Universidade do Minho
Escola de Engenharia

José Henrique Machado Pires

Virtual Reality for Imbalance Induction and Analysis of Neuromuscular Postural Reactivity

Universidade do Minho
Escola de Engenharia

José Henrique Machado Pires

Virtual Reality for Imbalance Induction and Analysis of Neuromuscular Postural Reactivity

Master's Dissertation

Master's in Electronics Engineering

Work supervised by

Professora Doutora Cristina P. Santos

Nuno Ferrete Ribeiro

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 Universidade do Minho.

Acknowledgements

Na realização desta dissertação, que me permitiu crescer tanto a nível profissional como pessoal, contei com o apoio de várias pessoas. Quero expressar os meus profundos agradecimentos. Em primeiro lugar, quero agradecer à minha orientadora, professora Doutora Cristina P. Santos pela oportunidade de trabalhar neste projeto, pela confiança e acompanhamento, bem como por todos os conselhos. Quero agradecer especialmente ao Nuno Ribeiro, por me ter acompanhado do primeiro ao último momento neste desafio, partilhando conhecimento valioso e muitas vezes sacrificando o seu tempo pessoal em prol deste projeto. Deixo também um agradecimento aos colegas investigadores do BiRD Lab, por estarem sempre dispostos a ajudar. Não poderia deixar de agradecer à minha família, nomeadamente à minha mãe e avós que me proporcionaram todas as condições para a minha formação, sem nunca desistirem de mim. Por fim, o meu muito obrigado a todos, mesmo que injustamente não mencionados, que contribuíram direta ou indiretamente para a concretização deste trabalho.

Abstract

Virtual Reality for Imbalance Induction and Analysis of Neuromuscular Postural Reactivity

The occurrence of falls in a projected continuously growing elderly population, together with its impact on mortality and reduced quality of life in the over-65 age group, turned the problem of falls a public health concern. It is estimated that worldwide 684,000 people die due to a fall, which occurs an average of 37.3 million times a year. The irregularity and diversity of a fall event results in the scarcity of available datasets that incorporate biomechanical and physiological data from actual falls. Due to this limitation, the researchers adopted a procedure that simulates falls in a controlled laboratory environment. However, these simulations are not representative of a real-world fall, given the multifactorial nature of a fall. This results in a gap involving the lack of adequate data for improving fall prediction and detection algorithms. Using virtual reality and drawing on its immersive characteristics, it was possible to include the concepts of place illusion and plausibility in the virtual environment. The participant can enjoy the sensation of being in the virtual environment and behave naturally, as they would if they were experiencing the real world. To achieve this, it is necessary to design a realistic virtual environment that simulate the everyday home environment. In this virtual environment, a protocol based on visual perturbations has been designed to recreate animations to induce different types of falls. Participants in the experimental protocol will be instrumented with inertial sensors that collect full-body motion kinematic data, electromyography muscle activity sensors, and electrodermal activity sensors. With this, a strategy to collect data from compensatory postural reactions similar to real-world falls induced by visual perturbations only was proposed. A protocol introducing several visual perturbations per session, which allows the collection of kinematic and physiological data representative of reaction patterns to perturbation during gait was proposed. This action resulted in a dataset construction contributing towards a practical way to solve the real-world fall data scarcity. Therefore, it provides a direct contribution to the mitigation of the occurrence of falls by providing data for fall prediction and detection algorithms. Statistical analysis of postural dynamic reactions reveals that the developed balance perturbation tool is effective and approaches results obtained in previous work with mechanical perturbations.

Keywords: Virtual Reality, Motion Analysis, Balance Perturbation, Real-World Falls, Fall Prevention and Prediction.

Resumo

Realidade Virtual para Indução de Desequilíbrio e Análise da Reatividade Postural Neuromuscular

A ocorrência de quedas numa população idosa que se projeta em contínuo crescimento, juntamente com o seu impacto na mortalidade e redução da qualidade de vida na faixa etária acima dos 65 anos, transformaram o problema das quedas numa preocupação de saúde pública. Estima-se que mundialmente 684 mil pessoas morram devido a uma queda, que ocorre em média 37.3 milhões de vezes por ano. A irregularidade e diversidade de um evento de queda resulta na escassez de datasets disponíveis que incorporem dados biomecânicos e fisiológicos de quedas reais. Devido a esta dificuldade, os investigadores adotaram um procedimento que consiste em simular quedas em ambiente controlado de laboratório. Porém, estas simulações não são representativas da queda do mundo real. Utilizando realidade virtual e recorrendo às suas características imersivas, pode incluir-se no ambiente virtual os conceitos de ilusão de local e de plausibilidade. Assim, o participante pode usufruir da sensação de estar no ambiente virtual e comportar-se de forma natural. Criou-se um ambiente virtual realista e aproximado ao ambiente doméstico do dia-a-dia. Nesse meio virtual, concebeu-se um protocolo baseado em perturbações visuais com animações que tentam induzir todos os tipos de quedas. Os participantes serão instrumentados com um conjunto de sensores inerciais que recolhem dados cinemáticos, sensores de eletromiografia e sensores de atividade eletrodérmica. Posto isto, propõe-se uma estratégia de recolha de dados de reações posturais compensatórias semelhantes a quedas do mundo real, induzidas apenas por perturbações visuais. A estratégia engloba um protocolo de introdução de perturbações visuais que permite a recolha de dados cinemáticos e fisiológicos representativos dos padrões de reação à perturbação durante a marcha. Desta ação resultou a construção de um dataset que minimiza a escassez de dados relativos a quedas do mundo real. Logo, contribui-se diretamente para a mitigação da ocorrência de quedas, fornecendo dados para algoritmos de previsão e deteção de quedas. A análise estatística das reações dinâmicas posturais revela que a ferramenta de perturbação do equilíbrio desenvolvida é eficaz e aproxima-se de resultados obtidos em trabalhos prévios com perturbações mecânicas.

Palavras-chave: Realidade Virtual, Análise de Movimento, Perturbação do Equilíbrio, Quedas do Mundo Real, Prevenção e Previsão de Queda.

Contents

List of Figures

List of Tables

Listings

Acronyms

AD Alzheimer's Disease

ADL Activities of Daily Living

ALS Amyotrophic Lateral Sclerosis

ANOVA Analysis of Variance

AP Anterior-Posterior

APA Anticipatory Postural Adjustment

BBS Berg Balance Scale

BoS Base of Support

BPPV Benign Paroxysmal Positional Vertigo

BW Backward

CAVE Cave Automated Virtual Environment

CCI Co-contraction Index

CCW Counter Clockwise

CMEMS Center of MicroElectroMechanical Systems

CNS Central Nervous System

CoG Center of Gravity

CoM Center of Mass

CoP Center of Pressure

CP Cerebral Paresis

CPA Compensatory Postural Adjustment

CW Clockwise

DHI Dizziness Handicap Inventory

DoP Direction of Progression

EEG Electroencephalography

EKG Electrocardiography

EMG Electromyography

EXTO External Oblique

FR Functional Reach

FW Forward

GMH Gastrocnemius Medial Head

GSR Galvanic Skin Response

GVS Galvanic Vestibular Stimulation

HMD Head-Mounted Display

HS Heel-Strike

IDE Integrated Development Environment

IMU Intertial Measurement Unit

IPQ Igroup Presence Questionnaire

KPIs Key Performance Indicators

LCD Liquid Crystal Display

LGM Left Gastrocnemius Medial Head

LoS Limits of Stability

LTA Left Tibialis Anterior

MANOVA Multivariate Analysis of Variance

MDS-UPDRS-III Movement Disorder Society Unified

ML Medial-Lateral

MoCA Montereal Cognitive Assessment

MoS Margin of Stability

MS Multiple Sclerosis

MSSQ Motion Sickness Susceptibility Questionnaire

MVC Maximum Voluntary Contraction

PAR-Q Physical Activity Readiness Questionnaire

PBT Perturbation-based balance training

PD Parkinson's Disease

PROMIS Patient-Reported Outcomes Measurement Information System

PSD Power Spectrum Density

RF Rectus Femoris

RGM Right Gastrocnemius Medial Head

RMS Root Mean Square

ROM Range of Motion

RPY Roll Pitch Yaw

RTA Right Tibialis Anterior

SCI Spinal Cord Injury

SCM Sternocleidomastoid

SD Standard Deviation

SL Stride Length

SOT Sensory Organization Test

SSQ Simulation Sickness Questionnaire

ST Semitendinosus

SV Stride Velocity

SW Stride Width

TA Tibialis Anterior

TIS Trunk Impairment Scale

TO Toe-Off

TUG Timed Up and Go

UPDRS Unified Parkinson’s Disease Rating Scale

VE Virtual Environment

vHIT video Head Impulse Test

VR Virtual Reality

VRR Virtual Reality Rehabilitation

WBB Wii Balance Board

WHO World Health Organization

Chapter 1

Introduction

This dissertation presents the research carried out in the scope of the fifth year of the Integrated Master's in Industrial Electronics and Computer Engineering during the academic year of 2020/21. This dissertation was developed at BiRD LAB (Biomedical Robotic Devices Laboratory) of the Center of MicroElectroMechanical Systems (CMEMS), a research center of the Department of Industrial Electronics (DEI) at University of Minho, Braga, Portugal. The project is divided into four phases, namely: i) virtual environment design and implementation; ii) visual perturbation protocol elaboration for multisensory data collection; iii) data processing and assembling into a dataset; and iv) statistical analysis. The dissertation focuses on the development of a protocol for introducing visual perturbations that induce loss of balance to collect kinematic and electrophysiological data on postural reactions, which are expected to approximate a real-world fall. The data collection will be compiled into an extensive dataset.

1.1 Motivation

Falls and unstable balance control are among the most challenging clinical problems faced by older adults. They are a cause of substantial rates of mortality and morbidity as well as major contributors to immobility [1] and premature nursing home placement [2]. An estimated 684 000 fatal falls occur each year, making it the second leading cause of unintentional injury death, after road traffic injuries [3]. In the United States, about three-fourths of deaths due to falls occur in the 13% of the population age ≥65, indicative of primarily a geriatric syndrome. About 40% of this age group living at home will fall at least once each year, and about 1 in 40 of them will be hospitalized. Of those admitted to hospital after a fall, only about half will be alive a year later [4]. Repeated falls and instability are very common indicators of nursing home admission [2].

The problem of falls in the elderly population is a combination of a high incidence together with a high susceptibility to injury, because of a high prevalence of clinical diseases (e.g., osteoporosis) [5, 6] and age-related physiological changes (e.g., slowed protective reflexes) [7] that make even a relatively mild fall particularly dangerous. In addition, recovery from fall injury is often delayed in older persons, which in turn increases risk of subsequent falls through deconditioning. Another complication is the post-fall anxiety syndrome, in which an individual down-regulates activity in a perhaps overcautious fear of falling; this in turn further contributes to deconditioning, weakness and abnormal gait and in the long run may actually increase risk of falls [8].From a financial point of view, the fall-related injuries costs are substantial. For people aged 65 years or older, the average health system cost per fall injury in the Republic of Finland and Australia are US$ 3611 and US$ 1049, respectively. Evidence from Canada suggests the implementation of effective prevention strategies with a subsequent 20% reduction in the incidence of falls could create a net savings of over US$ 120 million each year [3]. To overcome this social and economic burden, existing systems mainly focus on detecting a fall [9] with little emphasis on fall prediction and prevention [10]. Hence, there is an urgent need for developing monitoring systems that can minimize this cost and improve the quality of life for persons who suffer from falls. Fall prediction and prevention systems are of utmost importance to accomplish this task and can help reduce the financial, physical, and emotional consequences of a fall. However, fall prediction is a challenging problem due to the combination of intrinsic and extrinsic fall risk factors that contribute to a fall [11].

A lot of research is centered around developing techniques to identify normal Activities of Daily Living (ADL) either at a basic level (e.g., walking, running, cycling) or at a higher level (e.g., preparing breakfast, washing hands). These techniques are generally applied to monitor a subject's movements, assess physical fitness, and provide feedback. Though this research is useful, scenarios may exist where detection of abnormal activities become important, challenging, and relevant. Missing out such abnormal activities can impose health and safety risks on an individual. Falling is one of the most common type of abnormal activity and the most studied [12]. Most falls are caused by a sudden loss of balance due to an unexpected slip or trip, or loss of stability during movements such as turning, bending, or rising.

The occurrence of falls is infrequent and diverse. The rarity of their occurrence lead to a lack of data to train classifiers. More than one type of fall may also occur, and their unexpectedness make it difficult to detect falls in advance. Collecting fall data can be cumbersome because it may require the person to actually undergo a real fall which may be harmful and unsafe. Alternatively, artificial fall data can be collected in controlled laboratory settings. However, that may not be the true representative of actual falls [13, 14]. Analyzing artificially induced fall data can be good from the perspective of understanding and developing insights into falls as an activity but it does not simplify the difficult problem of fall prediction and prevention [15]. Moreover, the classification models built with artificial falls are more likely to suffer from over-fitting and may poorly generalize on actual falls [12].

On an average, nursing home residents incur 2.6 falls per person per year [4]. If an experiment is to be set up to collect real-world falls and assuming an activity is monitored every second by a sensor, around

31.55 million normal activities per year are gathered in comparison to only 2.6 falls. The data for real falls may be collected by running long-term experiments in nursing homes or private dwelling using wearable sensors or video cameras. However, the fall data generated from such experiments will still be skewed towards normal activities and it is difficult to develop generalizable classifiers to identify falls efficiently [16]. In addition to very few or no labelled data, the diversity and types of falls further make it difficult to model them efficiently [12]. It would also be more efficient to merge information from different wearable and environmental sensors to cover extrinsic risk factors. This merger would add to the clinical value of a fall risk assessment [17]. Some studies have already shown to be efficient using Electromyography (EMG) for gait analysis [18, 19]. Despite the advantages and widespread use of inertial sensors, there is a lack of research that explores the collection of biomechanical and physiological data (e.g., galvanic skin response, blood pressure, heart beat).

1.2 Problem statement and scope

Due to the rarity of the natural occurrence of falls and the inherent difficulties in collecting biomechanical and physiological data in a non-obstructive and user-friendly way in community-dwelling older adults, there is a lack of public falls datasets. Developing immersive VR environments with the concepts of place illusion and plausibility illusion [20] creates a medium in which people respond with their whole body, treating what they perceive as real. A HMD is able to present scenarios endowed with these concepts. In such device, the displays are mounted close to the eyes, and head tracking ensures that the left and right images update according to the head movements of the participant concerning the underlying VE. The separated left and right images for each eye ensure stereo vision. The participant has the illusion of moving through a surrounding, three-dimensional environment that contains static and dynamic objects.

Balance control is a complex skill composed of three subsystems: the proprioceptive, vestibular, and vision system, which work together to keep us aware of our surroundings and give us the ability to react to current conditions and prepare for future changes. Immersive virtual reality completely changes our visual perception of the surrounding environment, providing a loss of balance by itself [21]. Inducing balance disturbances through audiovisual stimuli to impose changes in postural control is the core action point, attempting to mimic a realistic fall. These postural reactions induced by imposing a conflict between the visual and proprioceptive systems, are going to be recorded and analyzed to cover the gap in existing data regarding real falls. These postural reactivity patterns to visual disturbances are recorded using inertial sensors as a motion tracking system, capturing biomechanical and kinematic data, and physiological sensors such as EMG and Galvanic Skin Response (GSR) to record muscle and electrodermal activity signals.

After signal acquisition from multiple sensors, a feature extraction technique is applied to retrieve meaningful information. Since data gathered from sensors contain undesired information, filtering techniques are essential. After filtering the collected data, appropriate features are selected. Since analyzing a high number of features requires a large amount of memory, finding the optimal feature set can improve the system's performance [11]. The variables collected will build a large dataset, satisfying the problem presented initially. In addition to its extension, it has the advantage of incorporating kinematic, muscular and physiological data, collected in a virtual environment endowed with ecological validity and naturalism. These points give the data collection a very strong connection with a collection in a real-world environment. Innovatively, while the occasional use of visual perturbations speeds up the gathering of data about imbalanced circumstances, their ongoing usage encourages the training of postural reactions, contributing significantly for biomechanical studies on how to deal with falls or imbalances. From the perspective of mitigating the occurrence of falls in the elderly, this work allows a future design of a balance training tool strictly based on visual perturbations.

1.3 Goals and research questions

1.3.1 Goals

The project's overall purpose is to reduce the occurrence of falls, especially in the older community. The concrete goal of this dissertation is to prove that one can collect more realistic data of postural reactions and pre-impact falls using virtual reality rather than simulated falls in a laboratory setting, and match this data to real-world falls. To be able to prove this, several step-goals were set.

Goal 1. Conduct a comprehensive literature review of publications carried out with HMD that introduce both visual or physical disturbances with the intent of disrupting balance and analyze the compensatory reactions. From this survey, understand which virtual environments and virtual reality equipment have been used the most. In addition, understand which visual disturbances are most commonly used to cause imbalances, and their parameters. This review also provides account for which sensor systems are most common and their locations, the calculated metrics and outcome measures used to assess the imbalance caused. The process that led to the achievement of this initial goal is described in Chapter 3. Key Performance Indicators (KPIs): state of the art that provides information on i) frequently used visual perturbations and ii) specifications for experimental protocols with virtual reality.

Goal 2. Design and build a virtual environment in Unity software. In addition, create an automatic delivery mechanism for visual disturbances, with a temporal registry in a log from the perturbations onset and end time, in order to assist in the labeling process; Establish the connection between the motion capture system and the software, in real time, to allow the representation of an avatar of the participant's whole body. Chapter 4 defines the requirements for the virtual environment design and describes the scripts used to operate the VE with a subject. Chapter 5 demonstrates the motion capture system that allows real-time representation of the avatar. KPIs: i) SteamVR average quality performance index: High; ii) hardware to software latency < 30ms.

Goal 3. Draw an experimental protocol for strictly visual disturbances that encompasses all categories of falls. To do this, prepare a list of visual disturbances to be introduced, outline the number of times the participant will experience these disturbances, and calculate an average data collection session time. In addition to this, make the choice of the sensors to be used and their location, especially the muscle activity sensors due to their limitation in number (8 sensors for 8 muscles). Finally, prepare a random sequence of perturbations to be introduced during the protocol, with some constraints, namely the impossibility of sequentially using the same perturbation to avoid an habituation effect. The experimental protocol is designed in Chapter 5, section 5.3. KPIs: i) cover all relevant identified fall types with a corresponding visual perturbation that is capable of causing them.

Goal 4. Carry out the data collection, recruiting 12 participants to undertake the experimental protocol, outlined in Chapter 5. KPIs: i) synchronous data collection (latency < 1ms); ii) Simulation Sickness Questionnaire (SSQ) total score from negligible (<5) to minimal (5-10).

Goal 5. Perform raw data processing, filtering and down-sampling to achieve synchronization between the sensor systems. Once the data is processed, ensure the same number of samples between the various systems and compile all the metrics coming from the various systems into a dataset. This dataset will have to be labeled through an automatic process by Matlab script, process that is described in Chapter 5. KPIs: i) build a fully labeled dataset with kinematic and physiological variables, comprising ADL, normal gait and disturbed gait.

Goal 6. Statistical analysis to validate the effectiveness of the visual perturbations introduced. Understand the differences introduced by visual disturbances in the created dependent variables. Statistically infer the significance of the visual disturbance condition relative to undisturbed gait. Further, to understand which visual disturbances had the most effect on the kinematic and muscle metrics, thus creating an ordered list of the most efficient disturbances. Chapter 6 addresses this objective. KPIs: i) one-way MANOVA p-value < 0.001 indicating a statistical effect of the perturbation factor on all independent variables analyzed multivariately; ii) one-way ANOVA p-value < 0.001 to reveal the effect of the different visual disturbances on each dependent variable.

1.3.2 Research Questions (RQs)

Once the six stages described in the previous section have been achieved, it remains to answer the research questions that will be clarified.

- **Research question 1:** Can a virtual reality headset introduce imbalances through visual perturbations? Can they cause postural reactions typical of a fall?

Statistical analysis will reveal the answer to the first question. If statistical relevance that differentiates representative parameters of postural imbalance in gait between undisturbed and disturbed situations is found, the question is validated. Furthermore, a qualitative comparison will be made between the parameters collected in this study and those mentioned in studies with physical disturbances. If this comparison shows similarities, the validity of the first question is reinforced. This first research question is related to Goal 6, and it is explored in Chapter 6.

- **Research question 2:** Which visual perturbation challenged the participants' balance the most?
  In order to design a balance training tool or assess the risk of falling due to exposure to the disturbance, it is necessary to rank the visual perturbations that challenge balance the most and those that do not present any difficulty. Thus, only the most effective perturbations are selected in designing such a tool, depending on the target population. Chapter 6 details the answer to this question, which is posed by Goal 6.

- **Research question 3:** Which virtual situation placed the participant over the most anxiety?
  With the electrodermal data, one can first identify the situations that put the participant under stress and associate this state with the deterioration of the ability to maintain balance. In addition, it gives insight into the perceived reality of the virtual environment. This question is addressed in the statistical analysis chapter, Chapter 6.

- **Research question 4:** What influence does the real-time representation of the avatar have in situations of virtual heights?
  The influence of the avatar in situations that may place the participant under stress is an understudied issue in the literature. Perhaps because of the lack of studies that incorporate the avatar in real time. For this reason, an answer to this question will be provided by comparing vertigo situations where the participant can see the own body and in those where they cannot. If there is statistical significance between these two conditions, one can answer the role of the avatar in maintaining balance or not. This question is addressed in Chapter 6.

- **Research question 5:** Is it possible for a habituation phenomenon to occur to visual disturbances?
  The importance of this question is to validate the results obtained throughout the trials and understand if there is a training effect component throughout the experimental procedure.

1.4 Contribution to knowledge

This dissertation has contributed to further scientific knowledge about the patterns of compensatory postural reactions when an individual is exposed to external disturbances, in this case, visual only. More precisely, it contributed the following content:

- Research in the literature into the history and importance of virtual reality in various fields of knowledge, with particular focus on neurorehabilitation and the elderly population. A brief financial analysis was also conducted that reveals the emergence of this technology and its growing use in participants with neurological problems or gait rehabilitation. Furthermore, this review distinguishes between non-immersive and fully immersive virtual reality.
- A second literature review, systematically following the PRISMA search method, brings together the studies on the imposition of balance perturbation employing a head-mounted display. The result is an analysis highlighting the existing virtual environments and the most commonly used apparatus and a selection of the most commonly used visual disturbances. Besides, it was possible to understand the existing weaknesses in this type of investigation and the future direction proposed by the authors. From here, the lack of sensor fusion of motion and muscle activity acquisition with sensors that collect biological signals was identified. Furthermore, introducing a single visual disturbance was identified as a weakness in the investigations conducted.
- From our knowledge, the virtual environment is a novelty in terms of the variety of virtual locations that the participant can be immersed in, resulting in a realistic virtual environment endowed with high ecological validity and resembling a home-living scenario.
- The visual perturbation protocol introduced by the headset is also a novelty concerning the variety of perturbations that induce all the types of falls most prone to occur.
- From this perturbation protocol, it was possible to compile a comprehensive dataset with kinematic, muscle, and electrodermal activity parameters.
- The results derived from the compensatory reactions when trying to induce a fall show strong evidence of the effectiveness of the perturbation protocol in mimicking an actual fall.
- Suggestion of an extended perturbation-based training protocol specific to train certain muscle groups and instigate particular kinematic parameters.

1.5 Thesis outline

This dissertation intends to demonstrate that the virtual environment and the applied visual perturbations can trigger compensatory postural reactions comparable to those identified from a pre-impact fall. To this end, this dissertation is organized into seven chapters: Chapter 2 provides a historical and conceptual introduction to VR, non-immersive devices, and more significant headsets. This chapter contains specific information regarding neurorehabilitation, following a rational explanation of the advantages of VR in motor learning and rehabilitation. This review, divided into two parts: immersive and non-immersive VR, gives an overview of the most commonly used devices, the patients most subject to intervention, the existing balance and fall risk assessments, the most commonly used sensor systems, and the most common types of intervention. Chapter 3 is the result of a systematic review of the literature. This review aims to assemble existing works introducing visual perturbations through VR-HMD to participants to investigate their compensatory postural adjustments. This chapter points out the specific purposes of the studies included in the eligibility criteria: visual perturbation delivered through HMD and analysis of compensatory postural adjustments. It investigates the most commonly used headsets and sensor configurations. It also explores the collected and calculated metrics indicative of imbalance and the main results from this data collection. Finally, it makes a critical appraisal of the limitations and strengths of the studies included and future research paths. Chapter 4 presents a solution proposal for the problem explained. This proposal comprises a conception and development of a virtual environment. This virtual environment must comply with certain characteristics that are outlined. For its construction, a specific software was used, which requires an introduction to it and an explanation of related concepts. The purpose of the virtual environment is to have a platform to introduce visual disturbances to the participant through the headset. From the visual disturbances described in the studies in Chapter 3, the selection of visual disturbances to present to the participant is justified, with the intent of eliciting compensatory reactions that approximates several types of real-world falls. Chapter 5 describes the participants who were part of the experimental protocol, all the equipment used to do the data collection, and more specifically the tasks performed in the virtual reality balance perturbation protocol. Once the data has been collected, it is then time to process it in order to compile a dataset. Chapter 6, the statistical analysis, presents the statistical method chosen and its assumptions to allow the analysis. In addition, it will present the statistical variables that were adapted from the parameters calculated earlier. A statistically relevant result that can help to support the answers to the research questions is presented at the end of the chapter. A discussion section of the results completes the chapter. This section attempts to approximate the results obtained with results obtained in the literature. Finally, Chapter 7 outlines conclusions supported by the results presented so far. At this point, it concludes on the efficiency of the proposed strategy and the feasibility of its use in future work. In addition, some limitations encountered during the course of the project and indications for future improvements are pointed out.

Chapter 2

Virtual Reality for Rehabilitation

2.1 Introduction

(a) HTC Vive full kit with basestations and controllers [22].

(b) HTC Vive Pro [23]

Figure 1: HTV Vive and Vive Pro products.

The concept of VR consists of a technology capable of allowing users to view and interact directly with a graphically simulated 3D environment. The essential components for obtaining a VR system are a computer that allows the design of 3D images, visualization devices for the user to see the virtual environment, and potentially hardware that monitors their movements or produces haptic and force responses as feedback. The term virtual reality was first coined in the late 1980s by Jaron Lanier, and since the creation of the first head-mounted display in 1965 by Ivann Sutherland [24], in 1993 Nintendo entered the market, making this technology accessible to the public, with the launch of Virtual boy (VR-32) [25, 26]. In 2012, Oculus Rift was presented for the first time and later bought by Facebook. In 2013, Valve discovered low-persistence displays, making a lag-free and smear-free display of VR content. In early 2014, Valve showed off their SteamSight prototype, the precursor to both consumer headsets released in 2016. It shared major features with the consumer headsets, including 1000 displays per eye, low persistence, positional tracking over a large area, and Fresnel lenses. HTC and Valve announced the virtual reality headset HTC Vive and controllers in 2015. The set included lighthouse's tracking technology from wall-mounted base stations for positional tracking using infrared light, as seen in Figure 1 (a). In 2014, Sony announced a virtual reality headset for the PlayStation 4 video game console. In 2015, Google announced Cardboard. Also, in 2015 Razer unveiled its open-source project, OSVR. By 2016 at least 230 companies were developing VR-related products. Amazon, Apple, Facebook, Google, Microsoft, Sony, and Samsung had dedicated VR groups. In 2016, HTC shipped its first units of the HTC Vive SteamVR headset. They marked the first major commercial release of sensor-based tracking, allowing for the free movement of users within a defined space. On 8 January 2018, HTC unveiled an upgraded Vive model known as HTC Vive Pro - Figure 1 (b). It features higher-resolution displays, now at 1440x1600 resolution per eye, along with a second outward-facing camera, attachable headphones, a microphone for noise cancellation analysis, and a refreshed design with a more balanced form, lighter weight, and a sizing dial. Since the construction of the proclaimed first head-mounted display, the temporal evolution of these devices is graphically visible in Figure 2 through a timeline.

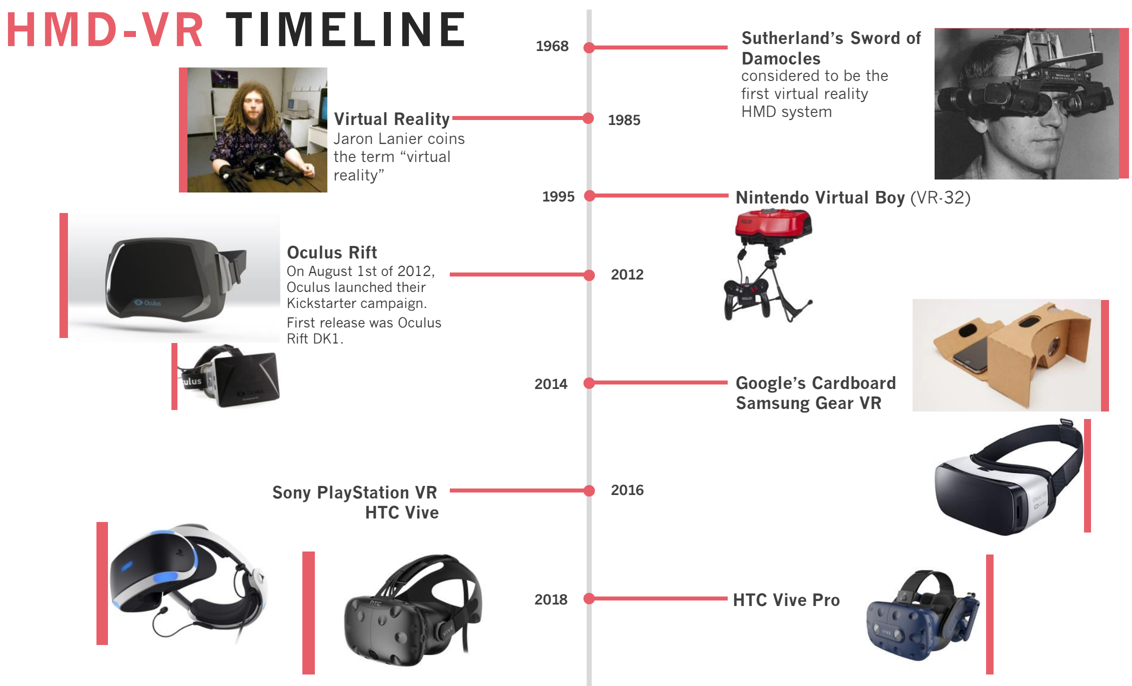


Figure 2: Inforgraphy of the evolution over time of virtual reality, specifically head-mounted displays.

The development history of the devices cannot be dissociated from the market analysis to determine whether it is a promising and profitable industry. Virtual reality continues to be a dominant force in the gaming industry. Nevertheless, it is also a tremendous asset for businesses. The increasingly sophisticated and diverse uses such as employee training, remote collaboration, testing, and prototyping, combined with creative use, have sparked industry interest, even though it is a recent technology, as seen in the Figure 2 timeline. It has also aroused the interest of big-tech and large manufacturing and software development companies, who have bet and placed themselves in this market. A market research report published in May 2022 [27] states that the global virtual reality market has been valued at $11.64 billion for the year 2021. What captivates the interest in conducting research with this technology, particularly with the contribution of knowledge to the automotive industry, healthcare, and education, among others, is the projected market size growth from 16.67 billion USD in 2022 to 227.34 billion USD in 2029. The VR market exhibits a compound annual growth rate of 45.2% over the forecast period. This report includes devices and software such as Quest 2, Google Cardboard, Unity Virtual Reality Development Software, and others. This paragraph highlights the potential for short-term investment in virtual training, engineering, and maintenance and outlines potential growth in the healthcare industry. Improved healthcare provision, patient care, and medical training place this area as a trend in the use of virtual reality. Since this dissertation fits into healthcare and utilizes virtual reality technology, it is a factor that makes the investigation relevant and with possible future profitability.

It is common to separate virtual reality into two categories. The concept of immersion underlies the decision between categories. Virtual reality can be categorized into two types: immersive and non-immersive virtual reality. Table 1, taken from [28], lists the immersion categories and a few examples of the associated technology. This separation does not mean that there cannot be immersion of an individual in a virtual environment just because a projection in three dimensions is used. The concept of immersion, along with that of place and plausibility illusion will be clarified to better understand the interest of more immersive virtual reality systems in a variety of sectors and contexts. To be considered an immersive experience, the system must include a set of displays and a tracking system, at least from the head position, so that the displayed visual information varies according to the user's movement. Such systems allow the user to move freely through the virtual environment, orient his head arbitrarily, and perceive space. It gives the sense of being in a real space called place illusion and that the presented scene is happening – plausibility illusion [20]. Immersion is an objective aspect of virtual reality environments, whereas presence is a psychological, perceptual element - the "feel of being there." So, immersion is an essential feature of VR research because it influences a user's VR experience and affects their sense of presence [20, 29].

Table 1: Immersion categories

	Non-Immersive	**Semi-Immersive**	**Fully Immersive**
Viewing Mediums	Computer monitor, TV screen	Panoramic TV	Head Mounted Display (HMD), CAVE
Cost	Low	Medium	From low (HMD) to high (CAVE)
Sense of Immersion	Low	Medium-High	High

Due to the unique immersive characteristics of virtual reality, this technology has been used in a wide range of domains, namely in the: i) training of military practices [30], ii) education [31], iii) surgeons training [32], iv) architecture, v) high-performance athlete training [33], and vi) healthcare research, including the physical, mental and social aspects [34, 35]. Within the health field, motor rehabilitation is one of the areas that is advancing rapidly and has several benefits when combined with VR systems. Once again, it is important not to divide the statement about the increasing use of virtual reality in neurorehabilitation from a social and financial perspective. Neurological disorders have a high prevalence together with short and long-term impairments and disabilities. A stroke or even Alzheimer's Disease (AD) can be fatal. Others, such as chronic headaches or seizure disorders, cause significant disabilities. Therefore, they are an emotional, financial, and social burden to the patients, their families, and their social network. Globally, in 2016, neurological disorders were the second leading cause of death (9 million deaths). The absolute number of deaths from all neurological disorders combined increased by 39% between 1990 and 2016 [36]. Neurological disease is the number one disease category causing disability globally and ranks number three in Europe. The total cost of brain disorders was estimated at €798 billion in 2010. Direct costs constitute the majority of costs (37% direct healthcare costs and 23% direct non-medical costs), whereas the remaining 40% were indirect costs associated with patients' production losses. On average, the estimated cost per person with a brain disorder in Europe ranged between €285 for headaches and €30,000 for neuromuscular disorders. In terms of the health economy, brain disorders likely constitute the number one economic challenge for European health care now and in the future [37]. The highlighted economic challenge puts

Table 2: Neurological disease cost estimation from 2004 to 2010 [36]

	Estimates in 2010			**Estimates in 2004**		
	Number of subjects (million)	Costs per subject (€PPC, 2010)	Total costs (million €PPP, 2010)	Number of subjects (million)	Costs per subject (€PPC, 2004)	Total costs (million €PPP, 2004)
Addiction	15.5	4227	65,684	9.2	6229	57,275
Anxiety disorders	61.3	1076	65,995	41.4	999	41,372
Brain tumor	0.24	21,590	5174	0.14	33,907	4586
Dementia	6.3	16,584	105,163	4.9	11,292	55,176
Epilepsy	2.6	5221	13,800	2.7	5778	15,546
Migraine	49.9	370	18,463	40.8	662	27,002
Mood disorders	33.3	3406	113,405	20.9	5066	105,666
Multiple sclerosis	0.54	26,974	14,559	0.38	23,101	8769
Parkinson's disease	1.2	11,153	13,993	1.2	9251	10,722
Psychotic disorders	5.0	5805	29,007	3.7	9554	35,229
Stroke	1.3	21,000	26,641	1.1	19,394	21,895
Traumatic brain injury	1.2	4209	5085	0.71	4143	2937
Total	178.5	2672	476,911	127.0	3040	386,175

the market for neurorehabilitation-related products on the rise. The neurorehabilitation devices market is expected to register a Compound Annual Growth Rate of 16.4% during the forecast period of 2019 to 2025 and was valued at 793 million USD in 2018 [38]. The demand for neurorehabilitation devices is increasing due to the rising number of neurological disorder cases, the emergence of robotic rehabilitation, a surge in the geriatric population, and the effectiveness of gaming systems in neurorehabilitation. However, strict regulatory policies and increasing requirements of skilled professionals are expected to hamper the market growth during the forecast period. The market is expected to witness profitable growth due to the rise in neurological disorder cases such as Parkinson's Disease (PD), AD, stroke, epilepsy, and Multiple Sclerosis (MS). Furthermore, researchers are engaged in evaluating the effectiveness and the appropriateness of adopting commercial games for neurorehabilitation. In the global neurorehabilitation market, Europe held a substantial share. This can be attributed to the rising prevalence of neurological diseases, product approvals, and the presence of developed economies. Approximately 85,000 neurologists in Europe currently provide care to these patients, which corresponds to approximately 10,000 patients per neurologist. However, there are huge disparities across European countries, ranging from 2,500 patients to 46,000 patients per neurologist [39]. For this reason it is important to develop models using new technologies that make neurorehabilitation accessible. Virtual reality, as pointed out, is a technology that has been implemented in the treatment of neurological diseases. This leads to the need to investigate the possibilities of complementing existing treatments, requiring brain research funding to flatten these social gaps of treatment accessiblilty. The following paragraph identifies a sub area of neurorehabilitation that has been receiving increasing attention.

There are several reasons why training with VR is of greater interest in motor learning and has better results [40–42]. The key fundamentals of motor rehabilitation that can be improved with the contribution of virtual reality are repetitive practice, feedback (proprioceptive and exteroceptive), and motivation [43, 44]. The advantage of repetitive practice in motor learning was already known. However, these advantages, which translate into changes in the cerebral cortex, do not happen just by massive practice - there must be a designated task or an objective, which must be achieved by the patient by trial and error [44], receiving feedback on their performance [45]. Motivation and dedication are necessary for a repetitive task, two typical characteristics in a virtual reality user experience [46]. Feedback in virtual environments can be provided in real-time in a very intuitive way or known right after a training block; it was shown that it results in changes in the level of cortical plasticity and subcortical cells and synaptic connections [47, 48]. However, repetition alone is not enough to induce changes in the motor cortex related to motor learning. Just increasing the frequency of use of a member does not produce significant changes. It is necessary to produce skilled limb movements [49]. Humans can learn motor skills in a virtual environment and transfer that motor learning to a real-world scenario [50].

There are several ways to classify Virtual Reality Rehabilitation (VRR). An obvious one is related to the specific patient population destined. Thus, one can distinguish musculoskeletal VRR, post-stroke VRR, and cognitive VRR. Musculoskeletal (orthopedic) patients who suffered a bone or muscle/ligament injury are younger and more numerous than other patients needing rehabilitation. The cognitive patient population groups individuals with various psychological disorders range from attention-deficit/hyperactivity to eating disorders to post-traumatic stress and phobias [51, 52]. Injuries, degenerative diseases, and structural defects can impair the nervous system. Some of the conditions that may benefit from neurological rehab may include ischemic or hemorrhagic strokes, trauma such as brain injury and Spinal Cord Injury (SCI), peripheral neuropathy, degenerative disorders such as PD, MS, Amyotrophic Lateral Sclerosis (ALS) or AD. Neurorehabilitation virtual reality interventions, with the paradigm shift of neurologic care, hopefully, is a junction that can facilitate the development of applications in the treatment of neurological patients with a significant impact on patient wellbeing [53]. Neurologic diseases interventions turned away from the primitive idea of a brain injury permanent effect on function and activity and became aware of the brain's regenerative potential, as well as dynamic brain reorganization [54, 55]. Optimal brain changes and recovery can happen, requiring controlled and intensive stimulation of affected brain networks [56]. On the other hand, human factors problems are associated with VR applications [57]. While it might seem intuitive that more immersive virtual environments would be best, they can generate cybersickness [58]. Common symptoms include nausea, ocular problems such as eye strain and blurred vision, disorientation and balance disturbances, and altered eye-hand coordination.

Although the common denominator of this literature review is the use of virtual reality in neurorehabilitation, there are two main sections: non-immersive and immersive virtual reality. It is important to clarify that there is frequently an inconsistency in the terminology used in published studies [59]. The term virtual reality is used when referring to any type of computerized rehabilitation, whether it is a 3-D representation of a virtual environment or 2-D monitors. Due to the lack of clarity in the use of the term VR, Figure 3 presents the taxonomy of rehabilitation systems that use virtual reality, which was taken into account when separating the chapter into two key sections. The descriptions made in the non-immersive virtual reality section 2.2 and in the immersive virtual reality section 2.3 will be concordant with this taxonomy. In this figure, semi-immersive reality represents non-immersive virtual reality in this text.

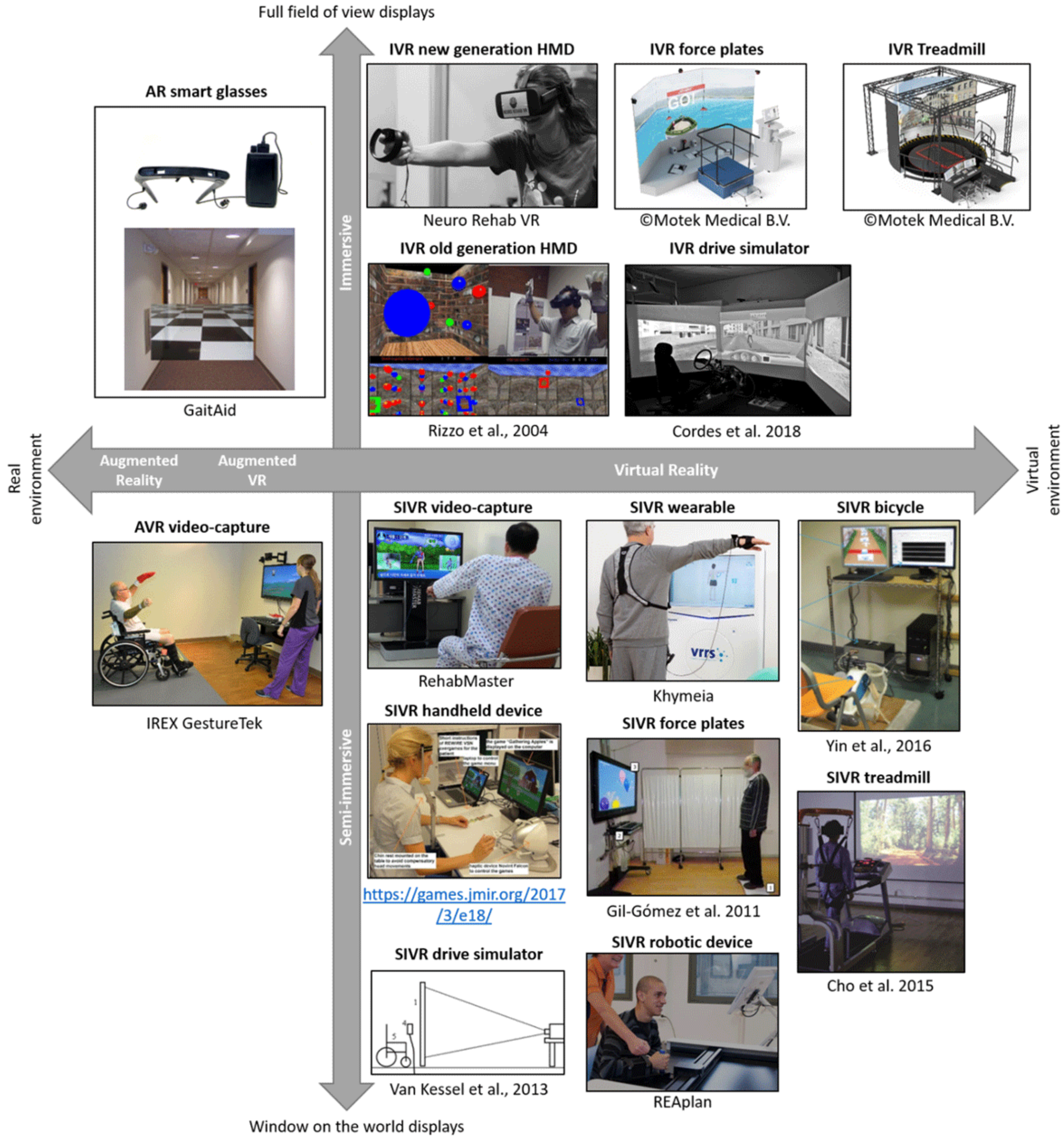


Figure 3: Taxonomy of virtual reality rehabilitation systems, based on the extent to which real and virtual information is mixed, the level of immersion and the main input device. AR = augmented reality, AVR = augmented virtual reality, IVR = immersive virtual reality, SIVR = semi-immersive virtual reality. HMD = head-mounted display. Figure reproduced from [59]

Each of these sections follows the same structure: i) equipment and sensors used; ii) outcome measures evaluated and iii) intervention protocols. At the end of the chapter, limitations and future directions are presented. Finally, conclusions are presented. These two final considerations encompass both categories of virtual reality.

2.2 Non-immersive Virtual Reality

Non-immersive virtual reality, in which visual stimuli are delivered using two-dimensional representations, is usually identified in the literature as exergaming (exercise + gaming), or serious games. Non-immersive VR rehabilitation programs are applied to develop four primary outcomes: motor control, balance, gait, and strength. Improving these outcomes is an expected goal in the rehabilitation of post-stroke, patients with Cerebral Paresis (CP), SCI, PD, MS, patients with impaired coordination like ataxia, or even epileptic and dyslexic patients. In addition to these conditions, VR has been proposed and used as an assistive rehabilitation technology for individuals suffering from severe burns [60] or Guillain-Barré syndrome [61]. Stroke patients receive the most treatment with non-immersive VR, perhaps because it is the disease with the highest incidence in developed countries and ranks number 5 among all causes of death [36]. Most studies in post-stroke patients who used VR technology attempted to assess the efficiency in improving upper limb functions. Lower limb functions rehabilitation, posture rehabilitation - static and dynamic balance - cognitive and motor rehabilitation, finger fine motor movements, and sensorimotor function were also carried out [42]. There is a research increase with non-immersive VR using screen-based virtual reality and off-the-shelf gaming consoles with accompanying games related to motor rehabilitation in the elderly population. It focuses on balance training, strengthening the lower limbs, promoting mobility and quality of life, with the ultimate aim of preventing falls and treating fear of falling [62]. When discussing neurorehabilitation studies whose population is the elderly, it makes more sense to think about rehabilitating the detoriating effects that aging has on balance control. Such effects may be the consequence of some pathology [63] or simply the aging of sensory systems and neuromuscular control mechanisms [64, 65]. Once it was clarified what type of systems and what level of immersion is referred to when introducing non-immersive virtual reality information, the groups of patients most intervened by this technology were identified. Qualitative information gathered in the following points is provided, with the structure previously suggested.

2.2.1 Equipments

In this type of rehabilitation with VR, a feature connecting all the systems is their similar construction. They consist of control units, elements projecting a VR environment, and various peripheral devices (force platforms or movement sensors). The most straightforward visual display device is a computer monitor. Using a Liquid Crystal Display (LCD) projector and a large wall screen as the monitor will enhance the sense of depth perception and thus a sense of presence [66]. These set-ups are relatively cheap and easy to use, do not require glasses or wired headsets, and allow both therapist and patient to view the same scene. Consumer-driven forces for new ways to interact with video games have led to the development of video capture (infrared and RGB optical sensors/cameras), inertial sensing devices, and pressure sensors for

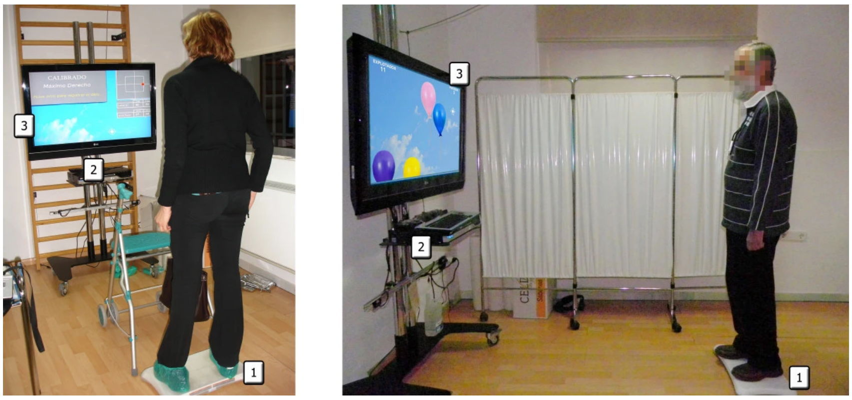


Figure 4: Patients playing with the system. The system consists of: 1) a WBB; 2) a PC; 3) Video display. Figure reproduced from [72]

measuring body movement. The most widely used sensors in exergame input devices include accelerometers, gyroscopes, infrared and RGB optical sensors, cameras, and pressure sensors [67]. The following describes the main consumer videogame console systems and exergames that health researchers and clinicians have used. The Nintendo Wii game console (Nintendo; Redmond, WA, USA), which Nintendo introduced in 2006, is the most commonly used technology and a well-known commercial off-the-shelf game system that uses inertial sensors [68]. The system's popularity is primarily due to the new approach to videogame interaction enabled through the Wii Remote™ (or Wiimote; Nintendo of America Inc. WA, USA), a wireless hand-held controller that embeds a three-axis accelerometer and a single- and dual-axis gyroscopes [69]. By fusing the sensor data from the gyroscopes and the accelerometers, the Wii Remote can measure changes in direction, speed, and acceleration with a sensitivity of ±1% [70]. The remote has an additional optical sensor on the controller that measures the position of a sensor bar mounted on the television, which emits two infrared light signals. The Wii controller can measure both rapid (using inertial sensing) and slow (using the optical sensor) movements using these sensors [71]. Figure 4 exemplifies a setup from a neurorehabilitation intervention. As noted, it follows a common design. Contains a control unit, a display element and a peripheral device - a force platform.

Inertial sensors have also been used in wobble boards for balance training [73, 74]. These boards consist of an unstable plate, which causes the user to wobble while standing on the plate, thereby controlling the game by shifting his weight. The movements are measured using a single orientation tracker (Xsens MTx Motion Tracker, Xsens Technologies, The Netherlands), consisting of three gyroscopes. The Wii Balance Board™ (WBB; Nintendo of America Inc., WA, USA) is a peripheral device (51x31 cm) that consists of four force transducers allowing calculation of the center of pressure used for game control. Exergames such as Wii Sports™ (Nintendo of America Inc., WA, USA) require players to use the Wiimote to mimic actions performed in real-life sports. Wii Fit™ (Nintendo of America Inc., WA, USA), another exercise-based game for the Nintendo Wii, makes use of the WBB. The game contains over 40 activities designed to engage the player in physical exercises that focus on maintaining the center of balance. When a player shifts their Center of Pressure (CoP), their onscreen avatar shifts its position on screen accordingly [67]. Comparable systems are pressure mats or panels with pressure sensors [75]. Inertial and pressure sensors hold the limitation that the user is directly contacting a controller. Alternatively, camera systems allow playing games without holding or wearing input devices, such as the gesture recognition system Eye Toy developed for Sony PlayStation II (Sony Computer Entertainment, Foster City, CA, US) [76, 77]. The disadvantage of this system is that the camera system does not provide the accuracy necessary for playing faster games or taking high-resolution measurements. Commercially available webcams are also used to control exergames [78]. Kinect motion-detecting camera system from X-Box360 (Microsoft Corporation, One Microsoft Way, Redmond, WA, US) captures depth and color information and generates a point cloud of colored dots. The software is able to calculate the 3D position of the dots, thereby creating a 3D image of the environment [79]. Kinect motion detection camera has also been used in neurorehabilitation [80–82]. Some studies use custom-designed exergames using GestureTek's Interactive Rehabilitation and Exercise System (IREX®) (GestureTek Health, Toronto, Ontario, Canada) [83], the Balance Rehabilitation Unit (BRU™, Medicaa™, Montevideo, Uruguay) [84], and different pressure mats or force platforms.

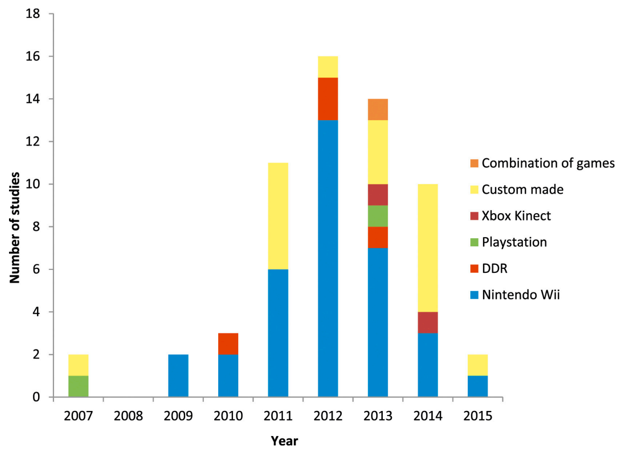


Figure 5: Distribution of game consoles in exergames studies until 2015. Graphic taken from [85].

Nintendo Wii has a significant interest in the neurorehabilitation research domain. The Wii Fit software and WBB have provided an increasingly attractive way of assessing and training individual balance ability, particularly in balance control studies [68, 86, 87]. Undoubtedly, The use of the Nintendo Wii console and Wii Fit software in balance control rehabilitation related research has been very popular and with great acceptance by the scientific community [68]. From the graph reproduced from [85] (Figure 5), the prevalence of the Nintendo Wii console is noticeable. In 2012 the HMD Oculus was launched, which may be the cause of the decrease in studies conducted with this console.

2.2.2 Outcome Measures

There are two types of outcome measures to assess the patient's progression through the rehabilitation program. The internal measures during the game and the external measures evaluated after the game. Intra-game, the measures are objective and measured using instrumentation devices, such as force plates, which have the advantage of providing feedback to the user while playing [88]. Outside the game, external measures can be clinical assessments to assess balance or sway variability after the intervention period. Clinical balance and mobility tests like the Berg Balance Scale (BBS) [89] and Timed Up and Go (TUG) [90] are abundantly used to quantify the effect of an exergame intervention on postural control and are considered external measures [75, 84, 87, 91]. In addition to universal clinical tests, some disease-related specific tests are made. To assess the disability in PD, most authors use the Unified Parkinson's Disease Rating Scale (UPDRS) [92, 93]. To examine trunk motor deficit of stroke patients, use the Trunk Impairment Scale (TIS) [94]. The external balance measures based on sensor data quantify sway variability and CoP displacement in ML and AP directions during quiet stance using force plates [95]. Internal outcome measures include, for example, the percentage of missed targets, and the total movement range of CoP in ML and AP directions, all measured using pressure mats [75]. The CoP controls the game in several exergame studies but these measurements are not considered internal or external outcome measures as they do not quantify balance ability but only are used to play the game [83, 96]. Regarding assessment schedule, outcome measurements were typically performed at baseline and immediately after or shortly after the intervention. Some adopt follow-up measures [67].

2.2.3 Intervention Protocol

The intervention standard protocol normalizes the structure of the virtual rehabilitation intervention, namely: i) number of sessions; ii) session content; iii) targeted population and iv) evaluation sessions. According to Burdea et al. [97], therapy in the inpatient or outpatient clinic is still preferred, as it presents a more structured environment free of home distractions. However, this forces the patient to travel to the clinical site, which is sometimes problematic. Training duration must be distinguished from session duration, including baseline assessments, equipment setup, and rest periods. Typically, the structure of a VR rehabilitation session is fixed for the first week, and then sessions get progressively longer and game harder over the remaining weeks. The games and how many times patients play can be fixed or selected during each session [97]. In Viñas-Diz and Sobrido-Prieto systematic review [98] about non-immersive VR rehabilitation on post-stroke patients, most of the studies applied the same treatment intensity in all groups, with some exceptions, in which the study group received more intensive treatment i.e., a more significant number of sessions and a longer session duration. VR rehabilitation programs analysis revealed a large discrepancy in the planned volume of exercises. The number of sessions fluctuated between 12 and 40 in stroke patients, 8 and 18 in PD, and 9 to 36 in CP. The duration of a single session ranged from 20 to 90 minutes in stoke, 20 to 50 minutes in PD, and 30 to 50 minutes in CP [99]. This review [99] concludes that a gold standard for improving rehabilitation in patients with stroke or PD or CP cannot be determined because of the immense diversity of implemented VR training in the inspected investigations. Furthermore, regarding the methodological quality of experimental protocol, which assesses whether a trial is subject to systematic errors (bias), it was found by several authors, using scales such as the PEDro score [100], that most studies had several methodological weaknesses [85, 99, 101].

2.3 Immersive Virtual Reality

Immersive Virtual Reality aims to completely immerse the user inside the computer-generated world, giving the impression to the user that it has stepped inside the virtual world. In Pimentel and Teixeira book [102], they inquired if for a virtual world to be considered immersive, whether it is honest enough to suspend the participant's disbelief for a while. It means that an immersive experience does not require the virtual world to be as natural as the physical one to achieve a sense of presence. Compared with VR on a traditional computer screen or tablet, immersive virtual reality provides an enhanced sense of immersion in the virtual world. Immersive headsets, or those that incorporate motion-tracked stereoscopic HMDs, and potentially motion controllers are known to induce a stronger sense of presence and potentially a sense of realism, embodiment, memory, and spatial understanding than non-immersive devices [20, 103]. A perceived embodiment can influence an individual's sense of agency, where a greater embodiment may result in increased control. A lack of embodiment may result in the sense of decreased control or distress and lead to a distortion of capabilities [104]. The above makes immersive virtual reality especially applicable in motor rehabilitation. A recent scoping review [105] identified the clinical populations targeted in physical rehabilitation using immersive VR. Patients' groups included stroke, SCI, disorders of the vestibular system, impaired balance, Cervical Range of Motion (ROM) impairments, PD, and congenital limb deficiency. The most common use of immersive VR rehabilitation was for upper extremity movement rehabilitation, vestibular training, and gait and balance rehabilitation. Fall prevention appears to be an area where immersive VR using HMD can significantly affect. In general terms, the patients who can benefit from immersive VR are the same ones who use non-immersive VR. The difference lies in the level of presence which may have better effects on rehabilitation at the level of motor learning [104, 106].

2.3.1 Equipments

Two immersive virtual reality systems are significantly associated with neurorehabilitation investigation: head-mounted displays and the CAVE system. The Cave Automated Virtual Environment, or CAVE, is a room-based, fully immersive virtual reality system. It was invented in 1992 by researchers at the University of Illinois's Electronic Visualization Lab [107]. The side walls are rear-projection screens, whereas the floor is a down-projection screen. Modern CAVE systems can also project scenes to the ceiling, creating a six-wall CAVE [108]. A CAVE user's movements are tracked by the sensors typically attached to the 3D glasses, and the video continually adjusts to retain the viewer's perspective [107]. Researchers acknowledge that the CAVE system is one of the most important systems of the evolution of virtual reality [109].

(a) CAVE at EVL, University of Illinois at Chicago. (b) A recent CAVE commercial system [110]

Figure 6: CAVE System

The head-mounted display (HMD) or headset is the other device capable of making a virtual reality experience fully immersive. When the history of virtual reality was explored in the introductory section, some commercially available HMD devices were introduced, which are widely used for studies in neurorehabilitation [105]. The technology-mediated simulation of enriched ecologically valid environments [111, 112] and controlled sensory stimulation is especially interesting for serious games, rehabilitation, and health-related applications [113]. Concurrent with renewed interest in VR, many systems were presented in the past years. As described by the manufacturers, these new devices provide improved features, such as higher resolution, faster refresh rate, and a wider field of view, at dramatically lower costs than early systems [114]. In addition, new HMDs include built-in tracking mechanisms to estimate the position and orientation of the head in real-time, which enables perspective correction and allows users to freely move, at least to some extent, as in the real world in a room-size environment [115]. From the clinical point of view, the possibility of moving and walking as in the real world could maximize the ecological validity of virtual reality environments and tasks requiring users to move, such as walking and navigating simulations or exergaming, and serious games for motor rehabilitation [105]. A study conducted by Adrián Borrego and colleagues [116] compare the last generation of head-mounted displays, namely the Oculus Rift (Oculus VR, Irvine, CA) and the HTC Vive (HTC Corporation, Taoyuan City, Taiwan) in terms of the operating range of the head tracking and the working area, accuracy, and jitter in a room-size environment. Following their conclusions, relevant technical information can be extracted. Their manufacturers have provided the characteristics of both headsets (Table 3).

Table 3: Comparison of the specifications of Oculus Rift and HTC Vive, retrieved from [116]

	Oculus Rift	HTC Vive
Display	OLED	OLED
Display size	90 mm x 2456 ppi	91.9 mm x 2447 ppi
Resolution	2160 x 1200 pixels	2160 x 1200 pixels
Refresh rate	90 Hz	90 Hz
Field of view		
Horizontal	94°	110°
Vertical	93°	113°
Lens type	Fresnel	Fresnel
Sensors	Accelerometer, gyroscope, magnetometer	Accelerometer, gyroscope
Integrated camera	No	Yes
Audio	Microphone, integrated supra-aural 3D spatial audio headphones (removable)	Microphone, jack for external headphones
Weight	470g	563g

Concerning tracking, the Oculus Rift uses a microphone-shaped infrared camera designed to be fixed on a conventional table and tilted toward the HMD. This infrared camera locates a constellation of infrared LEDs embedded in the headset but not directly visible. The HTC Vive, in contrast, uses two base stations designed to be fixed on a wall above head height, ideally more than 2 meters. Each base station must be placed facing the other at a maximum of 5m and tilted toward the HMD [116]. The base stations include two stationary LED arrays that flash at 60 Hz and two active spinning laser emitters that throw a light beam after each flash. The headset has a constellation of photodiodes that estimate the series of flashes. The headset position/orientation can be estimated from the delay between the flash emitted by the LED array and that emitted by the spinning emitter in each photodiode [116].

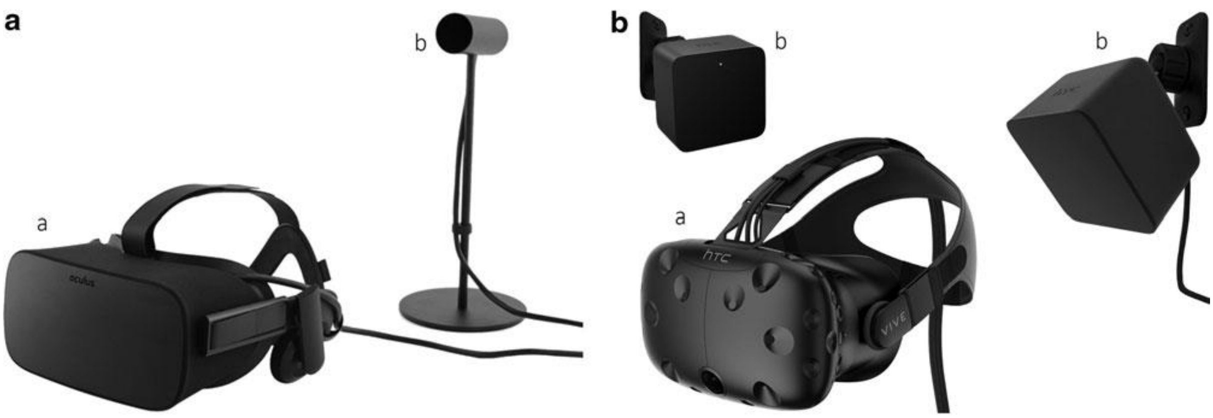


Figure 7: (a) Oculus Rift and (b) the HTC Vive, were analyzed in this study [116]. Both devices consisted of a display (a) and a tracking module (b).

The experimental results from [116] showed a working area of $11.75m^2$ for the Oculus Rift and

24.87m^2 for the HTC Vive at all heights, which significantly exceeded the maximum recommended working areas of 2.75 and 6.25m^2 for both devices, respectively. Regarding accuracy and jitter, better results on both parameters were detected inside the recommended working area for both devices. Similar values were obtained for accuracy and jitter at both heights for both devices, except for the HTC Vive's accuracy, which was better at lower heights. Both devices showed comparable performance at sitting height, both within and beyond the recommended working area. However, the HTC Vive presented worse accuracy and jitter at standing height. The operating range of the Oculus Rift's tracking system was 4.25m at both heights, starting at 0.25m away from the camera. The HTC Vive presented higher working ranges at both heights, up to 6.75m at a sitting height and 7m at a standing height. The experimental results for working area, accuracy, and jitter of the latest-generation HMDs' tracking support the use of these devices not only as a feasible alternative for VR based health applications, such as serious games, exergames, and motor rehabilitation systems, but also to enable the navigation and exploration of real-life size virtual environments, without any remarkable adverse effects [116].

2.3.1.1 Virtual Feedback

When using devices like HMDs, the internal measures are often given as real-time feedback [117]. Therefore, in this segment several types of feedback common to the practice of neurorehabilitation with immersive virtual reality are discussed. In immersive VR rehabilitation, visual feedback was used to interact with children with CP, creating different scenarios with different spectra of difficulties [118]. This type of visual feedback is also used in balance and postural training [119, 120]. VR can create conflict between information from the somatosensory, vestibular, and visual systems, forcing the patient to train the different sensory systems. As the training progresses, this visual feedback can gradually increase in speed and complexity to disturb static and dynamic postural control [121]. A significant development in virtual environments was incorporating tactile information and interaction forces into what was previously a visual experience [122]. Robots of varying complexity are being interfaced with more traditional VE presentations to provide haptic feedback that enriches the sensory experience and adds physical task parameters [123]. Biofeedback was also used in rehabilitation after injury [124]. It is the technique of providing biological information to patients in real-time. This information can sometimes be called augmented or extrinsic feedback instead of the sensory (or intrinsic) feedback that provides self-generated information to the user from various intrinsic sensory receptors [125].

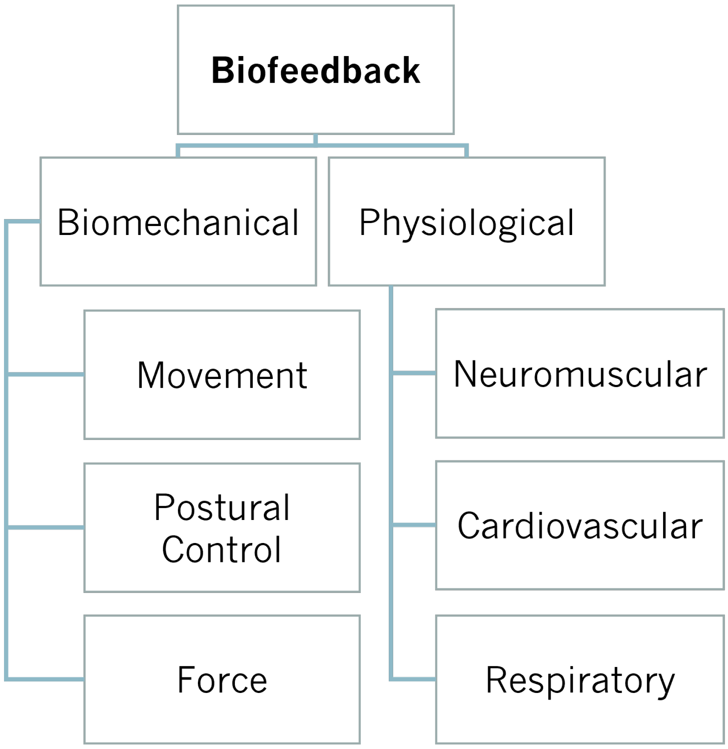


Figure 8: Categories of biofeedback used in physical rehabilitation, reproduced from [124]

Biofeedback in physical rehabilitation with virtual reality is classified into two main categories (Figure 8). Physiological and biomechanical biofeedback [124]. Electromyography biofeedback is the most widespread form. It is a method of retraining muscle by creating new feedback systems resulting from converting muscular signals into visual and auditory cues [126, 127].EMG uses surface electrodes to detect skeletal muscle activity changes, then fed back to the user. Biofeedback is most commonly delivered using visual, auditory, or haptic signals [124, 128]. However, recent years have witnessed the emergence of immersive, VR biofeedback signals. The existing evidence suggests a role for VR-based biofeedback in rehabilitation [128].

2.3.2 Outcome Measures

As discussed in the outcome measures section of non-immersive virtual reality, there are real-time calculated metrics to provide feedback to the participant and metrics to evaluate the intervention. This section focuses on the most commonly used outcome measures to evaluate the effectiveness of immersive virtual reality rehabilitation intervention. Dermody and colleagues [129] conducted a review including studies that evaluated the use of VR applications delivered via immersive headsets. Commercial headsets including Oculus Rift or HTC Vive and older models encompassing the field of view and motion tracking were included. This review considered the following outcomes: effectiveness of the VR intervention via immersive headset using physical, mental, or psychosocial outcome measures. These outcomes were measured using kinematic and kinetic computer applications or statistical software. Immersive VR therapy was also implemented to improve older adults' cognition, memory, and psychological aspects. Several outcomes were measured, including general cognitive abilities, verbal memory, executive functions, and visuospatial processing. Several studies focused on falls regarding the physical health outcomes targeted in the interventions. Some evaluated the use of VR training to improve postural and muscular adjustment to enable the subject to compensate for induced perturbation - balance-recovery, the goal of fall prevention [130].

2.3.3 Intervention Protocol

The frequency and duration of immersive VR exposure, similar to what was found for non-immersive VR, varied considerably between studies [129]. VR exposure ranged from five 10-second displays of stereoscopic 3D images to 55 minutes of serious gaming, with 15 minutes being the most common exposure time [129]. VR therapy was administered as a solo session in some studies [130–132]. Other studies completed VR therapy more regularly, with Levy et al. [133] administering it weekly over twelve weeks. Another study completed it twice a week for six weeks [134], and another completed it three times per week over seven weeks [135].

2.4 Limitations and Future Directions

A division was made between non-immersive and immersive virtual reality. After identifying the main equipment used in each of the categories, the outcome measures most used in the evaluation of neurorehabilitation intervention, and the intervention protocols, this section points out the evident limitations in VR neurorehabilitation and suggestions for future directions. This section refers to both types of virtual reality, since it is important to understand the overall implications for neurorehabilitation and especially for the rehabilitation of motor tasks and balance control.

2.4.1 Limitations

Rehabilitation with VR is a tool that brings many benefits in the design of therapy that targets neuroplastic mechanisms of the nervous system. However, it has developed more slowly than other technologies in other areas of health. Some weaknesses are pointed out:

- A great deal of specific knowledge between hardware and software is required to develop these VR systems, which is expensive.
- The multidisciplinary nature of neurological rehabilitation with VR brings together neuroscience, biomedical engineering, computer science, and orthopedics knowledge, presenting challenges.

- The high risk of bias in these studies, often geared toward feasibility or proof of concepts, lacks scientific rigor to prove their positive effect.
- The absence of randomization and blinding during the outcome assessment, together with very diverse intensities and training durations, in small sample size groups, often without a control group, can overestimate the effects.

Researchers must have an excellent acquaintance with the mechanisms inherent to motor learning to overcome these setbacks. Thus, they find the best exergame or the best therapy routine, concerning intensity and duration, appropriate to the characteristics of each clinical group. As noted, a commercial non-immersive virtual reality device can improve aspects of walking and balance. Identifying which type of patients would benefit most from VR therapy is still necessary. Skjæret [85] detects some flaws and determines some points that this rehabilitation investigation has to establish. Exergames should be customized for each patient according to their objectives and performance level. For that, inclusive designs must be created, listening to patients' goals, needs, and motivation: user-centered design. This would increase engagement with the exergame training. The progression should be automatic regarding the game, increasing difficulty and load when performance improves. On the contrary, it decreases the difficulty when performance worsens [85]. Adherence must be included in the study's methods so that there is scientific evidence that VRR programs have more adherence than traditional forms of exercise. More game data can also be collected to translate into more significant insight into exercise routines than traditional exercise logs. This understanding opens up the opportunity to give personalized feedback to each player/patient. Finally, in line with other authors, studies with a longer follow-up time should be conducted to have long-term effects. Include larger intervention groups, more unified follow-up assessments, protocols and standardized outcome measures. Most of the studies described in Saldana and colleagues' scoping review [105] provided low methodological quality (e.g., no randomization of participants into groups), limiting the findings' impact. The average sample size of included studies was 23.75 ± 17.6. In addition, study protocols varied widely in terms of treatment dosage, length, and incorporating HMD-VR methods. Although some studies under review [129] reported positive VR outcomes in older adults, it is important to highlight several practical challenges. Older adults are often vulnerable. So, it is necessary to translate VR interventions as a standard of care in the aged care sector. Investigations into using VR interventions with older adults who are frail or unwell are needed. In most of the studies, it was questionable if the researchers used a participatory approach by inviting older adults to contribute to the design and conduct of the intervention. No study designed or incorporated gaming technology in VR to support engagement and promote enjoyment for participants, contrary to what was found to be common practice in rehabilitation with non-immersive VR. Longitudinal studies should be conducted to determine the long-term effect on the outcome variable under study. All studies in this review [129] included a small sample size, which impacted the generalizability of the findings. Powered Randomized Controlled Trials are needed to determine the effectiveness of VR therapy compared with standard care practices. Scholars should increase the public availability of their VR simulations, questionnaires for measuring VR outcomes, and the datasets with accompanying code that informed their central conclusions. As mentioned in the motivation section, where the serious problem of falls is exposed, the existing systems to counteract this scourge focus mainly on fall detection, with little emphasis on predicting and preventing fall events. For this reason, it is important to mention that in addition to research efforts to promote balance training, it is also important to implement fall prevention strategies that capture the multifactorial nature of falling in order to reliably estimate the risk of falling. This strategy should be built on data acquired from various scenarios surrounding fall-related events. This information has been collected through questionnaires, fall diaries, and telephone calls. This information should be augmented with data collected from various sensors to improve the reliability and accurracy of fall detection and prediction systems [10]. The unpredictable nature of a fall puts barriers in the collection of data regarding these events. In addition to occurring infrequently, it is necessary for the subject to be instrumented at the time of the fall. The goal would be to collect data regarding real-world falls and build datasets with this information to complement fall prevention strategies. However, the strategy used to circumvent these barriers is to simulate falls in a controlled environment [136]. These simulations are not representative of a real-world fall [14]. Therefore, it is necessary to collect data from several sensors that best approximate a real-world fall [12].

2.4.2 Future Directions

Future studies could explore the use of HMD-VR applications in clinical settings and home-based training and as a component of telerehabilitation. Notably, although many of the earlier studies used expensive HMD devices that are no longer commercially available, some of the most recent studies used commercially available and low-cost devices. For example, Google Cardboard (Google, Mountain View, CA) retails online for $15 USD, or the Oculus Rift, which retails online for several hundred dollars. The accessibility of these devices should help make future studies using HMD-VR with larger populations more feasible [105]. HMD-VR may enable some patients to experience greater enjoyment and motivation during therapy with few risks or side effects, suggesting that HMD-VR can be a low-risk, fun adjunct to therapy. To further substantiate VR as a useful tool in rehabilitation, one reinforces that higher-quality studies with larger sample sizes are needed.

2.5 Conclusions

In conclusion, immersive HMD-VR is a commercially available, relatively inexpensive tool that can be used in rehabilitation with adult patients. It can also be used appropriately in assessment and intervention for patients with various diagnoses. A meta-analysis of virtual reality rehabilitation programs carried out by Howard [121] answers the following question: ”Are Virtual Reality Rehabilitation programs effective?”. Comparing groups of patients that completed a VRR program with control groups receiving conventional rehabilitation therapy without VR, this author support that, in most cases, VRR is effective. Analyzing all outcomes, the patients in VRR programs improved their physical abilities 0.397 standard deviations above those who did not receive any VRR training. VRR programs that develop gait abilities demonstrated the most remarkable effects. Those programs developed to improve strength showed a significant effect beyond comparison groups. Conversely, motor control VRR programs and those designed to improve balance demonstrated the slightest effects of all. To conclude this meta-analysis, Howard reveals that VRR programs are more effective than comparable rehabilitation programs. VRR programs are apt for developing strength and gait. Similarly, VRR programs were more effective than alternatives for developing motor control and balance, despite having less pronounced effects. The scientific reasons that lead to the efficiency of these programs are essentially related to the concepts of repetition or mass practice, feedback, and motivation. The neurophysiological and functional benefits of movement observation, imagery, repetitive mass practice, and imitation therapies that facilitate voluntary production of movement can be easily incorporated in VR, allowing the clinician to use sensory stimulation targeting specific brain networks like motor areas, critical for neural and functional recovery, promoting neuroplastic changes early in the recovery phase [137]. There are several critical factors to learn or relearn motor skills: quantity, duration, and the intensity of training sessions. The repetition itself does not build motor learning. The repeated practice must be linked to incremental success at some task or goal. In the typical nervous system, this is achieved by trial-and-error practice, with feedback about performance success provided by the senses (e.g., vision, proprioception). Nevertheless, to practice movements over and over, participants must be motivated. VR provides a powerful tool endowed with all these elements – the possibility of repetitive practice, feedback about performance, and motivation to endure practice. In particular, in a virtual environment, the feedback about performance can be augmented and enhanced relative to feedback that would occur in real-world motor skills practice. Real-time feedback immediately after a trial improves the learning rate [138]. Proponents of VR believe that outcomes will be enhanced following practice in VR because of making tasks more accessible, less dangerous, more customized, more fun, and easier to learn because of the salient feedback provided during practice. Some studies examined this real-world versus VR practice issue in a controlled fashion [139–141] providing some experimental evidence that motor learning in a virtual environment may be superior.

There is evidence that the proprioceptive and exteroceptive feedback associated with the execution of skilled tasks induces profound cortical and subcortical changes at the cellular and synaptic levels. Several studies provide neurophysiological evidence that motor repetition alone is not enough to induce cortical correlates of motor learning. VR also allows programming to display a virtual teacher who repeatedly performs the task, enhancing ”learning by imitation”via mirror neuron inputs. VR offers the unique capability for real-time feedback to the participant during practice in a very intuitive and interpretable form. In contrast, many potential distracters exist in real-world situations and may slow down learning. Even if it proves to be the case that VR offers no performance advantage over real-world practice, it is still a powerful new tool that can be used to test different methods of motor training, types of feedback provided, and different practice schedules for comparative effectiveness improving motor function in patients. The technology provides a convenient mechanism for manipulating these factors, setting up automatic training schedules, training, testing, and recording participants' motor responses [142]. Howard also answered the question, "why are VRR programs effective?"He pointed out three main mechanisms suggested as causes of success in this type of rehabilitation program's outcomes. The first cause is attributed to increased excitement. VR has been suggested to remove monotony and boredom from conventional therapy. Interacting with VR, whether they are exploring a new world or in a family environment, patients find it fun and exciting. Adding the competitive and challenging component of an exergame can increase the excitement. It leads to an increase in motivation to complete the tasks requested by the therapist. In addition, there is increased physical fidelity because the patient performs tasks closer to reality than abstract exercises. In gait rehabilitation, patients will walk through virtual environments instead of repeating movements with the knee or heel. The last mechanism pointed out by Howard [121] as a cause of the effectiveness of VRR programs is the increase in cognitive fidelity. In order to cognitively stimulate patients while undergoing treatment, these programs try to get as close to real-life as possible, whether by asking the patient to have a conversation while performing the exercise or solving simple math problems. This component is essential for the transfer of training to the real world to be more consistent. Thus, patients are better prepared for the distractions they face outside the clinical setting. Schulteis and Rizzo [143] suggested several additional advantages for using VR in cognitive rehabilitation, including control and consistency in the delivery of stimulation, immediate feedback through different senses, ability to intervene during practice in order to provide further instruction or guidance, the opportunity for self-training and learning in a safe environment, documentation of the patient's performance and the ability to create customized training environments at low cost.

Immersive VR allows the creation of individualized environments, while in non-immersive VR, most exergames follow the one size fits all paradigm. Using an HMD based therapy, training a task can be conducted in a simulated version of the individual's home or vocational setting. Additionally, HMD and CAVE system allows the user to receive immediate feedback from its responses, behaviors, or physiological body reactions. This supports the notion that rehabilitation using fully immersive virtual environments can improve training tasks' ecological validity and generalizability [143]. Several advantages that immersive VR use can offer to rehabilitation are directly relevant to rehabilitation research, assessment, and treatment. Some of these assets could exist with previously developed tools (non-immersive VR). However, VR goes beyond simply automating the past paradigms through its dynamic interactive and immersive three-dimensional features. One key benefit is VR's more naturalistic or real-life environment. This advantage is twofold. First, the experience of being immersed within a VE allows users to forget that they are in a testing situation which may subsequently allow for assessment of behaviors under more natural conditions and provide insight into individuals' typical behavior. These characteristic features contrast with the presentation of more artificially laden test-taking behaviors that may influence results obtained in traditional testing environments. Second, with the application of more real-life VR scenarios, it can be answered the question regarding the ecological validity of current assessments and interventions. Complete control over stimulus presentation and response measurement, according to the specifications of the clinician or researcher, is another strategic advantage of using immersive VR technology. Increased generalization of learning is another potential advantage. Better generalization of learning occurs with an increased similarity between training tasks and criterion targets.

Chapter 3

Balance Perturbations and Compensatory Postural Adjustments induced by Immersive Virtual Reality

The present review aims to find studies that use immersive VR through a HMD to introduce sensory disturbances in subjects, triggering dynamic or static anticipatory and compensatory postural behaviors. The reviewed articles must report postural adjustments with spatial-temporal parameters of gait or standing, either kinematic or electrophysiological. These parameters are indicators of the effectiveness of the disturbance inducing compensatory responses to maintain balance. This effectiveness will be the target of the systematic review, to investigate the possibility of applying sensory stimuli via HMD to provoke imbalances or real-world falls. As mentioned in Chapter 2, scientific research in the field of motor rehabilitation and the study of human gait and balance has undergone an intervention of VR. However, most studies use non-immersive virtual environments, the so-called exergames, and another large part uses systems that consider themselves immersive (e.g., CAVE system). Currently, the devices that offer greater capacity for immersion and presence in the virtual environment are HMDs. Due to their increasing economic accessibility, experiences with these devices have grown, and therapies are even being implemented in rehabilitation clinics using these devices. All existing reviews that focus on virtual reality via HMD are specific to a population of patients, as to their application. In this chapter, a synthesis of all studies that introduce visual perturbations using an HMD is made.

3.1 Introduction

As age increases, so does the frailty and the fall incidents worsened mobility or ADL disability [144]. There is an age-related decline in sensory systems and reduced ability to adapt to changes in their environment to maintain balance. Degeneration can be exacerbated by vestibular impairments and neurologic diseases [145]. Such a statement only reinforces the necessity of developing and applying effective VR balance training programs that target control of posture and balance through efficaciously integration of sensory information. For fall prevention, this integration requires rapid recalibration of visual, vestibular and somatosensory information [146]. The concept of virtual reality is built on the natural combination of two words: the virtual and the real. It refers to a range of computing technologies that present artificially generated sensory information in a form that people perceive as similar to real-world objects or events [147]. Can be defined as an approach to user-computer interface that involves real-time simulation of an environment that allows for user interaction via multiple sensory channels [148]. Similarly, Schultheis and Rizzo [143] considered virtual reality to be an advanced form of human–computer interface that allows the user to interact with and become immersed in a computer-generated environment in a naturalistic fashion [143]. Regardless of its definition specificities, the richly complex multisensorial experience offers a practice environment that can be ecologically valid [149], can elicit a substantial feeling of realness and agency if the concepts of immersion, place illusion, plausibility illusion and the fusion of these last two in the notion of a virtual body are archived. As it uses visual, sensory, and auditory feedback, there is extensive evidence that the proprioceptive and exteroceptive feedback associated with the execution of skilled tasks induces profound cortical changes associated with motor learning and that humans can transfer motor learning to a real-world environment [142]. In addition, gives an opportunity to increase the duration, intensity and mass practice needed to induce neuroplasticity, as it also strongly increases the participant's motivation [99]. For these reasons, exploring interventions that use VR may offer unique opportunities to address areas of health need. VR is evolving at a rapid rate and presents an opportunity to enhance and support older adults' physical and cognitive issues [129]. The variety of technologies that are considered VR range from non-immersive applications to completely immersive applications. Immersive VR can be produced by combining computers, HMDs, body tracking sensors, specialized interface devices, and real-time graphics to immerse a participant in a computer-generated simulated world that changes in a natural way with head and body motion [150]. VR can be used in several ways to re-train postural control and balance. First, VR can be used to manipulate visual feedback to produce conflicts between visual, somatosensory, and vestibular information as a way to train different sensory systems. Second, VR feedback can be systematically graded (in terms of speed and complexity) to challenge a person's static and dynamic postural control over the course of sensorimotor training [146]. Recently, VR therapies have been introduced in the field of neurorehabilitation, especially in patients who suffered from stroke or Parkinson's disease and in children with cerebral palsy. Previous systematic reviews have evaluated VR for balance training in patients with stroke [151, 152]. There are also reviews assessing the general effectiveness of VR-based rehabilitation for Parkinson's disease patients [92] and in patients with cerebral palsy [153, 154]. Perturbation-based balance training (PBT) is a balance training intervention that incorporates exposure to repeated postural perturbations to evoke rapid balance reactions, enabling the individual to improve control of these reactions with repeated practice [155]. The above-mentioned studies follow this paradigm, adding the perturbation on the VE. The objective of this study is to determine the evidence of visual perturbations induction of compensatory postural adjustments, and to determine VR technology viability preventing fall frequency. With this purpose, VR interventions in recent studies will be analyzed with regard to equipment's used, features extracted from measurement systems, VR protocols comprehending the perturbation type, their amplitude or frequency and study population. Finally, a discussion section is reserved for the results found to answer two research questions: RQ1 "Can a virtual reality headset introduce imbalances through visual perturbations? Can they cause postural reactions typical of a fall?"and RQ2 "Which visual perturbation challenged the participants' balance the most?". Addressing these questions will provide a foundation to understand the impact of visual perturbations delivered by HMD-VR on balance control system.

3.2 Materials and Methods

3.2.1 Search Strategy

A literature search was carried out using the boolean search strategy in the IEEEXplore, SCOPUS, Web Of Science, and PubMed databases. Combinations of the following key terms were used for each database: "virtual reality" OR "virtual environment" OR "immersive" AND perturbation. No filters were applied in each database to restrict searches. The studies were chosen under the following inclusion criteria: (1) the study was conducted using a HMD, (2) sensory disturbances were delivered to the participants (3) physiological, neuromuscular, or kinematic data were reported to study participant's reaction to perturbations. Following this inclusion criteria, a PRISMA flowchart is shown in Figure 9.

3.3 Results

3.3.1 Search Results

The initial search resulted in a total collection of 1101 publications. A selection was made according to PRISMA, detailed in the flowchart of Figure 9. From this total, 386 publications were imported from Scopus, 130 from IEEE, 386 from Web of Science, and 199 from PubMed. 144 were identified as duplicates and 855 publications were excluded by title or abstract that did not meet the inclusion criteria. Of the 102 complete articles or conference papers that remained, 51 were excluded for using VR that they consider immersive but that does not use HMDs. 42 publications underwent a full reading, a process that excluded 6 articles for not evaluating the subjects' physiological, kinetic or kinematic parameters after disturbances. In this full reading process, 2 articles were included that fulfilled the inclusion requirements, by references. Altogether, 40 articles were included in the subsequent analysis. Figure 9 presents the flowchart for the

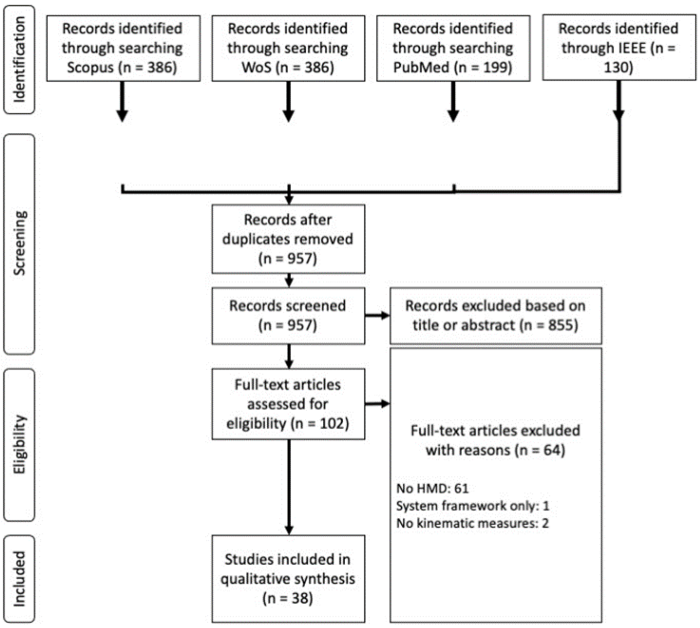


Figure 9: PRISMA flowchart

literature search process. A summary of the included studies is presented in Table 4. This summary presents the following characteristics: i) study author and year of publication; ii) study goal; iii) virtual reality system used and iv) sensors.

Table 4: Summary of the main characteristics of the included studies: reference, study objective, headset and sensors used.

Reference	Study goal	VR-HMD	Sensors
[156]	Age-related postural reactions	Kaiser Optics ProView XL5	Force-plate; Optical motion tracking
[157]		HTC Vive	Instrumented walkway
[158, 159]		Nvisor	Optical motion tracking
[160]		Glasstron LDI–100B, Sony	Optical motion tracking
[131, 161]	Patients postural reactions	Oculus Rift	IMU
[162]		Samsung Gear VR	Force-plate
[163]		Oculus Rift	Force-plate
[164]		Oculus Rift	Force-plate
[165]	Visual and Proprioceptive conflict	Oculus Rift	Optical motion tracking
[166]		Oculus Rift	Optical motion tracking; EMG
[167]		PlayStation VR	Optical motion tracking
[168]	Balance function and threatening scenarios	HTC Vive	EKG; EMG; EEG
[169]		Sensics piSight	Force-plate; EMG
[170]		HTC Vive	IMU
[171]		HTC Vive	Force-plate
[172]		iWear VR920	Force-plate; EMG; Accelerometer
[173]	Visual Perturbation postural reactions	HTC Vive	Instrumented walkway
[174]		Google Cardboard	IMU
[175]		HTC Vive	Optical motion tracking
[176]		HTC Vive	Optical motion tracking
[177]		HTC Vive Pro	Force-plate
[178]		HTC Vive	Force-plate; Optical motion tracking; EMG
[179]		Oculus Rift	Force-plate
[180]		HTC Vive	Instrumented walkway
[181]		Oculus Rift	Force-plate
[182]		Oculus Rift	Force-plate
[183]		Oculus Rift DK2	Optical motion tracking; EMG; EEG
[184]		Oculus Rift DK2	Force-plate
[185]	Balance Training	Oculus Rift DK2	Optical motion tracking; EMG; EEG; GSR
[186]		Glasstron LDI–100B Sony	Instrumented treadmill; Optical motion tracking; EMG
[132]		Glasstron LDI–100B Sony	Instrumented treadmill; Optical motion tracking
[130]		Glasstron LDI–100B Sony	Instrumented treadmill; Optical motion tracking; EMG
[187]	Balance Assessment	Oculus Rift DK2	Force-plate
[188]		Samsung Gear VR	Force-plate; EMG
[189]		Virtual Research V8	Force-plate

3.3.2 Study goals

The diagram in Figure 10 shows the broader separation of the studies' purposes. All aim to investigate compensatory reactions induced by visual perturbations. Some channel this analysis of postural adjustments to understand differences introduced by advancing age or by diseases. Others analyze these postural reactions when visual information discordant with proprioceptive information is introduced, when placing the participant in threatening situations, or to perform an objective balance assessment, or balance training. The objectives of the studies are mostly divided into two aspects: gait analysis or standing postural analysis (Figure 11). Both can study anticipatory and compensatory reactions to visual disturbances introduced by an optical flow in the VE. More specifically, 18 studies attempt to understand changes in gait patterns, 8

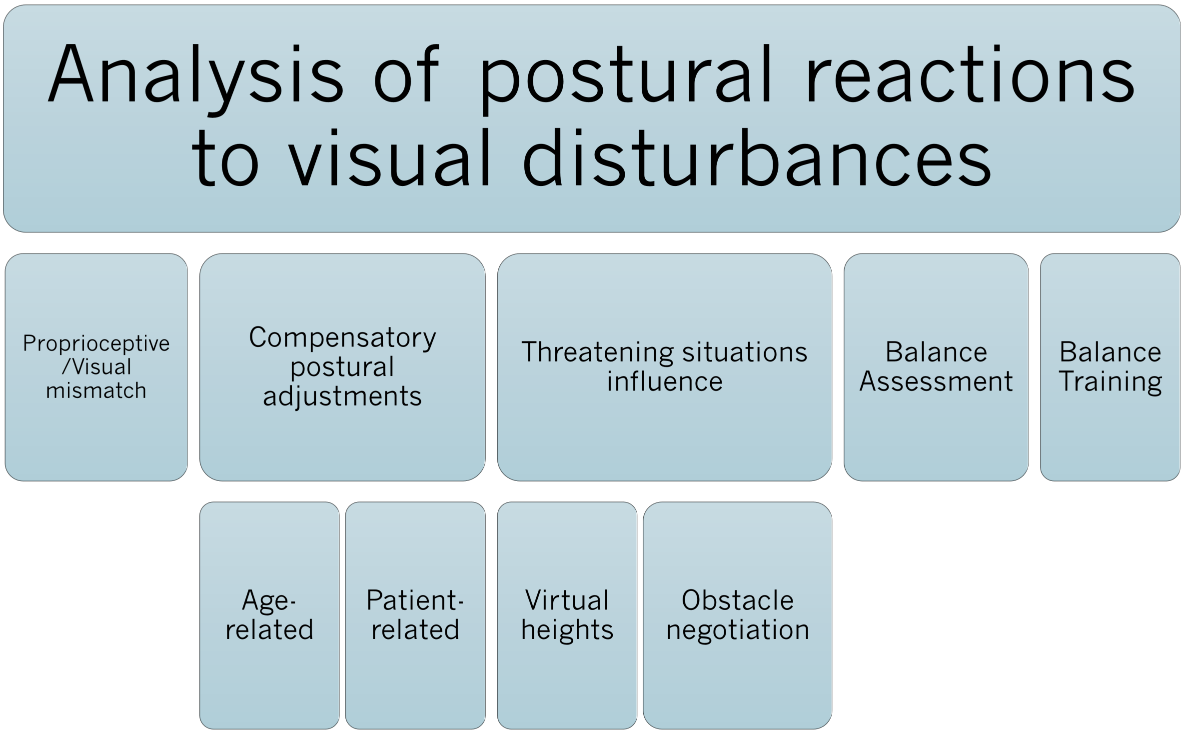


Figure 10: Diagram describing the separation of study goals.

of which during overground walking [157–159, 167, 170, 173, 174, 180] and 10 during treadmill walking [130, 132, 160, 165, 166, 175, 176, 183, 185, 186] while the remaining studies analyze postural reactions in upright stance to visual perturbation [131, 156, 161, 163, 168, 169, 171, 172, 177–179, 181, 182, 184, 187–193], either in bipedal stance or one leg stance. Whether walking or standing upright, the analysis is carried out without exception when the participant is exposed to visual or physical disturbances of different types, namely: i) continuous and multidirectional; ii) transient; iii) discrete; iv) RPY axes rotations (single or multi-axes); and v) dissonant or concordant proprioceptive and visual information. The following divisions do not distinguish between disturbed gait analysis or standing balance reactions analysis, since it is intended to present the purposes of the studies.

3.3.2.1 Discordant proprioceptive and visual feedback

Frost [165] and Drolet [166] focus on the interplay of visual feedback and proprioception. There is a crucial relationship between proprioception and visual feedback. While previous studies have separately investigated proprioceptive and visual feedback in gait, they have not shown the evoked muscle responses from visual feedback alterations. Understanding and modeling this relationship could lead to patient-specific interventions for motor rehabilitation. Frost and colleagues [165] quantify the functional role of expected vs. actual proprioceptive feedback for planning and regulation of gait measuring contralateral leg kinematics whereas Drolet and colleagues [166], through the manipulated visual and proprioceptive feedback, focus on the muscle activation patterns before, during, and after surface changes that are both visually informed

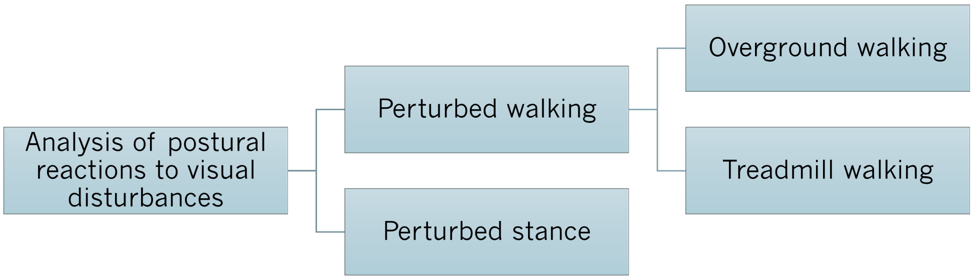


Figure 11: Differentiation between studies that analyze perturbed gait and those that analyze perturbed stance.

and uninformed.

3.3.2.2 Influence of threatening situations on postural control

Another group of studies intend to understand the influence of fear of falling caused by elevated heights or obstacle negotiation. Sun et al. [168] assesses the interaction between balance control and anxiety created by elevated heights and balance-demanding perturbation shifts, during standing. Cleworth and colleagues' study [169] investigates the effects of threat-related factors (fear, anxiety, and lack of confidence) on dynamic postural reactions. An analysis of kinetic, kinematic, and muscular responses was conducted to cover limitations of other related work, which they consider essential to consolidate current knowledge about the influence of threat on dynamic balance. Habibnezhad et al. [170, 171] contributed with two virtual height-related studies. In one of the studies [170], they empowered the participant with a first-person model that displayed virtual first-person legs. They measured the impact of heights on gait characteristics that may indicate postural sway. The other study conducted by the same author [171], entitled "Neurophysiological testing for assessing construction workers' task performance at virtual heights", follows the same line of research, by producing knowledge about construction workers who have to perform tasks at heights. The aim of this study is to introduce tasks with cognitive load and examine the postural stability of the workers, and to assess their task performance while standing at virtual heights. Still on the influence of fear of falling and anxiety on static or dynamic balance, the transposition of obstacles in virtual reality can be fitted into this branch since, the participant being immersed loses the visual feedback of the real world. This can pose challenges to balance since it can trigger anxiety, caused by fear of falling. Stepping over an obstacle involves lifting one leg to allow foot clearance while the supporting leg bears the entire body weight. By itself, this movement is an internal introduction of a balance perturbation. To understand the neuromuscular control strategy that is used to clear a virtual obstacle, Ida et al. [172] investigated the control of vertical posture during object avoidance tasks and further compared two supplementary conditions: clearing obstacles in immersed and non-immersed virtual reality. In previous studies on obstacle avoidance, lead and trail legs in humans have been considered to be controlled independently. However, this perspective is not sufficiently proven because the visuomotor transformation that occurs in the lead leg can have an influence on the trail leg. For this reason, Hagio and Kouaki [167] try to elucidate this interplay between the visuomotor transformation at the lead leg and the trajectory of the trail leg, using a virtual obstacle crossing task.

3.3.2.3 Compensatory postural reactions analysis

In this review, there are 20 studies that report postural reactions arising from the introduction of visual disturbances. In some cases, mechanical disturbances are introduced concurrently. One paper [194] selected for inclusion criteria presents an experiment that attempts to overcome the difficulties encountered by university education in teaching gait analysis based on nonlinear time series techniques. Due to the high cost of motion capture systems, motorized treadmills and virtual reality devices, this group was motivated to build an educational platform for the assessment of gait stability. Martelli et al. [173] focused on understanding the extent to which healthy young adults modify and adapt gait patterns during overground walking in a virtual environment that introduces multidirectional and continuous visual perturbations. The intent is to investigate the modifications that occur in spatiotemporal gait parameters and the propensity to make adaptations due to prolonged walking in the virtual environment and due to the introduction of visual perturbations. Ketterer et al. [192] investigated if continuous sensory conflicts induced by optic flow perturbations can challenge the postural system sustainably. Interestingly, a recent study [193] has been included whose primary aim is to test the effectiveness of electrical stimulation in improving reactive balance control after the introduction of visual disturbances. Riem and his colleagues contribute with two studies. In 2018, the first study [176] aims to examine whether an immersive virtual reality system is able to induce compensatory responses to the visual disturbances it imposes, similar to a physical disturbance. The second study [175] in 2020 investigates the effectiveness of applying repeated discrete visual perturbations to modulate balance in order to create a concept of balance perturbation system without the costs associated with mechanical perturbation mechanisms. Chander et al. (2019) [195] also conducts a study with the same goal of determining whether visual disturbances impact postural stability. However, it does a differentiated study of the impact of non-expected versus expected visual disturbances. Mohebbi et al. [178] investigates the effect of visual information on dynamic postural responses (postural sway) in upright stance. More specifically, it aims to investigate how the speed and amplitude of the visual perturbations it presents influence joints. The same author, more recently, developed a methodological approach to study human postural control applying visual perturbations whose amplitude and velocity can be modulated independently to investigate how the amplitude and velocity of visual field motion influences the postural responses evoked [190]. In another similar study [181] in the sense that it also evaluates the speed of the visual perturbation, Santoso and Phillips investigate the effect of the velocity of the optical flow in a virtual environment on the participant's postural control. Dennison and D'Zmura [179] investigated postural sway induced by visual disturbances with the intention of noticing whether participants reported motion sickness. Stramel et al. [180] using a robotic device with a light touch paradigm, explores whether this person-following robot can act as a proprioceptive reinforcer to stabilize overground gait when intoducing visual perturbations. Peterson and Ferris [183] objective of the study is to identify, through electroencephalography and independent component analysis, cortical responses to the visual disturbances they introduce during walking and standing. Lubetzky et al. [184] in one of the studies test the feasability of the virtual reality system in relation to the participants' ability to be capable of withstand cyber sickness with the execution of the protocol. Then the goal is to report the test-retest reliability of the participants' postural response to conditions of disturbed optical flow. With this, it finally describes the changes in measures of postural control and temporal dynamics in response to visual and somatosensory cues.

3.3.2.4 Age-related compensatory reactions analysis

Still with the main purpose of reporting on the postural reactions arising from the introduction of visual perturbations sometimes in conjunction with mechanical perturbations, three studies are identified that focus on the differences in these changes between young and the elderly population. Bugnariu and Fung [156] wanted to understand what are the effects of aging and what is the capacity of the central nervous system to make an adaptation after repeated exposure to visual disturbances, selecting pertinent sensory information and resolving sensory conflicts. Also, Osoba and colleagues [157], by introducing continuous visual perturbations, investigated age-related modifications and the propensity of either group to make visuomotor adaptations when faced with these conflicts. With a young population and an elderly group, Berard et al. [158] investigated the ability of both groups of participants to reweight visual cues, i.e., down-regulate optic flow cues or perturbations, while walking at normal and faster speeds.

3.3.2.5 Patients compensatory reactions analysis

In addition to studying compensatory reactions to visual flow perturbations, 6 studies had been identified that do this with populations with neurological diseases or impairing balance control diseases. Gago et al. [131] study compensatory postural adjustments caused by a paradigm that introduces sensory conflict. In addition, it makes an analysis of the risk of falling in patients suffering from AD. Yelshyna and colleagues [161] has the same goal as the study described above. Through kinematic analysis and spectral analysis, they evaluate compensatory postural adjustments in patients with Idiopathic PD, and assess the role of dopaminergic medication in these processes. Diniz-Filho et al. [163] evaluated postural reactivity in glaucoma patients using a dynamic virtual environment and also evaluated the relationship between postural metrics and patients' fall history. Three studies conducted with patients with some kind of vertibular loss were identified from this review. Lubetzky and colleagues [182] make a comparison of weighting and reweigting of optical flow given a slight somatosensory change (stable or compliant surface) between patients with vestibular dysfunction and age-matched healthy contol subjects. Again the same author published a study [164] whose objective was to compare the dynamic mechanism of balance in terms of postural sway and head kinematic movement strategy between patients with vestibular loss and healthy controls. In addition to studying the effect of aging on balance performance using visual disturbances introduced to healthy subjects, Chiarnovano et al. [162] also studies the effect of vestibular loss on balance performance in the same way. Apart from studying balance in both groups, it also makes an evaluation of a possible balance assessment tool.

3.3.2.6 Objective balance assessment tools

In 2018, Chiarovano et al. [188] publishes an investigation whose goal is to understand the difference between subjective and self-reported measures of instability or dizziness. The goal was to evaluate more specifically the relationship between the DHI score and objectively measured balance performance using the created visual perturbations. The same author [187], has developed a low-cost system to objectively quantify balance and investigate to what extent balance maintenance is an automatic mechanism or a challenging task that requires attention. In the same line of goals, but more focused on balance evaluation, Tossavaien et al. [189] proposed a system that easily implements various mechanical and visual perturbations to serve as a basis for clinical investigations or evaluations of balance and the interactions between the visual and vestibular systems. This system combines visual stimulation in virtual reality with posturography on an moving platform.

3.3.2.7 Balance training protocols

The next group of studies comprises efforts towards the development of balance training protocols. Two studies by the same authors [130, 132] design and evaluate the effectiveness of a virtual reality training in improving the recovery reactions of a visually induced slip and the implication on fall reduction, in a group of older patients. In one of the studies of this sequence, the objective remained the same, but the evaluation of the effect of the training was done based on kinematic and muscular responses. Liu and colleagues [186], test the effectiveness of the virtual slips induction training protocol compared to a moving platform training control group. Finally, the study by Peterson et al. [183] may fit a balance training proposal since their goal is to determine how virtual reality-induced transient visual perturbations modify electrocortical activity and behavioral outcomes while practicing a balance task during gait.

3.3.3 Equipments

3.3.3.1 Display devices

Regarding the device responsible for the VE display, the HMD, there is a domain of two devices: HTC Vive [157, 168, 170, 173, 175, 176, 178, 180] and Oculus Rift [131, 161, 163–166, 179, 182]. Other devices have been found that require the coupling of a smartphone, such as Google Cardboard [174] and Samsung Gear VR [188]. In addition to these, the Playstation VR headset [167], the Oculus Rift DK2 [183–185, 187] and a more advanced versions of the HTC Vive, the HTC Vive Pro [177] and HTC Vive Pro eye [192], were also used. The most recent study uses the Valve Index VR [193]. In older studies, the iWear VR920 [172] and Kaiser Optics ProView XL5 [156] are the VR systems. Four studies use Sony's Glasstron LDI-100B [130, 132, 160, 186], the oldest uses Virtual Research V8 [189], and headsets such as Sensis piSight [169] and NVisor [158, 159] are also found.

3.3.3.2 Sensors

All of the reviewed studies use the inertial sensors that the HMDs already have built-in or optical tracking systems for the participant to move the VE according to the rotations and displacements of his head and body, except for one study, in which the virtual environment was navigated with using an X-box controller [179]. There are fundamental properties of the devices that directly influence the sensation of immersion and presence in the user [196]. The most important feature is the image quality, which relates to the level of detail that the user perceives. The quality of the visual experience and the level of immersion is associated with the positive transfer of skills to the real world. Technical aspects contribute to superior quality such as field of view (HTC Vive: 110°; Oculus Rift: 110°), display resolution (HTC Vive: 1080 x 1200; Oculus Rift: 1080x1200) and refresh rate (HTC Vive: 90Hz; Oculus Rift: 90Hz) [116]. To keep track of subject's reactions to balance perturbations, sensors of movement and pressure, physiological sensors of muscular and neurological activity were used. The most used configuration combines a force platform or a motion tracking system with a physiological sensor, mainly EMG. It was also noted that many studies use only the force platform or the optical tracking system. 24 of the included studies used force plates to calculate ground reaction forces, with some of them motorized and capable of sliding and tilting to introduce physical perturbations. The WBB use in 4 studies [162, 179, 187, 188] should be highlighted, while 17 research studies used infrared optical tracking systems to obtain a 3D representation of the body [130, 132, 156, 158–160, 165–167, 169, 170, 175, 176, 183, 185, 186, 192]. Regarding the physiological sensors found, the use of EMG stands out and is used in 12 studies [130, 156, 166, 168, 169, 172, 178, 183, 185, 186, 188, 190]. In order to preserve the equilibrium, the Central Nervous System (CNS) uses two types of alterations in the activity of trunk and leg muscles. The first type is Anticipatory Postural Adjustment (APA)s. APAs are associated with the activation or inhibition of trunk and leg muscles prior to the current perturbation of balance in order to minimize the negative consequences of a predicted postural perturbation. The second type of adjustment in the activity of postural muscles deals with the current perturbations of balance and is termed as compensatory reactions or Compensatory Postural Adjustment (CPA)s. Therefore, to uncover the patterns of muscle activations responsible for keeping postural balance, electromyography sensor systems are extensively used in the studies present in this review. As part of the investigation of balance in threatening situations, such as at high altitudes, Electrocardiography (EKG) and electrodermal activity GSR systems were used [168, 185], that quantifies the anxiety created in the participants. In addition to these physiological sensor systems, neuronal activity monitoring was also found, using Electroencephalography (EEG) [168, 183, 185]. Another configuration to consider is the choice between the treadmill or overground walking. Some studies prefer to use instrumented mats and ask participants to do a natural walk because the gait differences introduced when walking on a treadmill are well documented [197–200], which can mask some changes in gait parameters. As a constituent part of the equipment used in the experiments, the use of safety harnesses is highlighted when the risk of falling is high, thus preserving the safety of the participants and the apparatus used. This additional component must be used with caution in order not to support the subject's weight, influencing their compensatory movements [201].

3.3.3.3 Electromyography

In total, the sensor locations of 12 studies were surveyed. In the article by Sun et al. [168] they placed 8 EMG sensors on the left and right Tibialis Anterior, Soleus, Gastrocnemius Lateral Head and Gastrocnemius Medial Head. Physically, a moveable platform, randomly under certain conditions introduced a 10-degree toe-down perturbation. Studies have shown that with increasing fall-related anxiety, co-contraction of the ankle joint agonist/antagonist muscle pairs intensifies. Cleworth et al. [169] also used exposure to heights and added a physical perturbation in the AP direction, through support-surface translations. EMG was collected from pairs of surface electrodes placed 2 cm apart along the muscle bellies of the right Soleus, Gastrocnemius Medial, Tibialis Anterior, Biceps Femoris, Rectus Femoris, L3 paraspinals, External Obliques, and Middle Deltoid. The results for the translation of the bearing surface at elevated height showed a significantly greater amplitude in the Middle Deltoid, with a similar trend for the Gastrocnemius Medial. Of the studies introducing pitch perturbations, Bugnariu and Fung [156], in addition with roll perturbations, rotated a removable platform concordantly or discordantly with the virtual environment. That is, the perturbation was concordant if visual perturbation was combined with synchronized surface perturbation in the opposite direction resembling the real life perception. Eight bilateral muscles were instrumented with bipolar Ag-AgCl surface electrodes to record electromyographic signals using a Noraxon system: Tibialis Anterior, Gastrocnemius Medial Head, Vastus Lateralis, Semitendinosus, Tensor Fascia Latae, Erector Spinae at the L3 level, neck extensor and neck flexor Sternocleidomastoideus. In young adults, muscle recruitment generally followed a distal-to-proximal sequence, regardless of perturbation direction or sensory conflicts. The ankle muscles, Tibialis Anterior and Gastrocnemius Medial, were first to be activated during toes-up and toes-down tilt, respectively, at a latency of 80-100 ms. Vastus Lateralis and Semitendinosus were activated 110- 130 ms, followed by Tensor Fascia Latae and Erector Spinae approximately 30-50 ms later, while the neck muscles Sternocleidomastoideus and neck extensor were activated at 150-170 ms. Peterson and Ferris [183] introduce discrete visual perturbations on the roll, in both directions of rotation. In addition to using EEG, they recorded 8 lower-leg EMG channels. Four muscles on each leg: tibialis anterior, soleus, medial gastrocnemius, and peroneus longus. During walking, the significant intermuscular connectivity increases occurred primarily between peroneus longus and soleus muscles on either leg. Parijat et al. [130] with an eight-channel EMG telemetry Myosystem 2000 (Noraxon, USA), was used to record bilateral temporal activations from vastus lateralis, medial hamstring, tibialis anterior, and medial gastrocnemius muscles of the lower extremity. To quantify the reactive responses to the virtual slip, and to quantify training effects, the coactivity index was calculated based on the EMG activity ratio of the antagonist/agonist muscle pairs (TA/MG and VL/MH). Proactive responses were found by muscle onset of MG, TA, MH and VL activation of the slipping limb. During VR training of slips, the proactive and reactive strategies in terms of muscles, showed early activation of all slipping limb muscles by the fifth trial. In the subsequent trials, early onsets were only detected for the VL and TA. Likewise, Liu et al. in 2015 [186] use the same type of EMG sensors and the same locations. Drolet et al. in 2020 [166], through the manipulated visual and proprioceptive feedback, focuses on the muscle activation patterns before, during, and after surface changes (through a variable stiffness treadmill) that are both visually informed and uninformed. The muscle activity of both legs was obtained using surface electromyography via a wireless surface EMG system (Delsys, Trigno Wireless EMG). Electrodes were placed on the tibialis anterior, gastrocnemius and soleus muscles. These muscles were selected as they play a primary role in ankle motion and stability, in which the gastrocnemius and soleus muscles produce plantar flexion of the foot and the tibialis anterior produces dorsiflexion of the foot [202]. The results of this study provide strong evidence that anticipation of walking surface compliance changes affects both muscle activation and kinematics of the leg preparing to land on the surface of different compliance. That accelerated swing phase is also supported by increased activity on the dorsi flexor muscles (tibialis anterior) during the swing phase before the landing of the foot. Moreover, the false expectation of transitioning to the new compliant surface, i.e., being prepared to step on a compliant surface but eventually stepping on a rigid one, elicits delayed responses on the plantar flexor muscles (Soleus and Gastrocnemius). Finally, Ida et al. in 2017 [172] records surface electromyography signals using disposable surface electrodes (Red Dot, 3M, USA). The electrodes were bilaterally attached to the bellies of the tibialis anterior, medial gastrocnemius, rectus femoris, biceps femoris, external oblique, rectus abdominis, and erector spinae muscles. These leg and trunk muscles are known to be involved in the control of upright stance when dealing with symmetrical and asymmetrical perturbations [203, 204]. The results supported that the obstacle avoidance task involves the activation of postural muscles in the early preparatory phase. Furthermore, the results on EMG integrals also supported that a VR setting modulates the components of postural adjustments during the avoidance action. The findings suggest that VR potentially deteriorates natural perceptual motor reaction and postural control strategy, which may relate to a participant's sense of presence in computer-simulated environment.

3.3.4 VR Protocols

3.3.4.1 Visual and Physical perturbations

Given the restriction of the use of VR through an HMD, the most common type of disturbance induced in the subjects was visual disturbance, also called optical flow [205]. It induces modulations in gait patterns and in postural control performance [206]. Can be stabilizing when appropriate or destabilizing when conflicting. During locomotion, there is a decreased stabilization pattern due to artificial changes in the optical flow [207, 208]. In contrast to mechanical disturbances, these visual disturbances do not impose inertial effects that move the position of the body in relation to the support surface. On the other hand, they create the illusion that a correction is necessary. This perception of the need for correction makes the responses to visual disturbances highly idiosyncratic, while the responses to mechanical disturbances are very consistent. This variability from subject to subject makes the introduction of visual disturbances a proper manner to assess the risk of falling, since the reaction reveals the dependence on visual feedback. Visual processing is a critical component of balance and gait stability [209]. For this reason, all studies use visual disturbances that consist of a change in the HMD display. Some studies combine optical flow with a physical disturbance [130, 156, 164–166, 168, 169, 172, 176, 182–184, 186–189]. Depending on the objective, these disturbances can be concordant or dissonant, i.e., to simulate a fall as in the real world, the subject has to visualize a rotation around the roll direction contrary to the rotation that occurs on the support surface. In most articles, visual disturbances consist of rotations or tilts on the Roll, Pitch and Yaw axes, translations on the vertical axis, and in ML and AP directions, or overlapping oscillations in the visual field, in the AP or ML axis directions. The use of moving objects in the virtual environment can also be considered a visual disturbance since it alters the sensory feedback information. For example, in Lubetzky and colleagues studies [164, 184], animated objects with variable speeds were used to trigger participant's postural changes, or virtual blocks that intend to resemble the movement of passers-by on a street. Place the participant virtually at heights can also be considered a visual disturbance because this threatening condition alters balance in a static position or during gait, increasing the weight of proprioceptive and vestibular reflexes [210]. Finally, two studies [131, 161] also used a visual field perturbation to simulate a fall on stairs: instantaneous translations on the vertical axis of the VE, which transported the subject from the top of a staircase to some steps below.

Some authors have introduced physical disturbances in conjunction with visual disturbances to investigate the interconnection between different sensory systems. More specifically, 2 studies [165, 166] have changed the stiffness of the gait support surface, in accordance with the visual feedback or not. Another study used a tilting platform to study the same interconnection of sensory systems [156]. To complement the threatening situation of high heights in the virtual environment, some researchers [168, 169] used platforms to create a bi-directional translational AP movement or a toe-down pitch rotation to increase anxiety. In a control group, mechanical disturbances in the form of pulls to one side in the direction were applied to compare compensatory responses with similar visual disturbances [183]. The use of compliant surfaces was also found in order to hinder the process of maintaining balance, namely stability trainers and a BOSU balance trainer [164, 182, 184, 187, 188]. To induce slips, Parijat and colleagues used a slippery floor surface that moved after heel contact to make baseline measurements and to quantify training transfer effects [130, 132].

3.3.4.2 Virtual environments

Most of the VE present in these studies were created with VR game engine software, mainly Unity and Unreal Engine. The latest headsets are equipped with inertial sensors that track the position and rotation of the head. Without exception, in real time, they alter the VE accordingly with the movement of the head, with a high refresh rate to avoid cyber sickness. The headsets that did not contain this functionality were equipped with an inertial sensor or used motion tracking systems information to synchronize the movements with the VE. Most of the VEs created by the researchers are realistic enough to make the experience immersive and create a sense of presence. Three of the oldest virtual environments have monochromatic representations: two simulate a simple corridor and the other only shows moving patterns. Interestingly, 2 studies [183, 185] capture video in real time from a webcam mounted on the HMD and works as a VE. As a general rule, the VEs did not use the subject's own representation in the form of an avatar. In 2 studies [169, 170] focused on the relationship between high heights and balance, studied the influence of the presence of an avatar that represented the movement of the legs or hands, more precisely while performing cognitively challenging tasks. When the studies were carried out on treadmills, the VE altered at the same speed as the subject's gait pace, whether self-chosen or imposed by the experimental protocol. Otherwise, the speed of ground walking determines the change in the VE representation. This synchronization is significant. If three-dimensional objects are presented in a dynamic, consistent, and precise manner, the VE is an ecologically valid platform for presenting dynamic stimuli in a manner that allows for both the veridical control of laboratory measures and the verisimilitude of naturalistic observation of real-life situations [112].

3.3.4.3 Protocol specifications

The diversity in the intervention duration or the exposure time to VR scenarios has already been pointed out as a limitation in several literature reviews that investigated the use of immersive VR for rehabilitation [99, 101, 105]. In the studies included in this work, the same trend can be seen. In addition, as the objectives are mainly to study compensatory adjustments triggered by visual perturbations, many of the authors do not even disclose the duration of the intervention. Some reveal the duration of exposure to the VE and whether there are pauses between trials to minimize the effects of exhaustion or lack of concentration. It is common to find pre-test periods to get the participant used to the HMD and the VE thus preventing a mask on the effects of disturbances and to assess whether there is simulation sickness. This pre-test period also increases the sense of presence. Habituation time was found in several studies, mainly in those that involved walking or interacting with the VE. In slips induction and balance training studies, follow-up measures were taken in order to quantify the transfer of skills acquired in training. These measurements were made on different days, in contrast to the majority, which collected the data in just one session, on the same day.

3.3.4.4 Population

Regarding the studies populations, 22 of them used only healthy young adults for the experiments, while 5 studies used healthy older adult participants only. In 4 studies, healthy young people were assigned to the control group, while older adults participated in tests with VR. Only one conference paper does not specify the type of subjects, nor their demographic characteristics. The scarcity of control groups is evident, being common practice only in studies with pathological subjects. From this pathological population, 4 studies were found with subjects diagnosed with vestibular impairments. One of these studies with patients with vestibular loss was the only one in which the participants were appointed by their neurologists to be part of this rehabilitation and experimental assessment. Other pathological subjects with AD, PD, Mènière disease and glaucoma were subjected to trials with VR.

3.3.5 Measures

The outcomes assessed in the studies are presented. The section arrangement is not chosen according to the objective of the study. However, it should be noted that the text is organized in order to place the measures from the articles that share common goals. The measurements identified are sorted by the evaluation mode of balance: static or dynamic. These metrics can be directly measured, calculated, or estimated. They can originate from a force plate or a motion tracking system, and through EMG, EEG or EKG systems. In addition, this section also presents traditional clinical tests performed and questionnaires.

3.3.5.1 Static balance measures

As mentioned in the taxonomy of the studies, there is a separation regarding the way the experience is carried out: standing or walking. This subsection presents the measures performed by the studies when evaluating the balance of the subjects when standing. In the group of studies that relate the anxiety created by threatening scenarios, namely heights, in one study [168] is carried out a Sensory Organization

Test (SOT) [211] through a force plate that objectively measures the subject's AP and ML body sway. In addition, by placing an EKG on the chest, anxiety on stance was quantified, detecting heart rate variability. Co-contraction Index (CCI) of the ankle dorsiflexion is used in this study to measure subject's dynamic anxiety during the trials, as it has been shown in previous studies that, with the increase in fall related anxiety, co-contraction of the ankle joint agonist/antagonist muscle pairs intensifies [212]. Through the 16 front electrodes of the EEG, the Power Spectrum Density (PSD) was calculated for each of the five frequency bands, which served as a feature to classify, through supervised learning classifiers, the balance function states. In a second study [169], with a motorized platform that rises virtually to a height, postural reactions to platform accelerations in AP directions were quantified by displacements and CoP speeds, namely peak and time to peak. With the IRED markers, angular displacements in the AP and ML directions were calculated to quantify the relative movement of the arms in relation to the trunk. Here, using EMG, the CCI was also calculated for four muscle pairs in the lower leg, upper leg and trunk. The amplitudes of the muscle signals were measured by the integrals. To characterize the obstacles negotiation [172], mechanical quantities were measured based on coordinates of the rigid body such as the maximum elevation of the toes, measured vertically. Using an accelerometer, it was possible to measure the accelerations of the foot on the various coordinated axes, while overcoming an obstacle. In conjunction with a force plate, it was possible to estimate CoP displacements while EMG data was collected, for example the onset muscle activation and integrated muscle activity.

Lubetzky et al. [182], with vestibular dysfunction patients and age-matched controls to quantify sensory entrainment with a force platform embedded in the floor, measured the postural sway in the AP direction. The Power Spectrum of the CoP time series was calculated at four different frequencies, each corresponding to a perturbed system. The gain and phase (modulus and argument) of the frequency response function were also calculated to, respectively, indicate the ratio of the amplitude of the response to the amplitude of the stimulus, and to express the relative timing of the response compared to the stimulus.

Studies were found [187–189] that aim to validate alternative objective assessment systems to replace traditional clinical assessments of balance or dizziness, more specifically in the diagnosis of individuals with Mènière compared to the Romberg Test [213, 214]; in assessing vestibular impairments compared to the Dizziness Handicap Inventory (DHI) [215]; or in the assessment of balance through objective measures to the detriment of the gold standard Equitest [216]. In the first study mentioned [189], this alternative system uses a force platform built with three force transducers arranged in an isosceles triangle to measure the vertical ground forces and the displacements in the AP and ML directions, CoP and its stabilogram - which calculates the respective ML and AP range, standard deviation and mean velocity as a function of time [217]. With these measurements, the Romberg Quotient is calculated, which is the ratio between the mean velocity with eyes closed and the mean velocity with eyes open. A feature that derives from these measurements is the Vertical ground reaction Force Power Fraction which detects slow changes and drifting, such as hanging on the safety harness. In the study that follows [188], it is intended to assess the effectiveness of objectively measuring the parameters evaluated by DHI through a WBB, in patients suffering from vestibular impairments. The metrics calculated here are the postural sway, the ocular speed and the vestibulo-ocular reflex gain. With EMG, cervical and ocular vestibular evoked myogenic potentials are measured. Finally, in a study conducted by the same author [187], the effectiveness of the gold standard balance test Equitest is compared with the measurements made also with a WBB, namely the length path and speed of the Center of Gravity (CoG) of the body estimated from values obtained by the CoP. Another outcome of the assessment is the presence or absence of a fall during the test, and the time elapsed until the fall.

In a research [179] to understand the extent to which visual disturbances induce cyber sickness in a VE, using a WBB, the postural sway was measured in the AP and ML directions, a time series was performed, and standard deviations were calculated for all participants. When determining the impact of an expected or non-expected visual disturbance [177], introduced by VR, CoP excursions were measured using a force plate, i.e., the average displacements of the CoP sway and its speed, which allowed an estimate of 95% ellipsoid area, which represents an area where changes in the CoP are such that 95% of the data is within that ellipse, and 5% outside. In an experiment to examine the contribution of vision to balance [178], laser range finders for the hip were used to calculate the Root Mean Square (RMS) of body angles to quantify body motion variability. The frequency response was analyzed with a discrete Fourier transform to determine the transfer and coherence function. These functions describe the dynamics of the body's response to the visual stimulus. The gain of the transfer function shows the sensitivity response as a function of frequency; the transfer function phase expresses the relative timing of the response compared with the stimulus at each frequency. Bugnariu and Fung [156] analyzed the selection of strategies in the regulation of upright balance related to aging. With two force platforms, they measured the CoM displacement, the resulting force of the CoP in the horizontal plane and the moments in the AP and ML directions. From the EMG signals, the muscle latencies were registered. In another work [162], aiming to understand: i) the effect of age on balance performance; and ii) the effect of complete vestibular loss on balance performance using VR, using a WBB, instead of considering the 3-dimensional trajectory of the CoM, considering one coordinate of the projection on a horizontal plane of the CoP, determined the so-called statokinesigram. In addition to this calculation, they determined the body's CoG trajectory as a function of time, normal limits and Limits of Stability (LoS). Santoso and Phillips [181], to investigate the optical flow effect on a subject's postural control, measured with a force plate the ground reaction forces and moments that allowed the calculation of CoP in the AP direction, and estimated the CoP distance traveled. Similarly, most studies that attempt to understand what influence induced visual perturbations have on postural dynamics, using a force plate, it is measured the postural sway in the AP and ML directions using the CoP. Less frequently, these parameters were also calculated using motion capture systems, particularly when the objective was to identify head or trunk deviations [183, 185].

3.3.5.2 Dynamic balance measures

When the study requires participants to walk, whether on a treadmill or overground, gait parameters are measured, mainly using motion capture systems, whether they are infrared optical tracking systems or, less frequently, non-optical motion tracking systems e.g., IMUs. The most present parameters in the studies that classify gait stability were joint angles position and speeds, feet positions that allow estimating the Heel-Strike (HS) and Toe-Off (TO); spatio-temporal gait parameters such as Stride Length (SL), Stride Width (SW), Stride Time (ST), Stride Velocity (SV), Direction of Progression (DoP) (the angle of the stride vector from two successive heel strikes of the same foot) and DoP deviation. From these parameters their variability, means and standard deviations were calculated, which are representative of instability in human gait [218, 219]. The CoM can also be calculated for each gait cycle, which allows estimating the Margin of Stability (MoS). Using an inertial sensor, it was also possible to measure the acceleration signals in all axes, which allows the calculation of CoM excursions. Body orientation and body displacement were also estimated, and temporal and frequency analyzes were used to monitor sudden changes due to postural adjustments [131, 161]. The most considered EMG signals during locomotion were the activation times of the muscles and their integrals, to distinguish between anticipatory or compensatory muscle activations.

3.3.5.3 Traditional Clinical Tests and Questionnaires

Questionnaires or traditional clinical tests were carried out in various circumstances. To assess the subject's eligibility in the study, the Physical Activity Readiness Questionnaire (PAR-Q) and the SSQ were used. Before and after the tests, questionnaires were performed, such as weekly exercise minutes, sitting hours per day, the DHI or the Motion Sickness Susceptibility Questionnaire (MSSQ), the Patient-Reported Outcomes Measurement Information System (PROMIS): a set of person-centered measures that evaluates and monitors physical, mental, and social health in adults and children; confidence questionnaires to assess perceived anxiety, Witmer & Singer Presence Questionnaire and Slater-Usoh-Steed Questionnaire for VR presence and the Montereal Cognitive Assessment (MoCA). Other tests that were done before and after the trials were the Functional Balance and Mobility test, the Romberg test, the ability to maintain tandem stance, Time repeated Sit-to-stand, and again the MSSQ. Visual acuity tests (Snell's chart) and cognitive deficits Mini-Mental State Examination were also conducted. In the case of patients with glaucoma, they underwent ophthalmological examination including review of medical history, visual acuity, slit-lamp bio microscopy, intraocular pressure, gonioscopy, ophthalmoscopic examination, stereoscopic optic disc photography, and Standard Automated Perimetry; history of falls with Falls Screening and Referral Algorithm. For vestibular patients, Activities-specific Balance Confidence Scale and again DHI. In the PD study, idiopathic PD patients had normal clinical postural stability measured by the retropulsion test (item 12 on Movement Disorder Society Unified (MDS-UPDRS-III)) and had a Hoehn & Yahr of 2 (OFF state). Clinical data were also collected, including years of disease duration, MDS-UPDRS-III (scored as either an in the OFF or the ON state). A brief neuropsychological examination was performed using the Portuguese version of the MoCA with scores normalized to the Portuguese population. Additionally, self-assessment of fear and acrophobia questionnaires were seen - James Geer Fear Questionnaire, Cohen's Acrophobia Questionnaire and International Physical Activity Questionnaire.

3.4 Discussion

There has been a growing interest in balance control mechanisms investigation using VR. More specifically, virtual reality using HMDs, replacing the usual three-dimensional projections, LCD screens, and robust and difficult to access systems such as CAREN and CAVE. HMDs provide an effective means of introducing visual disturbances due to the technical qualities that allow the user to feel a high level of presence and immersion. In addition, it requires low physical space, not much expertise from researchers and a decrease in the price of these devices was observed, which makes them very attractive for this type of studies discussed in this review. As seen, these investigations focus mainly on the balance system of the human body. Postural balance is studied either during standing or gait. It would be interesting to combine the two types of static and dynamical analysis, approaching studies closer to the real conditions that a subject is confronted with on a daily basis. One can either suffer a threat to balance during walking or standing.

Additionally, as mentioned in studies whose objective was to analyze anxiety and its relationship with balance, the threatening situations shown in the VE could be extended to other threatening scenarios, other than just the heights. By identifying situations and environments in which falls normally occur, these can be replicated to the VE, increasing the spectrum of physiological, neuronal, and muscular reactions beyond the fear of heights. The level of complexity and detail of the VEs created by the researchers falls short of the computational resources and tools that are currently available, which allow the creation of photorealistic scenarios. Therefore, the level of presence and immersion would substantially increase, which can trigger more realistic reactions. In addition, since most current HMDs have built-in headphones, it would increase the experience of VR if auditory stimuli were introduced along with visual stimuli. Not only to enrich the experience but also to become another stimulus that defies balance. None of the 40 studies seen added audio to their VEs. Finally, another sensory conflict that could be explored would be the haptics to simulate touching objects and collecting data from the reaction to intentional or unintentional touch.

With regard to the subjects chosen for the studies, the small sample size, the lack of control groups and non-random allocation, and the failure of follow-up measures are repeatedly mentioned. This leads to the fact that these studies lack methodological quality, which can introduce bias in the results. Even so, the vast majority claim clear evidence on the effectiveness of the introduction of optical flow and the induction of compensatory behaviors, as well as on the effectiveness and reproducibility of balance assessment systems using VR. The study of these behaviors could be extended to a wide range of neurological patients as done for patients with PD, AD, and those with problems in the vestibular system, and also to those with locomotion issues related to cardiovascular accidents or acquired brain injuries. Even more importantly, it would be interesting to test these systems in the elderly, the age group in the most fragile situation and the most exposed to falls that can cause serious damage.

That said, the addition of physiological sensors (EKG and GSR) would bring additional validity to the results: a neurological (EEG) and muscular (EMG) with motion tracking systems sensor fusion. Thus, the tendency to study only the kinematics of anticipatory or compensatory adjustments would decay, and knowledge about neuromuscular changes would increase. Another interesting point in addition to the auditory stimuli inclusion, as pointed out, would be the introduction of electrical stimulation of the vestibular system with Galvanic Vestibular Stimulation (GVS), responsible for creating the illusion of movement in the head and for creating balance losses. GVS systems are simple, economical, and safe. They can contribute to an easy and controlled induction of imbalance, which can be advantageous when used in conjunction with disturbances of the visual field. Experimental protocols with VR must go through a standardization process mainly to define gold standards in the period of habituation to the device and the VE, define the duration of the exposure to avoid creating habituation to disturbances and thus mask some behaviors, define the type of disturbance or more efficient combination of disturbances to evoke imbalances, setting breaks during the trial to avoid exhaustion and cyber sickness and, finally, adopt follow-up measures in a post-intervention period to validate the reproducibility of the created system and evaluate its effectiveness.

Due to space restrictions, many studies use treadmills instead of asking subjects to walk naturally on the floor. As discussed previously, the treadmill itself influence some important gait parameters. Therefore, it is advisable and more naturalistic to analyze the overground gait. Finally, the calculated metrics could be generalized throughout the studies since they all use very similar systems, mainly the force platforms and the optical tracking systems, which share even the same brand and data analysis software. This would allow generalization in the interpretation of results. These analyzes could be conducted in real-time, which required more computational capacity and synchronization between systems. However, this could enable the participant to adjust his behavior according to the visual or haptic feedback of the own compensatory movements. That is, the orientation and acceleration of the headset would no longer be the only element that closed the loop, and the movements and forces exerted on the floor would have an influence on the VE. Likewise, data processing could be done online, with automatic labeling since machine learning algorithms already allow for greater ease in this process than previous statistical methods. Another aspect to be modified is related to the public accessibility of the data generated, which could be valuable information for fall prediction classifiers.

3.5 Conclusions

This review presents a broad spectrum in the application of VR perturbations to balance, as can be seen by the variability of the study goals. On the other hand, contrary to what is common in this type of review, it focuses only on VR delivered by HMDs, considering that only these devices configure a totally immersive experience. Instead of including articles that use VR only for motor rehabilitation, this review extends to any type of analysis that is based on postural reactions to balance disturbances, which may be visual, physical, or a combination of both. From the included studies, it can be said that the study of gait, posture, and neuromuscular reactions of individuals to external disturbances has aroused a growing interest in the scientific community, largely due to the improvement of VR systems and their accessibility. It was possible to verify that, using an HMD, it is easy and safe to introduce visual disturbances. Given its portability, and the enthusiasm it can bring to the elderly, it is estimated that it will become a widely used tool in therapy, motor rehabilitation and fall prevention training both in households and at home.

The main objective of this review, which was to confirm the efficiency that the disturbances imposed on the HMD had in provoking sensory conflicts that trigger imbalance, was achieved. In addition, important information was collected for the design of an standard experimental protocol that aims to induce imbalances and near-fall situations, producing kinematic, physiological, neuronal and muscular data and to build a data set that is equipped with more realistic falls than simulated and ideally serve as a classifier for fall prediction systems, mitigating the problem of falls in the elderly community.

Chapter 4

Virtual Environment and Project Overview

After the actual state of the literature survey of the use of virtual reality in neurorehabilitation, more specifically in the introduction of visual pertrubations through HMD to assess postural reactions, it was pointed out in both the introductory chapter and the conclusion of Chapter 2 that there is a lack of datasets regarding real-world falls. [10, 12] In this chapter, it is created a novel approach which aims to address this scarcity. Throughout the chapter the tools and implemented methods to achieve the desired solution are presented.

4.1 Introduction

Briefly, the project's primary purpose is to contribute to preventing and predicting falls. Achieving this presupposes diminishing their occurrence and detecting them prematurely. To achieve this is necessary to overcome the problems identified in the literature. Thus, virtual reality interventions should be conducted in a standardized manner so that the interpretation of the results can be generalizable in the scientific community dedicated to decreasing the problem of falls. Hence, the data collections from these studies investigating the induction of postural imbalance through perturbations could complement or replace existing data from simulated falls. The foundation of this project is to make use of an immersive virtual environment to introduce visual disturbances into it. Based on the reactions of the participants, it will be worthwhile to collect data close to real-world fall events. Using virtual reality and its immersive properties, a naturalistic virtual environment will serve as a medium for the punctual introduction of sensory disturbances to the participant. These perturbations intend to challenge balance and elicit compensatory reactions. The compensatory adjustments will be recorded primarily in kinematic parameters and muscle activation patterns. The survey of articles that used virtual reality via a headset and introduced visual perturbations also revealed a tendency to use only one category of visual perturbation in each study [181, 183, 194]. This may restrict the compensatory responses variety of the subjects [220, 221]. For this reason, a varied list of visual perturbations is proposed to be delivered in the VE. The software where the virtual environment was built, the automatic way to introduce the perturbations is then explored, and finally, the stages of the developed project are outlined and what each step wants to solve and its interconnection.

4.2 Virtual Environment Design

First, it is fundamental to present some prepositions that the virtual environment must respect. Some research in the neuroscience community overlooks the relevant aspects of real-world activities and interactions and applies simplistic stimuli without complexity in their studies [222]. If the goal of research is to predict real-life phenomena, there is a need for a reasonable balance between strict experimental regimes and the inclusion of naturalistic, dynamic and contextually embedded stimuli [223]. Virtual reality environments combine experimental measurements control with attractive backgrounds to enhance the participant's experience. Zaki and Ochsner [224] dictate three specificities that real-world information differs from laboratory stimuli: i) cues are multimodal; ii) they include visual, semantic, and prosodic information; and iii) it is dynamic in the sense that information is presented serially or concurrently. It is embedded in a context that can influence the interpretation of the study.

The use of HMDs to present virtual environments that digitally recreate real-world activities allows the potential for laboratory control to be coupled with everyday functioning [111]. Nowadays, there is a computational capability for efficient administration, stimulus presentation, and automatic logging of participant responses. Since virtual environments allow experimental control and dynamic stimulus presentation in ecologically valid scenarios, background narratives engage participants and enhance the experience [225]. The future of VR in neuroscience is strongly tied to developments in technology that help to immerse the user in convincing, life-like sensorimotor illusions [149].

According to Franzen and Wilhelm's [226] definition of ecological validity, one of the requirements is verisimilitude - the test conditions have to resemble ADL. Ecological validity refers to experimental conditions reasonably similar to those in a real-world setting. In virtual environments, contextually rich simulations with multiple sensory cues might be considered to have greater ecological validity than environments that are limited to only the necessary and sufficient features for an experiment [112]. To achieve ecological validity and thus ensure transfer of motor skills to the real world and compensatory reactions to perturbation as natural as possible, the creation of a virtual scenario as close to reality as possible was the priority.

The VE was developed in Unity software, version 2020.3.2f1. Unity, commonly known as Unity3D, is a game engine and a Integrated Development Environment (IDE) for interactive media creation, typically

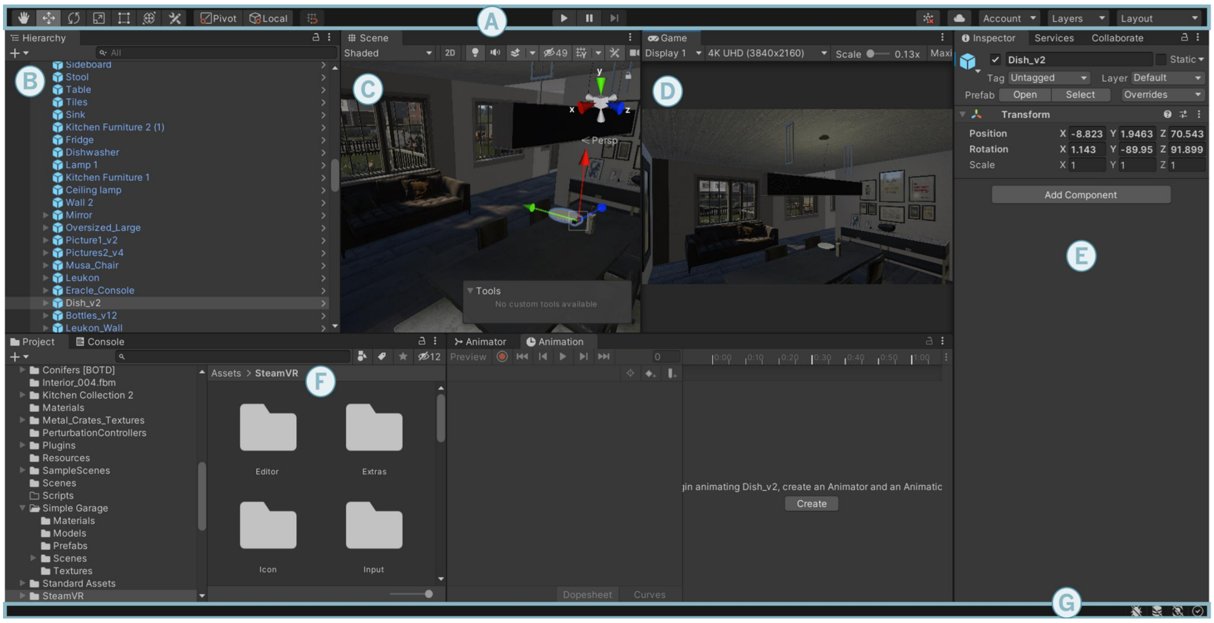


Figure 12: A - Toolbar; B - Hierarchy window; C - Scene View; D - Game Preview; E - Inspector; F - Project Window; G - Status Bar

video games. As CEO David Helgason put it [227], Unity "is a toolset used to build games, and it is the technology that executes the graphics, the audio, the physics, the interactions, the networking. Unity is famous for its fast-prototyping capabilities. Overall, users found Unity to be slightly easier to use than Unreal Engine, thanks to its native C# coding language, which should be relatively familiar for all developers and its overall workspace layout. It is a more accessible platform to start creating over Unreal Engine, with a slightly steeper learning curve [228]. Figure 12 displays the default view from Unity IDE.

The workspace layout is organized into two main windows for previewing the environment visually, two bars, and three important tabs. The enumeration below gives a brief description of each element of the layout.

A The Toolbar provides access to essential working features. The left contains the basic tools for manipulating the scene view and the game objects within it. In the center are the play, pause, and step controls. The buttons to the right give access to Unity Collaborate, Unity Cloud Services, and Unity Account, followed by a layer visibility menu. Finally, the Editor layout menu, which provides some alternate layouts for the Editor windows and allows to save custom layouts.

B The hierarchy window is a hierarchical text representation of every GameObject in the Scene. The hierarchy reveals the linking structure of the game objects.

C The scene view allows to visually navigate and edit your scene. This view can show a 3D or 2D perspective, depending on the type of project.

D The game view simulates how final rendered game will look like through the scene cameras. Pressing the play button, the simulation begins.

E The inspector Window allows viewing and editing all the properties of the currently selected game object. Because different types of game objects have different sets of properties, the layout and contents of the inspector window change each time one selects a different game object.

F The project window displays your library of assets available to use in your project. When one imports Assets into the project, they appear here.

G The status bar provides notifications about various Unity processes and quick access to related tools and settings.

For the construction of the virtual environment it is necessary to import assets. The assets can contain game objects, 3D models, textures, animations, among other elements. These will be converted into visual information when imported to the project. The key asset is the SteamVR plugin, used for a direct interface between the HTC Vive Pro and Unity. Unity's Prefab system allows one to create, configure, and store a game object complete with all its components, property values, and child game objects as a reusable asset, acting as a template from which it is possible to create new Prefab instances in the scene. From SteamVR plugin, "Player"prefab was used, allowing to set up the main player component. It also deals with all the relevant SteamVR Input actions needed since it contains the necessary scripts. An example of input is the pose-type action, a representation of position and rotation in three-dimensional space. This action is used to track HTC Vive Pro controllers and to estimate the position and orientation of the head. This allows the view in the virtual world to be adjusted to the participant's movement in the real world. This workflow is briefly described in the diagram in Figure 13.

The base virtual environment started with a typical suburban block. As in neuro-rehabilitation with virtual reality, to increase engagement, the virtual environment was designed as close to reality and as familiar as possible. Being able to place participants into an environment enhancing their presence allows to better study human-environment interactions by doing so within a controlled context [229]. Neo and colleagues [230] have identified important points regarding the level of realism that a virtual environment must have in order to examine human behaviors. First, the need for realistic details and textures must be evaluated according to the type of investigation. Next, they point out the need for context to exist, i.e., the opposite of an abstract world. Real-time representation of the participant's avatar and continuous positional tracking of head movement, and sensory feedback, are other important points to draw out the most natural reactions from participants [230]. The VE will be designed as realistic as possible, constrained only by the rendering capabilities of the equipment. The context is as naturalistic as possible in the sense that it will take place in a typical street and house. Beyond that, sensory feedback will be introduced with physical elements in the real world during the immersion, and a motion capture system based on inertial sensors

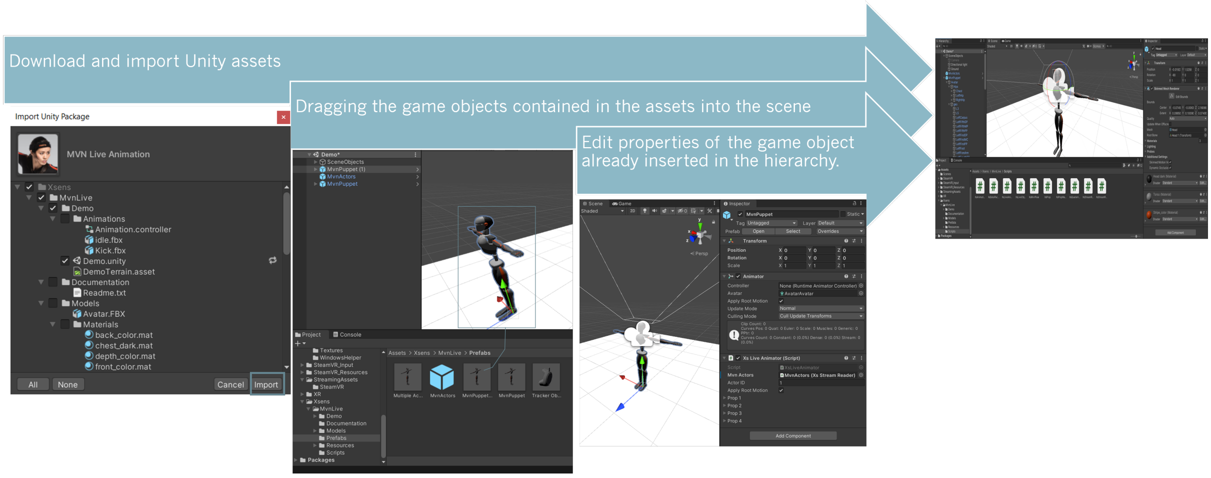
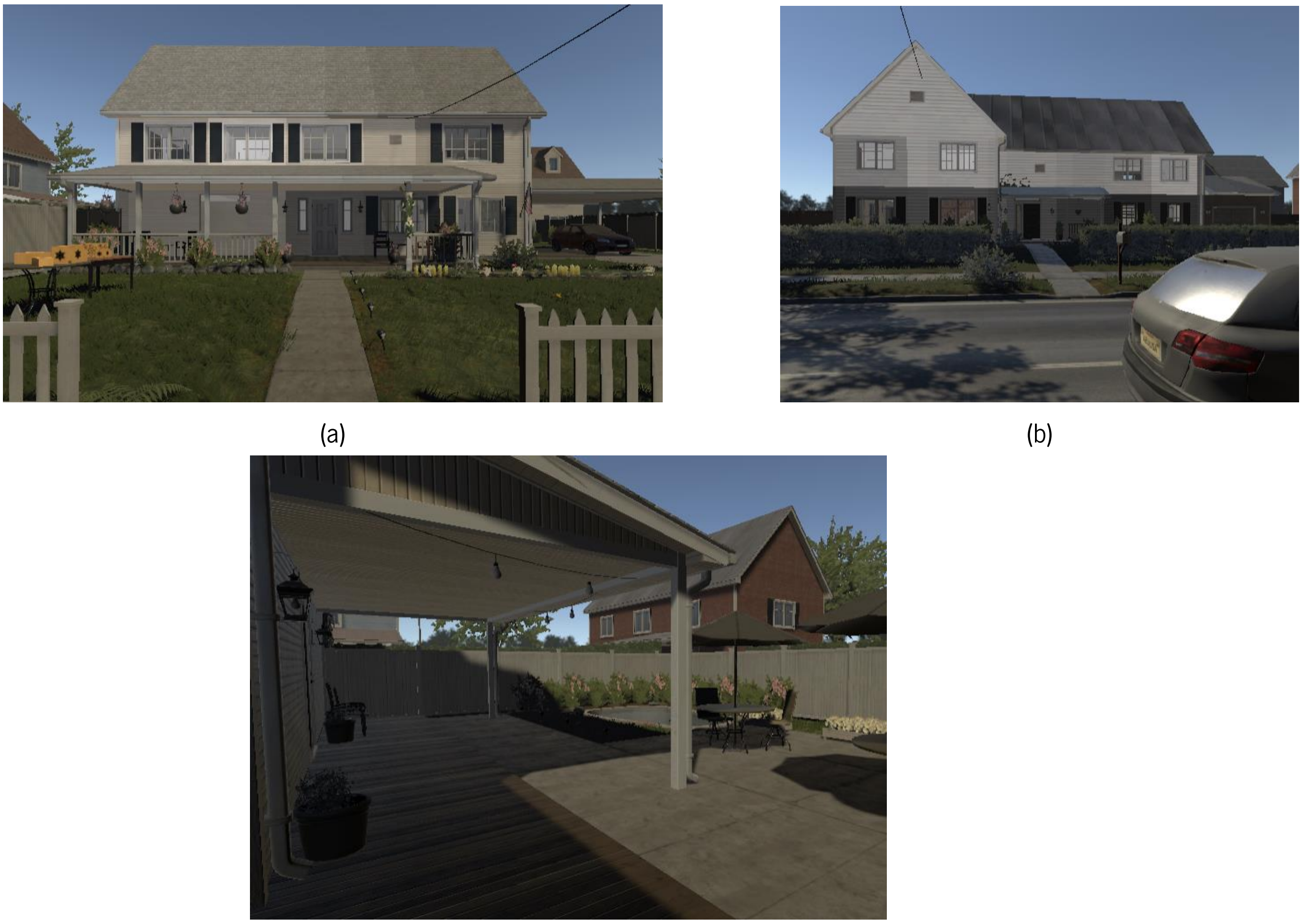


Figure 13: Basic workflow for creating a scene: import assets into the project, drag their desired components into the scene, and set their properties.

will allow the real-time representation of a real-time body avatar, embedded with head movement from the HMD tracking system.

Following the assumptions that were suggested earlier to extract natural reactions from the participants, the VE has the following structure: it contains a suburban block that encompasses the roads and their integral units such as parked cars, sidewalks and the accesses to the houses, and two fully furnished houses. The two distinct houses are organized into rooms just as they exist in the real world. These rooms include a kitchen, a living room, two bedrooms, a bathroom, an outdoor area with a pool and garden, and the commuting areas such as hallways and stairs. Screenshots of the house rooms created are shown in Figure 14. More reproductions of the divisions can be found throughout the document.



(a) (b) (c)

Figure 14: Snaphots from the virtual environment. Representations of exterior views of houses: a) House 1, b) House 2; c) House 1 backyard.

4.3 Visual perturbations

The most important decision in this virtual reality perturbation-based solution was the selection of visual perturbations to be used. Since the goal of the perturbations is to provoke a postural reaction as close as possible to a real-world fall, visual perturbations must induce all fall categories during normal walking and seizing the virtual environment created. These falls are the backward and forward fall, the lateral falls, the slip, the trip, and a syncope, as shown in Table 5. To collect data that best approximates an occurrence of these types of falls, Chapter 3 categorizes the visual disturbances found into: i) translations or rotations on one or several axes simultaneously; ii) visual and proprioceptive disturbance; iii) visual field oscillations and v) predefined trajectories object movement. This categorization serves to support the choices of the visual disturbances conceptualized.

Table 5: Visual disturbances and corresponding fall categories

Visual Perturbation	Fall category
Roll Rotation ML Translation	Lateral
Object Avoidance	Lateral/Forward (FW)/Backward (BW)
AP Translation FW/BW	FW/BW
Pitch Rotation	Slip
Virtual/Real Object	Trip
Vertigo	Fall from heights
RPY Rotation	Syncope

4.3.1 Visual and proprioceptive mismatch

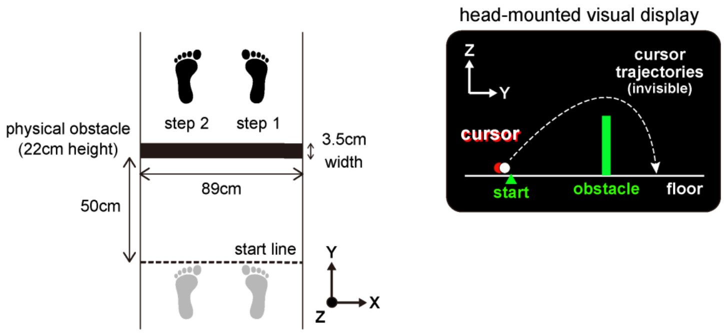


Figure 15: Setup and view of the participant during the virtual obstacle crossing. Partially reproduced figure from [167].

Hagio and Kouzaki [167] introduced a discordance between the visual and proprioceptive systems by showing on the HMD a virtual representation of leg height different from the real one, while the participant was crossing an obstacle, as shown in Figure 15. In other words, a discordant proprioceptive feedback was introduced. This description brought up the idea of presenting an object in the VE that the participant was instructed to cross. This object would have a virtual height smaller than a real object placed in the participant's path, on the floor. Trying to cross the obstacle, the participant would be confident that nothing will collide with his foot. In reality, probably will hit the real object because the height of this object is greater than what the subject sees in the virtual environment. This visual disturbance would cause a trip, with a mixture of physical disturbance, since it would hit a real object. For this disruption, the edge of the sidewalk is used to instruct the participant to climb the sidewalk in a natural way (Figure 16). Frost et al. [165] and Drolet et al. [166] also introduced a mixture of physical and visual disturbance. This perturbation is visual information discordant with proprioception, by varying the stiffness of the ground via a mechanical device.

Figure 16: Sidewalk trip induction via visual and proprioceptice mismatch.

Thus, participants were asked to walk in a straight line on virtual stone, randomly interspersed with pieces of sand or grass. Under certain conditions, when the participant stepped on a material they expected to be soft, a surface of stone-like hardness was introduced, creating a inconsistent sensory information, as can be understood from Figure 17 and 18. This perturbation caused muscular and kinematic changes when the subject was preparing to land with their leg on a softer surface. For the purpose of this study, this visual disturbance is of no interest since none of the authors reported loss of balance.

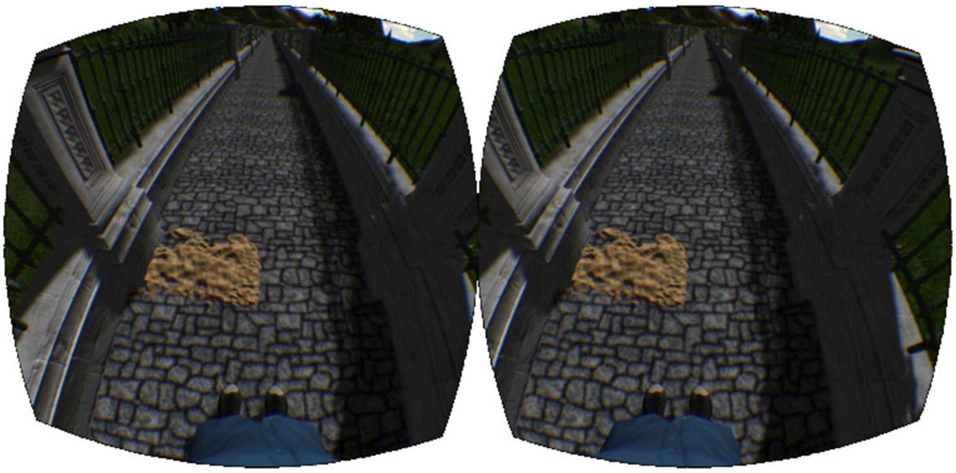


Figure 17: The virtual reality environment consisting of the walkway, sand patch, and virtual avatar, taken from [165]

Figure 18: Figure from Drolet et al. [166] setup consisting of a treadmill of variable stiffness and virtual environment delivered by a HMD.

4.3.2 Vertigo

Translations in the vertical direction are expected to place the subject in a situation of virtual high height, which can cause fear and anxiety, and has been shown to influence postural control. In this framework, four studies [168–171] placed their participants in different virtual height places, causing a sensation of vertigo and anxiety, to study the influence of these conditions on gait patterns and standing balance, as seen on Chapter 3. Figures 19 to 21 depict examples of the virtual environments created to place the participant at virtual heights. Habibnezhad and colleagues [170] studied in these conditions of elevated heights the influence of having or not having the real-time representation of an avatar while the subject walked along a narrow path on a scaffold at the top of an unfinished building.

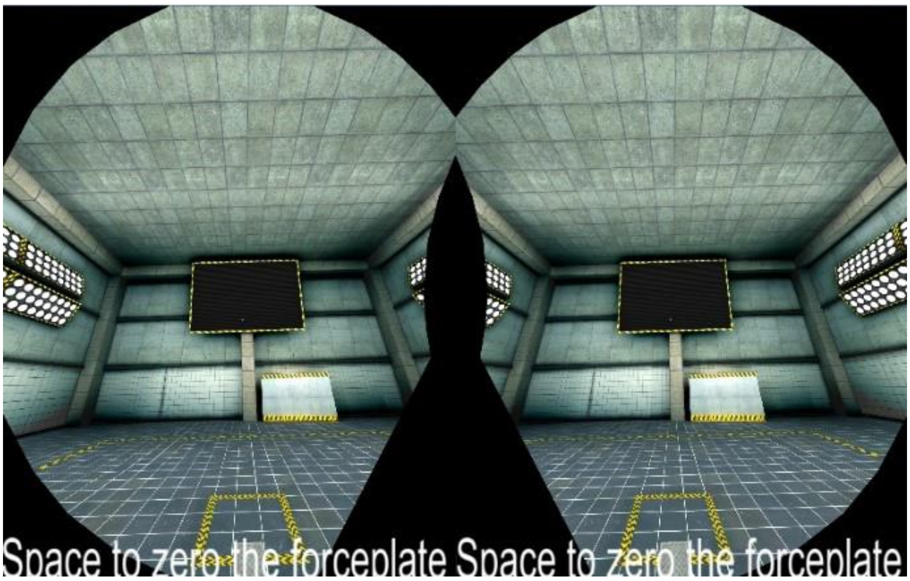


Figure 19: During the experiment, the platform in the VE went up to 2.5 meters, 5 meters and 7.5 meters, or a pit around the participants went down to 2.5 meters, 5 meters and 7.5 meters, such that all participants experienced seven distinct height conditions [168].

Figure 20: Avatar in view is a representation of where the individual stood, while the view for the subject was a first person view from the perspective of this avatar. Subjects were positioned at a virtual Low height (0.4 m) then a High height (3.2 m) during the experiment [169].

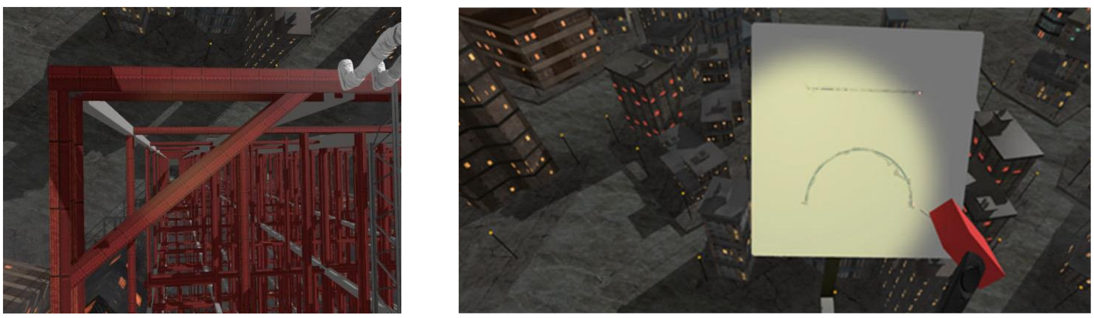
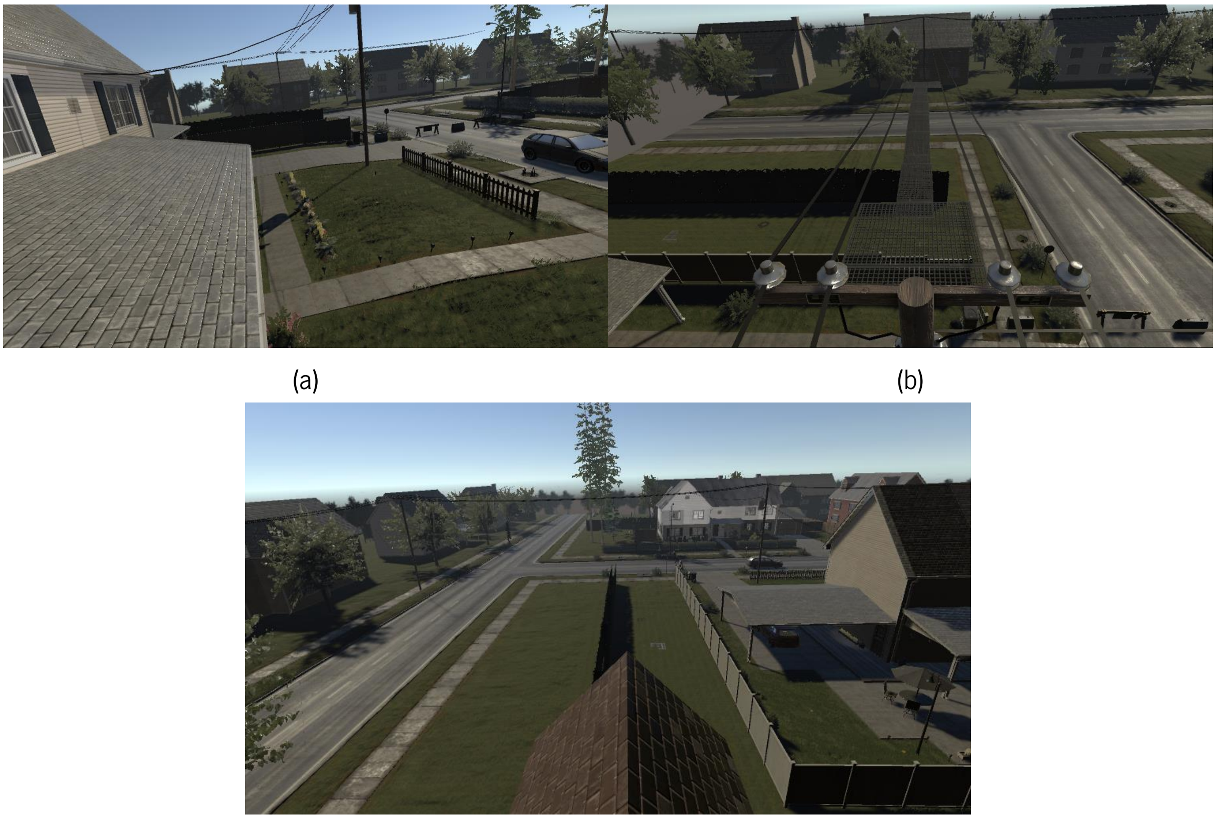


Figure 21: The participants were subjected to height by locating the triangular beam path on the 17th floor of an unfinished building, on the left [170]. On the right, the same author placed the participants on the edge of an unfinished building [171].

These studies have led to the development of three distinct situations of vertigo. The higher places in the virtual environment were used. The first vertigo location is the roof of one of the houses (Figure 22a).

The second is on another ceiling, in a very narrow area to walk, and that forces the participant to walk as on a beam to keep his virtual representation of the avatar from falling off (Figure 22b). The last location is between two high voltage poles (Figure 22c). This way, three distinct situations of vertigo were created: a) the participant can only fall to one side; b) can fall to both sides, and c) can fall and slide if the path in not followed. In order to induce more fear and stress in the participant, the avatar was influenced by gravitational force just like a human body.



(a) (b)

(c)

Figure 22: Snaphots from the Vertigo places: a) Simple roof, b) Electricity Pole c) Window Roof Beam Walking

Gago et al. [131] and Yelshyna et al. [161] used the vertical plane translation to simulate a fall from the top of a staircase. Figure 23 shows the virtual environment and the translation sequence. These two studies originated the idea of inducing a virtual fall in the participant. This fall should happen from an upper floor to a lower floor of the house, opening up the floor beneath the participant. This disturbance is called a Free Fall and its setup in the virtual environment can be seen in Figure 38.

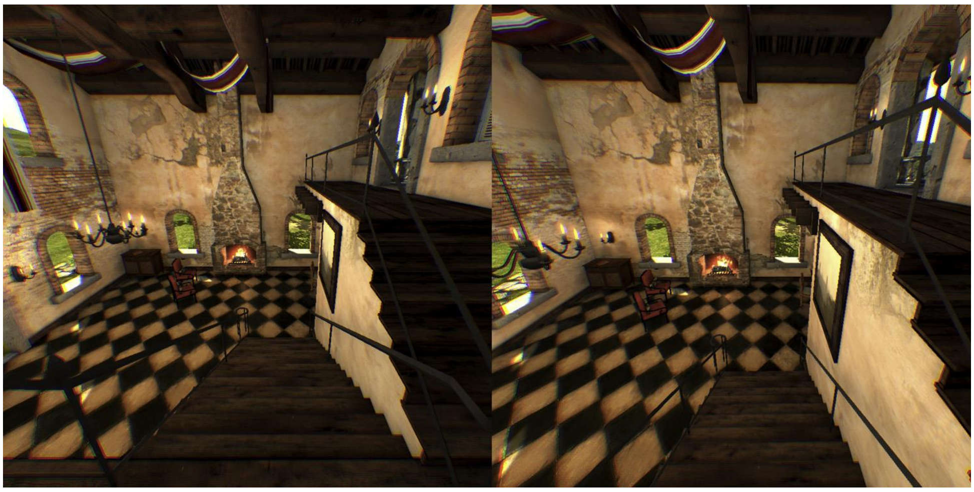


Figure 23: Figures reproduced from [131, 161]. Top of the staircase (left) and after a visual displacement of 1.17 meters along the vertical axis, translating to the middle of the stairs (right).

4.3.3 ML-Axis translation

Visual perturbations consisting of translations in the axial plane are intended to replicate a physical perturbation that pulls or pushes the body in a left or right direction. Guzman and colleagues [194] replicated a school hallway and applied continuous oscillations in the mediolateral axis to produce foreground movement whereas Riem et al. [175] applies discrete shifts in the virtual environment with a speed of 1m/s in the mediolateral axis. The participant is standing on a bridge with a pole serving as a visual reference. Participant viewpoint and virtual position are moved to the left or right, held for 9 seconds, and returned to the original viewpoint in the middle of the bridge for one second, as depicted in Figure 24. Dennison and D'Zmura [179] created a virtual environment that represented a space station that contained long corridors. While freely navigating the corridors, approximately every 2 seconds a visual disturbance would occur for 260ms and pull the participant's body in a left or right direction. These directions were calculated relative to the direction in which the participant was looking. This study also applies visual perturbations that pull the participant's body forward or backward. This type of perturbation also falls into the category of AP axis translations. With this literature support, the ML plane perturbation was created, consisting of a floor moving bidirectionally, at a speed of 1m/s. The ML Translation perturbation virtual setup can be seen in Figure 37.

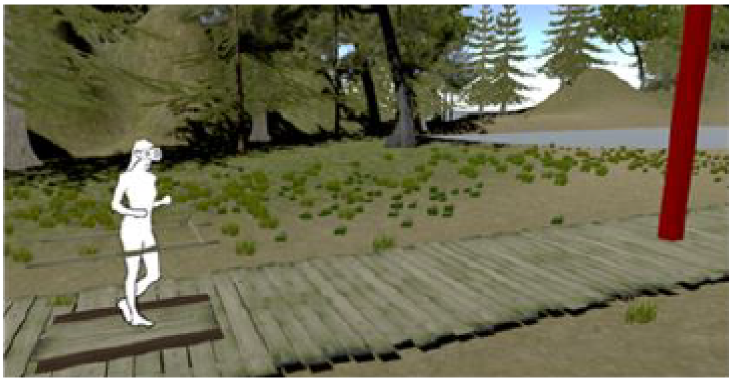


Figure 24: Experimental setup and virtual scene with the treadmill and red pole. Reproduced from [175].

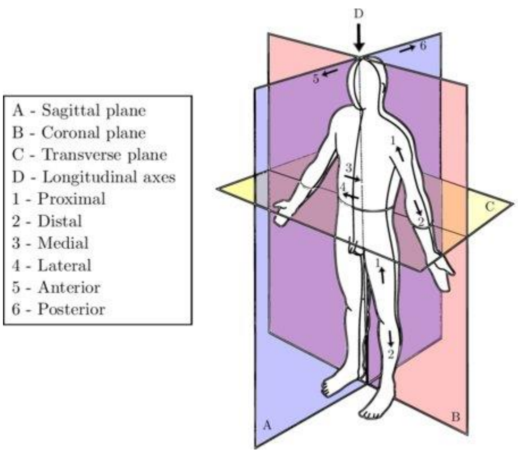
4.3.4 AP-Axis translation

As noted in the previous point, Denison and D'Zmura's study [179] also uses this type of translation in the sagital plane. These visual perturbations give the sensation of pulling or pushing the participant's body, simulating a backward or forward fall. Santoso and Philips [181] used the same visual perturbation by making a hallway shift abruptly for three meters, at three different speeds, in both directions. Another study [163] uses this translation. The axis is located in the participant's eyes, virtually located inside a tunnel. The perturbation occurs in peripheral vision and is a sum of sinusoids, meaning that the perturbation is continuous, and causes the participant to travel forward and backward on the AP axis. The two perturbations created to cover this kind of backward or forward fall were translations in a corridor. These translations occur at different locations but both have a translation speed of 1m/s and are not bidirectional: one forward, one backward. The corridor of the virtual environment that will serve as the medium for one of these perturbations is shown in Figure 25.

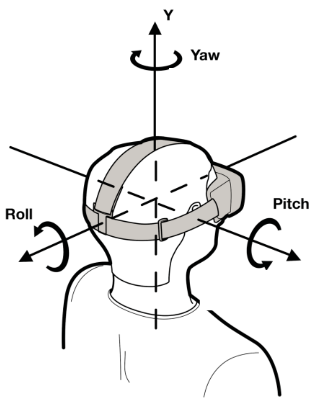
Figure 25: AP Translation corridor.

4.3.5 Axis rotations

Euler angles were used to express the rotation direction for this visual perturbations. Euler angle notation consists of the pitch, yaw, and roll angle formed in three directions. A rotation about the x-axis, or about the frontal plane, is refered to as a perturbation in the roll. A rotation about the y-axis, which is the same as rotating the coronal or frontal plane, is a pitch disturbance. Similarly, rotation about the z-axis is a disturbance in yaw. Figure 26 describes the notations used for translations and rotations in the anatomical planes of the human body.



(a)



(b)

Figure 26: a) Human body anatomical planes. Figure taken from [231]. Directional terminology is also displayed and will be used to denominate translations; b) RPY angles used in the notation of visual disturbances, adapted from [232].

Roll rotation is intended to imply in the human body a lateral swing from left to right or vice versa. Bugnariu and Fung [156] apply this type of visual disturbance along with an associated physical disturbance, concordant or discordant in vestibular sensory levels. The perturbation has an amplitude of 8 degrees and a speed of 36 degrees per second. The stimulus is applied in each direction, separately. Peterson and Ferris [183] also applied this kind of rotation while the subject was trying to balance by walking on a beam (Figure 28). This perturbation had an amplitude of 20 degrees, and a duration of half a second. The direction CW or CCW was random. Riem and colleagues [176] also used a perturbation on the roll in the form of pseudo random sinusoidal stimulus. The perturbation had three different intensities: 3, 6, and 11 degrees. In a similar way to these works, two directions of rotation (CW and CCW) are used, and three different amplitudes (10, 20 and 30 degrees). The rotation speed is constant. Figure 27 represents one of the perturbations in Roll.

Figure 27: The optical flow undergoes a rotation around the axis that points in the direction the participant is facing. In this example, the virtual environment rotates through a maximum amplitude of 30 degrees in half a second, returning to its normal in one second.

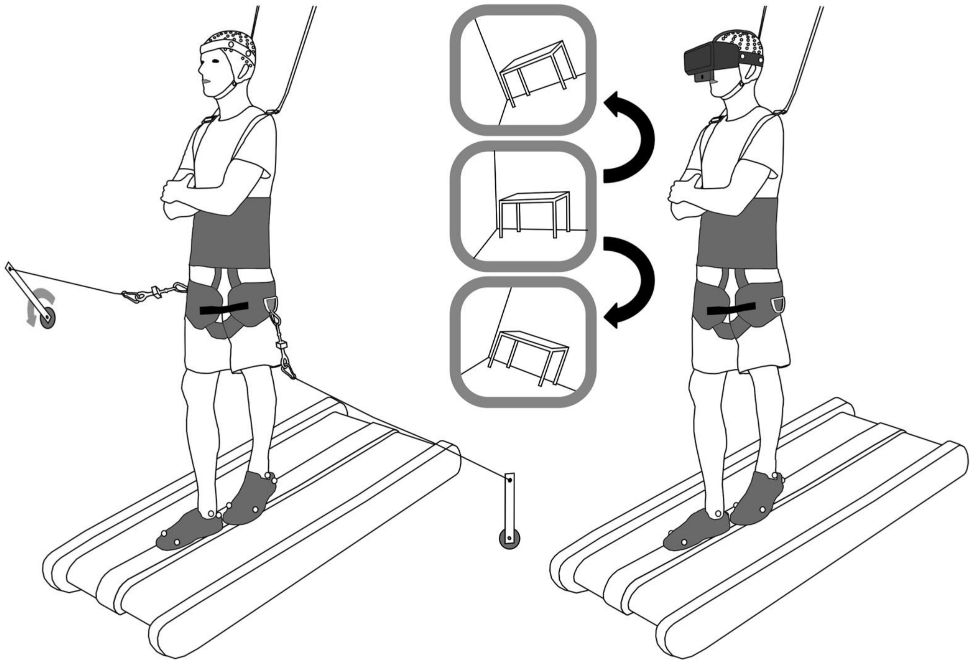


Figure 28: Subject walking on the beam, exposed to pull (left) and visual rotation (right) perturbations. Inset sketches show example 20 degree perturbations in CCW (top) and CW (bottom) directions [183].

Regarding rotations in pitch, The work by Bugnariu and Fung [156] described in the previous point also uses pitch perturbations. The intensity and speed are the same: 8 degrees and 36 degrees per second. Liu [186] and Parijat and colleagues [130, 132], put together a paper that aimed to address slips. To do so, they used a visual perturbation that tilt the virtual environment 25 degrees at a speed of 60 degrees per second. This perturbation was introduced approximately at the heel contact of the right foot. The pitch disturbance developed in this project to simulate a slip has an amplitude of 25 degrees, at a speed of 60 degrees per second (Figure 40).

For the rotation of the optical flow in the yaw, two studies [158, 159] use the visual perturbation around the z-axis. These rotate the focus of vision to the left or right, gradually, over a range of 40 degrees while the subject walks forward. This perturbation wanted to show the subjects' behavior when forced to look away from the direction of walking. The elderly in the study showed deviations in the gait trajectory while the young adults were able to down-regulate the visual information. This type of perturbation was not developed since the participants are all healthy young adults. Therefore, no relevant compensatory reactions are antecipated that would regulate the pathway [158, 159].

This paragraph refers to perturbations that are the result of simultaneous rotations in all three axes: roll, pitch and yaw, called RPY rotation. Three studies evaluate the risk of falling with a group of patients suffering from bilateral vestibular loss [162, 187, 188]. The velocities of each axis were independent, and governed by a pseudo random sum of sines, as shown in Figure 29.

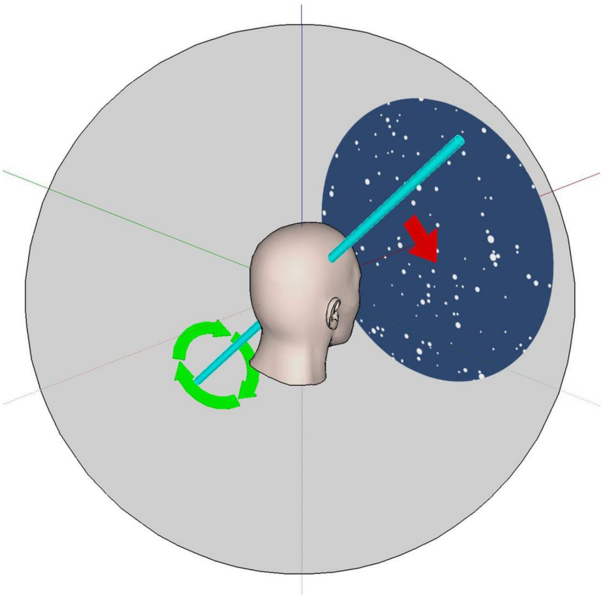


Figure 29: In the present study, one of the visual stimulus was a rotation around the subject at a rate of 20°/s. The entire three-dimensional field (globe with all spheres) was rotated at a velocity of 20°/s and around different axes [187]

Syncope is a temporary loss of consciousness usually related to insufficient blood flow to the brain. It's usually called fainting. Syncope and fall are two common and interrelated geriatric syndromes that cause considerable mortality and morbidity among older adults [233]. These studies gave the idea of creating a disturbance that rotated on all three axes the participant's field of view. This disturbance is achieved by getting up from a chair that physically exists. Virtually, the participant is sitting on a bed, and has to get up. As the participant stands up, suffers the perturbation that simulates a fainting spell or a drop in blood pressure. Figure 30 shows the participant's point of view while sitting on a chair in the real world, and sitting on a bed in the virtual world.

Figure 30: The participant is sitting on a bed in this viewpoint, is instructed to stand up, and undergoes the perturbation that rotates the camera in the three axes, as described in the perturbation with code VP008 in Table 6

4.3.6 Visual field oscillations

This type of visual disturbance is separated from translations and rotations because it describes a superimposition of visual oscillations on the flow of normal vision. The oscillations consisted of a pseudo-random sum of sines of four different frequencies. They occur in the ML direction or in the AP direction. Since the results obtained with this type of perturbation [157] were small changes in gait patterns and gait speed, this type of oscillation in the background was excluded. The camera rotations were prefered.

4.3.7 Predefined trajectories

Programmed trajectories of virtual objects or movements of the background is considered a visual perturbation. In this type of visual disturbance, the participant's point of view is not abruptly changed. Some element from the scene moves, or the whole scene moves, while keeping the viewpoint intact. To analyze the influence of a visual disturbance induced with or without warning, a virtual wall moves against the subject [177]. Ida and her team [172] virtually placed an object coming toward the participant through the floor. The instruction was to lift one foot and avoid contact with the virtual object. Lubetzky and colleagues used object movements in the virtual world. A ball coming towards the participant's head, with the addition of movements in the background of moving cars and cubes representing pedestrians [191, 234]. From the studies that put objects taking off against the participant, the idea of throwing bottles against the participant's head was conceived. Thus, it is possible to record compensatory reactions typical of deflecting

Table 6: Visual perturbations code, description, velocity and intensity description

Code	Perturbation	Parameters
VP001	Roll Axis Tilt - Clockwise	[10°, 20°, 30°] during 0.5s
VP002	Roll Axis Tilt – Counter-Clockwise	[10°, 20°, 30°] during 0.5s
VP003	Support Surface ML Axis Translation - Bidirectional	Discrete Movement (static pauses between movements) – 1 m/s
VP004	AP Axis Translation - Front	1 m/s
VP005	AP Axis Translation - Backwards	1 m/s
VP006	Pitch Axis Tilt	0°-25°, 60°/s
VP007	Virtual object with lower height than a real object	Variable object height
VP008	RPY Axis Tilt	Sum of sinusoids drive each axis rotation [162]
VP009	Scene Object Movement	Objects fly towards the subject's head. Variable speeds
VP010	Vertigo Sensation	Walk at a comfortable speed. With and without avatar. House's height
VP011	Axial Axis Translation	Free fall

an object, which can result in unbalance and falling. This visual disturbance setup is illustrated in Figure 39. Table 6 describes all the visual perturbations choices supported by the literature search mentioned beforehand.

4.4 Triggers and Scripts

This point will cover the method employed to automatize the delivery of the visual perturbations. A system of fixed spaces called triggers was used. When the participant enters these spaces virtually, an animation seen by the user consisting of the visual perturbation is activated. It is necessary to activate the visual perturbation when the subject enters a particular virtual space. The perturbation must remain active while in that location. Unity's built-in 3D physics engine is a system that simulates aspects of physical systems so that objects can accelerate correctly and be affected by collisions, gravity, and other forces. For this purpose, physics engine concepts that must be explained are collisions of rigid bodies and the concept of a trigger. A Rigidbody is the main component that enables physical behavior for a GameObject. With a Rigidbody attached, the object will immediately respond to gravity. If one or more Collider components are also added, the GameObject is moved by incoming collisions. Collider components define the shape of a GameObject for physical collisions. A box collider is a cube-shaped collider component that handles collisions for GameObjects. A box bollider example is shown in Figure 31. The box colliders created to demarcate the activation zone of the visual perturbations has a dimension on the x-axis of 2 Unity measurement units, which will correspond to 2 meters in reality (Figure 32).

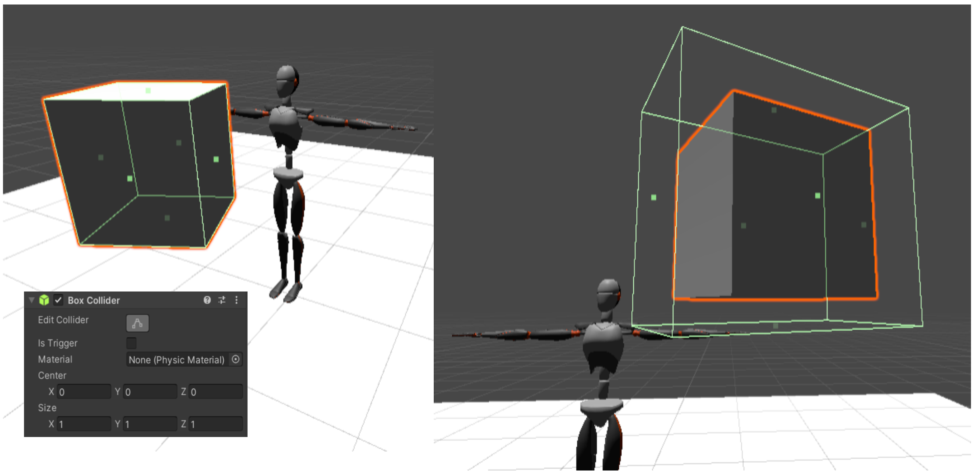


Figure 31: Properties and representation of a box collider. By default, when creating an object like a cube, its collider fits its shape and boundaries.

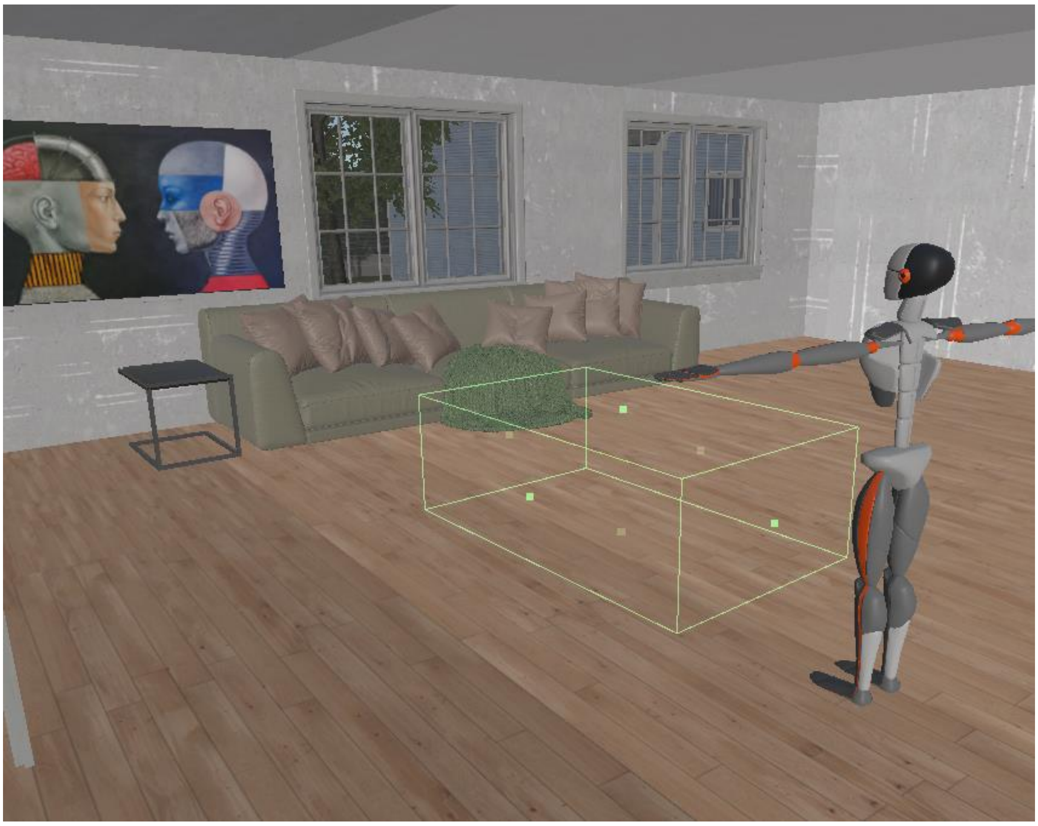


Figure 32: Exemplification of a box collider with the described dimension (2 units of measurement) that corresponds to 2 meters in the real world.

The scripting system can detect when collisions occur and initiate actions using the *OnCollisionEnter* function. However, one can also use the physics engine simply to detect when one collider enters the space of another without creating a collision. A collider configured as a Trigger, using the *Is Trigger* property, as seen in Figure 33 checkbox *(Is Trigger)*, does not behave as a solid object and will simply allow other colliders to pass through. When a collider enters its space, a trigger will call the *OnTriggerEnter* function on the trigger object's scripts.

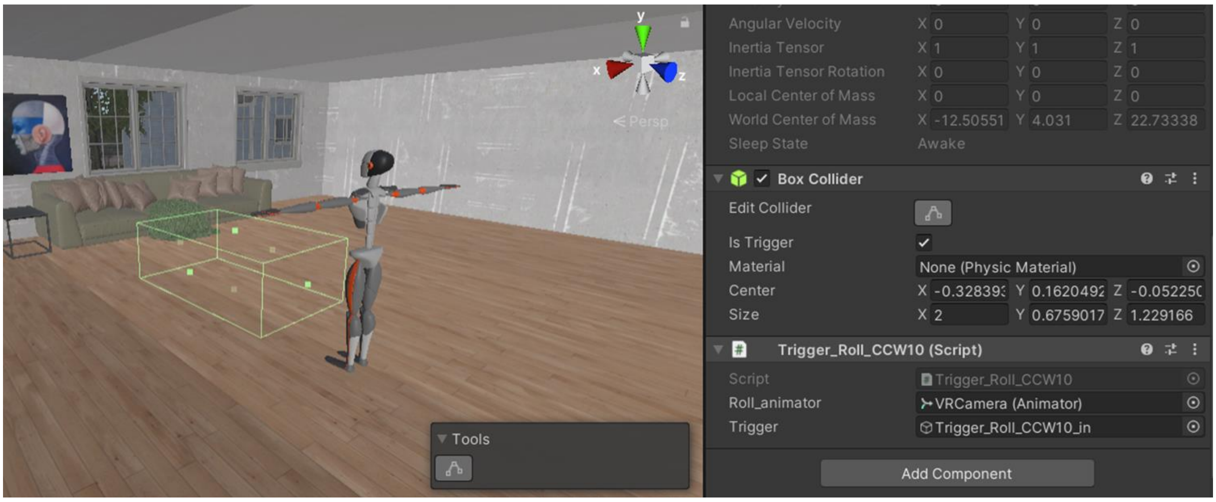


Figure 33: Setting the box collider as a trigger and the associated script visible in the inspector.

4.4.1 Scripts

Scripting governs GameObjects' behavior. The interaction between the scripts and components attached to the GameObjects creates the "game" flow and consequently modifies the view of the virtual environment that the participant experiences in the headset. Unity runs in a big loop. Cyclically reads all of the data that's in the game scene. For example, it reads through the lights, meshes, and behaviors and processes this information. Three main functions run automatically inside Unity: first, *Start* will be called if a GameObject is active, but only if the component is enabled; *Update* is called once per frame. Inside this function is where code defines the logic that runs continuously and other parts of the game that must be updated; *FixedUpdate* is used when one wants to do physics work. Running a Unity script executes some event functions in a predetermined order. Flowchart from Figure 34 describes those event functions and explains how they fit into the execution sequence. A script must be attached to a GameObject in the scene to be called by Unity, and the programming language is C#. All the languages that Unity operates with are object-oriented scripting languages. Scripting can be edited using Visual Studio.

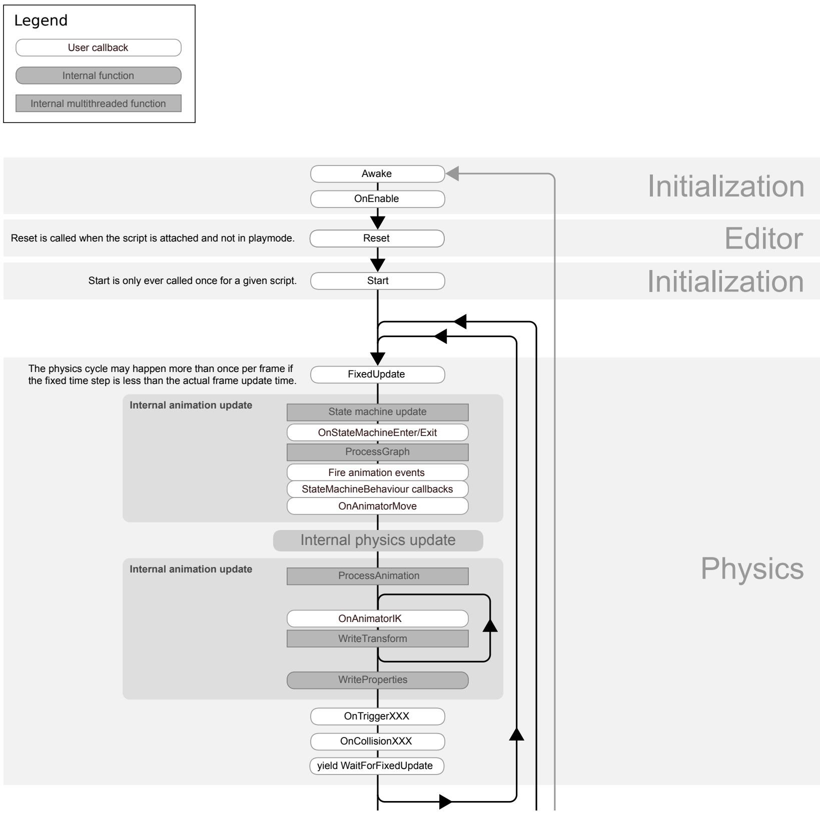


Figure 34: Script lifecycle flowchart, partially reproduced from the Unity manual [235]

With the colliders set up as Triggers to activate the animations representing the visual perturbations, the next step was to edit and set the animations. The following paragraphs briefly describe Unity's animation system. Unity's animation system is based on the concept of Animation Clips, which contain information about how particular objects should change their position, rotation, or other properties over time. For the current project purposes, the object to animate is be the VRCamera, a child object of the Player. The participant will see any changes made to this camera, defined as our Main Camera, as if it were his view. Animation Clips are then organized into a structured flowchart-like system called an Animator Controller. The Animator Controller acts as a state machine that keeps track of which clip should currently be playing and when the animations should change or blend. Figure 35 illustrates a simple Animator Controller.

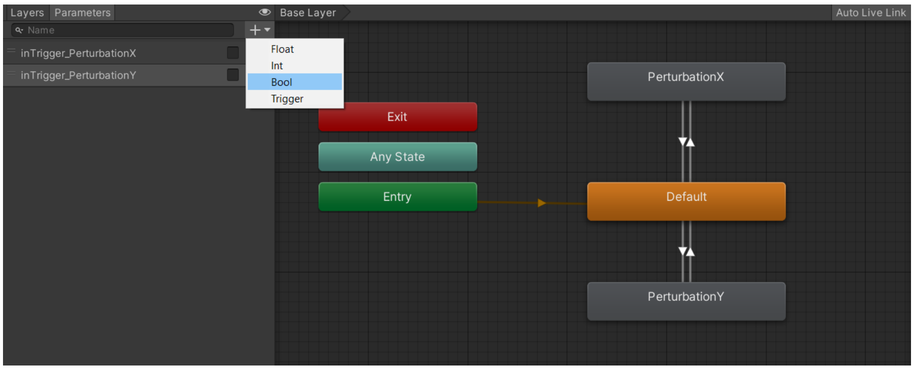


Figure 35: Simple animator controller with two states and two bool variables to control the transitions between states.

The PerturbationX and PerturbationY blocks from Figure 35 represent animation clips. The state that is unconditionally activated after pressing the run button is Default, the orange color block. From the Default state, the arrows indicate conditional transitions. For these transitions boolean variables are created such as the *inTrigger_PerturbationX* and *inTrigger_PerturbationY* variables. When one of these variables gets the value "true", set by script, it transitions the state and plays the animation clip. When it becomes "false", it returns to the Default state. To illustrate an animation clip, Figure 36 shows the example of editing a simple animation that, for the duration of one second, rotates an object by 30 degrees on the x-axis, and returns to the starting position.

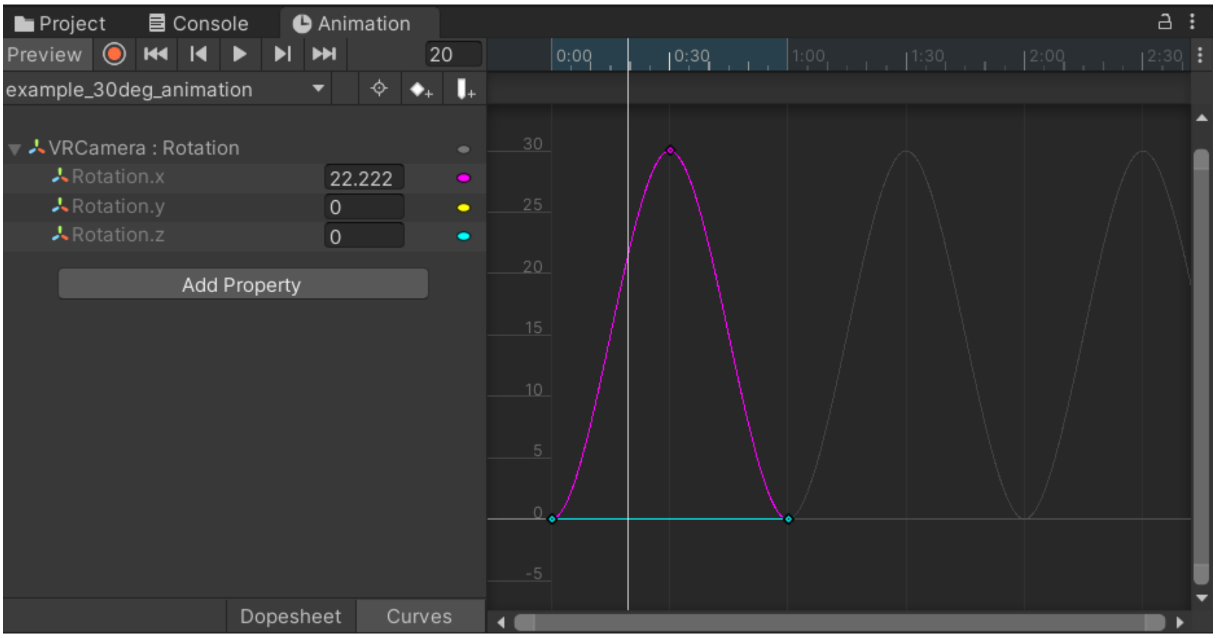


Figure 36: Example of an animation clip edit. When this animation is activated, it will apply a rotation on the x-axis of 30 degrees and return to the initial orientation in one second. This transformation is represented by the purple line.

After, the script that detects the collision between the Player and the box collider with trigger property is added to the gameObject with the collider component. This script has the function of activating the boolean variable that transitions state in the animator controller and thus starts playing the visual animation clip. Also, this script should save in the Unity Console Log a message with the exact time when the collision occurred and when it left the box collider, giving the perturbation onset time and its final time, along with the perturbation name.

Listing 1: Code handling the trigger entering.

```
[SerializeField] private Animator animator_name;
public GameObject trigger_name;
private bool onLog = false;

private void OnTriggerStay(Collider other)
{
   if (other.CompareTag(”Player”))

   {
      animator_name.SetBool(”inTrigger_PerturbationX”, true);
      if (!onLog)
      {
      UnityEngine.Debug.Log(”PerturbationX onset: ” + System.DateTime.Now.ToString((”HH:mm:ss:fff”)));
      onLog = true;
      }

   }
}
```

Using the *MonoBehaviour.OnTriggerStay(Collider)* class, where the parameter “other” refers to the Player, the boolean variable set in the animator controller is activated. In our example listed is the variable *in_Trigger_PerturbationX* which causes an animator controller state transition and makes the animation start.

Listing 2: Code handling the trigger exit.

```
private void OnTriggerExit(Collider other)
{
   if (other.CompareTag(”Player”))
   {
      roll_animator.SetBool(”inTrigger_PerturbationX”, false);
      UnityEngine.Debug.Log(”PerturbationX final: ” + System.DateTime.Now.ToString((”HH:mm:ss:fff”)));
      StartCoroutine(TimeDelay());
   }
}
```

OnTriggerExit is called when the collider "other"that is Player in practice, no longer has contact with the trigger. When this function is called, the boolean control variable takes the value zero, and the animation stops. The same happens for the time registration, making a calculation between the onset and the end, which gives us the duration of the disturbance.

Listing 3: 30 seconds delay script.

```
IEnumerator TimeDelay()
{
    //UnityEngine.Debug.Log("Delay started");
    trigger.GetComponent<Collider>().enabled = false;
    yield return new WaitForSeconds(30);
    //UnityEngine.Debug.Log("Timer gone");
    trigger.GetComponent<Collider>().enabled = true;
}
```

A 30 second delay was applied that deactivates the collider component of the gameObject so that the participant after suffering the visual disturbance can turn around and return to the starting point by passing over the trigger without it having any effect. The visual perturbations that have a different control flow are those corresponding to translations in the AP plane in both directions. The logic used does not activate a boolean variable. When the participant hits the trigger, the Player is affected of a position transformation every frame, giving the illusion that the participant is floating, at a speed of $1m/s$. Below is shown an excerpt of code that controls this perturbation. The variable "speed"is set to a value of 1 and the "target"is the target location at the end of the corridor. This time a delay is not used. Instead, the gameObject that has the trigger is destroyed.

Listing 4: AP perturbations script.

```
float step = speed * Time.deltaTime; // calculate distance to move
myPlayer.transform.position = Vector3.MoveTowards(myPlayer.transform.position, target.
    ↪ transform.position, step);
if(myPlayer.transform.position == target.transform.position)
    {
    Destroy(trigger);
    }
```

Finally, to conclude the explanation of VE creation and the method of introducing the perturbations, a description will be given of the animations created to achieve the effect of the chosen visual disturbances from Table 6.

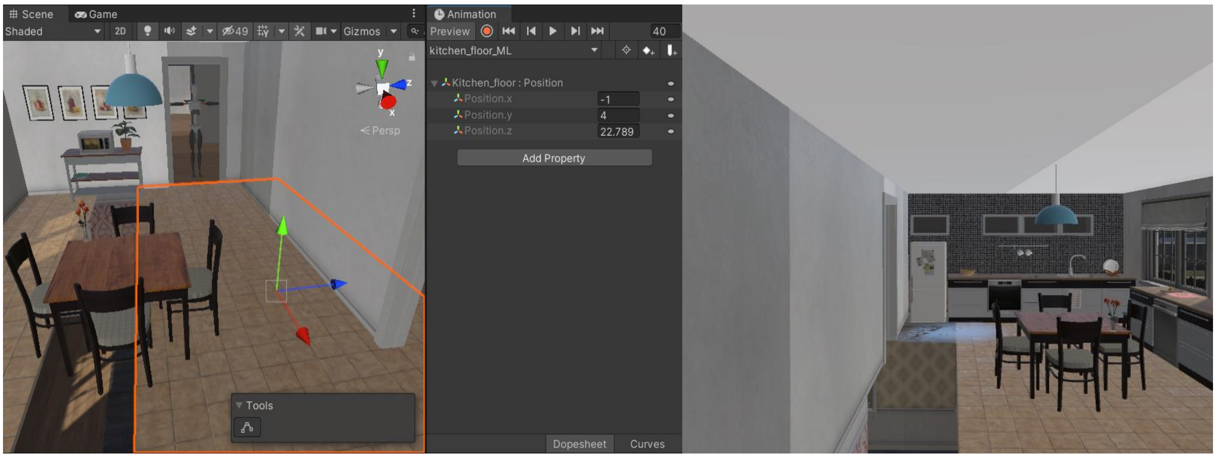


Figure 37: ML floor translation, continuous and bi-directional.

The visual translation perturbation (VP003) in the ML plane consists of an animation applied to the gameObject representing the floor. This animation is bi-directional, continuous, and has a speed of 1m/s. Figure 37 shows the floor being animated, as well as the first-person view that the participant will experience while undergoing this balance perturbation. For free fall (VP011), an animation is also used that transforms the position of the floor, sliding under the participant's feet and opening a void for the room below. In order for the avatar representing the participant's body to fall to the lower floor in the virtual environment, it is necessary to make the box collider that is part of the floor move with it. In the animation edit in Figure 38 this collider control is depicted.

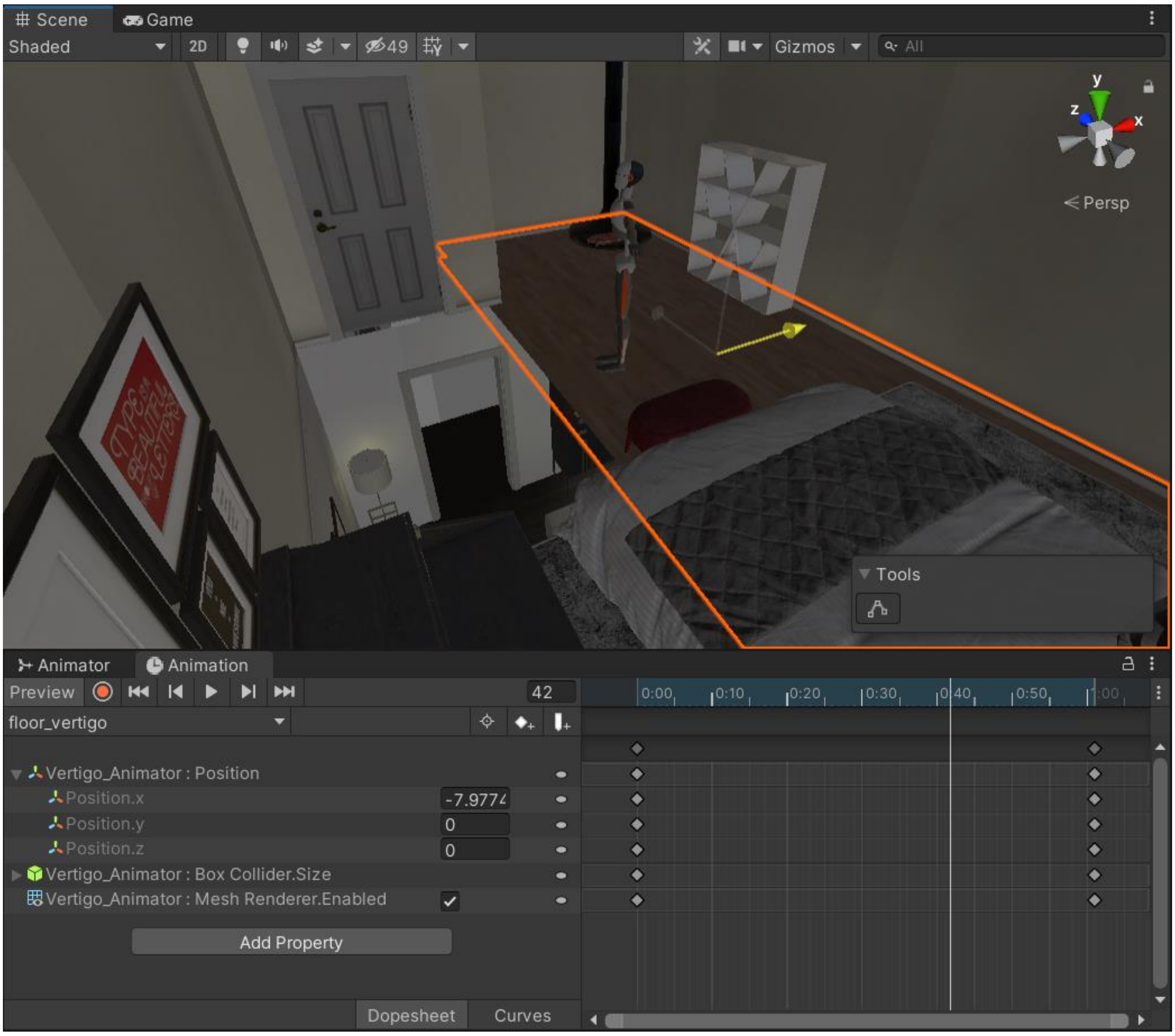


Figure 38: Scene view from the free fall perturbation setup.

The other animation that transforms gameObjects is the animation that corresponds to Object Avoidance (VP009). The bottles that are placed on the virtual floor start flying against the participant's head, applying only a position and rotation transformation to the bottles. Figure 39 shows the animation clip only with transformations in the position and rotation of the bottles. The target of this transformation is the avatar's head.

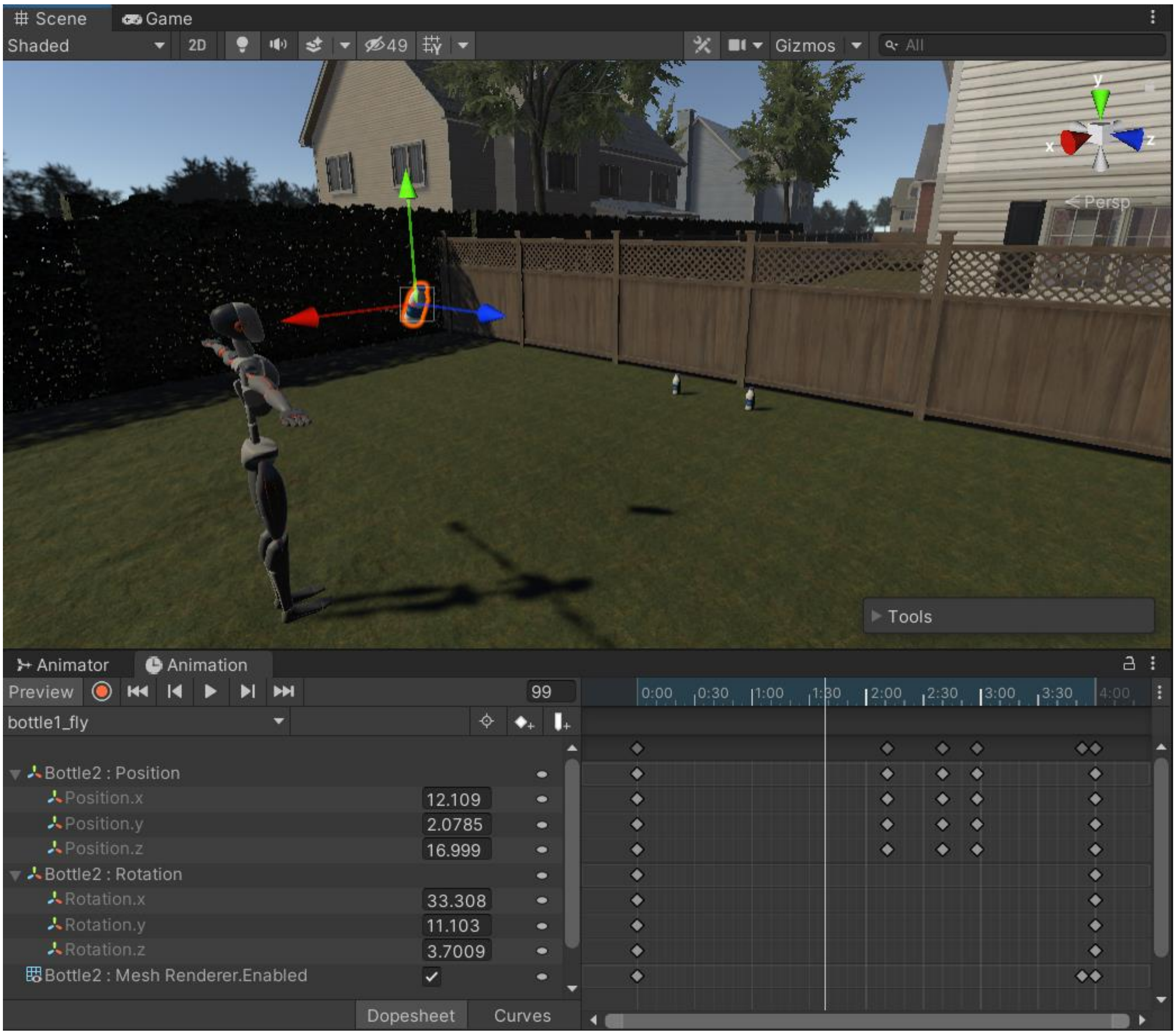


Figure 39: Object avoidance clip animation setup.

There is no animation governing the AP translation but with the movement of the whole Player, avoiding the camera moving and the participant having the view of his own static body while feeling himself moving in the AP plane. The Player's motion is made by script, as has already been described. One more rotation of the VRCamera is accomplished by animation, the pitch rotation. Like the roll rotation, only in one direction, with an amplitude of 25 degrees and a speed of 60 degrees per second.

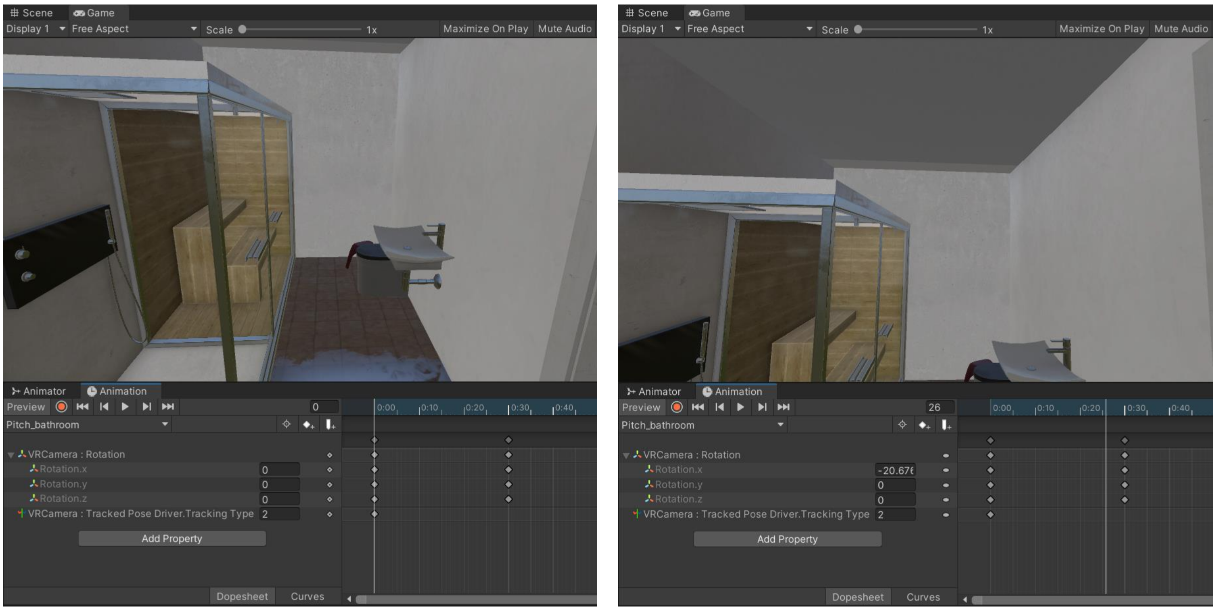


Figure 40: Animation of the pitch rotation. The example in the figure is depicted in one of the places that this disturbance can occur, in a bathroom. This animation is intended to simulate a slip.

In all animations that applied a camera rotation, the HMD tracking was turned off so that the participant's head movement would not overlap with the animation playing. The *TrackedPoseDriver* component applies the current pose value of a tracked device to the transform of the GameObject. In these cases, the rotation tracking type was disabled while the animation was running, leaving only the position tracking so that the participant's movement would be detected and consequently its trigger exit. This setting is visible in Figure 40, where the value of TrackingType is set equal to 2. In order for the subject to be placed in a position to access the trigger in a few steps, the participant is placed in a position about a unit of measurement that corresponds to one meter. As several studies prove, a person wearing a HMD takes a shorter stride length than they do in a comparable real world condition [236]. On the other hand, more recent studies evidence that gait parameters such as stride length can be quantitatively equivalent in a VE and in the real world [237, 238]. The script responsible for placing the Player in the defined positions consists of a series of conditions that check whether a given keyboard key has been pressed. If it was pressed, it looks for the gameObject that has the information of the position and rotation that the Player must be placed and assigns it these transformations. In the code snippet of the lines below, an example can be seen for the 'I' key:

Listing 5: Excerpt of code that places the participant in the start position, which will correspond to a distinct location for each keyboard key pressed.

```
if (Input.GetKeyDown(KeyCode.I))
      {
         Vector3 position = this.transform.position;
         Quaternion rotation = this.transform.rotation;

```

```
        position = GameObject.Find("Position_Roll_CW20_in2").transform.position;
        rotation = GameObject.Find("Position_Roll_CW20_in2").transform.rotation;

        this.transform.position = position;
        this.transform.rotation = rotation;
    }
```

The ”this”attribute refers to the gameObject that the script is attached to, the Player. When the key is pressed, the Player becomes oriented toward the trigger, about one meter or a mean step away from entering the box collider. In the case of vertigo situations, it is placed in the high-height locations.

4.5 Project Overview

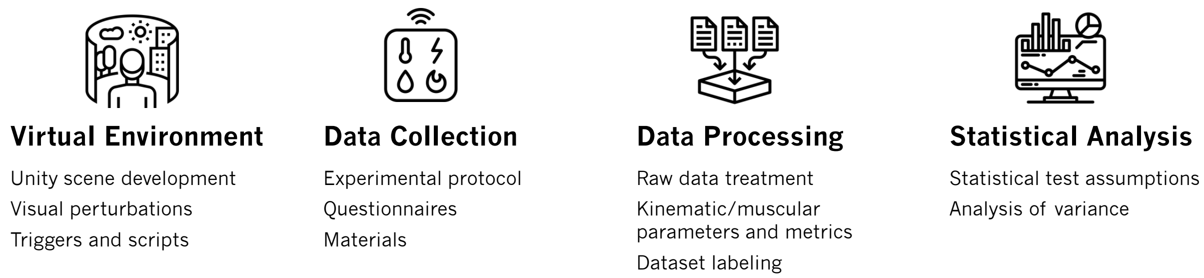


Figure 41: Project Overview: essential phases.

The strategy description used to approach a solution to the posed problem, it was initiated by conceptualizing and creating a virtual environment as a necessary medium to automatically introduce visual perturbations. Depending on this necessary implementation, the project comprises four main phases: i) virtual environment design and implementation; ii) visual perturbation protocol elaboration for multisensory data collection; iii) data processing and assembling into a dataset; and iv) statistical analysis. The virtual environment design has been discussed in detail. From conceptualization to construction, the software that enabled the development of the virtual environment was briefly explored. In addition, Unity allowed the development of the strategy to introduce disturbances at moments that will be dictated by the experimental protocol: execution of scripts when a trigger is detected, which starts an animation. The data collection through the experimental protocol will record kinematic, muscular and electrodermal data of the participants in response to the perturbations introduced. The explanation of this step will be provided in Chapter 5, where the systems used will be detailed. The systems are both the virtual reality setup and the sensor systems: i) motion capture system; ii) electromyography and iii) galvanic skin response. Each of these sensors will collect data in a synchronized manner. The raw data that arises and its processing will be explained. The whole process from the collected metrics to the construction of a dataset with this information from the three systems will be described. Additionally, the labelling of the pertrubations on the dataset is illustrated. Finally, in Chapter 6, a statistical analysis is carried out, which has to follow a set of assumptions. These assumptions are verified and statistical methods are applied to answer the research questions posed in Chapter 1, section 1.3.2.

Chapter 5

Materials and Methods

The collection of meaningful human motion data while dealing with visual perturbations caused by a HMD is essential to answer the proposed RQs in Chapter 1. In this regard, a protocol was designed and conducted to induce imbalance on the participants using visual perturbations while extracting relevant information from kinematic and physiological sensors. The collection of data regarding compensatory reactions while dealing with balance disturbances is important in balance control analysis, in the design of balance training protocols and, even more importantly, for compiling collection of data from various sensors into a dataset. In Chapter 2 and Chapter 3, several balance training intervention protocols were identified that disturbed the participant through visual perturbations. Two limitations found were the lack of fusion of inertial sensors with neurophysiological and biological sensors, and the introduction of only one visual perturbation in the protocol. With these limitations in mind, the protocol suggested in this chapter attempts to overcome the shortcomings by using various types of sensors and introducing a varied list of perturbations, in a fully immersive virtual environment, described in Chapter 4. This protocol will allow these perturbations to be introduced randomly, in a virtual environment endowed with ecological validity that aims to study compensatory reactions that are expected to resemble those induced by a real-world fall.

Recalling the scarcity of datasets with data on real-world falls and the time-consuming collection process that inherently requires an actual fall on a properly instrumented individual, the aforementioned virtual environment will play a key role in collecting the compensatory reactions in the experimental protocol. Many authors achieve this through physical perturbations. Others choose to simulate falls in a controlled laboratory environment [16, 239]. After the collection, the main goal that this project proposes to achieve is completed, which is to build a dataset while investigating the possibility of using visual disturbances to induce near-fall reactions.

5.1 Participants and Equipment

Twelve healthy young subjects (age: 25.09 ± 2.81; height: 167.82 ± 8.40 cm; weight: 64.83 ± 7.77 kg; males = 6; females = 6) were enrolled in the experimental protocol.

To participate in these trials, healthy subjects were chosen in a amount of more than ten subjects. The more specific inclusion criteria for these healthy subjects are: i) healthy locomotion; ii) No balance loss; iii) greater than 18 years of age; iv) body mass below 135 kg; and v) subjects have to voluntarily accept in the data acquisition experiment by signing an informed consent. The choice not to use the elderly was not based on their susceptibility to cybersickness. In fact, several studies show that the elderly suffer from less cybersickness than the youngest, and a higher presence attribute is noted [240–242]. However, aging causes neuromuscular degradations that decrease muscle strength, balance, proprioception and reaction time. Aging may be accompanied by adjustments in muscle activation such as a decrease in voluntary activation. This progressive decline in physical capacities reduces the ability of older adults to perform complex motor tasks and is associated with impaired mobility. For this reason, and to avoid the risk of a fall and injury, older adults were not used.

The volunteer subjects in this study would be excluded if they met any of these criteria: i) disease or deficit affecting locomotion; ii) epilepsy, vesibulopathy or other neurological condition resulting in potential instability during trials; iii) have recently undergone surgical procedures affecting mobility; iv) are included in another experimental protocol intervention; v) are under judicial protection/guardinship; and vi) have complications from using virtual reality with HMD (e.g., motion sickness). All subjects must be blinded to the protocol because otherwise it would introduce bias in anticipation to visual perturbations. For this reason, it was decided to present the visual perturbations at different locations and in a random order. In this way, the subjects never know when they will be disturbed and if they will be disturbed, strengthening the component of unpredictability and naturalism in the reactions.

5.2 Equipment and Sensors

This section describes the equipment used during the experimental protocol. The main equipment involved was the virtual reality headset, and then the sensor systems to capture the participants' gait, muscle contractions, and electrodermal activity. Emphasis is laid on the choice of sensor locations for the electromyography sensors. A safety harness was used in addition to this equipment. The harness system consisted in a vest that was attached to a structure in the ceiling through a rope. The length of the harness rope was adjusted in order to register a minimum of 15cm between the knees and the treadmill belt. This procedure was accomplished by asking participants to raise their feet, which led to the application of all the body weight into the harness system [243].

5.2.1 Virtual Reality Equipment

The kit used contains the headset, the controllers, and the base stations. This pack is illustrated in Figure 42. SteamVR Base Stations The base stations are responsible for head-tracking, which can change the participant's field of view with the movement of the head. The kit comes with two base stations, synchronized wirelessly. Precisely track the headset and the controllers in a 5 m^2 area. This head-tracking used technology is infrared beam emission. The technical specifications of this headset only are described in the table 8. The controllers were used only for setting up the virtual space. The participant did not hold the controllers during the protocol. The motion tracking system contained sensors in the hands, which replaced the commands in this home-living context.

Table 7: Headset technical specifications.

Headset Specs	
Screen	Dual AMOLED 3.5"diagonal
Resolution	1440 x 1600 pixels per eye (2880 x 1600 pixels combined)
Refresh rate	90 Hz
Field of View	110 degrees
Audio	Hi-Res certificate headset
Connections	Bluetooth, USB-C port for peripherals
Sensors	SteamVR Tracking, G-sensor, gyroscope, proximity, Eye Comfort Setting (IPD)

Figure 42: HTC VIVE Pro Full Kit. Images reproduced from [244]

5.2.2 Sensors

5.2.2.1 Motion tracking system

The motion capture system must be able to accurately record kinematic data and transport it in real time to the Unity software to have a full-body avatar representation in the virtual environment. This feature makes the experience endowed with a greater sense of presence, reinforcing the realism of the virtual environment and thus the naturalism of the postural reactions desired to provoke [245]. Using a markerless system with an RGB-D camera such as the Microsoft Kinect would have disadvantages because it would be affected by jitter, and inconsistent tracking, resulting in high latency. For an experimental protocol where the participant is moving most of the time, constantly changing their position relative to the capture device, several such devices would be needed. For this reason, this system was not used. Another available option would be to use a marker-based motion capture system from Optitrack. This system, while accurate, was not chosen because of the data collection place configuration and the possibility of the markers coming loose in a very dynamic activity. By exclusion, a motion capture system based on IMU was used. An IMU typically consists of an accelerometer, a gyroscope, and a magnetometer. Xsens released to the market in 2016 the second generation motion tracking system, the MTw Awinda with wireless sensors [246]. All motion trackers wirelessly transmit their data to the computer, via the Awinda Master (station or USB dongle) connected to a recording PC. The main MTw system components are shortly introduced: the MTw is a miniature IMU with a package size of 47mm × 30mm × 13mm and a weight of 16g. The MTw is designed to be robust, easy and comfortable in usage, with easy placement on the body based on flexible hook and loop straps.

Figure 43: An example of a participant using the MTw Awinda system [247].

Figure 44: a) MTw motion tracker (IMU); b) Awinda dongle; c) Awinda station; d) MTw body strap [246].

The accelerometer and gyroscope data are captured at a sampling frequency of 1000 Hz and low-pass filtered at a bandwidth of 184 Hz. This bandwidth is sufficient for motion analysis and guarantees high fidelity of the recorded signals. The available output data rates are provided in the document. 17 sensors were used, whose positions will be revealed by means of a diagram. Therefore, the maximum frame rate of the output data is 60 Hz. The user has access to different types of data: calibrated data - 3D acceleration, 3D angular velocity, 3D magnetic field, and 3D free acceleration; orientation data - 3D orientation per sensor. Orientation is provided in Euler representation, unit quaternions and rotation matrix. The sensors are placed in the recommended locations, and secured with Velcro strips, to minimize unintentional movement of the sensors. The recommended locations have been followed: the placement of the sensors at the extremities, for example forearm, is placed near the wrist, which provides more information about 3D motion. Also, the wrist area has less fatty tissue, lowering the chances of skin motion artifact; for the upper arm, it is recommended to place the sensor on the side of the upper arm, between the biceps and triceps muscle groups; on the lower leg, the literature describes two locations as good for placing the sensor. The lateral part of the lower leg was selected, aligned with the fibula, 6cm above the lateral malleolus; for upper leg measurements, it is recommended to place the MTw near the knee, on the lateral tigh, because it has less fat; for the pelvis sensor, the strap around the pelvis bone was tightened, at the height of the anterior superior iliac spine. Following these indications, Figure 45 shows a diagram with the location of the sensors.

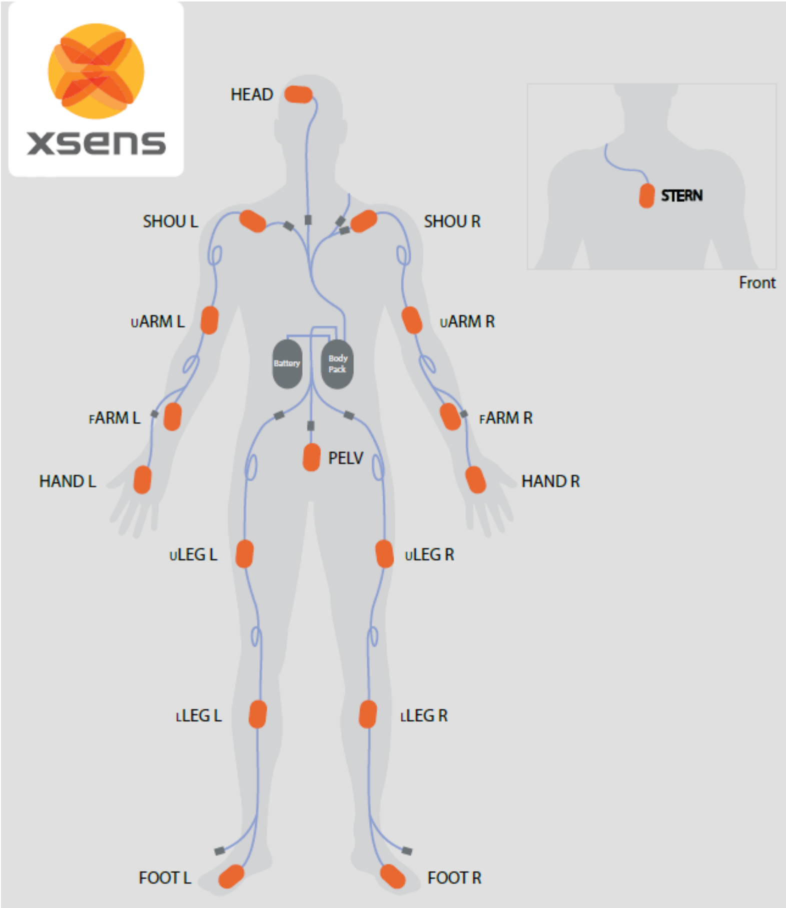


Figure 45: Sensor placement diagram, taken from [248].

5.2.2.2 Electromyography

Using kinematic measures to validate model output is an obvious and necessary step, but may be a very blunt instrument for validating neuromechanical systems. Kinematics alone is insufficient to distinguish different neural control strategies that result from different forces or patterns of muscle activation, potentially indicating different mechanisms of sensorimotor control. The measurements produced by muscle activation signals are representative, but are not equivalent to muscle tension. For this reason, one can make the mistake of trying to collect only kinematic information. Although good information of joint moments and powers is obtained, only the net moment created by all forces crossing the joint are obtained, and therefore the contribution of each muscle cannot be determined without additional information. This additional component is only accessible through EMG. Muscle activity as recorded through electromyography can provide important information as it represents amplified motor neuron pool activity and is also related to muscle force [249–251]. While it may be considered ideal to access neural signals directly, there are substantial limitations in current invasive techniques: they are intrusive, uncomfortable, painful and difficult to setup. There are two predominant forms of EMG measurement; surface and intramuscular EMG [252]. Non-invasive surface EMG is widely used for superficial, large, and easily accessible muscles.

This is the case for the muscles studied, detailed later in muscles item. In most studies, the surface myographic electrical signals are collected with Ag/AgCL electrode arrays arranged in a bipolar configuration and connected by cables to signal processing apparatus that are supported on a wearable. The connection made between the signal acquisition module and the receiving module can be made by cables, fiber optics or radio-telemetry. To obtain a temporal and spatial distribution description of multiple muscle activities, it is necessary to use multichannel recording. The decision between using surface EMG or internal EMG recording in gait analysis has to do with the accessibility of the target muscle by surface electrodes. If there is no other way, it is necessary to use fine-wire probes. Muscles such as the ilio-psoas and tibialis posterior are examples of locations inaccessible by surface electrodes. Further, the interference of indwelling electrodes on the gait pattern can be substantial, with an impact on step length, cadence and gait speed [253]. The Trigno Wireless Biofeedback System is a device designed to make electromyographic and biofeedback signal detection reliable and easy. The system transmits signals from Trigno Avanti sensors to a receiving base station using a time-synchronized wireless protocol that minimizes data latency across sensors. The core architecture of the Trigno System is designed to support high fidelity EMG signals, along with complementary biofeedback signals such as movement data, force signals, contact pressure events, timing, and triggering information. Trigno Avanti sensors support a low noise, high fidelity sensing circuit for detecting electromyographic biofeedback signals from the surface of the skin when muscle contract. Sensor bandwidth is selectable between 10-850Hz and 20-450 Hz and the input range of the sensor can be selected between 22mV or 11mV depending on user needs.

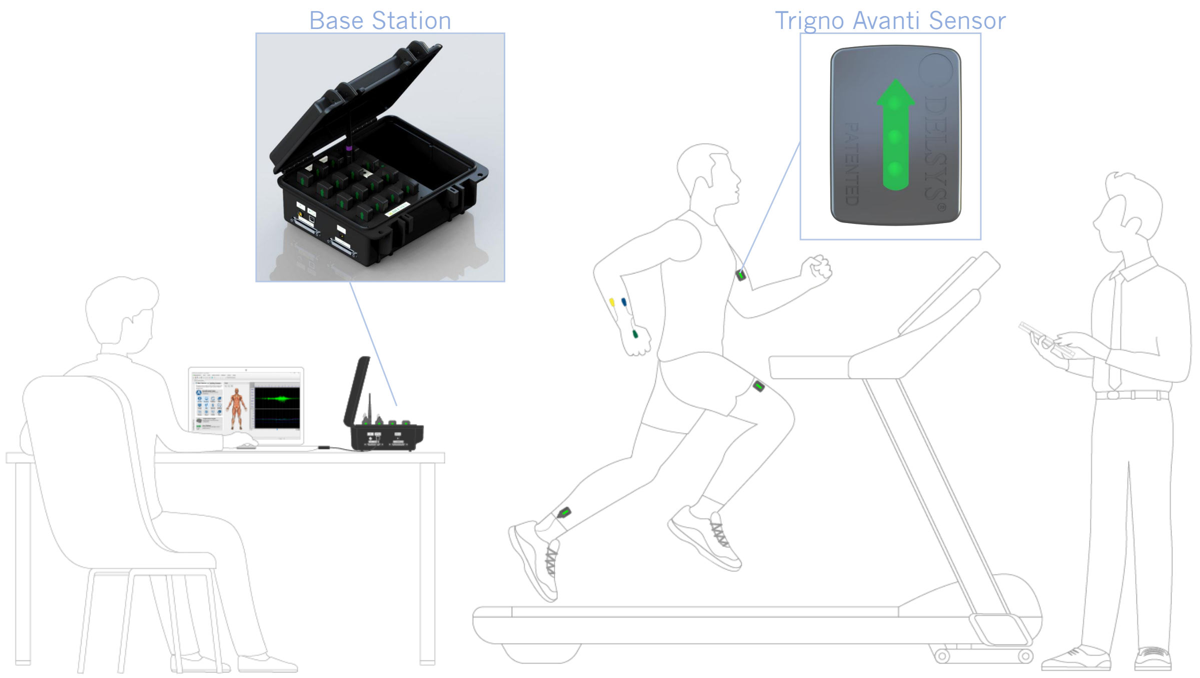


Figure 46: Overview of the system used, with an application example. The EMG sensor and the base station are highlighted. Images reproduced from [254]

After presenting the studies that used electromyography to interpret postural reactions provoked by visual or visual and physical disturbances in Chapter 3, a diagram was elaborated with the indication of the muscles used, grouped by type of disturbance. The color code corresponding to the perturbations is located at the bottom of the diagram. Since the available number of sensors is 8, the choice fell on placing sensors in those muscles that, according to the literature review, are the most comprehensive with respect to the various types of disturbance chosen. Therefore, in order to collect muscle activation data illicited by most compensatory reactions, the following muscles were chosen: i) Tibialis Anterior (both legs); ii) Gastrocnemius Medial Head (both legs); iii) Semitendinosus; iv) Rectus Femoris; v) External Oblique; and vi) Sternocleidomastoid.

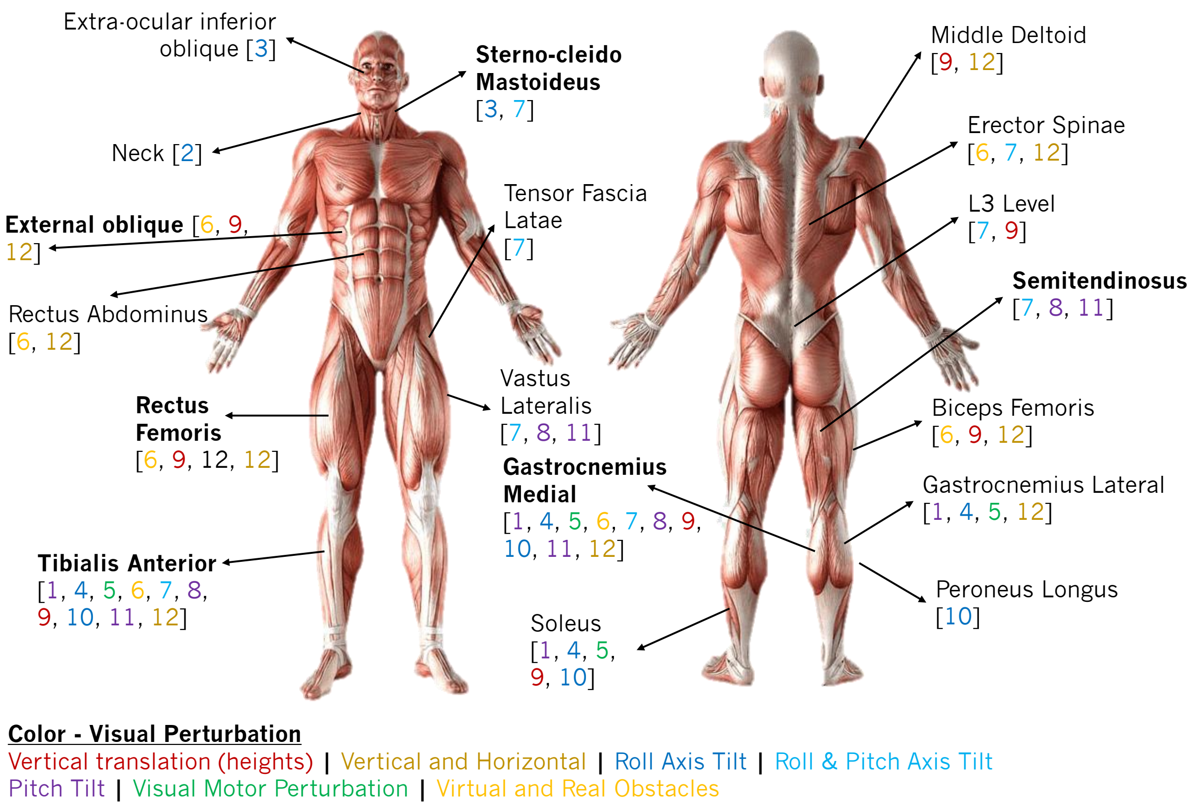


Figure 47: 1 - Sun et al. 2019 [168]; 2 - Peterson et al. 2018 [185]; 3 - Chiarovano et al. 2018 [188] ; 4 - Mohebbi et al. 2020 [178]; 5 - Drolet et al. 2020 [166]; 6 - Ida et al. 2017 [172]; 7 - Bugnariu and Fung 2007 [156]; 8 - Liu et al. 2015 [186]; 9 - Cleworth et al. 2016 [169]; 10 - Peterson and Ferris 2018 [183]; 11 - Parijat et al. 2015 [130]; 12 - Porras et al. 2021 [255].

5.3 Balance Perturbation Protocol

To test the effect of visual perturbations on biomechanical variables that characterize loss of balance, or in other words, that induce compensatory postural reactions, an experimental protocol was developed to introduce the perturbations and to collect data. The design of this experimental protocol is intended to achieve a level of control that will not allow for other interpretations about its success or failure. Virtual reality is a technology that offers advantages in the control and reproducibility of the experiment. Four fundamental components in the design of the experimental protocol are brought together, namely: i) stimulus control; ii) experiment reproducibility; iii) ecological validity and iv) real-world learning transfer.

Before developing the experimental protocol, a survey of articles using VR via HMD and introducing visual disturbances was conducted in Chapter 3. Information was extracted about the virtual environments and what types of visual disturbances were presented, the timing of the stimuli onset and their duration. An uncontrolled trials type study was designed. To this end, a list of tasks to be carried out with each participant was drawn up, visible in the "Tasks"section.

5.3.1 Tasks

The experimental protocol was designed on top of performing sequential tasks. First of all, it is imperative to collect the user's demographic data: age, height, weight, gender and citizen card ID for methodological reasons and to report the study. The areas intended for the placement of the electromyography sensors were cleaned with alcohol in order to maximize the quality of the electrical signal relayed and to fix the sensors well to the skin. The participant was then asked to be instrumented with the sensors, identifying the chosen muscles: i) Tibialis Anterior; ii) Gastrocnemius Medial Head; iii) Rectus Femoris. The sensors were firmly glued to the skin. Three trials of Maximum Voluntary Contraction (MVC) were performed for each muscle for further normalisation of EMG envelope. Further, participants were equipped with the full body configuration of Xsens MVN Awinda wearable inertial system that collected data at 60 Hz, 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. Following the sensor placement, participants underwent the N-Pose calibration of the system. Lastly, participants also worn the Shimmer GSR device on the dominant forearm with the electrodes placed on the index and middle fingers. PPG sensor must also be placed on the extremity of the index finger. To ensure safety, the harness was placed on the participant. For safety reasons, the length of the harness rope is adjusted in order to register a minimum of 15cm between the knees and the floor. This procedure is accomplished by asking the participant to raise their feet, which will lead to the application of all the body weight into the harness system [243] Then, equip the participant with the HTC Vive Pro headset and advise the subject to ask for help whenever necessary, such as episodes of motion sickness. The subject must adjust the distance between the lenses and the distance between the face and the lenses for a proper and more comfortable usage. The subject may still have a period of habituation to the virtual reality device and the environment, performing a familiarization trial without perturbations while using the entire setup, instructing the subject to stay within the playing area restricted by a virtual blue boundary. The subject performed the following activities without perturbations: walk front, turn around and ending at the starting position (3 times). In order achieve synchronous data acquisition from all the sensor systems, Sync Lab Desktop App was used. This team-developed desktop application for Windows OS is capable of synchronously start and stop data collection from the above mentioned systems and save the collected data in the computer that runs the app. The trigger signals sent by the Desktop application are electronic or wireless pulses. The former are either sent via Syncbox, which is a team-developed hardware interface that connects to the Xsens and Delsys systems or by direct USB communication, which is used to connect to both Kinect and Optitrack cameras. The wireless communication with Shimmer GSR system is performed directly from the computer running the app. Most importantly, the subjects performed the virtual perturbations available on 6. Each perturbation from Table 6 must be performed three times during the entire experimental protocol. These perturbations must be performed sequentially with a different order for different subjects. A Matlab script will generate randomly the virtual perturbations and non-perturbations introduction order. At the end, participants were asked to fill out a questionnaire about motion sickness during the experimental activity, the SSQ. Other questions will be asked to evaluate the immersion level and the effectiveness of the visual perturbations.

5.3.1.1 Questionnaires

The SSQ was used to assess the occurrence of cyber sickness episodes. To try to make a subjective measurement of the sense of presence achieved by the participants and to assess the degree of immersion in the created virtual environment, Igroup Presence Questionnaire (IPQ) was used. To understand at what level the simulation experience was convincing, users have to be comfortable, engaged and experience the phenomenon of presence and place illusion. Assessing simulator sickness is widely adopted in VR research. In SSQ, each item is rated with the scale from none, slight, moderate to severe. Through some calculations, four representative scores can be found: nausea-related subscore (N), Oculomotor-related score (O), Dysorientation-related subscore (D) are the scores for the symptoms for the specific aspects. Total Score (TS) is the score representing the overall severity of cybersickness experienced by the users of virtual reality systems. The questionnaire and how the score is calculated is described in Appendix 1.

The IPQ items can be presented in question form and the response is based on a scale of the degree of agreement or disagreement with the questions or statements. Questionnaire is presented in Appendix 1.

5.4 Data Processing

Diagram of Figure 48 summarizes the procedure for data processing from the three separate systems until the final dataset is compiled for each participant.

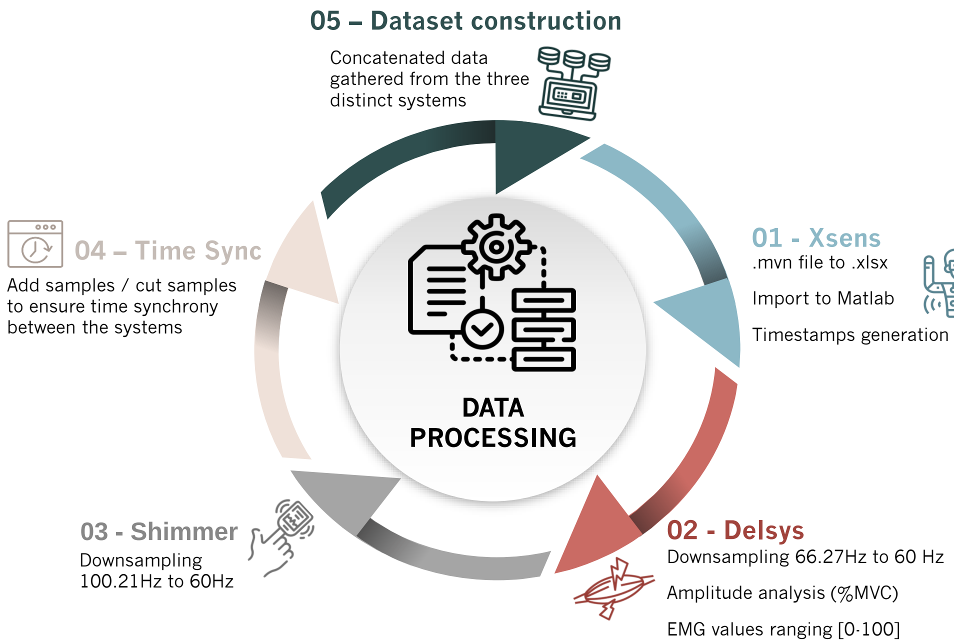


Figure 48: Data processing phases.

To explain which data to process, the raw data collected is reminded. From Xsens, at 60Hz, positional and accelerometry data from the IMUs. This system will serve as a reference so that the data collected from the other sensors will be resampled and presented with the same timestamp as Xsens. The system responsible for the temporal synchronization of the sensors' acquisition creates a text file called StartRecord_X with date, hours, seconds and milliseconds. The start and end time stamps serve as time boundaries for the first step: placing timestamps on the samples collected from Xsens, which will be the reference for all other data collected. After exporting the file saved in .mvn format to .xlsx, the following data is accessible. In each tab, the first column concerns the frame. Then, the orientation in quaternion and euler angles of the segments of the Pelvis, L5, L3, T12, T8, neck, head, right and left shoulder, upper arm, forearm, upper and lower leg, foot and toe. The segment position, linear and angular velocity, linear and angular acceleration in all three dimensions are also available. The Joint angles and ergonomic joint angles in reference ZXY and XZY, the center of mass, sensor free acceleration and magnetic field. Finally, sensor orientation in queternion and euler angles in three dimensions. All these kinematic features were saved in the dataset. In this initial phase, the available data was compiled from the excel file into Matlab table, adding the timestamp. This timestamp has a synchronizing role. All other systems will be time aligned with Xsens.

The Delsys system collected muscle data at a sampling frequency of 1111Hz. Before downsampling to the desired frequency of 60Hz, amplitude analysis was done with scripts built into the EMGworks software. This process is a normalization of the EMG data to MVC. This post-processing method uses a maximum RMS value from a recording to normalize subsequent EMG data series. The output is displayed as a percentage of the MVC (%MVC) value, which can be used to easily establish a common ground when comparing data between subjects. This normalized data is now downsampled in a Matlab script. Additionally, the script takes negative values as zero and values above 100% as 100%. This problem stems from the MVC trials recordings. In this situation, the participant may have done less force than they actually could. Although there are three trials and the maximum value is chosen from those three values, there may be variability because of posture and breathing [256]. These values between 0 and 100 were stored in the dataset for all 8 muscles.

Also, the GSR Shimmer data was downsampled from 100.21Hz to 60Hz. In case any system is delayed in starting the data record or finishing the recording, a Matlab script adds samples or cuts samples. After the necessary data processing to synchronize the data resulting from the three systems, the dataset was built concatenating the datasets of each system that were created and explained in the previous paragraphs. The Matlab script then joins the data together. It deletes the timestamps from all the systems and keeps only the Xsens one, which serves as a reference. The developed code identifies the size of the Xsens, Delsys and Shimmer samples. A state machine aligns the data if it is too much or too little. Finally, there is a concatenation of datasets, resulting in the final dataset. There is one dataset for each of the 12 subjects but at this stage, these data collections are unlabeled datasets.

5.4.1 Labelling

The following paragraphs will show how the labbeling of each dataset was done. Each perturbation was associated with a number. With the exception of the trip perturbation, which has an additional number to identify the foot strike. The undisturbed walking situations also have a label, as show in Table 9

Table 9: Label encoding.

Roll Indoor 1 CW10	**1**	Roll Indoor 2 CCW20	**11**	AP Axis Trans - Corridor Backward	**21**
Roll Indoor 1 CW20	**2**	Roll Indoor 2 CCW30	**12**	Pitch Indoor - Bathroom	**22**
Roll Indoor 1 CW30	**3**	Roll Outdoor CW10	**13**	Pitch Indoor - Fridge	**23**
Roll Indoor 1 CCW10	**4**	Roll Outdoor CW20	**14**	Window Roof Beam Walking - Vertigo	**24**
Roll Indoor 1 CCW20	**5**	Roll Outdoor CW30	**15**	Window Roof Beam Walking - Vertigo No Avatar	**25**
Roll Indoor 1 CCW30	**6**	Roll Outdoor CCW10	**16**	Simple Roof - Vertigo	**26**
Roll Indoor 2 CW10	**7**	Roll Outdoor CCW20	**17**	Simple Roof - Vertigo No Avatar	**27**
Roll Indoor 2 CW20	**8**	Roll Outdoor CCW30	**18**	Pitch Outdoor - Near Car Oil	**28**
Roll Indoor 2 CW30	**9**	ML Axis Trans - Kitchen	**19**	Trip - Sidewalk / Trip Shock	**29 / 290**
Roll Indoor 2 CCW10	**10**	AP Axis Trans - Corridor Forward	**20**	Bedroom Syncope	**30**
Garden - Object Avoidance	**31**	Electricity Pole - Vertigo No Avatar	**33**	Stairs	**35**
Electricity Pole - Vertigo	**32**	Free Fall	**34**		

The random sequence with the list of perturbations ordered chronologically served as a reference. For each perturbation that appeared in the sequence, one looked for the text that identifies it in the Unity log file. This Log marked the onset and end times of the disturbances that actually occurred, in the format HH:mm:ss.ms. When the string "onset"was identified, the associated datetime was removed. This instant of time was compared with the timestamps in the dataset. At the position that found the match, it was filled with the proper label until the end of the perturbation, with a similar process. It was necessary to make some corrections and manual entries of onsets and finals of perturbations. For this, the frames associated with the timestamps were useful for visual inspection in Xsens MVN software. Additionally, a text file was used that was filled in during the execution of the experimental protocol, which pointed out failures such as errors of the Unity software not activating the visual perturbations. This meant that the perturbation was recorded but the subject did not experience the perturbation. The scripted failure detection system would count the number of perturbations that were supposed to have occurred, detect whether there was a lack of information in an onset or an ending, and give the opportunity to manually enter the correct timing of the perturbation, to correctly label the dataset. The label plot shown in Figure 49 helped to correct flaws, as it gives an overview of the perturbations that occurred throughout the experimental protocol, whether there was overlap or if there were any perturbations of unusual duration. The horizontal axis represents the samples collected and the vertical axis represents the labels.

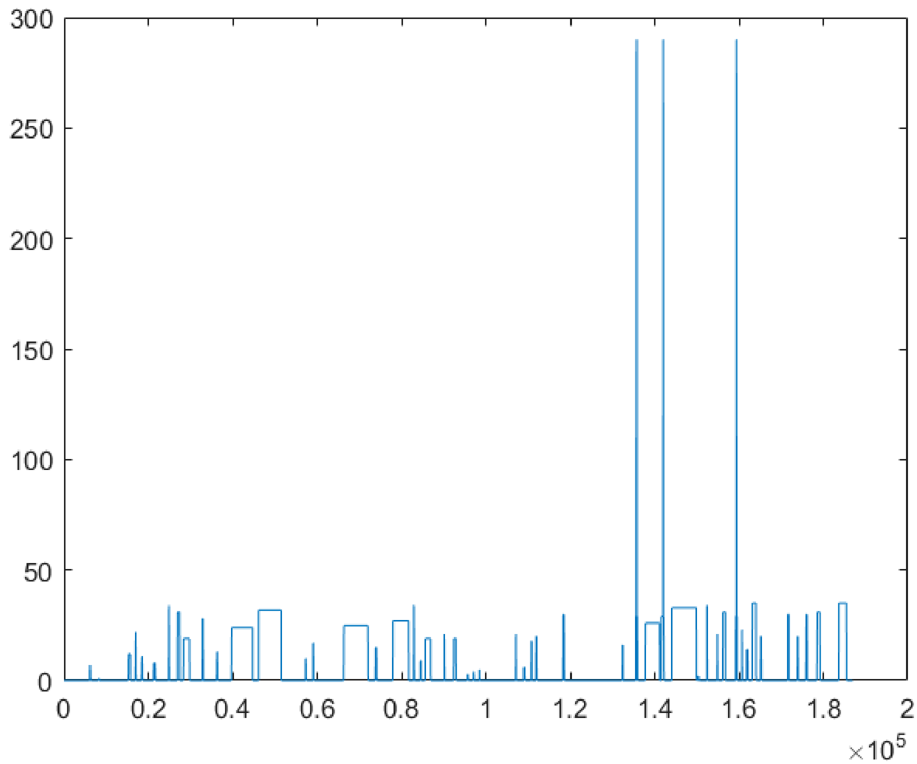


Figure 49: Ploted labels throughout the experimental protocol, for one subject. X-axis: Samples; Y-Axis: Labels.

Once the automatic labeling was done and the manual corrections and additions were completed, an error control script counted the occurrence of each disturbance, now represented by numeric labels. Figure 50 shows an example of identification of the trip disturbance by visual inspection.

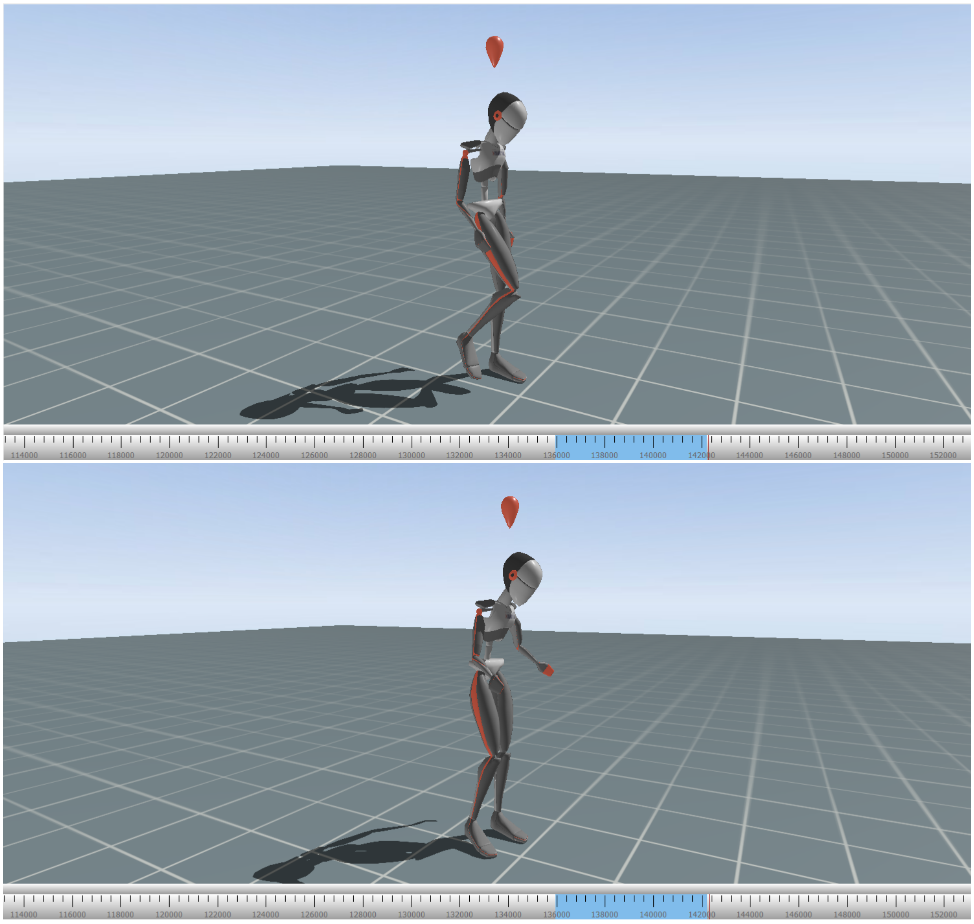


Figure 50: Visual inspection of the onset of the trip, annotation of the shock frame and subsequent detection of the end of the disturbance.

At this point, the 12 subjects datasets are fully labeled. In Figure 50 is an example of a transition from an undisturbed walking situation or neutral situation to a disturbance with label 33.

186943x1008 table

	1 Label	2 Timestamp_Xsens	3 Frame	4 Pelvisq0SegOQ	5 Pelvisq1SegOQ	6 Pelvisq2SegOQ	7 Pelvisq3SegOQ	8 L5q0SegOQ	9 L5q1SegOQ	10 L5q2SegOQ	11 L5q3SegOQ
144025	0	'12:41:08.958'	144229	0.9541	-0.0019	-0.0577	-0.2940	0.9580	0.0029	-0.0279	-0.2855
144026	0	'12:41:08.975'	144230	0.9541	-0.0018	-0.0576	-0.2940	0.9580	0.0026	-0.0277	-0.2855
144027	0	'12:41:08.992'	144231	0.9540	-0.0017	-0.0574	-0.2941	0.9580	0.0024	-0.0275	-0.2855
144028	0	'12:41:09.008'	144232	0.9540	-0.0016	-0.0573	-0.2942	0.9580	0.0022	-0.0274	-0.2855
144029	0	'12:41:09.025'	144233	0.9540	-0.0015	-0.0571	-0.2943	0.9580	0.0020	-0.0273	-0.2855
144030	0	'12:41:09.042'	144234	0.9540	-0.0014	-0.0569	-0.2944	0.9580	0.0018	-0.0272	-0.2855
144031	0	'12:41:09.058'	144235	0.9539	-0.0013	-0.0568	-0.2946	0.9580	0.0016	-0.0272	-0.2855
144032	0	'12:41:09.075'	144236	0.9539	-0.0012	-0.0566	-0.2948	0.9580	0.0015	-0.0273	-0.2856
144033	0	'12:41:09.092'	144237	0.9538	-0.0011	-0.0564	-0.2950	0.9580	0.0014	-0.0273	-0.2856
144034	0	'12:41:09.108'	144238	0.9538	-9.9850e-04	-0.0562	-0.2952	0.9579	0.0014	-0.0274	-0.2857
144035	0	'12:41:09.125'	144239	0.9537	-8.8401e-04	-0.0561	-0.2953	0.9579	0.0015	-0.0275	-0.2857
144036	0	'12:41:09.142'	144240	0.9537	-7.7342e-04	-0.0559	-0.2954	0.9579	0.0016	-0.0275	-0.2858
144037	0	'12:41:09.158'	144241	0.9537	-6.7064e-04	-0.0557	-0.2955	0.9579	0.0019	-0.0275	-0.2858
144038	0	'12:41:09.175'	144242	0.9537	-5.7176e-04	-0.0555	-0.2955	0.9579	0.0021	-0.0275	-0.2859
144039	33	'12:41:09.192'	144243	0.9538	-4.7395e-04	-0.0554	-0.2954	0.9579	0.0024	-0.0275	-0.2859
144040	33	'12:41:09.208'	144244	0.9538	-3.7401e-04	-0.0552	-0.2953	0.9579	0.0027	-0.0274	-0.2859
144041	33	'12:41:09.225'	144245	0.9539	-2.7515e-04	-0.0551	-0.2952	0.9579	0.0030	-0.0274	-0.2858
144042	33	'12:41:09.242'	144246	0.9539	-1.8418e-04	-0.0549	-0.2950	0.9579	0.0033	-0.0273	-0.2858
144043	33	'12:41:09.258'	144247	0.9540	-1.0429e-04	-0.0548	-0.2948	0.9579	0.0035	-0.0273	-0.2857
144044	33	'12:41:09.275'	144248	0.9541	-3.2293e-05	-0.0547	-0.2946	0.9579	0.0038	-0.0272	-0.2856
144045	33	'12:41:09.292'	144249	0.9541	3.4544e-05	-0.0545	-0.2943	0.9580	0.0040	-0.0272	-0.2856
144046	33	'12:41:09.308'	144250	0.9542	1.0161e-04	-0.0543	-0.2942	0.9580	0.0042	-0.0271	-0.2854

Figure 51: Labelled datset.

5.5 Discussion

Regarding the experimental protocol, there are two issues that should be highlighted that both serve to improve a future protocol and possible introduction of variability. First, the distances between the starting point of the gait and the trigger were fixed. An ideal situation would calculate at baseline the average stride lenght of the undisturbed gait and adjust this distance so that the participant would take about two steps before experiencing the perturbation. The second issue is related to the randomness of the introduction of the perturbations. Although a script was run that listed the perturbations randomly and did not allow for consecutive perturbations of the same type and intensity, some participants experienced the perturbations that were anticipated to be most severe at the beginning of the protocol and may have experienced a training effect for the next perturbations in the protocol. This presumption will be discussed when presenting the results of the statistical analysis. Overall, the protocol was run on schedule, generally with no failures in the introduction of the perturbations and no severe episodes of motion sickness. The data labeling processing phase deserves a brief note. Some of the visual perturbations did not write their beginnings or endings in the log. Therefore, it was necessary to manually check, frame by frame, the beginnings and ends of some visual perturbations: trip and object avoidance. A time-consuming process, requiring meticulous attention.

Chapter 6

Statistical Analysis

Given the amount of data collected and the observed differences in kinematic reactions in the presence of perturbations, the most cost-efficient way to analyze the effect of visual perturbations on inducing imbalances during gait was through statistical analysis. Moreover, the goal was not to detect gait abnormalities, but rather to detect whether the visual disturbance was an efficient way to introduce gait variability that denotes imbalance. To this end, several ANOVAs were conducted as a follow-up to an initial MANOVA. In these separate ANOVAs, the dependent variable could be kinematic, muscular or from the GSR. The independent variables were those representing perturbation and those representing non-perturbation. Therefore, will be statically inferred if there were significant differences in the metrics evaluated when subjected to visual disturbance. Further on, using post-hoc tests, attribute these differences created to specific disturbances. These results allow us to answer the research questions: i) can a virtual reality headset introduce imbalances through visual perturbations? Can they cause postural reactions typical of a fall?; and after the multiple comparisons ii) which visual perturbation challenged the participants' balance the most? This chapter is organized in order to explain the statistical analysis that was employed, the statistical variables created for this purpose, and the assumptions that must be verified in the data in order to use this analysis of variance tool. The subsequent section presents the results obtained, and finally, a discussion of these results is presented.

6.1 Multivariate Analysis of Variance

Tom Chau [257], in a review of gait data analysis methods, found that statistical methods are the most applied and understood. Analysis of quantitative gait data traditionally encounters challenges such as: i) high-dimensionality - the dataset contains kinematic, kinetic, muscular and metabolic variables. New and standardized methods are needed that better summarize massive gait time series, to allow for quantitative identification of important correlations such as perturbed versus undisturbed gait; ii) time dependence - for computational feasibility, time curves are usually parameterized by mean values or peak amplitudes. By doing so, explicit time dependence is ruled out, possibly along with valuable time-dependent patterns; iii) high variability - gait recordings exhibit intrasubject, intersubject, and between-trial variabilities, as well as variability caused by instrumentation. For example, poor placement of EMG sensors in less accessible areas can introduce this variability. Along the trial, there may be an accidental dislocation of either the EMG sensor or an IMU; iv) nonlinear relationships - the nonlinear dynamics intrinsic to human movement causes gait variables to interact in a complex nonlinear manner. To address these challenges, researchers have sought new ways to manipulate and interpret gait data: fuzzy, multivariate statistical analysis and fractal analyses [**Phinyomark2018**]. The statistical approach is based on multivariate statistical analysis, followed by multiple ANOVAs to answer the research questions. To perform this analysis, statistical measures of three groups of variables were calculated: muscular, kinematic (accelerometry, gyroscopy, and center of mass velocity), and electrodermal. For each of the dependent variables, the mean, standard deviation, minimum and maximum values, kurtosis and skewness values, and Lyapunov exponents were calculated.

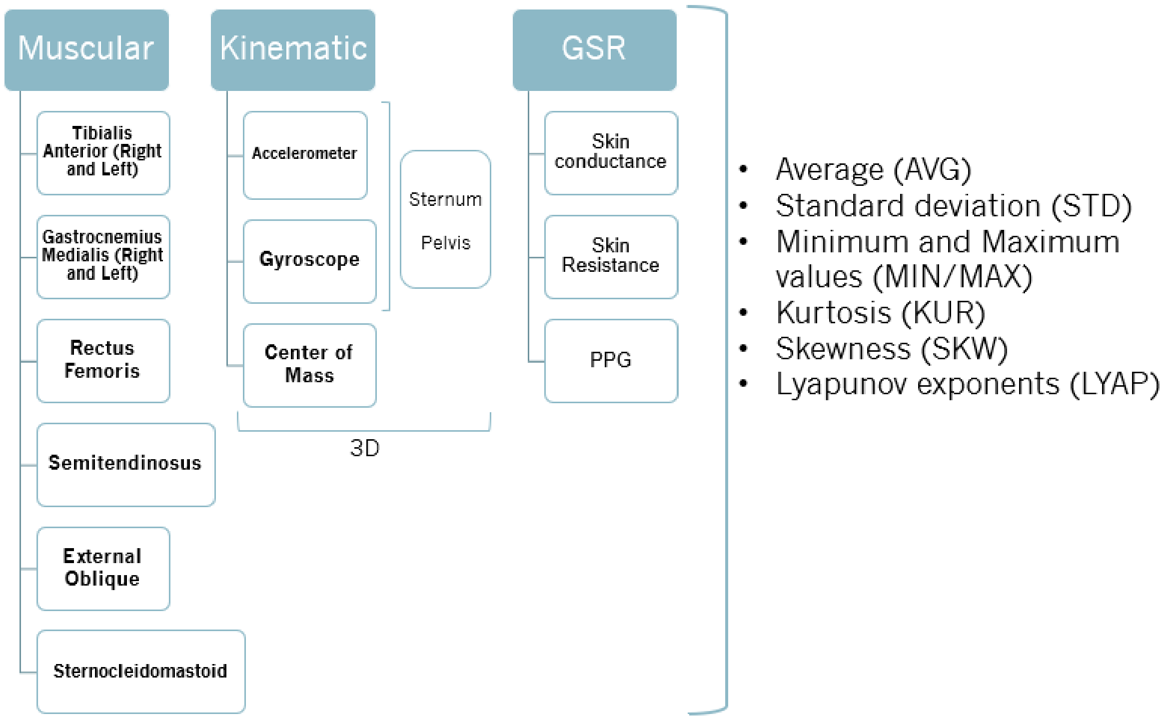


Figure 52: Statistical metrics computed for each of the chosen parameters: muscular, kinematic, and electrodetric.

This division into variable groups and subgroups is summarized in the diagram in Figure 52. The variables calculated for the 8 muscles were used, and from the electrodermal sensor the skin conductance, skin resistance and PPG. From the kinematic variables, those that are representative of sternal and pelvis acceleration and rotation were chosen, as they reflect trunk bend that may be a consequence of hip strategy, body sway, and trunk rotation [258]. CoM velocity was also chosen for obvious reasons, since CoM is the determining factor of balance [259]. The diagram from Figure 53 displays in more detail, the example of muscle variables and the chosen notation. The chosen metrics were divided into two categories: the variables measured under perturbation and those without perturbation occurring when the label has value zero. Figure 54 demonstrates the data view in SPSS software for the subject with ID=1. It is possible to see the encoding of the labels that correspond to undisturbed gait - BinaryLabel = No Perturbation/Perturbation.

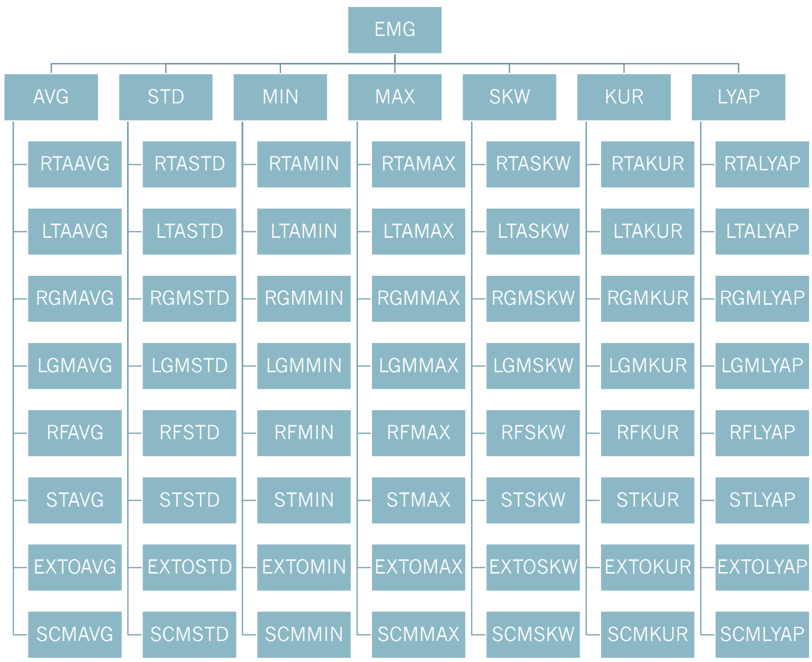


Figure 53: Key: RTA - Right Tibialis Anterior, LTA - Left Tibialis Anterior, RGM - Right Gastrocnemius Medial Head, LGM - Left Gastrocnemius Medial Head, RF - Rectus Femoris, ST - Semitendinosus, EXTO - External Oblique, SCM - Sternocleidomastoid.

	SubjectID	Label	BinaryLabel	RTAAVG	LTAAVG
1	1	No Perturbation	No Perturbation	4.478325908594441	7.796642854615499
2	1	Pitch Indoor - Fridge	Perturbation	8.977173813604805	9.804279414749479
3	1	Pitch Outdoor - Near Car Oil	Perturbation	4.199872739316477	9.116580734714804
4	1	No Perturbation	No Perturbation	6.341762672302301	9.179859582282228
5	1	Trip - Sidewalk	Perturbation	6.310912292626946	13.217460659130790
6	1	Pitch Indoor - Bathroom	Perturbation	5.482354665474841	7.066260442678666
7	1	Trip - Sidewalk	Perturbation	7.106970481204554	13.788438549536280
8	1	No Perturbation	No Perturbation	3.934568007205269	7.133397483563507
9	1	Roll Indoor 1 CCW20	Perturbation	8.049516380323944	10.122752226047773
10	1	AP Axis Trans - Corridor Forward	Perturbation	3.406052169586131	6.874530361344246
11	1	AP Axis Trans - Corridor Backward	Perturbation	3.705652481693888	7.535315697910128
12	1	AP Axis Trans - Corridor Backward	Perturbation	3.419800090369880	7.234557300990201
13	1	Trip - Sidewalk	Perturbation	11.056449649221086	10.089884311813282
14	1	Bedroom Syncope	Perturbation	6.824509396102412	9.807793076329862
15	1	Bedroom Syncope	Perturbation	6.855292075004845	11.493512365120680
16	1	No Perturbation	No Perturbation	3.666522198462367	7.390704161837284
17	1	Roll Indoor 2 CCW20	Perturbation	6.211731736973355	9.329882677571149

Figure 54: Data view: SubjectID column refers to the idenfification of the subject; the Label column has the name of the visual perturbations; BinaryLevel column aggregates all perturbation and no-perturbation labels, making a binary partition. The following columns have the values of the dependent variables.

6.1.1 Assumptions for Multivariate Analysis of Variance

MANOVA and ANOVA requires that the observations are independent, the response variables are normally distributed, and the covariance matrix of the response variables is homogeneous across groups. When the assumptions of normality and homogeneous covariance matrices are not met, past research has shown that the type I error rate of the standard MANOVA test statistics can be inflated while their power can be attenuated [260]. In addition, multivariate outliers' existence was examined, the absence of multicollinearity, and linearity among the dependent variables for each level of the independent variables.

6.1.1.1 Normality

The standard normal distribution is the most important continuous probability distribution. Has a bell-shaped density curve described by its mean and Standard Deviation (SD). In this distribution, extreme values in the data set have no significant impact on the mean value. If a continuous data follows normal distribution, then 68.2%, 95.4%, and 99.7% observations are lie between mean ± 1 SD, mean ± 2 SD, and mean ± 3 SD, respectively [261]. With large enough sample sizes (> 30 or 40), the violation of the normality assumption should not cause major problems [262]. This implies that parametric procedures can be used even when the data are not normally distributed [263]. If the samples consist of hundreds of observations, the distribution of the data can be ignored [261]. According to the central limit theorem, if the sample data are approximately normal then the sampling distribution too will be normal; dealing with large samples (> 30 or 40), the sampling distribution tends to be normal, regardless of the shape of the data [264]; Although for meaningful conclusions, assumption of the normality should be followed irrespective of the sample size [265].

There are various methods available to test the normality of the continuous data. The most popular methods are the Shapiro–Wilk test [266], Kolmogorov–Smirnov test, skewness, kurtosis, histogram, box plot, P–P Plot, Q–Q Plot, and mean with SD. The two well-known tests of normality, namely, the Kolmogorov–Smirnov test and the Shapiro–Wilk test, are the most widely used methods to test the normality of the data [267]. Normality tests were conducted in the statistical software SPSS. The Shapiro–Wilk test is a more appropriate method for small sample sizes (<50 samples), although it can also handle a larger sample size while the Kolmogorov–Smirnov test is used for n≥50. The null hypothesis states that data arises from a normally distributed population for both of the above tests. When p>0.05, the null hypothesis is accepted, and data is normally distributed. For convenience and ease of interpretation, Table 10 shows only two variables. The one colored red has a non-normal distribution (p<0.05). The blue one, a normally distributed variable (p>0.05). A distribution is called approximate normal if skewness or kurtosis of the data are between -1 and $+1$. Although, this is a less reliable method in the small-to-moderate sample size, i.e., n<300, because it cannot adjust the standard error. To overcome this problem, a z-test is applied for normality test using skewness and kurtosis. A z-score could be obtained by dividing the skewness values or excess kurtosis value by their standard errors. For small sample sizes (n<50), z-value ± 1.96 is sufficient to establish normality of the data. However, medium-sized samples (50≤n<300), at absolute z-value ± 3.29, conclude the distribution of the sample is normal. For sample size n>300, the normality of the data is dependent on the histograms and the absolute values of skewness and kurtosis. Either an absolute skewness value ≤2 or an absolute kurtosis ≤4 may be used as reference values for determining considerable normality [268]. z-value is calculated by the expression $\text{Z-value} = \frac{\text{Skewness/Kurtosis}}{\text{std. error}}$.

Table 10: Tests of normality

Tests of Normality

BinaryLabel		Kolmogorov-Smirnov			Shapiro-Wilk		
		Statistic	df	Sig.	Statistic	df	Sig.
RTAAVG	No Perturbation	0.043	553	0.015	0.989	553	0.001
	Perturbation	0.067	543	0.000	0.968	543	0.000
RTASTD	No Perturbation	0.026	553	0.2	0.997	553	0.301
	Perturbation	0.082	543	0.000	0.954	543	0.000

Table 12 show the normal variables following the rule for sample size over 300 (skewness value ≤2 or kurtosis ≤4). This table shows the example of the muscle variables that follow normal distribution either in perturbation or non-perturbation. Those in bold are those that do not obey the normal distribution. All the others do. Due to variables extension, only muscle variables were reproduced in the table. From this point on, the metrics evaluated will only be the average (AVG), standard deviation (STD), and minimum and maximum (MIN/MAX) values. From here on the statistical effect of Skewness, Kurtosis and Lyapunov exponents are no longer analyzed.

Table 12: EMG normal variables. No perturbation and with perturbation. The variables in bold represent a non-normal distribution.

EMG	SK<=2	K<=4	SK<=2	K<=4
	No Perturbation	Visual Perturbation	No Perturbation	Visual Perturbation
	RTAAVG	RTAAVG	RTAMIN	RTAMIN
	LTAAVG	LTAAVG	LTAMIN	LTAMIN
	RGMAVG	RGMAVG	RGMMIN	RGMMIN
	LGMAVG	LGMAVG	LGMMIN	**LGMMIN**
	RFAVG	RFAVG	RFMIN	RFMIN
	STAVG	STAVG	STMIN	STMIN
	EXTOAVG	**EXTOAVG**	EXTOMIN	**EXTOMIN**
	SCMAVG	SCMAVG	SCMMIN	SCMMIN
	RTASTD	RTASTD	RTAMAX	RTAMAX
	LTASTD	LTASTD	LTAMAX	LTAMAX
	RGMSTD	RGMSTD	RGMMAX	RGMMAX
	LGMSTD	LGMSTD	LGMMAX	LGMMAX
	RFSTD	**RFSTD**	RFMAX	RFMAX
	STSTD	**STSTD**	STMAX	STMAX
	EXTOSTD	**EXTOSTD**	EXTOMAX	EXTOMAX
	SCMSTD	**SCMSTD**	**SCMMAX**	SCMMAX

The Skewness and Kurtosis values were inspected in Table 13. When it comes to inferences about associations between variables, as intended to do in analysis of variance, it is recognized that in large samples, these statistical methods rely on the Central Limit Theorem, which states that the average of a large number of independent variables is approximately normally distributed around the true population mean. Thus, violation of the normality assumption is not sufficient to use another type of statistical test [269].

Table 13: Skewness and Kurtosis absolute values and z-values for each muscular variable, with or without perturbation.

Descriptives			Statistic	std. error	z-value
RTAAVG	No Perturbation	Skewness	-0.200	0.104	-1.92462074
		Kurtosis	-0.277	0.207	-1.3332817
	Perturbation	Skewness	0.644	0.105	6.145592741
		Kurtosis	1.276	0.209	6.095852283
LTAAVG	No Perturbation	Skewness	0.162	0.104	1.562132743
		Kurtosis	-0.696	0.207	-3.353548154
	Perturbation	Skewness	1.130	0.105	10.7821742
		Kurtosis	3.313	0.209	15.8305142
RGMAVG	No Perturbation	Skewness	1.009	0.104	9.712540647
		Kurtosis	0.725	0.207	3.497357039
	Perturbation	Skewness	1.578	0.105	15.05613193
		Kurtosis	6.822	0.209	32.59831173
LGMAVG	No Perturbation	Skewness	1.815	0.104	17.46800409
		Kurtosis	3.329	0.207	16.05392318
	Perturbation	Skewness	1.755	0.105	16.74056563
		Kurtosis	3.260	0.209	15.57825029
RFAVG	No Perturbation	Skewness	-0.239	0.104	-2.296615259
		Kurtosis	-0.038	0.207	-0.184753307
	Perturbation	Skewness	1.889	0.105	18.02081296
		Kurtosis	5.004	0.209	23.91069225
STAVG	No Perturbation	Skewness	0.336	0.104	3.235309874
		Kurtosis	-0.139	0.207	-0.671756261
	Perturbation	Skewness	1.249	0.105	11.91366701
		Kurtosis	4.057	0.209	19.38650919
EXTOAVG	No Perturbation	Skewness	2.015	0.104	19.39606869
		Kurtosis	3.166	0.207	15.26556396
	Perturbation	Skewness	3.130	0.105	29.85476694
		Kurtosis	17.261	0.209	82.47762381
SCMAVG	No Perturbation	Skewness	-1.524	0.104	-14.67238138
		Kurtosis	2.654	0.207	12.79916042
	Perturbation	Skewness	-1.424	0.105	-13.58047507
		Kurtosis	2.612	0.209	12.48164025

6.1.1.2 Multivariate outliers

The detection of multivariate outliers relies on different methods than the detection of univariate outliers. Univariate outliers have to be detected as values too far from a robust central tendency indicator, while multivariate outliers have to be detected as values too far from a robust ellipse that includes most observations [270]. In order to detect multivariate outliers, most researchers compute the Mahalanobis distance

Table 15: Mahalanobis distance critical values.

Dependent variables	**Critical Value**
1	10.8275662
2	13.8155106
3	16.2662362
4	18.466827
5	20.5150057
6	22.4577445
7	24.3218863
8	26.1244816
...	...
182	246.695142

Table 16: Residuals statistics - Mahal. Distance

Residuals Statistics

	Minimum	Maximum	Mean	Std. Deviation	N
Predicted Value	-0.64	1.46	0.50	0.443	1096
Std. Predicted Value	-2.558	2.168	0.000	1.000	1096
Standard Error of Predicted Value	0.047	0.253	0.094	0.035	1096
Adjusted Predicted Value	-5.25	328.41	0.90	10.420	1096
Residual	-0.816	0.812	0.000	0.232	1096
Std. Residual	-3.227	3.211	0.000	0.918	1096
Stud. Residual	-3.525	3.811	0.002	1.013	1096
Deleted Residual	-328.412	6.251	-0.407	10.425	1096
Stud. Deleted Residual	-3.548	3.839	0.002	1.014	1096
Mahal. Distance	37.570	1093.998	172.842	156.170	1096
Cook's Distance	0.000	9696.091	9.771	294.142	1096
Centered Leverage Value	0.034	0.999	0.158	0.143	1096

[271, 272]. This method is based on the detection of values too far from the centroid shaped by the cloud of the majority of data points. To calculate the Mahalanobis distance, a linear regression was performed. This step adds a new column to the dataset with the Mahalanobis distance, called 'MAH_1'. The critical value was calculated with the formula CHISQ.INV, with the probability parameter having a value of 0.999 and the degrees of freedom corresponding to the number of dependent variables. Table 15 shows the critical values calculated for 182 dependent variables.

Table 17 is the SPSS output for linear regression residuals statistics. Mahalanobis distance is described by row "Mahal. Distance". To calculate the Mahalanobis probability the compute variable is used with the expression p_MAH = 1-CDF.CHISQ(MAH_1,182). If $p<0.001$ it is a significant outlier. 175 significant outliers are identified in a total of 207144 samples, therefore, can be neglected. Outliers descriptive statistical analysis was performed by subject, by label of disturbance, and by the binary perturbation or no-perturbation distinction. Figure 55 reproducted from SPSS indicates a presence of significant outliers mostly in the labels corresponding to the disturbances. More specifically, 56.6 percent of the significant outliers appear in the disturbances. The AP-Backward Translation perturbation (label = 21) and Trip (Label = 290) represent 7.4% of these outliers.

Label

		Frequency	Percent	Valid Percent	Cumulative Percent
Valid	0	76	43.4	43.4	43.4
	1	1	.6	.6	44.0
	2	1	.6	.6	44.6
	5	5	2.9	2.9	47.4
	6	2	1.1	1.1	48.6
	8	2	1.1	1.1	49.7
	9	1	.6	.6	50.3
	11	1	.6	.6	50.9
	12	3	1.7	1.7	52.6
	13	1	.6	.6	53.1
	15	1	.6	.6	53.7
	17	1	.6	.6	54.3
	18	2	1.1	1.1	55.4
	20	4	2.3	2.3	57.7
	21	13	7.4	7.4	65.1
	22	5	2.9	2.9	68.0
	23	7	4.0	4.0	72.0
	24	2	1.1	1.1	73.1
	25	1	.6	.6	73.7
	27	1	.6	.6	74.3
	28	7	4.0	4.0	78.3
	30	5	2.9	2.9	81.1
	31	9	5.1	5.1	86.3
	32	4	2.3	2.3	88.6
	33	2	1.1	1.1	89.7
	34	4	2.3	2.3	92.0
	35	1	.6	.6	92.6
	290	13	7.4	7.4	100.0
	Total	175	100.0	100.0	

Figure 55: Descriptive statistics of outliers identified in perturbation.

6.1.1.3 Multicolinearity

Multicollinearity occurs when one dependent variable is almost a weighted average of the others - are correlated. If the degree of correlation between variables is high enough, it can cause problems interpreting the results. The correlation can be calculated with the Pearson correlation coefficient. These numbers measure the strength and direction of the linear relationship between the two variables. The correlation coefficient can range from -1 to +1, with -1 indicating a perfect negative correlation, +1 indicating a perfect positive correlation, and 0 indicating no correlation at all. In SPSS, the calculation of Pearson's correlation is done with a bivariate correlation analysis by choosing the Pearson coefficients option. Table 19 exemplifies the Pearson correlation values for the first eight variables, corresponding to the averages of muscle activations.

Table 18: Pearson Correlation values for the muscle variables averages. AVG = Average; Key: RTA; LTA; RGM; LGM; RF; ST; EXTO; SCM.

		RTAAVG	**LTAAVG**	**RGMAVG**	**LGMAVG**	**RFAVG**	**STAVG**	**EXTOAVG**	**SCMAVG**
RTAAVG	Pearson Correlation	***1***	0.323	0.286	0.184	0.377	0.252	0.109	-0.071
LTAAVG	Pearson Correlation	0.323	***1***	0.271	-0.004	0.179	-0.048	-0.187	0.198
RGMAVG	Pearson Correlation	0.286	0.271	***1***	0.243	-0.061	0.171	-0.097	0.174
LGMAVG	Pearson Correlation	0.184	-0.004	0.243	***1***	-0.164	0.123	0.697	0.314
RFAVG	Pearson Correlation	0.377	0.179	-0.061	-0.164	***1***	0.115	0.039	-0.423
STAVG	Pearson Correlation	0.252	-0.048	0.171	0.123	0.115	***1***	0.017	-0.49
EXTOAVG	Pearson Correlation	0.109	-0.187	-0.097	0.697	0.039	0.017	***1***	0.248
SCMAVG	Pearson Correlation	-0.071	0.198	0.174	0.314	-0.423	-0.49	0.248	***1***

To interpret the values of the coefficients, positive values denote positive linear correlation, negative values denote negative linear correlation, a null value denotes no linear correlation and the closer the value is to 1 or -1, the stronger the linear correlation. Most researchers agree that a coefficient of <0.1 indicates a negligible and >0.9 a very strong relationship [273]. Strong correlations occur between variables that do not fluctuate much, so the mean value will be positively correlated with its minimum and maximum value. The accelerometry and gyroscopy values of the sternum and pelvis, whether in mean, standard deviation, or maximum and minimum values, are highly correlated, as the displacement and tilt of the trunk and hip are also positively correlated in biomechanical terms. Therefore, in the statistical model, multicollinearity does not present a problem since there is no need to reduce the independent variables. Only caution is required in interpreting the results.

6.1.1.4 Linear relationship between the dependent variables for each level of the independent variables

To check this linear relationship assumption, in SPSS, produced a Scatter/Dot graph (Matrix Scatter). For simplicity, Figure 56 represents two dependent variables - RTAAVG and LTAAVG. The independent variable is the Perturbation/No Perturbation level, named BinaryLabel.

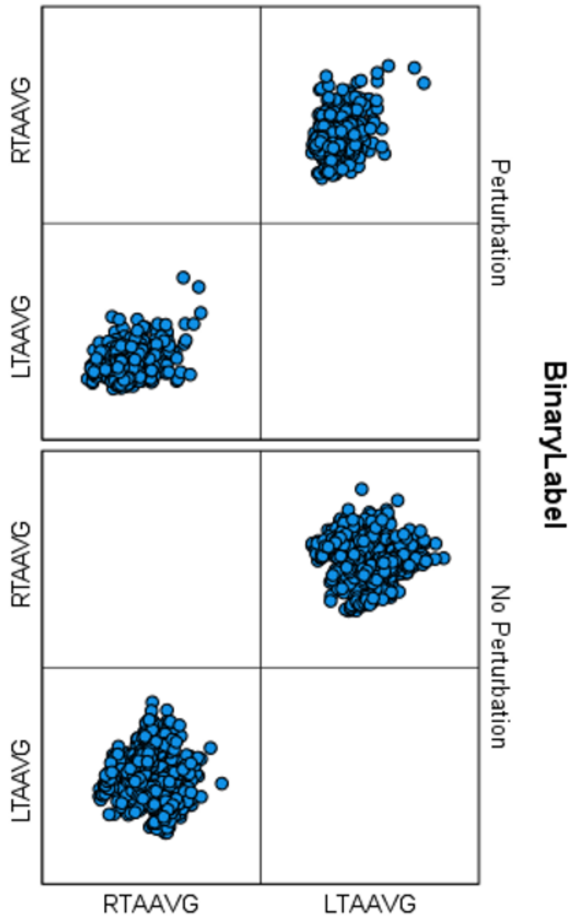


Figure 56: Linear relationship between the dependent variables for each level of the independent variables.

6.1.1.5 Homogeneity of variance-covariance matrices

Variance-covariance matrices should be compared between groups using Box's M-test, also called Box's Test for Equivalence of Covariance Matrices, is sensitive to non-normality. Box's M test is a parametric test used to compare variation in multivariate samples. More specifically, it tests if two or more covariance matrices are equal or homogeneous. The null hypothesis for this test is that the observed covariance matrices for the dependent variables are equal across groups, i.e., a non-significant test result indicates that the covariance matrices are equal. The generated test statistic is called Box's M statistic [274].

Box's Test of Equality of Covariance Matrices[a]

Box's M	10302.394
F	2.358
df1	2640
df2	46676.994
Sig.	<.001

Figure 57: Box's M Test.

Box's M-test p<0.001. The null hypothesis of equal (homogeneous) covariances matrices is rejected: the assumption of homogeneity of covariance is violated. However, real data in behavioral science often deviates from this assumption [275], [276]. Previous studies have found that analysis of variance is sensitive to violations of homogeneity assumptions [277–280], and several procedures have been proposed for dealing with heteroscedasticity. This suggests that heterogeneity has a greater effect on MANOVA robustness than does non-normality [278]. A desirable characteristic of a test is that while it is powerful, i.e., sensitive to changes in the specified factors under test, it is robust, i.e., insensitive to changes in extraneous factors not under test. Specifically, a test is called robust when its significance level (Type-I error probability) and power (one minus Type-II error probability) are insensitive to departures from the assumptions on which it is derived [281].

6.1.2 Post-hoc Test

If a significant F-value in an analysis of variance is obtained, it merely indicates that there are differences between the population means. This significant value is an indicator that there is an influence of visual perturbations on gait parameters. That is, visual perturbations effectively invoke imbalances. However, it does not inform in which specific perturbations the averages are significantly different. This separation will allow to infer which disturbances challenged the participant's balance the most. At this point, after conducting the analysis of variances, differences between means individually will be inspected. The purpose is to isolate significant differences. Much of the work on multiple comparisons is based on Tukey's work. In the case of this experimental protocol, an important comparison will be between the cases where there is no perturbation and the cases where there is visual perturbation. In this case, the undisturbed gait will serve as a control. When one group is to be compared against several, the recommended test is Dunnett's test [282]. Dunnett's test is performed by computing a Student's t-statistic for each group where the statistic compares the treatment group (perturbation labels) to a single control group (no perturbation). Since each comparison has the same control in common, the procedure incorporates the dependencies between these comparisons. In particular, the t-statistics are all derived from the same estimate of the error variance which is obtained by pooling the sums of squares for error across all groups. The formal test statistic for Dunnett's test is either the largest in absolute value of these t-statistics, or the most negative or most positive of the t-statistics.

6.2 Results and Discussion

6.2.1 one-way MANOVA

An initial one-way MANOVA examined muscular, kinematic and electrodermical metrics as dependent variables, and binary label perturbation/no perturbation as independent variables. Skewness, kurtosis, and Lyapunov exponents of the calculated metrics were excluded from this multivariate analysis. Since the factor is the binary label perturbation/non-perturbation, no post-hoc tests are performed. Four conventional statistics for MANOVA are typically reported: Pillai's Trace, Wilks' Lambda, Hotelling-Lawley Trace, and Roy's Greatest Root [283]. Figure 58 reports the four multivariate tests mentioned, as it was reported in SPSS. All multivariate tests show a significant effect of the perturbation/no-perturbation factor difference on the means of the muscular, kinematic, and electrodermal dependent variables. **Pillai's Trace = .725, F = 25.059, df = (104), p <.001.**

This means that there is a statistical effect of the perturbation factor on all independent variables analyzed multivariately, which has to be investigated with multiple follow-up ANOVAs. Pillai's trace test statistic gives more robust results in the case of homogeneous and heterogeneous variances, in unbalanced samples. Regarding the one-way MANOVA results, Can Ateş et al. [284] investigate how Wilks' lambda, Pillai's trace, Hotelling's trace, and Roy's largest root test statistics can be affected when the normal and homogeneous variance assumptions of the MANOVA method are violated. Since the observations obtained without perturbation are 553 and the observations with perturbation account for 543 observations for each dependent variable, indicates an unequal sample size. This is not a mandatory assumption to proceed with variance analysis but there are two potential problems that can arise: reduced statistical power and reduced robustness to unequal variance.

In practice, after detecting a statistically significant multivariate effect, researchers frequently resort to univariate ANOVAs and multiple pairwise t-tests to explain what is accounting for these group differences. Huberty and Morris [285] indicated that 96% of MANOVA follow-ups were univariate ANOVAs. In educational research, 84% of studies followed that strategy as well [286]. The popularity of the multivariate–univariate approach is not surprising as several leading textbooks on multivariate methods have suggested this strategy [287, 288]. The MANOVA test functions as a gatekeeper and is used with the hopes of controlling for Type I error rates when analyzing multiple dependent variables [289].

Multivariate Tests[a]

Effect		Value	F	Hypothesis df	Error df	Sig.
Intercept	Pillai's Trace	1.000	398863.942[b]	104.000	991.000	.000
	Wilks' Lambda	.000	398863.942[b]	104.000	991.000	.000
	Hotelling's Trace	41858.577	398863.942[b]	104.000	991.000	.000
	Roy's Largest Root	41858.577	398863.942[b]	104.000	991.000	.000
BinaryLabel	Pillai's Trace	.725	25.059[b]	104.000	991.000	<.001
	Wilks' Lambda	.275	25.059[b]	104.000	991.000	<.001
	Hotelling's Trace	2.630	25.059[b]	104.000	991.000	<.001
	Roy's Largest Root	2.630	25.059[b]	104.000	991.000	<.001

a. Design: Intercept + BinaryLabel

b. Exact statistic

Figure 58: Four conventional statistics for MANOVA reported: Pillai's Trace, Wilks' Lambda, Hotelling-Lawley Trace, and Roy's Greatest Root.

6.2.2 one-way ANOVA

A one-way ANOVA was performed to evaluate the effect of visual perturbations (labels) on muscle activation, kinematic parameters, CoM velocity and galvanic skin response dependent variables. One may recall the statistical variables calculated for the metrics available from the schematic in Figure 52. Again it is emphasized that only the metrics "average - AVG", "standard deviation - STD", "minimum - MIN"and "maximum - MAX"are in use. Overall, there are 32 variables from muscle activity, 48 variables from accelerometry and gyroscopy of the pelvis and sternum in the 3-axis, 12 variables from CoM velocity metrics, and 9 variables from galvanic skin response. A one-way ANOVA revealed that there was a statistically significant difference in the dependent variables listed in Table 20 between at least two groups. F-value and p-value are presented in Table 20 for each dependent variable. The table is arranged to show the variables with the most significant p-value first. There was no significant difference in the dependent variables showing the p-value with red text color ($p>0.05$). Purple corresponds to CoM velocity variables, blue to muscle variables, green to gyroscopy variables, and orange to accelerometry. (F(between groups df, within groups df) = [F-value], p = [p-value])

As an example of reporting, a one-way ANOVA was performed to compare the effect of visual perturbations (different labels) on Right Tibialis Anterior average muscle activation (RTAAVG). A one-way ANOVA revealed that there was a statistically significant difference in Right Tibialis Anterior average muscle activation between at least two groups ($F(35,1060) = 11.181$, $p<0.001$). The variables referring to the galvanic response of the skin that reflect physiological activity related to anxiety and fear of falling did not obtain significant values. Thus, were not included in the aforementioned table.

Table 20: One-way ANOVA results: F-value and p-value. The order of this table is done in order to understand which variables had the highest F-value, that is, which were most influenced by the situations where there were visual disturbances. Additionally, variables are separated by color to get an idea of which groups of variables were most affected. Blue - muscle variables; Green - Gyroscope; Orange - Accelerometer; Purple - CoM Velocity.

F-value	p-value	Independent Variable	F-value	p-value	Independent Variable	F-value	p-value	Independent Variable
44.3876	4.9562E-181	COMVelXMIN	7.398	1.67488E-31	RGMSTD	3.349156	3.9924E-10	StrGyrzAVG
30.3299	2.9712E-134	COMVelXAVG	7.397	1.68877E-31	RTASTD	3.095718	7.0099E-09	StrAccyAVG
22.269	1.3568E-102	StrGyrxSTD	7.307	5.07651E-31	RGMAVG	3.05817	1.0655E-08	StrAcczMIN
20.1081	2.34E-93	StrGyrzMAX	6.589	3.55611E-27	PelGyrzSTD	3.03915	1.3164E-08	PelAccyMIN
19.552	6.33223E-91	PelGyrzMAX	6.083	1.83851E-24	PelAccyMAX	3.017882	1.6667E-08	StrAccyMAX
18.9544	2.7722E-88	PelAccxSTD	5.544	1.40609E-21	PelAcczMIN	3.012572	1.7677E-08	StrAcczSTD
17.8613	2.22145E-83	PelAccySTD	5.367	1.2499E-20	RGMMAX	3.007972	1.8601E-08	StrAccxMIN
14.9473	7.81648E-70	PelGyrxSTD	5.22	7.57E-20	PelGyrxMAX	3.004329	1.9366E-08	StrAccyMIN
12.9882	2.50855E-60	COMVelYMAX	5.003	1.07675E-18	PelAccxMAX	3.00356	1.9532E-08	StrAcczMAX
12.8724	9.36689E-60	PelAcczSTD	4.979	1.45508E-18	LGMSTD	3.00045	2.0216E-08	StrAccxMAX
11.6453	1.26888E-53	StrGyrzMIN	4.744	2.56E-17	StrGyryMAX	2.980041	2.5328E-08	StrAccxSTD
11.18100	2.87403E-51	RTAAVG	4.724	3.25173E-17	COMVelZAVG	2.969282	2.8519E-08	StrAccxAVG
11.035	1.5833E-50	PelGyrzMIN	4.458	8.16166E-16	PelAcczMAX	2.966565	2.9386E-08	StrAccySTD
10.8063	2.33108E-49	LTAMAX	4.417	1.35E-15	StrGyryMIN	2.916839	5.07E-08	PelGyryMIN
10.5536	4.60525E-48	COMVelZSTD	4.292	6.04921E-15	PelAccxMIN	2.741791	3.38E-07	PelGyryMAX
10.4218	2.19047E-47	StrGyrySTD	3.934	4.34E-13	PelGyrxMIN	2.683527	6.2881E-07	PelAccxAVG
9.80508	3.385E-44	PelGyrySTD	3.781	2.66E-12	SCMMAX	2.680819	6.4714E-07	EXTOMAX
9.26007	2.35024E-41	RTAMAX	3.693	7.44721E-12	COMVelYSTD	2.662867	7.8266E-07	StrGyrxAVG
8.7332	1.37279E-38	StrGyrzSTD	3.678	8.9323E-12	LGMMAX	2.660458	8.0286E-07	EXTOSTD
8.64069	4.21946E-38	COMVelXSTD	3.663	1.06239E-11	COMVelYAVG	2.640012	9.9626E-07	COMVelZMAX
8.42246	5.9924E-37	LTASTD	3.619	1.77103E-11	StrGyryAVG	2.5719	2.0341E-06	PelAccyAVG
8.21407	7.60E-36	StrGyrxMAX	3.618	1.79964E-11	STMAX	2.501778	4.2048E-06	COMVelXMAX
7.4828	5.88E-32	StrGyrxMIN	3.596	2.32318E-11	COMVelYMIN	2.415235	1.0169E-05	RFMIN
7.47933	6.13649E-32	LTAAVG	3.531	4.92824E-11	RFMAX	2.400227	1.1833E-05	LGMAVG
			3.467	1.03E-10	RFSTD	2.332105	2.3405E-05	STAVG
			3.465	1.05448E-10	SCMSTD	2.123562	0.0001764	STSTD
			3.366	3.27661E-10	PelGyrzAVG	2.063500	0.00030898	RTAMIN
						2.045522	0.00036468	PelGyryAVG
						2.03077	0.0004175	RFAVG
						1.959229	0.00079695	RGMMIN
						1.906913	0.00126532	PelAcczAVG
						1.894578	0.00140912	LTAMIN
						1.572945	0.01891391	StrAcczAVG
						1.558554	0.02101814	PelGyrxAVG
						0.955329	0.54398981	COMVelZMIN
						0.836843	0.73760913	LGMMIN
						0.709917	0.89614266	STMIN
						0.580864	0.97617683	EXTOMIN
						0.181758	0.99999998	EXTOAVG
						0.14402	1	SCMMIN
						0.120651	1	SCMAVG

6.2.3 Dunnett Post Hoc

Dunnett's t-test with "No Pertubation"level as control group for multiple comparisons found that the mean values of independent variables in table 21 were significantly different between control group "No Perturbation"and the perturbations described in the same table. Post hoc test p-values and confidence intervals lower and upper bound can be consulted in Appendix 3. In the same way that ANOVA results were presented, to illustrate the results obtained in the post hoc test, a muscle variable was picked - *RTAAVG*, which represents the average value of the percentage of activation of the *Right Tibialis Anterior* muscle during the occurrence of the various perturbations. Table 21 shows the results of this test. The greatest differences between the averages when comparing the control group (no disturbance) with the visual disturbances are colored green. The red colors indicate the opposite, meaning that there was minimal interference of the visual disturbance on the average activation of this muscle. From the mentioned table, it is apparent that the visual disturbance that most affects the contraction of this muscle is Roll Indoor 2 CCW30. For illustrative purposes, Figure 59 represents a subject during the protocol experiencing this perturbation. The labels in bold represent the other most effective perturbations.

Table 21: Dunnett t-test (2-sided) result - Right Tibialis Anterior. The color gradation means a higher value for the green-tone colors and a lower value for the red ones, i.e., in green are the entries that the corresponding visual disturbance introduced the most difference in means.

Dependent Variable: **RTAAVG**
Dunnett t (2-sided): control group "No Perturbation"

(I) Label	(J) Label	Mean Difference (I-J)	Sig.
Roll Indoor 1 CW10	No Perturbation	2.619624074	0.014
Roll Indoor 1 CW20	No Perturbation	2.640917425	0.013
Roll Indoor 1 CW30	No Perturbation	2.82942145	0.005
Roll Indoor 1 CCW10	No Perturbation	3.068273265	0.001
Roll Indoor 1 CCW20	No Perturbation	3.182434604	0.001
Roll Indoor 1 CCW30	No Perturbation	4.154222882	0.000
Roll Indoor 2 CW10	No Perturbation	3.229842329	0.000
Roll Indoor 2 CW20	No Perturbation	1.720165350697620	0.505
Roll Indoor 2 CW30	No Perturbation	3.045723908	0.001
Roll Indoor 2 CCW10	No Perturbation	3.591367025	0.000
Roll Indoor 2 CCW20	No Perturbation	4.049127951	0.000
Roll Indoor 2 CCW30	No Perturbation	4.903273231	0.000
Roll Outdoor CW10	No Perturbation	4.198288474	0.000
Roll Outdoor CW20	No Perturbation	1.597146008609180	0.662
Roll Outdoor CW30	No Perturbation	2.031776183747670	0.191
Roll Outdoor CCW10	No Perturbation	3.05686819	0.001
Roll Outdoor CCW20	No Perturbation	4.553190306	0.000
Roll Outdoor CCW30	No Perturbation	3.400362578	0.000
ML Axis Trans - Kitchen	No Perturbation	1.896113135	0.001
AP Axis Trans - Corridor Forward	No Perturbation	2.020538563	0.000
AP Axis Trans - Corridor Backward	No Perturbation	2.004027816	0.000
Pitch Indoor - Bathroom	No Perturbation	2.541397364	0.021
Pitch Indoor - Fridge	No Perturbation	2.696144191	0.010
Window Roof Beam Walking - Vertigo	No Perturbation	3.453513534	0.000
Window Roof Beam Walking - Vertigo No avatar	No Perturbation	2.612068727	0.015
Simple Roof - Vertigo	No Perturbation	1.518723556932080	0.830
Simple Roof - Vertigo No avatar	No Perturbation	1.620855505233680	0.632
Pitch Outdoor - Near Car Oil	No Perturbation	2.304171835665670	0.063
Bedroom Syncope	No Perturbation	2.101540562	0.000
Garden - Object Avoidance	No Perturbation	1.778019862	0.002
Electricity Pole - Vertigo	No Perturbation	2.432576435	0.035
Electricity Pole - Vertigo No avatar	No Perturbation	2.280966349176490	0.070
Free Fall	No Perturbation	2.213991566	0.000
Stairs	No Perturbation	2.392414359	0.001
Trip - Sidewalk	No Perturbation	3.86578572	0.000

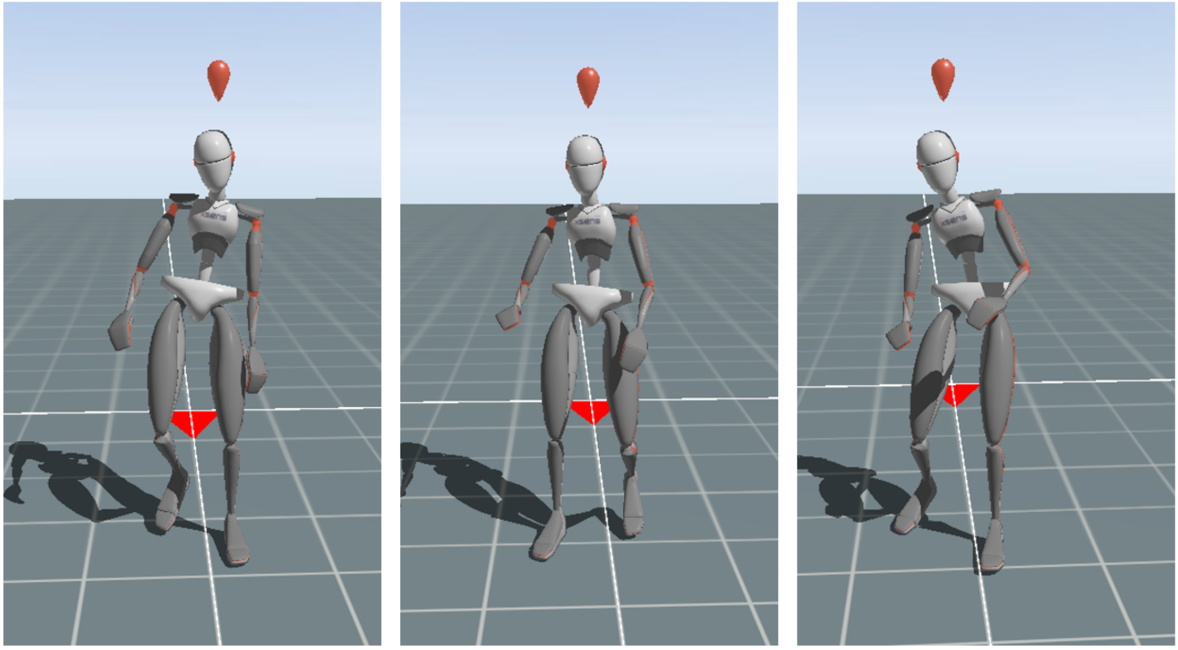


Figure 59: Visual inspection of a visual disturbance (Roll Indoor 2 CCW30), depicted as the disturbance that most strongly affected the activation of the Right Tibialis Anterior muscle.

6.2.3.1 Muscle variables

To prevent falling, adequate and timely muscle reactions are required to counteract the disturbance. Studies of balance and posture during quiet or perturbed standing have identified the dominance of the ankle muscles (plantarflexors/dorsiflexors) in the AP direction and hip abd/adductor muscles in the ML direction [258]. Kim et al. in 2022 [290] summarize reports of muscle activations during perturbed walking. All studies analyzed the Tibialis Anterior, a muscle responsible for ankle dorsiflexion. Its antagonist muscle, i.e., the one that relaxes when the Tibialis Anterior contracts, and vice versa, is the Gastocnemius. The results of this review [290] suggest that the plantarflexors and knee flexors muscles play a key role in disturbed gait. From the muscle variables, it is worth noting the massive presence at the top of the Table 20 of those from the lower leg muscles: Right and Left Tibialis Anterior and Right and Left Gastrocnemius Medial Head. By itself, this finding points favorably in the direction intended to prove. The muscles responsible for maintaining and restoring balance after gait disturbance are the most recruited by the visual disturbances introduced in this experimental protocol.

Dunnett's post hoc results show that there was a significant interference of the Pitch visual perturbations on the Gastrocnemius muscles. This findings are indicative of a similar muscle pattern induction between a physical slip induction perturbation and the visual perturbation Pitch applied in this project. Similar EMG activation pattern in Gastrocnemius is also observed in a physical disturbance consisting of backward platform translation. Once again the strong interaction of the visual perturbation AP Translation Backward with the activation of the Gastrocnemius is verified by statistical analysis establishing once again a strong association between the visual perturbation imposed and the related physical perturbation. Table 29 shows Dunnett's test for the maximum values of Tibialis Anterior of both legs.

In line with the results obtained in [291], also the Tibialis Anterior and Gastrocnemius Medial muscles of the dominant leg showed significant differences in the means between the mediolateral perturbation and the non-perturbation condition. In the RTAAVG and RGMAVG variables, the significance was p=0.0005 and p=0.0019, respectively. This shows that the ML Translation visual perturbation applied had a significant impact on the average change in activation of the ankle stabilizer muscles Tibialis Anterior and Gastrocnemius Medial. Also, in line with the results of Hallal et al. article [292], the results of Dunnett's post hoc test on the variables relating to the predominant thigh muscles (Table 22) show that the most effective perturbations at eliciting these muscles were Trip and Object Avoidance.

Table 22: Thigh muscles (Rectus Femoris and Semitendinosus) from proeminent leg. Dunnett post hoc results.

Dunnett t (2-sided): control group "No Perturbation"

Dependent Variable: **RFAVG**			
(I) Label	(J) Label	Mean Difference (I-J)	Sig.
Trip	No Perturbation	1.51143602471482	0.000

Dependent Variable: **RFSTD**			
(I) Label	(J) Label	Mean Difference (I-J)	Sig.
Trip	No Perturbation	3.29341839990663	0.000
Object Avoidance	No Perturbation	1.42813713948654	0.004

Dependent Variable: **STAVG**			
(I) Label	(J) Label	Mean Difference (I-J)	Sig.
Trip	No Perturbation	2.37444950421642	0.000
Object Avoidance	No Perturbation	1.34921899759266	0.015

Dependent Variable: **STSTD**		Mean Difference (I-J)	Sig.
(I) Label	(J) Label		
Trip	No Perturbation	2.51118468036652	0.000
Object Avoidance	No Perturbation	1.27082186434166	0.043

6.2.4 Kinematic variables

In order to assist in the comprehension of the topics discussed about kinematic measurements, it is useful to recall the directions of the axes defined by the Xsens motion capture system. Figure 60 shows a representation of the global frame on which this section is based.

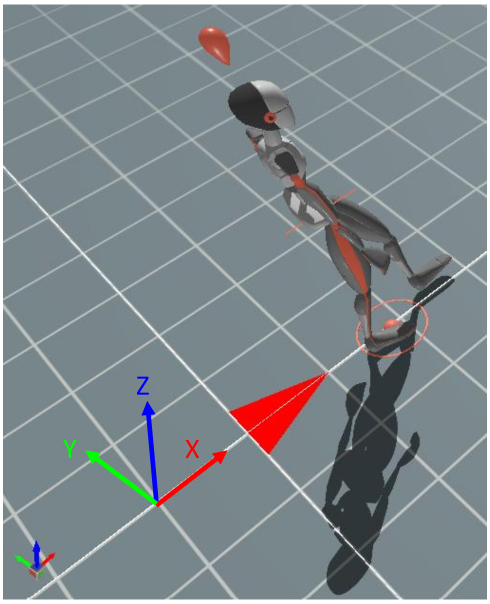


Figure 60: Global frame representation of the Xsens motion capture system. The center of mass is also represented in this image.

A movement about the X-axis can be called a movement in the sagittal or longitudinal plane. Its directions are Anterior or Posterior. It is also referred to as movement in the AP or anteroposterior plane. A movement about the Y-axis can be called movement in the horizontal, axial, or transverse plane. Its directions are Medial and Lateral. This movement is also referred to as movement in the ML or mediolateral plane. A movement about the Z-axis can be called movement in the frontal or coronal plane. Its directions are Inferior and Superior. It also refers to this movement as movement in the vertical plane.

The average value of the gyroscope in the y-axis of the inertial sensor placed on the Pelvis, is represented by the variable PelGyryAVG. This variable anatomically represents trunk flexion average, which reflects a strategy of the hip to counteract external perturbations in the anteroposterior plane. Its placement in Table 20 is in the last entries. As seen, the table is organized in such a way as to present first the variables that exhibited the most differences in the means with the introduction of the perturbations. Dunnett's post hoc test is presented for this variable in Table 23. As can be seen from the results in this table, the visual perturbations of backward translation in the anteroposterior plane and free fall are those that induce the greatest difference in means compared to no perturbation. The sequence in Figure 61 shows this trunk flexion, or a compensatory rotation reaction around the hip, during the *AP Axis Trans - Corridor Backward* perturbation.

Table 23: Dunnett t-test (2-sided) result - Pelvis Gyroscope Y-axis average. The color gradation means a higher value for the green-tone colors and a lower value for the red ones, i.e., in green are the entries that the corresponding visual disturbance introduced the most difference in means, in the given variable.

Dependent Variable: **PelGyryAVG**
Dunnett t (2-sided): control group "No Perturbation"

(I) Label	(J) Label	Mean Difference (I-J)	Sig.
Roll Indoor 1 CW10	No Perturbation	-0.019156068744117	1.000
Roll Indoor 1 CW20	No Perturbation	0.046585652433260	0.995
Roll Indoor 1 CW30	No Perturbation	-0.050398606785153	0.985
Roll Indoor 1 CCW10	No Perturbation	0.061744678548020	0.850
Roll Indoor 1 CCW20	No Perturbation	-0.011246342351866	1.000
Roll Indoor 1 CCW30	No Perturbation	-0.097596640962541	0.077
Roll Indoor 2 CW10	No Perturbation	0.024695424285692	1.000
Roll Indoor 2 CW20	No Perturbation	-0.030283073216625	1.000
Roll Indoor 2 CW30	No Perturbation	0.032938450919864	1.000
Roll Indoor 2 CCW10	No Perturbation	-0.068498746982669	0.678
Roll Indoor 2 CCW20	No Perturbation	-0.004423284802159	1.000
Roll Indoor 2 CCW30	No Perturbation	0.029774188599725	1.000
Roll Outdoor CW10	No Perturbation	-0.002263402923174	1.000
Roll Outdoor CW20	No Perturbation	0.016581080937336	1.000
Roll Outdoor CW30	No Perturbation	-0.000245233930631	1.000
Roll Outdoor CCW10	No Perturbation	-0.002883735760655	1.000
Roll Outdoor CCW20	No Perturbation	-0.048104335049574	0.992
Roll Outdoor CCW30	No Perturbation	0.068576448958735	0.676
ML Axis Trans - Kitchen	No Perturbation	0.004693273921583	1.000
AP Axis Trans - Corridor Forward	No Perturbation	0.013369126863331	1.000
AP Axis Trans - Corridor Backward	No Perturbation	0.102917761	0.000
Pitch Indoor - Bathroom	No Perturbation	0.015340377584652	1.000
Pitch Indoor - Fridge	No Perturbation	-0.074799266690465	0.493
Window Roof Beam Walking - Vertigo	No Perturbation	0.006742537685738	1.000
Window Roof Beam Walking - Vertigo No avatar	No Perturbation	0.008481912128048	1.000
Simple Roof - Vertigo	No Perturbation	0.002655106833035	1.000
Simple Roof - Vertigo No avatar	No Perturbation	0.001509060161740	1.000
Pitch Outdoor - Near Car Oil	No Perturbation	0.047790385795662	0.993
Bedroom Syncope	No Perturbation	0.036001969523966	0.862
Garden - Object Avoidance	No Perturbation	0.029512236710898	0.986
Electricity Pole - Vertigo	No Perturbation	0.004071963758247	1.000
Electricity Pole - Vertigo No avatar	No Perturbation	0.001497742267666	1.000
Free Fall	No Perturbation	0.070848516	0.009
Stairs	No Perturbation	0.010569542077887	1.000
Trip - Sidewalk	No Perturbation	-0.000118785402072	1.000

Table 24: ANOVA results - CoM velocity variables.

F-value	p-value	Independent Variable (CoM Velocity)
44.3876	4.96E-181	COMVelXMIN
30.3299	2.97E-134	COMVelXAVG
12.9882	2.509E-60	COMVelYMAX
10.5536	4.605E-48	COMVelZSTD
8.64069	4.219E-38	COMVelXSTD
4.72416	3.252E-17	COMVelZAVG
3.69318	7.447E-12	COMVelYSTD
3.6628	1.062E-11	COMVelYAVG
3.59569	2.323E-11	COMVelYMIN
2.64001	9.963E-07	COMVelZMAX
2.50178	4.205E-06	COMVelXMAX
0.95533	0.5439898	COMVelZMIN

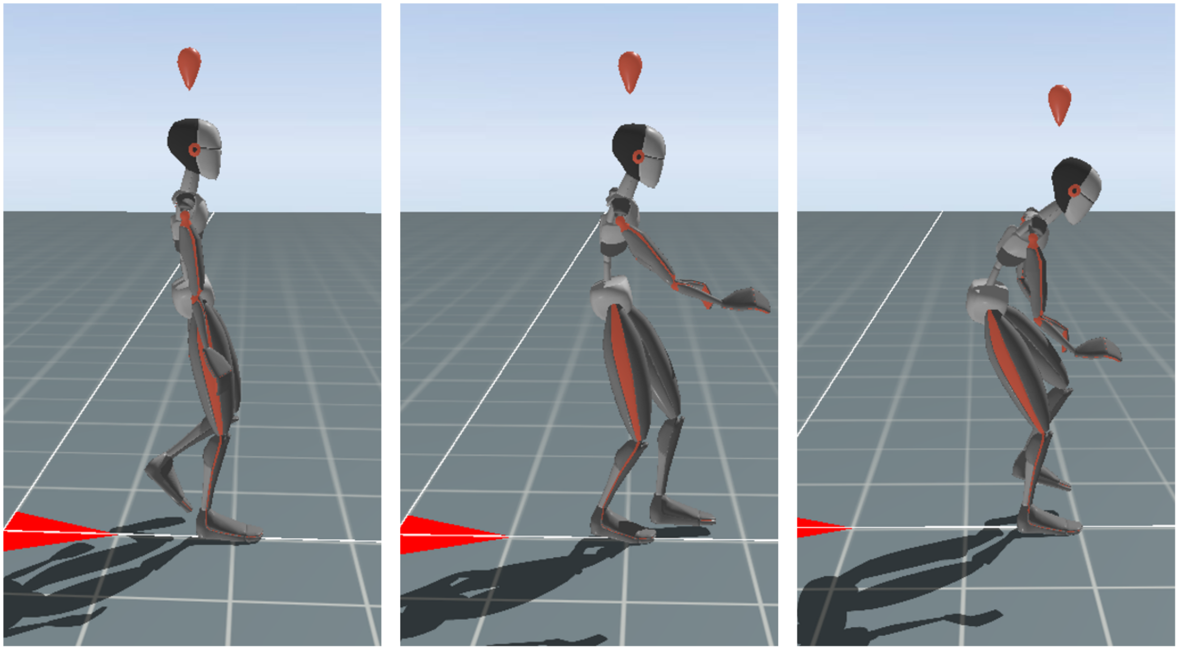


Figure 61: Illustration of a typical reaction to perturbation *AP Axis Trans - Corridor Backward*. This sequence is intended to validate the results shown in Table 23.

The horizontal velocity of the CoM must be considered when describing the range of possible motions. Balance will not be maintained when there is sufficient velocity of the center of mass in the horizontal plane, even if the position of the CoM is within the base of support [293]. Both model simulations and human experimental data have previously indicated that the CoM velocity plays an important role in regulating stable walking [294]. Table 24 provides the ordered F-values of the ANOVAs that relate to the CoM velocity variables.

Table 24 presents the results of Dunnett's test, which detects which visual perturbations, compared to the "No Perturbation"control, caused the variable CoMVelX to show the greatest difference in the means, representing a greater influence of the perturbations that have a larger (I-J) Mean Difference value. In this case, the variable represents the average velocity of the CoM in the X-axis, which depicts faster translations of the projection of the CoM along the AP direction. As expected, the perturbations that appear in bold in Table 24 are those that have a mean difference at the 0.05 level. That is, they were strongly influenced by these perturbations. The ones with a larger value of mean difference are the perturbations in Roll and Pitch. It is natural that perturbations that cause the participant to incur a faster gait or make abrupt stops using the hip strategy, are those that most manipulate the variable CoM velocity X-axis, on average. This happens in all Roll perturbations that, despite being perturbations that try to induce a lateral drop, cause the compensatory reaction of rapid oscillations in the horizontal plane during the perturbed gait. This inference will be demonstrated via a sequential display of frames. Figure 62 shows a sequence during the experimental protocol of a subject experiencing a Roll perturbation.

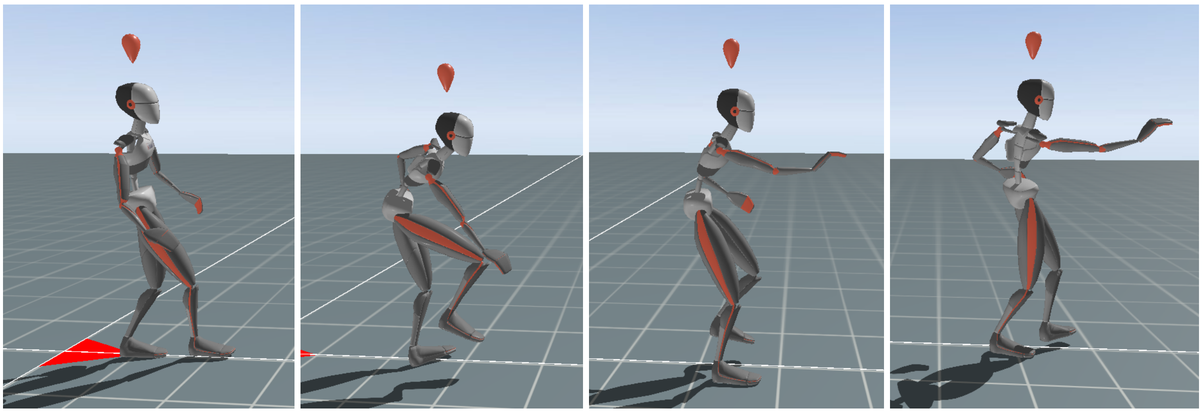


Figure 62: Participant experiencing a Roll perturbation.

It is noticeable that the participant, trying to maintain balance, is using a hip strategy and foot positioning that causes a velocity increase on the X-axis CoM. On the other hand, from the same table (Table 24 it follows that a perturbation ML Axis Translation has no statistical significance in this variable. This is because it induces a lateral sway that is not as abrupt as Roll. From Table 24 it appears that in the Y-axis or ML plane direction, although there was a significant effect of the perturbations on the average CoM velocity, it had much lower values than in the previous analysis in the X-axis. With the exception of the maximum value. This means that the strategies employed by the participants were sufficient to react to the visual perturbation and did not enter into CoM velocities in the ML direction that could lead to a fall. However, from the maximum value, inferences can be made that there were quite strong visual perturbations that caused the CoM to reach maximum velocities in this direction.

Regulation at hip level become more important in the case of stronger perturbations (hip strategy) [295]. In order to attempt to have a better understanding of the compensatory postural reactions, the

3-dimensional accelerometry and gyroscopy variables from the inertial sensor of the pelvis and sternum were used. Table 25 gathers the accelerometry variables.

Table 25: ANOVA results - acceletometry variables (pelvis and sternum).

F-value	p-value	Independent Variable (Acc)
18.95441747	2.7722E-88	PelAccxSTD
17.8613447	2.22145E-83	PelAccySTD
12.87239219	9.36689E-60	PelAcczSTD
6.08277026	1.83851E-24	PelAccyMAX
5.544332578	1.40609E-21	PelAcczMIN
5.003434221	1.07675E-18	PelAccxMAX
4.458436344	8.16166E-16	PelAcczMAX
4.292283176	6.04921E-15	PelAccxMIN
3.095717802	7.00993E-09	StrAccyAVG
3.058169938	1.06551E-08	StrAcczMIN
3.039150338	1.31642E-08	PelAccyMIN
3.01788209	1.66675E-08	StrAccyMAX
3.012572308	1.76773E-08	StrAcczSTD
3.00797208	1.86011E-08	StrAccxMIN
3.004329145	1.93663E-08	StrAccyMIN
3.003559771	1.95319E-08	StrAcczMAX
3.000449732	2.02156E-08	StrAccxMAX
2.980041256	2.53282E-08	StrAccxSTD
2.969281813	2.85192E-08	StrAccxAVG
2.966565049	2.93859E-08	StrAccySTD
2.683527474	6.28808E-07	PelAccxAVG
2.571899896	2.03414E-06	PelAccyAVG
1.906912998	0.001265315	PelAcczAVG
1.572945117	0.018913906	StrAcczAVG

It is obvious, by inspecting the table, the first three positions to be occupied by the variables of acceleration in the IMU of the pelvis, namely those representing the standard deviation. Regarding the variables representing mean values, it is noticeable that only the acceleration in the Y-axis of the sternum occupies top positions in the table. Extra attention is required when inspecting the results for variables representing acceleration and rotation of a sensor, since sensor placements may be different. Previously, the CoM directions used the global frame, since it is a calculated variable. The axes of the inertial sensor are as shown in Figure 63.

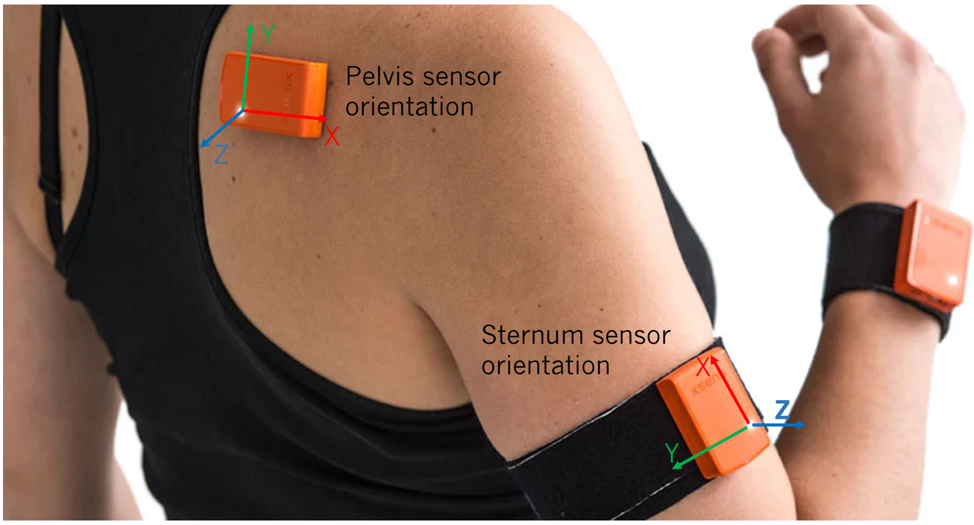


Figure 63: Axis orientations for Pelvis and Sternum intertial sensors.

6.2.5 The most effective visual disturbances

In the muscular variables, the muscles of the lower leg stand out. Of these variables, which visual disturbances most influenced their variability are identified. To be representative, the average and maximum values will be inspected, since the analysis is dealing with muscle activations. The average value represents the amount of time that the muscle is recruited and the maximum value reached is representative of how strong the disturbance was.

About the average activation of the Tibialis Anterior, the visual disturbances that had the most influence on this activation were Roll perturbations and Trip, in a general way. Of the Roll perturbations, in the muscles of both legs, the Roll CCW30, CCW20 and CW10 were the most effective. Additionally, the perturbation Window Roof Beam Walking had a high statistical significance regarding the average activation of the RTA and LTA. Regarding maximum values of Tibialis Anterior in both legs, there are impressive values in maximum muscle activation in vertigo situations. Especially in the two situations where the participant has the notion that he can fall on both sides.

Regarding the average activation of the Right Medial Gastrocnemius, disturbances in the Roll in general were the most influential. Roll CCW20 and CCW30 clearly stand out. Interestingly, two pitch perturbations (Pitch Bathroom and Pitch Fridge) were also among the most effective at activating this dominant leg muscle. Trip also obtained relevant significance. Strangely, the average activation of the Left Gastrocnemius Medial of the non-dominant leg was only influenced by the Free Fall and Trip perturbations, comparable to the contact of the right leg. On the maximal values of Gastrocnemius Medial Head, the right leg muscle did not exhibit significant differences for any visual perturbations. In the left leg, an interesting result shows that both perturbations in the AP direction: forward and backward, had a great influence on the maximum contraction value of this muscle of the non-dominant left leg. Additionally, in the Bedroom

Syncope perturbation there is also a large influence on the maximum contraction value variable of the LGM muscle.

The focus of the discussion is now the visual perturbations that are most effective at introducing postural reactions into the kinematic parameters. To deduce the strength of the visual disturbance on the kinematic changes, it seems appropriate to analyze which perturbation had the greatest influence on the variables representing the CoM velocity: average and maximum value. In the mean value change of the velocity of the CoM in AP direction, it can be stated that all Roll perturbations proved to be quite homogeneously influential on this parameter, i.e., only Roll CCW30 and Roll CW10 stood out slightly in terms of significance. Apart from these, all perturbations in the Pitch and Translation AP were highly significant in introducing velocity into the CoM in the AP direction, as one would expect, since these perturbations try to induce a forward or backward drop. In relation the maximum value of the CoM velocity in the AP direction, the only perturbation that stands out is Bedroom Syncope, which indicates the effectiveness and strength of the perturbation to induce aneroposterior sway. Following the same reasoning and sequence of the previous paragraph, the present will analyze the influence of visual disturbances on the speed of CoM (average and maximum value), this time in the ML direction. The perturbations with the greatest influence on the mean value were the AP translations, the Free Fall and the Trip. On the maximum value, there was a general influence of the Roll perturbations and a main influence of the AP translation backward perturbation. All Pitch perturbations are effective at producing a change in the maximum value of the CoM velocity in the ML direction, as well as Free Fall and Trip. Finally, regarding the analysis of the velocity of the CoM in the Z-axis, the Bedroom Syncope perturbation proved to be highly influential of this parameter both in average and maximum values, showing its effectiveness and strength in inducing hip strategy, making the participant abruptly lower the center of mass to regain equilibrium, as happened with AP Translation in both directions.

Chapter 7

Conclusion

As presented in the introductory chapter, preventing the occurrence of falls and detecting them prematurely, and consequently minimizing the health and economic harms arising therefrom, is the main motivation of this work. To achieve this purpose, it is necessary to implement both fall prediction and fall detection systems and algorithms. Due to the low frequency of fall occurrence, these algorithms are often implemented with data from simulated falls in a controlled laboratory environment. There is effectively a lack of datasets containing real falls, and very few data available. The existing ones contain only experimental data from an inertial sensor.

During an initial literature review, this work was contextualized in the advances that have been developed with virtual reality in the field of neurorehabilitation and with elderly subjects. A brief market analysis was done to highlight the emergence of this technology. It was possible to separate these interventions into immersive and non-immersive. Due to the increasing affordability of acquiring a fully immersive device, a headset, there has been an escalating number of studies of human balance with this apparatus. For this reason, in the third chapter a literature search is conducted, following a systematic approach, focused entirely on virtual reality delivered via head-mounted display. These studies had to introduce visual disturbances and examine the compensatory reactions of the participants. A survey was made of the most commonly used virtual reality equipment, the visual stimuli used and the experimental protocol in which they are inserted - duration of the intervention, pauses and habituation time. The most common types of participants in these interventions were analyzed, as well as the objectives of the studies, the most commonly used sensor systems, and finally, the metrics obtained and the evaluated outcomes. An additional section was included on the use of electromyography in these studies to support the choice of sensor location EMG in Chapter 5. It was found that the vast majority of studies employed a visual perturbation only and often in conjunction with mechanical perturbations. Furthermore, the use of just one sensor system is a common practice. As shown, compensatory reactions are dependent on the directions and intensities of the perturbations. For this reason, there was an urgency to create multiple visual perturbations and record the compensatory reactions at various levels: physiological and kinematic. Therefore, to address the gaps identified in the literature, a proposal was presented to approach the problem. This proposal was based on the design of a virtual environment endowed with a high level of realism due to its ecological validity and presence characteristics, conferred by a home-living paradigm very close to reality. In this virtual environment, animations were created that materialize the visual perturbations. These visual disturbances are randomly presented to the subjects in an experimental protocol that will allow data collection for the construction of the dataset. This dataset intends to contain data very close to those collected in real situations. The limitations encountered in the execution of the protocol were the lack of physical space that limited the movement of the subject to a back and forth path. Furthermore, due to the nature of the triggering mechanism of the animations, it was impossible to apply all the perturbations at exactly the same gait cycle for all subjects. On the other hand, the statistical analysis covers this deficit. This statistical analysis was presented in the sixth chapter as a way to verify if the visual perturbations introduced variations in the means of the dependent variables in comparison to the no-perturbation conditions. These differences were first assessed with a multivariate analysis of variance. To understand which variables changed the most with the introduction of visual disturbances, we conducted follow-up ANOVAs. These analysis of variance showed that variables of muscle groups strongly linked to balance restoration after external perturbations were the most statistically significant. The same was true for kinematic variables that mirror the loss of balance. With these results and strong correlations with compensatory reactions resulting from physical perturbations, it is possible to claim that the visual perturbation alone was sufficient to induce the participant into imbalance.

With this, RQ1 is answered affirmatively: "Can a virtual reality headset introduce imbalances through visual perturbations? Can they cause postural reactions typical of a fall?"The results presented in Chapter 6 statistically support that it is possible and effective to induce postural reactions similar to a real-world fall by introducing visual disturbances via an HMD. The RQ2, "Which visual perturbation challenged the participants' balance the most?"was answered in the same chapter 6. The most effective perturbations to induce imbalances are rotations in the Roll, in the CCW direction and with amplitudes of 20 and 30 degrees. The visual disturbance that tries to simulate a syncope, called Bedroom Syncope, also excelled in the effectiveness and strength of inducing imbalance, as well as AP Translation perturbations, albeit to a lesser extent. The RQ3, "Which virtual situation placed the participant over the most anxiety?"was not possible to be answered since the electrodermal variables did not show sufficient statistical relevance to infer in which situations the participants were under greater anxiety and stress. Regarding RQ4, "What influence does the real-time representation of the avatar have in situations of virtual heights?", with the statistical tests performed, a significant difference in postural reactions with and without avatar was not detectable. Whenever vertigo situations had an influence on parameters indicating loss of balance, they were with similar values in situations with and without avatar. Finally, for the RQ5 ”Is it possible for a habituation phenomenon to occur to visual disturbances?”, no reduction in reactions was detected throughout the protocol. However, the answer to this research question lacks a statistic analysis that includes the exposure time as a variable.

7.1 Future Work

During the execution of the project, flaws were identified that could be improved in future work that will surely be scientifically relevant. An appointment for future work would be to make labeling fully automated. In situations like Trip it is possible, detecting the shock through the IMU placed on the foot. In other situations, a simple analysis of gait parameters such as toe off and heel strike can be taken to detect the second step and there introduce the perturbation, according to the participant’s gait steps. In this way, labeling would be fully automatic and in real time. The allocation of a control group would not bring advantages in answering the research questions. In the future, this protocol could be labeled as balance training for fall prevention if the main hypothesis that visual disturbances are sufficient to cause balance loss leading to posture recovery mechanisms is proven. In that framework, the control group would make sense to evaluate the efficiency of the perturbation-based balance training protocol. Once answered affirmatively to RQ1 and specified the visual disturbances that were more effective for certain muscle groups or kinematic parameters, there are conditions to create a more specific protocol with fewer disturbances to introduce as a balance training tool. Fundamentally, the objective would be to subject elderly people to this training tool in the future and evaluate its effectiveness in preventing falls.

Bibliography

[1] P. Tasheva, P. Vollenweider, V. Kraege, G. Roulet, O. Lamy, P. Marques-Vidal, and M. Meán. “Association Between Physical Activity Levels in the Hospital Setting and Hospital-Acquired Functional Decline in Elderly Patients.” In: *JAMA network open* 3.1 (Jan. 2020). issn: 2574-3805. doi: 10.1001/JAMANETWORKOPEN.2019.20185. url: https://pubmed.ncbi.nlm.nih.gov/32003817/.

[2] J. E. Gaugler, S. Duval, K. A. Anderson, and R. L. Kane. “Predicting nursing home admission in the U.S: A meta-analysis.” In: *BMC Geriatrics* 7.1 (June 2007), pp. 1–14. issn: 14712318. doi: 10.1186/1471-2318-7-13/FIGURES/3. url: https://bmcgeriatr.biomedcentral.com/articles/10.1186/1471-2318-7-13.

[3] *Falls*. url: https://www.who.int/news-room/fact-sheets/detail/falls (visited on 01/04/2021).

[4] B. Moreland, R. Kakara, and A. Henry. “Trends in Nonfatal Falls and Fall-Related Injuries Among Adults Aged ≥65 Years — United States, 2012–2018.” In: *MMWR. Morbidity and Mortality Weekly Report* 69.27 (July 2020), pp. 875–881. issn: 0149-21951545-861X. doi: 10.15585/MMWR.MM6927A5. url: https://www.cdc.gov/mmwr/volumes/69/wr/mm6927a5.htm.

[5] A. Bergland. “Fall risk factors in community-dwelling elderly people.” In: *Norsk Epidemiologi* 22.2 (2012), p. 1. url: www.profane.org.uk.

[6] G. Bergen, M. R. Stevens, R. Kakara, and E. R. Burns. “Understanding Modifiable and Unmodifiable Older Adult Fall Risk Factors to Create Effective Prevention Strategies.” In: *American journal of lifestyle medicine* 15.6 (Nov. 2019), pp. 580–589. issn: 1559-8284. doi: 10.1177/1559827619880529. url: https://pubmed.ncbi.nlm.nih.gov/34916876/.

[7] A. Steiner. “Effects of physical activity on postural stability.” In: *Age and ageing* 30 Suppl 4.SUPPL. 3 (2001), pp. 33–39. issn: 0002-0729. doi: 10.1093/AGEING/30.SUPPL_4.33. url: https://pubmed.ncbi.nlm.nih.gov/11769787/.

[8] L. Z. Rubenstein. “Falls in older people: Epidemiology, risk factors and strategies for prevention.” In: *Age and Ageing* 35.SUPPL.2 (2006), pp. 37–41. issn: 00020729. doi: 10.1093/ageing/afl084.

[9] R. Igual, C. Medrano, and I. Plaza. "Challenges, issues and trends in fall detection systems." In: *BioMedical Engineering Online* 12.1 (2013). issn: 1475925X. doi: 10.1186/1475-925X-12-66.

[10] R. Rajagopalan, I. Litvan, and T. P. Jung. "Fall prediction and prevention systems: Recent trends, challenges, and future research directions." In: *Sensors (Switzerland)* 17.11 (2017), pp. 1–17. issn: 14248220. doi: 10.3390/s17112509.

[11] M. Hemmatpour, R. Ferrero, B. Montrucchio, and M. Rebaudengo. "A review on fall prediction and prevention system for personal devices: Evaluation and experimental results." In: *Advances in Human-Computer Interaction* 2019 (2019). issn: 16875907. doi: 10.1155/2019/9610567.

[12] S. S. Khan and J. Hoey. "Review of fall detection techniques: A data availability perspective." In: *Medical Engineering and Physics* 39 (2017), pp. 12–22. issn: 18734030. doi: 10.1016/j.medengphy.2016.10.014. arXiv: 1605.09351. url: http://dx.doi.org/10.1016/j.medengphy.2016.10.014.

[13] E. Stack. "Falls are unintentional: Studying simulations is a waste of faking time." In: *Journal of Rehabilitation and Assistive Technologies Engineering* 4 (2017), p. 205566831773294. issn: 2055-6683. doi: 10.1177/2055668317732945.

[14] M. Kangas, I. Vikman, L. Nyberg, R. Korpelainen, J. Lindblom, and T. Jämsä. "Comparison of real-life accidental falls in older people with experimental falls in middle-aged test subjects." In: *Gait & Posture* 35.3 (Mar. 2012), pp. 500–505. issn: 0966-6362. doi: 10.1016/J.GAITPOST.2011.11.016.

[15] E. Casilari, J. A. Santoyo-Ramón, and J. M. Cano-García. "Analysis of public datasets for wearable fall detection systems." In: *Sensors (Switzerland)* 17.7 (2017). issn: 14248220. doi: 10.3390/s17071513.

[16] X. Yu, J. Jang, and S. Xiong. "A Large-Scale Open Motion Dataset (KFall) and Benchmark Algorithms for Detecting Pre-impact Fall of the Elderly Using Wearable Inertial Sensors." In: *Frontiers in Aging Neuroscience* 13 (July 2021), p. 399. issn: 16634365. doi: 10.3389/FNAGI.2021.692865/BIBTEX.

[17] R. N. Ferreira, N. F. Ribeiro, and C. P. Santos. "Fall Risk Assessment Using Wearable Sensors: A Narrative Review." In: *Sensors* 22.3 (Feb. 2022). issn: 14248220. doi: 10.3390/S22030984.

[18] F. Di Nardo, A. Mengarelli, E. Maranesi, L. Burattini, and S. Fioretti. "Assessment of the ankle muscle co-contraction during normal gait: A surface electromyography study." In: *Journal of Electromyography and Kinesiology* 25.2 (Apr. 2015), pp. 347–354. issn: 18735711. doi: 10.1016/J.JELEKIN.2014.10.016.

[19] S. H. Roy, M. S. Cheng, S. S. Chang, J. Moore, G. De Luca, S. H. Nawab, and C. J. De Luca. "A combined sEMG and accelerometer system for monitoring functional activity in stroke." In: *IEEE Transactions on Neural Systems and Rehabilitation Engineering* 17.6 (2009), pp. 585–594. issn: 15344320. doi: 10.1109/TNSRE.2009.2036615.

[20] M. Slater. "Place illusion and plausibility can lead to realistic behaviour in immersive virtual environments." In: *Philosophical Transactions of the Royal Society B: Biological Sciences* 364.1535 (2009), pp. 3549–3557. issn: 14712970. doi: 10.1098/rstb.2009.0138.

[21] A. M. Gonzalez and A. B. Raposo. "Fall risk analysis during VR interaction." In: *Proceedings - 19th Symposium on Virtual and Augmented Reality, SVR 2017* 2017-Novem (2017), pp. 18–28. doi: 10.1109/SVR.2017.11.

[22] *HTC Vive Virtual Reality Headset Review | Time*. url: https://time.com/4280792/htc-vive-review/ (visited on 12/13/2020).

[23] *VIVE Pro | VIVE United States*. url: https://www.vive.com/us/product/vive-pro/ (visited on 06/25/2022).

[24] I. Sutherland. "The ultimate display." In: (1965).

[25] A. Yogasingam. "Virtual Boy: Nintendo's (red and) black sheep." In: *Electronic Engineering Times* 1559 (Apr. 2009), pp. 5–6. issn: 01921541.

[26] *Virtual Boy | Nintendo | Fandom*. url: https://nintendo.fandom.com/wiki/Virtual{_}Boy (visited on 06/27/2022).

[27] *Virtual Reality Market Size and Share Forecast Report, 2029*. url: https://www.fortunebusinessinsights.com/industry-reports/virtual-reality-market-101378 (visited on 06/27/2022).

[28] T. S. Mujber, T. Szecsi, and M. S. Hashmi. "Virtual reality applications in manufacturing process simulation." In: *Journal of Materials Processing Technology* 155-156.1-3 (Nov. 2004), pp. 1834–1838. issn: 09240136. doi: 10.1016/j.jmatprotec.2004.04.401.

[29] M. Slater and S. Wilbur. "A framework for immersive virtual environments (FIVE): Speculations on the role of presence in virtual environments." In: *Presence: Teleoperators and Virtual Environments* 6.6 (Dec. 1997), pp. 603–616. issn: 10547460. doi: 10.1162/pres.1997.6.6.603. url: https://www.mitpressjournals.org/doi/abs/10.1162/pres.1997.6.6.603.

[30] K. Bhagat, W.-K. Liou, and C.-Y. Chang. "A cost-effective interactive 3D virtual reality system applied to military live firing training." In: *Virtual Reality* 20.2 (2016), pp. 127–140. doi: 10.1007/s10055-016-0284-x.

[31] H.-K. Wu, S.-Y. Lee, H.-Y. Chang, and J.-C. Liang. "Current status, opportunities and challenges of augmented reality in education." In: *Computers and Education* 62 (2013), pp. 41–49. doi: 10.1016/j.compedu.2012.10.024.

[32] R. Reznick and H. MacRae. "Teaching surgical skills - Changes in the wind." In: *New England Journal of Medicine* 355.25 (2006), pp. 2664–2669. doi: 10.1056/NEJMra054785.

[33] B. Bideau, R. Kulpa, N. Vignais, S. Brault, F. Multon, and C. Craig. "Using virtual reality to analyze sports performance." In: *IEEE Computer Graphics and Applications* 30.2 (Mar. 2010), pp. 14–21. issn: 02721716. doi: 10.1109/MCG.2009.134.

[34] P. M. Emmelkamp and K. Meyerbröker. "Virtual Reality Therapy in Mental Health." In: *Annual Review of Clinical Psychology* 17 (2021), pp. 495–519. issn: 15485951. doi: 10.1146/annurev-clinpsy-081219-115923.

[35] D. Freeman, S. Reeve, A. Robinson, A. Ehlers, D. Clark, B. Spanlang, and M. Slater. "Virtual reality in the assessment, understanding, and treatment of mental health disorders." In: *Psychological Medicine* 47.14 (2017), pp. 2393–2400. issn: 14698978. doi: 10.1017/S003329171700040X.

[36] V. L. Feigin et al. "Global, regional, and national burden of neurological disorders, 1990–2016: a systematic analysis for the Global Burden of Disease Study 2016." In: *The Lancet Neurology* 18.5 (May 2019), pp. 459–480. issn: 14744465. doi: 10.1016/S1474-4422(18)30499-X.

[37] A. Gustavsson, M. Svensson, F. Jacobi, C. Allgulander, J. Alonso, E. Beghi, R. Dodel, M. Ekman, C. Faravelli, L. Fratiglioni, B. Gannon, D. H. Jones, P. Jennum, A. Jordanova, L. Jönsson, K. Karampampa, M. Knapp, G. Kobelt, T. Kurth, R. Lieb, M. Linde, C. Ljungcrantz, A. Maercker, B. Melin, M. Moscarelli, A. Musayev, F. Norwood, M. Preisig, M. Pugliatti, J. Rehm, L. Salvador-Carulla, B. Schlehofer, R. Simon, H.-C. Steinhausen, L. J. Stovner, J.-M. Vallat, P. Van Den Bergh Ap, J. Van Os, P. Vos, W. Xu, H.-U. Wittchen, B. Jönsson, and J. Olesen. "Cost of disorders of the brain in Europe 2010." In: (2011). doi: 10.1016/j.euroneuro.2011.08.008.

[38] *Neurorehabilitation Devices Market Segment, Size, Share, Global Trends, 2025 | MRFR*. url: https://www.marketresearchfuture.com/reports/neurorehabilitation-devices-market-2210 (visited on 11/30/2020).

[39] *FACT SHEET: The burden of neurological diseases on Europe*. Tech. rep. url: www.ean.org.

[40] D. Cano Porras, P. Siemonsma, R. Inzelberg, G. Zeilig, and M. Plotnik. "Advantages of virtual reality in the rehabilitation of balance and gait: Systematic review." eng. In: *Neurology* 90.22 (May 2018), pp. 1017–1025. issn: 1526-632X (Electronic). doi: 10.1212/WNL.0000000000005603.

[41] M. Covarrubias, M. Bordegoni, M. Rosini, E. Guanziroli, U. Cugini, and F. Molteni. “VR system for rehabilitation based on hand gestural and olfactory interaction.” In: *Proceedings of the ACM Symposium on Virtual Reality Software and Technology, VRST* 13-15-Nove (Nov. 2015), pp. 117–120. doi: 10.1145/2821592.2821619.

[42] K. E. Laver, B. Lange, S. George, J. E. Deutsch, G. Saposnik, and M. Crotty. “Virtual reality for stroke rehabilitation.” In: *The Cochrane database of systematic reviews* 11.11 (Nov. 2017). issn: 1469-493X. doi: 10.1002/14651858.CD008349.PUB4. url: https://pubmed.ncbi.nlm.nih.gov/29156493/.

[43] “Virtual Reality for Physical and Motor Rehabilitation.” In: Virtual Reality Technologies for Health and Clinical Applications (2014). Ed. by P. L. Weiss, E. A. Keshner, and M. F. Levin. doi: 10.1007/978-1-4939-0968-1. url: http://link.springer.com/10.1007/978-1-4939-0968-1.

[44] J. A. Kleim and T. A. Jones. “Principles of experience-dependent neural plasticity: implications for rehabilitation after brain damage.” In: *Journal of speech, language, and hearing research : JSLHR* 51.1 (Feb. 2008). issn: 1092-4388. doi: 10.1044/1092-4388(2008/018). url: https://pubmed.ncbi.nlm.nih.gov/18230848/.

[45] P. Kiper, L. Piron, A. Turolla, J. Stozek, and P. Tonin. “The effectiveness of reinforced feedback in virtual environment in the first 12 months after stroke.” In: *Neurologia i neurochirurgia polska* 45.5 (2011), pp. 436–444. issn: 0028-3843. doi: 10.1016/S0028-3843(14)60311-X. url: https://pubmed.ncbi.nlm.nih.gov/22127938/.

[46] P. Dias, R. Silva, P. Amorim, J. Laíns, E. Roque, I. Serôdio, F. Pereira, B. S. Santos, and M. Potel. “Using Virtual Reality to Increase Motivation in Poststroke Rehabilitation: VR Therapeutic Mini-Games Help in Poststroke Recovery.” In: *IEEE Computer Graphics and Applications* 39.1 (Jan. 2019), pp. 64–70. issn: 15581756. doi: 10.1109/MCG.2018.2875630.

[47] J. A. Kleim. “Neural plasticity and neurorehabilitation: teaching the new brain old tricks.” In: *Journal of communication disorders* 44.5 (Sept. 2011), pp. 521–528. issn: 1873-7994. doi: 10.1016/J.JCOMDIS.2011.04.006. url: https://pubmed.ncbi.nlm.nih.gov/21600589/.

[48] R. J. Nudo, G. W. Milliken, W. M. Jenkins, and M. M. Merzenich. “Use-dependent alterations of movement representations in primary motor cortex of adult squirrel monkeys.” In: *Journal of Neuroscience* 16.2 (Jan. 1996), pp. 785–807. issn: 02706474. doi: 10.1523/jneurosci.16-02-00785.1996. url: https://pubmed.ncbi.nlm.nih.gov/8551360/.

[49] D. E. Levac, M. E. Huber, and D. Sternad. “Learning and transfer of complex motor skills in virtual reality: A perspective review.” In: *Journal of NeuroEngineering and Rehabilitation* 16.1 (2019), pp. 1–15. issn: 17430003. doi: 10.1186/s12984-019-0587-8.

[50] D. Sternad, M. E. Huber, and N. Kuznetsov. "Acquisition of novel and complex motor skills: Stable solutions where intrinsic noise matters less." In: *Advances in Experimental Medicine and Biology* 826 (2014), pp. 101–124. issn: 22148019. doi: 10.1007/978-1-4939-1338-1_8/COVER/. url: https://link.springer.com/chapter/10.1007/978-1-4939-1338-1{_}8.

[51] M. M. North and S. M. North. "Virtual reality therapy for treatment of psychological disorders." In: *Career Paths in Telemental Health*. Springer International Publishing, Jan. 2016, pp. 263–268. isbn: 9783319237367. doi: 10.1007/978-3-319-23736-7_27.

[52] M. Krijn, P. M. Emmelkamp, R. P. Olafsson, and R. Biemond. *Virtual reality exposure therapy of anxiety disorders: A review*. July 2004. doi: 10.1016/j.cpr.2004.04.001.

[53] E. Schiza, M. Matsangidou, K. Neokleous, and C. S. Pattichis. "Virtual Reality Applications for Neurological Disease: A Review." In: *Frontiers in Robotics and AI* 6 (Oct. 2019), p. 100. issn: 22969144. doi: 10.3389/FROBT.2019.00100/BIBTEX.

[54] S. E. Nadeau. "A paradigm shift in neurorehabilitation." In: *The Lancet. Neurology* 1.2 (Feb. 2002), pp. 126–130. issn: 1474-4422. doi: 10.1016/S1474-4422(02)00044-3. url: https://pubmed.ncbi.nlm.nih.gov/12849517/.

[55] V. S. Ramachandran. "Plasticity and functional recovery in neurology." In: *Clinical Medicine* 5.4 (July 2005), p. 368. issn: 14702118. doi: 10.7861/CLINMEDICINE.5-4-368. url: /pmc/articles/PMC4954210/?report=abstracthttps://www.ncbi.nlm.nih.gov/pmc/articles/PMC4954210/.

[56] L. J. Gonzalez Rothi and A. M. Barrett. "Treatment innovation in behavioral rehabilitation of stroke: Removing limits on recovery." In: (). doi: 10.1682/JRRD.2006.08.0086.

[57] C. H. Lewis and M. J. Griffin. "Human factors consideration in clinical applications of virtual reality." In: *Studies in Health Technology and Informatics* 44 (1997), pp. 35–56. issn: 18798365. doi: 10.3233/978-1-60750-888-5-35.

[58] S. Weech, S. Kenny, and M. Barnett-Cowan. "Presence and Cybersickness in Virtual Reality Are Negatively Related: A Review." In: *Frontiers in psychology* 10.2 (Feb. 2019). issn: 1664-1078. doi: 10.3389/FPSYG.2019.00158. url: https://pubmed.ncbi.nlm.nih.gov/30778320/.

[59] H. Huygelier, E. Mattheus, V. Vanden Abeele, R. Van Ee, and C. R. Gillebert. "The use of the term virtual reality in post-stroke Rehabilitation: A scoping review and commentary." In: *Psychologica Belgica* 61.1 (June 2021), pp. 145–162. issn: 2054670X. doi: 10.5334/PB.1033/METRICS/. url: http://www.psychologicabelgica.com/articles/10.5334/pb.1033/http://www.psychologicabelgica.com/article/10.5334/pb.1033/.

[60] J. Haik, A. Tessone, A. Nota, D. Mendes, L. Raz, O. Goldan, E. Regev, E. Winkler, E. Mor, A. Orenstein, and I. Hollombe. “The use of video capture virtual reality in burn rehabilitation: The possibilities.” In: *Journal of Burn Care and Research* 27.2 (Mar. 2006), pp. 195–197. issn: 1559047X. doi: 10.1097/01.BCR.0000200890.34084.76.

[61] S. Albiol-Pérez, M. Forcano-García, M. T. Muñoz-Tomás, P. Manzano-Fernández, S. Solsona-Hernández, M. A. Mashat, and J. A. Gil-Gómez. “A novel virtual motor rehabilitation system for guillain-barré syndrome: Two single case studies.” In: *Methods of Information in Medicine* 54.2 (Jan. 2015), pp. 127–134. issn: 2511705X. doi: 10.3414/ME14-02-0002. url: http://www.thieme-connect.de/DOI/DOI?10.3414/ME14-02-0002.

[62] S. Vogt, N. Skjaeret-Maroni, D. Neuhaus, and J. Baumeister. “Virtual reality interventions for balance prevention and rehabilitation after musculoskeletal lower limb impairments in young up to middle-aged adults: A comprehensive review on used technology, balance outcome measures and observed effects.” In: *INTERNATIONAL JOURNAL OF MEDICAL INFORMATICS* 126 (June 2019), pp. 46–58. issn: 1386-5056. doi: 10.1016/j.ijmedinf.2019.03.009.

[63] F. A. Barbieri and R. Vitório. “Locomotion and Posture in Older Adults: The Role of Aging and Movement Disorders.” In: *Locomotion and Posture in Older Adults: The Role of Aging and Movement Disorders* (Feb. 2017), pp. 1–457. doi: 10.1007/978-3-319-48980-3.

[64] R. D. Seidler, J. A. Bernard, T. B. Burutolu, B. W. Fling, M. T. Gordon, J. T. Gwin, Y. Kwak, and D. B. Lipps. “Motor control and aging: Links to age-related brain structural, functional, and biochemical effects.” In: *Neuroscience and Biobehavioral Reviews* 34.5 (Apr. 2010), pp. 721–733. issn: 01497634. doi: 10.1016/J.NEUBIOREV.2009.10.005.

[65] J. H. Van Dieën and M. Pijnappels. “Balance control in older adults.” In: *Locomotion and Posture in Older Adults: The Role of Aging and Movement Disorders* (Feb. 2017), pp. 237–262. doi: 10.1007/978-3-319-48980-3_16/COVER/. url: https://link.springer.com/chapter/10.1007/978-3-319-48980-3{_}16.

[66] M. K. Holden and T. Dyar. “Virtual environment training: A New Tool for Neurorehabilitation.” In: *Neurology Report* 26.2 (2002), pp. 62–71. issn: 1085049X. doi: 10.1097/01253086-200226020-00003.

[67] M. van Diest, C. J. C. Lamoth, J. Stegenga, G. J. Verkerke, and K. Postema. “Exergaming for balance training of elderly: state of the art and future developments.” In: *JOURNAL OF NEUROENGINEERING AND REHABILITATION* 10 (Sept. 2013). doi: 10.1186/1743-0003-10-101.

[68] D. J. Goble, B. L. Cone, and B. W. Fling. *Using the Wii Fit as a tool for balance assessment and neurorehabilitation: the first half decade of ”Wii-search*

. Tech. rep. 2014. doi: 10.1186/1743-0003-11-12. url: http://www.jneuroengrehab.com/content/11/1/12.

[69] E. Tsekleves, A. Warland, C. Kilbride, I. Paraskevopoulos, and D. Skordoulis. "The use of the nintendo wii in motor rehabilitation for virtual reality interventions: A literature review." In: *Intelligent Systems Reference Library* 68 (2014), pp. 321–344. issn: 18684408. doi: 10.1007/978-3-642-54816-1_17.

[70] *ADXL330 Datasheet and Product Info | Analog Devices*. url: https://www.analog.com/en/products/adxl330.html{\#} (visited on 12/23/2020).

[71] "Wii Operations Manual System Setup." In: (2009).

[72] J. A. Gil-Gómez, R. Lloréns, M. Alcñiz, and C. Colomer. "Effectiveness of a Wii balance board-based system (eBaViR) for balance rehabilitation: a pilot randomized clinical trial in patients with acquired brain injury." In: *Journal of NeuroEngineering and Rehabilitation 2011 8:1* 8.1 (May 2011), pp. 1–10. issn: 1743-0003. doi: 10.1186/1743-0003-8-30. url: https://jneuroengrehab.biomedcentral.com/articles/10.1186/1743-0003-8-30.

[73] D. Fitzgerald, N. Trakarnratanakul, B. Smyth, and B. Caulfield. "Effects of a wobble board-based therapeutic exergaming system for balance training on dynamic postural stability and intrinsic motivation levels." In: *Journal of Orthopaedic and Sports Physical Therapy* 40.1 (2010), pp. 11–19. issn: 01906011. doi: 10.2519/jospt.2010.3121. url: https://pubmed.ncbi.nlm.nih.gov/20044704/.

[74] D Fitzgerald, N Trakarnratanakul, L Dunne, B Smyth, and B Caulfield. "Development and user evaluation of a virtual rehabilitation system for wobble board balance training." In: *2008 30th Annual International Conference of the IEEE Engineering in Medicine and Biology Society*. 2008, pp. 4194–4198. isbn: 1558-4615 VO -. doi: 10.1109/IEMBS.2008.4650134.

[75] A. L. Betker, T. Szturm, and Z. Moussavi. "Development of an interactive motivating tool for rehabilitation movements." In: *Annual International Conference of the IEEE Engineering in Medicine and Biology - Proceedings* 7 VOLS (2005), pp. 2341–2344. issn: 05891019. doi: 10.1109/IEMBS.2005.1616935.

[76] A. Reinthal, K. Szirony, C. Clark, J. Swiers, M. Kellicker, and S. Linder. "ENGAGE: Guided activity-based gaming in neurorehabilitation after stroke: A pilot study." In: *Stroke Research and Treatment* (2012). issn: 20420056. doi: 10.1155/2012/784232.

[77] G Yavuzer, A Senel, M. B. Atay, and H. J. Stam. ""Playstation eyetoy games" improve upper extremity-related motor functioning in subacute stroke: a randomized controlled clinical trial." In: *Eur. J. Phys. Rehabil. Med.* 44.3 (Sept. 2008), pp. 237–244.

[78] B. Lange, S. Flynn, C. Y. Chang, W. Liang, Y. Si, C. Nanavati, C. L. Chieng, and A. S. Rizzo. "Development of an interactive stepping game to reduce falls in older adults." In: *International Journal on Disability and Human Development* 10.4 (Nov. 2011), pp. 331–335. issn: 21910367. doi: 10.1515/IJDHD.2011.056/MACHINEREADABLECITATION/RIS. url: https://www.degruyter.com/document/doi/10.1515/IJDHD.2011.056/html.

[79] K. Khoshelham and S. O. Elberink. "Accuracy and Resolution of Kinect Depth Data for Indoor Mapping Applications." In: *Sensors 2012, Vol. 12, Pages 1437-1454* 12.2 (Feb. 2012), pp. 1437–1454. issn: 1424-8220. doi: 10.3390/S120201437. url: https://www.mdpi.com/1424-8220/12/2/1437/htmhttps://www.mdpi.com/1424-8220/12/2/1437.

[80] D. S. Park, D. G. Lee, K. Lee, and G. C. Lee. "Effects of Virtual Reality Training using Xbox Kinect on Motor Function in Stroke Survivors: A Preliminary Study." In: *Journal of Stroke and Cerebrovascular Diseases* 26.10 (Oct. 2017), pp. 2313–2319. issn: 15328511. doi: 10.1016/j.jstrokecerebrovasdis.2017.05.019.

[81] N. G. Schaham, G. Zeilig, H. Weingarden, and D. Rand. "Game analysis and clinical use of the Xbox-Kinect for stroke rehabilitation." In: *International journal of rehabilitation research. Internationale Zeitschrift fur Rehabilitationsforschung. Revue internationale de recherches de readaptation* 41.4 (Dec. 2018), pp. 323–330. issn: 1473-5660. doi: 10.1097/MRR.0000000000000302. url: https://pubmed.ncbi.nlm.nih.gov/29994922/.

[82] S. Almasi, H. Ahmadi, F. Asadi, L. Shahmoradi, G. Arji, M. Alizadeh, and H. Kolivand. "Kinect-Based Rehabilitation Systems for Stroke Patients: A Scoping Review." In: *BioMed Research International* 2022 (2022). issn: 23146141. doi: 10.1155/2022/4339054.

[83] E. Bisson, B. Contant, H. Sveistrup, and Y. Lajoie. "Functional balance and dual-task reaction times in older adults are improved by virtual reality and biofeedback training." In: *Cyberpsychology and Behavior* 10.1 (2007), pp. 16–23. issn: 10949313. doi: 10.1089/cpb.2006.9997.

[84] N. M. Kosse, S. R. Caljouw, P. J. Vuijk, and C. J. Lamoth. "Exergaming: Interactive balance training in healthy community-dwelling older adults." In: *Journal of Cyber Therapy and Rehabilitation* 4.3 (2011), pp. 399–407. issn: 17849934.

[85] N. Skjæret, A. Nawaz, T. Morat, D. Schoene, J. L. Helbostad, and B. Vereijken. "Exercise and rehabilitation delivered through exergames in older adults: An integrative review of technologies, safety and efficacy." In: *International Journal of Medical Informatics* 85.1 (2016), pp. 1–16. issn: 18728243. doi: 10.1016/j.ijmedinf.2015.10.008.

[86] R. A. Clark, A. L. Bryant, Y. Pua, P. McCrory, K. Bennell, and M. Hunt. "Validity and reliability of the Nintendo Wii Balance Board for assessment of standing balance." In: *Gait Posture* 31.3 (Mar. 2010), pp. 307–310. issn: 09666362. doi: 10.1016/j.gaitpost.2009.11.012.

[87] J. C. Nitz, S. Kuys, R. Isles, and S. Fu. "Is the Wii Fit a new-generation tool for improving balance, health and well-being? A pilot study." In: *Climacteric* 13.5 (Oct. 2010), pp. 487–491. issn: 13697137. doi: 10.3109/13697130903395193.

[88] R. E. Barclay□Goddard, T. J. Stevenson, W. Poluha, M. Moffatt, and S. P. Taback. "Force platform feedback for standing balance training after stroke." In: *The Cochrane Database of Systematic Reviews* 2004.4 (Oct. 2004). doi: 10.1002/14651858.CD004129.PUB2. url: /pmc/articles/PMC6464938//pmc/articles/PMC6464938/?report=abstracthttps://www.ncbi.nlm.nih.gov/pmc/articles/PMC6464938/.

[89] K. Berg, S Wood-Dauphinee, J Williams, and B. Maki. "Measuring balance in the elderly: Validation of an instrument." In: *Canadian journal of public health. Revue canadienne de santé publique* 83 Suppl 2 (1991), S7–11.

[90] P. D and R. S. "The timed "Up & Go": a test of basic functional mobility for frail elderly persons." In: *Journal of the American Geriatrics Society* 39.2 (1991), pp. 142–148. issn: 0002-8614. doi: 10.1111/J.1532-5415.1991.TB01616.X. url: https://pubmed.ncbi.nlm.nih.gov/1991946/.

[91] M. A. Williams, R. L. Soiza, A. M. E. Jenkinson, and A. Stewart. "EXercising with Computers in Later Life (EXCELL) - pilot and feasibility study of the acceptability of the Nintendo® WiiFit in community-dwelling fallers." In: *BMC research notes* 3 (2010). issn: 1756-0500. doi: 10.1186/1756-0500-3-238. url: https://pubmed.ncbi.nlm.nih.gov/20831836/.

[92] G. Barry, B. Galna, and L. Rochester. *The role of exergaming in Parkinson's disease rehabilitation: A systematic review of the evidence*. Mar. 2014. doi: 10.1186/1743-0003-11-33.

[93] W. C. Yang, H. K. Wang, R. M. Wu, C. S. Lo, and K. H. Lin. "Home-based virtual reality balance training and conventional balance training in Parkinson's disease: A randomized controlled trial." In: *Journal of the Formosan Medical Association = Taiwan yi zhi* 115.9 (Sept. 2016), pp. 734–743. issn: 0929-6646. doi: 10.1016/J.JFMA.2015.07.012. url: https://pubmed.ncbi.nlm.nih.gov/26279172/.

[94] M. M. Lee, D. C. Shin, and C. H. Song. "Canoe game-based virtual reality training to improve trunk postural stability, balance, and upper limb motor function in subacute stroke patients: a randomized controlled pilot study." In: *Journal of physical therapy science* 28.7 (July 2016), pp. 2019–2024. issn: 0915-5287. doi: 10.1589/JPTS.28.2019. url: https://pubmed.ncbi.nlm.nih.gov/27512255/.

[95] T. In, K. Lee, and C. Song. "Virtual Reality Reflection Therapy Improves Balance and Gait in Patients with Chronic Stroke: Randomized Controlled Trials." eng. In: *Medical science monitor : international medical journal of experimental and clinical research* 22 (Oct. 2016), pp. 4046–4053. issn: 1643-3750 (Electronic). doi: 10.12659/msm.898157.

[96] B. Lange, S. Flynn, R. Proffitt, C.-Y. Chang, and A. S. Rizzo. "Development of an interactive game-based rehabilitation tool for dynamic balance training." eng. In: *Topics in stroke rehabilitation* 17.5 (2010), pp. 345–352. issn: 1074-9357 (Print). doi: 10.1310/tsr1705-345.

[97] G. C. Burdea and K. Polistico. "A review of integrative virtual reality games for rehabilitation." In: *2017 E-Health and Bioengineering Conference, EHB 2017* (2017), pp. 733–736. doi: 10.1109/EHB.2017.7995528.

[98] S. Viñas-Diz and M. Sobrido-Prieto. "Virtual reality for therapeutic purposes in stroke: A systematic review." In: *Neurología (English Edition)* 31.4 (2016), pp. 255–277. issn: 21735808. doi: 10.1016/j.nrleng.2015.06.007.

[99] G. Juras, A. Brachman, J. Michalska, A. Kamieniarz, M. Pawłowski, A. Hadamus, D. Białoszewski, J. Błaszczyk, and K. J. Słomka. "Standards of Virtual Reality Application in Balance Training Programs in Clinical Practice: A Systematic Review." In: *Games for Health Journal* 8.2 (2019), pp. 101–111. issn: 21617856. doi: 10.1089/g4h.2018.0034.

[100] *PEDro scale - PEDro*. url: https://pedro.org.au/english/resources/pedro-scale/ (visited on 12/30/2020).

[101] S. G. Neri, J. R. Cardoso, L. Cruz, R. M. Lima, R. J. De Oliveira, M. D. Iversen, and R. L. Carregaro. "Do virtual reality games improve mobility skills and balance measurements in community-dwelling older adults? Systematic review and meta-analysis." In: *Clinical Rehabilitation* 31.10 (2017), pp. 1292–1304. issn: 14770873. doi: 10.1177/0269215517694677.

[102] "Virtual Reality: Through the New Looking Glass, Ken Pimentel and Kevin Teixeira. 1993. Windcrest/McGraw-Hill/TAB Books, Blue Ridge Summit, PA. 352 pages. ISBN: 0-8306-4065-7 (hc); 0-8306-4064-9 (pb). $32.95(hc)$;22.95 (pb." In: *Bulletin of Science, Technology & Society* 14.4 (1994), pp. 224–224. issn: 0270-4676. doi: 10.1177/027046769401400415.

[103] J. J. Cummings and J. N. Bailenson. "How Immersive Is Enough? A Meta-Analysis of the Effect of Immersive Technology on User Presence." In: *http://dx.doi.org/10.1080/15213269.2015.1015740* 19.2 (Apr. 2015), pp. 272–309. issn: 15213269. doi: 10.1080/15213269.2015.1015740. url: https://www.tandfonline.com/doi/abs/10.1080/15213269.2015.1015740.

[104] J. M. Juliano, R. P. Spicer, A. Vourvopoulos, S. Lefebvre, K. Jann, T. Ard, E. Santarnecchi, D. M. Krum, and S.-L. Liew. "Embodiment Is Related to Better Performance on a Brain–Computer Interface in Immersive Virtual Reality: A Pilot Study." In: *Sensors* 20.4 (Feb. 2020), p. 1204. issn: 1424-8220. doi: 10.3390/s20041204. url: https://www.mdpi.com/1424-8220/20/4/1204.

[105] D. Saldana, M. Neureither, A. Schmiesing, E. Jahng, L. Kysh, S. C. Roll, and S. L. Liew. "Applications of head-mounted displays for virtual reality in adult physical rehabilitation: A scoping review." In: *American Journal of Occupational Therapy* 74.5 (2020), pp. 1–15. issn: 19437676. doi: 10.5014/ajot.2020.041442.

[106] G. Tieri, G. Morone, S. Paolucci, and M. Iosa. "Virtual reality in cognitive and motor rehabilitation: facts, fiction and fallacies." In: *Expert review of medical devices* 15.2 (Feb. 2018), pp. 107–117. issn: 1745-2422. doi: 10.1080/17434440.2018.1425613. url: https://pubmed.ncbi.nlm.nih.gov/29313388/.

[107] C. Cruz-Neira, D. J. Sandin, T. A. DeFanti, R. V. Kenyon, and J. C. Hart. "The CAVE: Audio Visual Experience Automatic Virtual Environment." In: *Communications of the ACM* 35.6 (Jan. 1992), pp. 64–72. issn: 15577317. doi: 10.1145/129888.129892. url: https://dl.acm.org/doi/10.1145/129888.129892.

[108] M. C. Cabral, C. H. Morimoto, and M. K. Zuffo. "On the usability of gesture interfaces in virtual reality environments." In: *ACM International Conference Proceeding Series* 124 (2005), pp. 100–108. doi: 10.1145/1111360.1111370.

[109] M. A. Muhanna. "Virtual reality and the CAVE: Taxonomy, interaction challenges and research directions." In: *Journal of King Saud University - Computer and Information Sciences* 27.3 (2015), pp. 344–361. issn: 22131248. doi: 10.1016/j.jksuci.2014.03.023. url: http://dx.doi.org/10.1016/j.jksuci.2014.03.023.

[110] *Virtual Reality CAVE Solutions | Antycip*. url: https://steantycip.com/vr-cave/ (visited on 06/25/2022).

[111] Z. Campbell, K. K. Zakzanis, D. Jovanovski, S. Joordens, R. Mraz, and S. J. Graham. "Utilizing Virtual Reality to Improve the Ecological Validity of Clinical Neuropsychology: An fMRI Case Study Elucidating the Neural Basis of Planning by Comparing the Tower of London with a Three-Dimensional Navigation Task." In: *http://dx.doi.org/10.1080/09084280903297891* 16.4 (2009), pp. 295–306. issn: 15324826. doi: 10.1080/09084280903297891. url: https://www.tandfonline.com/doi/abs/10.1080/09084280903297891.

[112] T. D. Parsons. "Virtual reality for enhanced ecological validity and experimental control in the clinical, affective and social neurosciences." In: *Frontiers in Human Neuroscience* 9.DEC (Dec. 2015). issn: 16625161. doi: 10.3389/fnhum.2015.00660. url: /pmc/articles/PMC4675850/

/pmc/articles/PMC4675850/?report=abstracthttps://www.ncbi.nlm.nih.gov/pmc/articles/PMC4675850/.

[113] A. S. Rizzo and S. T. Koenig. “Is clinical virtual reality ready for primetime?” In: *Neuropsychology* 31.8 (Nov. 2017), pp. 877–899. issn: 1931-1559. doi: 10.1037/NEU0000405. url: https://pubmed.ncbi.nlm.nih.gov/29376669/.

[114] T. Rose, C. S. Nam, and K. B. Chen. “Immersion of virtual reality for rehabilitation - Review.” In: *Applied Ergonomics* 69.February 2017 (2018), pp. 153–161. issn: 18729126. doi: 10.1016/j.apergo.2018.01.009. url: https://doi.org/10.1016/j.apergo.2018.01.009.

[115] D. C. Niehorster, L. Li, and M. Lappe. “The accuracy and precision of position and orientation tracking in the HTC vive virtual reality system for scientific research.” In: *i-Perception* 8.3 (June 2017), pp. 1–23. issn: 20416695. doi: 10.1177/2041669517708205.

[116] A. Borrego, J. Latorre, M. Alcañiz, and R. Llorens. “Comparison of Oculus Rift and HTC Vive: Feasibility for Virtual Reality-Based Exploration, Navigation, Exergaming, and Rehabilitation.” In: *Games for Health Journal* 7.3 (2018), pp. 151–156. issn: 21617856. doi: 10.1089/g4h.2017.0114.

[117] H. Alhasan, V. Hood, and F. Mainwaring. “The effect of visual biofeedback on balance in elderly population: a systematic review.” In: *Clinical Interventions in Aging* 12 (Mar. 2017), p. 487. issn: 11781998. doi: 10.2147/CIA.S127023. url: /pmc/articles/PMC5345985//pmc/articles/PMC5345985/?report=abstracthttps://www.ncbi.nlm.nih.gov/pmc/articles/PMC5345985/.

[118] C. Gagliardi, A. C. Turconi, E. Biffi, C. Maghini, A. Marelli, A. Cesareo, E. Diella, and D. Panzeri. “Immersive Virtual Reality to Improve Walking Abilities in Cerebral Palsy: A Pilot Study.” In: (2039). doi: 10.1007/s10439-018-2039-1. url: https://doi.org/10.1007/s10439-018-2039-1.

[119] C. G. Horlings, M. G. Carpenter, U. M. Küng, F. Honegger, B. Wiederhold, and J. H. Allum. “Influence of virtual reality on postural stability during movements of quiet stance.” In: *Neuroscience Letters* 451.3 (Feb. 2009), pp. 227–231. issn: 03043940. doi: 10.1016/j.neulet.2008.12.057.

[120] H. Akizuki, A. Uno, K. Arai, S. Morioka, S. Ohyama, S. Nishiike, K. Tamura, and N. Takeda. “Effects of immersion in virtual reality on postural control.” In: *Neuroscience letters* 379.1 (Apr. 2005), pp. 23–26. issn: 0304-3940. doi: 10.1016/J.NEULET.2004.12.041. url: https://pubmed.ncbi.nlm.nih.gov/15814192/https://pubmed.ncbi.nlm.nih.gov/15814192/?dopt=Abstract.

[121] M. C. Howard. “A meta-analysis and systematic literature review of virtual reality rehabilitation programs.” In: *Computers in Human Behavior* 70 (2017), pp. 317–327. issn: 07475632. doi: 10.1016/j.chb.2017.01.013. url: http://dx.doi.org/10.1016/j.chb.2017.01.013.

[122] I. Bortone, M. Barsotti, D. Leonardis, A. Crecchi, A. Tozzini, L. Bonfiglio, and A. Frisoli. “Immersive Virtual Environments and Wearable Haptic Devices in rehabilitation of children with neuromotor impairments: a single-blind randomized controlled crossover pilot study.” In: *Journal of NeuroEngineering and Rehabilitation* 17.1 (Dec. 2020), pp. 1–14. issn: 17430003. doi: 10.1186/S12984-020-00771-6/TABLES/2. url: https://jneuroengrehab.biomedcentral.com/articles/10.1186/s12984-020-00771-6.

[123] Y. P. Chen and A. M. Howard. “Effects of robotic therapy on upper-extremity function in children with cerebral palsy: a systematic review.” In: *Develop Neurorehab* 19.1 (Jan. 2016), pp. 64–71. issn: 17518431. doi: 10.3109/17518423.2014.899648.

[124] O. M. Giggins, U. M. C. Persson, and B. Caulfield. “Biofeedback in rehabilitation.” In: *Journal of NeuroEngineering and Rehabilitation* 10.1 (2013), pp. 1–11. issn: 17430003. doi: 10.1186/1743-0003-10-60.

[125] J. A. Onate, K. M. Guskiewicz, and R. J. Sullivan. “Augmented feedback reduces jump landing forces.” In: *Journal of Orthopaedic and Sports Physical Therapy* 31.9 (2001), pp. 511–517. issn: 01906011. doi: 10.2519/jospt.2001.31.9.511. url: https://pubmed.ncbi.nlm.nih.gov/11570735/.

[126] M. Jahanshahi. “Biofeedback: Principles and Practice for Clinicians.” In: *Journal of Neurology, Neurosurgery & Psychiatry* 47.11 (Nov. 1984), pp. 1267–1267. issn: 0022-3050. doi: 10.1136/jnnp.47.11.1267.

[127] J. W. Yoo, D. R. Lee, Y. J. Cha, and S. H. You. “Augmented effects of EMG biofeedback interfaced with virtual reality on neuromuscular control and movement coordination during reaching in children with cerebral palsy.” In: *NeuroRehabilitation* 40.2 (2017), pp. 175–185. issn: 1878-6448. doi: 10.3233/NRE-161402. url: https://pubmed.ncbi.nlm.nih.gov/28222541/.

[128] R. Lüddecke and A. Felnhofer. “Virtual Reality Biofeedback in Health: A Scoping Review.” In: *Applied Psychophysiology Biofeedback* 47.1 (Mar. 2022), pp. 1–15. issn: 10900586. doi: 10.1007/S10484-021-09529-9/FIGURES/1. url: https://link.springer.com/article/10.1007/s10484-021-09529-9.

[129] G. Dermody, L. Whitehead, G. Wilson, and C. Glass. “The role of virtual reality in improving health outcomes for community-dwelling older adults: Systematic review.” In: *Journal of Medical Internet Research* 22.6 (2020). issn: 14388871. doi: 10.2196/17331.

[130] P. Parijat, T. Lockhart, and J. Liu. “EMG and kinematic responses to unexpected slips after slip training in virtual reality.” In: *IEEE Transactions on Biomedical Engineering* 62.2 (2015), pp. 593–599. doi: `10.1109/TBME.2014.2361324`.

[131] M. F. Gago, D. Yelshyna, E. Bicho, H. D. Silva, L. Rocha, M. Lurdes Rodrigues, and N. Sousa. “Compensatory Postural Adjustments in an Oculus Virtual Reality Environment and the Risk of Falling in Alzheimer’s Disease.” In: *Dementia and Geriatric Cognitive Disorders Extra* 6.2 (2016), pp. 252–267. issn: 16645464. doi: `10.1159/000447124`.

[132] P. Parijat, T. Lockhart, and J. Liu. “Effects of Perturbation-Based Slip Training Using a Virtual Reality Environment on Slip-induced Falls.” In: *Annals of Biomedical Engineering* 43.4 (2015), pp. 958–967. doi: `10.1007/s10439-014-1128-z`.

[133] F. Levy, P. Leboucher, G. Rautureau, O. Komano, B. Millet, and R. Jouvent. “Fear of falling: Efficacy of virtual reality associated with serious games in elderly people.” In: *Neuropsychiatric Disease and Treatment* 12 (2016), pp. 877–881. issn: 11782021. doi: `10.2147/NDT.S97809`.

[134] S. Benham, M. Kang, and N. Grampurohit. “Immersive Virtual Reality for the Management of Pain in Community-Dwelling Older Adults.” In: *OTJR Occupation, Participation and Health* 39.2 (Apr. 2019), pp. 90–96. issn: 19382383. doi: `10.1177/1539449218817291`. url: `https://journals.sagepub.com/doi/10.1177/1539449218817291`.

[135] P. J. White and Z. Moussavi. “Neurocognitive treatment for a patient with Alzheimer’s disease using a virtual reality navigational environment.” In: *Journal of Experimental Neuroscience* 2016.10 (Nov. 2016), pp. 129–135. issn: 11790695. doi: `10.4137/JEn.s40827`. url: `https://journals.sagepub.com/doi/10.4137/JEN.S40827`.

[136] J. Klenk, C. Becker, F. Lieken, S. Nicolai, W. Maetzler, W. Alt, W. Zijlstra, J. M. Hausdorff, R. C. van Lummel, L. Chiari, and U. Lindemann. “Comparison of acceleration signals of simulated and real-world backward falls.” In: *Medical engineering & physics* 33.3 (Apr. 2011), pp. 368–373. issn: 1873-4030. doi: `10.1016/J.MEDENGPHY.2010.11.003`. url: `https://pubmed.ncbi.nlm.nih.gov/21123104/`.

[137] E. L. Altschuler. “Interaction of vision and movement via a mirror.” In: *Perception* 34.9 (Sept. 2005), pp. 1153–1155. issn: 03010066. doi: `10.1068/p3409bn`. url: `http://journals.sagepub.com/doi/10.1068/p3409bn`.

[138] E. A. Bilodeau and I. M. Bilodeau. “Motor-Skills Learning.” In: *Annual Review of Psychology* 12.1 (Jan. 1961), pp. 243–280. issn: 0066-4308. doi: `10.1146/annurev.ps.12.020161.001331`. url: `http://www.annualreviews.org/doi/10.1146/annurev.ps.12.020161.001331`.

[139] E. Todorov, R. Shadmehr, and E. Bizzi. “Augmented feedback presented in a virtual environment accelerates learning of a difficult motor task.” In: *Journal of Motor Behavior* 29.2 (1997), pp. 147–158. issn: 00222895. doi: 10.1080/00222899709600829. url: https://www.tandfonline.com/action/journalInformation?journalCode=vjmb20.

[140] F. D. Rose, E. A. Attree, B. M. Brooks, D. M. Parslow, and P. R. Penn. “Training in virtual environments: Transfer to real world tasks and equivalence to real task training.” In: *Ergonomics* 43.4 (Apr. 2000), pp. 494–511. issn: 13665847. doi: 10.1080/001401300184378. url: http://www.tan.

[141] F. P. Brooks. “What’s real about virtual reality?” In: *IEEE Computer Graphics and Applications* 19.6 (Nov. 1999), pp. 16–27. issn: 02721716. doi: 10.1109/38.799723.

[142] M. K. Holden. “Virtual environments for motor rehabilitation: Review.” In: *Cyberpsychology and Behavior* 8.3 (2005), pp. 187–211. issn: 10949313. doi: 10.1089/cpb.2005.8.187.

[143] M. T. Schultheis and A. A. Rizzo. *The application of virtual reality technology in rehabilitation*. 2001. doi: 10.1037/0090-5550.46.3.296.

[144] L. P. Fried, C. M. Tangen, J. Walston, A. B. Newman, C. Hirsch, J. Gottdiener, T. Seeman, R. Tracy, W. J. Kop, G. Burke, and M. A. Mcburnie. *Frailty in Older Adults: Evidence for a Phenotype*. Tech. rep. 3. 2001, pp. 146–156. url: https://academic.oup.com/biomedgerontology/article/56/3/M146/545770.

[145] M. Y. Osoba, A. K. Rao, S. K. Agrawal, and A. K. Lalwani. “Balance and gait in the elderly: A contemporary review.” In: *Laryngoscope Investigative Otolaryngology* 4.1 (Feb. 2019), pp. 143–153. doi: 10.1002/lio2.252. url: https://pubmed.ncbi.nlm.nih.gov/30828632/.

[146] S. V. Adamovich, G. G. Fluet, E. Tunik, and A. S. Merians. “Sensorimotor training in virtual reality: A review.” In: *NeuroRehabilitation* 25.1 (2009), pp. 29–44. issn: 10538135. doi: 10.3233/NRE-2009-0497.

[147] P. N. Wilson, N. Foreman, and D. Stanton. *Virtual reality, disability and rehabilitation*. 1997. doi: 10.3109/09638289709166530. url: https://www.tandfonline.com/doi/abs/10.3109/09638289709166530.

[148] G. C. Burdea. “Virtual rehabilitation - benefits and challenges.” In: *Methods Inf Med* 42 (2003). doi: 10.1267/METH03050519.

[149] C. J. Bohil, B. Alicea, and F. A. Biocca. *Virtual reality in neuroscience research and therapy*. Dec. 2011. doi: 10.1038/nrn3122. url: https://www.nature.com/articles/nrn3122.

[150] B. Lange, S. Koenig, C. Y. Chang, E. McConnell, E. Suma, M. Bolas, and A. Rizzo. "Designing informed game-based rehabilitation tasks leveraging advances in virtual reality." In: *Disability and Rehabilitation* 34.22 (2012), pp. 1863–1870. issn: 09638288. doi: 10.3109/09638288.2012.670029. url: https://www.tandfonline.com/doi/abs/10.3109/09638288.2012.670029.

[151] L. Ke, B Lange, S George, D. Je, G Saposnik, and M Crotty. "Virtual reality for stroke rehabilitation (Review)." In: 11 (2018). doi: 10.1002/14651858.CD008349.pub4.www.cochranelibrary.com.

[152] A. Henderson, N. Korner-Bitensky, and M. Levin. *Virtual reality in stroke rehabilitation: A systematic review of its effectiveness for upper limb motor recovery*. Mar. 2007. doi: 10.1310/tsr1402-52.

[153] T. Massetti, T. Dias, T. B. Crocetta, R. Guarnieri, B. L. D. Freitas, P. B. Lopes, S. Watson, J. Tonks, C. Bandeira, and D. M. Monteiro. "The Clinical Utility of Virtual Reality in Neurorehabilitation : A Systematic Review." In: (2018). doi: 10.1177/1179573518813541.

[154] D. K. Ravi, N. Kumar, and P. Singhi. *Effectiveness of virtual reality rehabilitation for children and adolescents with cerebral palsy: an updated evidence-based systematic review*. Sept. 2017. doi: 10.1016/j.physio.2016.08.004.

[155] A. Mansfield, J. S. Wong, J. Bryce, S. Knorr, and K. K. Patterson. *Does Perturbation-Based Balance Training Prevent Falls? Systematic Review and Meta-Analysis of Preliminary Randomized Controlled Trials Background. Older adults and individuals with neurological conditions are at an*. Tech. rep. 2015. url: https://academic.oup.com/ptj/article/95/5/700/2686424.

[156] N Bugnariu and J Fung. "Aging and selective sensorimotor strategies in the regulation of upright balance." In: *Journal of NeuroEngineering and Rehabilitation* 4 (2007). doi: 10.1186/1743-0003-4-19. url: https://www.scopus.com/inward/record.uri?eid=2-s2.0-34447115927{\&}doi=10.1186{\%}2F1743-0003-4-19{\&}partnerID=40{\&}md5=f470f1937b1b2d72ed3166163e2ebc2a.

[157] M. Y. Osaba, D Martelli, A Prado, S. K. Agrawal, and A. K. Lalwani. "Age-related differences in gait adaptations during overground walking with and without visual perturbations using a virtual reality headset." In: *Scientific Reports* 10.1 (2020). doi: 10.1038/s41598-020-72408-6. url: https://www.scopus.com/inward/record.uri?eid=2-s2.0-85091321036{\&}doi=10.1038{\%}2Fs41598-020-72408-6{\&}partnerID=40{\&}md5=e4761b60be940a2b71b7b65ed4cf4318.

[158] J Berard, J Fung, and A Lamontagne. "Impact of aging on visual reweighting during locomotion." In: *Clinical Neurophysiology* 123.7 (2012), pp. 1422–1428. doi: 10.1016/j.clinph.2011.11.081. url: https://www.scopus.com/inward/record.uri?eid=2-s2.0-

84861682477{\&}doi=10.1016{\%}2Fj.clinph.2011.11.081{\&}partnerID=40{\&}md5=9aca5dc60d640a5da5769ae6506b9bf6.

[159] J. R. Berard, J. Fung, and A. Lamontagne. "Evidence for the use of rotational optic flow cues for locomotor steering in healthy older adults." In: *JOURNAL OF NEUROPHYSIOLOGY* 106.3 (Sept. 2011), pp. 1089–1096. issn: 0022-3077. doi: 10.1152/jn.00277.2011.

[160] X Zhang and T. E. Lockhart. "The impairment and recovery of dynamic walking stability during virtual environment exposure in the elderly." In: *Proceedings of the Human Factors and Ergonomics Society*. Vol. 2. 2010, pp. 1154–1158. doi: 10.1518/107118110X12829369834122. url: https://www.scopus.com/inward/record.uri?eid=2-s2.0-79952932407{\&}doi=10.1518{\%}2F107118110X12829369834122{\&}partnerID=40{\&}md5=eb46c78666db7a5f309

[161] D Yelshyna, M. F. Gago, E Bicho, V Fernandes, N. F. Gago, L Costa, H Silva, M. L. Rodrigues, L Rocha, and N Sousa. "Compensatory postural adjustments in Parkinson's disease assessed via a virtual reality environment." In: *Behavioural Brain Research* 296 (2016), pp. 384–392. doi: 10.1016/j.bbr.2015.08.017. url: https://www.scopus.com/inward/record.uri?eid=2-s2.0-84961147821{\&}doi=10.1016{\%}2Fj.bbr.2015.08.017{\&}partnerID=40{\&}md5=f5f5a49a7593052b4ad213be6781dddb.

[162] E Chiarovano, W Wang, S. J. Rogers, H. G. MacDougall, I. S. Curthoys, and C de Waele. "Balance in virtual reality: Effect of age and bilateral vestibular loss." In: *Frontiers in Neurology* 8.1 (2017). doi: 10.3389/fneur.2017.00005. url: https://www.scopus.com/inward/record.uri?eid=2-s2.0-85012041171{\&}doi=10.3389{\%}2Ffneur.2017.00005{\&}partnerID=40{\&}md5=1d87ca651894b886499db78330465b9d.

[163] A Diniz-Filho, E. R. Boer, C. P. B. Gracitelli, R. Y. Abe, N Van Driel, Z Yang, and F. A. Medeiros. "Evaluation of postural control in patients with glaucoma using a virtual reality environment." In: *Ophthalmology* 122.6 (2015), pp. 1131–1138. doi: 10.1016/j.ophtha.2015.02.010. url: https://www.scopus.com/inward/record.uri?eid=2-s2.0-84930040814{\&}doi=10.1016{\%}2Fj.ophtha.2015.02.010{\&}partnerID=40{\&}md5=a2a401d56c31af3a05c5e1f77bc9fdb

[164] A. V. Lubetzky, B. D. Hujsak, G. Fu, and K. Perlin. "An oculus rift assessment of dynamic balance by head mobility in a virtual park scene: A pilot study." In: *Motor Control* 23.1 (Jan. 2019), pp. 127–142. issn: 15432696. doi: 10.1123/mc.2018-0001. url: https://pubmed.ncbi.nlm.nih.gov/30518313/.

[165] R Frost, J Skidmore, M Santello, and P Artemiadis. "Sensorimotor control of gait: A novel approach for the study of the interplay of visual and proprioceptive feedback." In: *Frontiers in Human Neuroscience* 9.2 (2015). doi: 10.3389/fnhum.2015.00014. url: https://www.scopus.com/

inward/record.uri?eid=2-s2.0-84933679556{\&}doi=10.3389{\%}2Ffnhum.2015.00014{\&}partnerID=40{\&}md5=674cd67fd9e2cfce98aec3c37afed513.

[166] M Drolet, E. Q. Yumbla, B Hobbs, and P Artemiadis. "On the Effects of Visual Anticipation of Floor Compliance Changes on Human Gait: Towards Model-based Robot-Assisted Rehabilitation." In: *2020 IEEE International Conference on Robotics and Automation (ICRA)*. 2020, pp. 9072–9078. isbn: 2577-087X VO -. doi: 10.1109/ICRA40945.2020.9197536.

[167] S Hagio and M Kouzaki. "Visuomotor Transformation for the Lead Leg Affects Trail Leg Trajectories During Visually Guided Crossing Over a Virtual Obstacle in Humans." In: *Frontiers in Neuroscience* 14 (2020). doi: 10.3389/fnins.2020.00357. url: https://www.scopus.com/inward/record.uri?eid=2-s2.0-85084357262{\&}doi=10.3389{\%}2Ffnins.2020.00357{\&}partnerID=40{\&}md5=75fcdebe295114e2729de689fe31a1c3.

[168] R Sun, R Kaur, L Ziegelman, S Yang, R Sowers, and M. E. Hernandez. "Using Virtual Reality to Examine the Correlation between Balance Function and Anxiety in Stance." In: *2019 IEEE International Conference on Bioinformatics and Biomedicine (BIBM)*. 2019, pp. 1633–1640. isbn: VO -. doi: 10.1109/BIBM47256.2019.8983331.

[169] T. W. Cleworth, R. Chua, J. T. Inglis, and M. G. Carpenter. "Influence of virtual height exposure on postural reactions to support surface translations." In: *GAIT & POSTURE* 47 (June 2016), pp. 96–102. issn: 0966-6362. doi: 10.1016/j.gaitpost.2016.04.006.

[170] M Habibnezhad, J Puckett, M. S. Fardhosseini, and L. A. Pratama. "A mixed VR and physical framework to evaluate impacts of virtual legs and elevated narrow working space on construction workers' gait pattern." In: *Proceedings of the 36th International Symposium on Automation and Robotics in Construction, ISARC 2019*. 2019, pp. 1057–1064. doi: 10.22260/isarc2019/0141. url: https://www.scopus.com/inward/record.uri?eid=2-s2.0-85071481979{\&}doi=10.22260{\%}2Fisarc2019{\%}2F0141{\&}partnerID=40{\&}md5=19ed2a56cbe641607bcd4f9b779102b5.

[171] M Habibnezhad, J Puckett, H Jebelli, A Karji, M. S. Fardhosseini, and S Asadi. "Neurophysiological testing for assessing construction workers' task performance at virtual height." In: *Automation in Construction* 113 (2020). doi: 10.1016/j.autcon.2020.103143. url: https://www.scopus.com/inward/record.uri?eid=2-s2.0-85080081047{\&}doi=10.1016{\%}2Fj.autcon.2020.103143{\&}partnerID=40{\&}md5=4cf7e46fc474dbb9607ab27a2fca12d0.

[172] H. Ida, S. Mohapatra, and A. Aruin. "Control of vertical posture while elevating one foot to avoid a real or virtual obstacle." In: *EXPERIMENTAL BRAIN RESEARCH* 235.6 (June 2017), pp. 1677–1687. issn: 0014-4819. doi: 10.1007/s00221-017-4929-0.

[173] D. Martelli, B. Xia, A. Prado, and S. K. Agrawal. “Gait adaptations during overground walking and multidirectional oscillations of the visual field in a virtual reality headset.” In: *GAIT & POSTURE* 67 (Jan. 2019), pp. 251–256. issn: 0966-6362. doi: 10.1016/j.gaitpost.2018.10.029.

[174] J. D. Guzmán, D. E. Guzmán, C. F. Rengifo, J. D. Guzman, D. E. Guzman, and C. F. Rengifo. “Affordable Human Gait Analysis Using a Virtual Reality-Based Platform.” In: *IEEE Revista Iberoamericana de Tecnologias del Aprendizaje* 15.4 (2020), pp. 307–313. issn: 1932-8540 VO - 15. doi: 10.1109/RITA.2020.3033219. url: https://www.scopus.com/inward/record.uri?eid=2-s2.0-85096636411{\&}doi=10.1109{\%}2FRITA.2020.3033219{\&}partnerID=40{\&}md5=4606c0c30b0d22d5d9b6e31c41d80132.

[175] L. I. Riem, B. D. Schmit, and S. A. Beardsley. “The Effect of Discrete Visual Perturbations on Balance Control during Gait.” In: *2020 42nd Annual International Conference of the IEEE Engineering in Medicine & Biology Society (EMBC)*. 2020, pp. 3162–3165. isbn: 2694-0604 VO -. doi: 10.1109/EMBC44109.2020.9176303.

[176] L Riem, J. V. Dehy, T Onushko, and S Beardsley. “Inducing Compensatory Changes in Gait Similar to External Perturbations Using an Immersive Head Mounted Display.” In: *2018 IEEE Conference on Virtual Reality and 3D User Interfaces (VR)*. 2018, pp. 128–135. isbn: VO -. doi: 10.1109/VR.2018.8446432.

[177] H. Chander, S. N. K. K. Arachchige, C. M. Hill, A. J. Turner, S. Deb, A. Shojaei, C. Hudson, A. C. Knight, and D. W. Carruth. “Virtual-Reality-Induced Visual Perturbations Impact Postural Control System Behavior.” In: *BEHAVIORAL SCIENCES* 9.11 (Nov. 2019). doi: 10.3390/bs9110113.

[178] A Mohebbi, P Amiri, and R. E. Kearney. “Contributions of Vision in Human Postural Control: A Virtual Reality-based Study.” In: *2020 42nd Annual International Conference of the IEEE Engineering in Medicine & Biology Society (EMBC)*. 2020, pp. 3347–3350. isbn: 2694-0604 VO -. doi: 10.1109/EMBC44109.2020.9175605.

[179] M Dennison and M D’Zmura. “Effects of unexpected visual motion on postural sway and motion sickness.” In: *Applied Ergonomics* 71 (2018), pp. 9–16. doi: 10.1016/j.apergo.2018.03.015. url: https://www.scopus.com/inward/record.uri?eid=2-s2.0-85055078656{\&}doi=10.1016{\%}2Fj.apergo.2018.03.015{\&}partnerID=40{\&}md5=28946cb6c59f593bb38a0c9de9ca932c.

[180] D. M. Stramel, R. M. Carrera, S. A. Rahok, J Stein, and S. K. Agrawal. “Effects of a Person-Following Light-Touch Device During Overground Walking With Visual Perturbations in a Virtual Reality Environment.” In: *IEEE Robotics and Automation Letters* 4.4 (2019), pp. 4139–4146. issn: 2377-3766 VO - 4. doi: 10.1109/LRA.2019.2931267.

[181] M Santoso and D Phillips. “Optical Flow, Perturbation Velocities and Postural Response In Virtual Reality.” In: *2020 IEEE Conference on Virtual Reality and 3D User Interfaces Abstracts and Workshops (VRW)*. 2020, pp. 788–789. isbn: VO -. doi: 10.1109/VRW50115.2020.00245.

[182] A. V. Lubetzky, D. Harel, J. Kelly, B. D. Hujsak, and K. Perlin. “Weighting and reweighting of visual input via head mounted display given unilateral peripheral vestibular dysfunction.” In: *HUMAN MOVEMENT SCIENCE* 68 (Dec. 2019). issn: 0167-9457. doi: 10.1016/j.humov.2019.102526.

[183] S. M. Peterson and D. P. Ferris. “Differentiation in theta and beta electrocortical activity between visual and physical perturbations to walking and standing balance.” In: *eNeuro* 5.4 (2018). doi: 10.1523/ENEURO.0207-18.2018. url: https://www.scopus.com/inward/record.uri?eid=2-s2.0-85052150553{\&}doi=10.1523{\%}2FENEURO.0207-18.2018{\&}partnerID=40{\&}md5=24436190f5cc06730e58b5f442891f70.

[184] A. V. Lubetzky, E. E. Kary, D. Harel, B. Hujsak, and K. Perlin. “Feasibility and reliability of a virtual reality oculus platform to measure sensory integration for postural control in young adults.” In: *Physiotherapy Theory and Practice* 34.12 (Jan. 2018), pp. 935–950. issn: 15325040. doi: 10.1080/09593985.2018.1431344. url: https://pubmed.ncbi.nlm.nih.gov/29364733/.

[185] S. M. Peterson, E. Rios, and D. P. Ferris. “Transient visual perturbations boost short-term balance learning in virtual reality by modulating electrocortical activity.” In: *JOURNAL OF NEUROPHYSIOLOGY* 120.4 (Oct. 2018), pp. 1998–2010. issn: 0022-3077. doi: 10.1152/jn.00292.2018.

[186] J. Liu, T. E. Lockhart, P. Parijat, J. D. McIntosh, and Y.-P. Chiu. “Comparison of Slip Training in VR Environment And on Moveable Platform.” eng. In: *Biomedical sciences instrumentation* 51 (2015), pp. 189–197. issn: 0067-8856 (Print).

[187] E. Chiarovano, C. de Waele, H. G. MacDougall, S. J. Rogers, A. M. Burgess, and I. S. Curthoys. “Maintaining balance when looking at a virtual reality three-dimensional display of a field of moving dots or at a virtual reality scene.” In: *Frontiers in Neurology* 6.7 (2015), p. 164. issn: 16642295. doi: 10.3389/FNEUR.2015.00164/BIBTEX.

[188] E. Chiarovano, W. Wang, P. Reynolds, and H. G. MacDougall. “Imbalance: Objective measures versus subjective self-report in clinical practice.” In: *GAIT & POSTURE* 59 (Jan. 2018), pp. 217–221. issn: 0966-6362. doi: 10.1016/j.gaitpost.2017.10.019.

[189] T Tossavainen, E Toppila, I Pyykko, P. M. Forsman, M Juhola, and J Starck. “Virtual reality in posturography.” In: *IEEE Transactions on Information Technology in Biomedicine* 10.2 (2006), pp. 282–292. issn: 1558-0032 VO - 10. doi: 10.1109/TITB.2005.859874.

[190] A. Mohebbi, P. Amiri, and R. E. Kearney. “Identification of human balance control responses to visual inputs using virtual reality.” In: *Journal of Neurophysiology* 127.4 (2022), pp. 1159–1170. issn: 15221598. doi: 10.1152/jn.00283.2021.

[191] A. V. Lubetzky and B. D. Hujsak. "A virtual reality head stability test for patients with vestibular dysfunction." eng. In: *Journal of vestibular research : equilibrium & orientation* 28.5-6 (2018), pp. 393–400. issn: 1878-6464 (Electronic). doi: 10.3233/VES-190650.

[192] J. Ketterer, S. Ringhof, D. Gehring, and A. Gollhofer. "Sinusoidal Optic Flow Perturbations Reduce Transient but Not Continuous Postural Stability: A Virtual Reality-Based Study." In: *Frontiers in Physiology* 13.May (2022), pp. 1–9. doi: 10.3389/fphys.2022.803185.

[193] Z. Bassiri, C. Austin, C. Cousin, and D. Martelli. "Subsensory electrical noise stimulation applied to the lower trunk improves postural control during visual perturbations." In: *Gait and Posture* 96.May (2022), pp. 22–28. issn: 18792219. doi: 10.1016/j.gaitpost.2022.05.010. url: https://doi.org/10.1016/j.gaitpost.2022.05.010.

[194] J. D. Guzmán, D. E. Guzmán, and C. F. Rengifo. "Affordable Human Gait Analysis Using a Virtual Reality-Based Platform." In: *IEEE Revista Iberoamericana de Tecnologias del Aprendizaje* 15.4 (2020), pp. 307–313. issn: 1932-8540 VO - 15. doi: 10.1109/RITA.2020.3033219.

[195] H. Chander, A. Shojaei, S. Deb, S. Kodithuwakku Arachchige, C. Hudson, A. Knight, and D. Carruth. "Impact of Virtual Reality–Generated Construction Environments at Different Heights on Postural Stability and Fall Risk." In: *Workplace Health and Safety* (2020). doi: 10.1177/2165079920934000.

[196] A. Mehrfard, J. Fotouhi, G. Taylor, T. Forster, N. Navab, and B. Fuerst. "A Comparative Analysis of Virtual Reality Head-Mounted Display Systems." In: *arXiv* (Dec. 2019). arXiv: 1912.02913. url: http://arxiv.org/abs/1912.02913.

[197] D. H. Gates, B. J. Darter, J. B. Dingwell, and J. M. Wilken. "Comparison of walking overground and in a Computer Assisted Rehabilitation Environment (CAREN) in individuals with and without transtibial amputation." In: *Journal of NeuroEngineering and Rehabilitation* 9.1 (2012). issn: 17430003. doi: 10.1186/1743-0003-9-81.

[198] P. O. Riley, G. Paolini, U. Della Croce, K. W. Paylo, and D. C. Kerrigan. "A kinematic and kinetic comparison of overground and treadmill walking in healthy subjects." In: *Gait & Posture* 26.1 (June 2007), pp. 17–24. issn: 09666362. doi: 10.1016/j.gaitpost.2006.07.003. url: https://linkinghub.elsevier.com/retrieve/pii/S0966636206001457.

[199] S. J. Lee and J. Hidler. "Biomechanics of overground vs. treadmill walking in healthy individuals." In: *Journal of Applied Physiology* 104.3 (Mar. 2008), pp. 747–755. issn: 8750-7587. doi: 10.1152/japplphysiol.01380.2006. url: https://www.physiology.org/doi/10.1152/japplphysiol.01380.2006.

[200] F. Alton, L. Baldey, S. Caplan, and M. Morrissey. "A kinematic comparison of overground and treadmill walking." In: *Clinical Biomechanics* 13.6 (Sept. 1998), pp. 434–440. issn: 02680033. doi: 10.1016/S0268-0033(98)00012-6. url: https://linkinghub.elsevier.com/retrieve/pii/S0268003398000126.

[201] Y. Sun, H. M. Cullen, and E. P. Zehr. "Harnessing the Power of a Novel Program for Dynamic Balance Perturbation with Supported Body Weight." In: *Journal of motor behavior* 52.6 (Nov. 2020), pp. 643–655. issn: 1940-1027. doi: 10.1080/00222895.2019.1670129. url: https://pubmed.ncbi.nlm.nih.gov/31608808/.

[202] C. L. Brockett and G. J. Chapman. "Biomechanics of the ankle." In: *Orthopaedics and trauma* 30.3 (June 2016), pp. 232–238. issn: 1877-1327. doi: 10.1016/J.MPORTH.2016.04.015. url: https://pubmed.ncbi.nlm.nih.gov/27594929/.

[203] M. J. Santos, N. Kanekar, and A. S. Aruin. "The role of anticipatory postural adjustments in compensatory control of posture: 1. Electromyographic analysis." In: *Journal of Electromyography and Kinesiology* 20.3 (June 2010), pp. 388–397. issn: 10506411. doi: 10.1016/j.jelekin.2009.06.006.

[204] Y. J. Lee and A. S. Aruin. "Three components of postural control associated with pushing in symmetrical and asymmetrical stance." In: *Experimental Brain Research* 228.3 (July 2013), pp. 341–351. issn: 00144819. doi: 10.1007/S00221-013-3567-4/FIGURES/5. url: https://link.springer.com/article/10.1007/s00221-013-3567-4.

[205] K. Horiuchi, M. Ishihara, and K. Imanaka. "The essential role of optical flow in the peripheral visual field for stable quiet standing: Evidence from the use of a head-mounted display." In: *PLoS ONE* 12.10 (Oct. 2017). issn: 19326203. doi: 10.1371/JOURNAL.PONE.0184552. url: /pmc/articles/PMC5633140//pmc/articles/PMC5633140/?report=abstracthttps://www.ncbi.nlm.nih.gov/pmc/articles/PMC5633140/.

[206] A. Berencsi, M. Ishihara, and K. Imanaka. "The functional role of central and peripheral vision in the control of posture." In: *Human Movement Science* 24.5-6 (Oct. 2005), pp. 689–709. issn: 01679457. doi: 10.1016/J.HUMOV.2005.10.014.

[207] T. Prokop, M. Schubert, and W. Berger. "Visual influence on human locomotion. Modulation to changes in optic flow." In: *Experimental Brain Research* 114.1 (1997), pp. 63–70. issn: 00144819. doi: 10.1007/PL00005624. url: https://pubmed.ncbi.nlm.nih.gov/9125452/.

[208] M. Raffi and A. Piras. "Investigating the crucial role of optic flow in postural control: Central vs. peripheral visual field." In: *Applied Sciences (Switzerland)* 9.5 (2019). issn: 20763417. doi: 10.3390/APP9050934.

[209] K. Terry, E. H. Sinitski, J. B. Dingwell, and J. M. Wilken. "Amplitude effects of medio-lateral mechanical and visual perturbations on gait." In: *Journal of Biomechanics* 45.11 (July 2012), pp. 1979–1986. issn: 00219290. doi: 10.1016/j.jbiomech.2012.05.006.

[210] B. C. Horslen, C. J. Dakin, J. T. Inglis, J.-S. Blouin, and M. G. Carpenter. "Modulation of human vestibular reflexes with increased postural threat." In: *The Journal of Physiology* 592.16 (Aug. 2014), pp. 3671–3685. issn: 00223751. doi: 10.1113/jphysiol.2014.270744. url: http://doi.wiley.com/10.1113/jphysiol.2014.270744.

[211] *Sensory Organization Test | RehabMeasures Database*. url: https://www.sralab.org/rehabilitation-measures/sensory-organization-test (visited on 03/08/2021).

[212] Mark G. Carpenter, James S. Frank, and Cathy P. Silcher. "Surface height effects on postural control: A hypothesis for a stiffness strategy for stance." In: *Journal of Vestibular Research* 9.4 (Jan. 1999), pp. 277–286. issn: 0957-4271.

[213] N. Reicke. *DER KOPFSCHUTTEL-ROMBERG-TEST IM RAHMEN DER GLEICHGEWICHTSDIAGNOSTIK*. June 1992. url: https://europepmc.org/article/med/1634377.

[214] M. Nieschalk, K. W. Delank, and W. Stoll. "A quantitative interpretation of the Romberg test." In: *Laryngo- Rhino- Otologie* 74.8 (1995), pp. 489–494. issn: 09358943. doi: 10.1055/s-2007-997786.

[215] E. J. Formeister, R. Krauter, L. Kirk, T. R. Zhu, H. G. Rizk, and J. D. Sharon. "Understanding the Dizziness Handicap Inventory (DHI): A Cross Sectional Analysis of Symptom Factors That Contribute to DHI Variance." In: *Otology and Neurotology* 41.1 (Jan. 2020), pp. 86–93. issn: 15374505. doi: 10.1097/MAO.0000000000002438. url: https://pubmed.ncbi.nlm.nih.gov/31644479/.

[216] *SMART EquiTest CDP Dynamic System Count on objective, evidence-based information from the NeuroCom Family of Balance Manager Solutions NeuroCom ® SMART EquiTest ® CDP Objective Balance Assessment Dynamic Training Protocols*. Tech. rep. url: www.natus.com.

[217] R. Soangra and T. E. Lockhart. "Determination of stabilogram diffusion analysis coefficients and invariant density analysis parameters to understand postural stability associated with standing on anti-fatigue mats." In: *Biomedical Sciences Instrumentation* 48 (2012), p. 415. issn: 00678856. url: /pmc/articles/PMC3716259//pmc/articles/PMC3716259/?report=abstracthttps://www.ncbi.nlm.nih.gov/pmc/articles/PMC3716259/.

[218] O. Beauchet, G. Allali, C. Annweiler, S. Bridenbaugh, F. Assal, R. W. Kressig, and F. R. Herrmann. "Gait variability among healthy adults: Low and high stride-to-stride variability are both a reflection of gait stability." In: *Gerontology* 55.6 (2009), pp. 702–706. issn: 0304324X. doi: 10.1159/000235905.

[219] M. J. Toebes, M. J. Hoozemans, R. Furrer, J. Dekker, and J. H. Van Dieën. "Local dynamic stability and variability of gait are associated with fall history in elderly subjects." In: *Gait & Posture* 36.3 (2012), pp. 527–531. issn: 0966-6362. doi: 10.1016/J.GAITPOST.2012.05.016.

[220] C. L. Chen, S. Z. Lou, H. W. Wu, S. K. Wu, K. T. Yeung, and F. C. Su. "Effects of the type and direction of support surface perturbation on postural responses." In: *Journal of NeuroEngineering and Rehabilitation* 11.1 (Apr. 2014), pp. 1–12. issn: 17430003. doi: 10.1186/1743-0003-11-50/FIGURES/6. url: https://jneuroengrehab.biomedcentral.com/articles/10.1186/1743-0003-11-50.

[221] K. Takakusaki. "Functional Neuroanatomy for Posture and Gait Control." In: *Journal of Movement Disorders* 10.1 (Jan. 2017), pp. 1–17. issn: 2005-940X. doi: 10.14802/JMD.16062. url: http://www.e-jmd.org/journal/view.php?doi=10.14802/jmd.16062.

[222] M. Maier, B. Rubio Ballester, A. Duff, E. Duarte Oller, and P. F. Verschure. "Effect of Specific Over Nonspecific VR-Based Rehabilitation on Poststroke Motor Recovery: A Systematic Meta-analysis." In: *Neurorehabilitation and neural repair* 33.2 (Feb. 2019), pp. 112–129. issn: 1552-6844. doi: 10.1177/1545968318820169. url: https://pubmed.ncbi.nlm.nih.gov/30700224/.

[223] O. D. Kothgassner and A. Felnhofer. "Does virtual reality help to cut the Gordian knot between ecological validity and experimental control?" In: *https://doi.org/10.1080/23808985.2020.1792790* (2020), pp. 210–218. issn: 23808977. doi: 10.1080/23808985.2020.1792790. url: https://www.tandfonline.com/doi/abs/10.1080/23808985.2020.1792790.

[224] J. Zaki and K. Ochsner. "The Need for a Cognitive Neuroscience of Naturalistic Social Cognition." In: *Annals of the New York Academy of Sciences* 1167.1 (June 2009), pp. 16–30. issn: 1749-6632. doi: 10.1111/J.1749-6632.2009.04601.X. url: https://onlinelibrary.wiley.com/doi/full/10.1111/j.1749-6632.2009.04601.xhttps://onlinelibrary.wiley.com/doi/abs/10.1111/j.1749-6632.2009.04601.xhttps://nyaspubs.onlinelibrary.wiley.com/doi/10.1111/j.1749-6632.2009.04601.x.

[225] A. Gorini, C. S. Capideville, G. De Leo, F. Mantovani, and G. Riva. "The Role of Immersion and Narrative in Mediated Presence: The Virtual Hospital Experience." In: *https://home.liebertpub.com/cyber* 14.3 (Mar. 2011), pp. 99–105. issn: 21522723. doi: 10.1089/CYBER.2010.0100. url: https://www.liebertpub.com/doi/abs/10.1089/cyber.2010.0100.

[226] M. D. Franzen and K. L. Wilhelm. "Conceptual foundations of ecological validity in neuropsychological assessment." In: *Ecological validity of neuropsychological testing* (1996), pp. 91–112.

[227] *How Unity3D Became a Game-Development Beast*. url: https://insights.dice.com/2013/06/03/how-unity3d-become-a-game-development-beast/ (visited on 06/29/2022).

[228] *Unity vs Unreal Engine: Which Game Engine Should You Choose?* url: https://hackr.io/blog/unity-vs-unreal-engine (visited on 06/29/2022).

[229] M. Newman, B. Gatersleben, K. J. Wyles, and E. Ratcliffe. "The use of virtual reality in environment experiences and the importance of realism." In: *Journal of Environmental Psychology* 79 (Feb. 2022), p. 101733. issn: 0272-4944. doi: 10.1016/J.JENVP.2021.101733.

[230] J. R. J. Neo, A. S. Won, and M. M. Shepley. "Designing Immersive Virtual Environments for Human Behavior Research." In: *Frontiers in Virtual Reality* 0 (Mar. 2021), p. 5. issn: 2673-4192. doi: 10.3389/FRVIR.2021.603750.

[231] X. Roothaer. "Multi-scale study of the mechanical behaviour of bearing and bone-bearing bones : towards personalization of FE human models." Doctoral dissertation. 2019.

[232] F. Alonso-Martín, M. Malfaz, J. Sequeira, J. F. Gorostiza, and M. A. Salichs. "A Multimodal Emotion Detection System during Human–Robot Interaction." In: *Sensors* 13.11 (2013), pp. 15549–15581. issn: 1424-8220. doi: 10.3390/s131115549. url: https://www.mdpi.com/1424-8220/13/11/15549.

[233] P. Goyal and M. S. Maurer. "Syncope in older adults." In: *Journal of Geriatric Cardiology : JGC* 13.5 (2016), p. 380. issn: 16715411. doi: 10.11909/J.ISSN.1671-5411.2016.05.002. url: /pmc/articles/PMC4984568/https://www.ncbi.nlm.nih.gov/pmc/articles/PMC4984568/.

[234] A. V. Lubetzky and B. D. Hujsak. "A virtual reality head stability test for patients with vestibular dysfunction." In: *Journal of Vestibular Research: Equilibrium and Orientation* 28.5-6 (2019), pp. 393–400. doi: 10.3233/VES-190650. url: https://www.scopus.com/inward/record.uri?eid=2-s2.0-85064661796{\&}doi=10.3233{\%}2FVES-190650{\&}partnerID=40{\&}md5=ee899149c090765b887662caf4b69237.

[235] *Unity - Manual: Unity User Manual 2021.3 (LTS)*. url: https://docs.unity3d.com/Manual/UnityManual.html (visited on 05/25/2022).

[236] B. J. Mohler, J. L. Campos, M. B. Weyel, and H. H. Bülthoff. "Gait Parameters while Walking in a Head-mounted Display Virtual Environment and the Real World." In: *Eurographics Symposium on Virtual Environments, Short Papers and Posters*. Ed. by B. Froehlich, R. Blach, and R. van Liere. The Eurographics Association, 2007. isbn: 978-3-905673-64-7. doi: 10.2312/PE/VE2007Short/085-088.

[237] A. Canessa, P. Casu, F. Solari, and M. Chessa. "Comparing real walking in immersive virtual reality and in physical world using gait analysis." In: *VISIGRAPP 2019 - Proceedings of the 14th International Joint Conference on Computer Vision, Imaging and Computer Graphics Theory and Applications* 2 (2019), pp. 121–128. doi: 10.5220/0007380901210128.

[238] A. A. Soczawa-Stronczyk and M. Bocian. "Gait coordination in overground walking with a virtual reality avatar." In: *Royal Society Open Science* 7.7 (July 2020). issn: 20545703. doi: 10.1098/RSOS.200622. url: https://royalsocietypublishing.org/doi/full/10.1098/rsos.200622.

[239] J. Wang, Y. Chen, S. Hao, X. Peng, and L. Hu. "Deep learning for sensor-based activity recognition: A survey." In: *Pattern Recognition Letters* 119 (2019). Deep Learning for Pattern Recognition, pp. 3–11. issn: 0167-8655. doi: https://doi.org/10.1016/j.patrec.2018.02.010. url: https://www.sciencedirect.com/science/article/pii/S016786551830045X.

[240] A. T. Dilanchian, R. Andringa, and W. R. Boot. "A Pilot Study Exploring Age Differences in Presence, Workload, and Cybersickness in the Experience of Immersive Virtual Reality Environments." In: *Frontiers in Virtual Reality* 0 (Oct. 2021), p. 129. issn: 2673-4192. doi: 10.3389/FRVIR.2021.736793.

[241] L. Appel, E. Appel, O. Bogler, M. Wiseman, L. Cohen, N. Ein, H. B. Abrams, and J. L. Campos. "Older Adults With Cognitive and/or Physical Impairments Can Benefit From Immersive Virtual Reality Experiences: A Feasibility Study." In: *Frontiers in medicine* 6 (Jan. 2020). issn: 2296-858X. doi: 10.3389/FMED.2019.00329. url: https://pubmed.ncbi.nlm.nih.gov/32010701/.

[242] H. Huygelier, B. Schraepen, R. van Ee, V. Vanden Abeele, and C. R. Gillebert. "Acceptance of immersive head-mounted virtual reality in older adults." In: *Scientific reports* 9.1 (Dec. 2019). issn: 2045-2322. doi: 10.1038/S41598-019-41200-6. url: https://pubmed.ncbi.nlm.nih.gov/30872760/.

[243] Y. Okubo, D. L. Sturnieks, M. A. Brodie, L. Duran, and S. R. Lord. "Effect of Reactive Balance Training Involving Repeated Slips and Trips on Balance Recovery Among Older Adults: A Blinded Randomized Controlled Trial." In: *JOURNALS OF GERONTOLOGY SERIES A-BIOLOGICAL SCIENCES AND MEDICAL SCIENCES* 74.9 (2019), pp. 1489–1496. issn: 1079-5006. doi: 10.1093/gerona/glz021.

[244] *VIVE - VR Headsets, Games, and Metaverse Life | United States*. url: https://www.vive.com/us/ (visited on 06/09/2022).

[245] P. Caserman, A. Garcia-Agundez, and S. Gobel. "A Survey of Full-Body Motion Reconstruction in Immersive Virtual Reality Applications." In: *IEEE Transactions on Visualization and Computer Graphics* 26.10 (Oct. 2020), pp. 3089–3108. issn: 19410506. doi: 10.1109/TVCG.2019.2912607.

[246] M. Paulich, M. Schepers, N. Rudigkeit, and G. Bellusci. "Xsens MTw Awinda: Miniature Wireless Inertial-Magnetic Motion Tracker for Highly Accurate 3D Kinematic Applications." In: (). url: www.xsens.com,.

[247] M. Christofi, D. Michael-Grigoriou, and C. Kyrlitsias. "A Virtual Reality Simulation of Drug Users' Everyday Life: The Effect of Supported Sensorimotor Contingencies on Empathy." In: *Frontiers in Psychology* 11 (June 2020). doi: 10.3389/fpsyg.2020.01242.

[248] *Xsens - BMS LAB — Where Technology Meets Life*. url: https://bmslab.utwente.nl/info-hub/xsens/ (visited on 05/31/2022).

[249] H. Milner-Brown and R. Lee. "Synchronization of human motor units: Possible roles of exercise and supraspinal reflexes." In: *Electroencephalography and Clinical Neurophysiology* 38.3 (1975), pp. 245–254. issn: 0013-4694. doi: https://doi.org/10.1016/0013-4694(75)90245-X. url: https://www.sciencedirect.com/science/article/pii/001346947590245X.

[250] J Basmajian and C. J. De Luca. "Description and Analysis of the EMG Signal." In: *Muscles alive : their functions revealed by electromyography* (1985), pp. 65–100. issn: 0035-9157.

[251] D. A. Winter. "Biomechanics and Motor Control of Human Movement, Fourth Edition David A. Winter (cloth) 1. Human mechanics. 2. Motor ability. 3. Kinesiology. I. Title. QP303.W59." In: 7 (2009), pp. 6–22. url: www.wiley.com..

[252] D. Farina and F. Negro. "Accessing the Neural Drive to Muscle and Translation to Neurorehabilitation Technologies." In: *IEEE Reviews in Biomedical Engineering* 5 (2012), pp. 3–14. doi: 10.1109/RBME.2012.2183586.

[253] C. C. Young, S. E. Rose, E. N. Biden, M. P. Wyatt, and D. H. Sutherland. "The effect of surface and internal electrodes on the gait of children with cerebral palsy, spastic diplegic type." In: *Journal of orthopaedic research : official publication of the Orthopaedic Research Society* 7.5 (1989), pp. 732–737. issn: 0736-0266. doi: 10.1002/JOR.1100070515. url: https://pubmed.ncbi.nlm.nih.gov/2760746/.

[254] *Trigno® Research+ System - Delsys*. url: https://delsys.com/trigno/ (visited on 05/31/2022).

[255] D. Cano Porras, J. V. Jacobs, R. Inzelberg, Y. Bahat, G. Zeilig, and M. Plotnik. "Patterns of whole-body muscle activations following vertical perturbations during standing and walking." In: *Journal of NeuroEngineering and Rehabilitation* 18.1 (Dec. 2021), pp. 1–18. issn: 17430003. doi: 10.1186/S12984-021-00836-0/FIGURES/8. url: https://jneuroengrehab.biomedcentral.com/articles/10.1186/s12984-021-00836-0.

[256] S. Y. Lee and M. E. Jo. "Comparison of maximum voluntary isometric contraction of the biceps on various posture and respiration conditions for normalization of electromyography data." In: *Journal of Physical Therapy Science* 28.11 (Nov. 2016), p. 3007. issn: 09155287. doi: 10.1589/JPTS.28.3007. url: /pmc/articles/PMC5140790//pmc/articles/PMC5140790/?report=abstracthttps://www.ncbi.nlm.nih.gov/pmc/articles/PMC5140790/.

[257] T. Chau. “A review of analytical techniques for gait data. Part 1: fuzzy, statistical and fractal methods.” In: *Gait & Posture* 13.1 (Feb. 2001), pp. 49–66. issn: 0966-6362. doi: 10.1016/S0966-6362(00)00094-1.

[258] D. A. Winter. “Human balance and posture control during standing and walking.” In: *Gait & Posture* 3.4 (Dec. 1995), pp. 193–214. issn: 0966-6362. doi: 10.1016/0966-6362(96)82849-9.

[259] M. Vlutters, E. H. Van Asseldonk, and H. Van Der Kooij. “Center of mass velocity-based predictions in balance recovery following pelvis perturbations during human walking.” In: *Journal of Experimental Biology* 219.10 (May 2016), pp. 1514–1523. issn: 00220949. doi: 10.1242/JEB.129338/262127/AM/CENTER-OF-MASS-VELOCITY-BASED-PREDICTIONS-IN. url: https://journals.biologists.com/jeb/article/219/10/1514/14593/Center-of-mass-velocity-based-predictions-in.

[260] H. Finch. “Comparison of the performance of nonparametric and parametric MANOVA test statistics when assumptions are violated.” In: *Methodology* 1.1 (2005), pp. 27–38. issn: 16141881. doi: 10.1027/1614-1881.1.1.27.

[261] D. G. Altman and j. M. Bland. “Statistics notes: the normal distribution.” In: *BMJ (Clinical research ed.)* 310.6975 (Feb. 1995), p. 298. issn: 0959-8138. doi: 10.1136/BMJ.310.6975.298. url: https://pubmed.ncbi.nlm.nih.gov/7866172/.

[262] J Pallant. “SPSS survival manual: A step by step guide to data analysis using IBM SPSS.” In: (2020). url: https://www.taylorfrancis.com/books/mono/10.4324/9781003117452/spss-survival-manual-julie-pallant.

[263] A. Elliott and W. Woodward. “Statistical analysis quick reference guidebook: With SPSS examples.” In: (2007). url: https://books.google.com/books?hl=pt-PT{\&}lr={\&}id=SOsX0IbNxeIC{\&}oi=fnd{\&}pg=PR13{\&}ots=rp102ohz9N{\&}sig=E6LQNci{_}sUZlKJeq6RfAnSIH99M.

[264] A. Field. “Discovering statistics using IBM SPSS statistics.” In: (2013). url: https://books.google.com/books?hl=pt-PT{\&}lr={\&}id=c0Wk9IuBmAoC{\&}oi=fnd{\&}pg=PP2{\&}ots=LcDnKG-t0B{\&}sig=ucLnM5tLBQZXOVswukH3GS4y2A0.

[265] A. Ghasemi and S. Zahediasl. “Normality Tests for Statistical Analysis: A Guide for Non-Statisticians.” In: *International Journal of Endocrinology and Metabolism* 10.2 (Apr. 2012), p. 486. issn: 1726913X. doi: 10.5812/IJEM.3505. url: /pmc/articles/PMC3693611//pmc/articles/PMC3693611/?report=abstracthttps://www.ncbi.nlm.nih.gov/pmc/articles/PMC3693611/.

[266] S. S. Shapiro and M. B. Wilk. “An Analysis of Variance Test for Normality (Complete Samples).” In: *Biometrika* 52.3/4 (1965), pp. 591–611. issn: 00063444. url: http://www.jstor.org/stable/2333709 (visited on 06/15/2022).

[267] P. Mishra, C. M. Pandey, U. Singh, A. Gupta, C. Sahu, and A. Keshri. “Descriptive Statistics and Normality Tests for Statistical Data.” In: *Annals of Cardiac Anaesthesia* 22.1 (Jan. 2019), p. 67. issn: 09745181. doi: 10.4103/ACA.ACA_157_18. url: /pmc/articles/PMC6350423//pmc/articles/PMC6350423/?report=abstracthttps://www.ncbi.nlm.nih.gov/pmc/articles/PMC6350423/.

[268] H.-Y. Kim. “Statistical notes for clinical researchers: assessing normal distribution (2) using skewness and kurtosis.” In: *Restorative Dentistry & Endodontics* 38.1 (2013), p. 52. issn: 2234-7658. doi: 10.5395/RDE.2013.38.1.52. url: /pmc/articles/PMC3591587/https://www.ncbi.nlm.nih.gov/pmc/articles/PMC3591587/.

[269] T. Lumley, P. Diehr, S. Emerson, and L. Chen. “THE IMPORTANCE OF THE NORMALITY ASSUMPTION IN LARGE PUBLIC HEALTH DATA SETS.” In: (2002). doi: 10.1146/annurev.publheath.23.100901.140546. url: www.annualreviews.org.

[270] D. Cousineau and S. Chartier. “Outliers detection and treatment: a review.” In: *International Journal of Psychological Research* 3.1 (June 2010), pp. 58–67. issn: 2011-2084. doi: 10.21500/20112084.844.

[271] “Reprint of: Mahalanobis, P.C. (1936) ”On the Generalised Distance in Statistics.
.” In: *Sankhya A 2019 80:1* 80.1 (Mar. 2019), pp. 1–7. issn: 0976-8378. doi: 10.1007/S13171-019-00164-5. url: https://link.springer.com/article/10.1007/s13171-019-00164-5.

[272] C. Leys, O. Klein, Y. Dominicy, and C. Ley. “Detecting multivariate outliers: Use a robust variant of the Mahalanobis distance.” In: *Journal of Experimental Social Psychology* 74 (Jan. 2018), pp. 150–156. issn: 10960465. doi: 10.1016/J.JESP.2017.09.011.

[273] P. Schober and L. A. Schwarte. “Correlation coefficients: Appropriate use and interpretation.” In: *Anesthesia and Analgesia* 126.5 (May 2018), pp. 1763–1768. issn: 15267598. doi: 10.1213/ANE.0000000000002864. url: https://journals.lww.com/anesthesia-analgesia/Fulltext/2018/05000/Correlation{_}Coefficients{_}{_}Appropriate{_}Use{_}and.50.aspx.

[274] G. E. BOX. “A general distribution theory for a class of likelihood criteria.” In: *Biometrika* 36.3-4 (1949), pp. 317–346. issn: 0006-3444. doi: 10.1093/biomet/36.3-4.317. url: https://pubmed.ncbi.nlm.nih.gov/15402070/.

[275] L. M. Lix, J. C. Keselman, and H. J. Keselman. “Consequences of Assumption Violations Revisited: A Quantitative Review of Alternatives to the One-Way Analysis of Variance ”F”Test.” In: *Review of Educational Research* 66.4 (1996), p. 579. issn: 00346543. doi: 10.2307/1170654. url: /record/1997-02876-005.

[276] K. L. Tang and J. Algina. "Performance of Four Multivariate Tests Under Variance-Covariance Heteroscedasticity." In: *Multivariate behavioral research* 28.4 (Jan. 1993), pp. 391–405. issn: 0027-3171. doi: 10.1207/S15327906MBR2804_1. url: https://pubmed.ncbi.nlm.nih.gov/26801140/.

[277] P. J. Schneider and D. A. Penfield. "Alexander and Govern's Approximation: Providing an Alternative to ANOVA under Variance Heterogeneity." In: *The Journal of Experimental Education* 65.3 (1997), pp. 271–286. issn: 00220973, 19400683. url: http://www.jstor.org/stable/20152526 (visited on 06/15/2022).

[278] M. Blanca, R. Alarcón, J. Arnau, R. Bono, and R. Bendayan. "Non-normal data: Is ANOVA still a valid option?" In: *Psicothema* 29 (Oct. 2017), pp. 552–557. doi: 10.7334/psicothema2016.383.

[279] H. Büning. "Robust analysis of variance." In: 24 (Jan. 1997), pp. 319–332.

[280] S. Lee and C. Ph.D. "Modified ANOVA for unequal variances." In: *COMMUNICATIONS IN STATISTICS Simulation and Computation® Vol. 32* No. 4 (Jan. 2003), pp. 987–1004. doi: 10.1081/SAC-120023874.

[281] P. K. Ito. "7 Robustness of ANOVA and MANOVA test procedures." In: *Handbook of Statistics* 1 (Jan. 1980), pp. 199–236. issn: 0169-7161. doi: 10.1016/S0169-7161(80)01009-7.

[282] C. W. Dunnett. "A Multiple Comparison Procedure for Comparing Several Treatments with a Control." In: *Journal of the American Statistical Association* 50.272 (1955), pp. 1096–1121. doi: 10.1080/01621459.1955.10501294. eprint: https://www.tandfonline.com/doi/pdf/10.1080/01621459.1955.10501294. url: https://www.tandfonline.com/doi/abs/10.1080/01621459.1955.10501294.

[283] J. Ching-Hong Li, M. Nesca, R. Michael Waisman, Y. Cheng, and V. Man Chung Tze. "A robust effect size measure Aw for MANOVA with non-normal and non-homogenous data:" in: *https://doi.org/10.1177/2059* 14.3 (Nov. 2021). issn: 20597991. doi: 10.1177/20597991211055949. url: https://journals.sagepub.com/doi/full/10.1177/20597991211055949.

[284] C. Ateş, Ö. Kaymaz, H. E. Kale, and M. A. Tekindal. "Comparison of Test Statistics of Nonnormal and Unbalanced Samples for Multivariate Analysis of Variance in terms of Type-I Error Rates." In: *Computational and Mathematical Methods in Medicine* 2019 (2019). issn: 17486718. doi: 10.1155/2019/2173638.

[285] C. J. Huberty and J. D. Morris. "Multivariate Analysis Versus Multiple Univariate Analyses." In: *Psychological Bulletin* 105.2 (1989), pp. 302–308. issn: 00332909. doi: 10.1037/0033-2909.105.2.302. url: /record/1989-21163-001.

[286] H. J. Keselman, C. J. Huberty, R. A. Cribbie, L. L. Lowman, L. M. Lix, B. Donahue, M. D. Petoskey, S. Olejnik, R. K. Kowalchuk, J. C. Keselman, and J. R. Levin. "Statistical Practices of Educational Researchers: An Analysis of their ANOVA, MANOVA, and ANCOVA Analyses:" in: *http://dx.doi.org/10.3102/00346543068003350* 68.3 (June 2016), pp. 350–386. issn: 00346543. doi: 10.3102/00346543068003350. url: https://journals.sagepub.com/doi/10.3102/00346543068003350.

[287] K. Pituch and J. Stevens. *Applied Multivariate Statistics for the Social Sciences: Analyses with SAS and IBM's SPSS*. Routledge, 2016, p. 814. isbn: 9780415836661. url: https://www.routledge.com/Applied-Multivariate-Statistics-for-the-Social-Sciences-Analyses-with-SAS/Pituch-Stevens/p/book/9780415836661.

[288] B. G. Tabachnick and L. S. Fidell. *Using multivariate statistics*. Seventh edition. Vol. 87. 4. NY NY: Pearson, 2019, pp. 611–628. isbn: 9780134790541. url: https://lccn.loc.gov/2017040173%0Ahttps://www.vlebooks.com/Vleweb/Product/Index/437320?page=0.

[289] F. L. Huang. "MANOVA: A Procedure Whose Time Has Passed?:" in: *https://doi.org/10.1177/001698621988720* 64.1 (Dec. 2019), pp. 56–60. issn: 19349041. doi: 10.1177/0016986219887200. url: https://journals.sagepub.com/doi/10.1177/0016986219887200.

[290] H. K. Kim and L. S. Chou. "Lower limb muscle activation in response to balance-perturbed tasks during walking in older adults: A systematic review." In: *Gait and Posture* 93.July 2021 (2022), pp. 166–176. issn: 18792219. doi: 10.1016/j.gaitpost.2022.02.014. url: https://doi.org/10.1016/j.gaitpost.2022.02.014.

[291] S. A. Acuña, C. A. Francis, J. R. Franz, and D. G. Thelen. "The effects of cognitive load and optical flow on antagonist leg muscle coactivation during walking for young and older adults." eng. In: *Journal of electromyography and kinesiology : official journal of the International Society of Electrophysiological Kinesiology* 44 (Feb. 2019), pp. 8–14. issn: 1873-5711 (Electronic). doi: 10.1016/j.jelekin.2018.11.003.

[292] C. Z. Hallal, N. R. Marques, D. H. Spinoso, E. R. Vieira, and M. Gonçalves. "Electromyographic patterns of lower limb muscles during apprehensive gait in younger and older female adults." In: *Journal of Electromyography and Kinesiology* 23.5 (Oct. 2013), pp. 1145–1149. issn: 1050-6411. doi: 10.1016/J.JELEKIN.2013.06.006.

[293] Y. C. Pai and J. Patton. "Center of mass velocity-position predictions for balance control." In: *Journal of Biomechanics* 30.4 (Apr. 1997), pp. 347–354. issn: 00219290. doi: 10.1016/S0021-9290(96)00165-0.

[294] 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." In: (2016). doi: 10.1242/jeb.129338.

[295] F. Reynard, D. Christe Id, and P. T. Id. "Postural control in healthy adults: Determinants of trunk sway assessed with a chest-worn accelerometer in 12 quiet standing tasks." In: (2019). doi: 10.1371/journal.pone.0211051. url: https://doi.org/10.1371/journal.pone.0211051.t001.

Appendix

A

Appendix 1 - Questionnaires

A.1 SSQ - Simulator Sickness Questionnaire

None = 0

Slight = 1

Moderate = 2

Severe = 3

Symptoms	Weights for Symptoms Nausea	Oculomotor	Disorientation
General discomfort	1	1	
Fatigue		1	1
Headache		1	
Eye strain		1	
Difficulty focusing	1		1
Increased salivation	1		1
Sweating	1	1	1
Nausea	1		1
Difficulty concentrating		1	1
Fullness of head			
Blurred vision			
Dizzy (eyes open)			
Dizzy (eyes closed)			1
Vertigo			
Stomach awareness	1		
Burping	1		

Score

Nausea = [1] * 9.54

Oculomotor = [2] * 7.58

Disorientation = [3] * 13.92

Total Score = ([1] + [2] + [3]) * 3.74

A.2 IPQ - Igroup Presence Questionnaire

1. In the computer generated world I had a sense of ”being there”. Anchors: not at all-very much.
2. Somehow I felt that the virtual world surrounded me. Anchors: fully disagree-fully agree.
3. I felt like I was just perceiving pictures. Anchors: fully disagree-fully agree.
4. I did not feel present in the virtual space. Anchors: did not feel-felt present.
5. I had a sense of acting in the virtual space, rather than operating something from outside. Anchors: fully disagree-fully agree.
6. I felt present in the virtual space. Anchors: fully disagree-fully agree.
7. How aware were you of the real world surrounding while navigating in the virtual world? (i.e. sounds, room temperature, other people, etc.)? Anchors: extremely aware-moderately aware-not aware at all.
8. I was not aware of my real environment.

9. I still paid attention to the real environment.

10. I was completely captivated by the virtual world. Items 8, 9 and 10 anchors: fully disagree - fully agree.

11. How real did the virtual world seem to you? Anchors: completely real - not real at all.

12. How much did your experience in the virtual environment seem consistent with your real world experience? Anchors: not consistent-moderately consistent-very consistent.

13. How real did the virtual world seem to you? Anchors: about as real as an imagined world–indistinguishable from the real world.

14. The virtual world seemed more realistic than the real world. Anchors: fully disagree-fully agree.

Appendix B

Appendix 2 - Collected Data

Table 26: Pelvis Segment Orientation - Quaternion.

Frame	Pelvis q0	Pelvis q1	Pelvis q2	Pelvis q3
0	0.666697	0.02442	-0.01447	0.744788
1	0.667091	0.024479	-0.01424	0.744438
2	0.667484	0.024537	-0.01401	0.744088
3	0.667847	0.024594	-0.01382	0.743764
4	0.668167	0.024652	-0.01369	0.743476
5	0.668457	0.024708	-0.01361	0.743215

Table 27: Pelvis and L5 Segments Orientation - Euler Angles.

Frame	Pelvis x	Pelvis y	Pelvis z	L5 x	L5 y	L5 z
0	0.631554	-3.19136	96.31583	0.663594	1.132234	99.47792
1	0.657549	-3.1783	96.25477	0.646775	1.14189	99.42615
2	0.68353	-3.16521	96.19371	0.629946	1.151531	99.37438
3	0.705547	-3.15514	96.13746	0.613154	1.158478	99.32579
4	0.721944	-3.15034	96.08764	0.596836	1.160585	99.28118
5	0.73439	-3.14858	96.04261	0.580565	1.160002	99.23975

Table 28: Center of Mass Position, velocity and acceleration.

Frame	CoM pos x	CoM pos y	CoM pos z	CoM vel x	CoM vel y	CoM vel z	CoM acc x	CoM acc y	CoM acc z
0	0.065809	-0.06667	0.928615	-0.00546	-0.00532	0.000132	0.01831	0.019042	-0.00346
1	0.065655	-0.06681	0.928548	-0.0052	-0.00566	0.000513	0.016565	-0.00032	0.006886
2	0.0655	-0.06695	0.928481	-0.00499	-0.00599	0.000862	0.014002	0.019983	-0.00537
3	0.065338	-0.06709	0.928424	-0.0047	-0.00621	0.001087	0.012033	0.008562	-0.00424
4	0.065158	-0.06722	0.928383	-0.00459	-0.00677	0.000975	0.014816	0.020301	-0.03047
5	0.06497	-0.06735	0.928352	-0.00458	-0.00733	0.000696	0.010822	0.02937	-0.03554

Appendix

C

Appendix 3 - Dunnet post hoc results.

Table 29: Tibialis Anterior (both legs) maximum value during the perturbation of the AP forward translation.

Dependent Variable: **RTAMAX**
Dunnett t (2-sided): control group "No Perturbation"

(I) Label	(J) Label	Mean Difference (I-J)	Sig.
AP Axis Trans - Corridor Forward	No Perturbation	-13.134	0.001

Dependent Variable: **LTAMAX**
Dunnett t (2-sided): control group "No Perturbation"

(I) Label	(J) Label	Mean Difference (I-J)	Sig.
AP Axis Trans - Corridor Forward	No Perturbation	-16.1384306932541	0.00001

Table 30: CoM Velocity X-Axis

Dependent Variable: **CoMVelXAVG**
Dunnett t (2-sided): control group "No Perturbation"

(I) Label	(J) Label	Mean Difference (I-J)	Sig.
Roll Indoor 1 CW10	No Perturbation	.318863875510020	0.000
Roll Indoor 1 CW20	No Perturbation	.349096596655353	0.000
Roll Indoor 1 CW30	No Perturbation	.307255828343055	0.000
Roll Indoor 1 CCW10	No Perturbation	.367813571450228	0.000
Roll Indoor 1 CCW20	No Perturbation	.248719137715026	0.000
Roll Indoor 1 CCW30	No Perturbation	.377776350802918	0.000
Roll Indoor 2 CW10	No Perturbation	.351158191895926	0.000
Roll Indoor 2 CW20	No Perturbation	.246921181988535	0.000
Roll Indoor 2 CW30	No Perturbation	.296457025397596	0.000
Roll Indoor 2 CCW10	No Perturbation	.351668887744441	0.000
Roll Indoor 2 CCW20	No Perturbation	.346302387377758	0.000
Roll Indoor 2 CCW30	No Perturbation	.284650550019517	0.000
Roll Outdoor CW10	No Perturbation	.376979801261731	0.000
Roll Outdoor CW20	No Perturbation	.183797479404317	0.001
Roll Outdoor CW30	No Perturbation	.268730323696038	0.000
Roll Outdoor CCW10	No Perturbation	.352539228145275	0.000
Roll Outdoor CCW20	No Perturbation	.354344954867888	0.000
Roll Outdoor CCW30	No Perturbation	.304008073069146	0.000
ML Axis Trans - Kitchen	No Perturbation	-0.040411469423663	0.988
AP Axis Trans - Corridor Forward	No Perturbation	.278694482310105	0.000
AP Axis Trans - Corridor Backward	No Perturbation	.285070376934027	0.000
Pitch Indoor - Bathroom	No Perturbation	.311698696858576	0.000
Pitch Indoor - Fridge	No Perturbation	.319395995507304	0.000
Window Roof Beam Walking - Vertigo	No Perturbation	0.085262642881027	0.849
Window Roof Beam Walking - Vertigo No avatar	No Perturbation	0.110728980032425	0.347
Simple Roof - Vertigo	No Perturbation	0.017450523567697	1.000
Simple Roof - Vertigo No avatar	No Perturbation	0.014153569666839	1.000
Pitch Outdoor - Near Car Oil	No Perturbation	.406010110828113	0.000
Bedroom Syncope	No Perturbation	0.019130099829945	1.000
Garden - Object Avoidance	No Perturbation	0.063720280813013	0.392
Electricity Pole - Vertigo	No Perturbation	0.022847295742689	1.000
Electricity Pole - Vertigo No avatar	No Perturbation	0.024449460181626	1.000
Free Fall	No Perturbation	.281932062758319	0.000
Stairs	No Perturbation	0.043019863066841	1.000
Trip - Sidewalk	No Perturbation	.223563169946864*	0.000

Table 31: CoM Velocity Y-Axis

Dependent Variable: COMVelYAVG
Dunnett t (2-sided): control group "No Perturbation"

(I) Label		Mean Difference (I-J)	Sig.
Roll Indoor 1 CW10	No Perturbation	-0.084853092337285	0.403
Roll Indoor 1 CW20	No Perturbation	-0.062694268357597	0.922
Roll Indoor 1 CW30	No Perturbation	-0.060240450198819	0.950
Roll Indoor 1 CCW10	No Perturbation	-0.078661877421914	0.565
Roll Indoor 1 CCW20	No Perturbation	-0.074328944487546	0.682
Roll Indoor 1 CCW30	No Perturbation	-0.064391362850143	0.898
Roll Indoor 2 CW10	No Perturbation	-0.052176147103201	0.993
Roll Indoor 2 CW20	No Perturbation	-0.003912459106325	1.000
Roll Indoor 2 CW30	No Perturbation	-0.080620338473769	0.512
Roll Indoor 2 CCW10	No Perturbation	-0.083279409340202	0.442
Roll Indoor 2 CCW20	No Perturbation	-0.065898567670261	0.873
Roll Indoor 2 CCW30	No Perturbation	-0.049271548496592	0.997
Roll Outdoor CW10	No Perturbation	0.003983722760867	1.000
Roll Outdoor CW20	No Perturbation	-0.007124812100923	1.000
Roll Outdoor CW30	No Perturbation	-0.025759077723618	1.000
Roll Outdoor CCW10	No Perturbation	-0.098915839262086	0.145
Roll Outdoor CCW20	No Perturbation	-0.023311798096732	1.000
Roll Outdoor CCW30	No Perturbation	-0.027449480320436	1.000
ML Axis Trans - Kitchen	No Perturbation	0.012639840138936	1.000
AP Axis Trans - Corridor Forward	No Perturbation	-.075625096049279	0.010
AP Axis Trans - Corridor Backward	No Perturbation	-.105504817052303	0.000
Pitch Indoor - Bathroom	No Perturbation	-0.005393534653457	1.000
Pitch Indoor - Fridge	No Perturbation	0.015223449881657	1.000
Window Roof Beam Walking - Vertigo	No Perturbation	-0.019523558267776	1.000
Window Roof Beam Walking - Vertigo No avatar	No Perturbation	-0.021895421258025	1.000
Simple Roof - Vertigo	No Perturbation	-0.004653594330529	1.000
Simple Roof - Vertigo No avatar	No Perturbation	-0.004831762220440	1.000
Pitch Outdoor - Near Car Oil	No Perturbation	-.116327311846199	0.029
Bedroom Syncope	No Perturbation	-0.015415082268271	1.000
Garden - Object Avoidance	No Perturbation	-0.020385527811576	1.000
Electricity Pole - Vertigo	No Perturbation	-0.007068212806775	1.000
Electricity Pole - Vertigo No avatar	No Perturbation	-0.004765013645850	1.000
Free Fall	No Perturbation	-.137080069908033	0.000
Stairs	No Perturbation	-0.013523825208105	1.000
Trip - Sidewalk	No Perturbation	-.087567300753617	0.001

Table 32: Left Tibialis Anterior average activation during disturbances

Dependent Variable: LTAAVG
Dunnett t (2-sided): control group "No Perturbation"

(I) Label		Mean Difference (I-J)	Sig.
Roll Indoor 1 CW10	No Perturbation	2.97388521	0.001
Roll Indoor 1 CW20	No Perturbation	2.355169392	0.031
Roll Indoor 1 CW30	No Perturbation	2.801620994	0.003
Roll Indoor 1 CCW10	No Perturbation	1.944462761383190	0.192
Roll Indoor 1 CCW20	No Perturbation	2.075182315305000	0.113
Roll Indoor 1 CCW30	No Perturbation	2.614791683	0.008
Roll Indoor 2 CW10	No Perturbation	3.477905585	0.000
Roll Indoor 2 CW20	No Perturbation	2.352449382	0.032
Roll Indoor 2 CW30	No Perturbation	2.096222846435500	0.103
Roll Indoor 2 CCW10	No Perturbation	1.792548152887440	0.330
Roll Indoor 2 CCW20	No Perturbation	2.874962719	0.002
Roll Indoor 2 CCW30	No Perturbation	2.947880266	0.001
Roll Outdoor CW10	No Perturbation	2.3029704	0.040
Roll Outdoor CW20	No Perturbation	1.543005490338140	0.644
Roll Outdoor CW30	No Perturbation	1.853395372603340	0.268
Roll Outdoor CCW10	No Perturbation	2.29216105	0.042
Roll Outdoor CCW20	No Perturbation	2.74420739	0.004
Roll Outdoor CCW30	No Perturbation	2.646838422	0.007
ML Axis Trans - Kitchen	No Perturbation	1.313217169840910	0.056
AP Axis Trans - Corridor Forward	No Perturbation	1.629620586	0.004
AP Axis Trans - Corridor Backward	No Perturbation	0.930249896549105	0.594
Pitch Indoor - Bathroom	No Perturbation	2.314149133	0.038
Pitch Indoor - Fridge	No Perturbation	1.964383729034580	0.178
Window Roof Beam Walking - Vertigo	No Perturbation	3.56612488	0.000
Window Roof Beam Walking - Vertigo No avatar	No Perturbation	2.483360311	0.016
Simple Roof - Vertigo	No Perturbation	1.359584992083050	0.908
Simple Roof - Vertigo No avatar	No Perturbation	1.425731916335660	0.790
Pitch Outdoor - Near Car Oil	No Perturbation	1.267575806052360	0.928
Bedroom Syncope	No Perturbation	1.38901541	0.030
Garden - Object Avoidance	No Perturbation	1.258331013797960	0.086
Electricity Pole - Vertigo	No Perturbation	2.592656256	0.009
Electricity Pole - Vertigo No avatar	No Perturbation	2.033031547963360	0.135
Free Fall	No Perturbation	1.668146467	0.004
Stairs	No Perturbation	1.99481298	0.009
Trip - Sidewalk	No Perturbation	3.3731591	0.000

Table 33: Right Gastrocnemius Medial average activation

Dependent Variable: RGMAVG
Dunnett t (2-sided): control group "No Perturbation"

(I) Label		Mean Difference (I-J)	Sig.
Roll Indoor 1 CW10	No Perturbation	3.10630367	0.001
Roll Indoor 1 CW20	No Perturbation	2.747095743	0.007
Roll Indoor 1 CW30	No Perturbation	2.768976841	0.006
Roll Indoor 1 CCW10	No Perturbation	2.81419115	0.005
Roll Indoor 1 CCW20	No Perturbation	3.905388069	0.000
Roll Indoor 1 CCW30	No Perturbation	4.177641353	0.000
Roll Indoor 2 CW10	No Perturbation	1.888039350961260	0.302
Roll Indoor 2 CW20	No Perturbation	1.002126927622290	0.999
Roll Indoor 2 CW30	No Perturbation	2.159077418047470	0.111
Roll Indoor 2 CCW10	No Perturbation	1.738242039580120	0.470
Roll Indoor 2 CCW20	No Perturbation	2.754626194	0.006
Roll Indoor 2 CCW30	No Perturbation	3.158247565	0.001
Roll Outdoor CW10	No Perturbation	1.781387306731530	0.417
Roll Outdoor CW20	No Perturbation	0.607638351032706	1.000
Roll Outdoor CW30	No Perturbation	1.164642639985970	0.984
Roll Outdoor CCW10	No Perturbation	2.652007438	0.011
Roll Outdoor CCW20	No Perturbation	2.745629035	0.007
Roll Outdoor CCW30	No Perturbation	2.980962358	0.002
ML Axis Trans - Kitchen	No Perturbation	1.752899996	0.002
AP Axis Trans - Corridor Forward	No Perturbation	2.269476042	0.000
AP Axis Trans - Corridor Backward	No Perturbation	2.243209186	0.000
Pitch Indoor - Bathroom	No Perturbation	2.682294342	0.010
Pitch Indoor - Fridge	No Perturbation	2.509072552	0.023
Window Roof Beam Walking - Vertigo	No Perturbation	2.166129994942240	0.108
Window Roof Beam Walking - Vertigo No avatar	No Perturbation	1.698699839750240	0.519
Simple Roof - Vertigo	No Perturbation	1.059685103574840	0.998
Simple Roof - Vertigo No avatar	No Perturbation	0.886259206256088	1.000
Pitch Outdoor - Near Car Oil	No Perturbation	2.288986185282650	0.064
Bedroom Syncope	No Perturbation	0.048115826602937	1.000
Garden - Object Avoidance	No Perturbation	0.856300377740972	0.816
Electricity Pole - Vertigo	No Perturbation	1.911186316970140	0.280
Electricity Pole - Vertigo No avatar	No Perturbation	1.589684056623660	0.659
Free Fall	No Perturbation	1.727080879	0.004
Stairs	No Perturbation	0.948730365928233	0.964
Trip - Sidewalk	No Perturbation	2.492861318	0.000

Table 34: Left Gastrocnemius Medial average activation

Dependent Variable: LGMAVG
Dunnett t (2-sided): control group "No Perturbation"

(I) Label		Mean Difference (I-J)	Sig.
Roll Indoor 1 CW10	No Perturbation	2.707988975744050	0.264
Roll Indoor 1 CW20	No Perturbation	1.632868134061870	0.984
Roll Indoor 1 CW30	No Perturbation	2.548981309739350	0.379
Roll Indoor 1 CCW10	No Perturbation	1.292941216173430	1.000
Roll Indoor 1 CCW20	No Perturbation	1.046428891761260	1.000
Roll Indoor 1 CCW30	No Perturbation	1.505983662884160	0.995
Roll Indoor 2 CW10	No Perturbation	2.777656969657160	0.222
Roll Indoor 2 CW20	No Perturbation	1.638082599642550	0.984
Roll Indoor 2 CW30	No Perturbation	2.028548150944080	0.826
Roll Indoor 2 CCW10	No Perturbation	1.935543966484540	0.885
Roll Indoor 2 CCW20	No Perturbation	1.055948951075990	1.000
Roll Indoor 2 CCW30	No Perturbation	2.299600749186970	0.598
Roll Outdoor CW10	No Perturbation	2.275364145041300	0.620
Roll Outdoor CW20	No Perturbation	0.674592976984532	1.000
Roll Outdoor CW30	No Perturbation	1.432399082876200	0.998
Roll Outdoor CCW10	No Perturbation	1.454827092500390	0.997
Roll Outdoor CCW20	No Perturbation	2.196185949878020	0.691
Roll Outdoor CCW30	No Perturbation	2.443042758023660	0.468
ML Axis Trans - Kitchen	No Perturbation	1.471312816553630	0.419
AP Axis Trans - Corridor Forward	No Perturbation	1.238263898538520	0.793
AP Axis Trans - Corridor Backward	No Perturbation	1.424934737559330	0.487
Pitch Indoor - Bathroom	No Perturbation	3.200112290068350	0.067
Pitch Indoor - Fridge	No Perturbation	1.549403888596720	0.993
Window Roof Beam Walking - Vertigo	No Perturbation	1.623028559791330	0.986
Window Roof Beam Walking - Vertigo No avatar	No Perturbation	0.981380716626383	1.000
Simple Roof - Vertigo	No Perturbation	1.041716147376590	1.000
Simple Roof - Vertigo No avatar	No Perturbation	0.836904937521562	1.000
Pitch Outdoor - Near Car Oil	No Perturbation	2.087198816306320	0.783
Bedroom Syncope	No Perturbation	0.196001377559589	1.000
Garden - Object Avoidance	No Perturbation	0.687570354157092	1.000
Electricity Pole - Vertigo	No Perturbation	1.421963198631690	0.998
Electricity Pole - Vertigo No avatar	No Perturbation	1.379721029179060	0.999
Free Fall	No Perturbation	2.188941471	0.017
Stairs	No Perturbation	1.175490357323440	0.994
Trip	No Perturbation	2.674281756	0.000

Table 35: Right Rectus Femoris average activation

Dependent Variable: RFAVG
Dunnett t (2-sided): control group "No Perturbation"

(I) Label		Mean Difference (I-J)	Sig.
Roll Indoor 1 CW10	No Perturbation	0.699965740245334	0.996
Roll Indoor 1 CW20	No Perturbation	0.362430683544032	1.000
Roll Indoor 1 CW30	No Perturbation	0.581029049960165	1.000
Roll Indoor 1 CCW10	No Perturbation	0.859896130661107	0.935
Roll Indoor 1 CCW20	No Perturbation	0.199858839827132	1.000
Roll Indoor 1 CCW30	No Perturbation	0.578154500926549	1.000
Roll Indoor 2 CW10	No Perturbation	0.197206808280718	1.000
Roll Indoor 2 CW20	No Perturbation	0.627998980570361	0.999
Roll Indoor 2 CW30	No Perturbation	0.686347310736688	0.997
Roll Indoor 2 CCW10	No Perturbation	0.574718034902844	1.000
Roll Indoor 2 CCW20	No Perturbation	0.937550562419996	0.850
Roll Indoor 2 CCW30	No Perturbation	1.198278874732280	0.379
Roll Outdoor CW10	No Perturbation	0.552466348833032	1.000
Roll Outdoor CW20	No Perturbation	1.036299771749620	0.685
Roll Outdoor CW30	No Perturbation	0.926365820619946	0.865
Roll Outdoor CCW10	No Perturbation	0.188976661989746	1.000
Roll Outdoor CCW20	No Perturbation	0.894776590912832	0.902
Roll Outdoor CCW30	No Perturbation	0.696487277294276	0.996
ML Axis Trans - Kitchen	No Perturbation	0.206867154932963	1.000
AP Axis Trans - Corridor Forward	No Perturbation	0.340869722050998	1.000
AP Axis Trans - Corridor Backward	No Perturbation	0.614508560408416	0.670
Pitch Indoor - Bathroom	No Perturbation	0.480112706243617	1.000
Pitch Indoor - Fridge	No Perturbation	0.345444482841427	1.000
Window Roof Beam Walking - Vertigo	No Perturbation	0.839848295537213	0.950
Window Roof Beam Walking - Vertigo No avatar	No Perturbation	0.522092934826758	1.000
Simple Roof - Vertigo	No Perturbation	0.265704289830341	1.000
Simple Roof - Vertigo No avatar	No Perturbation	0.211429132814236	1.000
Pitch Outdoor - Near Car Oil	No Perturbation	0.697441399398723	0.996
Bedroom Syncope	No Perturbation	0.790835635311581	0.180
Garden - Object Avoidance	No Perturbation	0.865295718132902	0.083
Electricity Pole - Vertigo	No Perturbation	1.003414149300670	0.745
Electricity Pole - Vertigo No avatar	No Perturbation	0.693006143101604	0.997
Free Fall	No Perturbation	0.715932681350940	0.409
Stairs	No Perturbation	0.587368632512244	0.985
Trip - Sidewalk	No Perturbation	1.511436025	0.000

Table 36: Right Semitendinosus average activation

Dependent Variable: STAVG
Dunnett t (2-sided): control group "No Perturbation"

(I) Label		Mean Difference (I-J)	Sig.
Roll Indoor 1 CW10	No Perturbation	0.985773869318605	0.992
Roll Indoor 1 CW20	No Perturbation	0.411982605695332	1.000
Roll Indoor 1 CW30	No Perturbation	-0.015385553509775	1.000
Roll Indoor 1 CCW10	No Perturbation	0.320729053136658	1.000
Roll Indoor 1 CCW20	No Perturbation	0.707262382004043	1.000
Roll Indoor 1 CCW30	No Perturbation	1.147450248833310	0.939
Roll Indoor 2 CW10	No Perturbation	0.331422485985998	1.000
Roll Indoor 2 CW20	No Perturbation	0.716166858710562	1.000
Roll Indoor 2 CW30	No Perturbation	0.529206765997514	1.000
Roll Indoor 2 CCW10	No Perturbation	-0.125046302169388	1.000
Roll Indoor 2 CCW20	No Perturbation	0.571575265792170	1.000
Roll Indoor 2 CCW30	No Perturbation	1.044792613293530	0.981
Roll Outdoor CW10	No Perturbation	0.095475242259429	1.000
Roll Outdoor CW20	No Perturbation	0.488765449662409	1.000
Roll Outdoor CW30	No Perturbation	0.293867522345581	1.000
Roll Outdoor CCW10	No Perturbation	0.135950609353658	1.000
Roll Outdoor CCW20	No Perturbation	0.485625960332069	1.000
Roll Outdoor CCW30	No Perturbation	0.851983821897890	0.999
ML Axis Trans - Kitchen	No Perturbation	0.274995680933047	1.000
AP Axis Trans - Corridor Forward	No Perturbation	0.498894607454150	1.000
AP Axis Trans - Corridor Backward	No Perturbation	0.694395311266030	0.918
Pitch Indoor - Bathroom	No Perturbation	0.872613525542348	0.999
Pitch Indoor - Fridge	No Perturbation	0.363576607073626	1.000
Window Roof Beam Walking - Vertigo	No Perturbation	1.657811295525130	0.319
Window Roof Beam Walking - Vertigo No avatar	No Perturbation	0.949141310068473	0.995
Simple Roof - Vertigo	No Perturbation	1.194653133843610	0.944
Simple Roof - Vertigo No avatar	No Perturbation	0.799893397924553	1.000
Pitch Outdoor - Near Car Oil	No Perturbation	0.147205590545215	1.000
Bedroom Syncope	No Perturbation	0.986508474885123	0.298
Garden - Object Avoidance	No Perturbation	1.349218998	0.015
Electricity Pole - Vertigo	No Perturbation	1.929391276469530	0.103
Electricity Pole - Vertigo No avatar	No Perturbation	1.774717713264610	0.203
Free Fall	No Perturbation	0.772454004621497	0.833
Stairs	No Perturbation	0.709745995179707	0.997
Trip - Sidewalk	No Perturbation	2.374449504	0.000

Table 37: Average accelerometry value of the pelvis on the X-axis

Dependent Variable: PelAccxAVG
Dunnett t (2-sided): control group "No Perturbation"

(I) Label		Mean Difference (I-J)	Sig.
Roll Indoor 1 CW10	No Perturbation	-0.119308209	0.036
Roll Indoor 1 CW20	No Perturbation	0.001485443370296	1.000
Roll Indoor 1 CW30	No Perturbation	-0.035661038648688	1.000
Roll Indoor 1 CCW10	No Perturbation	-0.043105827068708	1.000
Roll Indoor 1 CCW20	No Perturbation	0.075951178022178	0.724
Roll Indoor 1 CCW30	No Perturbation	-0.061159441489911	0.964
Roll Indoor 2 CW10	No Perturbation	-0.009286873287281	1.000
Roll Indoor 2 CW20	No Perturbation	0.092930119623961	0.309
Roll Indoor 2 CW30	No Perturbation	0.054888556762268	0.992
Roll Indoor 2 CCW10	No Perturbation	0.079513158889391	0.634
Roll Indoor 2 CCW20	No Perturbation	0.031518526117218	1.000
Roll Indoor 2 CCW30	No Perturbation	-0.019055688769580	1.000
Roll Outdoor CW10	No Perturbation	0.089040065909324	0.393
Roll Outdoor CW20	No Perturbation	0.031134949478232	1.000
Roll Outdoor CW30	No Perturbation	0.048599460747291	0.999
Roll Outdoor CCW10	No Perturbation	0.008713664024468	1.000
Roll Outdoor CCW20	No Perturbation	0.065822077905075	0.918
Roll Outdoor CCW30	No Perturbation	-0.024539570096379	1.000
ML Axis Trans - Kitchen	No Perturbation	-0.036437670604969	0.958
AP Axis Trans - Corridor Forward	No Perturbation	-0.036880326791476	0.960
AP Axis Trans - Corridor Backward	No Perturbation	-0.044275084593301	0.741
Pitch Indoor - Bathroom	No Perturbation	-0.102161743843130	0.159
Pitch Indoor - Fridge	No Perturbation	-0.043181862891647	1.000
Window Roof Beam Walking - Vertigo	No Perturbation	-0.009837613695173	1.000
Window Roof Beam Walking - Vertigo No avatar	No Perturbation	-0.011226283943210	1.000
Simple Roof - Vertigo	No Perturbation	-0.008428145496137	1.000
Simple Roof - Vertigo No avatar	No Perturbation	-0.007873657885809	1.000
Pitch Outdoor - Near Car Oil	No Perturbation	0.014377110389235	1.000
Bedroom Syncope	No Perturbation	-0.059028182057526	0.184
Garden - Object Avoidance	No Perturbation	-0.044424426616709	0.735
Electricity Pole - Vertigo	No Perturbation	-0.005958299541883	1.000
Electricity Pole - Vertigo No avatar	No Perturbation	-0.003724161200502	1.000
Free Fall	No Perturbation	-0.077372938	0.016
Stairs	No Perturbation	-0.024232459141920	1.000
Trip - Sidewalk	No Perturbation	0.032384095163653	0.992

Table 38: Average accelerometry value of the pelvis on the Y-axis

Dependent Variable: PelAccyAVG
Dunnett t (2-sided): control group "No Perturbation"

(I) Label		Mean Difference (I-J)	Sig.
Roll Indoor 1 CW10	No Perturbation	-0.019156068744117	1.000
Roll Indoor 1 CW20	No Perturbation	0.046585652433260	0.995
Roll Indoor 1 CW30	No Perturbation	-0.050398606785153	0.985
Roll Indoor 1 CCW10	No Perturbation	0.061744678548020	0.850
Roll Indoor 1 CCW20	No Perturbation	-0.011246342351866	1.000
Roll Indoor 1 CCW30	No Perturbation	-0.097596640962541	0.077
Roll Indoor 2 CW10	No Perturbation	0.024695424285692	1.000
Roll Indoor 2 CW20	No Perturbation	-0.030283073216625	1.000
Roll Indoor 2 CW30	No Perturbation	0.032938450919864	1.000
Roll Indoor 2 CCW10	No Perturbation	-0.068498746982669	0.678
Roll Indoor 2 CCW20	No Perturbation	-0.004423284802159	1.000
Roll Indoor 2 CCW30	No Perturbation	0.029774188599725	1.000
Roll Outdoor CW10	No Perturbation	-0.002263402923174	1.000
Roll Outdoor CW20	No Perturbation	0.016581080937336	1.000
Roll Outdoor CW30	No Perturbation	-0.000245233930631	1.000
Roll Outdoor CCW10	No Perturbation	-0.002883735760655	1.000
Roll Outdoor CCW20	No Perturbation	-0.048104335049574	0.992
Roll Outdoor CCW30	No Perturbation	0.068576448958735	0.676
ML Axis Trans - Kitchen	No Perturbation	0.004693273921583	1.000
AP Axis Trans - Corridor Forward	No Perturbation	0.013369126863331	1.000
AP Axis Trans - Corridor Backward	No Perturbation	0.102917761	0.000
Pitch Indoor - Bathroom	No Perturbation	0.015340377584652	1.000
Pitch Indoor - Fridge	No Perturbation	-0.074799266690465	0.493
Window Roof Beam Walking - Vertigo	No Perturbation	0.006742537685738	1.000
Window Roof Beam Walking - Vertigo No avatar	No Perturbation	0.008481912128048	1.000
Simple Roof - Vertigo	No Perturbation	0.002655106833035	1.000
Simple Roof - Vertigo No avatar	No Perturbation	0.001509060161740	1.000
Pitch Outdoor - Near Car Oil	No Perturbation	0.047790385795662	0.993
Bedroom Syncope	No Perturbation	0.036001969523966	0.862
Garden - Object Avoidance	No Perturbation	0.029512236710898	0.986
Electricity Pole - Vertigo	No Perturbation	0.004071963758247	1.000
Electricity Pole - Vertigo No avatar	No Perturbation	0.001497742267666	1.000
Free Fall	No Perturbation	0.070848516	0.009
Stairs	No Perturbation	0.010569542077887	1.000
Trip - Sidewalk	No Perturbation	-0.000118785402072	1.000