

FACULDADE DE ENGENHARIA DA UNIVERSIDADE DO PORTO

Development of a methodology to design a lower limb assistive exoskeleton

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Resumo

Durante o envelhecimento natural do ser humano, o corpo perde massa muscular. Este fator regularmente conduz a uma mobilidade reduzida moderada e adoção de uma vida sedentária. Este é um dos motivos que leva algumas pessoas idosas a sofrer de isolamento e, conseqüentemente, depressão. Cerca de 20% da população sénior na União Europeia sofre de mobilidade reduzida moderada o que, em 2017, corresponde a cerca de 20 milhões de pessoas segundo dados do Eurostat.

Com o envelhecimento da população, este número de pessoas tende a incrementar, o que aumenta a importância da criação de soluções tecnológicas adequadas e focadas na mobilidade reduzida moderada na população sénior.

Esta tese consiste em estabelecer uma metodologia para desenvolver um exosqueleto ativo. Para esse efeito, foram criados ciclos de trabalho para várias fases de desenvolvimento de um exosqueleto assistivo. O trabalho está focado no desenvolvimento de um exosqueleto assistivo para os membros inferiores dedicado a idosos com mobilidade reduzida moderada.

Para desenvolver o suporte mecânico, foram feitos estudos ergonómicos e anatómicos de modo a poder construir um sistema que possa ser adaptável a várias anatomias possíveis do corpo humano. Outro objetivo foi desenvolver um sistema que pode ser usado junto do corpo, permitindo a utilização por baixo de roupas largas, possibilitando uma redução da ansiedade social por parte dos utilizadores.

Depois de desenvolvido o sistema de suporte mecânico, foram feitos testes biomecânicos usando captura de movimento e plataformas de força, simulando com pesos localizados o exosqueleto montado para se observar o comportamento dinâmico do sistema do corpo humano e do exosqueleto. Esta informação é usada para determinar os requisitos dos actuadores a aplicar no exosqueleto.

Foi desenvolvido também um sistema autónomo de aquisição de dados customizado para controlar protótipos funcionais do exosqueleto. O sistema é composto por um mini-computador, circuito de aquisição de sinais EMG, acelerómetro e giroscópio, e um sistema de baterias com circuito para carregamento. O software desenvolvido permite adquirir 8 sinais EMG e observar os mesmos em tempo real a partir de um dispositivo sem fios capaz de reproduzir a páginas web compatíveis com a norma HTML5.

Também foi feito um estudo sobre a colocação dos elétrodo EMG de superfície a partir de resultados experimentais, para usar como base num sistema de controlo para o exosqueleto.

Por fim, foi feita uma proposta de sistema de atuação do exosqueleto, tendo em conta os requisitos obtidos sobre binário, velocidade, peso e consumo energético dos actuadores, obtidos nas fases anteriores do trabalho executado.

Abstract

During a person's natural ageing process, the body loses muscle mass. This factor usually leads to a moderate mobility impairment and the adoption of a sedentary life as a result. This is one of the reasons that leads some elders to suffer from isolation and, consequently, depression.

Close to 20% of the elder population in the European Union suffers from moderate mobility impairment which, in 2017, corresponds to close to 20 million people according to the latest Eurostat studies.

With the population ageing happening in most developed countries, the number of people in this situation tends to grow larger, which increases the importance for creating adequate technological solutions that are focused on the moderate mobility impairment for the senior population.

This thesis consists in establishing the methodology to design an active exoskeleton. For that purpose, a number of workflows was created for the several development stages of an assistive exoskeleton. This work is focused on developing an assistive exoskeleton for the lower limbs, dedicated to elders who suffer from moderate mobility impairment.

To design the mechanical support frame, ergonomic and anatomical studies were performed to allow the construction of a system that can adapt to several human body shapes. Another objective was to develop a system that can be used close to the human body, allowing it to be worn beneath large clothes, which could reduce the social awareness and awkwardness from its users.

After the mechanical support frame was designed, biomechanical trials were performed using motion capture and pressure plates with the human subjects carrying weights placed along the lower limbs. This was performed to simulate the dynamic behaviour of a system with the human body and the exoskeleton. This information is used to determine the torque requirements for the actuators to apply in the exoskeleton.

An custom and autonomous EMG sensor acquisition module was developed to control future functional prototypes of the exoskeleton. The system is composed of a single-board computer, a data acquisition module for EMG, accelerometer and gyroscope, and a battery system with a charging circuit. The developed software allows the acquisition of 8 discrete EMG signals and their observation in real time through any wireless device capable of reproducing HTML5 compatible web pages as well as a

It was also made a study for the surface EMG electrode placement using experimental results, to use as groundwork for developing a prototype control system.

After this work, a preliminary proposition for the actuation system was made, that takes into consideration the requirements for torque, angular speed, weight and power consumption of the actuators, which were acquired during the previous phases of the work developed during the project.

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I would like to dedicate this thesis to my close family, my wife and daughter, Paula and Isabel. To Paula for the endless support, motivation and sheer patience during the harshest times of a several year long project with some "ups" and more "downs" than anyone ever expects. To young Isabel for the innocent giggles and hugs that can recharge a father's energies and keep up working for that very important additional couple of hours during the night.

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Daniel Sá Pina

Only a fool learns from his own mistakes. The wise man learns from the mistakes of others.

- Otto von Bismarck

Contents

1	Introduction	1
1.1	Motivation	1
1.2	Objective	4
1.3	Thesis Structure	5
2	State of the Art	7
2.1	Current Solutions	8
2.1.1	Classic Solutions	8
2.1.2	Active Exoskeletons or Orthoses for Lower Limb Assistance	9
2.2	A Database for Research on Active Exoskeletons and Orthoses for the Lower Limbs	11
2.2.1	Database Population and Construction	12
2.2.2	Results: Localization and General Statistics	15
2.2.3	Results: Actuation Technologies	18
2.2.4	Results: Sensor Technologies	19
2.2.5	Database Results Discussions	21
2.3	Actuators in Robotics for Human Interaction	21
2.3.1	DC Motors	22
2.3.2	Electroactive Polymers	22
2.3.3	Shape Memory Alloys	23
2.3.4	McKibben Actuator	24
2.3.5	Flexible Fluidic Actuators	25
2.3.6	Pneumatic Cylinders and Hydraulic Actuators	26
2.4	Conclusions	27
3	Mechanical Design for the Support Frame	29
3.1	Design Goals	29
3.1.1	Active and Passive Movements	29
3.1.2	Materials and Components	30
3.2	Design Cycle	32
3.3	Component Description	34
3.3.1	Belt to Hip Section	35
3.3.2	Hip to Knee Section	39
3.3.3	Knee to Heel Section	43
3.4	Torque and Force Tests	48
3.4.1	Hip Flexion and Extension	51
3.4.2	Knee Flexion and Extension	53
3.4.3	Extraordinary Force Loads	56
3.5	Conclusions	56

4	Multi Body Simulations from Motion Capture Results	59
4.1	Methodology	59
4.2	Motion Capture Data Acquisition	60
4.3	OpenSIM Workflow	63
4.4	OpenSIM Results	67
4.4.1	Gait Results	67
4.4.2	Stair Ascend and Descend Results	70
4.4.3	Chair Sitting and Standing Results	74
4.5	Conclusions	76
5	Sensor, Control and Actuation Systems	79
5.1	Sensor and Control Systems	79
5.1.1	Myoelectric Sensors and electrode placement	79
5.1.2	Experimental sEMG Results	81
5.1.3	Prototype for Sensor and Control Unit	88
5.2	Actuation System	91
5.3	Conclusions	94
6	Conclusions and Future Work	95
6.1	Project Achievements	95
6.2	Project Limitations	96
6.3	Future Work	96
A	Publications	99
A.1	Journal Publications	99
A.2	Conference Publications	99
	References	101

List of Figures

1.1	Cycle of symptoms and consequences during aging.	2
1.2	Proportions and number of people (in millions) in the European Union aged 65 years or older with moderate or severe mobility impairment, according to a 2014 EUROSTAT study [1].	3
1.3	Evolution of the age percentages between 2016 and 2040, according to EUROSTAT study from 2017 [2].	3
1.4	Example of clothes that would hide the active exoskeleton proposed.	4
2.1	On the left, a pair of crutches. On the right, an assistive walker.	8
2.2	Promotional pictures of exoskeletons for paraplegics or patients with severe motion disability. From left to right: Ekso, Indego, ReWalk.	9
2.3	Still pictures of promotions videos of the HAL "Body Suits". On the left and center, the lower limb version showing the ability to climb stairs. On the right, the full body version assisting the wearer in holding a heavy weight.	10
2.4	Promotional Pictures of the Keeogo exoskeleton [3].	11
2.5	Simplified representation of the database's entity-relationship model, using Crow's Foot Notation.	13
2.6	Screenshot of the starting page of the custom database for active orthoses and exoskeletons.	13
2.7	Screenshot of the database GUI page dedicated to editing and adding new articles.	14
2.8	Screenshot of the database GUI page showing the type of data that can be immediately extracted from the database	14
2.9	Screenshot of the data displayed in the custom database's GUI. In this case, the number of articles per publication is shown.	15
2.10	In blue, the number of articles related to active orthoses and exoskeletons gathered per year. In red, the number of authors associated to the gathered articles.	16
2.11	Number of articles associated to each Conference.	16
2.12	Number of articles associated to each Research Institution, divided by target limbs or body parts.	17
2.13	Proportions of articles associated to each country, divided by target limbs or body parts.	17
2.14	Number of orthosis/exoskeleton projects, grouped by the devices' mobility and showing the proportions between actuation technologies.	19
2.15	Number of occurrences for each sensor type, taken from the gathered articles	20
2.16	Proportion of sensor technologies, grouped by physical variable classes.	20
2.17	Basic Functioning Principle of a Dielectric Electro-Active Polymer [4]	22
2.18	Pictures of a working Spring-Roll shaped EAP actuator [5]	23

2.19	Examples of medical appliances using SMAs. On the left are grippers, scissors and tongs for laparoscopies; on the right, SMA bone plates fixed in a jaw. [6]	24
2.20	Example of a working McKibben actuator. On the left, the actuator is deflated. On the right, the actuator is inflated, bending the nail. [7]	24
2.21	Basic functioning of the Flexible Fluidic Actuator. On the left, the elastic chamber is deflated. On the right, the elastic chamber is inflated, creating a rotary motion between the plastic films [8].	25
2.22	Examples of current Flexible Fluidic Actuator [9].	26
3.1	Diagram representing the gait cycle for the right lower limb (highlighted in blue) from a healthy human being. The muscle groups that are generating work for the hip extension/flexion and knee flexion/extension during the Stance and Swing phases, can be seen highlighted in red. The direction of motion of each joint in the sagittal plane is represented by the yellow arrows. These movements are actively assisted in the proposed system.	29
3.2	Screenshot of the MakeHuman interface to create anatomically correct 3D models of human bodies.	31
3.3	Anatomically correct 3D human models created to develop the mechanical frame.	31
3.4	Cyclical development diagram for the mechanical frame design of the active exoskeleton.	33
3.5	Exoskeleton adaptable to a smaller female body to the left and a larger male body to the right, in the foot and ankle areas.	33
3.6	Planes of human anatomy. Sagittal plane in red, Transverse plane in green and Coronal plane in blue.	34
3.7	Render of the full exoskeleton frame. The components are divided into 3 sections: Belt to Hip in section 3.3.1, Hip to Knee in section 3.3.2 and Knee to Heel in section in 3.3.3	34
3.8	On the left, an image of component "Hip-Fit" with some measurements in millimeters. On the right, component "Hip-Fit" housing a spherical bearing and connected to component "Belt-Hip-B".	35
3.9	On the left, the components "Hip-Fit" and "Belt-Hip-B" are adapted to the waistline and hip joint of a female model. On the right, the same components are adapted to a male model.	36
3.10	On the left, component "Belt-Hip-A". On the right, component "Belt-Hip-B". Some measurements are shown in millimeters	36
3.11	Components "Belt-Hip-A" and "Belt-Hip-B" adapted to models with different distances between the hip joint and the waist.	37
3.12	Component "Belt-Distribute" on the left and component "Belt-Extension-A" on the right. Some measurements are shown in millimeters.	37
3.13	Components "Belt-Distribute", "Belt-Extension-A" and "Belt-Extension-B" adapted to two different examples. To the left, a model with more volume towards the lumbar region. To the right, a model with more volume towards the abdominal region.	38
3.14	From left to right, components "Waist-Extension-A", "Waist-Extension-B" and "Waist-Contact". Some measurements are shown in millimeters.	38
3.15	Components "Waist-Extension-A/B" and "Waist-Contact" adapted to a model with a slimmer abdominal area on the left and a model with a larger abdominal area to the right.	39
3.16	Component "Waist-Contact" with the surface that will need cushioning highlighted in red.	39

3.17	From left to right, components "Thigh-Up-A", "Thigh-Up-B" and "Thigh-Up-C". Some measurements are shown in millimeters.	40
3.18	On the left, components "Thigh-Up-A/B/C" are adapted to a model with a shorter hip-to-knee distance and slimmer thigh. On the right, the same group of components is adapted to a model with a longer hip-to-knee distance and a thicker thigh.	40
3.19	Component "Thigh-Up-C" with the surface that will need cushioning highlighted in red.	41
3.20	Components "Thigh-Mid-A" on the left and "Thigh-Mid-B" on the right. Some measurements are shown in millimeters.	41
3.21	Components "Thigh-Mid-A/B" adapted to a wider thigh on the left and to a narrower thigh on the right.	42
3.22	Surfaces that will need cushioning in "Thigh-Mid-A/B" highlighted in red.	42
3.23	Components "Knee-Up-A" on the left and "Knee-Up-B" on the right. Some measurements are shown in millimeters and angles in degrees.	43
3.24	Components "Knee-Up-A/B" adapted to a smaller knee volume on the left, and a larger knee volume on the right.	43
3.25	From left to right, components "Knee-Down-A", "Knee-Down-B" and "Knee-Down-C". Some measurements are shown in millimeters for distances and degrees for angles.	44
3.26	Components "Knee-Down-A/B/C" adapted to a smaller knee and shorter knee-to-ankle distance on the left, and a larger knee with longer knee-to-ankle distance on the left.	44
3.27	Components "Shin-Up-A/B". Some measurements are seen in millimeters.	45
3.28	Components "Shin-Up-A/B" adapted to a narrower upper shin on the left, and a wider upper shin on the right.	45
3.29	Surfaces that will need cushioning in components "Shin-Up-A/B", highlighted in red.	45
3.30	From left to right, components "Shin-Down-A", "Shin-Down-B" and "Shin-Down-C". Some measurements are seen in millimeters.	46
3.31	On the left, components "Shin-Down-A/B/C" adapted to a model with a shorter knee-to-ankle distance. On the right, the same components adapted to a model with a longer knee-to-ankle distance.	46
3.32	Surfaces from the "Shin-Down-A/B/C" that need cushioning, highlighted in red.	47
3.33	Component "Ankle-Up-A" in the top highlighted in blue and "Ankle-Up-B" in the bottom. Pictures from left and right show the same pair of components from two different perspectives.	47
3.34	Component "Ankle-Down-A" in the top highlighted in blue and "Ankle-Down-B" in the bottom. The picture from the left shows the spherical bearing fitted in "Ankle-Down-A" and the picture from the right shows the same pair of components from two different perspectives and without the bearing.	48
3.35	Components "Ankle-Up-A/B" and "Ankle-Down-A/B" adapted to a smaller foot on the left, and a larger foot on the right.	48
3.36	Render of the exoskeleton with the planned fastening belts highlighted in blue.	51
3.37	Results for the Hip Flexion Maximum Stress with a 60 N.m torque.	51
3.38	Results for the Hip Flexion Nominal Stress with a 30 N.m torque.	52
3.39	Results for the Hip Extension Maximum Stress with a 150 N.m torque.	52
3.40	Results for the Hip Extension Nominal Stress with a 80 N.m torque.	53

3.41	Instant of maximum Knee Flexion determined in a study from Hamner <i>et al</i> [10]. At 0.5 seconds during heel contact, the knee joint torque is 60 N.m at a 22° angle with the thigh.	53
3.42	Results for the Knee Flexion Maximum Stress, with hip flexion included.	54
3.43	Results for the Knee Extension Maximum Stress.	55
3.44	Results for the Knee Extension Nominal Stress.	55
3.45	Results for the Extraordinary Force Load in an emergency situation.	56
3.46	Render of the mechanical frame adapted to a female model with a height of 1.55 m and a male model with a height of 1.84 m.	57
4.1	Arrangement of the force plates in the biomechanics laboratory.	60
4.2	Distribution of the attached weights to approach the weight dynamics of the exoskeleton developed.	61
4.3	Pictures of the female subject. On top, with normal clothing and no added weights. On the bottom, with the attached weights and backpack.	61
4.4	Pictures of the female subject. On top, with normal clothing and no added weights. On the bottom, with the attached weights and backpack.	62
4.5	Pictures of the female subject. On top, with normal clothing and no added weights. On the bottom, with the attached weights and backpack.	62
4.6	Pictures of the male subject performing gait trials with normal clothing on the left and attached weights on the right.	63
4.7	Pictures of the female subject performing stair trials with normal clothing on the left and attached weights on the right.	64
4.8	Pictures of the female subject performing sit trials.	64
4.9	Workflow for the OpenSim Simulation Toolkit to achieve precise values for joint torques using RRA.	65
4.10	Maximum threshold values for the residual forces and error displacements for the simulation results when performing walking or running trials, according to OpenSIM's documentation.	66
4.11	From left to right: male subject with exoskeleton, female subject with exoskeleton, male subject without exoskeleton, female subject without exoskeleton.	66
4.12	Screenshot of the simulation with the male model with the exoskeleton on the left and without the exoskeleton on the right.	67
4.13	Hip Extension/Flexion torque values for the gait trials with the female subject. . .	68
4.14	Hip Extension/Flexion torque values for the gait trials with the male subject. . . .	68
4.15	Knee Extension/Flexion torque values for the gait trials with the female subject. .	69
4.16	Knee Extension/Flexion torque values for the gait trials with the male subject. . .	69
4.17	Screenshot of the simulation with the male model with and without the exoskeleton while climbing stairs on the left and descending stairs on the right.	70
4.18	Hip Extension/Flexion torque values for the stair ascend trials with the female subject	70
4.19	Hip Extension/Flexion torque values for the stair ascend trials with the male subject.	71
4.20	Knee Extension/Flexion torque values for the stair ascend trials with the female subject.	71
4.21	Knee Extension/Flexion torque values for the stair ascend trials with the male subject.	72
4.22	Hip Extension/Flexion torque values for the stair descend trials with the female subject.	72
4.23	Hip Extension/Flexion torque values for the stair descend trials with the male subject.	73

4.24	Left Knee Extension/Flexion torque values for the stair descend trials with the female subject.	73
4.25	Left Knee Extension/Flexion torque values for the stair descend trials with the male subject.	74
4.26	Screenshot of the simulation with the male model with the exoskeleton on the left and without the exoskeleton on the right, during the sitting and standing simulations.	74
4.27	Hip and Knee Extension/Flexion torque values for the sitting trials with the male subject.	75
4.28	Hip and Knee Extension/Flexion torque values for the standing trials with the male subject.	75
5.1	Diagram with the proposed electrode placement for EMG control.	80
5.2	Diagram with a proposed control system using EMG, potentiometers and accelerometers. In this diagram, the location for the potentiometers and accelerometers is symbolic.	81
5.3	Results for the EMG data acquisition during normal gait for the right lower limb highlighted in yellow. The readings show results for the Rectus Femoris, Fascia Latae, Biceps Femoris and Gluteus Maximus. The EMG data is synchronized with the motion capture data, retrieved into OpenSIM.	82
5.4	Results for the EMG data acquisition for the left lower limb highlighted in yellow, for stair ascending during the stance phase. The readings show results for the Rectus Femoris, Biceps Femoris and Gluteus Maximus. The EMG data is synchronized with the motion capture data, retrieved into OpenSIM.	83
5.5	Results for the EMG data acquisition for the right lower limb highlighted in yellow, for stair ascending during the swing phase. The readings show results for the Rectus Femoris and Fascia Latae. The EMG data is synchronized with the motion capture data, retrieved into OpenSIM.	84
5.6	Results for the EMG data acquisition for the left lower limb highlighted in yellow, for stair descending during the stance phase. The readings show results for the Rectus Femoris and Gluteus Maximus. The EMG data is synchronized with the motion capture data, retrieved into OpenSIM.	85
5.7	Results for the EMG data acquisition for the right lower limb highlighted in yellow, for stair descending during the swing phase. The readings show results for the Rectus Femoris, Fascia Latae and Biceps Femoris. The EMG data is synchronized with the motion capture data, retrieved into OpenSIM.	86
5.8	Results for the EMG data acquisition for the right lower limb, for stair standing and. The readings show results for the Rectus Femoris, Fascia Latae, Biceps Femoris and Gluteus Maximus. The EMG data is synchronized with the motion capture data, retrieved into OpenSIM.	87
5.9	Simplified diagram with the proposed hardware for the control system.	88
5.10	Picture of the several components in the functional prototype. 1 – Odroid C1 Mini-Computer; 2 – Data Acquisition Board; 3 – Batteries; 4 – Power Converter Board; 5 – Reference Electrode; 6 – Pairs of Differential Electrodes	89
5.11	Simplified diagram of the software architecture for the portable system for wireless sEMG Monitoring	90
5.12	Picture of the developed system in use with electrode pairs on the Rectus Femoris in each leg, plus a reference electrode in the knee.	90
5.13	Screenshot of the HTML5 webpage with the EMG readings, taken from an Android smartphone.	91

5.14	Diagram of a telemedicine consultation using the sensor system developed in the course of the project.	91
5.15	Angular velocities for the hip and knee joints recorded during the gait trials without weight.	92
5.16	Picture of the proposed Nanotec Brushless DB59M024035-A DC Motor.	93

List of Tables

2.1	Comparison between Actuator Technologies	28
3.1	Torque used in the FEM tests of the mechanical frame. Bibliographic references for values are also shown.	50
4.1	Actuation requirements for each assisted movement	76
5.1	Movement and Muscle Activation	80

Abbreviations and Symbols

AMOAD	Active Motion-Oriented Assistive Devices
EMG	Electromyography
sEMG	Surface Electromyography
RRA	Reduce Residuals Algorithm
GUI	Graphical User Interface
HTML5	Hypertext Markup Language, version 5
HAL	Hybrid Assistive Limb
FEM	Finite Element Method
CAD	Computer-Aided Design
ISO	International Organization for Standardization
EAP	Electro-Active Polymer
SMA	Shape-Memory Alloy
FFA	Flexible Fluidic Actuator
CNC	Computer Numerical Control
CMC	Computed Muscle Control
PC	Personal Computer

Chapter 1

Introduction

1.1 Motivation

This introductory section consists in explaining the various physical, psychological and social-demographic factors that contributed as motivation for initiating the project outlined in the thesis.

The motivation for the work developed throughout this thesis is based on two distinct factors: the physical and psychological effects of mobility impairment during ageing explained, and the population ageing in European countries according to recent studies.

The ageing process in the human being results in several changes in the musculoskeletal system. Among other effects, the muscles shrink and lose mass, the number and size of muscle fibers decrease, the tendons and cartilages become less tolerable to stress, the heart lowers the speed at which it can pump blood and the bones lose mass, becoming more prone to fractures [11]. One of the first major symptoms that appear with ageing is an irregular gait and decreased gait speed [12].

The appearance of these symptoms usually results from an increasingly sedentary life. Consequently, a sedentary life will aggravate the aforementioned changes in muscle and bone mass reductions. An absence of muscle stimulation results in loss of muscle mass [13] and the lack of stress applied to human bones will prevent the piezoelectric effect and mechanotransduction that maintains their density [14].

This results in a continuous self-feeding cycle where the symptoms contribute directly to worsening the conditions, as shown in Figure 1.1.

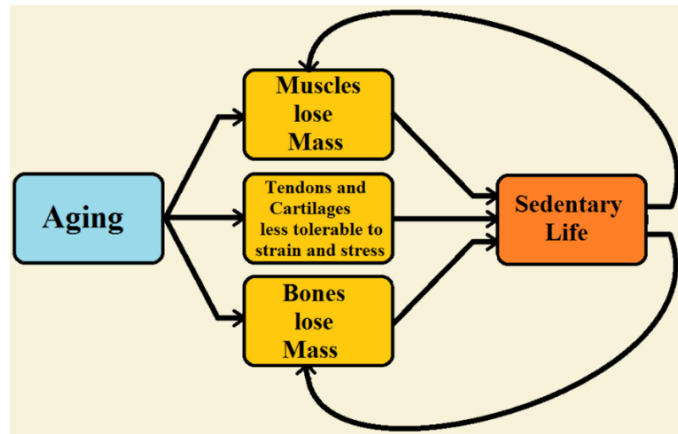


Figure 1.1: Cycle of symptoms and consequences during aging.

Additionally, adopting a sedentary life and moving less frequently also decreases a person's confidence and motor control, increasing the chances of suffering from falls [15]. The lack of an active life is also associated with social isolation [16]. The end result is a lower quality of life that can result in depression and other psychological disorders [17].

As referred in section 2.1.1, there are classical solutions for medium mobility impairment, such as crutches and walkers. However they occupy the upper limbs which results in a drastic change of lifestyle and the misuse of these solutions can cause injuries in the long term. Moreover, the mobility walkers can be difficult to use within tight spaces such as house interiors and bathrooms.

For this condition, new and innovative devices such as the HAL (Hybrid Assistive Limb), presented in 2.1.2, can be used as an evolution to modern-day crutches and walkers. However, the very high price of these devices (close to 1000 euro per month according to the product's website as of 2017) poses a problem for their adoption on a larger scale. Additionally, the device's large volume is eye-catching, which can cause social awkwardness within many European cultures.

Due to an increase in life expectancy and decrease of birth-rate, many developed countries are suffering from an ageing population. Europe is the continent with the oldest population in the world. According to a recent EUROSTAT report [1], close to 20% of the European population is aged 65 or above. This results in almost 85 Million people in that age bracket. Within these, 23% show a moderate difficulty while walking and 21% show severe difficulty, as seen in Figure 1.2.

Assuming the larger proportion of the population with severe levels of walking difficulties presented in the study is wheelchair-bound, the number of people who are in need of solutions like crutches is represented in the blue fraction of Figure 1.2, which is close to 20 Million.

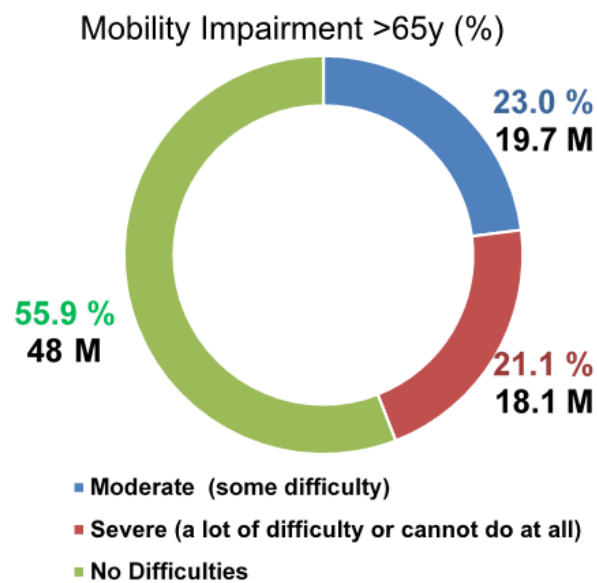


Figure 1.2: Proportions and number of people (in millions) in the European Union aged 65 years or older with moderate or severe mobility impairment, according to a 2014 EUROSTAT study [1].

Moreover, this population is bound to increase. Figure 1.3 shows the predicted age-related demographic changes predicted between 2016 and 2040.

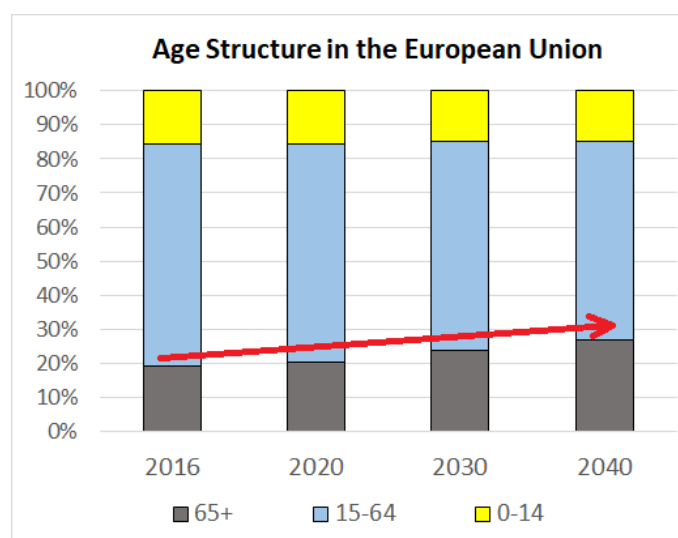


Figure 1.3: Evolution of the age percentages between 2016 and 2040, according to EUROSTAT study from 2017 [2].

EUROSTAT's projections show that, by 2040, the people aged 65 or more will account to almost one fourth of the population [2].

Therefore, the ratio between working age (18-65 years old) and non-working age people would

be close to 2 to 1, which raises questions regarding the sustainability of the European society [18]. Due to this growing concern, many studies [19, 20, 21] and initiatives [22, 23] have emerged in order to extend the active age for European citizens.

Moreover, the age evolution projections would increase the number of people aged 65 and over with moderate mobility difficulties from 20 Million to over 35 Million.

It is therefore of the utmost importance to develop solutions that contribute to solving the several problems of mobility within the 3rd and 4th ages, which are aggravated by the ageing population in the European Union.

1.2 Objective

The proposal for this work is to develop a low-profile active exoskeleton with a smaller range of utilization than other active exoskeletons, which is uniquely to assist an elderly person's gait and other daily life activities such as climbing stairs.

By focusing on these activities, it is possible that an exoskeleton that assists the lower limb activities can be achieved with a smaller frame than the examples found in section 2.1.2.

Given the different set of requirements for the present work, it is expected for this device to be able to assume a smaller frame, which could go unnoticed if the user is wearing loose clothes, such as the ones depicted in Figure 1.4.



Figure 1.4: Example of clothes that would hide the active exoskeleton proposed.

Another factor being taken into account is the cost of the exoskeleton. For example, the Keeogo exoskeleton is marketed as "low cost", is priced over 25000 euro, which is beyond reach for many European households. A recent survey carried out in Sweden [24] indicates that over 85% of the people agree that an exoskeleton for the lower limbs to assist elders in daily-life activities should cost up to 3000 euro.

Therefore, an active gait-assist exoskeleton with the aforementioned properties would be a welcome addition to a growing panoply of technological devices being researched for the growing population of elders in developed countries.

With the awareness that creating a fully functional prototype throughout the doctoral program period would be a very difficult task to achieve, the project proposes to complete the following stages for developing a Lower Limb Exoskeleton prototype:

1. Develop a thorough State of the Art research on exoskeletons in the market or during development to avoid overlapping efforts;
2. Design of the exoskeleton's mechanical support system (frame) to attach to a human's lower limbs and waist. The system must support its own weight in order to avoid encumbering its user, and have mechanical characteristics regarding degrees of freedom that will allow its user to perform all daily life activities. The mechanical support system must also be adaptable to users with different biometric measurements;
3. Through computational simulations using the Finite Element Method, certify that the mechanical frame is able to withstand the forces and torques needed to provide assistance to hip flexion/extension and knee flexion/extension to at least 50% of the values found in literature for gait and stair climbing;
4. Using a workflow of Motion Capture experimental results and multi-body simulations, determine the force and torque required for using the exoskeleton
5. Design and develop a sensory and computational system for the exoskeleton;
6. Propose an actuator and battery system according to the data previously acquired;

For all the stages, there is also an effort to use adequately priced off-the-shelf components and low-cost materials to focus on the development of a solution that can be widely adopted.

The final purpose is to create a hardware development platform from which future works may be aimed at studying control systems that will eventually allow using the exoskeleton to assist elders in gait, stair climb and sitting/lifting activities, in order to reduce a sedentary behavior and increase their quality of life.

1.3 Thesis Structure

Chapter 2 presents the work developed to determine a State of the Art relative to the project.

Section 2.1 presents the classic solutions being adopted today (in 2.1.1), as well as some examples of active exoskeletons for the lower limbs that are already in the market, in 2.1.2.

During the project, a custom database for active exoskeletons or orthoses was developed to produce statistical analyses regarding the development and scientific research of these devices. Section 2.2.1 depicts how the database was developed and populated. The statistical results from the database are presented in 2.2.2.

An overview for actuation technologies being used for robotics with human interaction is shown in section 2.3. This section shows a comparison between the types of actuators and each technology's advantages and disadvantages.

Chapter 3 describes the proposed mechanical support frame for the exoskeleton. Section 3.1 describes the mechanical design goals proposed for the project. The development cycle implemented to achieve these goals is shown in section 3.2. A description of each component and how the several components interact with each other is described in section 3.3. Section 3.4 shows the Finite Element Simulations made on selected sub-assemblies of the mechanical support, using torque and force values taken from the literature.

Chapter 4 shows the methodology used to test the force and torque demands required for the developed exoskeleton in a virtual environment, using experimental data. Section 4.1 explains the procedures for simulating the strength/torque requirements for using the developed mechanical frame and exoskeleton with two subjects. The results for gait, stair climbing and sitting/standing are presented in section 4.4.

The proposed Sensor, Control and Actuation Systems for the exoskeleton are studied in Chapter 5. An overview of sensors based on EMG (Electromyography) and their proposed placement based on literature in this project is presented in 5.1.1. Experimental EMG results for the physical activities studied in Chapter 4 are shown in section 5.1.2. Within the scope of the sensor and control system for the project, a functional prototype for a miniaturized computational system with EMG data acquisition was developed. The prototype is described in section 5.1.3. Section 5.2 shows the the proposed actuation and battery systems based on the requirements that resulted from the work described in section 4.5 from Chapter 4.

The general conclusions for the thesis are presented in Chapter 6. The subjects include an overview of the project achievements, the limitations of the project and proposals for future work.

Chapter 2

State of the Art

An orthosis is, by the definition of the International Organization for Standardization (ISO), an "externally applied device used to modify the structural and functional characteristics of the neuromuscular and skeletal systems" [25]. Its purpose is to correct, prevent or assist the mobility of a vertebrate body. An artificial exoskeleton, depending on its purpose, may be used as protection and/or with the same purpose as an orthosis. Through conventional naming, it may be assumed that an orthosis can be described as an exoskeleton for medical purposes.

Active orthoses and powered exoskeletons, among other denominations, are devices made to attach to one or several human limbs in order to assist its user's movement through means of electronically-controlled actuators and/or mechanical brakes. The technology developed for these devices can be used in many situations, such as rehabilitation [26, 27], industrial [28, 29] and general strength enhancement purposes [30, 31].

In order to simplify the reference to both kinds of devices, this work uses the term "Active Motion-Oriented Assistive Devices" (AMOADs) for Active Orthoses and Powered Exoskeletons in general.

Active orthoses and exoskeletons have suffered a sudden and recent increased interest throughout the scientific community. Various factors can be attributed to this change, such as general advances in control and actuation technologies, the increasingly larger trend of developing technological solutions for otherwise difficult tasks and the ageing of the population in certain countries.

2.1 Current Solutions

This section describes the solutions that are currently available in the market, as well as some that are in development.

2.1.1 Classic Solutions

In the present day, the most adopted solutions for moderate mobility impairment are walkers and crutches/canes, as presented in Figure 2.1.



Figure 2.1: On the left, a pair of crutches. On the right, an assistive walker.

Both solutions make use of the upper limbs to compensate for a lower strength and stability in the lower limbs. These are widely used because their cost is very accessible to all household and they are widely known and available.

The crutches and canes are normally used by people with a healthier degree of strength and dexterity in the upper limbs, as the level of mobility is higher than those using walkers. Crutches' users can easily rotate their bodies, and the devices require a small horizontal area to operate.

However, long-term use of crutches have been observed to cause nerve palsy [32, 33]. Canes and crutches are also prone to misuse, which can cause posture-related injuries [34].

The walkers or walking frames are used by people with lower strength in the upper limbs. They have four points of contact with floor, making them a better alternative for people with balance issues. Because the walkers are meant to be used with both arms at the same time, the mobility provided is more limited. The horizontal area to operate is large, so these devices are usually cumbersome to use indoors.

There are variations to the walkers with front or four wheels. For recovering patients there are evolutions to walkers being developed, with e.g. active wheels and learning algorithms to assist a recovering patient's gait patterns [35, 36].

2.1.2 Active Exoskeletons or Orthoses for Lower Limb Assistance

Most of the active exoskeletons that appeared recently in the market are dedicated to paraplegics. Examples of these devices are Indego [37], ReWalk [38] and Ekso [38], as seen in Figure 2.2.



Figure 2.2: Promotional pictures of exoskeletons for paraplegics or patients with severe motion disability. From left to right: Ekso, Indego, ReWalk.

These active exoskeletons have very different requirements from exoskeletons dedicated to people with mild to moderate lower limb weakness.

Paraplegics and people with severe locomotion disability need a device with different requirements than people who mild to moderate mobility impairments. These devices are meant to replace wheelchairs, so they need to be very robust and rigid from both mechanical and control standpoints. Additionally, their use require an additional set of crutches for balance.

Since they can also be used by people with no lower limb activity, they must be able to support the full weight of the user's body and completely replace the torques and forces originally meant to be made by the lower limbs.

There are examples of lower limb active exoskeletons for people with moderate mobility impairment, such as the Hybrid Assistive Limb (HAL) [39] in Figure 2.3.

The HAL-5 is a full-body modular exoskeleton, developed as part of the "Cyberoid Project" from the University of Tsukuba, led by Yoshiyuki Sankai [40]. HAL Lower Limb is composed of a belt/backpack and a frame from hip to the floor, being able to support its own weight.

It proposes to help elder and physically challenged people with everyday tasks like walking, sitting/standing and climbing stairs. The full body version is oriented at assisting human labor.



Figure 2.3: Still pictures of promotions videos of the HAL "Body Suits". On the left and center, the lower limb version showing the ability to climb stairs. On the right, the full body version assisting the wearer in holding a heavy weight.

It uses myoelectric sensors to analyse muscle activity and DC Servo Motors for assisting hip flexion/extension and knee flexion/extension. The full body version is estimated to operate continuously for close to 2 hours and 30 minutes.

There are two control modes for the HAL-5. One, called "Bio-Cybernic Control System", has an independent control for each motor, where the muscle activity of each limb is analyzed and translated into torque. The second is called "Robotic Autonomous Control System". Here, the signals from the muscle activity are aggregated and filtered, and afterwards compared to an existing database of predefined macro-movements like sitting, standing and walking. The second control system can be considered less versatile, but should be more convenient for cases where the wearer possesses an anomaly in the central nervous system.

The HAL Lower Limb has been studied for application in patients with spinal cord injury and stroke [41], and as of December 2017 it has obtained CE and FDA approval.

Another solution for moderate mobility impairment is the Keeego, seen in Figure 2.4. Keeego assists knee flexion and extension, so while it can assist gait, it is more effective at assisting stair climbing, sitting and standing. Its weight is fully supported by the user, so it's aimed at younger or generally stronger individuals who suffer from early stages of neurodegenerative diseases such as Parkinson's, multiple sclerosis or advanced stages of osteoarthritis. It has shown to be effective with patients suffering from hemiparesis [3].



Figure 2.4: Promotional Pictures of the Keeego exoskeleton [3].

2.2 A Database for Research on Active Exoskeletons and Orthoses for the Lower Limbs

In order to create a comprehensive and organized state-of-the-art bibliographic work, several articles related to active motion-oriented assistive devices were gathered from seven scientific databases. Afterwards, a custom database was created to contain, organize and cross the information gathered from each relevant article. This work presents the statistical results regarding the localization of research development, the actuation technologies and the man-machine interface sensors.

The purposes for creating the database are to consult and easily update a state-of-the-art work, as well as providing the means for researchers and physicians to better identify and locate the research and development of these devices.

2.2.1 Database Population and Construction

The following online search engines from scientific research databases were used to gather information into the state-of-the-art database:

1. "Science Direct" [42],
2. "IEEE Xplore" [43],
3. "MetaPress" [44],
4. "ACM Digital Library" [45],
5. "ASME Digital Library" [46],
6. "IOP Science" [47]
7. "Emerald" [48]

There is a lack of a consistent naming convention for these devices. Depending on various authors and associated research institutions, several different names have been given to the same kind of devices, regardless of their purpose, technology involved or target body parts. Therefore, in order to gather as much articles related to this subject as possible, the surveys in each of the aforementioned online search engines were repeated for each of the most recurrent terms used to describe these devices: "Exoskeleton", "Orthosis", "Orthesis" and "Rehabilitation Robot".

From the survey results, the articles containing a description or a study about an existing active orthosis or powered exoskeleton were retrieved. Articles that describe the planning stages of a prototype that is yet to assume a physical form were not retrieved, because they do not possess enough information to complete a database entry. The survey includes articles from journals and conferences found in the aforementioned search engines. Due to the intellectual property constraints related to the authors' research institution, only the articles made available under the "B-On" program were retrieved.

This survey was conducted during December 2010, therefore the statistical data shown below dates between 1974 and November 2010. The survey resulted in the retrieval of 203 articles. Out of these, 15 were state-of-the-art and/or comparison articles. From each of the remaining 188 articles, various technical and non-technical characteristics were gathered into a custom-made database, using Microsoft Office Access 2010.

The database can provide statistical information regarding the scientific development of active orthoses and exoskeletons, and automatically updates when more articles are added in the future. A simplified representation of the database's entity-relationship model can be found in Figure 2.5:

In order to increase the reliability of the statistics shown, the database differentiates between "Articles" and "Projects". When two or more articles describe a work based on the same orthosis or exoskeleton, they belong to the same "Project".

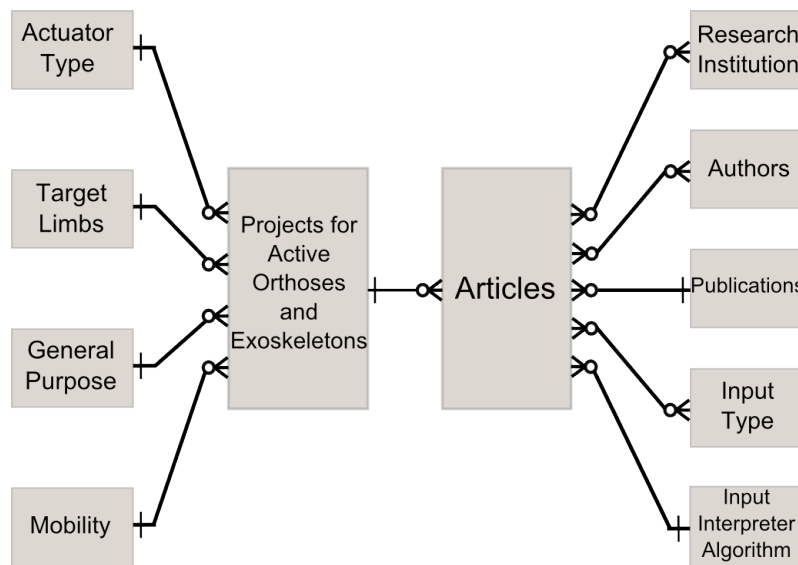


Figure 2.5: Simplified representation of the database's entity-relationship model, using Crow's Foot Notation.

A "Project" represents any device from a hardware prototype in development stage to a commercial product. The "Article" entry contains information about its associated research institutions, authors, publications and a number of technical specifications. The "Project" entry, to which one or more "Articles" belong to, contains the information about the target body parts, purpose, mobility and some technical specifications. Each entry from the database represented in Figure 2.5 is a separate table. From the aforementioned differentiation, 123 different "Projects" were identified within the 169 retrieved articles.

The database was developed with an accessible graphical user interface (GUI) that is meant to facilitate new article entries and specifications for each new project. The starting page is seen in Figure 2.6.

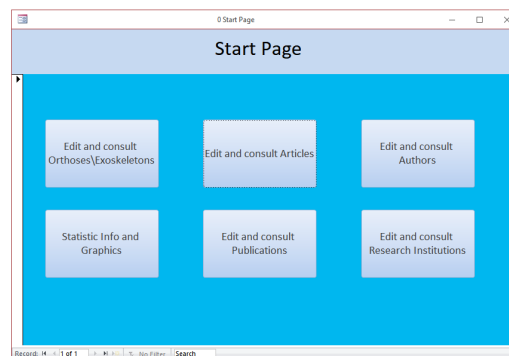


Figure 2.6: Screenshot of the starting page of the custom database for active orthoses and exoskeletons.

Each button in Figure 2.6 is a shortcut to a new page that allows editing or adding authors, publications, research institutions, articles and projects. Figure 2.7 shows an example of the interface dedicated to adding and editing new articles.

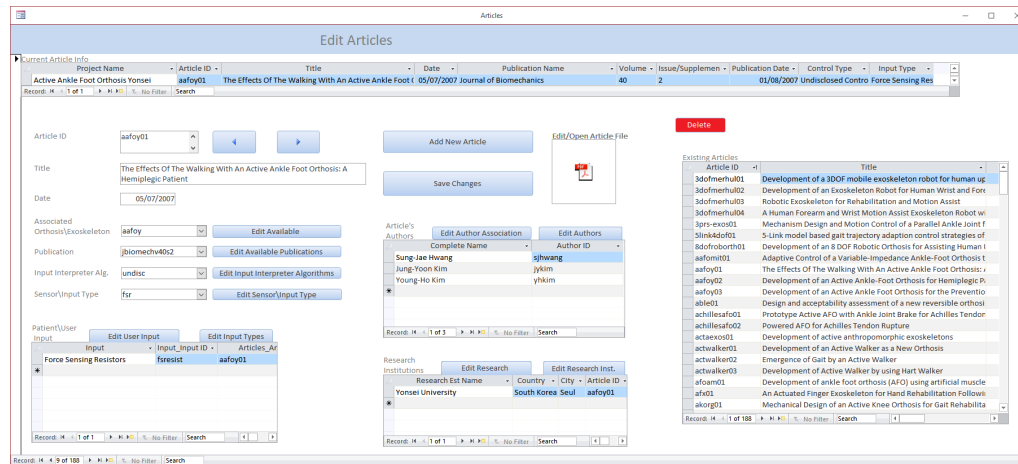


Figure 2.7: Screenshot of the database GUI page dedicated to editing and adding new articles.

The database also extracts data automatically for use in statistic studies in the future, as shown in the page shown in Figure 2.8.

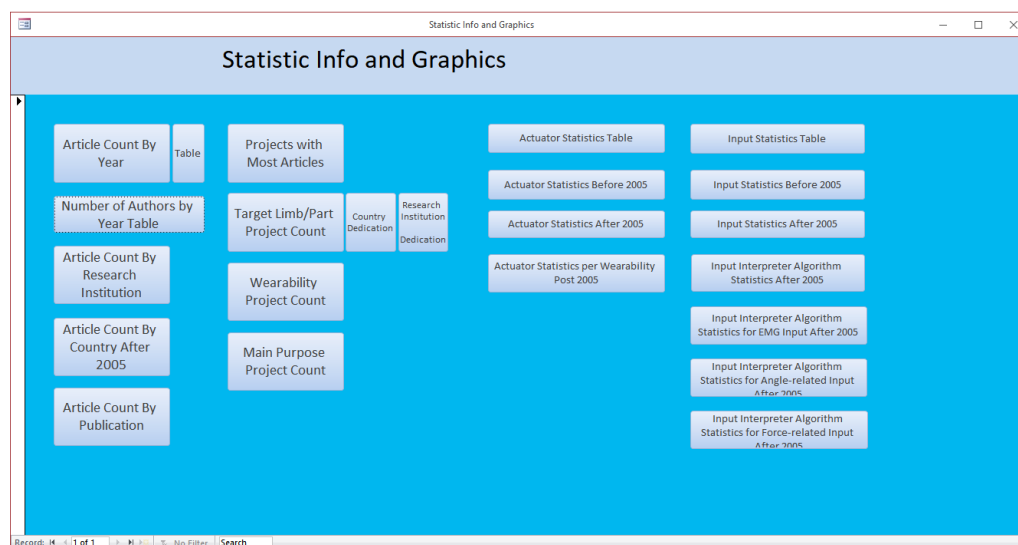
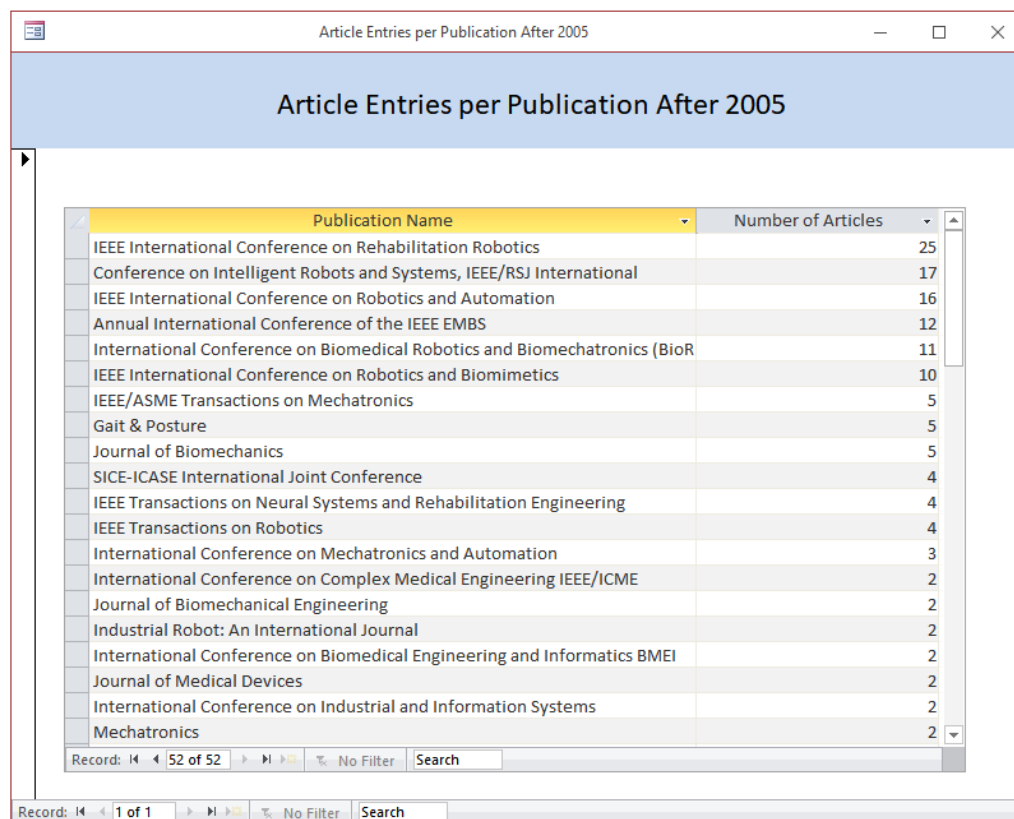


Figure 2.8: Screenshot of the database GUI page showing the type of data that can be immediately extracted from the database

This information is presented to the user in the form of a table that can be directly exported to an Office Excel or equivalent datasheet. An example can be seen in Figure 2.9.



Publication Name	Number of Articles
IEEE International Conference on Rehabilitation Robotics	25
Conference on Intelligent Robots and Systems, IEEE/RSJ International	17
IEEE International Conference on Robotics and Automation	16
Annual International Conference of the IEEE EMBS	12
International Conference on Biomedical Robotics and Biomechanics (BioR)	11
IEEE International Conference on Robotics and Biomimetics	10
IEEE/ASME Transactions on Mechatronics	5
Gait & Posture	5
Journal of Biomechanics	5
SICE-ICASE International Joint Conference	4
IEEE Transactions on Neural Systems and Rehabilitation Engineering	4
IEEE Transactions on Robotics	4
International Conference on Mechatronics and Automation	3
International Conference on Complex Medical Engineering IEEE/ICME	2
Journal of Biomechanical Engineering	2
Industrial Robot: An International Journal	2
International Conference on Biomedical Engineering and Informatics BMEI	2
Journal of Medical Devices	2
International Conference on Industrial and Information Systems	2
Mechatronics	2

Figure 2.9: Screenshot of the data displayed in the custom database's GUI. In this case, the number of articles per publication is shown.

It is the author's intent to update this database in the long term, in order to keep a relevant source of information on AMOADs.

2.2.2 Results: Localization and General Statistics

Regarding the orthosis'/exoskeleton's main purpose, this factor is rather challenging to identify in many of the articles. While some articles are very clear regarding the device's purpose [49, 50] others are often vague regarding this subject. Therefore, it is not possible to offer reliable statistics regarding each device's specific purpose.

Out of the 123 Projects, 89 refer their purpose as related to rehabilitation activities, from retraining motor movements to re-establishing or aiding the motion ability to target limbs due to injury or stroke. Strength enhancement is mentioned as main purpose for 14 Projects.

An immediate information that can be taken from the database is the general commitment from the scientific community regarding AMOADs. Figure 2.10 shows the number of articles gathered for each corresponding year of final draft and the number of related authors as well.

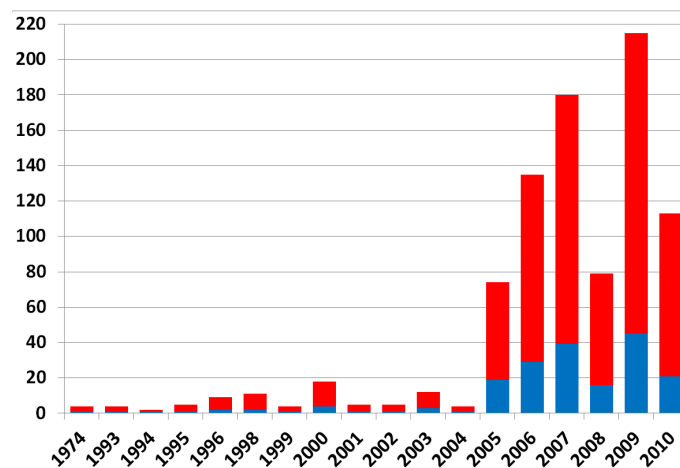


Figure 2.10: In blue, the number of articles related to active orthoses and exoskeletons gathered per year. In red, the number of authors associated to the gathered articles.

Given that 2005 is the year where the largest increase in scientific interest was observed, all the statistical studies presented in the article are, from this point, calculated using articles dated between 2005 and the end of the survey.

From the "Publication" table, it is possible to determine the publications with the largest amount of associated articles. This information can be of interest to a researcher looking to find a conference that is likely to have presentations on AMOADs. Only the results with 3 or more occurrences are represented in Figure 2.11:

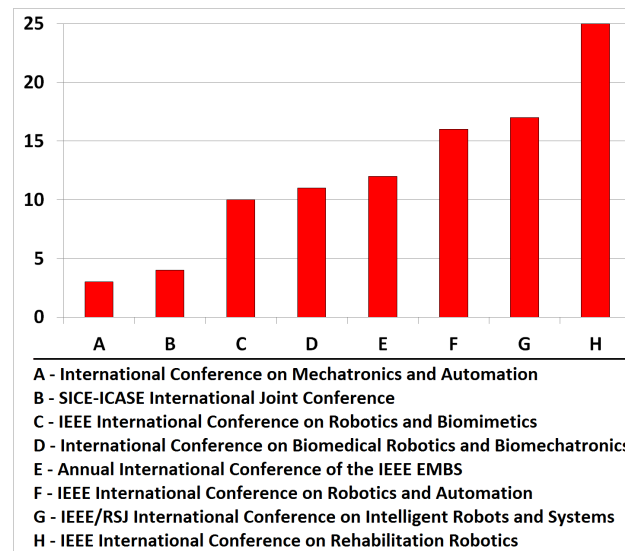


Figure 2.11: Number of articles associated to each Conference.

In order to identify where the research is located, depending on the target body parts, the "Research Institution" table was crossed with the "Target Limbs" table and target body parts, as

shown in Figure 2.12:

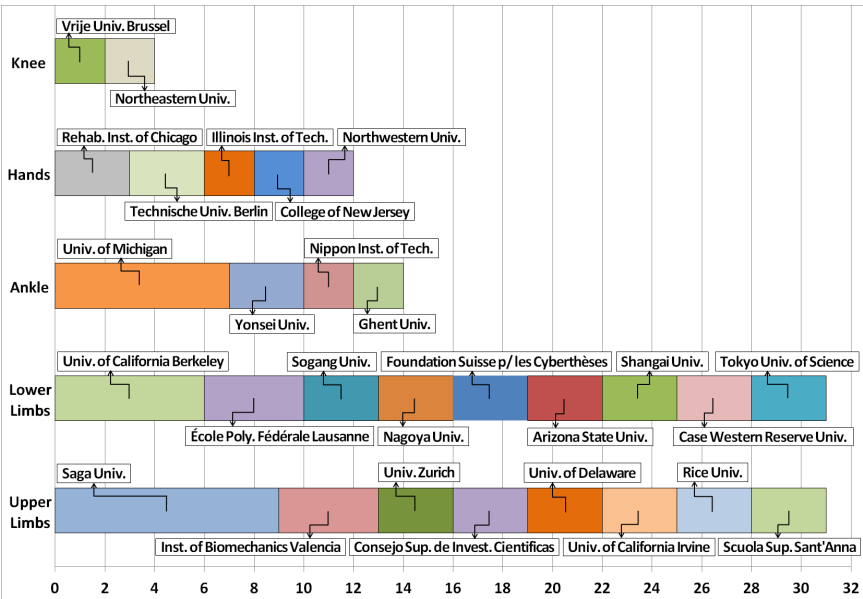


Figure 2.12: Number of articles associated to each Research Institution, divided by target limbs or body parts.

Figure 2.12 shows results with 2 or more occurrences. The "Lower Limbs" result represents projects that actuate and support both the ankles and the knees. The "Upper Limbs" represents projects that actuate and support the elbows and/or the shoulders.

Since the "Research Institution" table also contains information on the country where each institution is located, it is also possible to attain a comparison of research "dedication" between different countries, as shown in Figure 2.13.

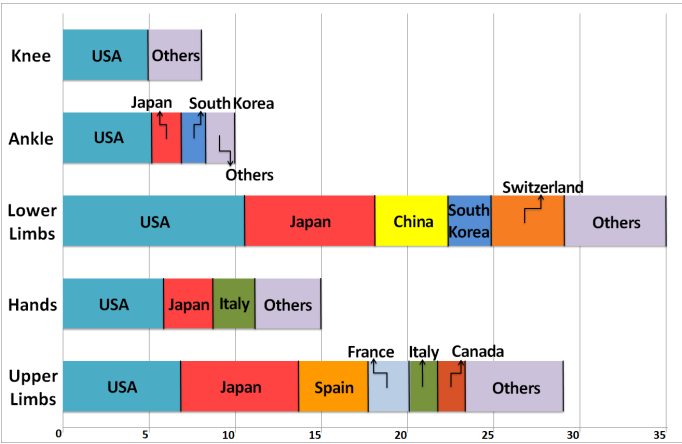


Figure 2.13: Proportions of articles associated to each country, divided by target limbs or body parts.

In Figure 2.13, countries with less than 4 associated articles were grouped into a single result called "Others".

This result shows the proportions of article associations between each country. An immediate result to be taken from the results above is that the research institutions from the United States of America have the largest amount of associated articles about AMOADs, followed by Japan.

Given the large difference in population between these countries, it can be assumed that Japan shows the highest dedication per capita. This result can be explained by the crescent ageing in Japan, which some of the projects tend to address.

2.2.3 Results: Actuation Technologies

As seen in Figure 2.5, the actuator statistics were taken from the "Projects" entry. Different actuation technologies may offer distinct advantages depending on the type of AMOAD. The DC Motor (servo-motor, stepper motor and others), is one of the oldest forms of actuation technologies used in active devices. Therefore, various and effective control methods are well studied. Also, pre-built motor controllers can be achieved from several manufacturers, rendering this technology easy to implement from a control standpoint. Some implementations use different kinds of gearboxes, increasing the smoothness and torque to the actuator.

However, other actuation technologies have a substantially larger torque-per-weight ratio [51], which should provide a practical advantage for weight-dependent, mobile orthoses and exoskeletons. This is the case of the McKibben actuators, despite the increased difficulty in developing control algorithms for a stable and safe operation [52].

For this reason, the "Actuator Type" table was crossed with the "Mobility" table (seen in Figure 2.5), in order to compare the same actuation technologies within 3 types of mobility:

1. Fixed , where the device is attached to a wall or a table, so the weight is less important [53, 54];
2. Mobile while sitting on a wheelchair, where the weight of the device is supported by a wheelchair [55, 56];
3. Mobile while walking, where the device allows (or is planned to allow, at some point in the prototype stage) the user to walk freely while attached. In this case, the torque-per-weight ratio holds the most importance since the user and/or the device itself have to support the increased weight [31, 40].

Figure 2.14 shows the results of crossing the aforementioned data, and it shows that the weight concerns, when projecting a mobile device, are driving the scientific community to experiment with other actuation technologies.

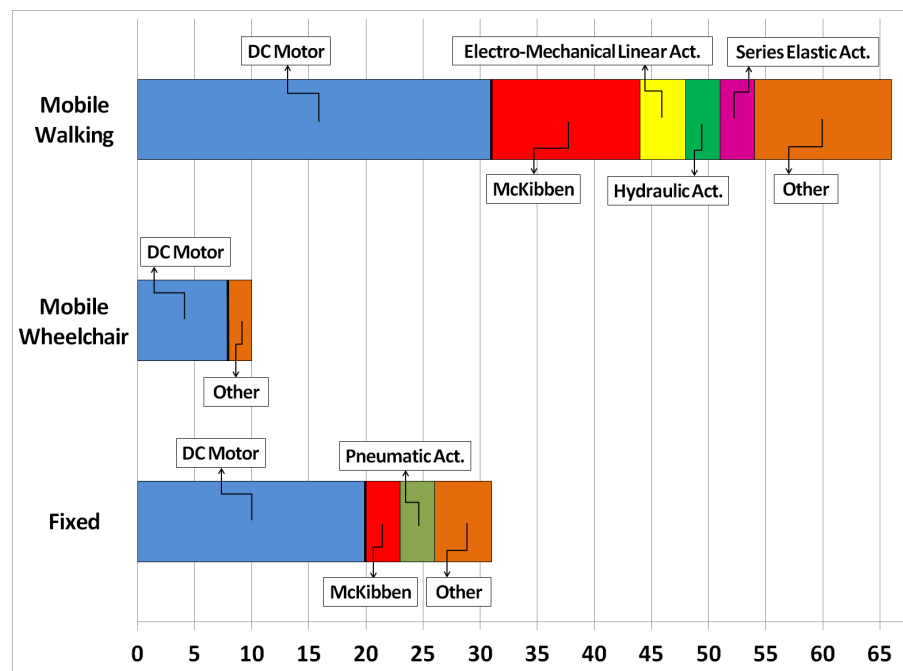


Figure 2.14: Number of orthosis/exoskeleton projects, grouped by the devices' mobility and showing the proportions between actuation technologies.

2.2.4 Results: Sensor Technologies

Figure 2.15 shows the number of occurrences of each sensor technology used in a device, throughout all analyzed articles. Within each article, all the sensor input types being used as an interface between the user and the device were gathered for these statistics.

The sensors exclusively used for internal actuator operation generally belong to a different part of the control loop and are more often documented within the scientific studies related to the actuation technology. As these studies may not relate directly to the development of active orthoses and exoskeletons, these sensors were not considered for the "Input Type" table.

This information, represented as "Input Type" in Figure 2.5, was taken from each article, as different articles in the same project may refer to different sensor input hardware [57, 58].

All the sensor types with less than three occurrences were gathered into a single group called "Other". "No Input" refers to devices using automated sequences for rehabilitation/training purposes, without having a direct connection between the user and the device [59, 60]. "Undisclosed Input" is the term used in the database when an article does not mention the sensor type used between the user and the device, which occurs in some articles dedicated to trial tests [61, 62].

Another useful information is to know which physical variables are preferred to be used in the control of active orthoses and exoskeletons. In order to retrieve this information, the sensor technologies were grouped into different classes, with each one determining the type of physical variables each sensor is measuring.

The list of gathered sensor types was divided into four physical variable classes:

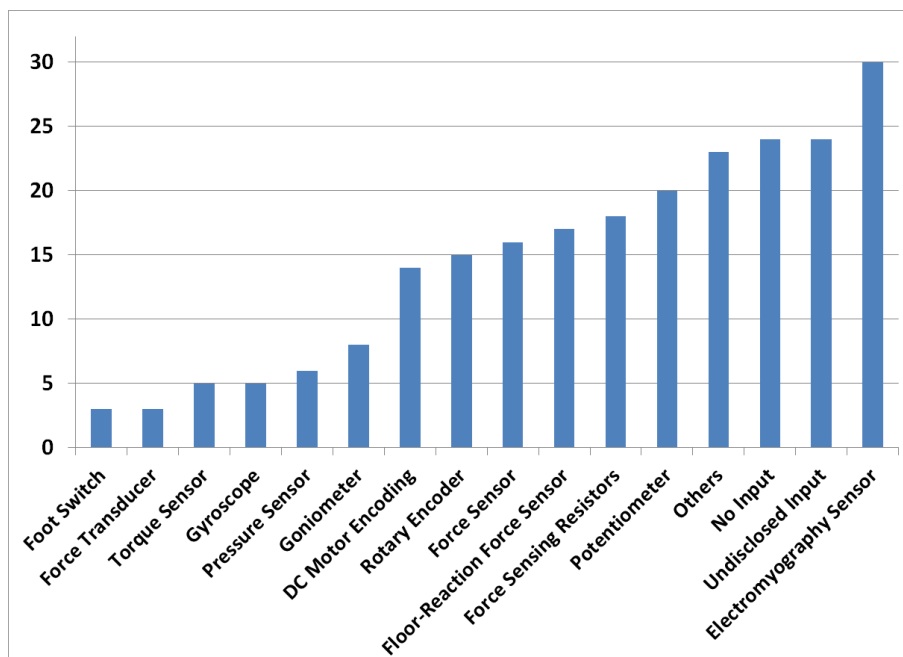


Figure 2.15: Number of occurrences for each sensor type, taken from the gathered articles

1. Electromyographic sensors,
2. Force-related sensors: consisted of torque sensors, force sensors, floor-reaction-force sensors, force-sensing-resistors and force transducers,
3. Angle-related sensors: consisted of potentiometers, rotary encoders, dc motor encoders, goniometers and gyroscopes,
4. Other sensors: consisted of less used sensors like accelerometers, on/off switches, optical tracking devices, among others.

Figure 2.16 shows the proportions between these sensor types. "Undisclosed Input" and "No Input" cases were not accounted for in the statistics represented.

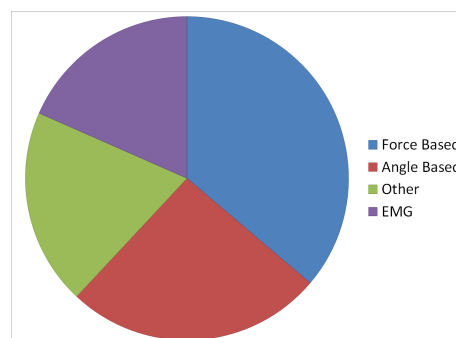


Figure 2.16: Proportion of sensor technologies, grouped by physical variable classes.

2.2.5 Database Results Discussions

United States of America is the country with the largest scientific dedication to AMOADs, followed by Japan. Within these results, Japan seems to be the country with the largest dedication per capita. An investment in AMOADs may be a reaction to the country's issues with ageing.

The actuator statistics show that the most popular actuator for active orthoses and exoskeletons is the DC Motor so far. Although this actuator presents a relatively low torque/weight and torque/volume ratio, its matured control methods makes it the most widely adopted choice.

However, when studying a mobile device with weight and volume constraints, as we see in Figure 2.14, the DC Motor is seen to be adopted in less than half of the studied projects. As actuation technologies with higher torque/weight ratios become mature, it may be expected to see DC Motors being gradually replaced for wearable devices, particularly by McKibben actuators.

However, at the time of writing of this article, there is not a definitive actuation technology to use in all mobility case. While having a higher torque/weight ratio, the McKibben actuators (as well as other kinds of fluidic/pneumatic actuators) still need a dedicated compressor and batteries to operate, which also raises the matter of acoustic noise. When taken into consideration the power needed for the dedicated compressor, the overall power efficiency of the McKibben actuators and other pneumatic solutions is also reduced.

Regarding the statistics relative to the sensors used, the EMG sensors appear as the most popular type of sensor. When grouping the sensors by "physical variable" classes, it can be seen that although force-related sensors are actually being tried out in larger proportions, the magnitude of occurrences is similar for the three main classes.

2.3 Actuators in Robotics for Human Interaction

An actuation system designed to function close to the human body must abide to many limitations. The torque/weight (N.m/kg) and force/volume ratios are determining factors. An overly heavy system would quickly fatigue the patient, whereas a bulky system would hinder the patient's movements.

Energy consumption is a major concern. The more energy the system needs, more weight and volume will be necessary for the power supply and other components. Acoustic noise (for example, from an air compressor) is also a limiting factor. A noisy system would disturb the user and his/her environment. Depending on the type of application, the precision and controllability provided by the actuation system should also determine if it's appropriate or not. However, the most important aspect for an actuation system working close to a human is overall safety. The system must have mechanical and electronic safety measures in order not to hurt or further encumber the patient, there must be no leakage of corrosive or other harmful materials and there can be virtually no risk of explosion or overheating of the actuator, batteries and power electronics.

These factors are studied for a range of practical examples in this section.

2.3.1 DC Motors

DC Motors can be divided into two main types: Brushed and Brushless.

In the Brushed DC motors the current commutation is made through a mechanical contact. These motors exist in large quantities and are inexpensive to produce.

However, the commutator mechanisms, the brushes, are under constant friction during its operation. As a consequence, these motors have low longevity and reliability, compared to the brushless counterparts.

The Brushless DC Motors have an outer component with permanent magnets which move through interaction with the electromagnets in the armature. The current in the armature windings is switched electronically, through a solid-state circuit. This allows for a more durable and reliable operation. Since there is no mechanical friction through the commutation system, these motors are also more power efficient and produce less noise, among other advantages.

For an usage in orthoses and prostheses, the building characteristics of DC motors represent a difficult dilemma. The energy conversion is made in the actuator itself, so it would require a shield in order to avoid direct contact with body parts.

From all the actuators studied, the DC Motors provide the weakest torque/weight relations. An example of a recent small and light-weight Brushless DC Motor has an Energy-to-Mass ratio of 23 N.m/kg [63]. In DC Motors this ratio can be changed using reduction gearboxes, at the cost of increased weight and consequently a lower Energy-to-Mass ratio.

Although the DC Motors can provide a wide rotary range of movement (the rotor can rotate virtually endlessly), this feature is of little importance for prostheses and orthoses because the human limbs do not require such a movement.

2.3.2 Electroactive Polymers

The Electro-Active Polymers (EAPs) are, as the name implies, polymers that change shape or volume when submitted to a voltage. Polymers reacting physically to a voltage only started being studied as actuators by the end of the 1980s [4].

In the beginning of the 1990s, new EAP materials allowed for higher strains, creating a bigger focus in the scientific community into this actuation technology.

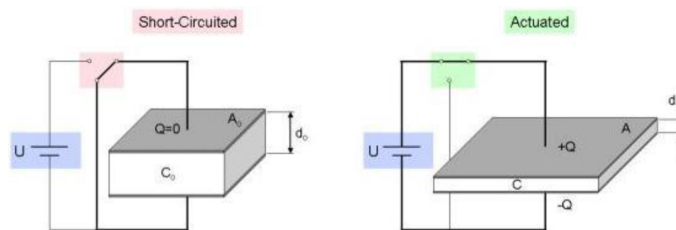


Figure 2.17: Basic Functioning Principle of a Dielectric Electro-Active Polymer [4]

There are two types of EAPs: the Electronic EAPs and the Ionic EAPs [64]. The Electronic EAPs change shape through an electric field whereas the Ionic EAPs actuate due to the mobility or diffusion of Ions. The Ionic EAPs are generally capable of larger forces than the Electronic EAPs.

The Dielectric EAPs, from the group of Electronic EAPs, need very high voltages (in the order of thousands of volts) to change shape. However, since no energy must be spent in order to maintain a certain shape, the energy consumption of an actuator based in a Dielectric EAP is very low. Also, the behavior of a Dielectric EAP is similar to a skeletal muscle.

For these reasons, Dielectric EAPs in the shape of springrolls (seen in Figure 2.18) are being thoroughly studied in order to create artificial muscles for prosthetics [65].

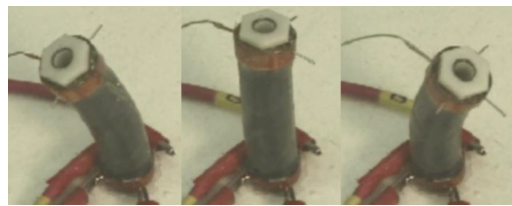


Figure 2.18: Pictures of a working Spring-Roll shaped EAP actuator [5]

Although the low power requirements and shapeshifting abilities of the EAPs make them sound more elegant than the alternatives, the high-voltages required for activation pose a problem related to the patient's general safety (due to risk of electric shocks), as well as having large voltage converters in a mobile appliance.

Also, the strength provided by these actuators is very small compared to other alternatives [66].

2.3.3 Shape Memory Alloys

Shape Memory Alloys (SMAs) are metallic alloys with the ability to change their shape when faced with a change in temperature. According to Kazuhiro Otsuka et al [67], the first observation of this behavior occurred in 1951, with a Gold-Cadmium alloy. In the present day, the most frequent SMA used is the Nickel-Titanium alloy (Ni-Ti, also known as "Nitinol") due to inexpensive manufacturing and relatively large strains (chapter 4.2.2 in [68]).

Actuators based in SMAs have an actuation strength of two orders of magnitude larger than EAP actuators [66] and the Ni-Ti alloy is considered a highly biocompatible material (can be implanted in an animal without harming the host). Therefore, these materials have been thoroughly used in medical instruments and orthopaedic implants such as laparoscopy tools, orthopaedic staples and bone implants [67], as seen in Figure 2.19.

However, using heat as an actuation trigger causes many limitations for the SMAs as actuators in active orthoses: the energy efficiency is very low (around 1% in practical uses), the actuation and activation speeds are slow and the lifetime can be significantly reduced when the actuator is bound to excessive strain and/or stress [67].

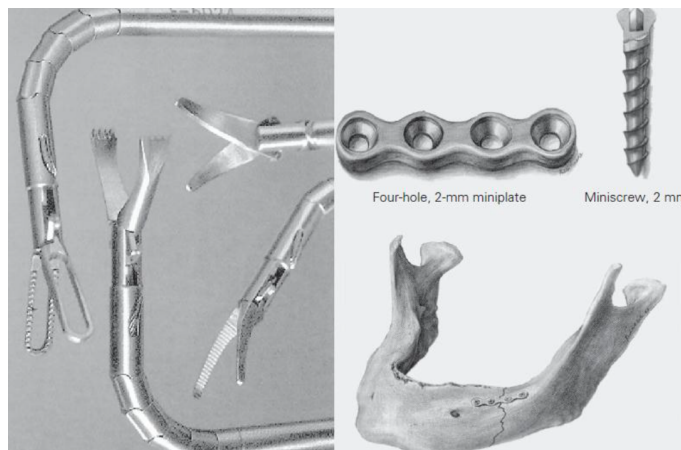


Figure 2.19: Examples of medical appliances using SMAs. On the left are grippers, scissors and tongs for laparoscopies; on the right, SMA bone plates fixed in a jaw. [6]

2.3.4 McKibben Actuator

The first concept of a fluid-driven actuator appeared in 1930 by the hands of a Russian inventor called S. Garasiev. In the 1950s, J.L. McKibben applied these actuators in active prostheses, hence the naming convention [69].

They were first commercialized in the 1980s by the Japanese company Bridgestone Corporation. The McKibben Actuator can also be called Pneumatic Artificial Muscle or Braided Muscle, among other names.

The McKibben Actuator consists of an internal rubber bladder surrounded by a braided sleeve, both of which are tightly fixed to both ends. When the internal bladder is inflated, the external braided sleeve forces the bladder to expand sideways, approaching both ends to one another, as seen in Figure 2.20. Like the EAP's, this actuation can be considered as biomimetic (a method applied in a synthetic system that was inspired by nature) because of its resemblance with the skeletal muscles.

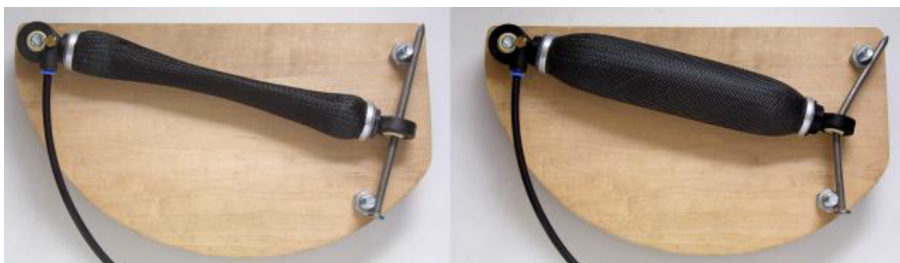


Figure 2.20: Example of a working McKibben actuator. On the left, the actuator is deflated. On the right, the actuator is inflated, bending the nail. [7]

These actuators are known for being light-weight and having a very high Energy-to-Mass ratio. As an example, the "Festo MAS-10-N Fluid Muscle" has a maximum Energy-to-Mass ratio

of 687,5 N.m/kg [51]. As long as the gas used for inflating the internal bladder is innocuous (in most applications, air is used), the McKibben actuators can be considered safe, without risk of fire or explosion. They can be found for sale for industrial applications by companies such as Festo [70] and Shadow Robotics [71]. Because these actuators have been in the market for more than 20 years, many robotic systems have been developed to take advantage of their biomimetic actuation.

The largest drawback of using McKibben actuators lies in the fact that it can only exert a compressive force between two ends and its control precision. The force of these actuators depends not only on the pressure inside the internal bladder, but also on the inflation level. There is also friction created between the internal bladder and the surrounding braided sleeve, which causes a hysteresis between inflation and deflation. Therefore, the actuation is nonlinear and hard to predict. For this reason, many control methods are now being studied in order to increase the usability of these actuators [69]. Moreover, these actuators also require a source of pressurized gas (pump or canister).

2.3.5 Flexible Fluidic Actuators

The Flexible Fluidic Actuator (FFA) was invented by Dr.-Ing. Stefan Schulz et al, in 1999 [8]. The functioning principle of the FFA is based on a very simple model, with two planar plastic films connected through a joint axis, with an elastic chamber in between. By inflating the elastic chamber, the two plastic films would separate, creating a rotary force in each front, as seen in Figure 2.21:

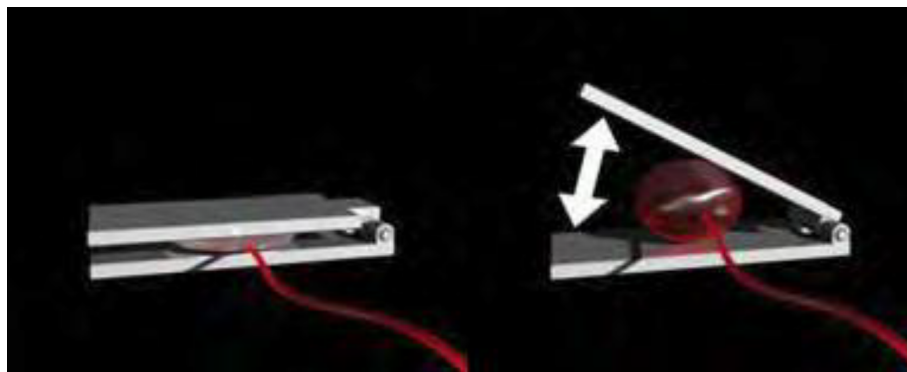


Figure 2.21: Basic functioning of the Flexible Fluidic Actuator. On the left, the elastic chamber is deflated. On the right, the elastic chamber is inflated, creating a rotary motion between the plastic films [8].

The FFA was first thought as a solution to find an actuation method that could be small enough to fit an artificial finger joint. The current FFAs, although preserving the same basic functioning as the model mentioned above, have a drastically different shape. The simple elastic chamber was replaced with an accordion-shaped bellows and the ends of the actuator are now separated, with the joint axis being mechanically connected to the ends, according to the application. These updated models can be observed in Figure 2.22:

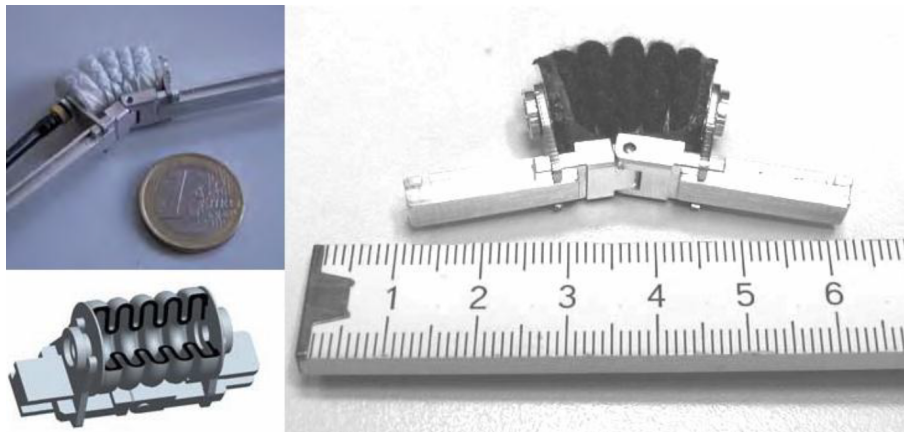


Figure 2.22: Examples of current Flexible Fluidic Actuator [9].

The FFAs can use gases and liquids as expansion medium. Like the McKibben actuator, the FFA's binary depends on the internal pressure and the inflation level which makes a precise control approach rather difficult. Depending on the inflation level (which translates directly into the rotation axis), the FFA has an Energy-to-Mass ration between 148 and 740 N.m/kg, using air as expansion medium [63].

The actuation motion of the FFAs can be considered non-biomimetic. While the skeletal muscle applies a "pull" force between two extremities, the FFAs apply a rotary "push" force.

However, their ability of keeping a high Energy-to-Mass ratio and functionality, even in very small form-factors, makes them suitable for developing light-weight and unobtrusive active orthoses.

These actuators also require the presence of a source of pressurized fluid to function, so a canister and/or a pump would be needed for continuous operation.

The research for smaller and stronger FFAs (supporting higher pressure levels) is supported by a parallel research for smaller and more power-efficient pressure pumps and other fluidic components [9], in order to apply the technology in portable systems.

2.3.6 Pneumatic Cylinders and Hydraulic Actuators

The Pneumatic Cylinders can be of single-acting and double acting [51]. The single-acting pneumatic cylinders use the air pressure in their interior to change the distance between its two edges, and have a spring in its interior that make them come back to the initial position. The double-acting cylinders consist of two pneumatic chambers at different pressures and have a valve system that controls the pressure in each cylinder as well as the speed and direction of air flow. Therefore, the linear movement of these actuators depends on the pressure on each chamber. The valves are generally controlled through DC motors.

Given the natural expandability and compressibility of the air and other gases, it is difficult to achieve a precise control using Pneumatic Cylinders.

The Hydraulic Cylinders convert hydraulic pressure energy into mechanical energy using oil as means of transmission. Their operation is similar to the pneumatic actuators.

The hydraulic actuators can be divided into hydraulic cylinders or hydraulic angular actuators. These have some advantages over pneumatic actuators, as they allow working at higher pressures so small actuators can provide large forces. The non-corrosive oil used in these actuators isn't as expansive/compressible as the gases used in pneumatic actuators, allowing a precise control over the mechanical components. Although they are generally safe to use, these actuators are also very expensive.

2.4 Conclusions

During the last few years, the global scientific community has shown a large increase in the overall interest around these devices. This phenomenon can be observed through the sudden increase of published scientific articles starting in 2005, as well as the increasing number of associated authors.

Regarding the bibliographic research task on the subject of active orthoses and exoskeletons, the first and foremost fact to point out is the lack of a coherent taxonomy adopted by the scientific community.

Nowadays, the same terms can be used to describe both, a medical device for rehabilitation purposes and a general strength enhancement device, difficulting the process of using regular scientific search engines for a bibliographic research on either type of devices. For example, an article describing an active orthosis designed for rehabilitation [72] may also be described as a "rehabilitation robot" [73], a "powered exoskeleton" [74] or a combination of the aforementioned terms [9]. A few authors also use the general term "assistive device" [75] to describe them, which can also be used to mention a cane or a pair of crutches.

For this reason, a thorough and relevant bibliographic search on active orthoses is hard to obtain. This factor may result in the creation of redundant research projects, subsequently delaying the technological evolution of these devices and increasing the associated costs for their implementation.

As for future trends, it is possible that different forms of actuators will be increasingly explored [76]. Advances in the field of Electroactive Polymers [77] may allow the usage of these actuators in AMOADs because of the variety of shapes that they can assume. Regarding mobile AMOADs, their developers will probably follow a trend where the design will tend to be unobtrusive and light, after most of the functional aspects are successfully implemented.

Table 2.1 shows a qualitative comparison between the technologies mentioned in this section [78].

Table 2.1: Comparison between Actuator Technologies

Actuator	Energy-to-Mass Ratio	Main Advantages	Main Disadvantages
DC Motor	23 N.m/kg	Price, availability and wide knowledge about control methods	Low Energy-to-Mass Ratio
Dielectric EAP	-	Very Low Power Consumption	High Voltage Required
Shaped Memory Alloys	-	Very High actuation force	Low energy efficiency, slow
McKibben	687 N.m/kg	High Energy-to-Mass ratio, availability	Difficult Control due to hysteresis, actuation on a single direction
Flexible Fluidic Actuators	Up to 740 N.m/kg	High Energy-to-Mass Ratio, effective in small form-factors	Force Decreases with flexion angle, actuation on a single direction
Pneumatic	100 N.m/kg	Light	Doesn't allow large forces because it doesn't support high pressures, actuation on a single direction
Hydraulic	150 N.m/kg	Small Actuators Provide Large Forces	High Price, actuation on a single direction

For this project, the actuators chosen are Brushless DC Motors. As mentioned in Chapter 1, this project is driven by its feasibility within cost limitations. Despite the lower Energy-to-Mass ratio, the DC Motors are inexpensive, they don't require an additional apparatus for fluid intake and the control methods are well studied.

Chapter 3

Mechanical Design for the Support Frame

3.1 Design Goals

3.1.1 Active and Passive Movements

The human gait cycle can be divided into two phases: Stance and Swing, as seen in Figure 3.1. During the Stance phase between 0% and 60% of the cycle, the body is moved forward when assisted by the front leg's hip extension. From 60% to 100% of the gait cycle, the Swing phase consists of moving the same leg to the front when assisted by hip flexion, which also moves forward the body's center of mass.

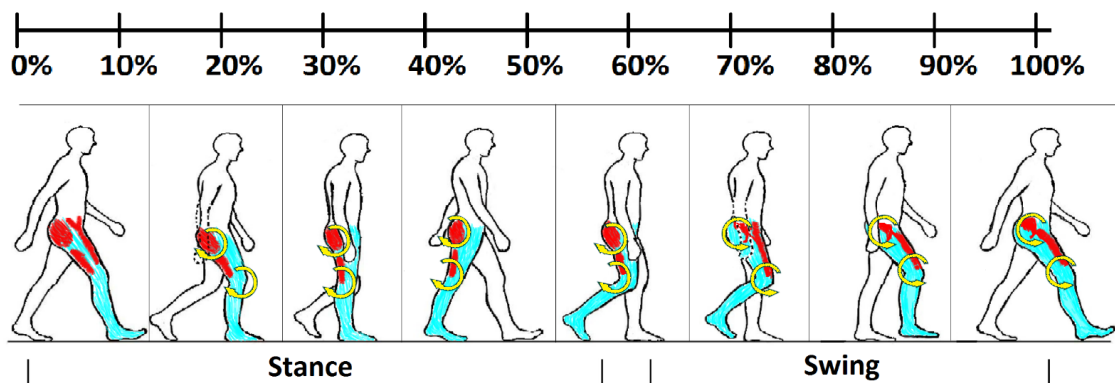


Figure 3.1: Diagram representing the gait cycle for the right lower limb (highlighted in blue) from a healthy human being. The muscle groups that are generating work for the hip extension/flexion and knee flexion/extension during the Stance and Swing phases, can be seen highlighted in red. The direction of motion of each joint in the sagittal plane is represented by the yellow arrows. These movements are actively assisted in the proposed system.

The prototype of the active lower limb exoskeleton is designed to feature actuators that assist hip extension/flexion and knee extension/flexion. Since both pairs of movements are antagonistic,

they can both be performed by two actuators for each leg if these are capable of generating torque in two directions, such as DC motors. The knee joint uses a regular ball bearing, and the hip and ankle joints use spherical bearings to provide a passive degree of freedom for rotation and abduction/adduction for the hip, and dorsiflexion, plantar flexion, eversion/Inversion and axial rotation for the ankle.

At the moment of heel contact (0% in Figure 3.1), most muscles in the lower limb are flexed to provide stability. During the Stance phase (0 to 60% in Figure 3.1), the exoskeleton assists the work generated by the Gluteus group of muscles for hip extension, and the Hamstrings group for knee flexion and hip flexion. During the Swing phase (60% to 70%), the exoskeleton assists the Iliopsoas group and Tensor Fasciae Latae muscles for hip flexion, and the Quadriceps group for knee extension and hip flexion [79]. These are muscles that suffer from atrophy in elders [80].

Due to the symmetry of the biomechanics between left and right legs during a healthy gait, the exoskeleton is also symmetric.

Through these movements, the exoskeleton is also potentially able to assist its wearer to roam through non-levelled terrain such as climbing stairs. The actuators and batteries are expected to be positioned in the lower back and will perform a push/pull force directly into the mechanical frame through bowden cables. At this stage, an added weight of 10kg is projected for the actuators and respective batteries.

Although the ankle movements are also commonly weak among elders, at this stage the exoskeleton is projected to assist hip flexion/extension and knee flexion/extension, because these movements involve higher torques during gait [81], and they are both essential for sitting/lifting up and climbing stairs. Therefore, assisting these movements may save more energy thus providing greater stamina to an elder user.

3.1.2 Materials and Components

The mechanical support for the active exoskeleton was designed using Solidworks 2016 from Dassault Systems, and the software's finite-element analysis module was used for simulating static loads.

The MakeHuman Open Source software [82] was used to produce anatomically correct 3D human models. This software uses algorithms taken from several studies regarding the body surface area of human subjects [83, 84, 85], allowing to create human models based on several user-defined properties such as gender, age, weight and height, as seen in Figure 3.2.

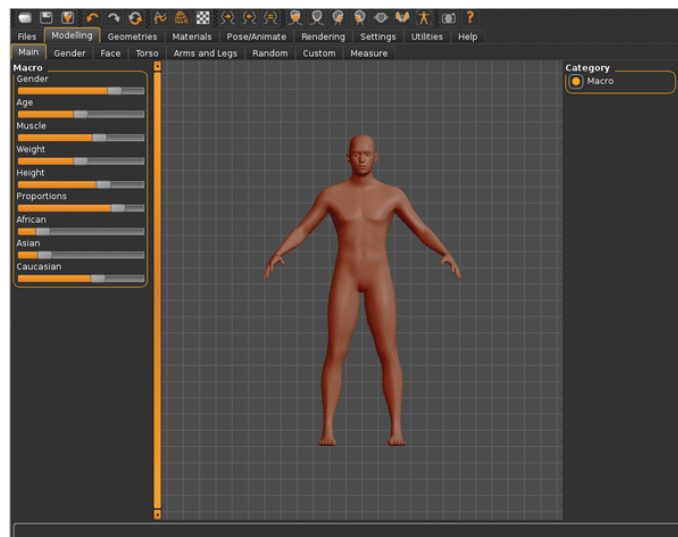


Figure 3.2: Screenshot of the MakeHuman interface to create anatomically correct 3D models of human bodies.

Considering the objectives of this project, several models were created with ages between 70 and 80 years old, heights between 1.50 m and 1.83 m and variable weight levels. Figure 3.3 shows the models created.

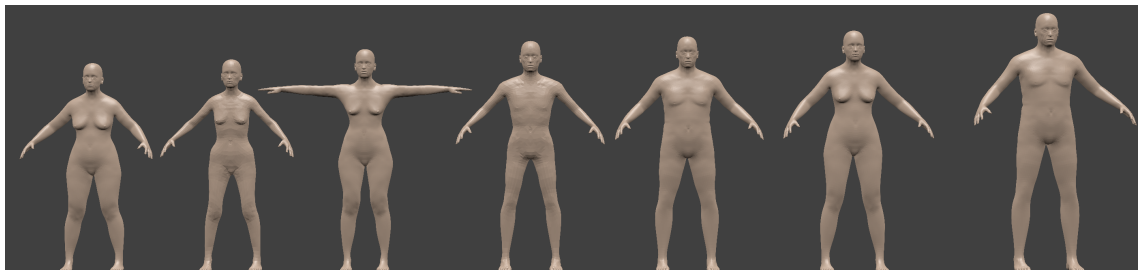


Figure 3.3: Anatomically correct 3D human models created to develop the mechanical frame.

The material chosen for designing the mechanical frame is Aluminium 6061 T6 with a typical yield strength of 275 MPa. The fatigue limit under cyclic load is 97 MPa for 500 million reversed cycles. The aluminium alloys are typically easy to machine in CNC have a lower weight than other metals. This specific alloy was chosen for its relatively high yield strength together with a low cost. The higher yield strength the material has, the smaller the frame can be. The material selection took cost and accessibility into consideration to produce a functional prototype at a later stage of the project. For a functional prototype, each component would need to be machined using CNC. Although the cost for CNC in general is high, the chosen material has a low cost and can be easily and rapidly machined compared to other metals.

Although composite materials (such as carbon fiber) could be studied to reach a smaller volume, the raw material for Aluminium is significantly cheaper in present day. Also, during a

possible production stage, using Aluminium allows to use mold injection which would make the mechanical frame cheaper for mass production.

3.2 Design Cycle

The exoskeleton's mechanical frame was designed with the following objectives:

- Be able to support itself, in order to not encumber its user;
- Avoid direct contact with zones close to the bone, use zones with muscles as “cushion” in order to avoid injuries;
- Low volume and weight;
- Avoid tight protrusions and indentations, so the components for early prototypes can be produced through fast prototyping;
- Avoid expanding the user's “personal area”, in order to keep a slim profile and avoid accidents like the user unintentionally bumping into objects or other people;
- Allow the user to have, to a certain extent, all degrees of freedom that are present in the hip, knee and ankle joints. This allows the user to perform a more natural gait and to be able to traverse different types of terrains;
- Modular approach with several components that can connect through variable distances. This allows the same exoskeleton to be adaptable to several body types;

The exoskeleton's mechanical frame design went through a cyclical development in order to reach an optimal form, as shown in Figure [3.4](#).

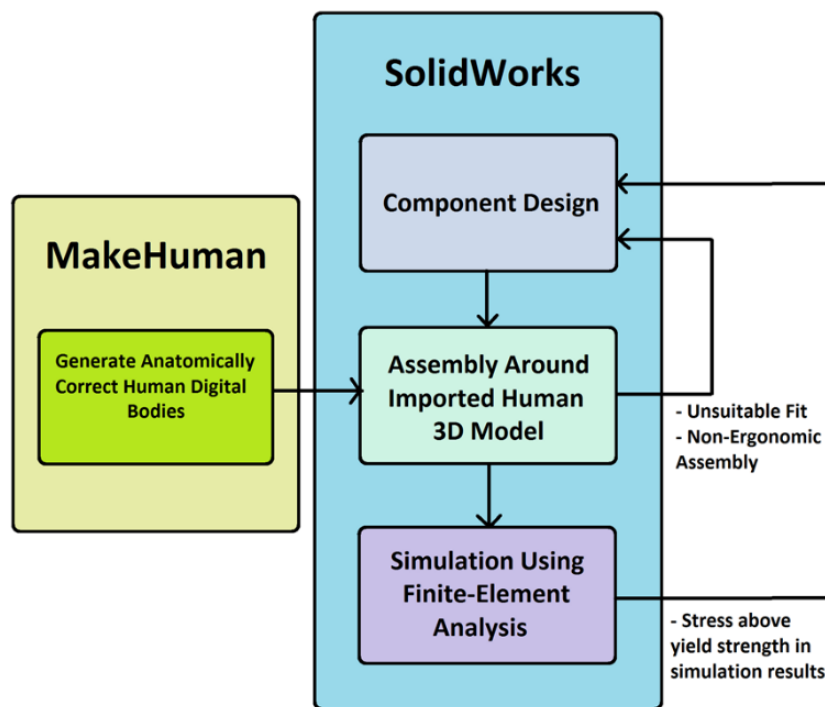


Figure 3.4: Cyclical development diagram for the mechanical frame design of the active exoskeleton.

The several components were designed through Solidworks. The human models represented in Figure 3.3 were converted into meshes and the components were gradually assembled around the imported models. With this procedure, the assembled components were guaranteed to fit closely to several human models, while also following the objectives mentioned above. Through this cycle, the mechanical frame can be used by people with different body measurements. An example of an adaptability study can be observed in Figure 3.5:

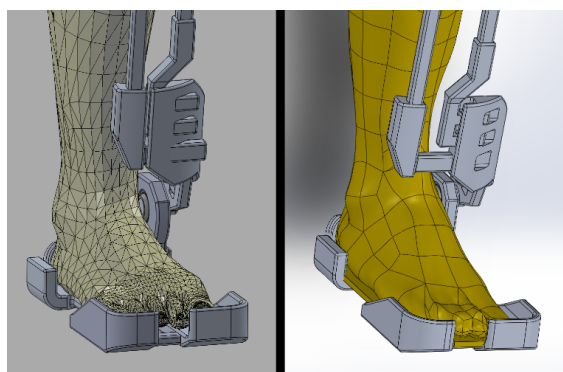


Figure 3.5: Exoskeleton adaptable to a smaller female body to the left and a larger male body to the right, in the foot and ankle areas.

3.3 Component Description

This section provides a short overview for each component that was designed for the mechanical frame. As mentioned before, the components connect through variable distances or angles to be adaptable to different human bodies. An explanation is provided regarding how the components connect between each other and the biomechanical, anatomical and ergonomic motives for each connection. The anatomical planes shown in Figure 3.6 are used to explain the connections.

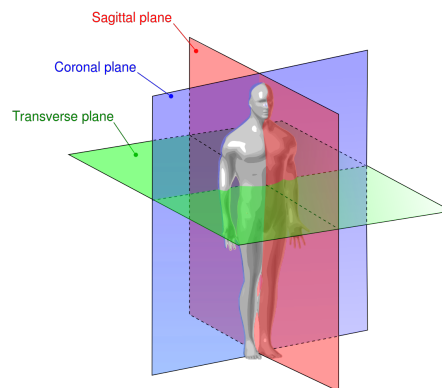


Figure 3.6: Planes of human anatomy. Sagittal plane in red, Transverse plane in green and Coronal plane in blue.

The system is composed of 25 distinct components, 13 of which are exclusive to each side and 12 can be used on either side. Figure 3.7 shows a render of the mechanical frame and the groups of components that are described in each following section.



Figure 3.7: Render of the full exoskeleton frame. The components are divided into 3 sections: Belt to Hip in section 3.3.1, Hip to Knee in section 3.3.2 and Knee to Heel in section in 3.3.3

The mechanical frame is symmetrical from the sagittal plane. Therefore, the components of the right side (left side of Figure 3.7 are described in this section, as the symmetrical counterparts of each component will have an identical purpose and functionality. Section 3.5 portrays other renders of the mechanical frame with other angles and human models.

The variable distances mentioned in this section are meant to be fixed after taking anatomic measurements from the exoskeleton's user. The type of bonding (e.g. direct screwing or adhesives) to apply between components after determining these distances is not studied within the scope of this thesis.

3.3.1 Belt to Hip Section

3.3.1.1 Hip-Fit

The component "Hip-Fit" shown in Figure 3.8 is parallel to the Sagittal plane and stands close to the hip joint. It is designed to house a spherical bearing from Schaeffler and connect to the "Belt-Hip-B" component through a variable angle within the Coronal plane.

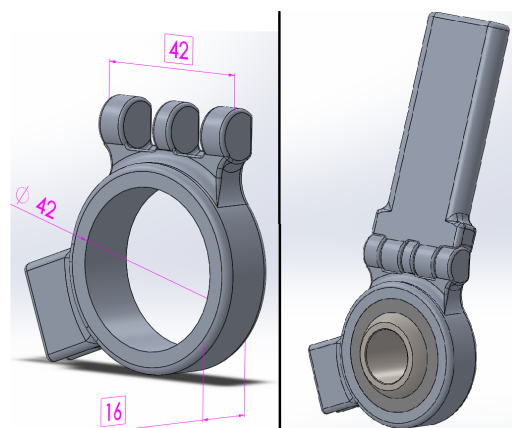


Figure 3.8: On the left, an image of component "Hip-Fit" with some measurements in millimeters. On the right, component "Hip-Fit" housing a spherical bearing and connected to component "Belt-Hip-B".

The variable angle between the components "Hip-Fit" and "Belt-Hip-B" has the purpose of adapting to the body area between the hip joint and the waist line. This parameter is usually different between males and females, as seen in Figure 3.9.

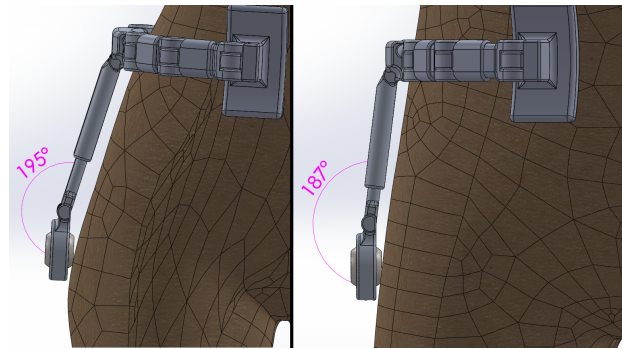


Figure 3.9: On the left, the components "Hip-Fit" and "Belt-Hip-B" are adapted to the waistline and hip joint of a female model. On the right, the same components are adapted to a male model.

3.3.1.2 Belt-Hip

The components "Belt-Hip-A" and "Belt-Hip-B" seen in Figure 3.10 serve the purpose of connecting the hip joint components to the waist area.

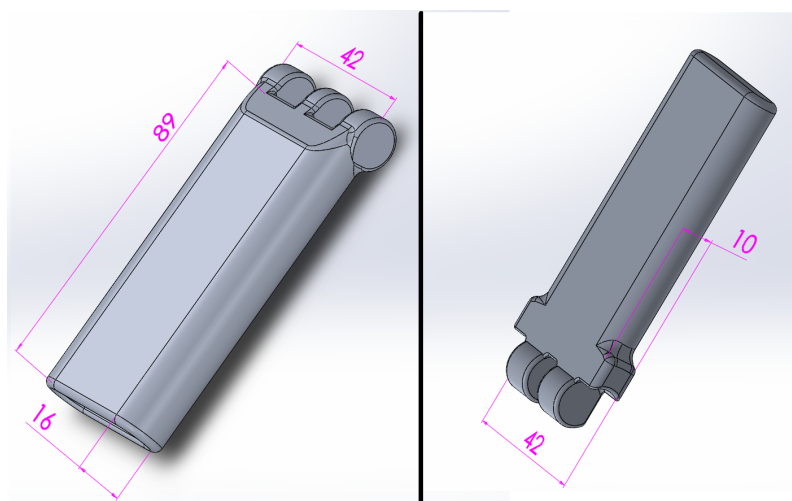


Figure 3.10: On the left, component "Belt-Hip-A". On the right, component "Belt-Hip-B". Some measurements are shown in millimeters

Components "Belt-Hip-A" and "Belt-Hip-B" have a socket connection, in a variable distance parallel to the Coronal plane. This distance allows the mechanical frame to adapt different distances between the hip joint and waist, as shown in Figure 3.11.

Component "Belt-Hip-A" connects to component "Belt-Distribute" mentioned in 3.3.1.3. The connection between "Belt-Hip-A" has a variable angle, though the angle is the same as the connection between "Hip-Fit" and "Belt-Hip-B" shown in 3.3.1.1, in the inverse direction. This way, component "Belt-Distribute" and the components that connect to it are kept parallel to the Transverse plane.

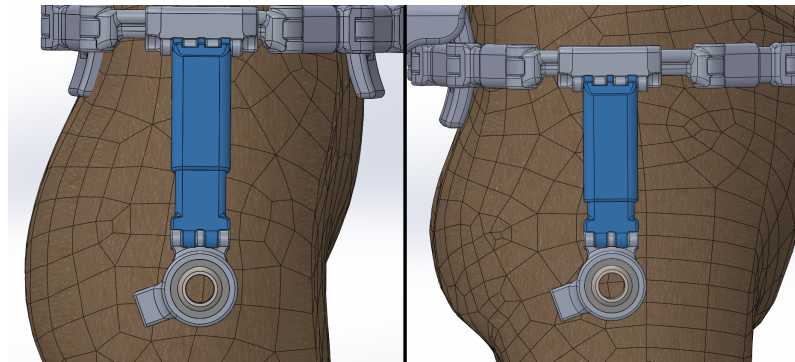


Figure 3.11: Components "Belt-Hip-A" and "Belt-Hip-B" adapted to models with different distances between the hip joint and the waist.

3.3.1.3 Belt-Distribute

Component "Belt-Distribute" is used as a hub to connect the waist components and the components leading to the hip section. It connects to "Belt-Hip-A" and to two smaller symmetrical components called "Belt-Extension-A" and "Belt-Extension-B". "Belt-Distribute" and "Belt-Extension-A" are shown in Figure 3.12.

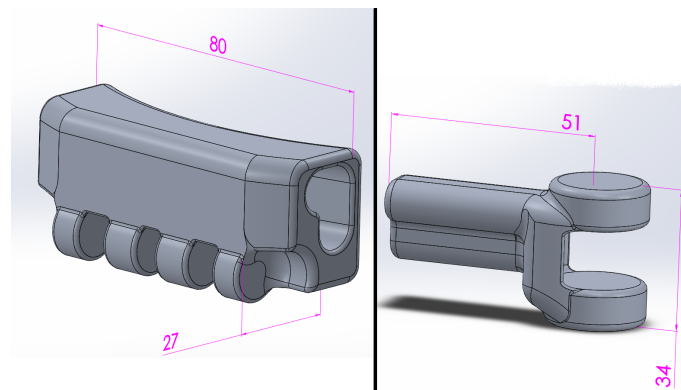


Figure 3.12: Component "Belt-Distribute" on the left and component "Belt-Extension-A" on the right. Some measurements are shown in millimeters.

Components "Belt-Distribution" and "Belt-Extension-A/B" have a sleeve fitting connection with variable distance, which is parallel to the Transverse plane and have a 12° angle to the Sagittal plane. The purpose of these variable connections is to better adapt to different types of waist, namely with volume distribution towards the abdominal or lombar regions. Different fitting distances are shown in Figure 3.13.

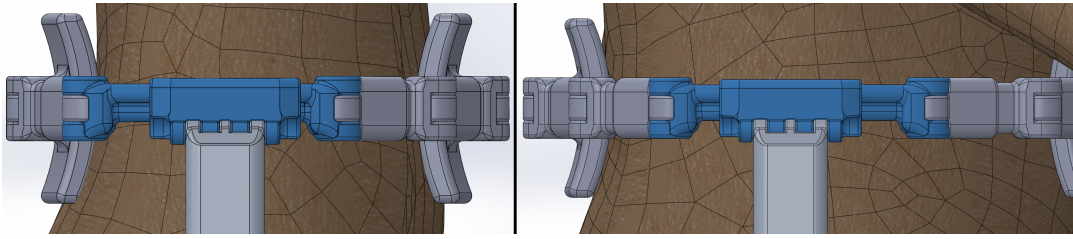


Figure 3.13: Components "Belt-Distribute", "Belt-Extension-A" and "Belt-Extension-B" adapted to two different examples. To the left, a model with more volume towards the lumbar region. To the right, a model with more volume towards the abdominal region.

3.3.1.4 Waist-Extension and Waist-Contact

Components "Waist-Extension-A" and "Waist-Extension-B" serve the purpose of surrounding the waist of the user. At the end of these components, the component "Waist-Contact" has a large area of contact to provide an interface with either the front abdominal region or the back lumbar region. These components are shown in Figure 3.14.

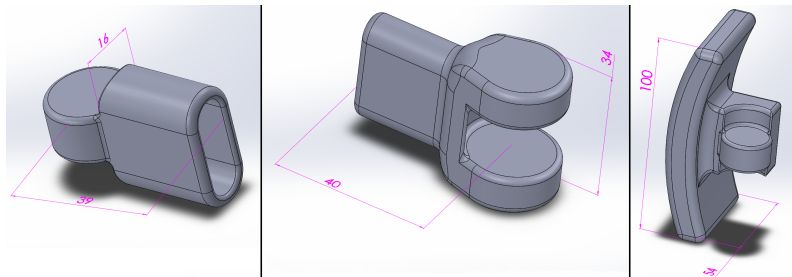


Figure 3.14: From left to right, components "Waist-Extension-A", "Waist-Extension-B" and "Waist-Contact". Some measurements are shown in milimeters.

The component "Waist-Extension-A" connects to "Belt-Extension-A/B" through a variable angle, and to "Waist-Extension-B" through a variable distance. Then, component "Waist-Extension-B" can either connect to the "Waist-Contact" component or to another pair of "Waist-Extension-A/B", through a variable angle parallel with the Transverse plane. The variable angles, distances and ability to chain more than one pair of "Waist-Extension-A/B" components allows the frame to maintain a close proximity to the waist of the user. An example is shown in Figure 3.15.

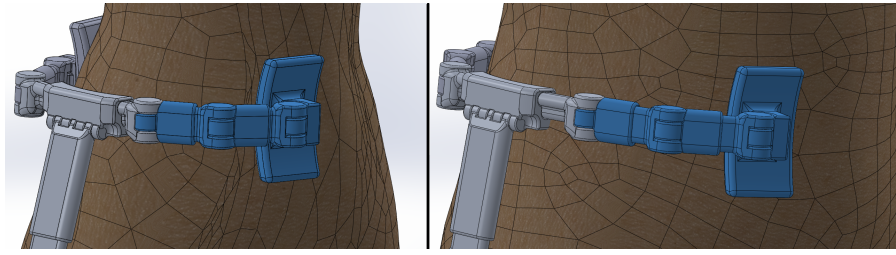


Figure 3.15: Components "Waist-Extension-A/B" and "Waist-Contact" adapted to a model with a slimmer abdominal area on the left and a model with a larger abdominal area to the right.

Since component "Waist-Contact" is supposed to press on the lumbar and abdominal regions when using the exoskeleton, the area of contact will need to have cushioning, as shown in Figure 3.16.

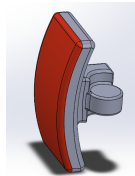


Figure 3.16: Component "Waist-Contact" with the surface that will need cushioning highlighted in red.

3.3.2 Hip to Knee Section

3.3.2.1 Thigh-Up

As mentioned in 3.3.1.1, component "Hip-Fit" houses a spherical bearing from Schaeffer, model GE20-FW. It allows for free movement of hip flexion and extension, and the spherical implementation allows for 17° for hip rotation and abduction/adduction. According to [86], angle values for hip rotation and abduction/adduction during normal gait are lower than 10° for each direction, so the level of rotation provided by the spherical bearing is deemed sufficient for normal gait.

Component "Thigh-Up-A" inserts directly into the bearing and has a socket connection with variable distance to component "Thigh-Up-B". Component "Thigh-Up-C" provides an interface with the front of the thigh and has two socket connections with variable distances, one to connect with "Thigh-Up-B" and another to connect with "Thigh-Mid-A" mentioned in 3.3.2.2. The components "Thigh-Up-A/B/C" are seen in Figure 3.17.

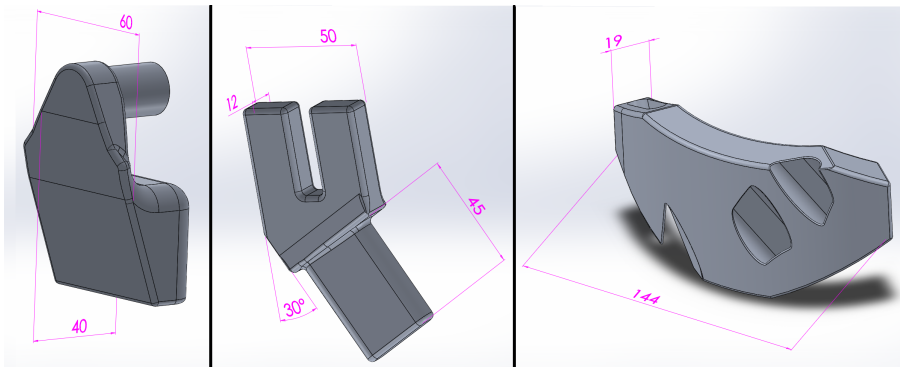


Figure 3.17: From left to right, components "Thigh-Up-A", "Thigh-Up-B" and "Thigh-Up-C". Some measurements are shown in millimeters.

The connection between "Thigh-Up-A" and "Thigh-Up-B" has a variable distance parallel to the Coronal and Sagittal planes and it serves to adapt to different distances between the hip and knee joints. The connection with a variable distance between "Thigh-Up-B" and "Thigh-Up-C" is parallel to the Sagittal plane and has an angle of 30° with the Coronal plane, allowing the frame to adapt to users with more volume in the thigh towards the front. Figure 3.18 shows an example of these components on two different models.

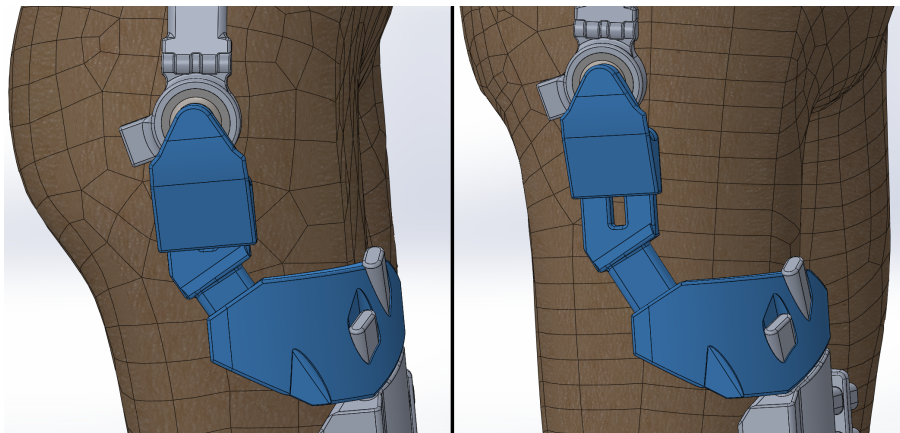


Figure 3.18: On the left, components "Thigh-Up-A/B/C" are adapted to a model with a shorter hip-to-knee distance and slimmer thigh. On the right, the same group of components is adapted to a model with a longer hip-to-knee distance and a thicker thigh.

Component "Thigh-Up-C" is supposed to press on the thigh when the exoskeleton is assisting with hip extension, so the area of contact shown in Figure 3.19 will need cushioning.

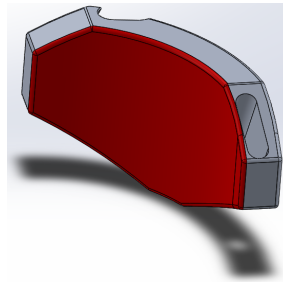


Figure 3.19: Component "Thigh-Up-C" with the surface that will need cushioning highlighted in red.

3.3.2.2 Thigh-Mid

The component "Thigh-Mid-A" connects to "Thigh-Up-C" and to "Thigh-Mid-B" through socket connections with a variable distance. "Thigh-Mid-A" also has a socket connection to fit "Knee-Up-A". Components "Thigh-Mid-A" and "Thigh-Mid-B" are seen in Figure 3.20.

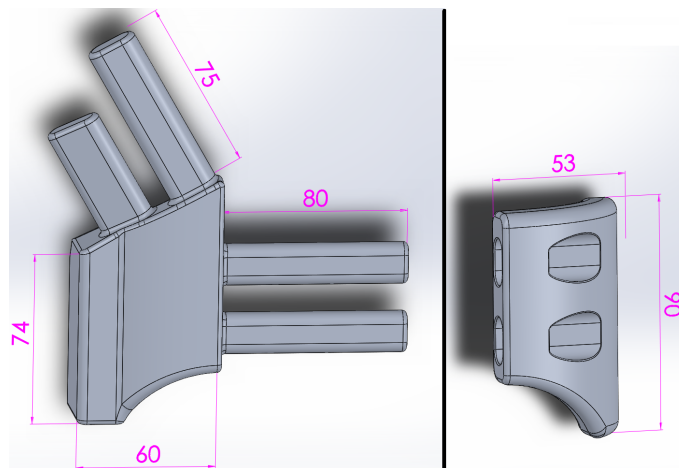


Figure 3.20: Components "Thigh-Mid-A" on the left and "Thigh-Mid-B" on the right. Some measurements are shown in millimeters.

"Thigh-Mid-A" and "Thigh-Mid-B" serve the purpose of adapting to the thigh's width and making contact with the front of the thigh. "Thigh-Mid-A" connects to "Thigh-Up-C" through a variable distance parallel to the Coronal plane and with an angle of 40° to the Sagittal plane. The connection between "Thigh-Mid-A" and "Thigh-Mid-B" is parallel to the Transverse and Coronal planes. Figure 3.21 shows these components adapted to thighs with different widths.

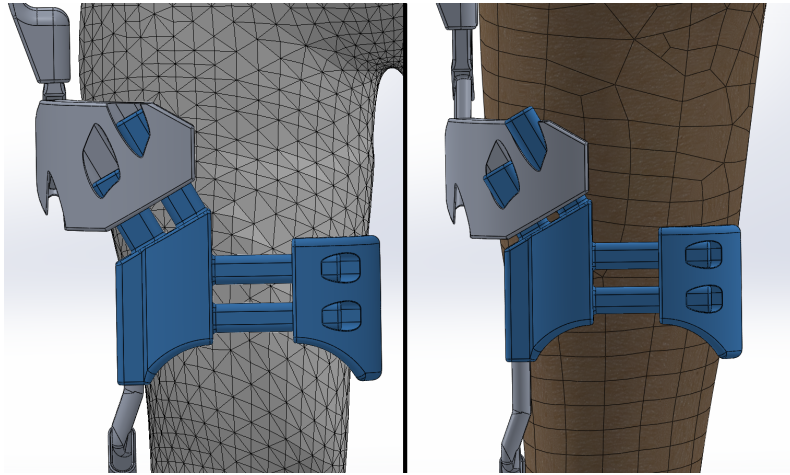


Figure 3.21: Components "Thigh-Mid-A/B" adapted to a wider thigh on the left and to a narrower thigh on the right.

Components "Thigh-Mid-A/B" also press on the thigh when the exoskeleton is assisting with hip extension, so the areas shown in Figure 3.22 need cushioning.

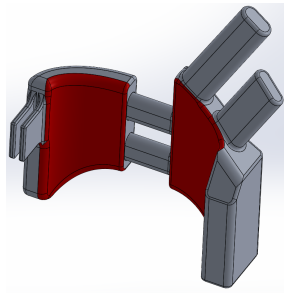


Figure 3.22: Surfaces that will need cushioning in "Thigh-Mid-A/B" highlighted in red.

3.3.2.3 Knee-Up

Component "Knee-Up-A" connects to "Thigh-Mid-A" and "Knee-Up-B". "Knee-Up-B" has a socket to house "Knee-Up-A" and also houses a regular bearing for knee flexion and extension. "Knee-Up-A" and "Knee-Up-B" are seen in Figure 3.23.

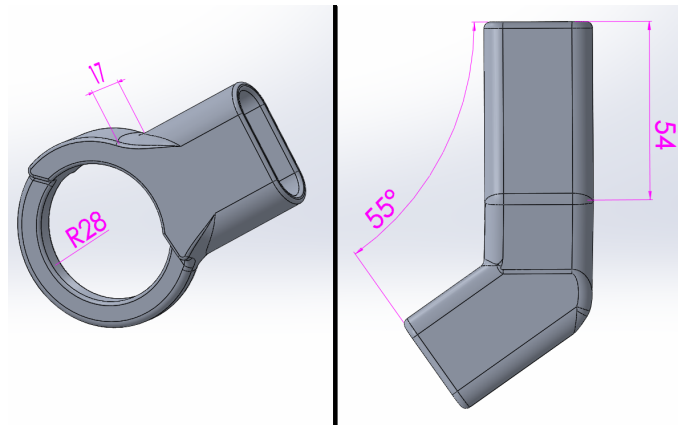


Figure 3.23: Components "Knee-Up-A" on the left and "Knee-Up-B" on the right. Some measurements are shown in millimeters and angles in degrees.

The variable distance between "Knee-Up-A" and "Thigh-Mid-A" has a direction parallel to the Sagittal and Coronal planes, and allows the frame to adapt to different hip-to-knee distances. The small distance between "Knee-Up-A" and "Knee-Up-B" is parallel to the Sagittal plane, and has a 125° angle with the Transverse plane. This distance allows the frame to adjust the rotation of the bearing depending on the knee volume, as seen in Figure 3.24.

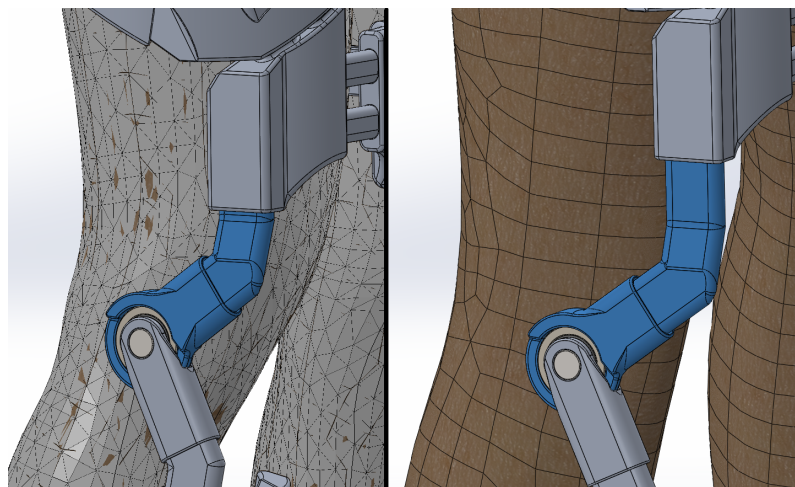


Figure 3.24: Components "Knee-Up-A/B" adapted to a smaller knee volume on the left, and a larger knee volume on the right.

3.3.3 Knee to Heel Section

3.3.3.1 Knee-Down

Component "Knee-Down-A" requires a shaft that enters the bearing that is fitted in component "Knee-Up-B". With this set, the mechanical frame can rotate freely with knee flexion and ex-

tension. "Knee-Down-A" also has a socket to connect "Knee-Down-B" with a variable distance. "Knee-Down-B" connects to "Knee-Down-A" and "Knee-Down-C". "Knee-Down-C" has a socket to connect "Knee-Down-B" with a variable distance and extensions to connect to "Shin-Up-A/B". Components "Knee-Down-A/B/C" are shown in Figure 3.25.

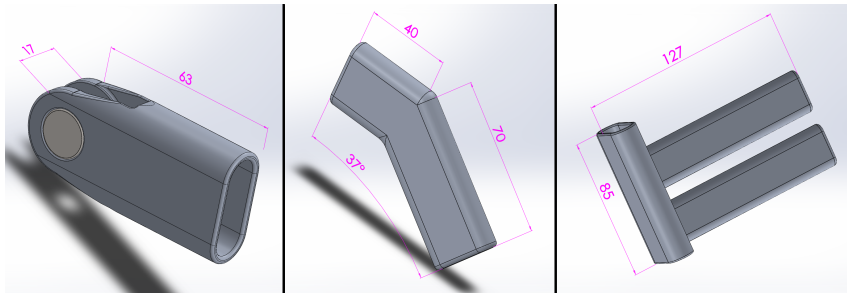


Figure 3.25: From left to right, components "Knee-Down-A", "Knee-Down-B" and "Knee-Down-C". Some measurements are shown in millimeters for distances and degrees for angles.

The distance between components "Knee-Down-A/B" is parallel to the Sagittal plane and has a 143° angle with the transverse plane, and its purpose to adapt to the knee's rotation axis, similarly to the "Knee-Up-A/B" distance. The distance between components "Knee-Down-B/C" is parallel to the Coronal and Sagittal planes, and adapts to different knee-to-ankle distances. An example is shown in Figure 3.26.

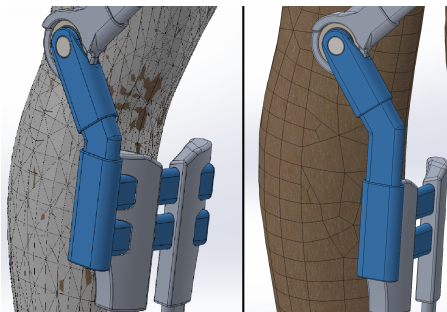


Figure 3.26: Components "Knee-Down-A/B/C" adapted to a smaller knee and shorter knee-to-ankle distance on the left, and a larger knee with longer knee-to-ankle distance on the right.

3.3.3.2 Shin-Up

Components "Shin-Up-A" and "Shin-Up-B" are symmetrical to the Sagittal plane. Both components have sockets that fit with "Knee-Down-C", sockets to fit with extension bars that connect to the "Shin-Down-A/B" components. Components "Shin-Up-A/B" have a large area of contact that press on the upper zone of the shin to assist knee flexion during normal gait, and to hamper the knee during stair descent. These components are shown in Figure 3.27

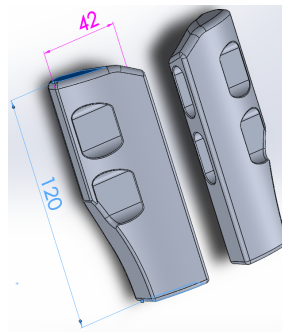


Figure 3.27: Components "Shin-Up-A/B". Some measurements are seen in millimeters.

The variable distance connection between "Shin-Up-A/B" and "Knee-Down-C" is parallel to the Coronal and Transverse planes, and allows the frame to adapt to different shin widths and slightly different alignments between the femur and tibia. An example is seen in in Figure 3.28.

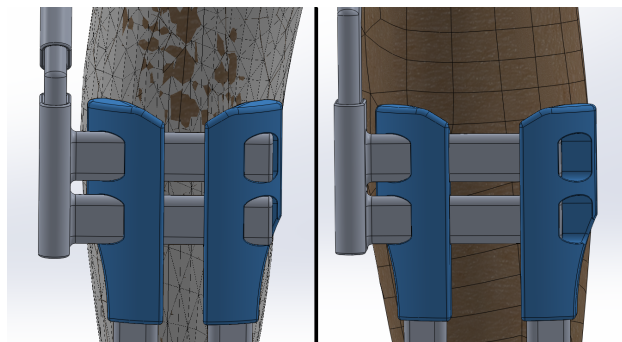


Figure 3.28: Components "Shin-Up-A/B" adapted to a narrower upper shin on the left, and a wider upper shin on the right.

Since components "Shin-Up-A/B" are designed to press the upper shin area during knee flexion for gait and stair descent, these will need cushioning within the areas highlighted in Figure 3.29.

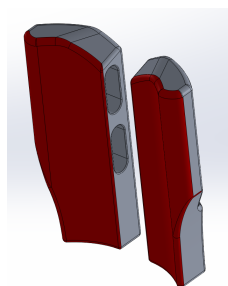


Figure 3.29: Surfaces that will need cushioning in components "Shin-Up-A/B", highlighted in red.

3.3.3.3 Shin-Down

Components "Shin-Down-A" and "Shin-Down-B" connect through vertical extension bars to components "Shin-Up-A" and "Shin-Up-B", respectively. Therefore, the positioning of "Shin-Down-A/B" is parallel to "Shin-Up-A/B" and the Coronal plane. "Shin-Down-C" has sockets to fit "Shin-Down-A/B" with a variable distance, has a shape that circles to the outer zone of the foot and has a disk-shaped socket to fit "Ankle-Up-A" from 3.3.3.4. Components "Shin-Down-A/B/C" are shown in Figure 3.30.

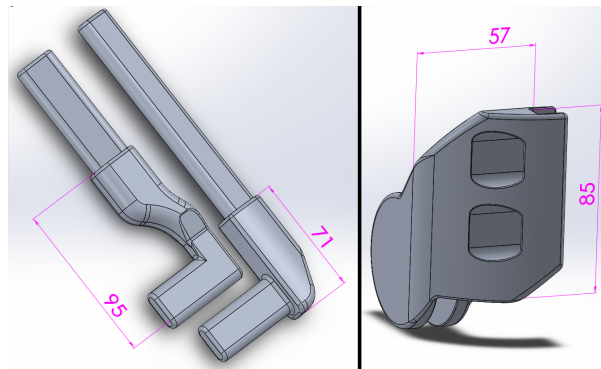


Figure 3.30: From left to right, components "Shin-Down-A", "Shin-Down-B" and "Shin-Down-C". Some measurements are seen in milimeters.

The variable distance connections between "Shin-Down-A/B" and "Shin-Up-A/B" are parallel to the Coronal and Sagittal planes and serve the purpose of adapting the frame to users with different knee-to-ankle distances. The connections between "Shin-Down-A/B" and "Shin-Down-C" are parallel to the Coronal and Transverse planes, and allow the frame to adapt to feet with different widths. An example of these components adapting to two different bodies is shown in Figure 3.31.

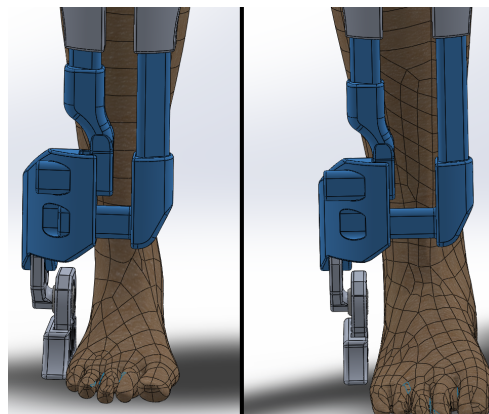


Figure 3.31: On the left, components "Shin-Down-A/B/C" adapted to a model with a shorter knee-to-ankle distance. On the right, the same components adapted to a model with a longer knee-to-ankle distance.

Although components "Shin-Down-A/B/C" are distant from the knee joint, it's possible that some force may be applied to the lower shin when the exoskeleton is assisting knee flexion, so the surfaces shown in Figure 3.32 must have cushioning.

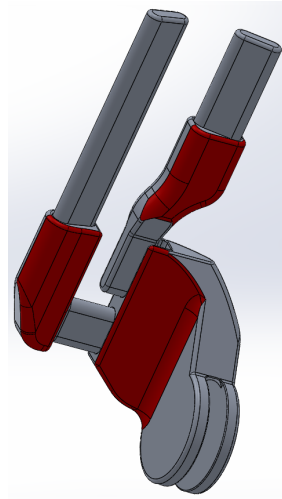


Figure 3.32: Surfaces from the "Shin-Down-A/B/C" that need cushioning, highlighted in red.

3.3.3.4 Ankle-Up and Ankle-Down

Components "Ankle-Up-A" and "Ankle-Up-B" fit inside the disk-shaped socket in component "Shin-Down-B" from section 3.3.3.3. Component "Ankle-Up-B" also fits in the bearing housed in component "Ankle-Down-A". Figure 3.33 shows components "Ankle-Down-A/B".

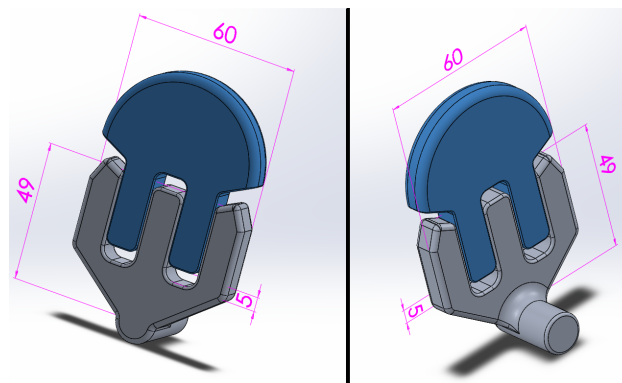


Figure 3.33: Component "Ankle-Up-A" in the top highlighted in blue and "Ankle-Up-B" in the bottom. Pictures from left and right show the same pair of components from two different perspectives.

Component "Ankle-Down-A" houses a spherical bearing from Schaeffer model GE15-FW, which allows for full ankle rotation and 16° ankle eversion/inversion. During normal gait, ankle eversion/inversion varies between -3° and 17° [86], so the limited ankle inversion may have some

impact in the natural movement. Considering that some types of footwear put a larger restriction of ankle inversion, this angle limitation should be negligible.

Component "Ankle-Down-B" has a variable distance to "Ankle-Down-A" and makes contact with the floor in order to support the weight of the exoskeleton. "Ankle-Down-B" is currently considered a "placeholder", since the surface that makes contact with the floor be made of a different material (such as rubber). Also, the variable distance between "Ankle-Down-A" and "Ankle-Down-B" must be able to change frequently, unlike the other components, because this distance will depend on the footwear chosen by the user. Figure 3.34 shows components "Ankle-Down-A/B".

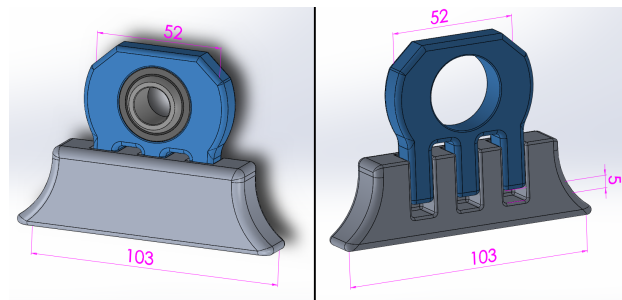


Figure 3.34: Component "Ankle-Down-A" in the top highlighted in blue and "Ankle-Down-B" in the bottom. The picture from the left shows the spherical bearing fitted in "Ankle-Down-A" and the picture from the right shows the same pair of components from two different perspectives and without the bearing.

Components "Ankle-Up-A/B", due to the disk-shaped socket, have a variable angle parallel to the Sagittal plane. Components "Ankle-Up-A/B" and "Ankle-Down-A/B" can also have a small variable distance between them (up to 15 millimeters each), parallel to the Sagittal plane. This allows these components to adapt to feet with different sizes and shapes, as shown in Figure 3.35.

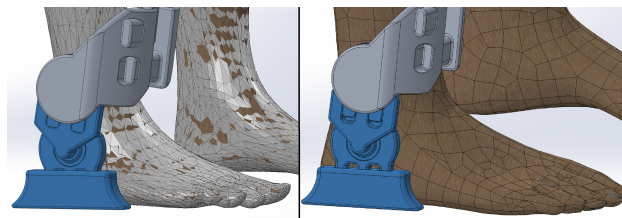


Figure 3.35: Components "Ankle-Up-A/B" and "Ankle-Down-A/B" adapted to a smaller foot on the left, and a larger foot on the right.

3.4 Torque and Force Tests

This section presents the results of the simulations made to evaluate the mechanical properties of the custom designed mechanical frame for the exoskeleton. These were made using the Solid-

works' Finite Element Method Simulation module, as static tests. The tests presented are presented into two different categories: Maximum Stress and Nominal Stress. The maximum stress consists of applying, to the mechanical frame, the typical torque values for hip and knee flexion/extension taken from other scientific studies that use young and healthy subjects, namely for horizontal gait [10] and stair climbing/descending [87]. These values are used as a reference to verify the safety margins for the mechanical behavior of the exoskeleton's frame. Each result presents a maximum Von Mises stress, which is compared to the maximum yield strength of the chosen material, Aluminium 6061-T6, 275 MPa.

The exoskeleton in this project is designed to assist the movement on elderly subjects, and not to completely replace the effort of hip and knee flexion/extension on young and healthy individuals. Elderly subjects generally move slower by applying lower angular speeds in the knee (-25%) and the hip (-20%) than young individuals [12]. This means the torque generated by these joints is also 25% to 20% lower, according to the torque formula that associates with moment of inertia and angular acceleration. It is then fair to assume:

$$\tau = I \times \alpha \quad (3.1)$$

$$\tau = I \times \frac{d\omega}{dt} \quad (3.2)$$

Where τ is the torque in N.m, I is the moment of inertia in $kg.m^2$, α is the angular acceleration and ω is the angular speed. If the angular speed ω is reduced, so will the resulting torque τ . It is then fair to assume that joint torques are lower in horizontal gait if the walking pace is also lower. On the other hand, the weight of the exoskeleton is expected to impose an excess torque to perform the daily activities proposed: horizontal gait, stair climbing/descending, sitting and getting up. This factor is explored in Chapter 4, through an workflow of experimental data and multi-body simulations.

Given the arguments mentioned above, the Nominal Stress tests take into account an estimated 50% assistance to the torques for hip and knee flexion/extension. In this section, the excess torques imposed by using the exoskeleton are expected to be at least partially compensated by the lower torques applied by elders who move at a slower pace. Therefore, the Nominal Stress tests use 50% of the torque values used in the Maximum Stress tests for hip and knee flexion/extension. While the Maximum Stress tests have the results compared with the maximum yield strength, the Nominal Stress tests compare the maximum Von Mises stress to the fatigue limit under cyclic load for Aluminium 6061-T6, which is 97 MPa.

The Nominal Stress tests are used as a reference for normal usage, whereas the Maximum Stress tests are used to evaluate the safety limits of the mechanical frame.

Stair climbing and descending are the activities that demand larger torque values. During stair climbing, a single limb is responsible for raising the whole human body mass against the direction

of the gravity. During stair descending, a single limb is responsible for stopping the body mass from an acceleration in the direction of the gravity. Table 3.1 shows the torque values for hip and knee extension/flexion being used for each test.

Table 3.1: Torque used in the FEM tests of the mechanical frame. Bibliographic references for values are also shown.

Movement	Activity	Value (N.m)
Hip Flexion Maximum Stress	Horizontal walking	60 [10]
Hip Flexion Nominal Stress	50% Horizontal walking	30
Hip Extension Maximum Stress	Stair Climbing	150 [87]
Hip Extension Nominal Stress	53% Stair Climbing	80
Knee Flexion Maximum Stress	Horizontal walking	60 [10]
Knee Flexion Nominal Stress	50% Horizontal walking	30 (<i>note: not needed</i>)
Knee Extension Maximum	Stair Descending	150 [10]
Knee Extension Nominal	50% Stair Descending	75

The tests also consider the maximum element displacement, because the mechanical frame must be stable enough to be used as an assistive device. This value is taken into consideration especially for the Nominal Stress tests, as large displacements could cause the user to loose balance or feel unsafe.

The components where the torque is applied are the "Thigh-Up-A" (described in 3.3.2.1) for hip flexion/extension and "Knee-Down-A" (described in 3.3.3.1) for knee flexion/extension.

The system is planned to use the actuators placed in a backpack which is supported by the mechanical frame. The actuators then connect to the components through non-elastic bowden cables. With this setup, the torque output from the actuators is transferred directly to the mechanical frame. Additional components will need to be developed to adapt the bowden cables to the joints and guarantee a pull force that is always perpendicular to the rotating arm in each joint. For the tests, the torque from the actuators is represented as a torque applied to the surfaces of the components that are aligned with the rotation.

The mechanical frame is designed to envelop the user from the front and sides, in order to be easy and fast to mount. Also not yet fully considered in the mechanical frame design is the fact that the system is supposed to use belts to fasten the exoskeleton to the lower limbs. The belts are used to fasten and propel the limbs in situations that require lower torque levels, namely hip flexion and knee extension during horizontal gait. A support belt for the thigh is planned to connect between components "Hip-Fit" (3.3.1.1) and "Thigh-Mid-B" (3.3.2.2). Another support belt for the lower knee region is planned to connect components "Shin-Up-A" and "Shin-Up-B". A picture of the system with the belts can be seen in Figure 3.36.



Figure 3.36: Render of the exoskeleton with the planned fastening belts highlighted in blue.

3.4.1 Hip Flexion and Extension

3.4.1.1 Hip Flexion Maximum Stress

For the Hip Flexion Maximum Stress, the simulation has a 60 N.m torque applied the normal surfaces of component "Thigh-Up-A" and gravity. The fixed geometry is set as the edges of components "Hip-Fit" and "Thigh-Mid-B" to represent the thigh belt, and in the contact surfaces of components "Waist-Contact" (Figure 3.16). Figure 3.37 shows the resulting Von Mises graphic.

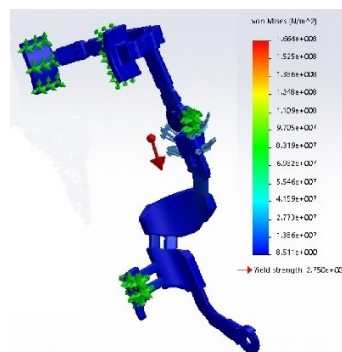


Figure 3.37: Results for the Hip Flexion Maximum Stress with a 60 N.m torque.

The maximum Von Mises Stress observed is 166 MPa and the maximum displacement observed is 0.24mm.

3.4.1.2 Hip Flexion Nominal Stress

The Hip Flexion Nominal Stress uses the same configuration as the Maximum Stress, with the only difference being the torque value, which is 30 N.m. The results can be seen in Figure 3.38

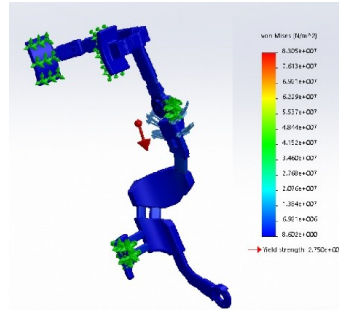


Figure 3.38: Results for the Hip Flexion Nominal Stress with a 30 N.m torque.

For this test, the maximum Von Mises Stress observed is 83 MPa and the maximum displacement is 0.11mm.

3.4.1.3 Hip Extension Maximum Stress

The Hip Extension Maximum Stress corresponds to stair climbing. For this test, "Hip to Knee" components are aligned with an angle of 65° to simulate the leg lift to climb a stair. The fixed zones are the contact surfaces of components "Waist-Contact" (Figure 3.16) and "Thigh-Mid-A/B" (Figure 3.22). A torque of 150 N.m is applied to component "Thigh-Up-A" in the same direction as hip extension. Figure 3.39 shows the results for this test.

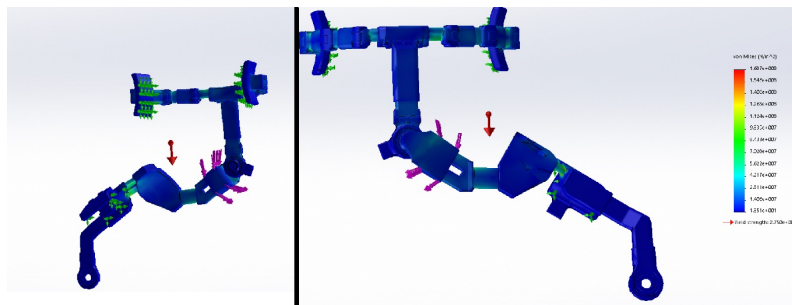


Figure 3.39: Results for the Hip Extension Maximum Stress with a 150 N.m torque.

The maximum Von Mises stress observed in this test is 169 MPa and the maximum displacement is 1.2 mm.

3.4.1.4 Hip Extension Nominal Stress

The Hip Extension Nominal Stress test uses the same parameters as the previous test, with the only difference being a reduced torque of 80 N.m. The results are shown in Figure 3.40.

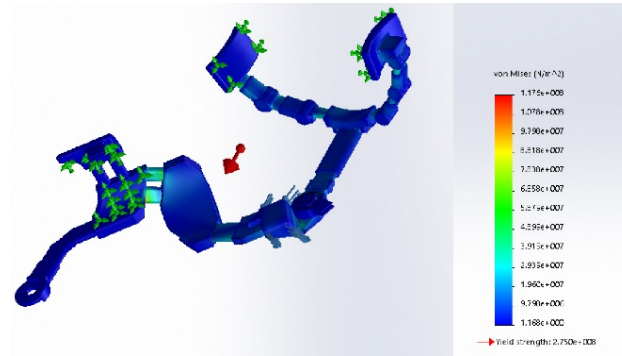


Figure 3.40: Results for the Hip Extension Nominal Stress with a 80 N.m torque.

For this test, the maximum Von Mises stress observed is 118 MPa and the maximum displacement is 0.8mm.

3.4.2 Knee Flexion and Extension

3.4.2.1 Knee Flexion Maximum Stress

The Knee Flexion Maximum Stress is performed within a more elaborate test that joins hip flexion and knee extension during the instant of largest knee torque for normal gait. In order to determine the moment and angle of highest knee flexion torque during horizontal gait, the experimental results from Hamner *et al* [10] were consulted, and the results obtain can be seen in Figure 3.41.

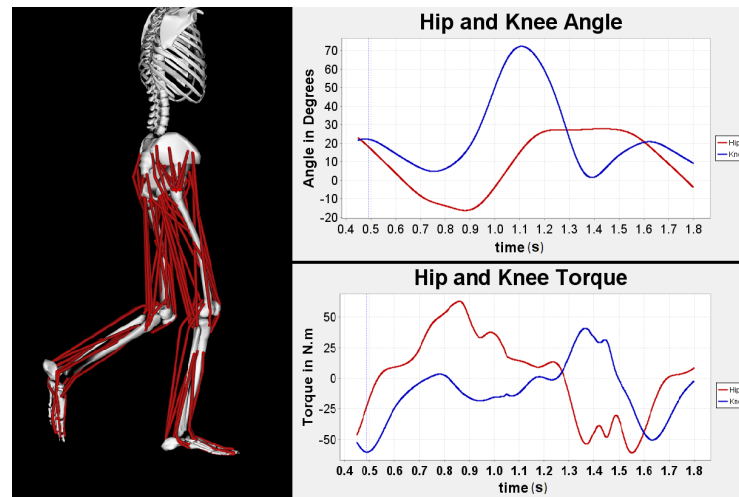


Figure 3.41: Instant of maximum Knee Flexion determined in a study from Hamner *et al* [10]. At 0.5 seconds during heel contact, the knee joint torque is 60 N.m at a 22° angle with the thigh.

For this test, all the exoskeleton components between the belt and the heel are included. To replicate the conditions in Figure 3.41, the "Hip to Knee" components have an inclination of 18° with the Coronal plane and the "Knee to Heel" components have an inclination of 22° with

the previous group. Component "Thigh-Up-A" has an applied torque of 25 N.m to simulate hip extension and component "Knee-Down-A" has an applied torque of 60 N.m to simulate knee flexion. The fixed geometry is applied to the contact surfaces of components: "Waist-Contact" (Figure 3.16), "Thigh-Mid-A/B" (Figure 3.22), "Shin-Up-A/B" (Figure 3.29) and "Shin-Down-A/B/C" (Figure 3.32).

Figure 3.42 shows the results for this test.

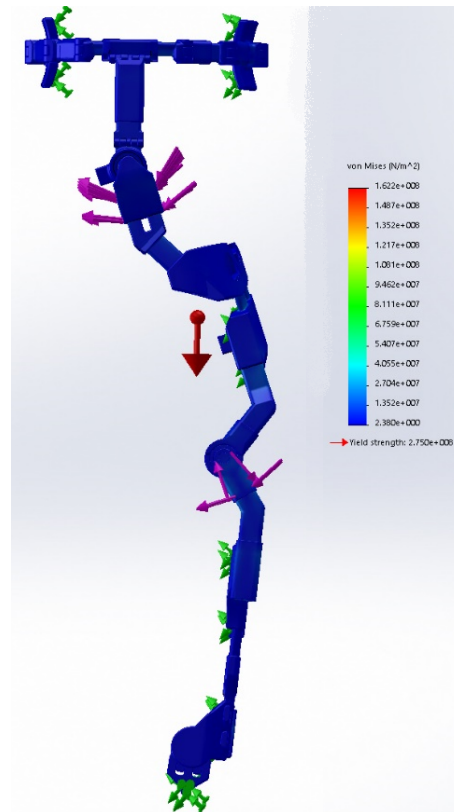


Figure 3.42: Results for the Knee Flexion Maximum Stress, with hip flexion included.

The maximum Von Mises stress observed is 162 MPa and the maximum displacement is 0.14mm. Given the relatively low displacement and stress values observed for a maximum stress test, performing a similar test with half the torque values is considered negligible to prove the frame's capacity for knee flexion.

3.4.2.2 Knee Extension Maximum Stress

The daily life activity that demands a maximum knee extension torque is stair descending. In this case, the muscles responsible for knee extension (Quadriceps group) perform negative work. The maximum value for knee extension in this case is close to 150 N.m [87]. For this test, a knee extension of 150 N.m is applied to component "Knee-Down-A" and the fixed geometry is applied to the contact surfaces of components "Thigh-Mid-A/B" (Figure 3.22) and "Shin-Up-A/B" (Figure 3.29).

The results can be observed in Figure 3.43

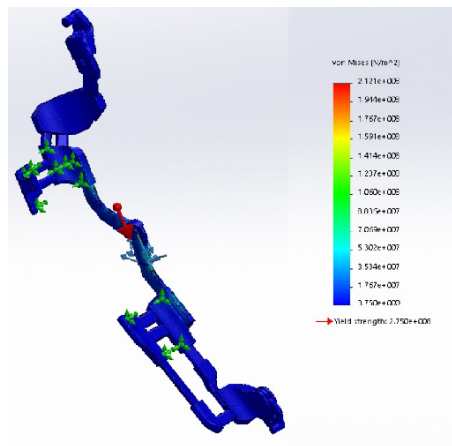


Figure 3.43: Results for the Knee Extension Maximum Stress.

The maximum Von Mises stress observed is 212 MPa and the maximum displacement is 0.3 mm.

3.4.2.3 Knee Extension Nominal Stress

The Knee Extension Nominal Stress is similar to the previous test, with the torque value applied on component "Knee-Down-A" being lowered to 75 N.m. The results are shown in Figure 3.44

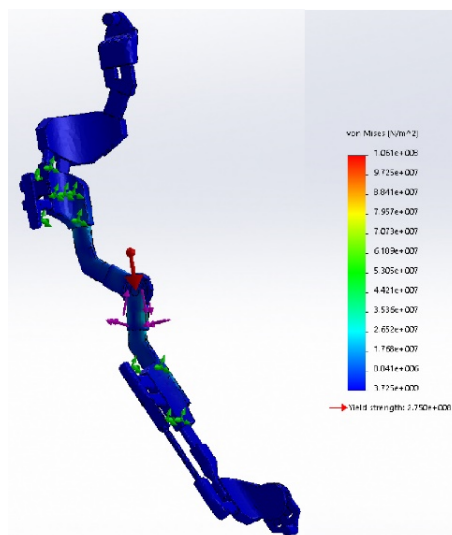


Figure 3.44: Results for the Knee Extension Nominal Stress.

For this test, the maximum Von Mises stress observed is 106 MPa and the maximum displacement is 0.13 mm.

3.4.3 Extraordinary Force Loads

This additional test has the purpose of evaluating the behavior of the full mechanical frame when subject to a large vertical force in addition to gravity. In the case of the user getting off balance or another unpredictable situation that may overload the mechanical frame, the exoskeleton should be able to allow the actuators to enter a temporary state of maximum rotor lock. This will prevent the joints from moving so the user can use the frame for full support.

Therefore, this test consists of using the exoskeleton from the belt to the heel and applying a downward force of 600 N. For a full exoskeleton frame with both sides, this would mean that the exoskeleton is supporting a total of 1200 N or 120 kg. All the joints are locked and the only fixed surface is in component "Ankle-Up-B". To the top of the belt components, a force of 600 N is applied, as well as gravity. The result can be seen in Figure 3.45

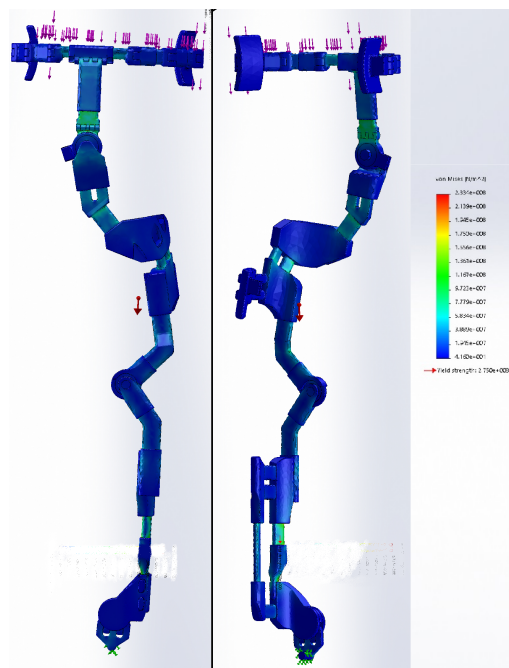


Figure 3.45: Results for the Extraordinary Force Load in an emergency situation.

In this test, the maximum Von Mises stress is 233 MPa and the maximum displacement is 37 mm.

3.5 Conclusions

The current mechanical frame design achieves most of the goals proposed in the beginning of this chapter. The mechanical frame is able to adapt to bodies of several sizes and shapes. Figure 3.46 shows the developed mechanical frame adapted to a female model with a height of 1.55 m and a male model with a height of 1.84 m.

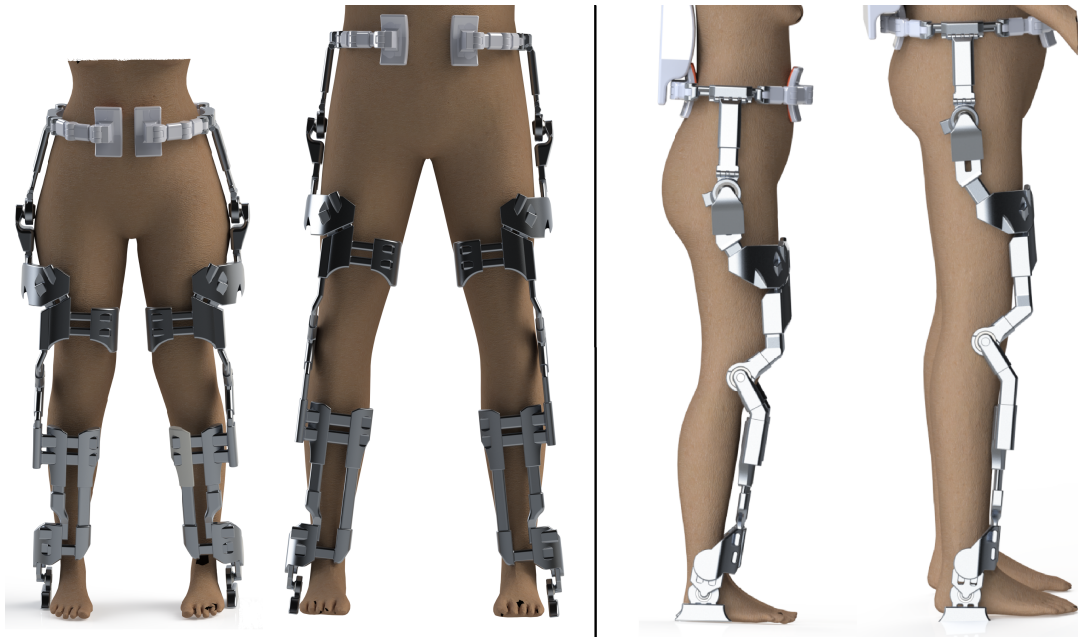


Figure 3.46: Render of the mechanical frame adapted to a female model with a height of 1.55 m and a male model with a height of 1.84 m.

Each "half frame" consists of 36 components and weights 3.9 kg. The full exoskeleton frame on both sides has a total weight of 7.8 kg. The maximum stress tests show that the exoskeleton can withstand the torques necessary to fully replace hip and knee flexion/extension. However, the the Von Mises stress levels are significantly above the maximum fatigue levels of the material, so these torques cannot be used on a daily basis. When performing a torque assistance of 50% for each movement, the values are adequate for daily usage, as the Nominal Stress tests show a maximum Von Mises stress between 83 and 118 MPa for a maximum fatigue level of 97 MPa of the material for 500 million cycles.

With the proposed components, the mechanical frame is designed to provide active hip and knee flexion/extension movements. Through the spherical bearings, the passive movements allowed are 17° abduction/adduction and rotation for the hip, as well as full plantar flexion/dorsiflexion and 16° eversion/inversion and axial rotation for the ankle.

However, the following design factors must still be studied:

- The bonding method between components must be studied, as using screws or adhesives might alter the mechanical behavior of the system;
- The heel to floor components must be redesigned and developed with means to be able to regularly change the distance to the floor, so they can adapt to different footwear;
- A method and additional components must be developed to hold the backpack into the belt, in a way that won't provoke a significant instability to the system;

- Additional components to hold and properly drive the bowden cables must be developed;
- A proper mechanical project must be developed to provide tolerance and precision levels needed to produce a full-scale prototype.

Chapter 4

Multi Body Simulations from Motion Capture Results

4.1 Methodology

This chapter describes the studies performed to determine the requirements for the actuation system.

The exoskeleton is designed to support its own weight. However, when the exoskeleton is equipped, the combined system composed of human body and exoskeleton is bound to demand additional torque than the human body alone. This happens because the total weight is higher, so the torque levels required to perform daily life activities will also be greater. The formula for the torque is expressed in equation 3.1, and given I as the moment of Inertia dependent on the mass m and distance r to the pivot:

$$I = m \times r^2 \quad (4.1)$$

therefore, the torque τ for a given movement depends on the mass m , distance r and angular acceleration α :

$$\tau = m \times r^2 \times \alpha \quad (4.2)$$

If the purpose of the exoskeleton is to provide the same natural movements that are performed on a daily basis, then the angular acceleration α is ideally the same, and r is constant because the exoskeleton adapts its length to the user's limbs. Therefore, an increment on the mass m of the bodies being moved results in an increment on the torque τ .

So the sum of torques required to move a system composed of human body and exoskeleton is greater than the sum of torques required to move the human body:

$$\sum \tau_{(human\ with\ exoskeleton)} > \sum \tau_{(human)} \quad (4.3)$$

The system proposes to assist the human movement in daily activities in close to 50%, so the torque provided by the exoskeleton actuators must be designed to perform 50% of the system combined of human with exoskeleton.

To achieve these values, a number of trials with human subjects was made with no weights and then with attached weights to simulate the weight of the exoskeleton.

4.2 Motion Capture Data Acquisition

The trials were performed in a laboratory capable of doing motion capture through cameras and markers placed on the subjects. There are also four force plates that exist to determine the ground reaction forces for gait trials, and are arranged according to Figure 4.1.

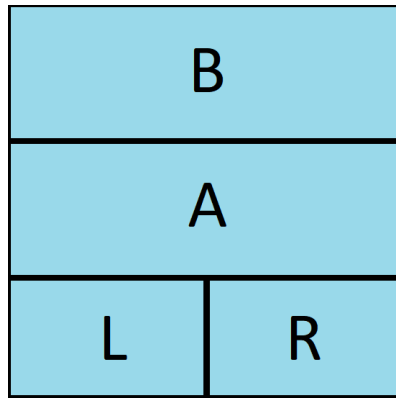


Figure 4.1: Arrangement of the force plates in the biomechanics laboratory.

During a gait trial, if the first force plate to be stepped on is plate "L", then it is assumed that plate A will be stepped on by the right foot and plate B will be stepped on by the left foot. This arrangement allows the system to acquire the ground reaction forces for a full gait cycle (seen in Figure 3.1).

The trials were performed by two subjects: a female with a weight of 52.4 kg and height of 1.52 m, and a male with a weight of 63.8 kg and a height of 1.73 m.

As seen in section 3.5, the mechanical frame of the exoskeleton weights close to 8 kg. In order to include the weight of the backpack with actuators, batteries and other subsystems, the exoskeleton simulation trials were performed with the subjects carrying an additional weight of 20 kg. The trials were performed with a weight distribution similar to the exoskeleton frame, in order to approach the dynamic behavior of the exoskeleton. The subjects attached different weights throughout the lower limbs and its distribution can be seen in Figure 4.2.

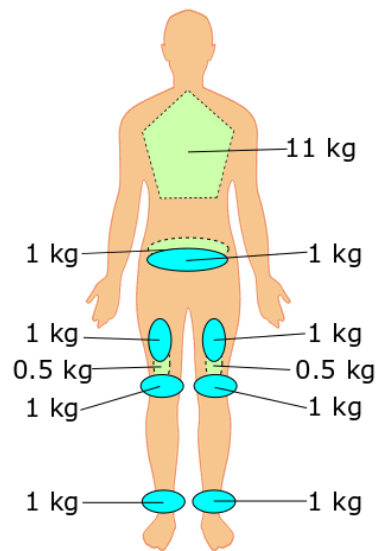


Figure 4.2: Distribution of the attached weights to approach the weight dynamics of the exoskeleton developed.

Pictures of the female and male subjects with and without the attached weights can be seen in Figures 4.3 and 4.4, respectively.

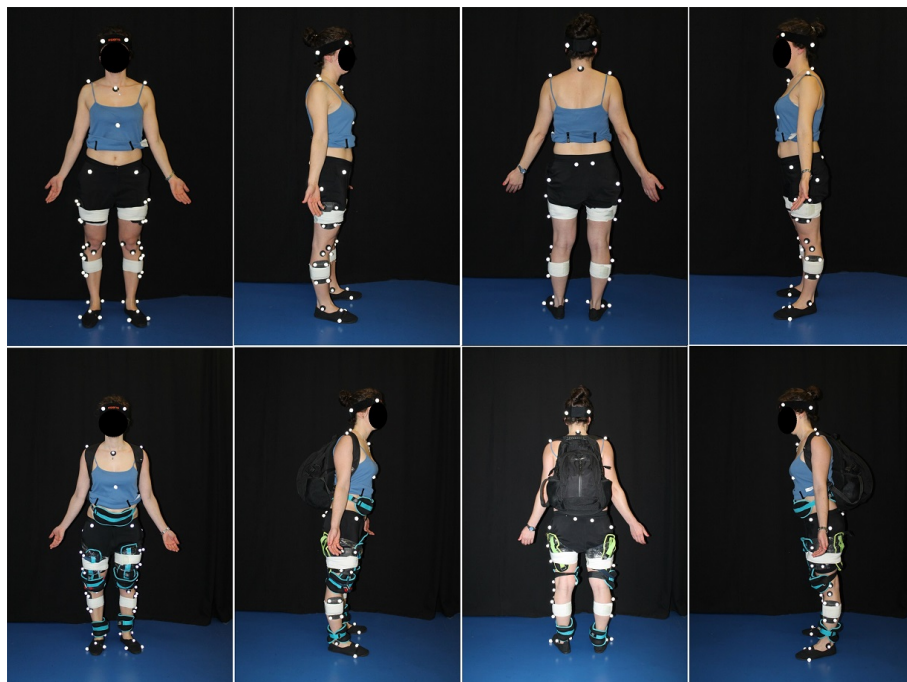


Figure 4.3: Pictures of the female subject. On top, with normal clothing and no added weights. On the bottom, with the attached weights and backpack.

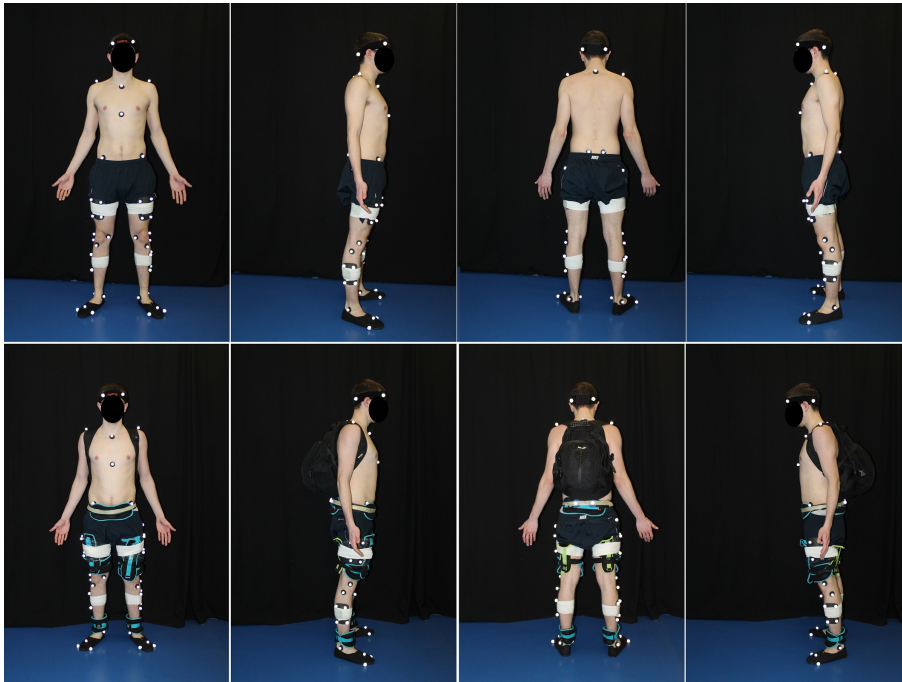


Figure 4.4: Pictures of the female subject. On top, with normal clothing and no added weights. On the bottom, with the attached weights and backpack.

The subjects performed the trials described in Figure 4.5.

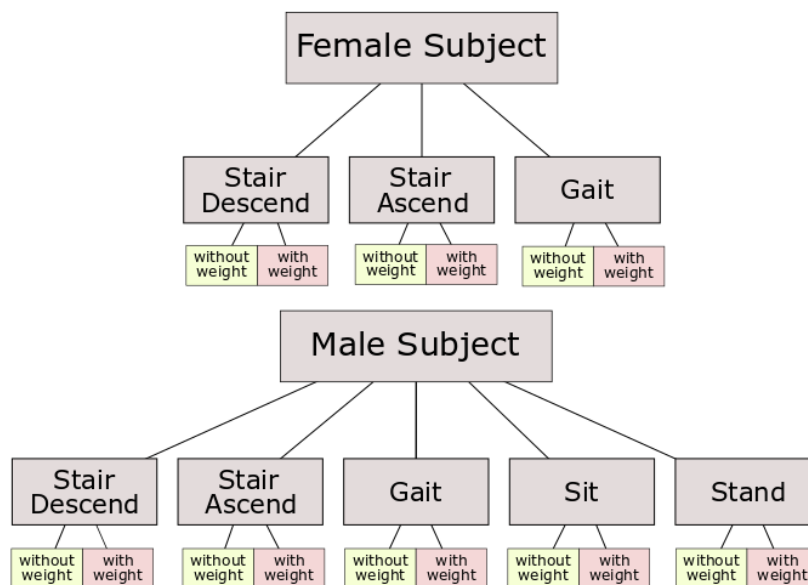


Figure 4.5: Pictures of the female subject. On top, with normal clothing and no added weights. On the bottom, with the attached weights and backpack.

Figure 4.6 shows a gait trial performed by the female subject, with and without the additional weights.



Figure 4.6: Pictures of the male subject performing gait trials with normal clothing on the left and attached weights on the right.

To perform the stair trials, a set of mock-up stairs was built with brick and wood to provide a safe method to simulate stair ascend and descend. The stair climb and descend cycles have the same description as gait cycles. Each leg performs a stance phase (between the foot making contact with one step and lifting off to the next step) and swing phase (between the foot lifting off one step and reaching the next). There are two mock-up steps: with a height of 40 cm and a second with a height of 80 cm. The lower set of stairs was placed in force plate A (from Figure 4.1) and the taller set was placed in force plate B. The acquisition software for the force plates was then tuned to subtract the weight of the mock-up steps. To complete the cycle after the taller step, there was a wooden block to complete the the ascend or initiate the descend movements. The wooden block was not placed over force plates.

In order to organize the final simulation results, the subjects were instructed to climb the mock-up stairs with the right leg on the lower step, and then to descend the stairs with the right leg first. Figure 4.7 shows pictures of the tests for stair ascend and stair descend.

For the sit and stand results, the subjects were instructed to stand backwards on the L and R platforms, and a regular bench was placed in platform A. Due to a human mistake, the ground reaction forces from the chair were not properly recorded, so the chair simulations only include the subject movements without touching the chair. Figure 4.8 shows the female subject performing a sit trial. The female subject did not perform sit and stand trials with the weights, and the trials were only used to gather EMG data.

4.3 OpenSIM Workflow

The software used to perform the multibody simulations is the OpenSim 3.3 Simulation Toolkit developed by Scott Delp *et al* [88]. The software is open source and is capable of performing multibody simulations for biomechanical applications. The musculoskeletal tridimensional model used for the simulations is the reference Gait2392 [89], which is based on anatomical data provided by several studies [90, 91]. Gait2392 has 23 degrees of motion and was designed to perform lower limb simulations.



Figure 4.7: Pictures of the female subject performing stair trials with normal clothing on the left and attached weights on the right.



Figure 4.8: Pictures of the female subject performing sit trials.

OpenSim takes the marker data and performs Inverse Kinematics to translate into motion data for the Gait2392 model, in the form of angular position per joint through time. There is also a Scaling function, which uses marker data and the subject's weight to customize the Gait2392 model. With the scaled model, motion data and ground reaction forces from the force plates, it is possible to perform Inverse Dynamics which provides the torque values for each joint.

However, the Inverse Dynamics procedure induces virtual residual actuators (consisted of torques and forces applied to the pelvis) to help stabilize the model due to data acquisition errors in the motion capture system and force plates. In complex models like the Gait2392 using complex motions, the residual actuators often assume large values, which significantly reduces the validity and precision of the calculated torques.

The Residual Reduction Algorithm (RRA) uses the Computer Muscle Control (CMC) algorithm to control the torque applied by point actuators placed in each joint. The CMC algorithm was

developed to calculate muscle activations for any given movement through power-saving paths, as a biomimetic approach [92]. At the same time, the RRA makes slight modifications the motion data and proposes mass changes to each body in order to reduce the values of the residual actuators.

Figure 4.9 shows the regular workflow for performing simulations using RRA:

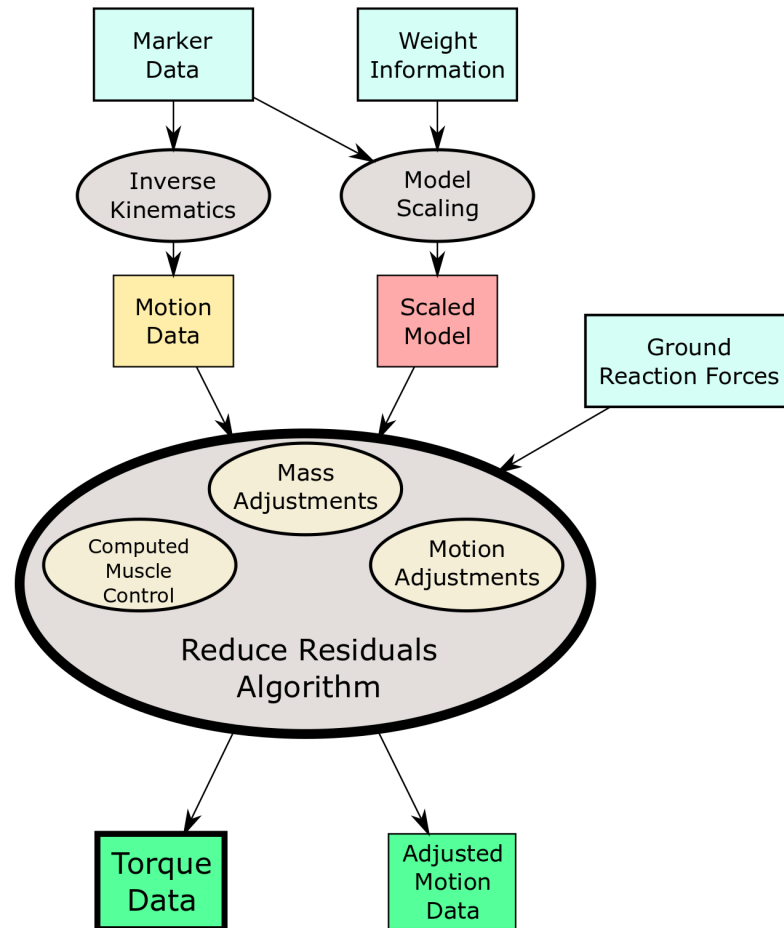


Figure 4.9: Workflow for the OpenSIM Simulation Toolkit to achieve precise values for joint torques using RRA.

According to the software developers, the results are only valid if the residual forces and error displacements assume the values seen in Figure 4.10. Therefore, the results obtained in this chapter were achieved by tuning the simulation timings to fit the thresholds for the residual actuators.

For performing the simulations with the additional weights, the exoskeleton components described in Chapter 3 were adapted to the body measurements of the male and female subjects. Afterwards, the exoskeleton components were ported and adapted into the Gait2392 model with the correct values for geometry, mass, center of mass and inertia. The weight distribution is different between the weights attached to the subjects in the trials and the virtual exoskeleton. To compensate for this difference, the RRA procedure for mass changes was repeated for each virtual model until the proposed changes were minimal. At the moment of writing, the mass change

Thresholds:	GOOD	OKAY	BAD
MAX Residual Force (N)	0-10 N	10-25N	> 25 N
RMS Residual Force (N)	0-5 N	5-10 N	> 10 N
MAX Residual Moment (Nm)	0-50 Nm	50-75 Nm	>75 Nm
RMS Residual Moment (Nm)	0-30 Nm	30-50 Nm	>50 Nm
MAX pErr (trans, cm)	0-2 cm	2-5 cm	>5 cm
RMS pErr (trans, cm)	0-2 cm	2-4 cm	>4 cm
MAX pErr (rot, deg)	0-2 deg	2-5 deg	> 5 deg
RMS pErr (rot, deg)	0-2 deg	2-5 deg	> 5 deg

Figure 4.10: Maximum threshold values for the residual forces and error displacements for the simulation results when performing walking or running trials, according to OpenSIM's documentation.

procedure is only possible through command line and is not available through the software's GUI.

The models used the OpenSim simulations can be seen in Figure 4.11.

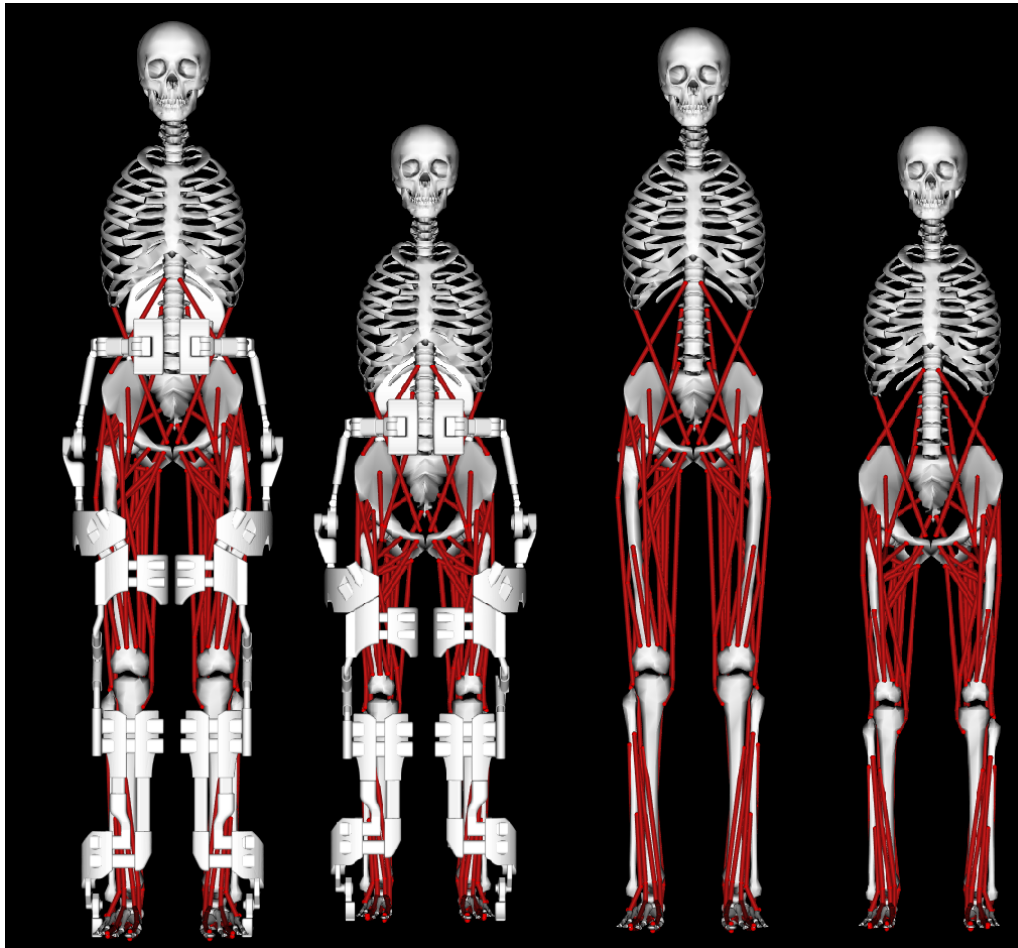


Figure 4.11: From left to right: male subject with exoskeleton, female subject with exoskeleton, male subject without exoskeleton, female subject without exoskeleton.

4.4 OpenSIM Results

The results presented in this section show the torque values for the hip and knee joints for each activity and for each subject. These joints are planned to be actuated and assisted by the exoskeleton. For some trials, the ankle torque is also shown. The ankle joint is not actuated by the system, but the torque differences may be a concern. However, a large proportion of the ankle plantar flexion torque originates from passive elastic forces from extending the Soleus muscle. Also, many types of footwear limit the plantar flexion so its importance in this system is considered partially reduced. Each result graphic shows a joint torque from a trial with weight and another joint torque from the equivalent trial without weight, to provide better means of comparison.

The OpenSim's graphics creator utility defines a positive torque for hip flexion and negative torque for hip extension. For the knee, flexion is negative and extension is positive.

In some results, the same trial for the same joint is split into two simulations. The reason for this is because in some simulations the values for the residual actuators and displacement errors would start to increase.

4.4.1 Gait Results

The following results were obtained through the OpenSIM RRA simulations with data taken from the gait trials. A screenshot of the male model with and without the exoskeleton can be observed in Figure 4.12

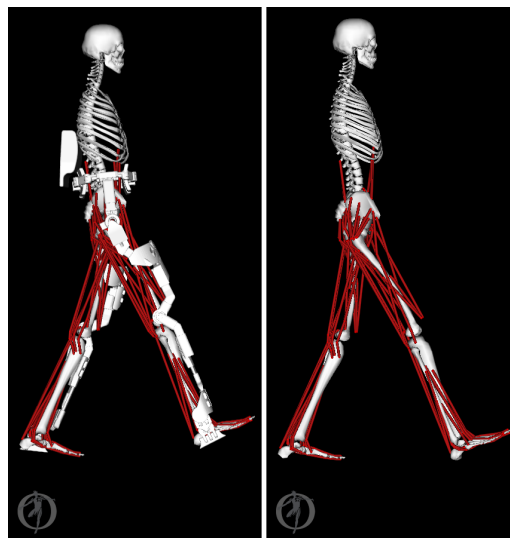


Figure 4.12: Screenshot of the simulation with the male model with the exoskeleton on the left and without the exoskeleton on the right.

Hip Joint

Figure 4.13 shows the torque results for the hip joint flexion/extension in the female subject, during the gait trials. The higher positive torque values in the hip flexion correspond to the beginning of the swing phase (60% to 70% in the gait cycle shown in Figure 3.1).

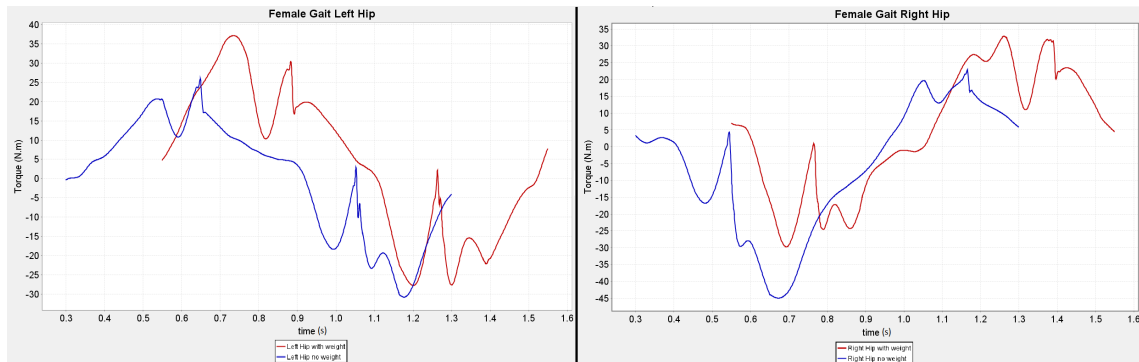


Figure 4.13: Hip Extension/Flexion torque values for the gait trials with the female subject.

With the female subject, the additional weight causes a difference of 12 N.m in the left hip flexion and 10 N.m in the right hip flexion. Hip extension is showing larger torque values without weight. A possible explanation for this is the subject assuming a faster walking pace.

Figure 4.14 shows the same trials for the male subject.

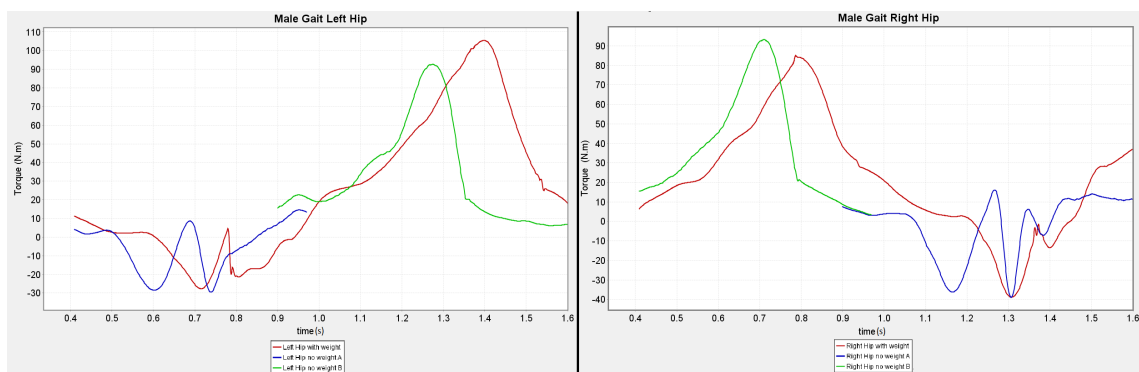


Figure 4.14: Hip Extension/Flexion torque values for the gait trials with the male subject.

The male subject shows a 12 N.m difference in the right hip flexion.

Knee Joint

The torque results for the female subject's knee joints during gait are shown in Figure 4.15. The higher positive torque values in the knee correspond to the moment of heel contact (0% in the gait cycle shown in Figure 3.1).

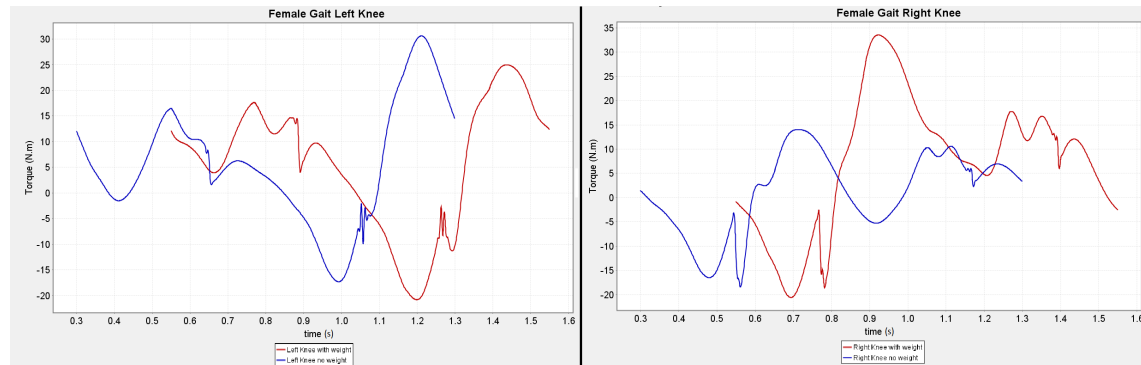


Figure 4.15: Knee Extension/Flexion torque values for the gait trials with the female subject.

The right knee results with no weight in the female trial is showing a significantly lower extension torque than the other knee results.

Figure 4.16 shows the results for the same joints from the male subject.

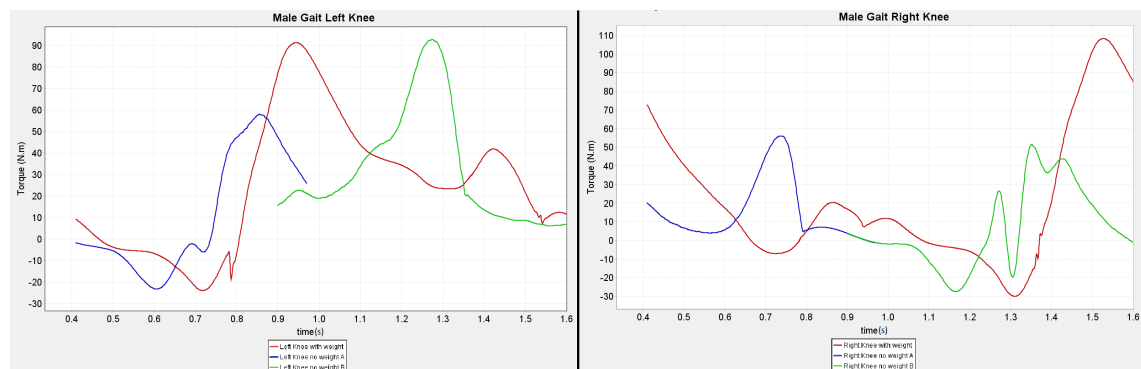


Figure 4.16: Knee Extension/Flexion torque values for the gait trials with the male subject.

Similarly to the female trial, the right knee results show a lower extension torque than the other results. Apart from this fact, there seems to be no discernible differences in the knee torque with the weight for the male subject.

4.4.2 Stair Ascend and Descend Results

The stair ascend and descend simulations could not be resolved for a full "stair" cycle. Due to the generally increased complexity of the movements, the simulation periods are reduced to keep the residual actuator values within the pre-established thresholds shown in Figure 4.10. Therefore, it is expected for the joint results in one side to match half a cycle and the other side to match the other half (e.g. left leg joints show the swing phase and the right leg joints show the stance phase).

Screenshots of the male model with and without the exoskeleton during the simulations with stairs can be observed in Figure 4.17

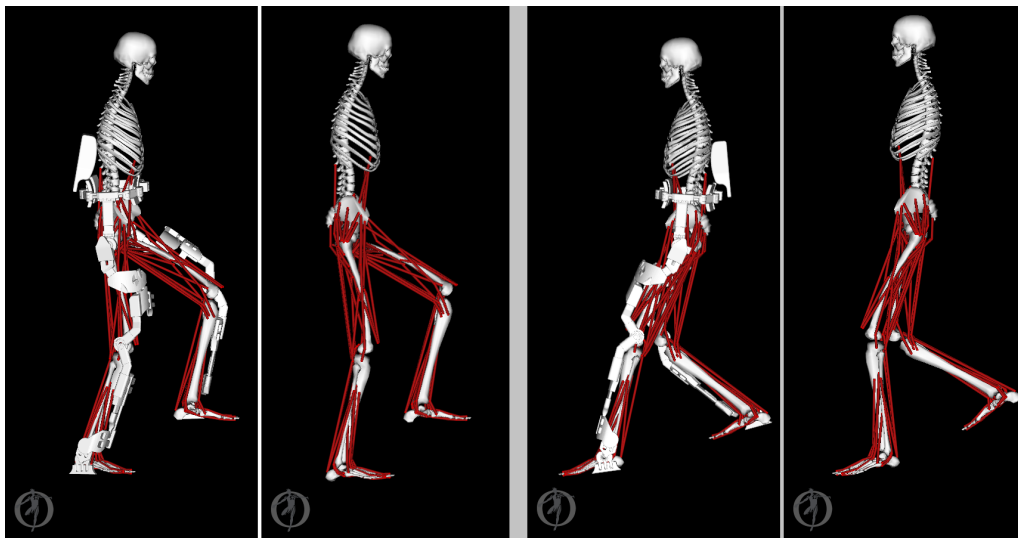


Figure 4.17: Screenshot of the simulation with the male model with and without the exoskeleton while climbing stairs on the left and descending stairs on the right.

Hip Stair Ascend

Figure 4.18 show the female hip results for stair ascend.

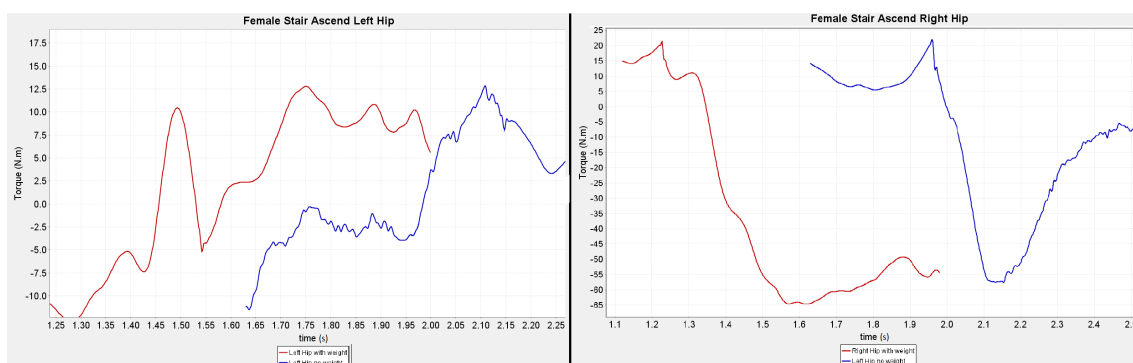


Figure 4.18: Hip Extension/Flexion torque values for the stair ascend trials with the female subject

The female hip results for stair ascend show similar torque levels for the trials with and without weight. However, the period of higher torque activation in the weight trial is significantly larger. This means that added weight resulted in a slower movement and the added mass was compensated with a lower acceleration.

Figure 4.19 shows the hip results for the male subject.

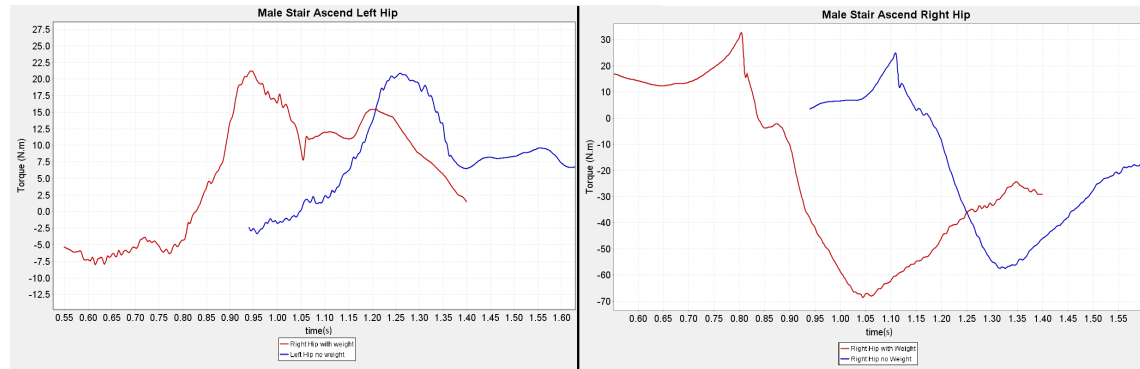


Figure 4.19: Hip Extension/Flexion torque values for the stair ascend trials with the male subject.

The maximum torque values for the male subject's hips show no discernible differences in the swing phase (left hip flexion), and the right hip shows a 10 N.m difference in extension during the stance phase.

Knee Stair Ascend

Figure 4.20 shows the knee torque values for stair ascend with the female subject.

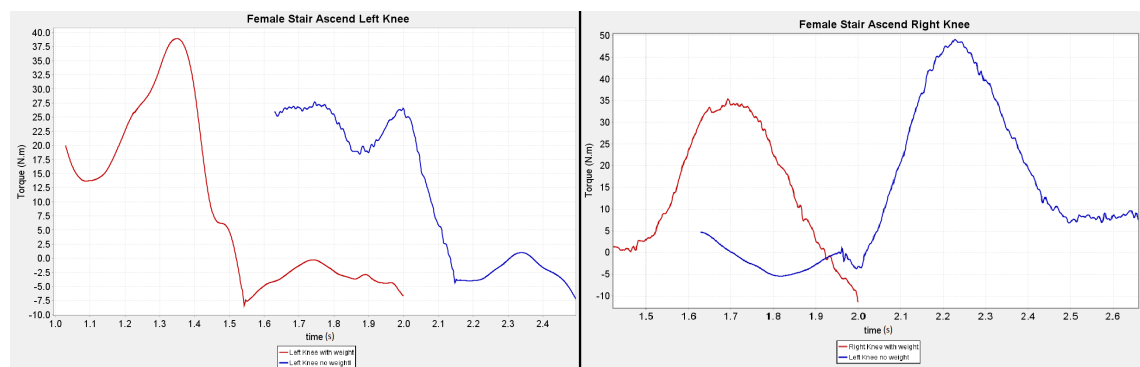


Figure 4.20: Knee Extension/Flexion torque values for the stair ascend trials with the female subject.

The weight trials show a 22 N.m increase in maximum torque for knee extension during the swing phase with the left knee. The stance phase in the right knee shows no torque increase.

Figure 4.21 show the same results for the male subject.

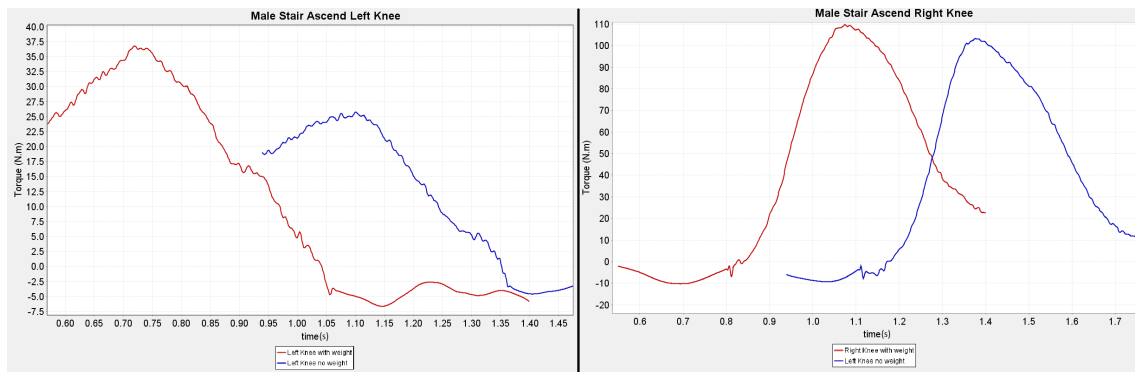


Figure 4.21: Knee Extension/Flexion torque values for the stair ascend trials with the male subject.

The male results show a 12 N.m increase for the swing phase in the left knee joint and a small torque increase for the stance phase in the right knee joint.

Hip Stair Descend

Figure 4.22 shows the hip joint torque values for the female subject during the stair descend trial.

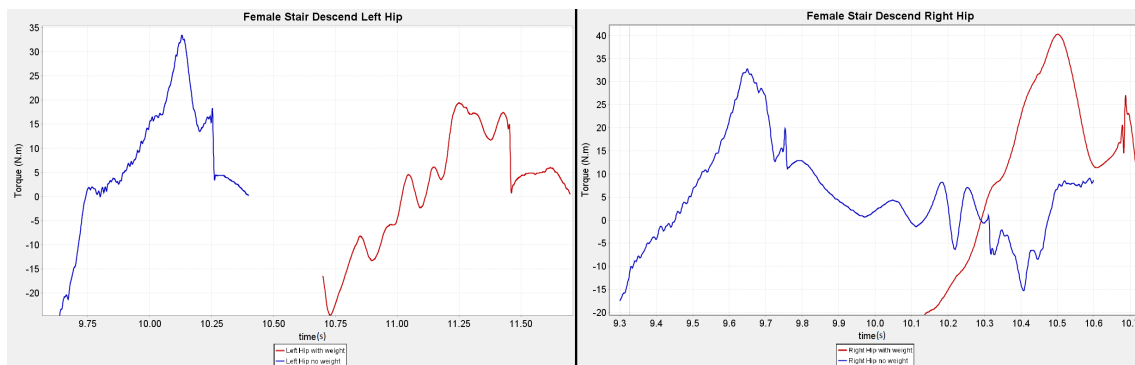


Figure 4.22: Hip Extension/Flexion torque values for the stair descend trials with the female subject.

The hip extension torque values are comparatively low during the stair descend trials because the gravity helps propel the body during the motion. The weight trials show small hip flexion torque differences from the trials with no weight.

Figure 4.23 shows the same test for the male subject.

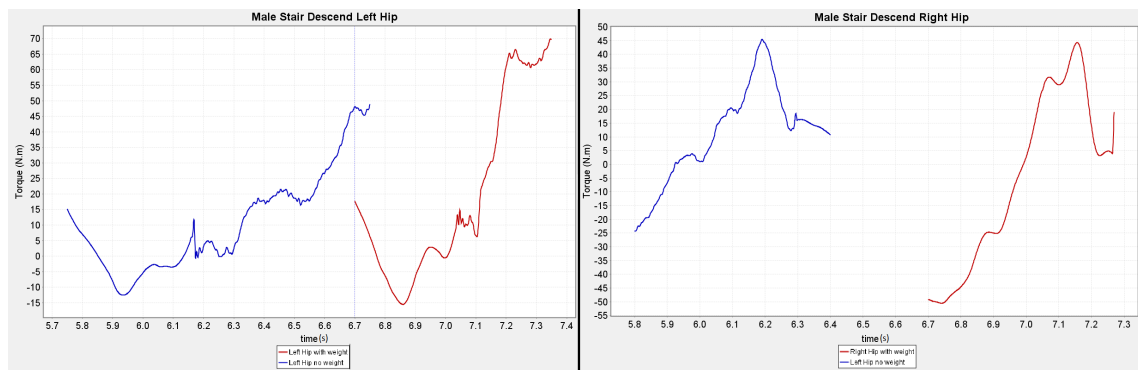


Figure 4.23: Hip Extension/Flexion torque values for the stair descend trials with the male subject.

The male subject results show a 20 N.m difference in the left hip extension and no discernible torque difference in the right hip. In the specific case of Figure 4.23, the large value observed in the right hip extension is due to the timing of the simulation and does not match the values from other simulation results.

Knee Stair Descend

Figure 4.24 show the knee torque values for stair descend with the female subject.

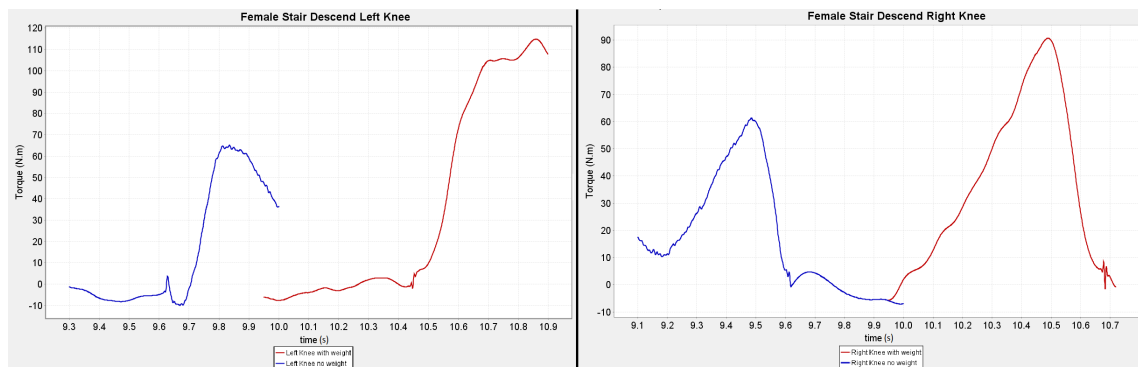


Figure 4.24: Left Knee Extension/Flexion torque values for the stair descend trials with the female subject.

The stair descend trial with the knee joint shows the largest torque differences between the results with and without weight. The knee extension peak values correspond to the moment when the subjects' feet make contact with the lower step. As noted in Table 3.1, this moment matches the highest torque observed in any of the movements studied in the trials. The largest joint torque difference observed is 30 N.m for the right knee and 50 N.m for the left knee.

Figure 4.25 shows the same trials for the male subject. The knee torque differences with and without weight are similar between the male and female subject. The male subject shows a difference of 50 N.m in both knees.

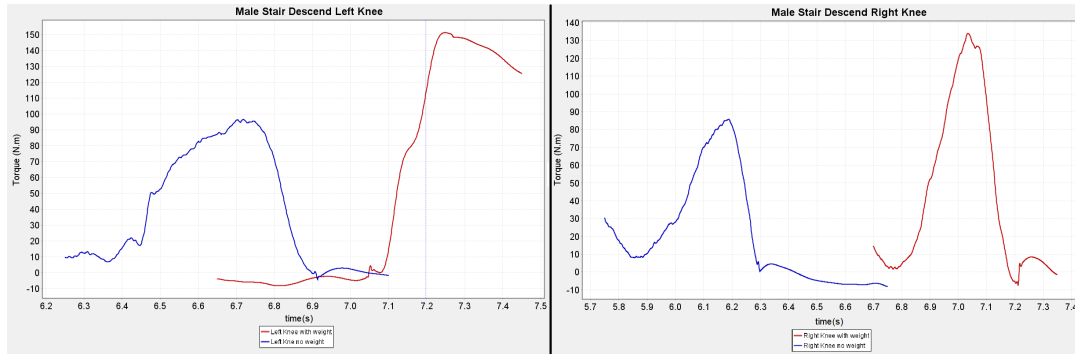


Figure 4.25: Left Knee Extension/Flexion torque values for the stair descend trials with the male subject.

4.4.3 Chair Sitting and Standing Results

As mentioned in Section 4.2, the chair simulations only include the subject movements without touching the chair. In order to produce the best possible results, the simulations were tuned to include time period between the body in a stand-up position and the moment with the largest recorded ground reaction forces, which corresponds to the larger torque values. These trials were only performed by the male subject. A screenshot of the male model with and without the exoskeleton during the sitting and standing simulations is seen in Figure 4.26.

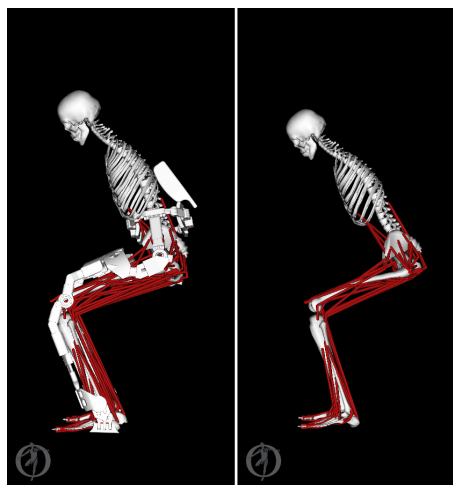


Figure 4.26: Screenshot of the simulation with the male model with the exoskeleton on the left and without the exoskeleton on the right, during the sitting and standing simulations.

Since both legs perform the same movement at the same time during sitting and standing, the following graphics show the results for each joint in both sides, with and without weight.

Figure 4.27 shows the hip and knee torque values for the sitting movement.

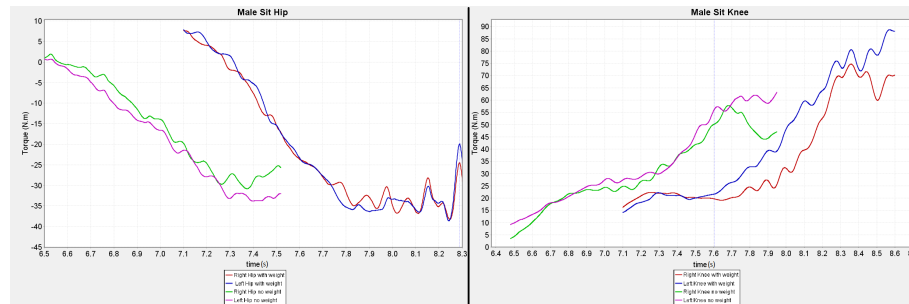


Figure 4.27: Hip and Knee Extension/Flexion torque values for the sitting trials with the male subject.

The difference in torque values for the hip movement is negligible with the added weight. For the knee joint, the largest observed torque difference is 26 N.m in the left knee joint between trials with and without weight.

Figure 4.28 shows the hip torque values for the standing movement.

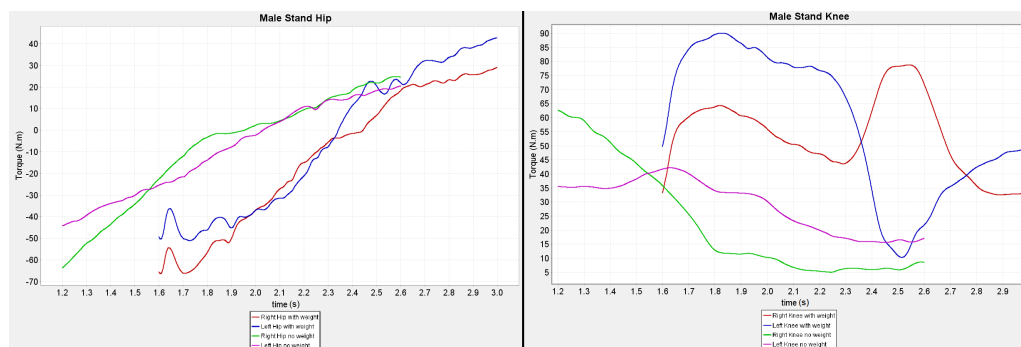


Figure 4.28: Hip and Knee Extension/Flexion torque values for the standing trials with the male subject.

The largest observed difference in hip joint is 20 N.m for the left hip. The results with the weight show that the subject stood up with higher torque from the left limb, lost balance at 2.3 seconds and afterwards compensated with right leg at 2.5 seconds. If the discrepancies seen at 2.5 seconds are ignored, the maximum torque difference observed is 27 N.m in the right knee torque.

4.5 Conclusions

Although the subjects were instructed to perform the same movements, the additional weights created a difference in the body dynamics, and therefore the movements themselves. For this reason, some tests show large differences in torque values and activation timings between trials with and without additional weight.

The Exoskeleton proposes to assist the movement in the daily activities by 50%. To achieve this, the maximum torque values from the actuation system should be able to perform 50% of the maximum torque values for a given movement without the weights ($\tau(t_{max})_{normal\ body}$), plus the difference in torque for the same movement with the weights ($\Delta\tau(t_{max})_{body\ with\ exoskeleton}$).

Therefore:

$$\tau(max)_{actuation\ system} = \tau(t_{max})_{normal\ body} \times 50\% + \Delta\tau(t_{max})_{body\ with\ exoskeleton} \quad (4.4)$$

The aforementioned torque values and torque differences are observed in the following trials:

Hip Extension:

Stair ascend with male subject: 60 N.m without weight and $\Delta\tau$ 10 N.m.

$$\tau(max)_{actuation\ system} = (0.5 \times 60) + 10 = \mathbf{40\ N.m}$$

Hip Flexion:

Stair Descend with male subject: 50 N.m without weight and $\Delta\tau$ 20 N.m.

$$\tau(max)_{actuation\ system} = (0.5 \times 50) + 20 = \mathbf{45\ N.m}$$

Knee Extension:

Stair Descend with male subject: 100 N.m without weight and $\Delta\tau$ 50 N.m.

$$\tau(max)_{actuation\ system} = (0.5 \times 100) + 50 = \mathbf{100\ N.m}$$

Knee Flexion:

Gait with male subject: 27 N.m without weight and $\Delta\tau$ 3 N.m.

$$\tau(max)_{actuation\ system} = (0.5 \times 27) + 3 = \mathbf{16.5\ N.m}$$

Table 4.1 shows an overview of the torque requirements for the actuation system.

Table 4.1: Actuation requirements for each assisted movement

Movement	Torque (N.m)
Hip Extension	40
Hip Flexion	45
Knee Extension	100
Knee Flexion	16.5

The largest requirement for torque assistance from the exoskeleton is the knee extension with a value of 100 N.m. Given the large difference between the torque requirement for this specific movement (knee joint in stair descend) and the torque requirements for the other movements, the 50% assistance value may not be met in this case, or additional types of solutions may be studied. For example, since the knee extension during stair descend is performing negative work, the same torque can be obtained through a controllable brake.

Chapter 5

Sensor, Control and Actuation Systems

5.1 Sensor and Control Systems

The proposed sensor system is based on surface electromyography. The sensor placement is studied, and experimental results from the proposed sensor placement were acquired during the trials mentioned in Chapter 4. A prototype for the control system was built, based on EMG acquisition and a low-cost single board computer.

5.1.1 Myoelectric Sensors and electrode placement

The use of surface electromyography (sEMG) to control exoskeletons has been thoroughly studied [93], and some exoskeletons in the market already use it as means of control, such as the HAL Exoskeleton mentioned in section 2.1.2. The main advantage of using sEMG is that the user's "will to move" is read directly as a biosignal, so the learning curve may be shortened and the overall system can be safer. A single sEMG signal requires a differential pair of electrodes that are placed throughout the muscle fiber length, and a reference electrode placed in a electrically neutral tissue (skin area that is close to a bone). The same reference electrode can be used for other differential pairs that are placed in other muscles.

For this project, the choice for electrode placement was based on the following guidelines:

1. Be able to detect hip flexion, hip extension, knee flexion and knee extension independently;
2. Use a minimum number of electrode pairs to avoid long setup times;
3. Place on muscles that are close to the skin so that surface electrodes can use;
4. Place on large muscles or muscle groups with the same function to avoid cross-talk from other muscles;
5. Easy to find through palpation;
6. Placed as close as possible to the waist, to avoid large wires and consequently safety hazards.

With the information above considered, the following muscles are identified:

- Biceps Femoris or Semitendinosus, with the function of performing knee flexion,
- Gluteus Maximums, with the function of performing hip extension and external rotation,
- Tensor fasciae latae, with the function of performing hip flexion and medial rotation,
- Rectus Femoris, with the function of performing knee extension and hip flexion,

Figure 5.1 shows the placement for each electrode pair:

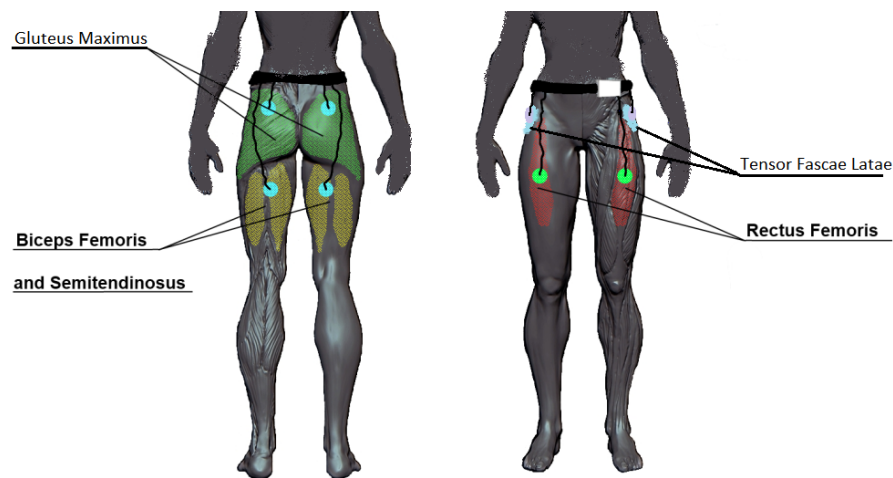


Figure 5.1: Diagram with the proposed electrode placement for EMG control.

Table 5.1 shows the predicted activation patterns among the chosen muscles for each movement [94]:

Table 5.1: Movement and Muscle Activation

Movement	Biceps Femoris	Gluteus Maximus	T. Fasciae Latae	Rectus Femoris
Heel Contact	X	X	X	X
Gait Stance	X	X	-	-
Gait Swing	-	-	X	X
Stair Stance	X	X	-	-
Stair Swing	-	-	X	-
Sit	-	X	-	X
Stand	-	X	-	X

With the data gathered through sEMG over the aforementioned muscles, it is possible to design a control system that operates mainly around muscle activation, as seen in Figure 5.2. Other sensors may be used to provide additional information, such as potentiometers to measure the rotation angle at each joint, or accelerometers to provide redundant information for increased safety.

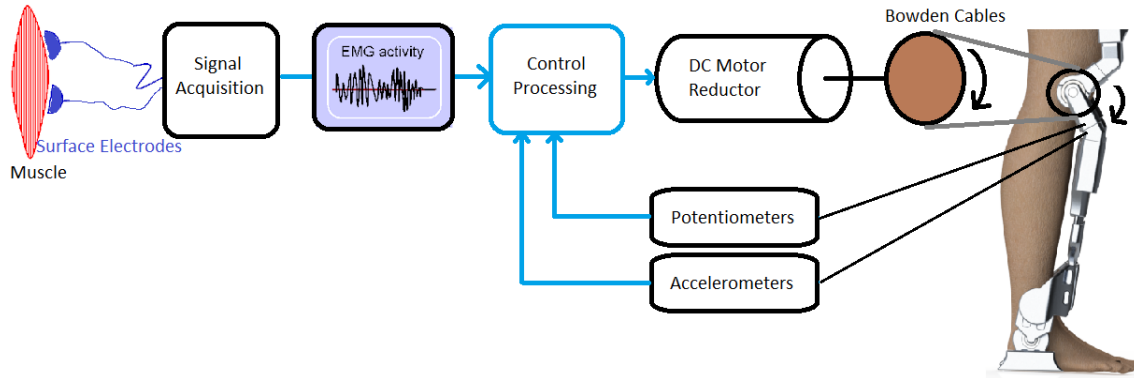


Figure 5.2: Diagram with a proposed control system using EMG, potentiometers and accelerometers. In this diagram, the location for the potentiometers and accelerometers is symbolic.

5.1.2 Experimental sEMG Results

During the laboratory trials for motion capture described in 4.2, the female subject wore surface EMG sensors for measuring muscle activation patterns. The sensors used were the "Trigno Wireless EMG" from "Delsys". Due to physical restrictions created by the weight placements described in Figure 4.2, the sensors could not be placed during the trials with weights.

The sensors were placed according to Figure 5.1, and the EMG data was synchronized with the respective motion data for gait, stair ascend/descend and sit/stand trials. For better understanding the muscle activations in the trials involving stair ascend and descend, the results are separated between stance (when the lower limb is in contact with the step and bears the weight of the human body) and swing (when the lower limb is moved forward).

Figure 5.3 shows the sEMG results for a gait trial. The movement is divided into sections "A" through "B1'".

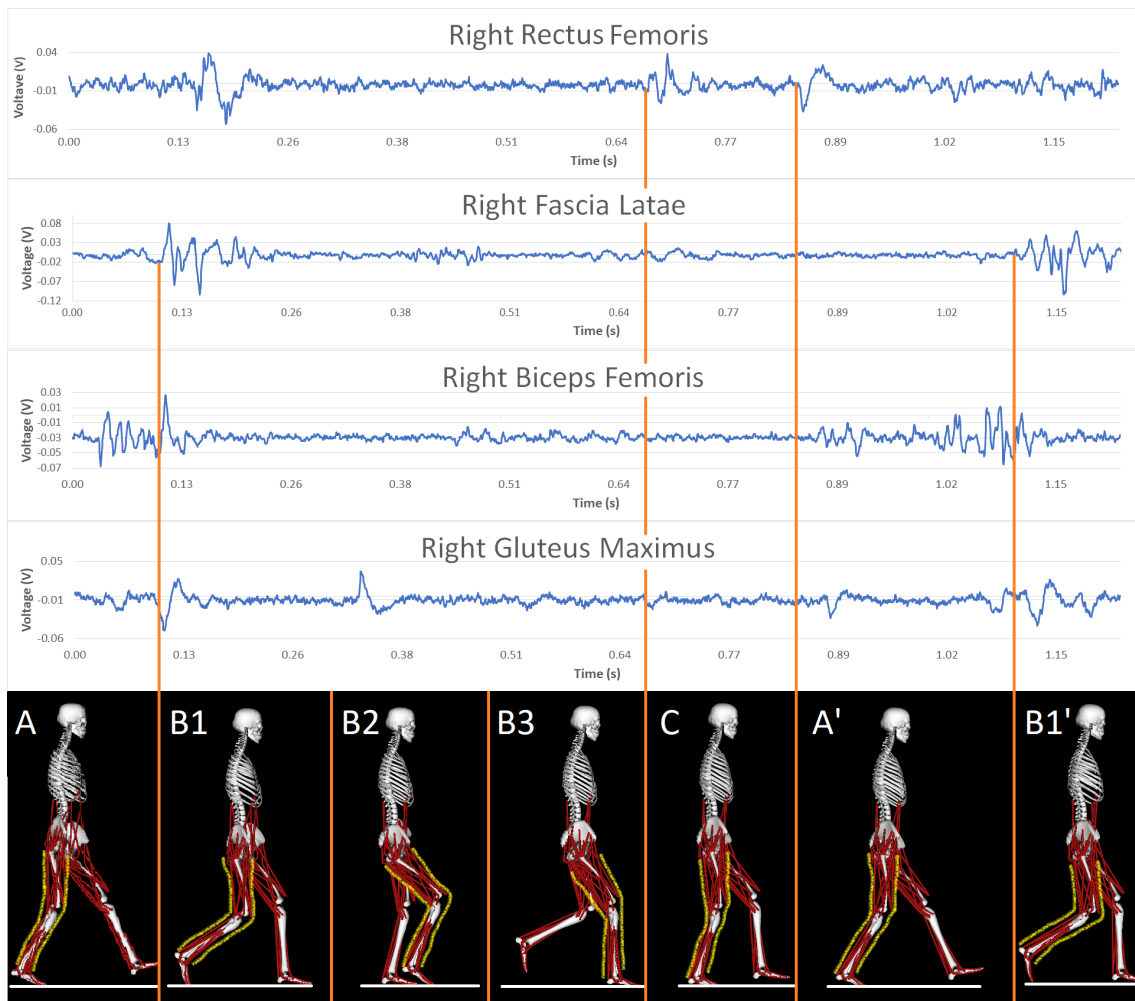


Figure 5.3: Results for the EMG data acquisition during normal gait for the right lower limb highlighted in yellow. The readings show results for the Rectus Femoris, Fascia Latae, Biceps Femoris and Gluteus Maximus. The EMG data is synchronized with the motion capture data, retrieved into OpenSIM.

Section "A" shows the end of the stance phase. Section "B1" shows the beginning of the swing phase, showing activations for the Fascia Latae and Rectus Femoris. Section "B3" ends the swing phase with the heel contact showing an activation for the Rectus Femoris. Sections "C" to "A'" show the stance phase with activations from Gluteus Maximus and Biceps Femoris. Section "B1'" shows the activity for Gluteus Maximus and Biceps Femoris ending with the Fascia Latae showing the beginning of a new swing phase.

Figure 5.4 shows the results for the stance phase during stair ascend, with the Rectus Femoris, Biceps Femoris and Gluteus Maximus muscles.

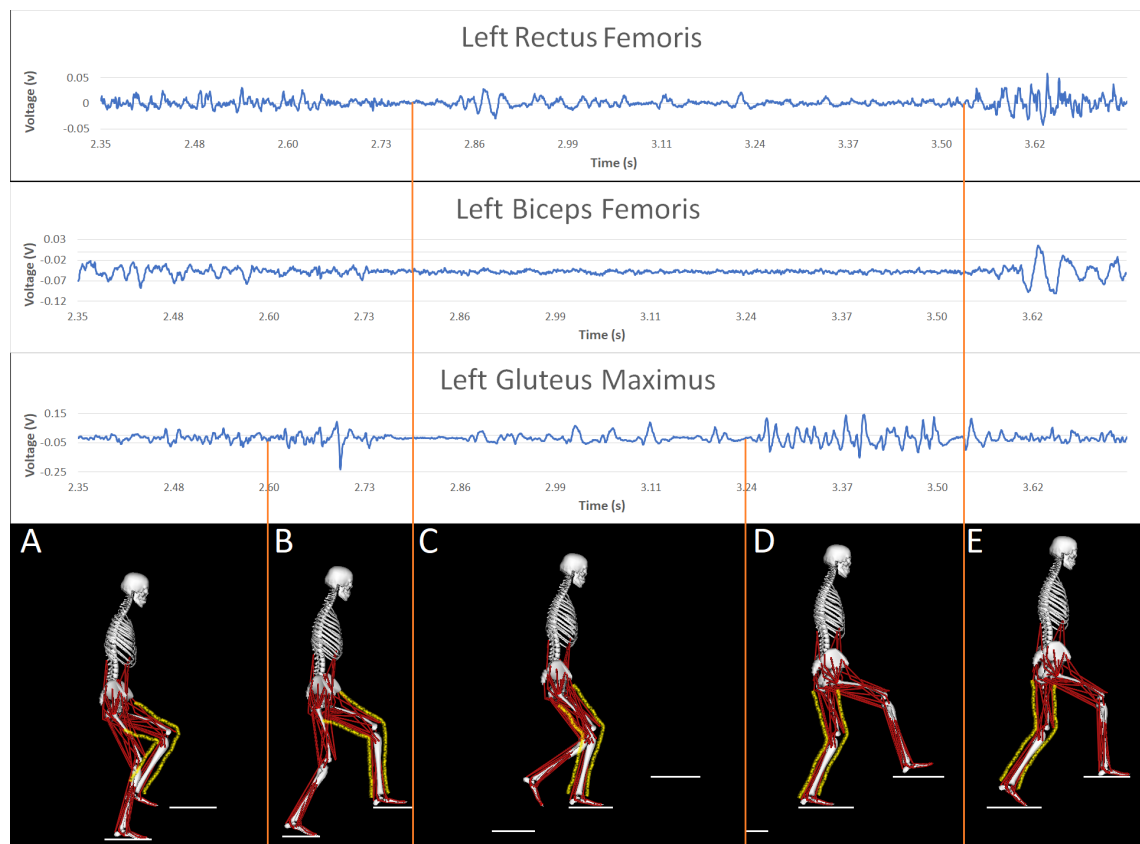


Figure 5.4: Results for the EMG data acquisition for the left lower limb highlighted in yellow, for stair ascending during the stance phase. The readings show results for the Rectus Femoris, Biceps Femoris and Gluteus Maximus. The EMG data is synchronized with the motion capture data, retrieved into OpenSIM.

The heel contact in section "B" shows a spike in activity for the Gluteus Maximus for hip extension. Section "C" shows activity in the Rectus Femoris for knee extension and continued activity in the Gluteus Maximus for hip extension. Section "D" shows more intensive activity in the Gluteus Maximus for hip extension. Section "E" shows a decrease in the Gluteus Maximus' activity when the hip is fully extended and an increase in the Rectus Femoris' activity for the final part of stage of the Stance phase.

The results for the swing phase during stair ascend are seen in Figure 5.5.

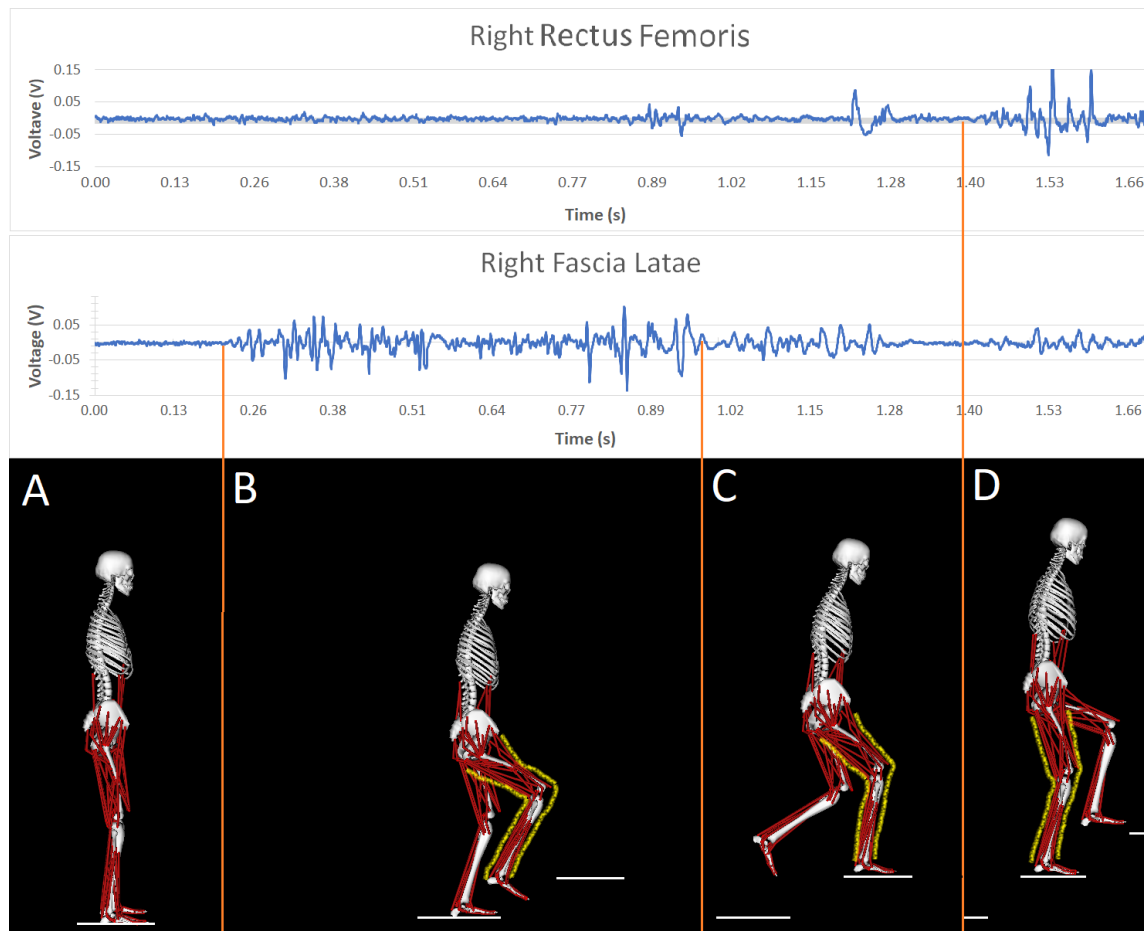


Figure 5.5: Results for the EMG data acquisition for the right lower limb highlighted in yellow, for stair ascending during the swing phase. The readings show results for the Rectus Femoris and Fascia Latae. The EMG data is synchronized with the motion capture data, retrieved into OpenSIM.

The subject starts in a stand-up stance in section "A". The beginning of the hip flexion in section "B" shows an activation of the Fascia Latae, which ends in section "D" when the right lower limb enters the stance phase.

Figure 5.6 shows the stance phase while descending a set of stairs.

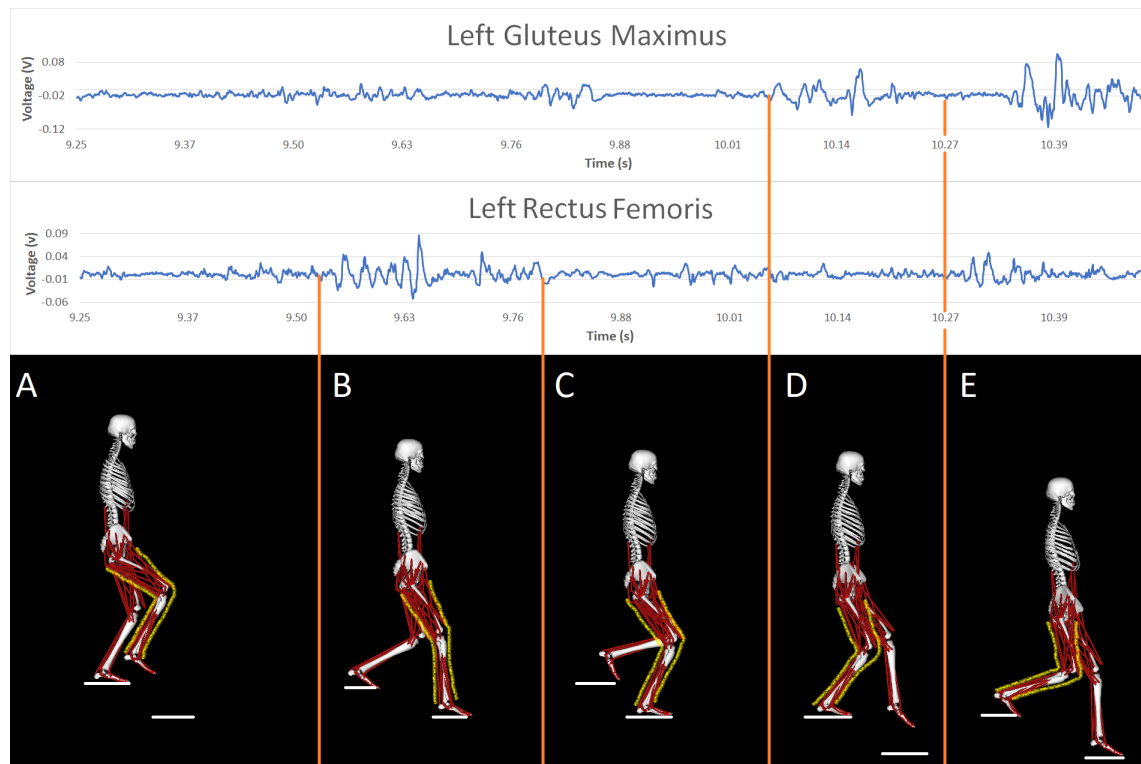


Figure 5.6: Results for the EMG data acquisition for the left lower limb highlighted in yellow, for stair descending during the stance phase. The readings show results for the Rectus Femoris and Gluteus Maximus. The EMG data is synchronized with the motion capture data, retrieved into OpenSIM.

The foot contact with the step is detected with through the activation of the Rectus Femoris in section "B". During section "D" the muscles show no relevant activity, and at the end of the stance phase in section "D", an activation of the Gluteus Maximus is observed.

The EMG results for the swing phase during stair descent are present in Figure 5.7.

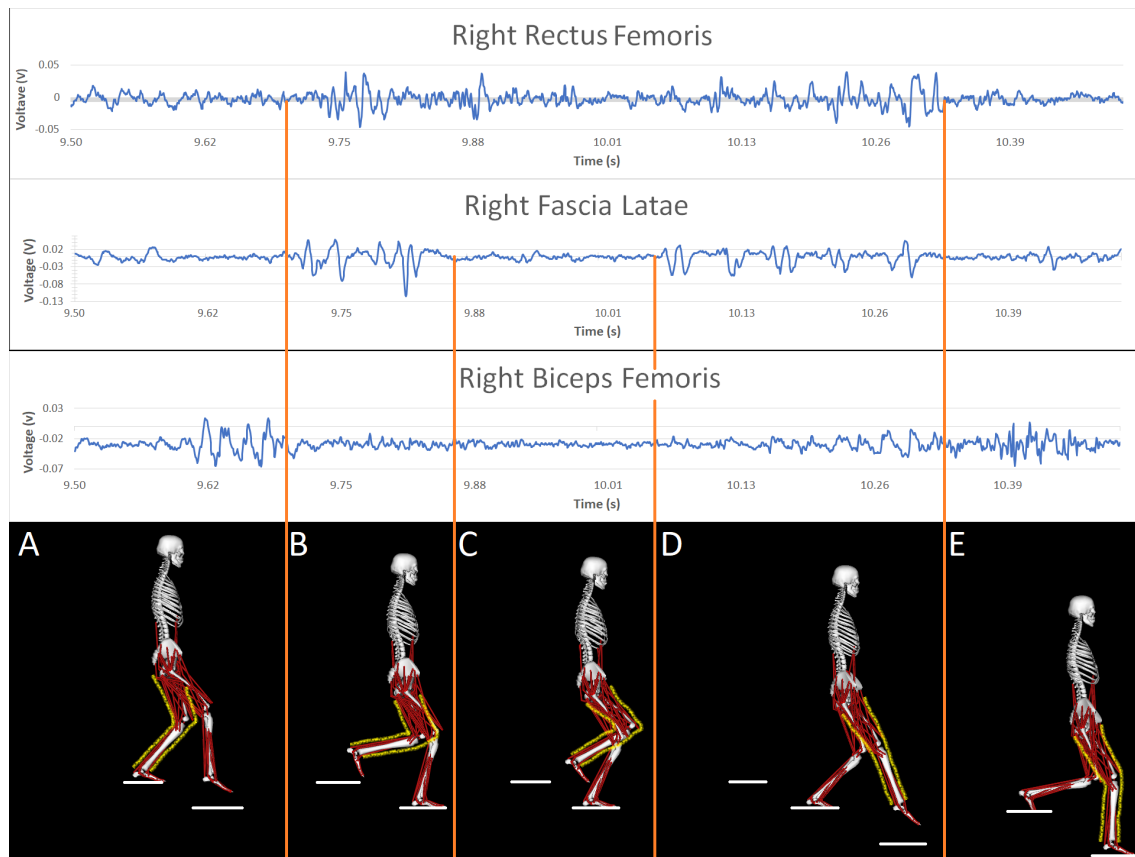


Figure 5.7: Results for the EMG data acquisition for the right lower limb highlighted in yellow, for stair descending during the swing phase. The readings show results for the Rectus Femoris, Fascia Latae and Biceps Femoris. The EMG data is synchronized with the motion capture data, retrieved into OpenSIM.

The beginning of the swing phase is seen in section "B" with the activation of the Fascia Latae and an increased activity in the Rectus Femoris. The Fascia Latae is activated again in section "D" together with the Rectus Femoris for both hip flexion and knee extension, in order to propel the lower limb towards the next step. Section "E" shows the transition to the stance phase, with an increased activity in the Biceps Femoris while the Rectus Femoris and Fascia Latae cease activity.

Figure 5.8 shows the movement of standing up from section "A" to "D", and sitting down from "D" to "A".

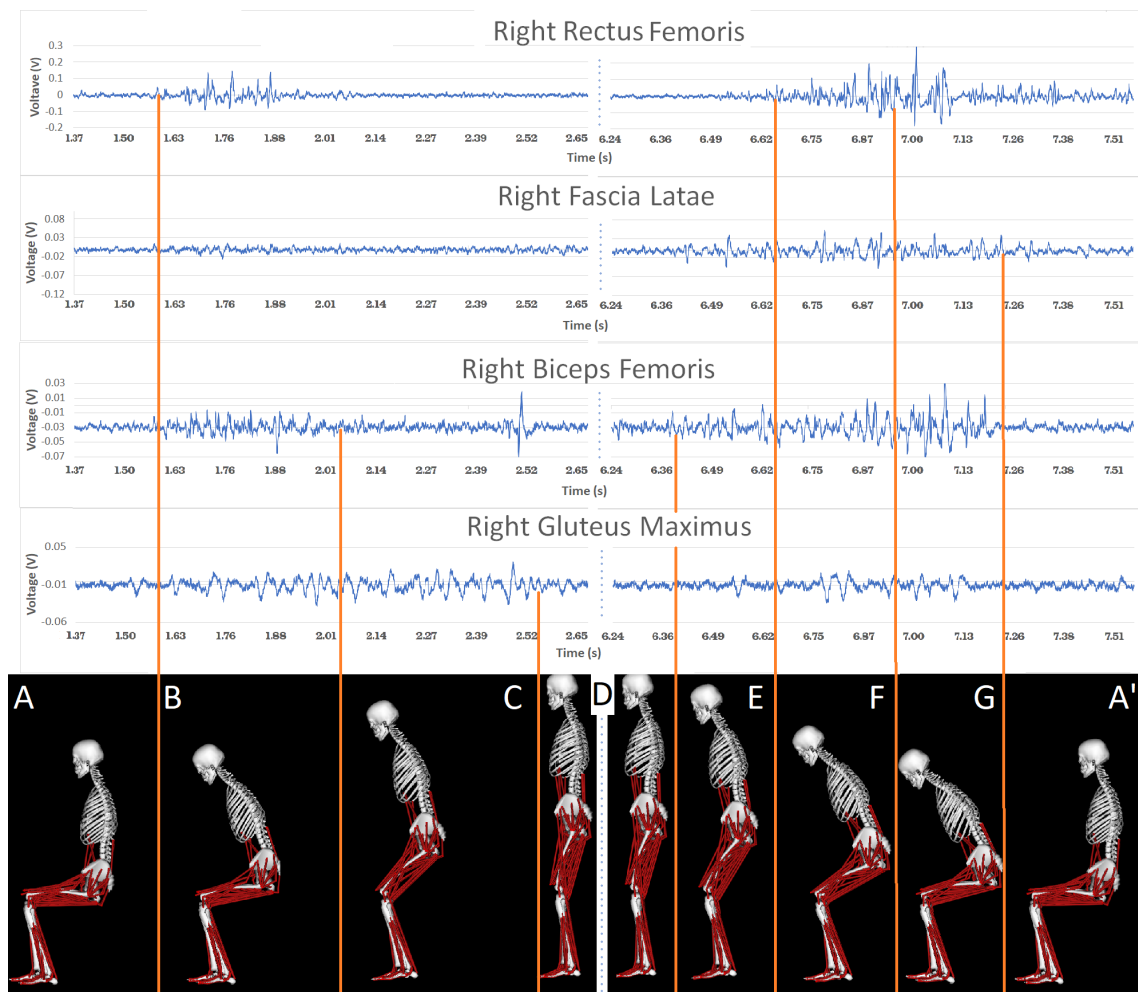


Figure 5.8: Results for the EMG data acquisition for the right lower limb, for stair standing and. The readings show results for the Rectus Femoris, Fascia Latae, Biceps Femoris and Gluteus Maximus. The EMG data is synchronized with the motion capture data, retrieved into OpenSIM.

The subjects starts in a sitting position in section "A". The beginning of the standing movement matches the activation of the Rectus Femoris and the Gluteus Maximus. The Rectus Femoris ceases activity during section "D", with continued activity from the Gluteus Maximus. Section "D" shows a standing and resting position. The sitting movement starts with activity from the Biceps Femoris seen in section "E". Sections "F" and "G" show continuous activity from the Rectus Femoris. Section "A'" shows a sitting and resting position.

The the separated activations for the analysed muscles are similar to the results expected in table 5.1. Therefore, it may be possible to develop a software control system that uses, as main input data, the sEMG signals from the Biceps Femoris, Gluteus Maximus Fascia Latae and Rectus Femoris.

5.1.3 Prototype for Sensor and Control Unit

For studying the control methods for exoskeleton prototypes in the future, a stand-alone sensor and control unit was developed.

Figure 5.9 shows a diagram with the proposed control system.

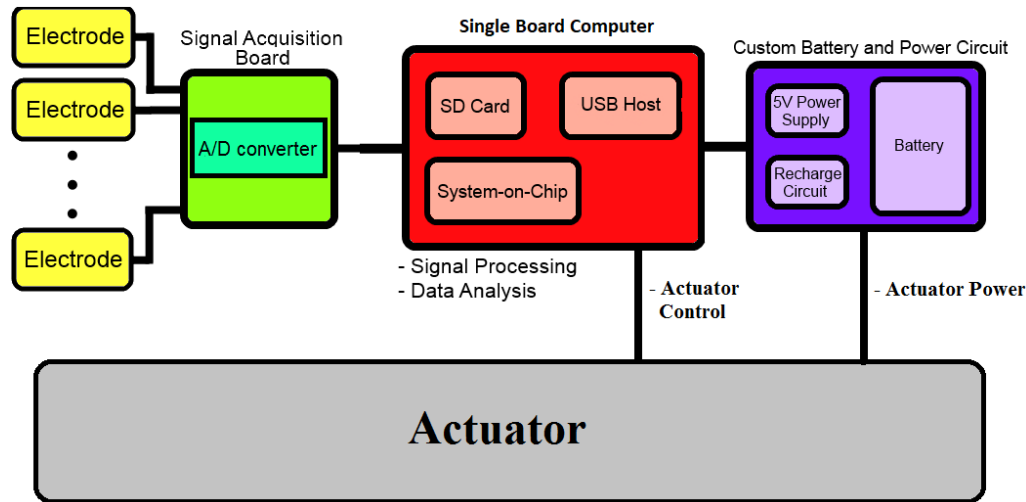


Figure 5.9: Simplified diagram with the proposed hardware for the control system.

The system's hardware is composed of an acquisition module, an Odroid-C1 single board computer from Hardkernel with a Wi-Fi dongle, a DC-DC power converter with batteries and wired sEMG electrodes. The hardware is fit inside a custom designed enclosure.

The acquisition module is a custom-made circuit capable of analyzing 8 different muscles and a reference electrode, at a 24bit resolution and 500 Hz sampling rate [95]. It also has a MEMS sensor with gyroscope and accelerometer. The Odroid-C1 has a quad-core Cortex A5 with 1GB of RAM with a compact Wi-Fi dongle using the Ralink RT5370 chipset for a 802.11n 2.4 GHz wireless communication. The Odroid-C1 connects to the acquisition module through an UART port at 500 kHz.

The system is powered by four 3.7V batteries connected to a DC-DC step-up converter, which provides a 5V power supply to the Odroid-C1 and allows charging the batteries through a micro-USB connector.

Figure 5.10 shows a picture of the system with the hardware components.

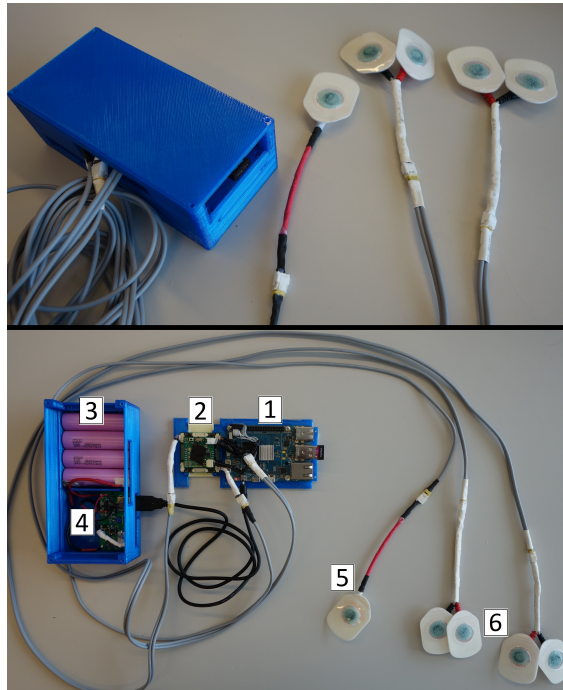


Figure 5.10: Picture of the several components in the functional prototype. 1 – Odroid C1 Mini-Computer; 2 – Data Acquisition Board; 3 – Batteries; 4 – Power Converter Board; 5 – Reference Electrode; 6 – Pairs of Differential Electrodes

The Odroid-C1 has the Ubuntu 14.04 operating system installed, thus it benefits from its large community support and stability.

All programming was done using Python version 2.7. Because the Python code is interpreted, the programs can run substantially slower than similar programs made using pre-compiled languages like C/C++. However, the simplified syntax and automatic memory management makes Python a much more appropriate language for prototyping and learning.

Also, for an academic research project, Python offers the advantage of being a simple to understand and widespread programming language. For example, it is relatively easy and fast to edit the program and add or alter functionalities while the system is being worn, ultimately saving time.

The developed software includes a module for data acquisition and parsing using *NumPy*, a digital filtering module using *Scipy* and a websockets communication module and webpage server using *Tornado*. The software also saves the results in compressed form with a time-date stamp into the SD card for storage and possible offline analysis.

This compact system has its own web server, so the sEMG readings can be easily accessed through a standard web browser from a multitude of devices (e.g. PC, smartphone, tablet), without the need for installing a third-party software using a dedicated PC.

Figure 5.11 shows the software architecture developed for the sEMG visualization.

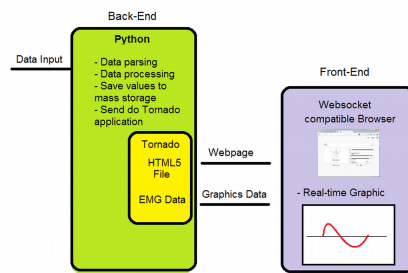


Figure 5.11: Simplified diagram of the software architecture for the portable system for wireless sEMG Monitoring

A webpage using standard HTML5 and Javascript code was developed and uses the “CanvasJS” charting library to render the sEMG readings in real-time. The webpage shows two sEMG readings from a total of 8 selectable channels in the form of continuously updated charts.

A 2-order Butterworth high-pass digital filter is implemented through SciPy with a cutoff frequency set to 20 Hz. This filter prevents the signal interference originated from body movement and other factors. The Butterworth filter was chosen for its flat response and low computational demands.

The program makes use of the “multiprocessor” module, to offload the digital filtering and web server functionality towards different CPU cores and free computational resources for the control program.

A picture of the system being used is seen in Figure 5.12

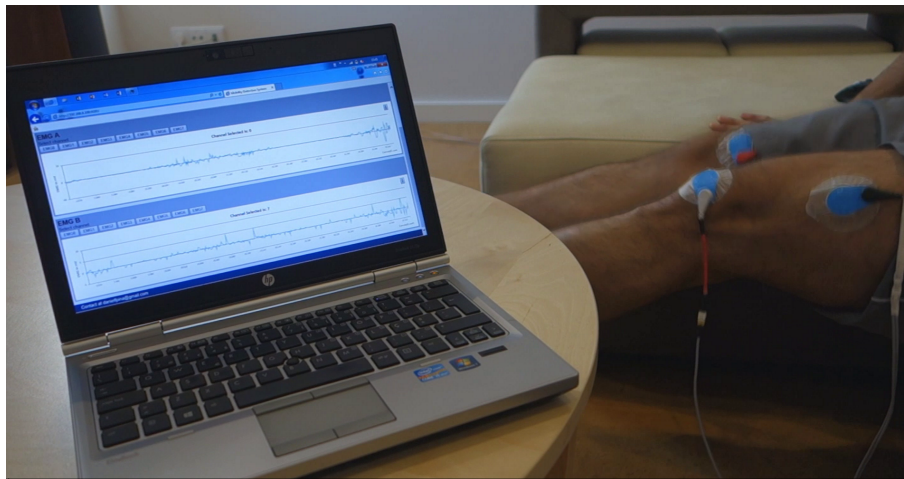


Figure 5.12: Picture of the developed system in use with electrode pairs on the Rectus Femoris in each leg, plus a reference electrode in the knee.

Because the system is portable, autonomous and provides a web page interface for signal monitoring, it can be used for other applications. Because there is no requirement for installing dedicated software in a dedicated computer, it can also be used as a cost-efficient surface EMG

equipment for clinics and hospitals. Figure 5.13 shows the web page with the real-time EMG readings using an Android smartphone with the included "Chrome" web browser.

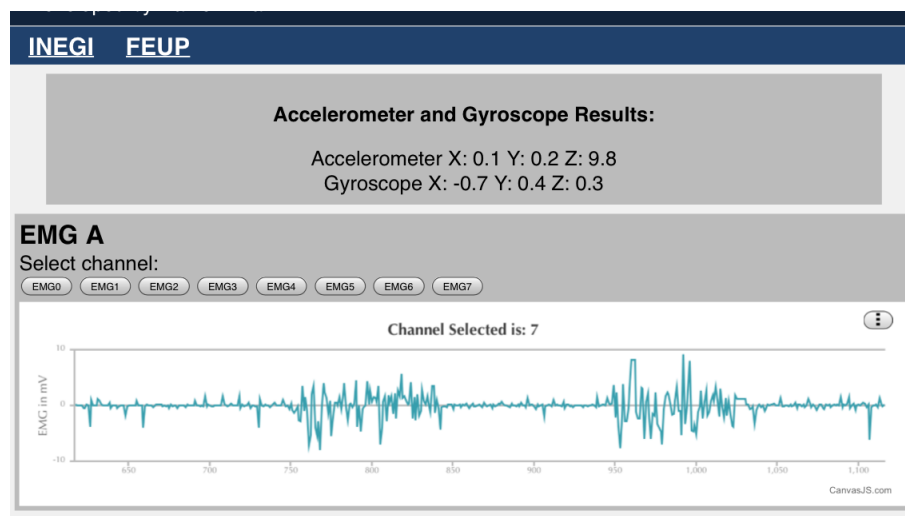


Figure 5.13: Screenshot of the HTML5 webpage with the EMG readings, taken from an Android smartphone.

Since the signal acquisition and processing is done in the portable unit, which also houses the web page, the system can also be used for telemedicine applications. Figure 5.14 shows a simplified diagram for a possible telemedicine consultation.

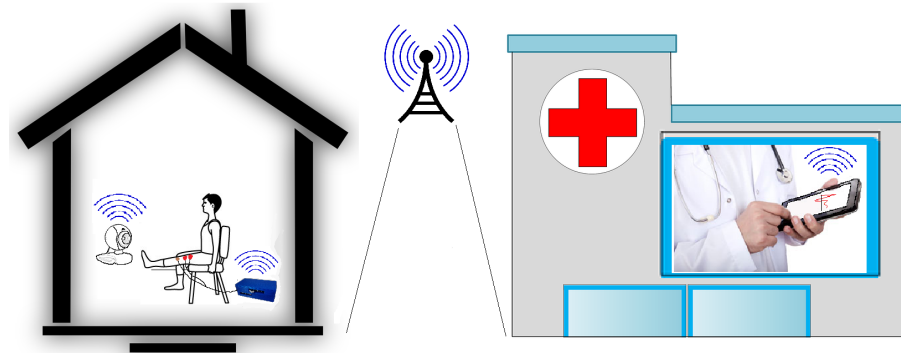


Figure 5.14: Diagram of a telemedicine consultation using the sensor system developed in the course of the project.

5.2 Actuation System

This section proposes an actuation system and a brief analysis of the chosen components. The components were chosen with the cost taken into consideration, as part of the objectives set in Chapter 2.

The following maximum torque values were obtained from the simulations in Chapter 4:

Hip Extension: **40 N.m**

Hip Flexion: **45 N.m**

Knee Extension: **100 N.m**

Knee Flexion: **16.5 N.m**

The choice of a brushless DC Motor involves evaluating the nominal torque, maximum torque and rotation speed. With the torque values known, it is now necessary to determine the rotation speed. Figure 5.15 shows the angular speeds recorded for the male and female gait trials without weight:

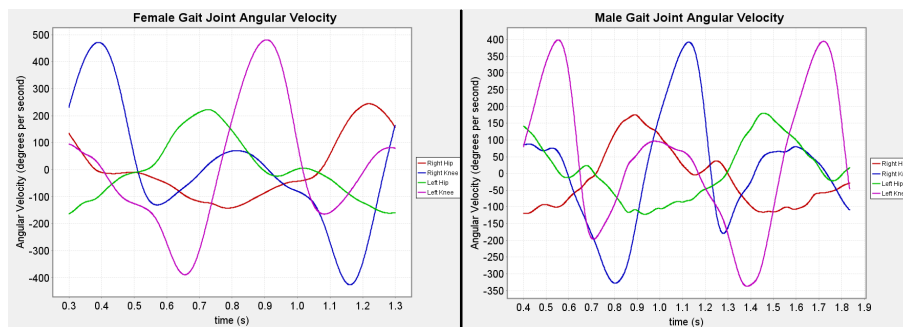


Figure 5.15: Angular velocities for the hip and knee joints recorded during the gait trials without weight.

The following maximum angular velocities are observed in the trial results:

Hip joint: 200 degrees per second = 33.33 Revolutions Per Minute (RPM); Knee joint: 480 degrees per second = 80 RPM.

To achieve an actuation system capable of the torque and angular velocity values shown above, a system composed of DC brushless motors and gearbox reducer is proposed. As an example, the following *off-the-shelf* components can be used:

DC Motor: Nanotec Brushless DB59M024035-A

- Nominal rotation speed: 3500 RPM (voltage controlled).
- Maximum rotation speed: 5000 RPM
- Nominal torque: 0.3 N.m
- Maximum torque: 1.11 N.m
- Nominal voltage: 24 V
- Nominal power: 135 W
- Weight: 0.65 kg

The proposed DC Motor is seen in Figure 5.16.

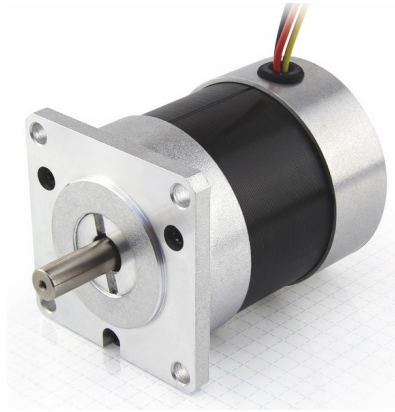


Figure 5.16: Picture of the proposed Nanotec Brushless DB59M024035-A DC Motor.

To reach the torque values at the given angular velocities the following reducers can be used:

For hip joint, Totel XSD-17-100-2UH-SP

- Reduction ratio: 100
- Repeated Peak Torque: 37 N.m
- Momentary Peak Torque: 71 N.m
- Average input angular velocity: 3500 RPM
- Maximum input angular velocity: 7300 RPM
- Weight: 0.46 kg

For the knee joint, Totel XSD-17-50-2UH-SP

- Reduction ratio: 50
- Repeated Peak Torque: 23 N.m
- Momentary Peak Torque: 48 N.m
- Average input angular velocity: 3500 RPM
- Maximum input angular velocity: 7300 RPM
- Weight: 0.46 kg

Using the mentioned components, the following output characteristics can be applied to the hip joint:

$$RotationSpeed = \frac{3500RPM}{100} = \mathbf{35\ RPM}$$

$$NominalTorque = 0.3N.m \times 100 = \mathbf{30N.m}$$

$$MaximumTorque = \mathbf{71N.m} \text{ (limited by the reducer output)}$$

For the knee joint , the following output characteristics apply:

$$RotationSpeed = \frac{3500RPM}{50} = \mathbf{70\ RPM}$$

$$NominalTorque = 0.3N.m \times 50 = \mathbf{15\ N.m}$$

$$MaximumTorque = \mathbf{48\ N.m} \text{ (limited by the reducer output)}$$

The total weight for 4 motors and 4 reducers is 4.44 kg. During the trials and simulations, the backpack where the actuation system is placed is set to weight 10 kg. This means the "weight budget" for the remaining components (compute system, batteries, cables and mechanical support system) is approximately 6.5 kg.

The torque output from the reducers is planned to be applied to the exoskeleton frame to selected components through bowden cables, as mentioned in Chapter 3. A study on this application is planned for future works.

These solutions can be applied to all movements except knee extension during stair descend. Since this torque output is applied during negative work (torque direction is opposite to the movement), a solution involving controllable brakes may be more adequate.

5.3 Conclusions

The placement for the EMG electrodes is determined, based on muscle activation and proper positioning guidelines. The electrode placement for an actuator control based on surface EMG was validated through experimental results.

A functional prototype for sensory system and computational unit capable of acquiring EMG data and controlling the exoskeleton is built and successfully tested. The system is capable of acquiring 8 EMG signals, which is the requirement previously established to gather sufficient data to properly control hip flexion/extension and knee flexion/extension during daily activities.

The system can be controlled and the data can be seen remotely and in real-time, which is relevant for saving time when developing the control for a functional exoskeleton prototype. It also uses an accessible programming language to reduce development time. Although these features are less relevant to the scope of the project mentioned in the thesis, the functional prototype for portable and autonomous sEMG signal acquisition with an embedded web server can be explored for other uses, such as telemedicine applications.

A preliminary study on possible actuators is also made, as grounds for a thorough work in the future.

Chapter 6

Conclusions and Future Work

6.1 Project Achievements

A custom-made database to gather information about the state-of-the-art was developed and described in Chapter 1. Along with this work, a necessity was identified, as the number of elders with mild to moderate mobility impairment in the European Union is large enough to justify the development of an assistive exoskeleton focused solely on that population.

The focus of the project was to develop an exoskeleton for the lower limbs to assist 3rd age users with medium mobility impairment. During the course of the work, the project established a series of steps to take when developing an assistive exoskeleton with human interaction.

The mechanical design for the support frame was developed through a cycle that included a thorough anatomical and ergonomic study using anatomically correct 3d models of human bodies. This approach is so far unprecedented, as most mechanical frames from current lower-limb exoskeletons are essentially made up of straight bars laterally attached to the outside of the legs, which renders them inevitably large and eye-catching.

The development cycle established in this worked allowed to design a mechanical frame that stays close to the human body and is able to adapt to different body shapes. The result is an exoskeleton frame that could be used beneath large clothes, which may be a very important factor when taking into account the social awkwardness and concerns from its users.

The laboratory trials and posterior multi-body simulations were used to establish the torque values of the exoskeleton when taking into consideration the additional weight from the exoskeleton itself. These results can now be used in the future for more elaborate studies on actuation systems. A preliminary study on a set of actuators compatible with the torque requirements is shown in Section 5.2.

A portable control and sensor unit was also developed to use on prototypes. The software in the custom sensor and control was developed specifically to speed up the research on future exoskeleton prototypes, as it supports the number of EMG channels required to control the proposed exoskeleton.

The system uses low-cost materials and off-the-shelf components, which may result in an exoskeleton that is accessible to most households.

Throughout the time period of the work presented in this thesis, the author published 2 articles in international journals, 3 articles/presentations in international conferences and 4 articles/presentations in national conferences.

6.2 Project Limitations

The simulations performed in Chapter 3 are important to perform during the early stages of development, but they are not as reliable as real-life trials. They can be used as a means to guide the development, but not as definitive results to apply in a final product.

Some components from the mechanical frame were not designed and therefore not included in the simulations, such as additional components to hold the backpack into the belt, components to hold and drive the bowden cables and heel-to-floor components that allow a variable distance change.

More simulations including a larger variety of forces and torques could be performed, however a significantly larger number of variations could have diminishing returns for the above reasons.

The opensim software and specifically the RRA function were observed to work on very limited time windows. As the multibody system grows more complex (particularly human and exoskeleton attached) and the motions involve tridimensional displacements (e.g. stair ascend/descend trials), the movement changes applied by the algorithm end up creating large displacement errors and larger residual forces. For the more complex simulations, the RRA results would become unstable after close to 1 second of simulation, which is not enough to include a full stair ascend/descend cycle.

6.3 Future Work

The next phase of development for the exoskeleton is to develop a proper mechanical project to provide tolerance and precision levels of the exoskeleton, to produce a full-scale prototype. Some additional components also need to be developed, such as the backpack and its adapters, bowden cable holders and the heel-to-foot components.

The information from the RRA results can now be used to perform muscle studies in OpenSim, and the software also allows to implement virtual actuators that can be used in the simulations. This study would provide a more precise information on which muscles will be "relieved" during the exoskeleton assistance and also which muscles may be overly stressed, such as the muscles responsible for hip rotation, adduction/abduction and heel plantar flexion.

The study for the actuation system is preliminary, and can be thoroughly expanded with actuator behavior, predicted power consumption and noise output. The rather fast rotation speed of the DC Motors used may induce noise levels that make them non-viable for use in the exoskeleton.

With a hardware platform for sensor and control having been developed and the actuation system also studied, it is now possible to start researching control algorithms for the DC Motors.

The project was developed with a scope directed at elders who suffer from moderate mobility impairment. This assumes that the target user has a fully functional motor control and balance, and only lacks in general lower limb strength. These factors allow for a control system using surface EMG. However, with further research into the control subsystem, the proposed exoskeleton may be useful for younger patients who suffer from early stages of neurodegenerative diseases, such as Parkinson's and Multiple Sclerosis.

Appendix A

Publications

A.1 Journal Publications

Daniel Pina, Antonio Augusto Fernandes, Renato Natal Jorge, Joaquim Gabriel, "A statistical study regarding the state-of-the-art of active motion-oriented assistive devices", Biomedical Engineering Applications Basis and Communications, 2014, DOI: 10.4015/S1016237214500471

Daniel Pina, Antonio Augusto Fernandes, Renato Natal Jorge, Joaquim Gabriel, "Designing the mechanical frame of an active exoskeleton for gait assistance", Advances in Mechanical Engineering, Vol.10, Issue 2, 2018, DOI: 10.1177/1687814017743664

A.2 Conference Publications

International Conferences

Daniel Pina, Antonio Augusto Fernandes, Renato Natal Jorge, Joaquim Gabriel, "A Technological and Statistical State-of-the-Art Study Regarding Active Motion-Oriented Assistive Devices", International Conference on Biomedical Electronics and Devices "Biodevices", 2012

Daniel Pina, Antonio Augusto Fernandes, Renato Natal Jorge, Joaquim Gabriel, "Development of a Portable System for Online EMG Monitoring", Online Experimentation "Exp.At", 2015

Daniel Pina, Antonio Augusto Fernandes, Renato Natal Jorge, Joaquim Gabriel, "Development and Testing of a Compact EMG Monitoring Device for Telemedicine Applications", European Congress of Physical and Rehabilitation Medicine, 2016

National Conferences

Daniel Pina, Joaquim Gabriel, Renato Natal Jorge, Antonio Augusto Fernandes, António Moreira, "Desenvolvimento de um Sistema de Orientação Dinâmica para um Atuador Fluídico Flexível", 5º Congresso Nacional de Biomecânica, 2013

Daniel Pina, Renato Natal Jorge, Joaquim Gabriel, Antonio Augusto Fernandes, "Desenvolvimento de um Sistema de Aquisição de Sinal para Controlo de um Exosqueleto Ativo de Reabilitação", 6º Congresso Nacional de Biomecânica, 2015

Daniel Pina, Renato Natal Jorge, Joaquim Gabriel, Antonio Augusto Fernandes, "Mechanical Design of a Low-Profile Assistive Exoskeleton", Mechanical Engineering Conference "CEM", 2016

Daniel Pina, Antonio Augusto Fernandes, Renato Natal Jorge, Joaquim Gabriel, "Desenvolvimento do Suporte Mecânico de um Exosqueleto Ativo para Assistência da Marcha", 7º Congresso Nacional de Biomecânica, 2017

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