek, C. Rodriguez-Guerrero, B. Vanderborght, and D. Lefeber, "Design and Development of Customized Physical Interfaces to Reduce Relative Motion between the User and a Powered Ankle Foot Exoskeleton," *Proc. IEEE RAS EMBS Int. Conf. Biomed. Robot. Biomechatronics*, vol. 2018-Augus, pp. 1083–1088, 2018, doi: 10.1109/BIOROB.2018.8487706.

[116] J. T. Meyer, S. O. Schrade, O. Lambercy, and R. Gassert, "User-centered design and evaluation of physical interfaces for an exoskeleton for paraplegic users," *IEEE Int. Conf. Rehabil. Robot.*, vol. 2019-June, pp. 1159–1166, 2019, doi: 10.1109/ICORR.2019.8779527.

[117] B. Kalita, J. Narayan, and S. K. Dwivedy, "Development of Active Lower Limb Robotic-Based Orthosis and Exoskeleton Devices: A Systematic Review," *Int. J. Soc. Robot.*, vol. 13, no. 4, pp. 775–793, 2021, doi: 10.1007/s12369-020-00662-9.

[118] J. L. Contreras-Vidal *et al.*, "Powered exoskeletons for bipedal locomotion after spinal cord injury," *J. Neural Eng.*, vol. 13, no. 3, 2016, doi: 10.1088/1741-2560/13/3/031001.

[119] S. Mohammed and Y. Amirat, "Towards intelligent lower limb wearable robots: Challenges and perspectives - state of the art," *2008 IEEE Int. Conf. Robot. Biomimetics, ROBIO 2008*, pp. 312–317, 2009, doi: 10.1109/ROBIO.2009.4913022.

[120] T. E. Lockhart, "Biomechanics of Human Gait – Slip and Fall Analysis," 2013.

[121] M. Grimmer, A. A. Elshamanhory, and P. Beckerle, "Human Lower Limb Joint Biomechanics in Daily Life Activities: A Literature Based Requirement Analysis for Anthropomorphic Robot Design," *Front. Robot. AI*, vol. 7, no. February, pp. 1–17, 2020, doi: 10.3389/frobt.2020.00013.

Appendices

Appendix I – Tukey-B post hoc results

In this appendix, the Tukey-B post hoc results for the IV's included in ANOVA's models, performed in Chapter 5, will be presented.

Table A.1 Tukey-B post hoc results for perturbation IV, concerning the data collected with protocol 1

Dependent Variable	***Independent Variable***	***Subsets based on dependent variables' similarity***		
		1	**2**	**3**
Right Hip AVG	Rope Pull	9.15		
	Steady Walking		12.74	
	Biomechanical Response			16.04
Left Hip AVG	Steady Walking	12.77		
	Rope Pull	13.29		
	Biomechanical Response		17.40	
Right Knee AVG	Steady Walking	26.82		
	Rope Pull	27.38		
	Biomechanical Response		31.26	
Left Knee AVG	Rope Pull	24.39		
	Steady Walking		26.86	
	Biomechanical Response			31.60
Right Ankle AVG	Rope Pull	4.16		
	Steady Walking		7.51	
	Biomechanical Response			10.41
Left Ankle AVG	Rope Pull	3.22		
	Steady Walking		8.65	
	Biomechanical Response			10.86
Right Hip Frontal AVG	Steady Walking	-1.66		
	Biomechanical Response	-1.45		
	Rope Pull		-0.54	
Left Hip Frontal AVG	Steady Walking	-1.15		
	Rope Pull		-0.21	
	Biomechanical Response		0.052	
xCoM velocity AVG	Biomechanical Response	0.78		
	Steady Walking	0.78		
	Rope Pull	0.94		
yCoM velocity AVG	Biomechanical Response	0.15		
	Steady Walking	0.19		
	Rope Pull	0.20		
zCoM velocity AVG	Steady Walking	4.17		
	Biomechanical Response	6.64		
	Rope Pull	6.68		
Distance Foot AVG	Biomechanical Response	0.36		
	Steady Walking		0.39	
	Rope Pull			0.59

Distance CoM–Left Foot AVG	Steady Walking	100.06		
	Rope Pull	114.14		
	Biomechanical Response	114.80		
Distance CoM–Right Foot AVG	Steady Walking	45.44		
	Rope Pull	49.61		
	Biomechanical Response	49.76		

Table A.2 Tukey-B post hoc results for speed IV, concerning the data collected with protocol 1

Dependent Variable	***Independent Variable***	***Subsets based on dependent variables' similarity***		
		1	**2**	**3**
Right Hip AVG	1.8 km/h	10.52		
	Self-selected		12.52	
	5.4 km/h			15.32
Left Hip AVG	1.8 km/h	12.53		
	Self-selected		14.05	
	5.4 km/h			15.63
Right Knee AVG	1.8 km/h	22.81		
	Self-selected		29.82	
	5.4 km/h			32.49
Left Knee AVG	1.8 km/h	23.43		
	Self-selected		28.46	
	5.4 km/h			31.11
Right Ankle AVG	5.4 km/h	4.67		
	Self-selected		6.61	
	1.8 km/h			10.37
Left Ankle AVG	5.4 km/h	5.40		
	Self-selected		7.31	
	1.8 km/h			10.46
Right Hip Frontal AVG	1.8 km/h	-1.90		
	Self-selected	-1.36		
	5.4 km/h		-0.69	
Left Hip Frontal AVG	1.8 km/h	-0.97		
	5.4 km/h	-0.52		
	Self-selected	-0.32		
xCoM velocity AVG	1.8 km/h	0.36		
	Self-selected		0.80	
	5.4 km/h			1.37
yCoM velocity AVG	1.8 km/h	-0.18		
	Self-selected		0.22	
	5.4 km/h			0.57
zCoM velocity AVG	Self-selected	-0.00		
	5.4 km/h	5.09	5.09	
	1.8 km/h		9.54	
Distance Foot AVG	1.8 km/h	0.37		
	Self-selected		0.45	
	5.4 km/h			0.48
Distance	1.8 km/h	36.50		

CoM–Left Foot AVG	Self-selected	46.22		
	5.4 km/h		61.23	
Distance CoM–Right Foot AVG	1.8 km/h	86.57		
	Self-selected	101.72		
	5.4 km/h		134.98	

Table A.3 Tukey-B post hoc results for perturbation IV, concerning the data collected with protocol 2

Dependent Variable	***Independent Variable***	***Subsets based on dependent variables' similarity***		
		1	**2**	**3**
Right Hip AVG	Steady Walking	10.23		
	Biomechanical Response	12.16	12.16	
	Rope Pull		13.72	
Left Hip AVG	Rope Pull	1.96		
	Steady Walking		9.33	
	Biomechanical Response		11.34	
Right Knee AVG	Rope Pull	14.10		
	Steady Walking		24,60	
	Biomechanical Response			28.36
Left Knee AVG	Rope Pull	20.66		
	Steady Walking	22.86		
	Biomechanical Response		25.35	
Right Ankle AVG	Rope Pull	2.25		
	Steady Walking	3.22		
	Biomechanical Response		6.36	
Left Ankle AVG	Steady Walking	3.02		
	Rope Pull	4.18		
	Biomechanical Response		5.91	
Right Hip Frontal AVG	Biomechanical Response	-1.92		
	Steady Walking	-1.72		
	Rope Pull		-0.37	
Left Hip Frontal AVG	Biomechanical Response	-3.04		
	Steady Walking	-2.42	-2.42	
	Rope Pull		-1.99	
xCoM velocity AVG	Rope Pull	0.10		
	Biomechanical Response	0.10		
	Steady Walking	0.98		
yCoM velocity AVG	Steady Walking	-0.06		
	Rope Pull	0.34		
	Biomechanical Response	0.36		
zCoM velocity AVG	Rope Pull	-0.07		
	Biomechanical Response	0.00		
	Steady Walking	0.21		
Distance Foot AVG	Biomechanical Response	0.35		
	Steady Walking	0.38		
	Rope Pull		0.56	
Distance CoM–Left	Biomechanical Response	0.92		
	Rope Pull	0.93		

Foot AVG	Steady Walking	0.94		
Distance CoM– Right Foot AVG	Biomechanical Response	0.92		
	Steady Walking		0.93	
	Rope Pull			0.97

Table A.4 Tukey-B post hoc results for perturbation intensity IV, concerning the data collected with protocol 2

Dependent Variable	*Independent Variable*	*Subsets based on dependent variables' similarity*		
		1	2	3
Right Hip AVG	Soft Perturbation	10.13		
	No – Perturbation	10.23		
	Severe Perturbation	13.27	13.27	
	Intermediate Perturbation		14.28	
Left Hip AVG	Severe Perturbation	2.03		
	Soft Perturbation		8.45	
	No – Perturbation		9.33	
	Intermediate Perturbation		9.50	
Right Knee AVG	Severe Perturbation	20.97		
	Intermediate Perturbation	21.31		
	Soft Perturbation	21.46		
	No – Perturbation		24.60	
Left Knee AVG	Soft Perturbation	19.55		
	Intermediate Perturbation	22.20	22.20	
	No – Perturbation		22.86	
	Severe Perturbation			26.35
Right Ankle AVG	No – Perturbation	3.22		
	Intermediate Perturbation	4.03		
	Severe Perturbation	4.48		
	Soft Perturbation	4.52		
Left Ankle AVG	No – Perturbation	3.02		
	Severe Perturbation	4.85	4.85	
	Intermediate Perturbation	5.04	5.04	
	Soft Perturbation		5.33	
Right Hip Frontal AVG	Soft Perturbation	-2.03		
	No – Perturbation	-1.72	-1.72	
	Intermediate Perturbation	-1.21	-1.21	
	Severe Perturbation		-0.46	
Left Hip Frontal AVG	Intermediate Perturbation	-2.87		
	No – Perturbation	-2.42		
	Severe Perturbation	-2.31		
	Soft Perturbation	-2.19		
xCoM velocity AVG	Intermediate Perturbation	-0.23		
	Soft Perturbation	0.00		
	Severe Perturbation	0.56		
	No – Perturbation	0.98		
yCoM velocity AVG	No – Perturbation	-0.06		
	Severe Perturbation	0.16		
	Soft Perturbation	0.32		

	Intermediate Perturbation	0.53		
zCoM velocity AVG	Severe Perturbation	-0.04		
	Intermediate Perturbation	-0.04		
	Soft Perturbation	-0.02		
	No – Perturbation	0.21		
Distance Foot AVG	No – Perturbation	0.38		
	Soft Perturbation	0.39		
	Intermediate Perturbation		0.46	
	Severe Perturbation		0.49	
Distance CoM–Left Foot AVG	Soft Perturbation	0.92		
	No – Perturbation	0.93		
	Severe Perturbation	0.94		
	Intermediate Perturbation	0.94		
Distance CoM–Right Foot AVG	Soft Perturbation	0.93		
	No – Perturbation	0.93		
	Severe Perturbation	0.95		
	Intermediate Perturbation	0.95		

Annexes

Annex 1 – GRF and torques estimation procedure using CusToM toolbox

The present annex intends to illustrate the steps conducted, in CusToM toolbox, to estimate both GRF and joint torques.

1) The first step consists in defining the musculoskeletal model parameters. Firstly, the mass (kg) and the height (m) of the subject should be defined in the CusToM Toolbox interface. To open this window, "GenerateParameters" command should be executed in Matlab ® command window.

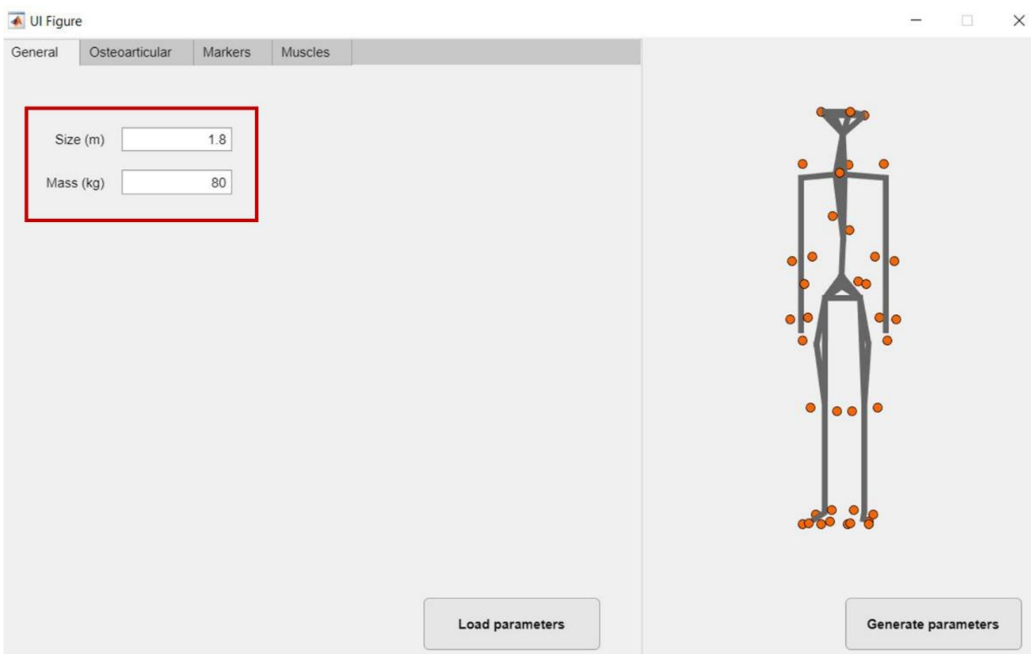


Figure A.1 Musculoskeletal model calibration: size (m) and mass (kg) parameters definition.

2) Then, it is also fundamental to guarantee that the markers of Xsens used during the protocol correspond to the markers defined in the musculoskeletal model of the toolbox. CusToM toolbox includes a pre-defined markers set list for Xsens previous collected data. When studying Xsens data, "Marker_set5" should be selected as shown in Figure A.2. This tab also allows to modify the pre-defined marker set by removing or adding a marker from the initial set.

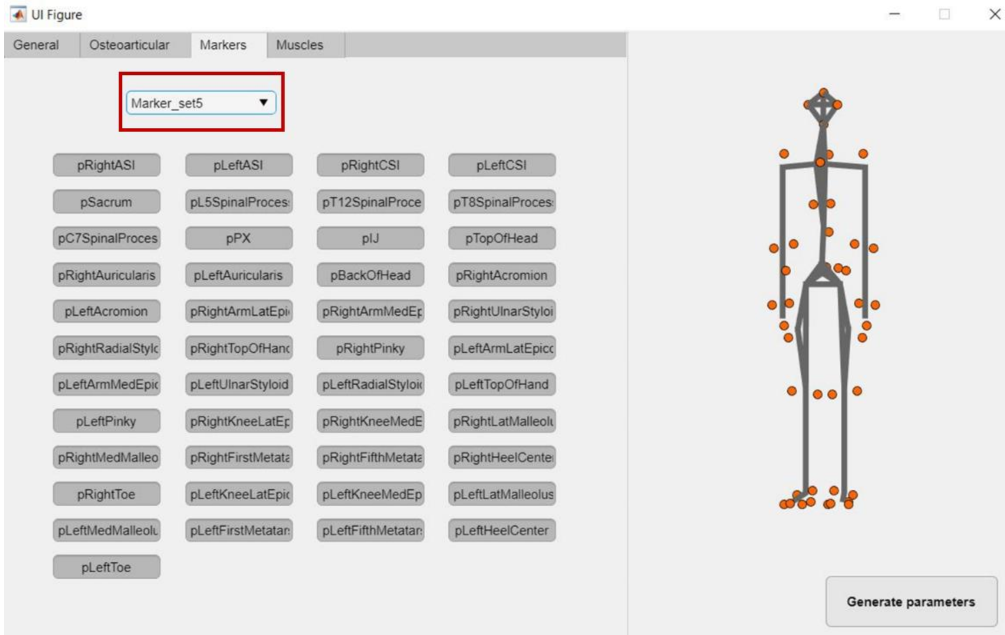


Figure A.2 Musculoskeletal model calibration: marker set definition.

3) After musculoskeletal model calibration, the Xsens file is imported by selecting "Load Parameters" button in the toolbox interface. CusToM toolbox allows to import Xsens files both in c3d and mvnx format.

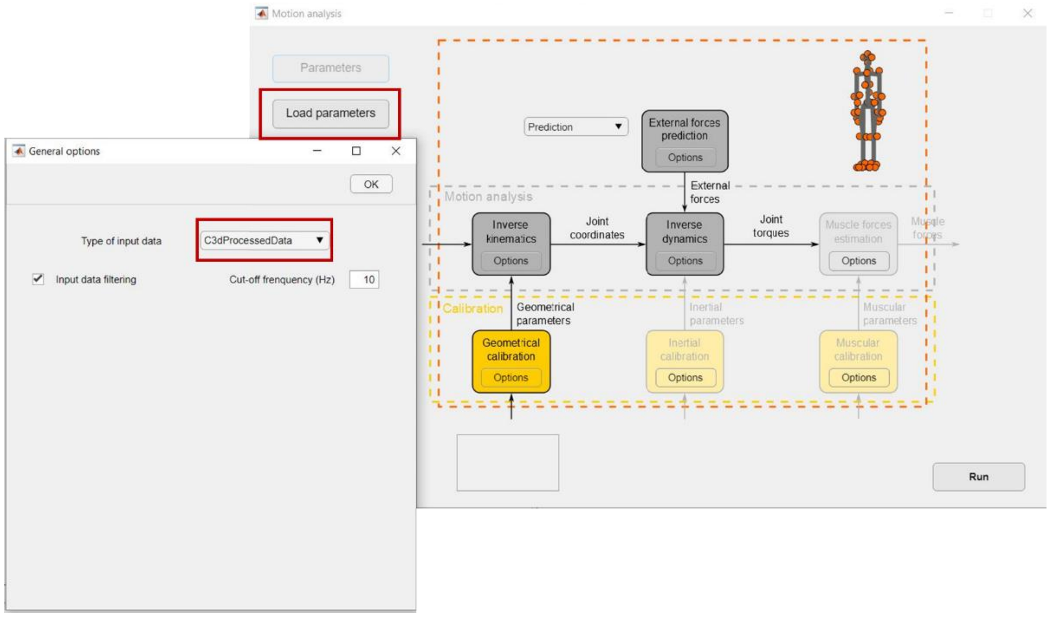


Figure A.3 Xsens file importing.

4) After importing the file, the option "Prediction" should be selected to allow the GRF and torques estimation using the inertial data. Concerning the window "External forces prediction options" shown in Figure A.4, CusToM toolbox, allows the definition of the contact points where external forces are applied. To GRF estimation, the contact points related to Right and Left feet's soles should be selected.

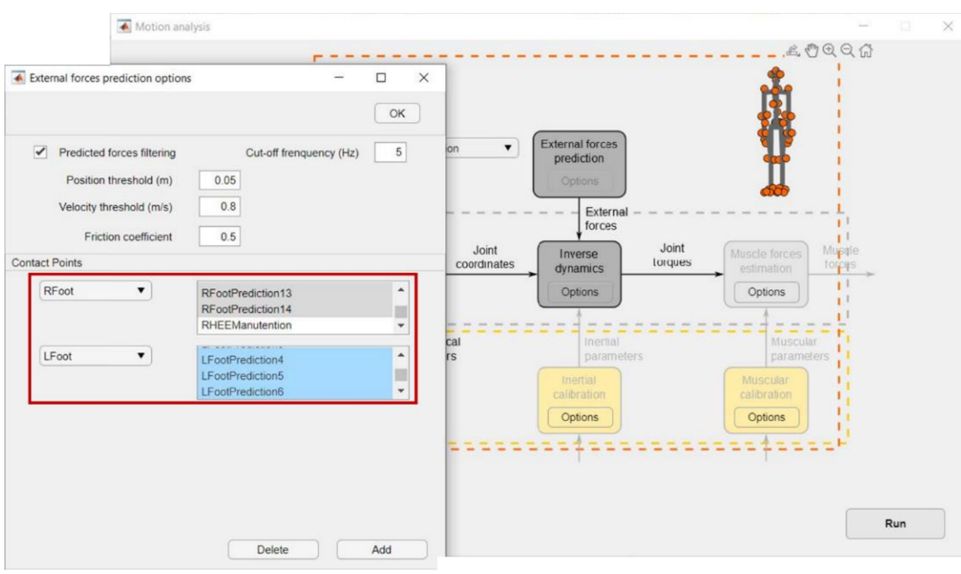


Figure A.4 GRF prediction options: contact points definition.