

FACULDADE DE ENGENHARIA DA UNIVERSIDADE DO PORTO

Human-Robotic manipulator's electromechanical interface

Gonalo Vieira Greg3rio da Fonseca e Costa



Mestrado em Engenharia Eletrot3cnica e de Computadores

Supervisor: Professor Doutor H3lio Mendes de Sousa Mendona

Second Supervisor: Professor Doutor Manuel Fernando dos Santos Silva

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Resumo

Esta dissertação apresenta o desenvolvimento de uma interface do tipo *end effector* destinada à reabilitação do membro superior em doentes pós-AVC, concebida para ser acoplada a um manipulador robótico colaborativo. O principal objetivo consistiu na conceção e validação de um *end effector* capaz de estabelecer interface com o braço, antebraço e mão humanas, permitindo simultaneamente o controlo ao nível articular do punho e do cotovelo. O sistema integra componentes mecânicos, elétricos e de software numa arquitetura unificada, preparada para integração com manipuladores robóticos de utilização médica.

O acidente vascular cerebral continua a ser uma das principais causas de incapacidade permanente em adultos a nível mundial, sendo o défice motor do membro superior um fator determinante na perda de autonomia e na diminuição da qualidade de vida. A crescente necessidade de sessões de reabilitação intensivas e repetitivas exerce uma pressão significativa sobre os sistemas de saúde, reforçando a necessidade de soluções robóticas capazes de apoiar os terapeutas, mantendo simultaneamente a eficácia terapêutica.

O sistema desenvolvido permite a atuação ativa da flexão–extensão do cotovelo e da pronção–supinação do punho, possibilitando a implementação de estratégias terapêuticas ativas, assistivas e resistivas. A estrutura mecânica foi desenhada de modo a equilibrar rigidez estrutural, adaptabilidade ergonómica e simplicidade de implementação. Foi ainda desenvolvida uma arquitetura de controlo embebido personalizada, permitindo o controlo de atuação em tempo real e assegurando condições de interação segura entre o utilizador e o sistema robótico.

Foram implementados e avaliados experimentalmente três modos de controlo (ACTIVE Assist, ACTIVE Resist e ASSIST-AS-NEEDED). A validação progressiva, incluindo ensaios sem carga, testes com a estrutura completa e testes com utilizadores, confirmou a estabilidade do comportamento de atuação, a fiabilidade na execução de comandos, a compensação eficaz de cargas e a correta implementação de mecanismos de segurança. Os resultados demonstram um desempenho consistente e repetível em todos os modos implementados, confirmando que o protótipo desenvolvido cumpre o seu papel operacional enquanto interface controlável para reabilitação do membro superior.

Embora a integração completa com o manipulador robótico e a realização de ensaios clínicos com doentes pós-AVC constituam etapas futuras, o trabalho apresentado estabelece um protótipo validado e uma base técnica robusta para a evolução do sistema rumo a uma solução de reabilitação robótica com potencial aplicação clínica.

Abstract

This dissertation presents the development of an end effector interface designed for upper limb rehabilitation in stroke patients, attachable to a collaborative robotic manipulator. The primary objective was to design and validate an end effector capable of interfacing with the human arm, forearm, and hand while enabling joint-level control of the wrist and elbow. The system integrates mechanical, electrical, and software components into a unified architecture capable of integration with medical robotic manipulators.

Stroke remains one of the leading causes of long-term adult disability worldwide, with upper-limb motor impairment significantly affecting independence and quality of life. The increasing demand for intensive and repetitive rehabilitation exercises places substantial pressure on healthcare systems, creating a need for robotic solutions capable of supporting therapists while maintaining therapeutic effectiveness.

The developed system enables active actuation of elbow flexion–extension and wrist pronation–supination, allowing implementation of active, assistive, and resistive therapeutic strategies. The mechanical structure was designed to balance rigidity, ergonomic adaptability, and implementation simplicity. A custom embedded control architecture was developed to enable real-time actuation control and safe human–robot interaction.

Three control modes (ACTIVE Assist, ACTIVE Resist, and ASSIST-AS-NEEDED) were implemented and experimentally evaluated. Progressive validation, including no-load and full-structure testing and user testing confirmed stable actuation behaviour, reliable command execution, effective load compensation, and correct safety enforcement. The results demonstrate consistent and repeatable performance across all implemented modes, confirming that the developed prototype fulfils its intended operational role as a controllable upper-limb rehabilitation interface.

Although full integration with the robotic manipulator and clinical testing with stroke patients remain future steps, the presented work establishes a validated prototype and a robust technical foundation for further development toward a clinically deployable robotic rehabilitation system.

United Nations' Sustainable Development Goals

The work towards the United Nations' Sustainable Development Goals has become one of the most pressing issues globally, striving to enable these goals in the shortest timespan possible is required of developed countries and companies. A contribution to Goal 3 of the United Nations' Sustainable Development Goals (*Ensure healthy lives and promote well-being for all at all ages*), specifically, target 3.8. (*Achieve universal health coverage, including financial risk protection, access to quality essential health-care services and access to safe, effective, quality and affordable essential medicines and vaccines for all*) is anticipated from this dissertation.

This goal was chosen due to the fact that the work being developed in this dissertation is intended to have a direct impact on patient's recovery efficiency and effectiveness with post stroke disability. It is also of notable consideration how a recovery with improved efficiency and effectiveness rate is expected to directly contribute to the patient's well-being and good health, while also ensuring access to a quality service for rehabilitation.

Table 1: Project Contribution to the United Nations' Sustainable Development Goals

SDG	Target	Contribution
3	8	In the development of this dissertation, it is intended to have a direct impact on patient's recovery efficiency and effectiveness with post-stroke disability.

Acknowledgments

I dedicate wholeheartedly this project and degree to my grandfathers who inspired me to pursue engineering.

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“Not everything that can be counted counts, and not everything that counts can be counted.”

- William Bruce Cameron

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Abbreviations & Symbols

AAN	Assist-as-needed
CIMT	Constraint-induced movement therapy
COBOT	Collaborative Robot
DALYs	Disability Adjusted Life Years
DoF	Degrees of Freedom
FEUP	Faculdade de Engenharia da Universidade do Porto
FOC	Field-Oriented Control
GRADE	Grading of Recommendations Assessment, Development and Evaluation
PAMs	Pneumatic Artificial Muscles
QoL	Quality of Life
SEAs	Series Elastic Actuators
UI	User Interface
VSAs	Variable Stiffness Actuators

Chapter 1

Introduction

In this chapter the context, motivation, goals and structure of this dissertation will be covered. The contextual focus will be on stroke and its causes and consequences, the motivation provides the reasoning for the development of an end effector for rehabilitation in stroke patients, and the goals showcase clear objectives necessary to achieve in the development of this dissertation. Finally, a small section will be focused on the structure presented in this document.

1.1 Context

A stroke is a syndrome of neurological deficit attributed to vascular injury (infarction, haemorrhage) of the central nervous system [1]. Stroke has been recognised as the second leading cause of death worldwide and the third leading cause of death and disability combined, Disability Adjusted Life Years (DALY). The DALY is a metric that merges the years lost due to early death as well as the years lived with not optimal health, commonly known as disability [2]. As of the latest available data (2019), approximately 12 million new stroke cases are recorded every year, and 6.35 million deaths from stroke are registered. Additionally, there are 101 million prevalent cases of stroke. An increase of 70% in the global incidence of stroke was observed from 1990 to 2019, and the prevalent cases of stroke have increased by 85%. Table 1.1 displays data related to stroke [3].

Table 1.1: Absolute number of incident and prevalent strokes, deaths from stroke and DALYs due to stroke in 2019, and percentage change globally for 1990–2019 [3]

Statistic Types	Stroke in Absolute Numbers (millions)	Percentage Change (1990-2019)
Incidence (95% UI)	12.22	70%
Deaths (95% UI)	6.35	43%
Prevalence (95% UI)	101.47	85%
DALYs (95% UI)	143.23	32%

The incidence of stroke cases is associated with risk factors. Particularly, high systolic blood pressure, high particulate matter pollution, high body mass index, high fasting plasma glucose and smoking are among the leading risk factors, with the top 19 risk factors being responsible for 87% of total stroke DALYs [3].

The vast majority (62%) of strokes are ischaemic, mostly caused by small vessel arteriosclerosis, cardioembolism and large artery atherothromboembolism. With the remaining cases resulting from a form of haemorrhage, mostly intracerebral or subarachnoid. [1]. Figure 1.1 represents an illustration of risks, causes and mechanisms of stroke.

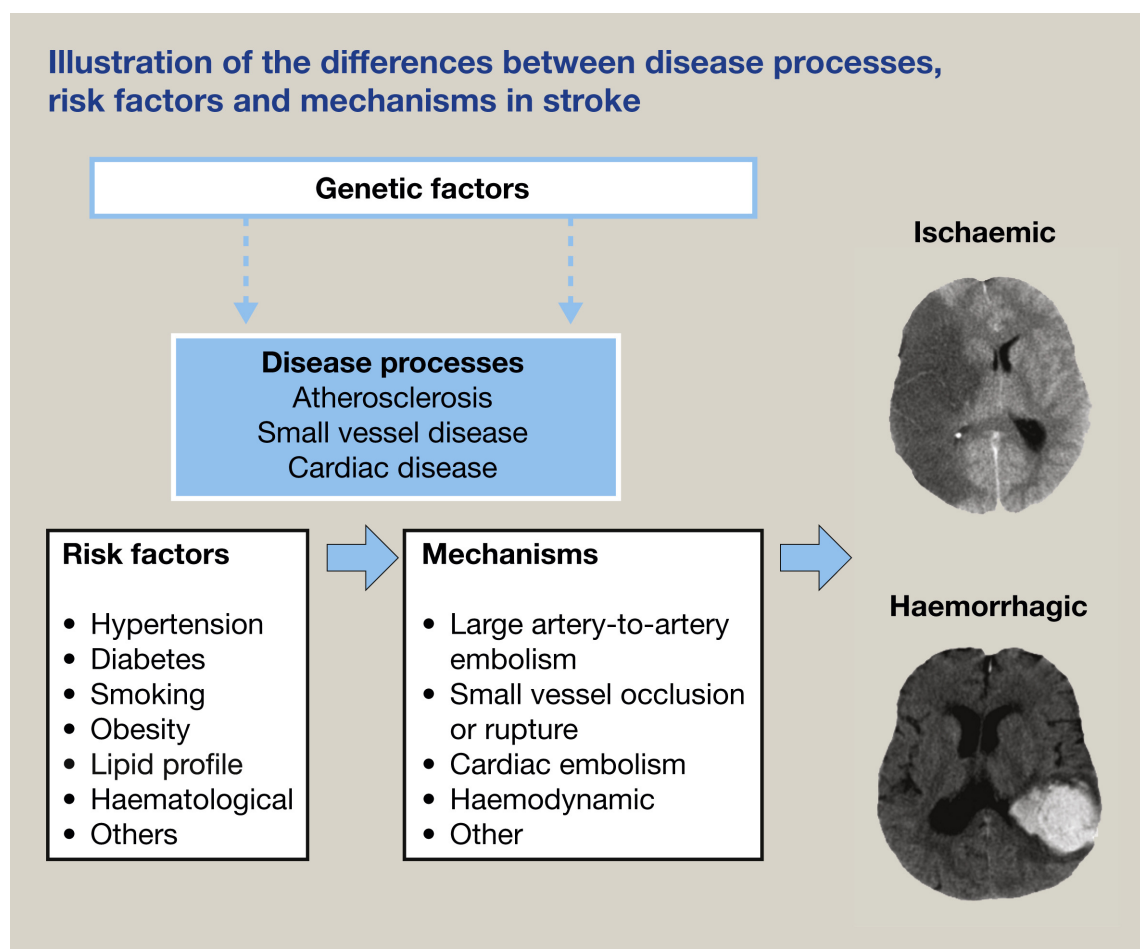


Figure 1.1: Illustration between disease processes, risk factors and mechanisms of stroke [1]

As mentioned above, stroke is one of the leading causes of adult disability [3]. The disability associated with stroke usually persists beyond the acute recovery period. This is the phase in which the patient is under intensive medical evaluation and care immediately after experiencing the stroke until stabilisation [4].

In the aftermath of a stroke, Valery L. Feigin et al. [3] reported that around 75% of patients report two to five continuous unmet needs following hospital discharge [3]. Namely, cognition, fatigue, emotion/mood, leisure/hobbies, mobility, pain and employment. Specifically, related to mobility, most people report deficits related to hemiparesis, upper-limb dysfunction, pain and

impaired balance [5]. These sequelae, which vary from neurological, physiological, physical and social deficiencies, have a significant impact on the patient's quality of life. It is estimated that 75% of stroke survivors require some form of rehabilitation after hospital discharge [6]. Hemiparesis, the weakness or inability to move one side of the body, is present in around 80% of stroke patients [7].

Stroke leaves many patients with issues related to upper limb motor function, and most improvements come in the first months after stroke. However, within 3 months 55%-78% of patients in rehabilitation still had yet to achieve age-specific norms for upper-limbs function [8] [9]. And even up to 4 years after experiencing a stroke, half of patients still report issues with arm function [10].

The growth of the stroke burden and prevalence over incidence has created the need for more rehabilitation resources for patients who lost physical motor capacity as a result of stroke [3]. In this dissertation, the focus will be on the motor function recovery segment of rehabilitation, specifically upper limb rehabilitation. The increased necessity of resources to rehabilitate stroke patients has created an opportunity for the integration of automated systems to reduce the workload placed on healthcare professionals and increase the availability and efficiency of motor recovery treatment [11]. Several studies and devices have been developed and deployed for that effect.

The aim of this dissertation is to develop an end effector that, in combination with a collaborative manipulator, automates physiotherapy exercises to restore mobility in the upper limbs.

1.2 Motivation

The mobility and dexterity of the upper limbs in humans are considered some of the features that distinguish humans from most animals. It is among the most critical motor functions needed to live one's daily life and perform their jobs. This implies its recovery is of the utmost importance to improve patients' quality of life and the quality of life of those surrounding them, as well as reduce their burden on society and healthcare systems.

With the growing incidence of stroke since 1990 [3], the demand for rehabilitation care has also been increasing [11]. Increased pressure has been placed on healthcare professionals and resources due to the growing incidence, with ever more people being affected by post stroke disability. Recovery has also become more important to help them integrate into society as a useful member of it.

The automation of certain repetitive recovery exercises with the help of robotic manipulators is intended to help diminish the amount of time spent per patient that each healthcare worker would require. Research on robotic-assisted stroke recovery demonstrates an improvement in the efficacy of post-stroke treatment, with patients undergoing robotic-assisted therapy showing a 6-7% increase in upper limb motor control capacity [12].

1.3 Goals

The development of an electromechanical end-effector for a collaborative manipulator (KUKA iiWA 14 R820) is the main goal of the dissertation. The adaptation of the KUKA iiWA collaborative manipulator for physical rehabilitation of the upper limbs in recovering stroke patients is required. The end effector is designed to interface with the upper limb of a patient and help perform a set of repetitive exercises in order to emulate and enable the patient to do the required movement. In order to achieve the required goal, the end-effector will also require active actuation for its movement in order to ensure the patient is able to fully complete the required exercises and movements of the upper limb in case the patient is unable to do the complete movement. As well as passive and semi-passive actuation in order to allow a more able patient to perform the required exercises utilising their own capacities to the fullest.

To achieve this objective, the development is required to have several complementary goals to complete the main goal. These can be summarized as:

- Design and 3D modelling of a electromechanical interface to interact with the patient's limb, ensuring ergonomics and comfort;
- The creation of a control system to actuate the electrical component of the system;
- Creation and testing of prototype units to ensure compliance with requirements; results will lead to production of final prototype;
- Production of final prototype design and final testing validation;
- Integration of final system with collaborative manipulator.

1.4 Dissertation Structure

The dissertation features 6 chapters and employs the following structure:

The **Introduction** chapter presents the context and background that led to the development of the dissertation, describing the motivation, rationale, and goals to be achieved. It also provides an overview of the structure of the dissertation. The **Theoretical Framework** chapter is dedicated to presenting the theoretical concepts and foundations that will support the Literature Review and the subsequent development work. In the **Literature Review** chapter, previous works, articles, dissertations, theses, and other scientific documents related to the subject of this dissertation are reviewed to highlight the results already achieved in the field. The **Methodology** chapter documents the methods employed in the development of the system. The **Tests & Results** chapter describes the tests conducted to validate the work and presents the data obtained from these tests. Finally, the **Conclusion** chapter summarises the results achieved and discusses the extent to which the goals set at the beginning of the dissertation have been met.

Chapter 2

Theoretical Framework

All the necessary theoretical concepts and definitions required for the understanding and clarity of the developed dissertation will be provided in the following chapter, specifically the concepts related to manipulators, robotics and rehabilitation.

This chapter develops the necessary theoretical details of the components of the developed dissertation. Specifically, the different types of manipulators, actuators and end effectors that are usually utilised in the field of rehabilitation robotics. A review of the most commonly employed electric motor actuators is also developed, along with brief technical expositions of the most common communication and control methods for actuation.

2.1 Stroke Rehabilitation

In order to properly understand the desired end goal of the project developed in this dissertation, a small review of the most effective and common types of rehabilitation practices needs to be studied. As such, this section is dedicated to reviewing common practices for stroke rehabilitation.

Stroke rehabilitation is an ever evolving area of research and development and is constantly being analysed for the most effective treatment methods. However, the research into this field still lacks a substantial quantity of high-quality evidence regarding what are the most effective conventional therapy methods and exercises to yield the best results, with the knowledge and judgement of each health professional still being the main decision factor. Nevertheless, there are significant studies on treatment efficacy based on moderate-quality evidence. Such studies have found that a few methods have gathered some evidence of being beneficial to the recovery of upper limb motor function. Some of these are: CIMT, mental practice, mirror therapy, interventions for sensory impairment, virtual reality, high quantity of repetitive exercises, intensive comparison therapy, robotic intervention [13][14][15].

Some evidence shows that a greater amount of intervention provides better results than lower amounts, but there is lacking evidence and research on the optimal amount of intervention for rehabilitation as [Figure 2.1](#) [13][14][15].

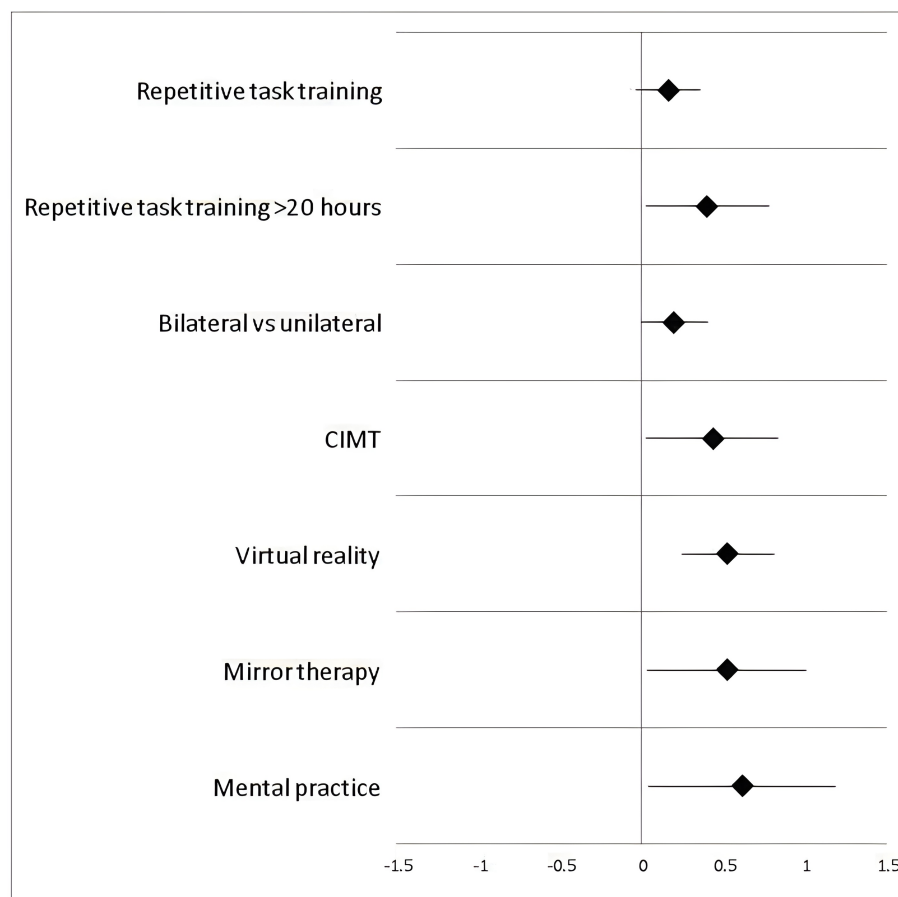


Figure 2.1: Moderate level GRADE evidence for the efficacy of conventional upper limb stroke rehabilitation treatments. Comparison versus control group with usual care (including no treatment) [13]

In the scope of this dissertation, the main focus for rehabilitation will be repetitive exercises. More specifically, robotic assisted repetitive exercises. Repetition is a key principle in motor skills' acquisition. Several reviews have found that repetitive task training is one of the most effective methods for post stroke rehabilitation [15][16]. With varying GRADE evidence levels from various reviews, nonetheless, the consensus seems to show that most evidence is low to moderate level GRADE evidence [17]. This repetition relates to "an active motor sequence performed repetitively within a single training session, with practice aiming towards a clear functional goal" as stated by Beverley French et. al. [18].

Robotic rehabilitation, through the use of repetitive exercises, has been shown to display results with the same efficacy or better than usual care, non robotic, repetitive exercises, with the efficacy of this treatment already described in the previous paragraph [15][19][20]. Several scores can measure the patient's capacity, and these studies display positive results for robotic rehabilitation, as seen in Figure 2.2.

The evidence for the benefit of an exoskeleton vs end effector solution is inconclusive, with both typologies enabling positive outcomes of therapy in different studies. However, the main

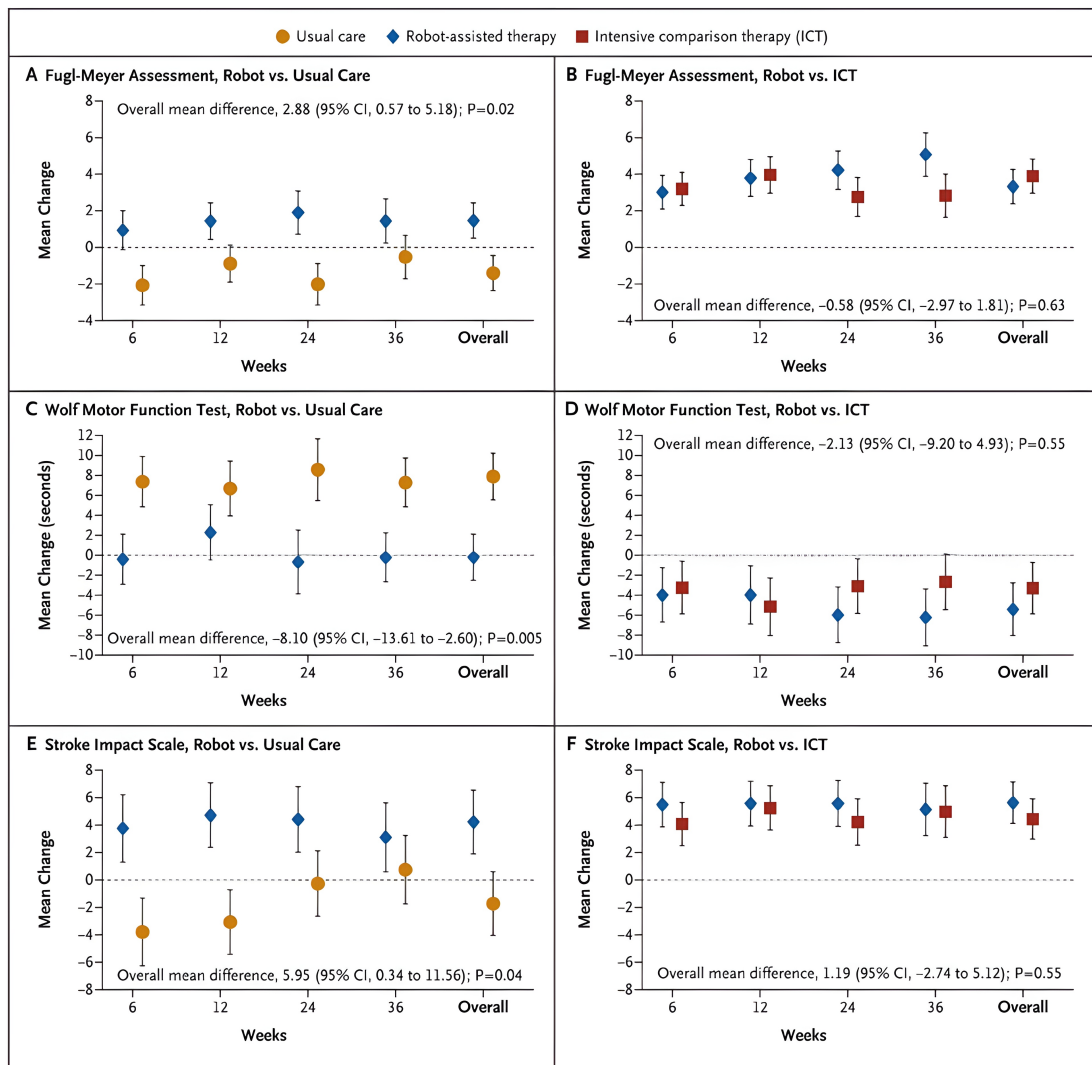


Figure 2.2: Changes in the outcomes during a 36 week period, comparing robotic rehabilitation, Usual Care and Intensive Comparison Therapy (ICT) [20]

benefit attributed to robotic rehabilitation is the capacity for increased repetition and intensity during therapy, with results displaying increased effectiveness in patients who were more severely impaired by stroke [19][21].

Robotic rehabilitation devices also can be associated with the prospect of unassisted rehabilitation, allowing for unsupervised rehabilitation and increasing positive outcomes. This would allow for at home rehabilitation, reducing the need for on-clinic therapy. Assistive devices derived or designed with similar concepts as rehabilitation devices could also provide a meaningful quality of life improvement for impaired stroke patients [22][23].

Rehabilitation robotics is still far from its potential. Most existing systems lack portability and the capacity to evaluate the neurological needs of the patients and provide personalised treatment. The inclusion of personalised treatment that enables the targeting of visuospatial, sensory and

cognitive function will likely yield better outcomes in all domains affected by stroke [24].

2.2 Collaborative Manipulator

A collaborative manipulator is one type of collaborative robot, or COBOT. Collaborative manipulators are one type of manipulator designed to interact directly with a human operator safely. They ensure that no harm or danger can be created for the operator.

A collaborative manipulator, as it's implied, is also a manipulator robot. A manipulator robot is a kind of robot that allows for the handling of objects with precision and reliability. Manipulators are a common sight in many industrial environments and are able to perform an array of tasks due to their versatility and adaptability. Manipulator robots can vary in their availability of DoF (Degrees of Freedom) and usually feature an end effector tool to realise their tasks. Manipulators can also be featured on mobile platforms or be stationary. In this document, the sole focus will be on stationary collaborative manipulators.

The next paragraphs will be focusing on commercially available solutions for collaborative manipulators. The COBOT market is vast, with many options for different requirements. However, it is necessary to select the specific requirements for that task of upper limb rehabilitation. According to 2016 data, the arm, forearm and hand combined account for around 10% of the human body mass, and the forearm and hand combination represent 4.4% and the hand 1.2% of the human body mass. In order to be able to carry out the required task of upper limb rehabilitation in the greatest percentage of the population possible, the manipulator requires a payload capacity of at least around 5.8 kg in order to accommodate bearing the weight of the forearm and hand combination and 13.1 kg to accommodate the full arm system (arm, forearm and hand). These values are used in order to accommodate an individual of 130.3 kg (95th percentile for the American population). These values drop significantly for the weight of the median American (90.6 kg, under the 75th percentile), with the forearm and hand combination having an approximate weight of 4 kg and the full arm, forearm and hand combination weighing approximately 9.6 kg as can be seen in Table 2.1[25][26].

Table 2.1: Arm and forearm weight distribution [25][26]

Percentile	Arm, Forearm, and Hand weight (kg)	Forearm and Hand weight (kg)	Hand weight in (kg)
Mean (90.6kg)	9.06	3.99	1.09
50th (87.4kg)	8.74	3.85	1.05
75th (101.9kg)	10.19	4.48	1.22
85th (110.6kg)	11.06	4.87	1.33
90th (119.4kg)	11.94	5.25	1.43
95th (130.3kg)	13.03	5.73	1.56

Based on the data presented above, it is possible to conclude that in order for a manipulator to support an upper limb, it requires at least a 6 kg payload capacity in order to support the forearm

of the patient. However, these calculations do not account for the weight of the end effector, which also requires support by the manipulator. For the 95th percentile, a 6 kg payload capacity manipulator is demonstrated to be clearly insufficient for the desired task, as it leaves only a 0.27 kg supported capacity for the end effector. As the end effector is required to have actuation, it should have a greater weight tolerance. Therefore, only manipulators with at least 10 kg of payload capacity will be considered.

In the above-mentioned payload range, the market features numerous options of COBOTs. In this document some of these will be analysed. The market also features medically certified COBOTs ready to be deployed in real life rehabilitation scenarios, such as the KuKa LBR Med, the first medically certified COBOT [27], which exists in two variants, but only one above the 10 kg payload capacity, the KuKa LBR Med 14 (Figure 2.3), with a 14 kg payload capacity [28]. This manipulator features characteristics nearly identical to the industrial manipulator Kuka LBR iiWA 14 [29]. As for other industrial COBOTs, brands such as Techman Robot, Yaskawa, ABB and Universal Robotics all feature manipulators with the necessary requirements. Techman Robot features several COBOTs in the necessary capacity range, with the TM12, TM14 (Figure 2.4) and TM16 featuring a payload capacity of 12 kg, 14 kg and 16 kg respectively [30]. Yaskawa provides the HC10DTP (Figure 2.5), a 10 kg load capacity COBOT [31], ABB the GoFa manipulator in two variants a 10 kg payload capacity variant (Figure 2.6) and a 12 kg load capacity variant (Figure 2.7)[32]. Finally, Universal Robotics has two models as well in this range, the UR10e and UR16e (Figure 2.8), with 12.5kg and 16kg payload capacities, respectively [33][34].



Figure 2.3: KuKa LBR Med [27]



Figure 2.4: Techman Robots TM14 [35]



Figure 2.5: Yaskawa HC10DTP [36]



Figure 2.6: ABB GoFa 10kg [37]

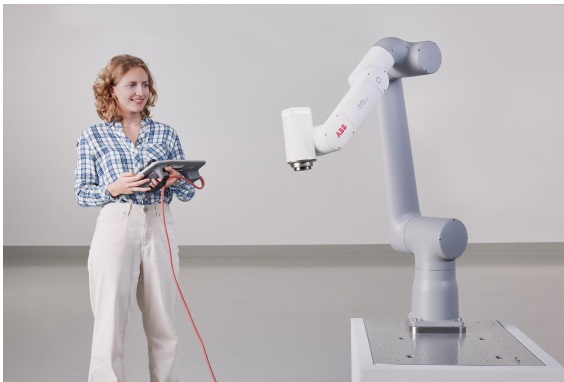


Figure 2.7: ABB GoFa 12kg [37]

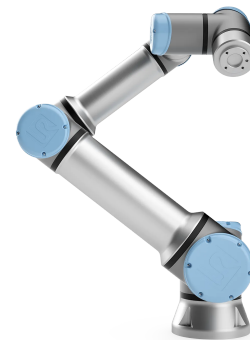


Figure 2.8: Universal Robotics UR16e [38]

These manipulators feature similar capabilities. However, all of them feature slight differences which distinguish them, making some stand out more than others based on the project's requirements. In the following table, more detailed specifications will be displayed comparing the manipulators.

Table 2.2: Manipulators Specifications [28][29][30][31][32][34][33]

Manipulator	Payload Capacity (kg)	DoF	Weight (kg)	Reach (mm)
KuKa LBR Med 14 [28]	14	7	32.7	820
KuKa LBR iiWA 14 [29]	14	7	32.7	820
TM12 [30]	12	6	32.8	1300
TM14 [30]	14	6	32.5	1100
TM16 [30]	16	6	32	917
HC10DTP [31]	10	6	58	1200
ABB GoFa 10 [32]	10	6	51	1520
ABB GoFa 12 [32]	12	6	48	1270
UR10e [33]	12.5	6	33.5	1300
UR16e [34]	16	6	33.1	900

Analysing the data from Table 2.2, it can be concluded that most available options feature a similar range of operational conditions, and a few trends can be observed. All manipulators bar the KuKa LBR Med and Kuka iiWA feature 6 DoF, and a correlation can also be observed between the reach capability of the COBOT and its maximum payload, with the manipulators that have a bigger payload capacity featuring a smaller maximum reach compared to similar models from the same brand. However, none of the reach values presented in this table are restrictive to choosing a capable manipulator. The human upper arm has an average length of around 330 mm for males and 300 mm for females [39], these values are within the expected parameters of operation of the reviewed manipulators.

2.3 End Effector

The work conducted in this dissertation is mostly based on the development of an end effector. Any tool, device or extension that couples to the wrist of a robotic manipulator can be described as an end effector, and they are the main way that. It can be viewed as the "hand" of a manipulator, the part of the robot that interacts with the environment and performs the desired task. However, it does not have to resemble, and usually doesn't, the human hand or any other biological "hand". It can employ a variety of purposes, from grasping, pushing, fastening, moving, sensing or any set of desired mechanical tasks. The term "end effector" is not interchangeable with the term "gripper", as "gripper" is one specific kind of end effector as well as the most common type of end effector[40].

End effectors can be categorised into three different categories based on their design and application:

- Grippers
- Sensors
- Process Tools

Grippers' main use case is grasping and holding objects securely and reliably, and they are the most common type of end effector in robotics and essential in many applications. Grippers can also be further subdivided into their own categories. Mechanical grippers are the simplest type of gripper. Although their complexity may vary, they usually use jaws or fingers to grasp and manipulate objects. They are simple and robust and used in applications where firm and stable grasping is required, such as in pick-and-place operations or assembly tasks. An example of a parallel jaw gripper can be seen in [Figure 2.9](#) and an example of a finger gripper can be seen in [Figure 2.10](#). Vacuum grippers utilise pneumatics and suction cups to hold objects. Their use cases are more specific, as they require smooth flat surfaces in order to have a firm and reliable grasp. They can be used in processes such as glass handling or PCB (Printed Circuit Board) manufacturing. Magnetic grippers make use of magnetic fields to manipulate ferromagnetic objects. They are most commonly used in the handling of heavy and/or irregular ferromagnetic objects, providing a strong and reliable grip. Servo grippers, or electric grippers, are a type of gripper that utilise electric motors and gearboxes to precisely control the force applied on the objects they are gripping. This kind of gripper provides high precision and flexibility and is usually used for delicate objects or tasks, such as quality control or inspection tasks [\[41\]\[42\]\[43\]](#). An example of this type of gripper can be seen in [Figure 2.11](#).



Figure 2.9: Parallel Jaw Gripper [\[44\]](#)



Figure 2.10: Finger Gripper [\[44\]](#)

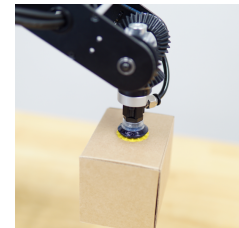


Figure 2.11: Vacuum Gripper [\[44\]](#)

The second category of end effector mentioned, sensors, are used as a way for the robot to gain perception of its environment. Sensors gather data on the robot's environment and can monitor variables such as the position and orientation of the robot or the object. Robots can be equipped with innumerable different types of sensors, but the ones most commonly used are cameras, ultrasonic sensors, force/torque sensors, light sensors, hall effect sensors and proximity sensors [\[41\]\[42\]\[43\]](#).

Lastly, end effectors can also be categorised as process tools. Process tools can be described as tools designed to perform specific tasks. They can be viewed as more advanced grippers who need to perform more precise and specific tasks besides gripping objects. Their complexity can greatly vary from being a simple screw placing tool to an entire complex assembly similar to a

robot. Some common examples of these tools are painting tools, welding tools, cutting tools, dispensers and grinding tools to name a few [41][42][43]. Rehabilitation tools usually align with this category of end effector, as they have specific requirements that require specific tools. The work developed in this dissertation falls into this subcategory of end effector.

In the field of rehabilitation robotics, there exist a few solutions for upper limb rehabilitation end effectors. They usually consist of a base where the patient rests his forearm, where it can be tied down with velcro straps, as seen in Figure 2.12 solution, or resting unrestrained. It is also a common solution to have a cylinder for the patient to grip with the hand. These feature unilateral movement of the forearm only, providing no further actuation on any other part of the upper limb. This can be seen in the MOTORE solution in Figure 2.13



Figure 2.12: BIONIK's Arm/Hand end effector solution for upper limb rehabilitation [45]



Figure 2.13: Humanware's MOTORE solution for upper limb rehabilitation post Stroke [46]

2.4 Actuators

The development of an end effector for the purpose of upper limb rehabilitation requires a form of actuation in order to deliver the required movements to the patient. In this section, a review will be performed on the different types of actuators available and suitable for end effectors used in the rehabilitation of the upper limbs for stroke patients. Providing analysis on the different kinds of actuation, their benefits and downsides, along with an explanation for the chosen actuation method.

The development of an end effector for rehabilitation is most commonly done using active methods [47]. For the purposes of end effector actuation, there are several solutions on the choice of actuator, with endless types of actuation available for this purpose; however, the constraints of the chosen system must also be taken into consideration in the selection of these actuation methods.

The most common method of actuation in end effector based rehabilitation is done using electric actuators, due to their versatility and adaptability [47]. Nevertheless, several other options can be and are used for this purpose. The most common options for actuation besides electric

actuators are usually pneumatic actuators, hydraulic actuators, Series Elastic Actuators (SEAs), Variable Stiffness Actuators (VSAs) and Pneumatic Artificial Muscles (PAMs) [47][48].

These actuation methods have their advantages and disadvantages, which can be observed in Table 2.3. Furthermore, VSAs and SEAs are categorised as compliant actuation. Compliant actuation is one kind of actuation that is able to deliver more dynamic adaptability and shock tolerance due to the elastic properties of these actuators. These actuators can more precisely reproduce forces similar to the ones experienced by the upper limb in its natural motion [48].

Table 2.3: Comparison of Actuation Types: Advantages and Disadvantages [47] [48] [49]

Actuation Type	Advantages	Disadvantages
Electric	• Versatile • High power	• High weight • High impedance
Pneumatic	• Lightweight • Lower impedance • High power-to-weight ratio	• Stationary system • Requires pressurized air • Slow dynamics • Hysteresis • Limited service area
Hydraulic	• High power • High payload capacity • High precision	• High weight • High impedance • Complex mechanical design • Fluid handling • Noise and size constraints
SEAs	• Energy efficiency • Higher user safety • Dynamic adaptability • Shock tolerance • Stability	• Fixed compliance • Low bandwidth • Nonlinear torque–weight relationship
VSAs	• Energy efficiency • Higher user safety • Dynamic adaptability • Shock tolerance • Good joint simulation • Adjustable compliance	• More complex mechanical design • Springs increase adjustment complexity • Nonlinear torque–weight relationship
PAMs	• Lightweight • Low cost • High specific force • High specific torque	• Slow response • Nonlinear response • Complex design and control

As observed in the above table, different actuation methods provide different challenges, benefits and disadvantages. With some solutions providing clear benefits in rehabilitation. Mainly the use of compliant actuators, which allow for better control and more precise movements that can more accurately simulate the natural movement of the upper limb [48]. However, these solutions are very complex mechanically and therefore expensive.

2.5 Electric Motor Review

In what regards electric motors for actuation, the choice is quite extensive. As such, it is necessary to analyse all available options in order to find the best available solutions.

As studied in [Subsection 2.4 Actuators](#), the available solutions of actuations are quite extensive. However, the most commonly used solutions are based on electric motor actuation. As such, a more thorough review of the electric motors utilised in these devices is required.

In order to accommodate the largest percentage of the population, the motor would require enough torque to accommodate the weight of the human arm system. As seen previously, the 90th percentile of the American population has an arm weight of 11.94 kg. However, the motor is only applied at the elbow, requiring actuation of the forearm and arm only [26] [25]. The forearm and hand combination is 5.25 kg for the same percentile [26] [25]. The average forearm length is 0.35 m or 350 mm [50]. With the normal torque values for elbow flexion and extension in exoskeletons being up to 9 Nm [51]. This restricts the types of electrical motors that are useful for the necessary task. Therefore, the motors reviewed will be the ones most commonly used: brushless DC motors, servo motors and stepper motors.

2.5.1 Brushless DC Motors

Brushless Direct Current Motors (or BLDC) are one type of synchronous electric motor. This motor's rotor and stator generated magnetic fields rotate at the same frequency (in sync). As suggested in the name, these kinds of motors do not utilise any brushes for commutation, with the commutation being done electronically. BLDC motors are commonly used in many industries, such as appliances, automotive, aerospace, consumer, medical, industrial automation, and equipment and instrumentation. This type of motor also provides many advantages over induction and brushed electric motors, such as better speed/torque characteristics, high dynamic response, higher efficiency, lower noise operation, higher speed ranges, longer lifespan and lower maintenance [52].

BLDCs are fabricated in single, two or three phase configurations, with the 3-phase configuration being the most common. The stator is composed of fixed wiring, while the rotor is composed of embedded permanent magnets. In order to detect rotor position, BLDC machines can utilise either Hall Effect (magnetic) encoders or derive the position through back EMF measurement [52].

These particular types of motors possess many advantages useful for utilisation in a human-robotic interface, making them an optimal choice for such an application. The integration of a BLDC into the end effector would provide precise, efficient and reliable control of movements in the system, providing good system performance.

2.5.2 Servo Motors

Servo motors aren't an actual classification of electric motors. The term 'servo motor' just implies any kind of electric motor that operates in a closed loop system, which utilises feedback in order to compare and correct the motor's position, torque or speed, allowing it to determine any of these parameters with high precision. The definition does not specify any kind of construction or operation specification. Servo motors can be both AC or DC motors, synchronous or asynchronous, brushless or brushed. However not all of these types of drives perform the same way when utilised as a servo motor[53][54].

Servo motors are most commonly BLDC motors in a closed loop system. This allows them to harness all the benefits above described, making them an ideal candidate for an application such as a rehabilitation end effector due to their capability to actuate with precision and fine tuned control [53].

2.5.3 Stepper Motors

Stepper motors are a type of digital actuator designed to rotate in discrete angular increments, making them particularly well-suited for precision control applications. A stepper motor is essentially a brushless DC motor whose rotor advances incrementally when its stator windings are energised in a predefined sequence. This type of construction allows the rotor to lock into position without the need for position feedback systems, relying instead on the controlled energisation of stator poles. Stepper motors are widely used in digital control systems and applications requiring high positioning accuracy, such as 3D printers, CNC machines, and robotic arms [55][56].

These motors come in different structural forms, including variable reluctance (VR), permanent magnet (PM), and hybrid types. In a VR stepper motor, the rotor is typically constructed from soft iron with salient poles and has no windings, relying purely on the interaction between the stator magnetic field and the rotor's shape. In contrast, permanent magnet stepper motors incorporate magnetic rotors that enhance torque and holding force. The hybrid type, which combines features of both VR and PM designs, is the most commonly used today due to its improved precision and performance. Regardless of type, the core principle remains the same—rotor movement is dictated by a sequence of electrical pulses sent to the stator, with each pulse corresponding to a specific angular displacement [55][56].

Because of their ability to provide open-loop control, hold position firmly, and achieve accurate positioning without encoders, stepper motors are widely adopted in industrial automation, mechatronics, and embedded systems. However, their performance is highly dependent on the drive electronics and pulse timing, as poor control can lead to resonance or missed steps. Nonetheless, their sturdy design, ease of integration, and cost-effectiveness continue to make them a valuable component in electromechanical system design [55][56].

In human-robotic interfaces and rehabilitation robotics, stepper motors are valued for their ability to provide precise, repeatable motion through open-loop control, eliminating the need for

encoders. This makes them ideal for guiding limb trajectories or performing structured rehabilitation exercises. Their firm position-holding, ease of integration, and cost-effectiveness have led to widespread use in mechatronics and embedded systems. However, performance depends on proper drive electronics, and timing issues like resonance or missed steps can arise if poorly configured. Despite these limitations, their reliability and simplicity make them a strong fit for controlled therapeutic applications.

2.6 PID Control Overview

One of the most common methods of controller schemes is the PID controller, or Proportional Integral Derivative Controller (PID). A PID controller operates with 3 different control variables, K_p , the proportional gain, K_i , the integral gain and K_d , the derivative gain. These 3 values form a 3 action control scheme, with this form of PID control being widely used in industrial automation due to its balance between simplicity and performance [57][58][59]. A control scheme for a typical PID controller can be seen in Figure 2.14.

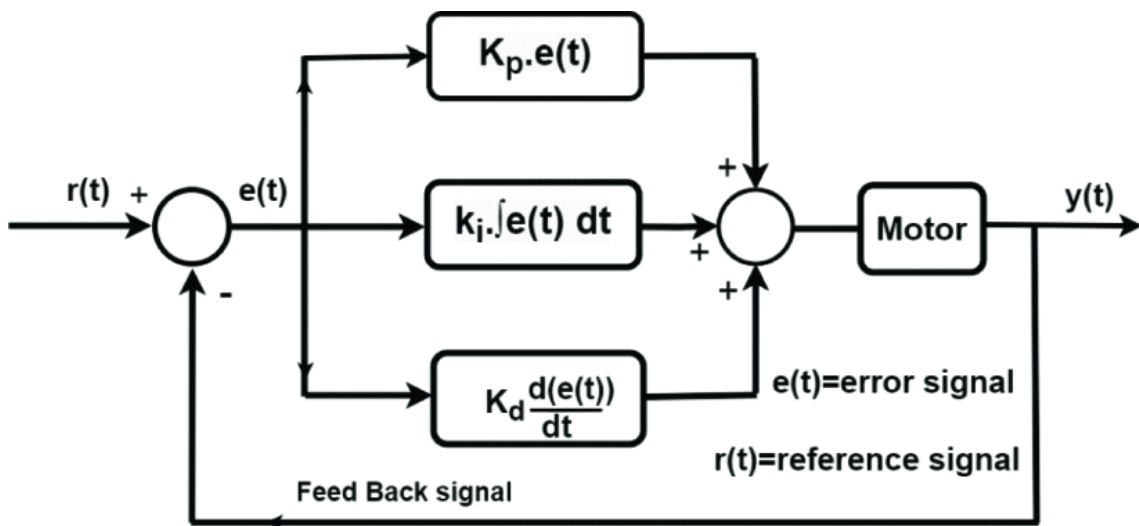


Figure 2.14: PID Control Scheme [57]

The input to the controller is the error signal $e(t)$, representing the difference between the desired reference signal and the actual system output. The proportional component (P) produces an output that is directly proportional to the current error value. It is responsible for correcting large, immediate discrepancies between the desired and actual values. However, proportional control alone cannot eliminate steady-state error. To address this limitation, the integral component (I) sums the past errors over time and adds this accumulated value to the control output, thereby eliminating residual offset and ensuring accurate steady-state performance. On the other hand, the derivative component (D) predicts future error based on its rate of change, enabling anticipatory action that helps dampen overshoot and reduce oscillations [57][58][59].

Together, these three terms form the standard PID control law:

$$u_c(t) = K_p e(t) + K_i \int e(t) dt + K_d \frac{de(t)}{dt} \quad (2.1)$$

where K_p , K_i and K_d are the proportional, integral, and derivative gains, respectively. The PID output $u(t)$ is used to adjust the actuator, such as a motor driver.

This structure makes PID controllers well suited for motor drive systems, where real-time adjustment of speed or position is required. The controller compares the feedback from an encoder or sensor to a reference trajectory and continuously adjusts the voltage, current, or torque commands to minimise error. In such applications, the PID control loop is typically implemented in microcontrollers or digital signal processors, with the gains K_p , K_i and K_d tuned based on system dynamics and performance requirements [57][58][59].

PID control remains a cornerstone of industrial and robotic control systems due to its simple structure and reliable performance across a varying range of applications. The combination of the three control variables (integral, derivative and proportional) enables the controller to improve both the transient and steady-state responses of a system, reducing rise time, steady-state error, and overshoot. In practical robotics applications, especially those involving noisy sensors or discrete-time implementations, the derivative term is often omitted, resulting in a PI controller that retains good performance while improving robustness [58].

PID control design typically involves a tuning process to determine appropriate gains for the proportional (K_p), integral (K_i), and derivative (K_d) terms. Classical tuning strategies include heuristic approaches like the Ziegler–Nichols method, which estimate gains based on the system's ultimate gain and oscillation period. These rules are straightforward but tend to produce aggressive responses with high overshoot [58][59]. To improve controller performance, frequency-domain tools such as Bode plots and phase/gain margin analysis have been employed to guide tuning more analytically [58]. Performance is often evaluated using criteria such as the Integral of Absolute Error (IAE), rise time, settling time, and overshoot, which provide quantitative measures for comparing different tuning strategies [58][59].

While classical methods remain widely used due to their simplicity and effectiveness, they are often not enough for highly nonlinear or time-varying systems. In such cases, modern alternatives—including optimisation-based techniques and intelligent algorithms—have been introduced to overcome the limitations of classical PID tuning. Nonetheless, the foundational principles of classical PID control continue to underpin much of modern control theory and practice [58].

In order to determine the gain's values for a motor, the actuator and its load can be represented as a second order system, an oscillatory system. These are described by the following differential equation: [58]

$$\ddot{y}(t) + 2\zeta\omega_n\dot{y}(t) + \omega_n^2 y(t) = Ku(t) \quad (2.2)$$

where ω_n is the natural frequency and ζ the damping ratio. This structure is representative of many mechanical systems, including rotary actuators, and serves as a foundation for analytical PID tuning. Assuming a proportional-derivative (PD) control structure acting on a system with known inertia J , the controller gains can be computed to match the closed-loop dynamics of the

system to a desired second-order response. This leads to the following tuning rules: [58]

$$K_p = J \cdot \omega_n^2, \quad K_d = 2J\zeta\omega_n \quad (2.3)$$

These equations ensure that the system's natural response (rise time, overshoot, damping) adheres to specifications suitable for the application. For rehabilitation robotics, critical damping ($\zeta = 1$) and low natural frequencies (e.g., $\omega_n = 5$ rad/s) are typically chosen to ensure safe and compliant movement. Integral action K_i is usually omitted in such applications, as steady-state torque error is not critical and the inclusion of integral gain may destabilise the interaction with a human user [58].

Integral action K_i is typically omitted in position control where steady-state torque error is not critical. However, in applications where slow drift or load disturbances are significant, a small K_i can be added to eliminate steady-state error without compromising stability. In such cases, K_i can be estimated as: [58]

$$K_i = \alpha \cdot K_p \cdot \omega_n \quad (2.4)$$

Where α is a small scalar (typically between 0.05 and 0.2) to avoid instability or wind-up. This approach is consistent with experimental findings in rehabilitation robotics, where K_i is kept significantly lower than K_p to preserve compliance and safety. Studies report that setting K_i to approximately 5–10% of $K_p \cdot \omega_n$ is sufficient to eliminate steady-state errors while maintaining smooth, user-friendly control dynamics [60].

These formulas ensure that the system's natural response (rise time, overshoot, damping) adheres to specifications suitable for the application. For rehabilitation robotics, critical damping ($\zeta = 1$) and low natural frequencies (e.g., $\omega_n = 5$ rad/s) are typically chosen to ensure safe and compliant movement [58][60].

In rehabilitation systems specifically, the ability of the PID controller to switch between passive and active modes by adjusting gains in real time is critical. This makes it best suited for applications that require both precise trajectory following and compliance with the human-robotic interface's movement, providing a foundation for assist-as-needed strategies and feedback control [61].

2.7 CAN Communication

In this section a small review of CAN communication will be performed. CAN communication is among the most utilised in automation and robotics applications. Therefore, a deeper review of this protocol will be performed.

The Controller Area Network (CAN) is a serial communication protocol originally developed by Robert Bosch GmbH in the 1980s for use in automotive systems and has since evolved into a robust and widely adopted fieldbus technology in robotics, automation, and distributed embedded systems. CAN is defined in ISO 11898 and specifies only the physical and data link layers, leaving the application layer to be implemented based on system needs. Unlike traditional point-to-point

interfaces like RS232 or RS485, CAN supports real-time, multi-master communication among nodes with built-in arbitration and error detection mechanisms [62] [63].

A defining feature of CAN is its use of non-destructive bitwise arbitration, which allows the highest priority message to proceed on the bus without collision, even when multiple nodes attempt to transmit simultaneously. Each message on the bus is assigned a unique identifier that determines its priority. Lower numerical values correspond to higher priorities. This design ensures deterministic timing and makes CAN particularly suited to time-sensitive and safety-critical applications [62]. CAN supports various communication topologies, including point-to-point, broadcast, and multicast, all using just two differential signal lines (CAN_H and CAN_L), contributing to its simplicity, low wiring cost, and strong noise immunity [63].

Physically, CAN is implemented using a transceiver and a controller with an integrated or external CAN controller. The maximum transmission rate is 1 Mbps for up to 40 m, but can reach up to 10 km at reduced speeds (e.g., 5 kbps). The protocol employs short frames with a maximum data payload of 8 bytes, which enhances real-time performance and reliability. All messages undergo CRC-15/CRC-17 error checking and automatic retransmission upon detection of failure. Fault confinement mechanisms further isolate malfunctioning nodes to maintain network integrity [62] [64].

The flexibility of CAN lies in its decentralised communication model: every node can initiate communication without needing a central master. This is crucial in modular systems such as underwater vehicles or robotic networks, where subunits operate semi-independently while responding to commands from a main controller. CAN's filter-and-mask mechanism enables nodes to selectively process only relevant messages, optimising CPU load and bandwidth usage. Identifiers can be structured to encode module address and message type, simplifying message parsing in large networks [63].

Due to these characteristics—robustness, real-time capability, and flexibility, CAN remains a preferred communication protocol in embedded and robotic systems requiring distributed intelligence and fault tolerance. Although the protocol was not originally designed for higher-layer standardisation (such as CANOpen), its physical and data link layers continue to provide a reliable foundation for customised application layer implementations tailored to specific system requirements [64].

2.8 Exoskeletons

One other common solution for robotic-assisted rehabilitation of the upper limbs in stroke patients comprises the use of exoskeletons. An exoskeleton solution provides many benefits over an end effector solution, as exoskeletons provide better joint fit and control. End effector based solutions cannot effectively control the posture of the upper limbs based off one interface, and controlling the torque on certain upper limb joints can prove a great challenge. As an effect of this generating isolated movement of single joints is also a great challenge, as the single interface usually causes

a combination of movements in more than one joint. This greatly limits the types of exercises that can be achieved for stroke recovery[65].

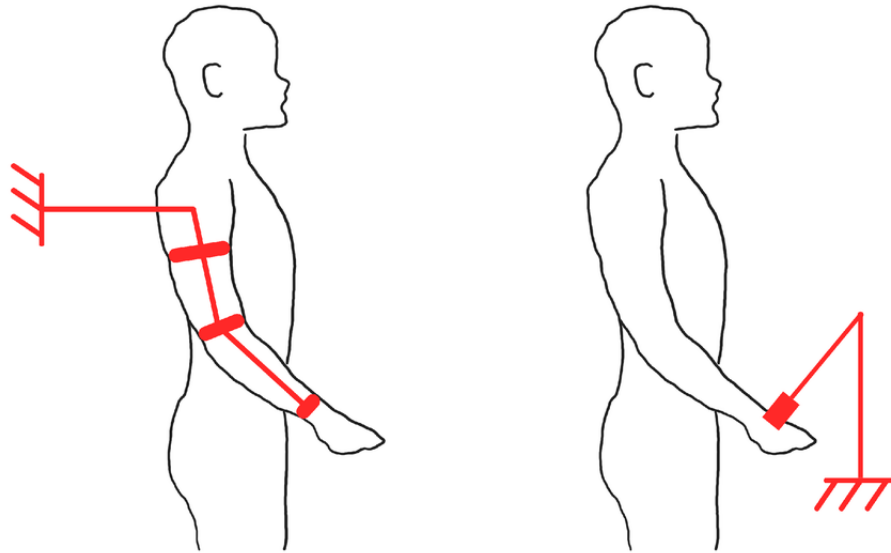


Figure 2.15: End effector based (right) solution vs exoskeleton based solution (left) [66]

An exoskeleton comprises a structure that is tailored to the human arm's anatomy. It is designed in a way that it ensures that the structure's joints match with the upper limb's joints, being able to interact with single specific joints independently of their connection with other joints in the arm [48][67]. However, this tailor made approach that the exoskeleton ensures has several disadvantages. The main one being that, similar to a tailor made suit, it is very hard to make a one size fits all approach, with it showing severe deficiencies in being able to adapt to different arm lengths [65]. An example of an exoskeleton system can be observed in Figure 2.16. The great complexity of these systems also adds one other disadvantage. Due to their complexity and lack of adaptability, these systems are commonly very expensive and able to fit only a small amount of the population. This makes the use of exoskeletons as a viable, universal solution for upper limb recovery unlikely at the current moment[65][67][68]. Figure 2.15 displays the difference between end effector and exoskeleton solutions.



Figure 2.16: Upper limb exoskeleton example: ARMEO Power [69]

2.9 Summary

This chapter presented an overview of robotic rehabilitation for the upper limb, establishing the clinical, mechanical, and technological background required for the analysis developed in this dissertation. The main principles of motor recovery following neurological injury were discussed, with emphasis on repetitive, task oriented training and the role of assist-as-needed strategies in promoting active patient participation. The primary categories of robotic rehabilitation devices, end effector based systems and exoskeletons were also introduced, along with their typical applications in clinical and research contexts. In addition, the chapter reviewed the most common control strategy used in rehabilitation robotics.

The study of collaborative manipulators showed that modern COBOTs are a valid tool for rehabilitation because they have enough payload capacity, workspace, and built-in safety features for people to interact with robots. In particular, manipulators such as the KUKA iiWA and similar systems were shown to meet the kinematic and operational requirements necessary for upper-limb rehabilitation tasks, while maintaining compliance with safety constraints.

Furthermore, the discussion of end-effector-based rehabilitation devices and actuation technologies revealed a clear trade-off between system complexity, motion fidelity, and cost. Exoskeletons offer better control at the joint level, but their complicated mechanics and cost hard to use and operate. End-effector-based solutions, although more constrained in joint specificity, offer a more practical and cost-effective alternative when appropriately designed and actuated.

Finally, the overview of electric actuators, control strategies, and communication methods clarified the suitability of brushless DC motors combined with digital control architectures for rehabilitation systems. These technologies provide the precision, adaptability, and controllability required to implement multiple rehabilitation modes, including passive, active, and assist-as-needed operation.

Overall, this chapter provides the conceptual and technical framework that informs the design choices presented in the subsequent chapters. The identified rehabilitation principles, robotic architectures, and actuation strategies.

Chapter 3

Literature Review

This chapter will focus on existing works regarding robotic rehabilitation of the upper limbs in stroke patients, more specifically, methods, practices, exercises.

3.1 Manipulator and End Effector based solutions

In the space that concerns the solutions for robotic assisted stroke rehabilitation based on end effectors and other robotic systems, there are several existing options. The purpose of this section is to analyse some of these solutions.

3.1.1 InMotion® ARM

In this first review, the focus of analysis will be the InMotion® ARM from BIONIK. The InMotion ARM is an end effector based system for rehabilitation of the upper limbs. This device provides task several specific movements, along with an interactive system in order to facilitate the patient's ability to perform the exercises correctly [70].

The InMotion ARM utilises impedance-resistance-based actuation, that is, a system that only actuates actively when the patient is not able or deviates from the required movement pattern in a given exercise, an assist-as-needed device. This method provides several advantages, with the main advantage being that it works as an assist-as-needed control scheme, that is, the patient initialises the movement independently and only actuates when the system detects any deviation or non-compliance with the full required exercise. Independent movement is critical to successful motor capacity rehabilitation, which reinforces that advantage of such a method. Although, these kinds of actuation can also trigger uncontrolled, unexpected movements which can result in undesired consequences [71][72][73]. Although this system can also utilise fully active or fully passive modes where the patient is required to do all of the movement independently or do it fully assisted [72].

This system consists of a shoulder-elbow unit for planar moments with 2 degrees of freedom. The device has an adjustable height workstation in order to ensure that the patient's arm is parallel to the floor and the patient is sited and aligned with the centre of the robot. The patient holds the



Figure 3.1: BIONIK InMotion ARM [70]

handle in the end effector with its hand and is provided arm support through a strap on the forearm [72].

This solution provides a wide range of movements, with training sessions usually being comprised of a repetitive set of tasks done with different assistance modes in order to complete the exercises provided by the interactive system. [72][70].

The InMotion ARM is built from a SCARA type robot. This configuration was chosen due to its low impedance in the horizontal plane and extremely high impedance in the vertical plane. It can easily support the weight of the patient's limb. It is also actuated by brushless motors, with 9.65 Nm of constant torque and encoders capable of measuring position and velocity [74]. As seen in Figure 3.1 the end effector of this system comprises an arm mount and built-in and velcro straps in order to ergonomically and comfortably support the patient's arm during movement.

As stated before, the InMotion ARM system utilises an impedance-resistance-based control. This type of control method allows for both passive and active utilisation of the system, as well as, in conjunction with an active mode, adding resistive forces or assistive forces according to the needs of the patient. This model establishes a relationship between the system's motion and the human-robot interaction forces. This relationship is based on a mass-spring-damper system. The

control loop diagram for this system can be observed in [Figure 3.2](#).

$$\tau_{\text{ext}}(t) = M_d \ddot{e}(t) + B_d \dot{e}(t) + K_d e(t) \quad (3.1)$$

Where the torque joint external vector is given by $\tau_{\text{ext}}(t)$, $e(t)$ represents the error from the measured joint angle q and the desired joint angle q_{des} considered without the application of external torque, with $\dot{e}(t)$ and $\ddot{e}(t)$ being the first and second derivatives of the error, respectively. The remaining parameters M_d , B_d and K_d represent the gains of the mass, spring and damper respectively. In this control system, q as system inputs, which display the system's motion state and $\tau_{\text{ext}}(t)$ torques as outputs. Impedance control also considers human interaction forces along with the motion tracking errors, achieving stable interaction. However, it cannot compensate for unaccounted dynamics, along with the instability inserted in the system at high impedances [\[73\]](#).

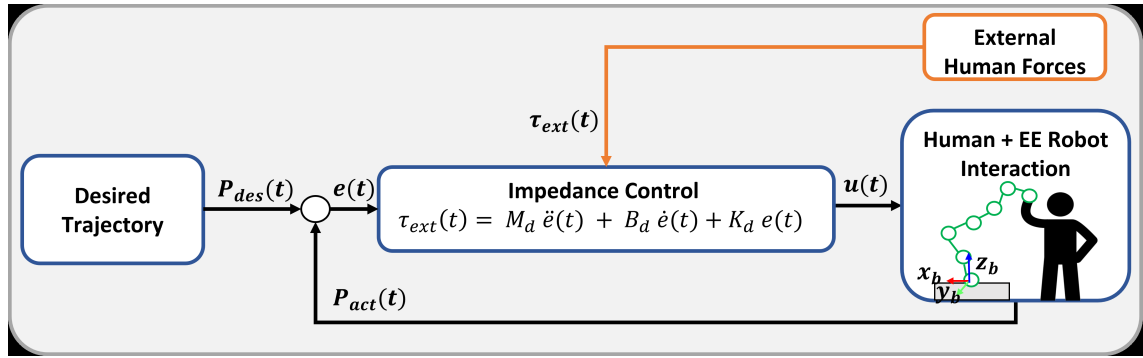


Figure 3.2: Impedance Control Block Diagram [\[73\]](#)

The InMotion system has faced extensive research and review. Due to this, several publications have been made regarding its results. Research shows that the methods used in the InMotion ARM system do not provide meaningful improvements in rehabilitation in primary metric scores such as FMA (Fugl-Meyer Assessment) and MRC (Medical Research Council) scales, with the results being similar to conventional therapy methods in both proximal and distal upper limb motor function and muscle strength [\[72\]\[75\]](#). While these results while do not show any improvements they prove themselves similar to conventional therapy, providing a meaningful alternative for an automated system of therapy.

3.1.2 REHAROB

The REHAROB or robotic rehabilitation system for upper limb motion therapy for the disabled, is one other end effector based solution for upper limb rehabilitation in stroke patients. The REHAROB system is composed of two manipulators attached to the patient arm and forearm through end effectors, as seen in [Figure 3.3](#). Similarly to the InMotion ARM, it is comprised of a shoulder-elbow support unit, and it supports passive or active assistance to the patient, but unlike the InMotion system, the REHAROB system provides full 3D motion capabilities and utilises indirect force control [\[76\]\[73\]](#).



Figure 3.3: REHAROB Robotic rehabilitation system for upper limb motion therapy for the disabled [77]

This system is designed to be able to execute a large set of slow movement exercises at constant speed in order to increase movement amplitude and decrease joint sparsity. This solution also has 3 DoF, enabling full 3D motion of the upper limb [71][78]. This method can provide a greater range of exercises. Enabling better modelling of regular treatment techniques on the patient. These exercises are usually applied with 2 different kinds of robot operation. One where the system is only providing resistance and the patient needs to be able to perform the upper limb movement against said resistance, and one where the patient must execute specific movements by performing specific trajectories actively guided by the REHAROB robot. In this mode, the robot is always performing some active actuation in order to guide the patient [79][78].

The design of this system is comprised of a pair of manipulators that are able to perform a varying range of exercises, attached to these, the system has two orthoses, or end effectors, that comprise the electromechanical interface with the patient's arm. These are 250mm long and include a safety relaxing mechanism, a 6 DoF force control sensor, a quick changer, a turn around handle with safeball®, a 6 DoF T/F force monitoring sensor, a second quick changer and a thermoplastic orthosis. It can also be seen on Figure 3.4 that the patient's arm is secured with cloth straps to the end effector in order to ensure a tight, ergonomic fit [80].

As stated above, the REHAROB system utilises indirect force control methods to perform actuation control on the patient. This control method ensures the safe operation of the manipulator as well as capabilities for human-based learning (programming) of exercises. This method utilises an outer loop force/torque sensor based PI controller in conjunction with an inner loop robot

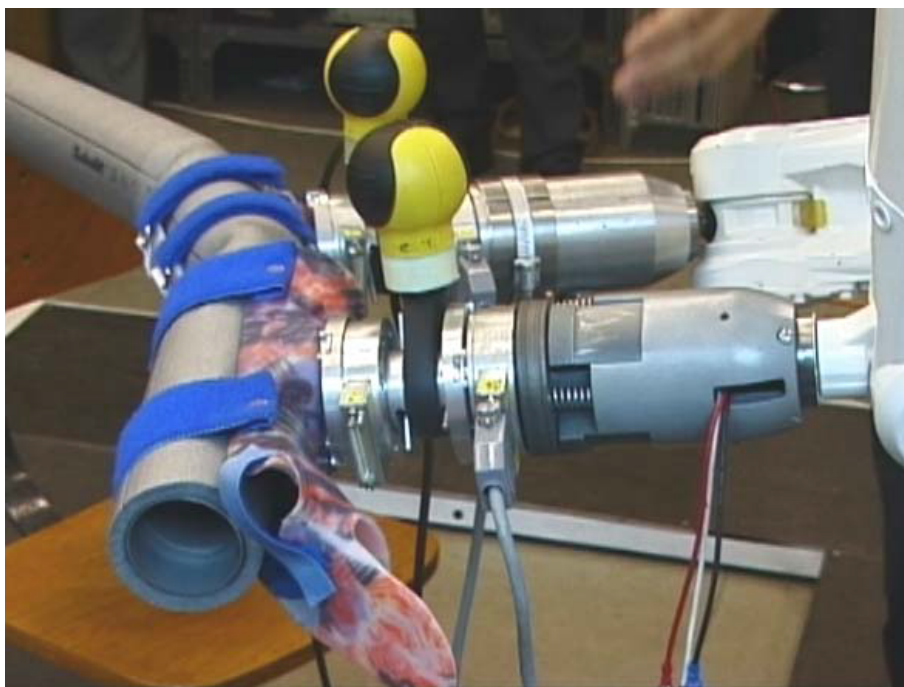


Figure 3.4: REHAROB's End effector [80]

internal position control. This works by reading the input force error signal (F_e) of the indirect force controller and comparing it with the force signal (F_s). If this is verified, the robot's force controller performs an incremental positional/orientation displacement of the end effector to the robot controller relative to the previous position ($X_d(k-1)$) of the robot arm at a 40Hz frequency. Force errors should also be minimised but are considered inevitable. This system ensures that excessive forces are never applied and user safety is maintained [78][73]. This control scheme can be observed in Figure 3.5

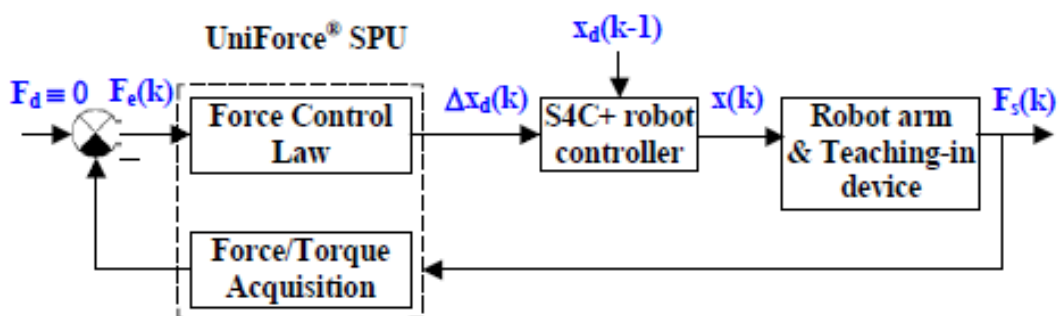


Figure 3.5: Block diagram of the Indirect Force Control used in the REHAROB system [78]

This system has also been the target of extensive reviews and research, providing ample data on its efficacy and uses. Studies have shown improvement in motor capacity in patients that used the REHAROB system compared to patients who performed conventional therapy. Particularly in the metrics of FMA shoulder-elbow, active range of elbow flexion (RMA) and FIMTM self-care.

However, no significant improvements were found in shoulder girdle anteflexion. This still proved to be a useful alternative to any conventional rehabilitation therapy [76].

3.1.3 MOTORE

The MOTORE, a mobile robot for upper limb neuro-ortho rehabilitation, is an end effector based solution for rehabilitation of the upper limbs. This solution, unlike the ones previously studied, comprises a mobile unit, designed to perform telerehabilitation, that is, rehabilitation away from a medically controlled environment (i.e. home). This system is composed of a single unit that does the actuation, requiring no manipulator arm. This was a key design principle in its development in order to make it portable [81]. The MOTORE rehabilitation device, similarly to the InMotion ARM is only capable of planar movement but is only able to provide support for the elbow and wrist. [73] [81] [82]. This system can be viewed in Figure 3.6



Figure 3.6: MOTORE Rehabilitation Device [83]

The MOTORE system is designed by Humanware to be able to execute sets of movements (exercises) with the patient's upper limb. In order to accomplish this, the MOTORE unit provides

force feedback during movement to ensure correct execution of movement but can, like the previous two reviewed systems, be passive. One other key feature of this system is its portability. One of the design requirements of the system was full portability in order to be able to operate in any place and at any table. To achieve this, the system is powered by an integrated battery, which is rechargeable with an AC adapter. The system with a full battery provides at least 70 min of operation, which is near the rehabilitation exercises execution times performed at a rehabilitation clinic [81].

The force feedback that this solution employs actuates to a maximum force of 30 Newton and an average velocity of 0.5 m/s , with acceleration limited to 1 m/s^2 . These values are based on data collected from conventional therapy exercises in order to keep the patient engaged with the treatment and be able to achieve desirable results. The MOTORE system also includes a contact sensor in order to detect user grasp of the end effector, as well as the incorporation of a support tool limiting the operational workspace [81].

The mechanical design of this end effector varies significantly compared to the previously analysed solutions. As stated before, this system is comprised of a single mobile unit that does not require a manipulator in order to perform its operation. In order to achieve this, Humanware designed a system that utilises three omnidirectional wheels, arranged in a 120° configuration, with their axes converging in the centre of the device. The wheels' rotation provides the necessary actuation. The handle that the patient grasps is off-centre towards the front of the device in order to provide more comfortable support for the arm and better weight distribution. The handle is supported by a flange that rotates utilising cylindrical bearings around the vertical handle axis. This solution enables friction free rotation of the forearm support, improving user comfort [81]. This can be observed in Figure 3.7.

The MOTORE unit utilises several control loops in order to perform the device's control. At the innermost level, it utilises PWM duty cycle loop (Figure 3.8). This loop is sampled at 5kHz frequency and utilises current monitors from H bridges. The current regulation allows for velocity control that enables the unit to follow specific reference velocities in the local coordinate system. With this control structure, the velocities can be accurately estimated based on only the proprioception system, without the need for Kalman filters. This configuration allows for a safer implementation in the absence of pen information (i.e. due to lags in the communication system). The outer loop control, seen in Figure 3.9 takes into consideration open loop compensations, such as the device inertia. The core of the outer loop system is a velocity generator. This module decides how to distribute the force that the patient applies on the handle to the velocity controller. This module is responsible for the different modes of operation, from no velocity to free motion or constrained movement.

The MOTORE system has been reviewed and researched and has shown efficacy in upper limb rehabilitation for stroke patients. Particularly with improvements in FMA scores, which display a significant reduction in upper limb impairment after treatment with this robotic rehabilitation system. Making it another viable alternative to conventional therapy [84].

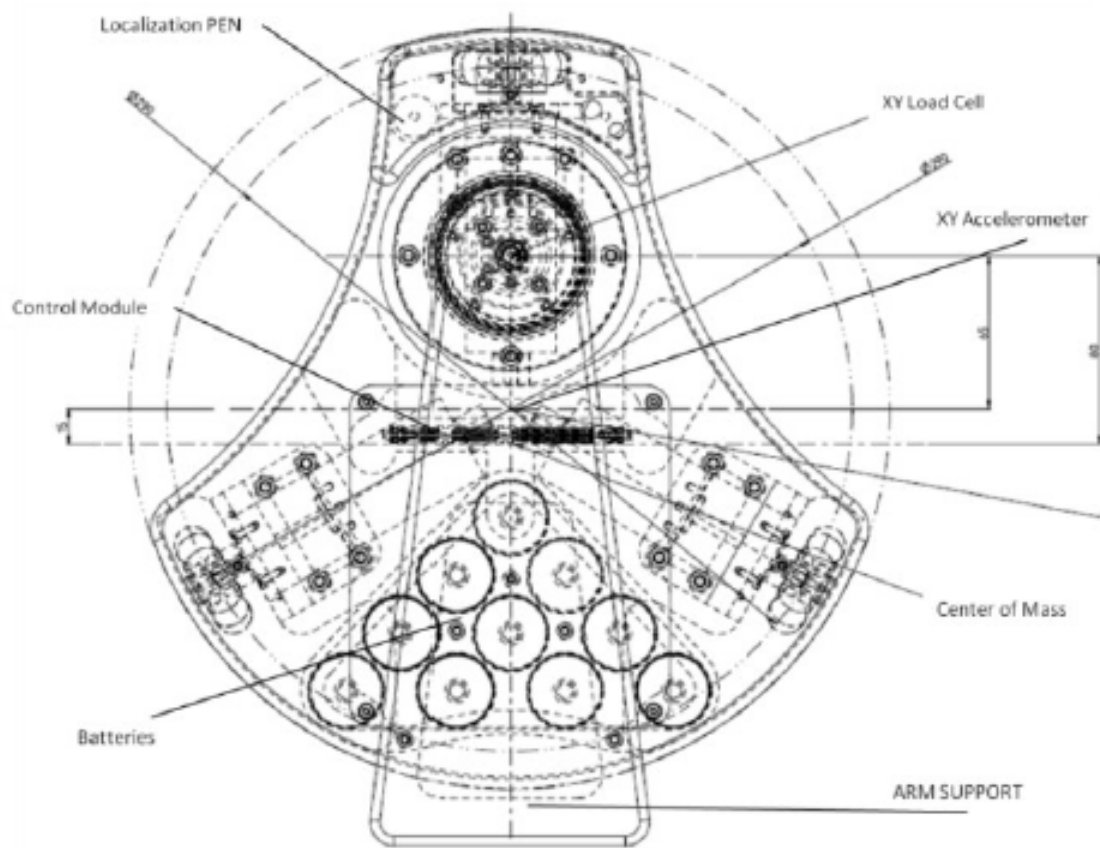


Figure 3.7: Diagram of mechanical structure of the MOTORE Rehabilitation device [81]

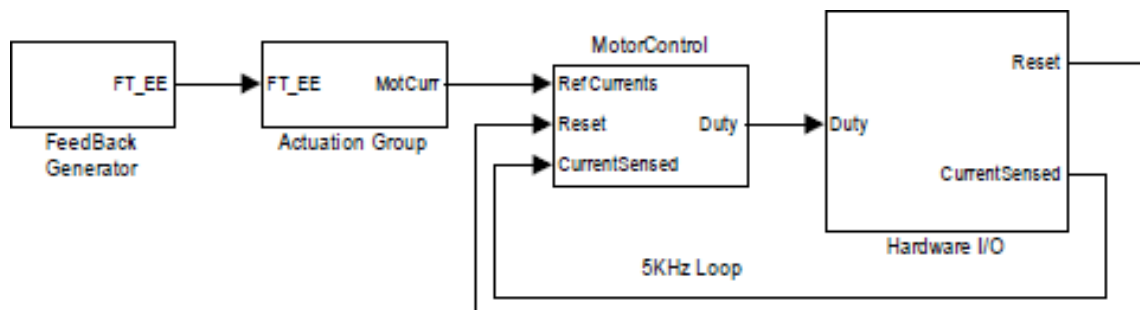


Figure 3.8: Inner loop control diagram of the MOTORE rehabilitation unit [81]

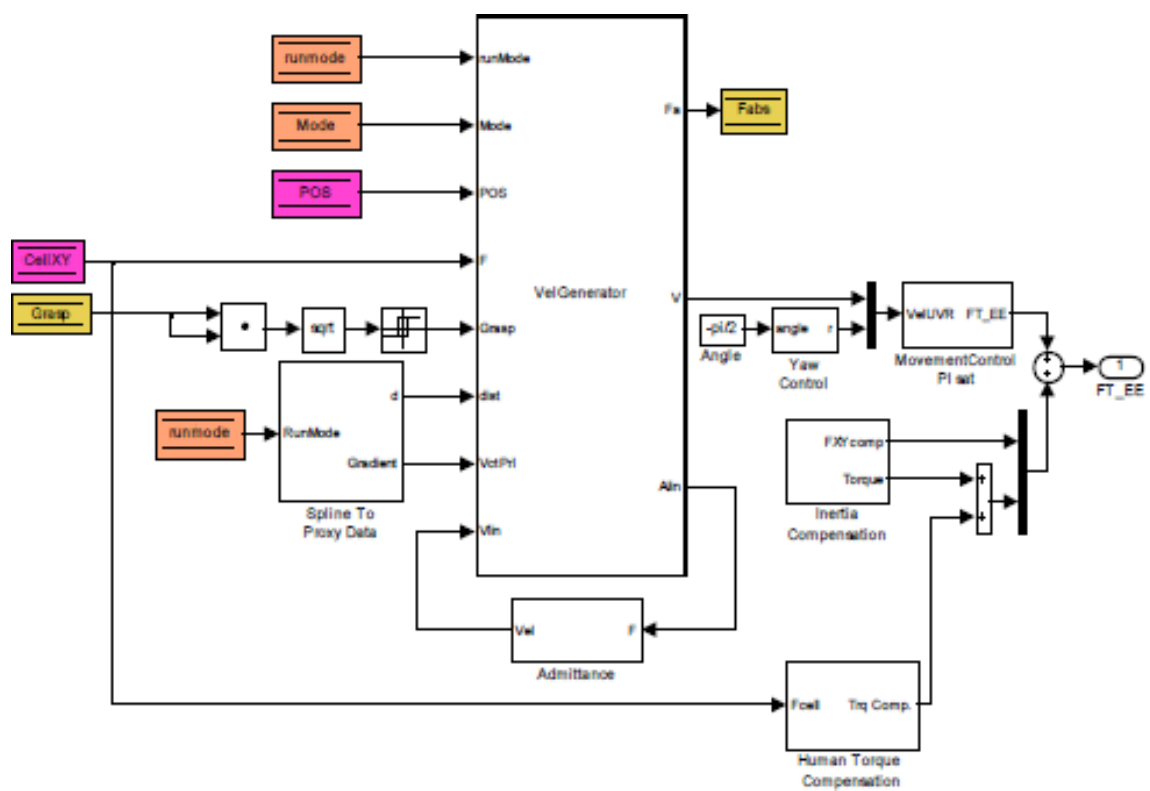


Figure 3.9: Outer loop control diagram of the MOTORE rehabilitation unit [81]

3.1.4 Amadeo

The Amadeo finger-hand rehabilitation device, seen in [Figure 3.10](#), is an end effector based solution for rehabilitation of the hand and fingers in stroke patients. This device is able to perform rehabilitation tasks for the fingers and hands of stroke patients. Similarly to the InMotion ARM this system also features an entertainment system in order to make exercises interactive and engaging [\[85\]\[86\]](#).

The Amadeo rehabilitation robot employs a system of position control, that always actuates a constant force on the patient's joints, regardless if the patient demonstrates having the required muscle strength for performing the given exercises [\[73\]\[86\]](#). This solution has 5 DoF, and it can provide independent movement of each finger from the passive rotation joint placed in between the fingertip and the laterally moving entity, with the thumb featuring 2 passive rotation joints. This solution provides ample coverage of the finger's workspace, although not all the space is covered [\[87\]](#).



Figure 3.10: Amadeo Finger rehabilitation End effector [\[88\]](#)

The Amadeo finger rehabilitation device is a hand unit composed of six individual finger slides, four for the straight fingers and 2 for the left and right thumbs. The device also contains a hand-arm support interface along with supports for each individual finger and tapes to hold the fingers in place. The system also includes a height adjustable framework and an operating panel operating on a computer. The finger supports are connected to the actuation sliders through magnets. These sliders carry out finger extension and flexion [\[89\]\[90\]\[87\]](#).

The Amadeo system's position based control is the primary control method of the Amadeo rehabilitation system. During therapy, linear trajectories with position and speed values drive the fingers along the rails in predefined motion profiles. This allows the system to deliver both passive and active assistive training. In passive training the Amadeo fully drives the patient's fingers, while in active assistive training the patient is encouraged to contribute to the robot's movement. However, if the patient is unable to complete the exercise, the robot ensures exercise completion by

acting on the finger, essentially an assist-as-needed mode. The system also allows for the degree of difficulty of the active training to be tuned for each individual finger [86][73][91].

The Amadeo control framework is inputted by surface electromyography (sEMG) for movement initiation. With this solution, surface electrodes placed over the finger flexor and extensor muscle groups capture any sEMG activity, which is processed to extract muscle activation levels relative to a baseline recorded during calibration. Once the sEMG signal is detected, the controller interprets this as voluntary motor intent and triggers the corresponding finger movement. This allows for patients with preserved neural activation but insufficient mechanical force generation to actively participate in therapy [91].

The Amadeo system has also been extensively reviewed and studied in clinical trials, with results showing efficacy in finger motor capability improvement in patients who utilised this system in conjunction with conventional therapy. With several studies displaying improvements in FMA scores when comparing patients who utilised this system with patients who only underwent conventional therapy methods [92][90][93].

3.2 Exoskeleton based solutions

This section is dedicated to the analysis of exoskeleton based solutions for upper limb rehabilitation of stroke patients, one of the most common types of robotic based solutions for rehabilitation.

3.2.1 ARMEO Power

The ARMEO Power is an exoskeleton designed for upper limb rehabilitation for a set of conditions, including stroke. This device, seen in [Figure 3.11](#) provides a plethora of features for a more efficient recovery, including full arm weight support, full 3D motion, passive and active actuation and an interactive system for patient engagement[93][71].

This exoskeleton provides 6 DoF: 3 for the shoulder, 1 for the elbow extension, 1 for the forearm supination and 1 for the wrist flexion, providing a wide range of actionable areas of the upper limb and enabling full 3D motion, not limiting exercises to just one single plane [94]. This device is designed to envelop the entirety of the patient's arm, providing adjustable settings for shoulder height and limb length [69][93][71].

The ARMEO Power, similarly to the InMotion ARM, provides an assist-as-needed method of rehabilitation. Being able to configure the level of resistance or actuation needed to help the patient overcome the challenges necessary for rehabilitation. Furthermore, this technology enables the stabilisation of specific joints, enabling a modular approach to rehabilitation. This ensures all joints can be worked on individually or as groups as is needed along the recovery process [93][95]. The control scheme for this exoskeleton is not available in any source as of the writing of this document.

Exoskeletons are able to provide better results in regard to the outcome of rehabilitation of stroke patients in the upper limbs. Patients who undertook rehabilitation with the ARMEO Power



Figure 3.11: Armeo Power Exoskeleton [69]

Exoskeleton have shown statistically significant improvements over those who went through conventional therapy methods. With patients mainly showing improvements in FMA (Fugl-Meyer Assessment), showing particular improvement over conventional methods in the first few sessions [96].

3.2.2 EXO-UL7

The EXO-UL7 is a dual-arm exoskeleton designed for upper limb rehabilitation in stroke patients. This system provides rehabilitation solutions for the forearm, featuring two distinct methods of actuation, unilateral and bilateral. In bilateral training, the unaffected limb acts as the master, while the impaired limb operates as the slave, mimicking the motion of the contralateral side through PID control. The assistance provided is partial, offering guidance without enforcing precise symmetry—and motor torques are capped in software to prevent excessive force. Gravity compensation is applied only to the exoskeleton's own structure, not to the patient's arm weight [97][98]. Figure 3.12 displays the EXO-UL7 Exoskeleton.

Each arm comprises seven actuated degrees of freedom: three for shoulder movements (abduction/adduction, flexion/extension, internal/external rotation), one for elbow flexion/extension, one for forearm pronation/supination, and two for wrist control (flexion/extension and radial/ulnar deviation). This configuration supports full 3D motion, allowing for complex, multi-joint exercises that replicate daily activities [97].



Figure 3.12: EXO-UL7 Exoskeleton [98]

To enable interactive therapy, the EXO-UL7 is equipped with optical encoders for joint position measurement and six-axis force/torque sensors at key anatomical locations. These sensors allow the system to collect real-time biomechanical data during therapy sessions, which can be used to control virtual objects or games on a computer, fostering patient engagement [97]. A linear PID controller custom for the EXO-UL7 ensures semi-global asymptotic stability without requiring detailed modelling of the exoskeleton dynamics. This enables precise, stable motion tracking even in the presence of non-linearities and disturbances [98].

Clinical studies using the EXO-UL7 have demonstrated significant improvements in upper limb function post-stroke, particularly in joint range of motion, control accuracy, and voluntary motor activation. These outcomes, often measured through kinematic metrics and standard scales such as the Fugl-Meyer Assessment, suggest the device is a viable alternative or complement to conventional therapy [99].

3.2.3 MyoPro

The MyoPro exoskeleton is a myoelectric upper-limb, wearable rehabilitation exoskeleton. It features 2 degrees of freedom. This system is able to actuate on the elbow joint and the hand. It performs elbow flexion and extension along with hand grasp and release. This device also features four surface electromyography sensors (sEMG). These sensors allow for the measurement

of muscle activity and allow the system to actuate on an assist-as-needed basis with this data [100][101][102]. This system can be seen in Figure 3.13.



Figure 3.13: MyoPro upper limb exoskeleton [101]

The MyoPro features 2 degrees of freedom. As it only performs planar actuation of the elbow (flexion/extension) and wrist (grasp/release) joints. This device transforms muscle activation signals, read from the electromyography sensors, into muscle commands. The MyoPro has four distinct control modes: standby mode, biceps mode, triceps mode and dual mode. In the standby mode the sEMG sensors' data is ignored, and the motors remain inactive. For the three remaining modes, the system utilises input from the sEMG sensors to perform assist actuation. For both mode, it reads from the biceps and triceps sensor, allowing both flexion and extension assistance. The biceps mode it only utilises data from the biceps sensor, only allowing flexion motion. And finally, for the triceps mode it only reads data from the triceps sEMG and only allows extension movement [101][102].

This device is available commercially and has been the subject of review in several studies[73]. These studies have demonstrated significant clinical and statistical improvements in stroke patients that utilised the MyoPro upper limb exoskeleton to perform rehabilitation. Particularly with significant increases in mean points scores in the Fugl-Meyer Scale [102][103][104].

3.3 Summary of existing solutions

In this section, Table 3.1 summarising the analysed existing solutions will be displayed in order to summarise the interest points of the analysed solutions:

Table 3.1: Summary of robotic rehabilitation solutions analysed in Chapter 3

Device Name	Year	Type	Target	DoFs	Control Philosophy Sensors	Control Modes	Actuation	Testing Status
InMotion® ARM [73]	2009	End-effector	Shoulder–Elbow	2	Impedance / resistance control; position sensing	Passive; Active; Assist-as-needed	Electric	Tested on stroke patients
REHAROB [73]	2007	End-effector	Upper limb	2	Position control; force sensing	Passive; Active	Electric	Tested on stroke patients
MOTORE [73]	2010	End-effector	Upper limb	3	Position control	Passive; Active	Electric	Tested on stroke patients
AMADEO [73]	2015	End-effector	Fingers / Hand	5	Position control; sEMG-based intent detection	Passive; Active; Assist-as-needed	Electric	Tested on stroke patients
ARMEO® Power [73]	2011	Exoskeleton	Upper limb	6	Assist-as-needed; control scheme not disclosed	Passive; Active; Assist-as-needed	Electric	Tested on stroke patients and commercially available
EXO-UL7 [73]	2016	Exoskeleton	Upper limb	7	PID position control; angular sensors	Active; Assist-as-needed	Electric	Tested on stroke patients
MyoPro [73]	2017	Exoskeleton	Elbow / Wrist	2	sEMG-based intent detection	Assist-as-needed	Electric	Tested on stroke patients and commercially available

The reviewed solutions display a clear division between end-effector and exoskeleton architectures. End-effector systems tend to have fewer degrees of freedom and simpler mechanical structures, focusing on task-level movement generation, whereas exoskeleton devices replicate anatomical joints and generally incorporate a higher number of DoFs. This reflects a trade-off between biomechanical fidelity and system complexity. Exoskeletons aim for joint-level control precision, while end-effectors simplify alignment issues but sacrifice anatomical isolation.

Although, high DoF exoskeleton systems attempt to replicate the full kinematic chain of the upper limb, this increases mechanical complexity, alignment requirements, and control burden. Conversely, lower DoF systems targeting specific joints (e.g., elbow or wrist) may offer improved mechanical robustness.

Most commercial and research systems incorporate assist-as-needed control modes, highlighting the importance of adaptive interaction. However, only a limited number of systems implement advanced intention-detection strategies (e.g., sEMG-based control), suggesting that many platforms still rely primarily on predefined trajectory or impedance-based assistance. Furthermore, all solutions more deeply analysed are comprised of electrical actuation, with this being the primary method for most solutions found.

While several devices have demonstrated clinical feasibility, there remains heterogeneity in testing protocols and outcome validation. The translation from controlled experimental studies to standardised clinical deployment remains inconsistent.

Despite advances in robotic upper-limb rehabilitation, there remains a need for systems that balance mechanical simplicity, sufficient biomechanical relevance, and adaptive assistive control while remaining clinically deployable and economically viable.

3.4 Limitations of existing solutions

The development of robots for rehabilitation of the upper limbs in stroke patients has provided remarkable advances in recent years. Still, it shows many signs of being a vanguard technology

with many improvements but many shortcomings and limitations as well. In this section, these key limitations will be evaluated.

1. **Reduced motion or reduced DoF:** Many of the existing solutions have a clear limitation in their applicability due to the fact that solutions such as the InMotion ARM have limited space motion capacity, this example in particular being only able to operate in one single plane of motion. This is a key shortcoming that needs to be overcome in order to equate robotic rehabilitation to conventional therapy methods
2. **Insufficient Joint Simulation:** End effector systems are usually limited in their actuation, being able to only actuate on a single point greatly reduces their capacity to fully simulate correctly the joints in the human upper limb, with many solutions such as the InMotion or REHAROB only being able to partially simulate joint movements, and often being unable to block the movement of certain joints in order to do specific exercises for specific joints
3. **Cost and complexity of exoskeletons:** Exoskeletons are able to provide the best results in what regards robotic rehabilitation. However, this comes at a great cost, not only in financial cost but but also in complexity costs, often being very voluminous machines that require complex control and actuation systems that highly inflate its costs. This complexity also increases maintenance complexity and costs, making these systems extremely expensive compared to end effector based systems.

3.5 Summary

This chapter reviewed the current state of robotic systems for upper-limb rehabilitation, with particular focus on the structural and functional limitations of existing solutions. End effector based devices were shown to be usually constrained by reduced degrees of freedom and limited joint specific actuation, which restrict their ability to accurately reproduce complex, multi-joint upper-limb movements. In the opposite direction, exoskeleton-based systems provide improved joint alignment and motion fidelity but are often associated with increased mechanical complexity, higher costs, larger physical footprints, and greater maintenance demands.

Overall, the literature indicates a clear trade-off between motion accuracy and system complexity across existing rehabilitation approaches. These findings underline the need for rehabilitation devices that better balance motion versatility, joint specificity, system simplicity, and cost-effectiveness, thereby improving practicality and accessibility in clinical environments.

Chapter 4

Methodology

This chapter is focused on the requirements, design, methods, and processes required in the development of this dissertation. It is broken down into 5 different sections regarding the different design areas and requirements, a full system overview, and its existing limitations.

The primary focus of the work conducted in this dissertation was the development of an end effector to assist rehabilitation of the upper limbs in stroke patients. In order to perform this, it was necessary to utilise a manipulator as a reference. As such, the Kuka iiWA was used, due to its similarities to the Kuka LBR, a medically certified manipulator.

The development of this dissertation required the development of a full mechanical system as well as a control algorithm for the CubeMars AK60 KV80 and CubeMars AK70-10 KV100 Brushless DC Motors that were utilised for the actuation of the system. The mechanical design had to take into consideration all requirements and was constructed using aluminium profiles and 3D printed components, with the system being fully designed for this dissertation. It was iterated upon digitally until a final model was in place where no significant flaws could be identified in a 3D virtual environment. For its design, the CAD software Fusion by Autodesk was the main tool utilised. The control of the actuators was done through custom-built software utilising CAN communication between the user interface, a microcontroller and the motor itself.

The following sections will develop a comprehensive overview of the requirements and architecture, display the mechanical and electrical actuation systems and conclude with the existing limitations within the system.

The developed system can be broken down into two main areas: mechanical design and electrical design. The mechanical design encompasses all the mechanical design choices, such as material selection, mechanical movements and couplings. The electrical design comprises all the software and hardware required for system control and actuation, such as the actuator, actuator control hardware and software, end of travel limits and all necessary software for interaction with the system. These design solutions will be further developed in the following subsections of **Mechanical Design** and **Electrical Design**.

4.1 Requirements

In order to develop this dissertation, a set of requirements was required. At a basic level the system would need to, at least, be able to support an average human arm, forearm and hand and actuate on it. Ensuring that the system would be able to at least perform the most basic task required. However, in order to properly develop the project, more specific and objective goals would be required regarding necessary motion ranges and movements performed.

In light of such requirements, an interview with physiotherapist Sérgio Carvalho from *Clínica Fisiátrica da Foz* was conducted, where a more technical assessment of the requirements could be conducted. In the interview, it was learnt that the most common approach for stroke rehabilitation in the forearm was extension and flexion of the forearm in relation to the arm, accompanied by adduction and abduction of the forearm. Along with this, it was learnt that the most common way auxiliary attachments used in physiotherapy are strapped to the patient's limbs, with a simple velcro strap being among the most common solutions.

From all these points, it is possible to gather the final general requirements for the project:

- Able to support an average human arm
- Perform automatic actuation on the patient's arm
- Be able to be easily attached to the patient's arm
- Be able to perform Extension and flexion concurrently with adduction and abduction

This platform allowed the project to be more easily developed with a clear focus and goal in mind. Enabling it to reach the final stage as seen in the section below.

4.2 System Architecture

The two main areas mentioned above comprise the two main areas of system architecture. However, these can be further broken down in order to further develop a concise system architecture overview.

The mechanical system consists of 3D printed assemblies with structural interconnecting aluminium profiles for rigidity and better support, along with ball bearings in order to allow smoother rotations of the elbow and wrist axis. The electrical system is composed of the CubeMars AK60 KV80 Brushless DC Motor, its 24V power supply, the control system hardware that ensures communication between the computer and the motor, and the control software and user interface. A more detailed overview of the system can be seen in the block diagram below.

As observed in [Figure 4.1](#) the two main sections of the system can be further divided into smaller, more focused subsections. These subsections allow a clearer and more detailed explanation of the system parts. The mechanical system is broken down into 4 different subsections, each regarding the body part they attach to. The upper arm contains the arm mount attachment and its respective profiles, and the forearm contains the forearm/wrist mount attachment and its

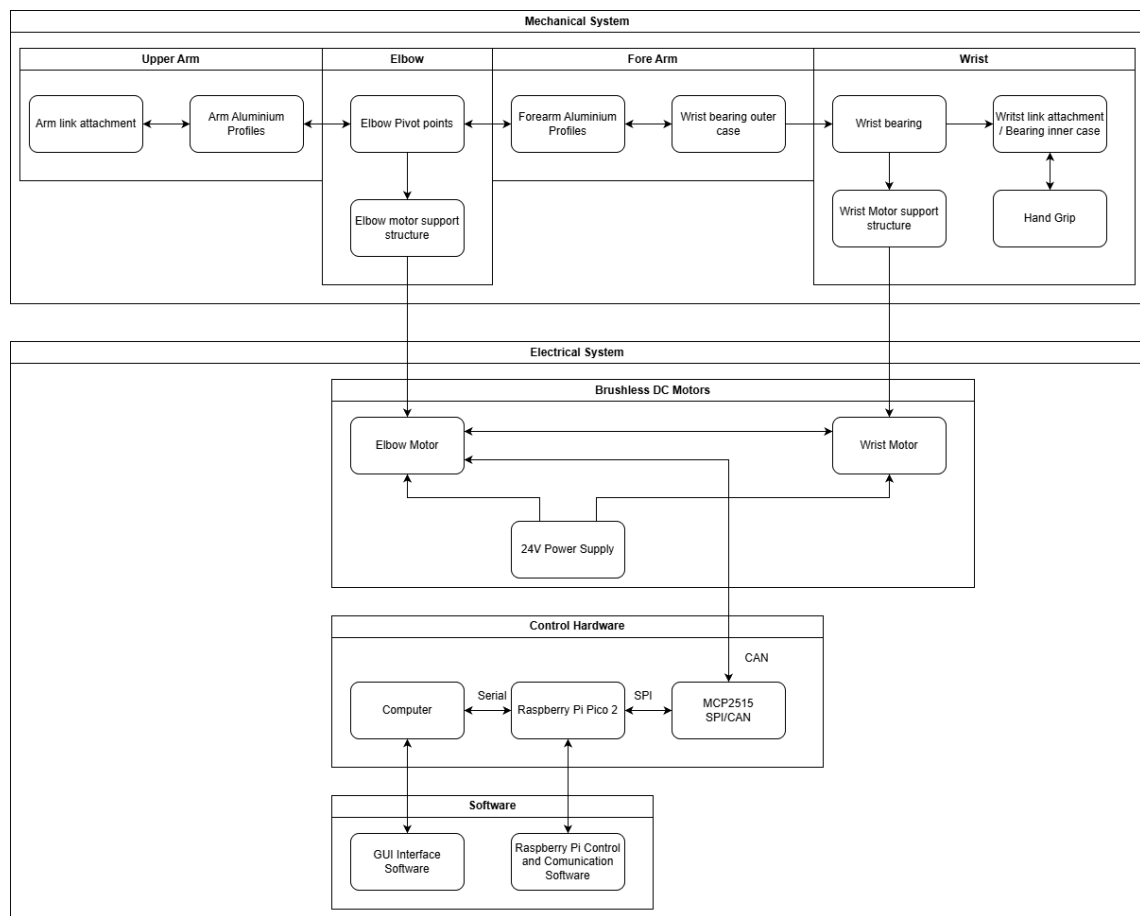


Figure 4.1: Block Diagram representing System Architecture

respective profiles. The remaining 2 subsections are in relation to the pivot points of the actuation system, the elbow and the wrist. These two points feature the two actuators, the elbow and the wrist actuators, which are responsible for the movements performed by the end effector.

The electrical system encompasses everything that relates to actuation, control and software. The 3 main parts of this system are the motors, the control hardware, and the software. The motors are comprised of the motor units themselves, including the hardware drivers, the power supply, and the CAN Bus that performs communication with the control hardware. The control hardware is comprised of the entire hardware loop that performs all communication between the user and the motor for control. This includes the interface device, the computer and the 2 communication modules, the microcontroller, Raspberry Pi Pico 2 and the SPI to CAN module MCP2515, this last one interfaces directly with the CAN Bus to communicate with the motor, it is comprised of a MCP2515 CAN controller and a TJA1050 CAN transceiver. That is the component that interfaces the CAN controller with the CAN Bus. Finally, the last subsection of the electrical system is the software that runs on the microcontroller to perform the motor control and the interface software that interacts with the operator and sends the respective commands to the microcontroller.

4.3 Mechanical Design

As stated in Section 4.1 **Requirements**, the developed end effector should be able to bear the weight of the human arm during actuation, as such the structure needs to be able to support the weight of the human arm, not only in a resting position but also during actuation, where a counterforce can be exerted. It should also be modular and easily fitted and adjusted to each patient. Due to the Kuka's 14kg payload limit, end effector weight is also among the key priorities for the end effector design, while keeping ergonomics and patient comfort in mind.

In pursuit of such goals, the design of the mechanical components of this end effector went through an iterative design while different solutions were evaluated. The entire mechanical assembly was designed and iterated utilising CAD software and analysed to check for flaws and failures of design. In the end, the design was finalised utilising 2 main materials. 3D printed PLA support and attachment parts and aluminium profiles for the beams.

4.3.1 Objectives and Constraints

The main mechanical objective in the development of this dissertation is the development of an ergonomic, practical and capable solution to actuate on the arm of a stroke patient while working with the existing constraints.

The end effector for this dissertation had the objective of actuating on two different joints. As such, early on, the decision to build a system that would actuate on the elbow and wrist was made. In order to perform this actuation, a pair of electric motors was also chosen to perform this actuation. The choice of electric motors allowed for a versatile and robust solution to perform actuation for different sets of requirements. Particularly, the necessity of active and resistive actuation that would allow the patient not only to be automatically actuated on but to apply force to the system in order for it to move.

The main constraint this system faced was the maximum payload capacity of 14 kg of the Kuka iiWA, this restriction limits the entire weight of the end effector plus human arm to this payload capacity. Which meant making a lightweight solution would be among the main objectives as well. This effort into reducing the system's weight would also mitigate the maximum required torque of the motors in order to perform meaningful actuation.

4.3.2 Design and Modelling Process

The main design tool used in the development of the mechanical parts of this system was Autodesk Fusion 360, selected for its robust parametric modelling capabilities and integrated assembly simulation tools, particularly suited for designing 3D printed components. Fusion allowed for rapid and iterative prototyping of structural elements, and its assembly suite allowed for comprehensive system simulation with constrained degrees of freedom. This ensured that all part fittings, clearances, and tolerances could be verified virtually before physical fabrication.

The mechanical design process was done through two major iterations. The first iteration was a fully 3D printed architecture and was structured using a top-down design approach. It began with the creation of the manipulator to end effector connection and the patient arm mount, followed by the elbow pivot mechanism and motor attachment, and concluded with the wrist interface. While this prototype served as a valuable source of inspiration and concept validation, it exhibited significant structural instability.

The second and final iteration introduced aluminium profiles as the main structural components while retaining some of the previously developed 3D printed pieces. This hybrid design significantly improved structural stability. A significant enhancement in this iteration was the introduction of an articulated wrist rotation system, allowing for independent wrist actuation relative to the elbow joint. This mechanism employed a large diameter ball bearing to support the wrist joint and an additional, smaller bearing at the opposite end of the elbow assembly to enable smooth, low friction motion. These improvements resulted in a system better aligned with the mechanical, ergonomic, and safety requirements defined in [Section 4.1](#).

The final design utilises six Bosch Rexroth 20×20 mm aluminium profiles as the main structural elements. These profiles run horizontally and parallel to the patient's arm, providing the necessary support for both of the motor housings and the wrist-bearing assembly. In addition, two smaller 20×11 mm profiles are included to allow for adjustable positioning of the hand grip, allowing the system to accommodate patients with different forearm lengths. The wrist and arm support pads are fully 3D printed. To secure the patient's arm in place, a simple velcro strap system is used at both contact points. [Figure 4.2](#) shows the CAD model of the built system.

Both motors are secured to the aluminium profiles using 3D printed mounting structures. These structures were designed to match the CubeMars motor's bolt pattern, utilising six M3 holes that accommodate standard M3 screws. The wrist motor is housed within a mounting structure fixed to the arm side aluminium profiles, while its rotor is connected to the forearm profile to enable forearm movement. The wrist motor assembly incorporates two additional 20×20 mm aluminium profiles, which are attached to the outer wrist-bearing structure and also supports the mechanical end stops that constrain wrist movement. Similarly, the elbow motor structure includes integrated mechanical end stops to limit the range of motion of the forearm and ensure safe actuation within appropriate limits.

The ball bearings are both press-fitted into 3D printed housings, using a crush rib design. This technique is commonly used when inserting metal components into plastic housings, as it allows for more forgiving tolerances while still achieving a tight fit. The cavity is modelled with a slight 3° outward angle, matching the bearing diameter at the base and gradually increasing towards the opening. Within this hole, six small ribs are symmetrically distributed to hold the bearing. During insertion, these ribs are crushed, allowing the bearing to be tightly and securely press fitted. A section of this design method can be seen in [Figure 4.3](#). This method ensures a tight press fit, ensuring mechanical stability and alignment. The design was validated through test prints to make sure that the tolerances would fit, with adjustments made to rib height and cavity tolerances as needed.

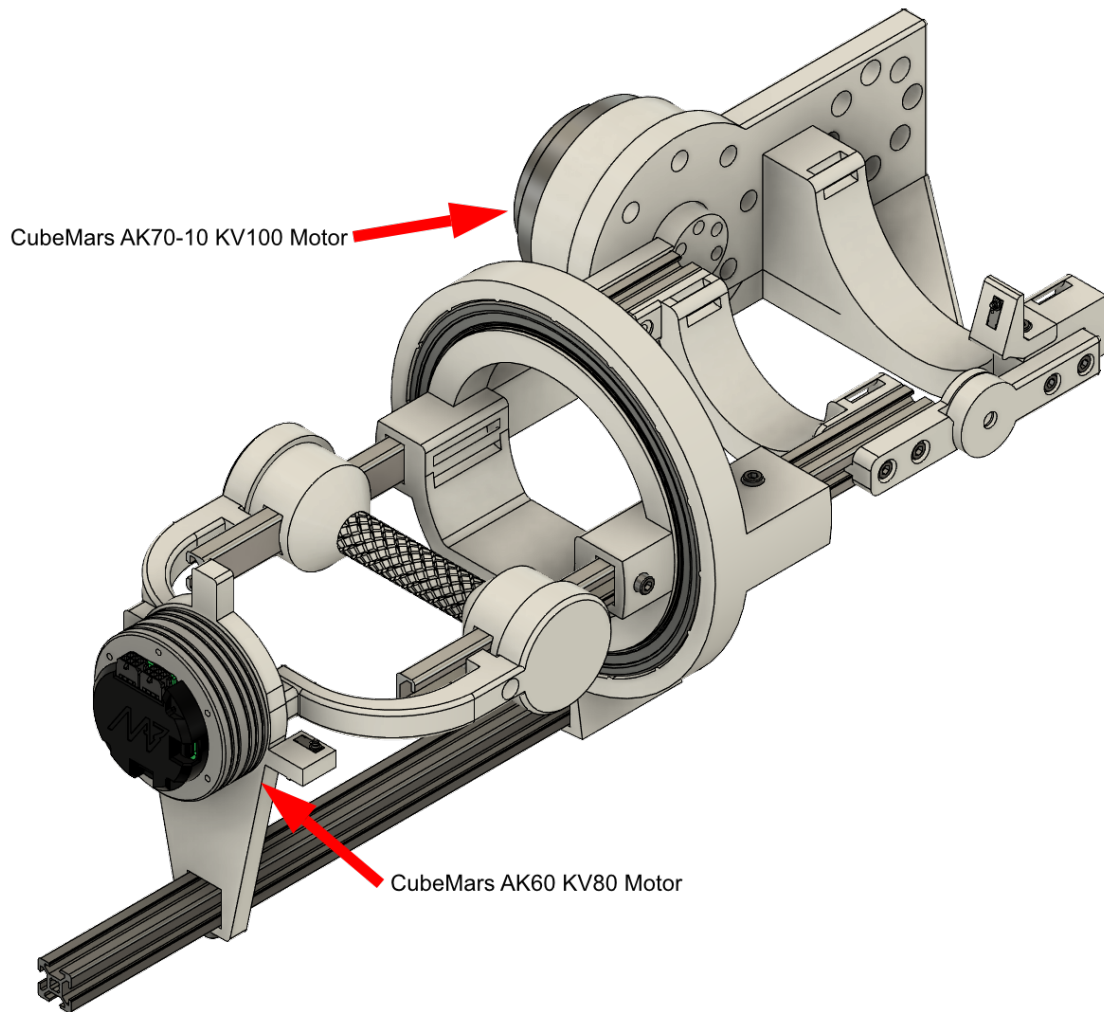


Figure 4.2: CAD Model of the end effector structure

4.3.3 Assembly and Integration

The final assembly of the end effector was executed in a modular way, allowing for easier part replacement and adjustment. Components were grouped into three main subassemblies: the upper arm structure, the forearm structure and wrist structure, along with the actuator and electronics module. Each subassembly was independently assembled prior to final integration.

All structural elements were fabricated using a combination of aluminium Bosch Rexroth 20×20 mm and 20×11 mm profiles and custom-designed 3D printed PLA components. The aluminium profiles were cut to the required length, and the 3D printed components, including motor mounts, bearing housings, and limb supports, were printed in the INESC TEC iiLab using a fused filament fabrication (FFF/FDM) 3D printer. All holes and fits were validated and manually post-processed as necessary to ensure mechanical alignment.

The 3D printed parts were attached to their respective aluminium profiles, utilising steel M5 Bosch Rexroth T-nuts for the respective profile size. In order to securely attach the T-nuts, profile

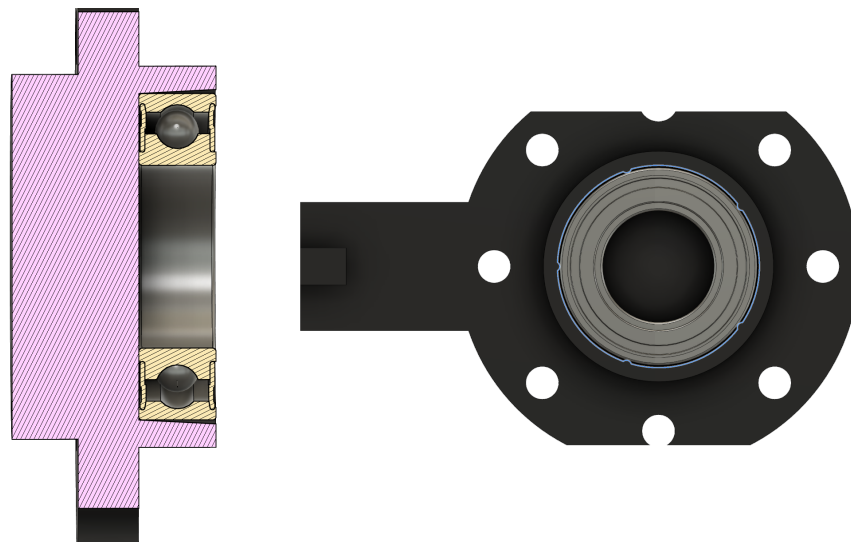


Figure 4.3: Crush rib design profile and front view

and 3D printed pieces, galvanised steel M5x10mm screws were utilised (DIN 912), this system also allows for part adjustment in order to accommodate different patient's arm lengths, although it requires loosening several screws in order to achieve this.

The CubeMars AK60-6 KV80 and AK70-10 KV100 motors were secured onto the structure utilising custom made supports that were fitted to the arm housing and wrist aluminium profile. Once secured, their rotors were attached to the respective moving segments, one driving the elbow joint and the other driving the wrist. Both the motor housing and the motor rotor were secured using 6 M3x6mm ball head galvanised steel screws (DIN 7985). The ball bearings were press-fitted into their housings using the crush rib technique, as described in [4.3.2 Design and Modelling Process](#), and alignment between moving and fixed joints was tested.

The electronics, seen in [Figure 4.4](#), featuring the Raspberry Pi Pico 2 and MCP2515 CAN controller, were located close to the end effector. However, it stayed separate from the end effector in order to keep the end effector's mass as light as possible. Special care was taken to ensure that wires did not interfere with the moving segments of the arm.

After physical integration, system-level tests were conducted to verify mechanical clearance, smooth motion of the joints, and correct actuation. Calibration routines were executed to set encoder offsets and end-stop detection, using the embedded microswitch for zeroing. Finally, communication with the user interface was established, and basic motion commands were issued to confirm full control loop functionality.

4.4 Electrical Design

As outlined at the start of this chapter, the chosen method of actuation for this end effector was a pair of brushless DC Motors, more specifically the CubeMars AK60-6 KV80 and CubeMars AK70-10 KV100 Brushless DC Motors. The choice of two electric motors would allow for key

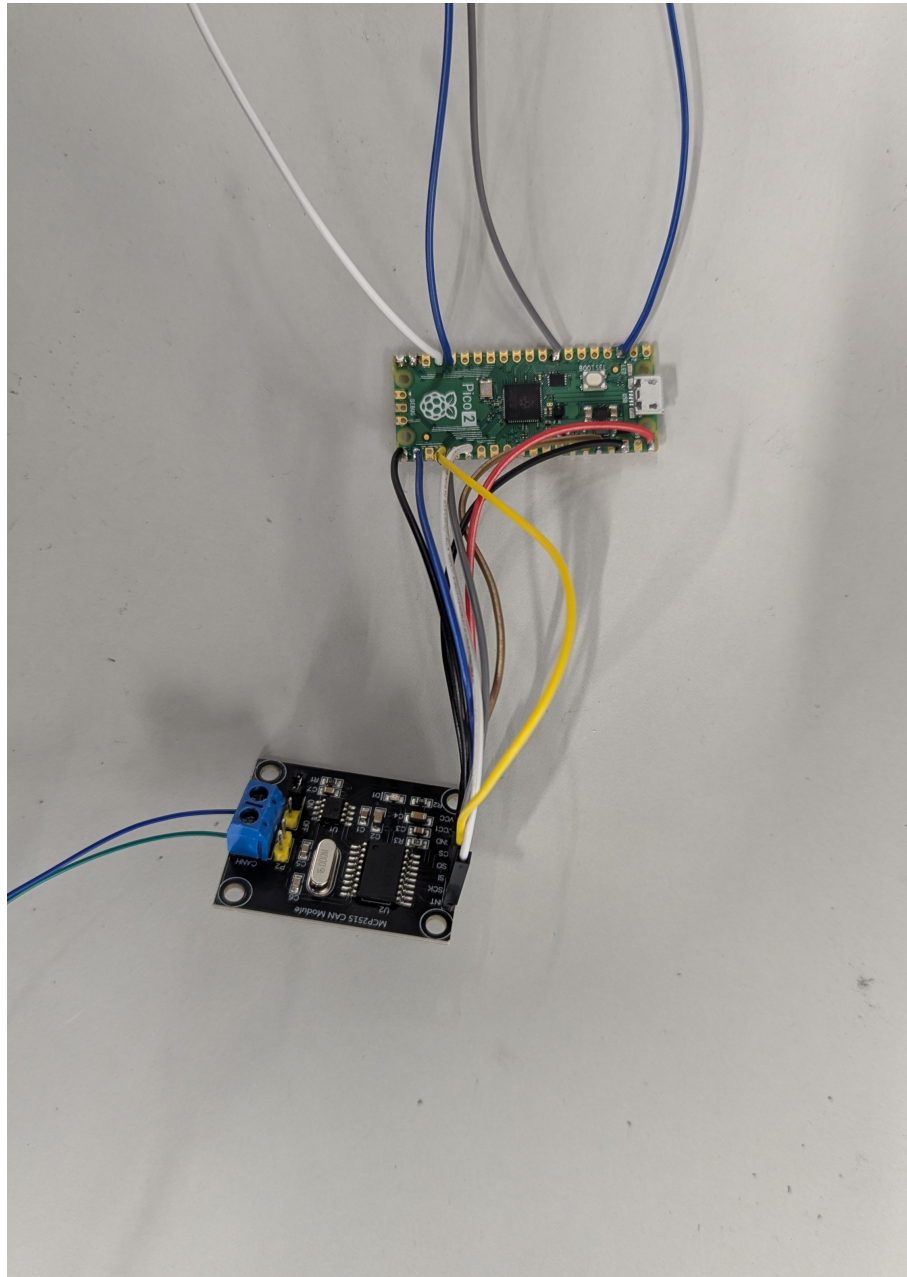


Figure 4.4: End effector Electronics

benefits such as controllable speed, position and resistance control, enabling the fulfilment of the project requirements for rehabilitation exercise versatility and variety.

To achieve these functional requirements, the two motors are integrated in an electrical system composed of three main subsystems: the motor power supply, drivers and communication, the control hardware and system software, as seen in [Figure 4.1](#). Each motor has its own driver, supplied by CubeMars, and is powered by a 24V DC Power Supply. Communication between these drivers and the microcontroller is handled through CAN bus protocol.

The microcontroller implemented is a Raspberry Pi Pico 2 microcontroller, which acts as

the main control unit for both motors, with CAN bus protocol communication being achieved through an MCP2515 SPI-CAN module to enable the microcontroller to directly emit CAN commands. This structure allows for reliable transmission of real-time commands. Finally, to enable the user real time control, the microcontroller software is interfaced with a computer through Serial communication, which runs a dedicated UI responsible for sending high level commands to the microcontroller and motors.

4.4.1 Actuation Requirements and Motor selection

The actuation requirements for this project were defined by the necessity to deliver controlled, safe and repeatable motion to the elbow and wrist joints to a recovering stroke patient during rehabilitation exercises. The system would also be required to perform both active and passive assistance to rehabilitation exercises, that is, active and resistive movement, supporting both modes in which the patient is either being passively moved by the end effector or requires applying above nominal force to move its arm. As such, the actuation solution would require precise control over speed, torque and position, with the ability to switch control modes for each exercise type.

Alongside the requirements, several constraints would also need to be considered, due to the mechanical structure employed and robotic manipulator used. The main constraint in this system would be weight, as the Kuka iiWA robotic manipulator employs a maximum payload capacity of 14 kg, with this capacity encompassing the end effector's mass along with the patient's arm mass. The motor would also be required to provide sufficient torque to the different segments so that they actuated on smoothly and safely, particularly accounting for patient applied forces, resistance and component friction.

In order to determine the torque requirements, it was necessary to calculate the maximum required torque that the system would need to have. In order to do so, a worst-case static analysis was performed to determine the required torque to move the elbow joint and wrist joint.

In this static analysis the forearm was modelled as a rigid body with a mass of 5.00 kg ($m_f = 5.00\text{kg}$) the 90th percentile reviewed in [Table 2.1](#). The length to the centre of mass of 0.18 m ($L_f = 0.175$). The end effector was calculated to have a mass of 3.04 kg ($m_e = 3.04\text{kg}$) and length to the forearm centre of mass of 0.21 m (0.21 m). The minimum required torque to perform any actuation on the forearm would be the torque required to overcome gravity in the horizontal plane ([Equation 4.1](#)), where the force of gravity is calculated by [Equation 4.2](#).

$$\tau_g = F_g \cos \theta \quad (4.1)$$

$$F_g = (m_f g + m_e g) \quad (4.2)$$

$$\tau_g(\theta) = \tau_f \cos \theta + \tau_e \cos \theta = (m_f g L_f + m_e g L_e) \cos \theta. \quad (4.3)$$

Computing this with the collected data, it is possible to get:

$$T_g(\theta) = \underbrace{(m_f g \frac{L_f}{2} + m_e g L_e)}_{\text{Nm at } \theta=0^\circ} \cos \theta = (5 \cdot 9.81 \cdot 0.175 + 3.04 \cdot 9.81 \cdot 0.21) \cos \theta = 14.85 \text{ Nm} \cos \theta. \quad (4.4)$$

As the force of gravity is higher in the horizontal plane ($\theta = 0.00^\circ$) the maximum required torque for the elbow joint is 14.85 Nm.

The wrist static analysis was computed using a similar method. However, in these calculations, only the wrist grip section of the end effector and the patient's hand need to be considered. Utilising the same data gathered in Table 2.1, we can gather that the hand mass is 1.43 kg ($m_h = 1.43 \text{ kg}$) and the end effector's wrist actuation pieces have a mass of 0.33 kg ($m_{ew} = 0.33 \text{ kg}$). For the wrist actuation the length of the hand is very small, as the handle and hand placement are almost in the same axis as the motor. This makes the necessary torque to overcome the force of gravity very small and therefore insignificant in this context.

In addition to overcoming the force of gravity, these motors are also required to overcome the minimum torque required to move these joints. Stoke, patients usually require more force to perform extension/flexion of the elbow joint. For healthy individuals the torque required for full extension of the passive elbow is around 3 Nm with stroke patients with severe spasticity requiring a value higher than that. But never higher than 5Nm [105]. As for wrist pronation/supination, it was found that 1.5 Nm of torque were necessary to perform full pronation/supination of the wrist joint in stroke patients [106].

From the data gathered, it can be concluded that the minimum torque requirements for the elbow joint are 19.85 Nm, and 1.5 Nm for the wrist joint. With the aim of fulfilling the requirements stated above, a motor study was performed.

Model	Manufacturer	Mass (g)	Nominal Torque (Nm)	Maximum Torque (Nm)	Rated Voltage (V)	Peak Current (A)	Communication Protocol
AK60-6 V1.1 KV80	CubeMars	368	3	9	24	22.7	CAN / UART
AK70-10 KV100	CubeMars	521	8.3	24.8	24/48	23.2	CAN / UART
AK10-9 V3.0 KV60	CubeMars	940	18	53	48	31.9	CAN / UART
RMD-X4-P12-10-C	My Actuator	330	4	10	24	19.5	CAN / EtherCAT
RMD-X4-P36-36-C	My Actuator	360	10.5	34	24	21.5	CAN / EtherCAT
RMD-X6-P20-60-C	My Actuator	820	20	60	48	29.1	CAN / EtherCAT
TS5003-10	RCDrone	250	3	4.5	24	>3.5 (not specified)	CAN / RS485
TS9012-6	RCDrone	763	12	37	48	>8.4 (not specified)	CAN / RS485
TS12020-10	RCDrone	1217	25	53	48	>11.5 (not specified)	CAN / RS485
YM070-210-R051-RH	Dynamixel	790	8.2	16.3	24	>11.9 (not specified)	CAN / RS485 / Modbus
YM070-210-R099-RH	Dynamixel	790	14.6	31.7	24	>11 (not specified)	CAN / RS485 / Modbus

Table 4.1: Comparison of Motors for End effector Applications (yellow lines specify the motors selected for this dissertation)

As displayed on [Table 4.1](#) a motor study was conducted in order to compare and evaluate the existing solutions on the market for collaborative actuation. Several different companies provide products in this market space, and some of the most commonly adopted solutions were reviewed. With varying different specs that would be able to fulfill the requirements.

In order to satisfy these requirements, the CubeMars AK60 KV80 Brushless DC Motor, shown in [Figure 4.5](#), was chosen for the wrist actuation and the CubeMars AK70-10 KV100 to actuate on the elbow joint. These motors provide an acceptable compromise between compact form factor, high torque output, weight, integrated sensors and control support along with good documentation and commonly used communication interfaces. The AK60 is rated for 3Nm of nominal torque and 9Nm of peak torque, making it capable of supporting typical wrist rehabilitation forces, even when accounting for wrist inertia or resistance. The motor also integrates a 6:1 planetary gear reduction with 0.55° of backlash, allowing to boost the torque output while maintaining smooth back drivability due to its relatively low reduction [\[107\]](#). Meanwhile the CubeMars AK70-10 KV100 BLDC Motor is rated for 8.3Nm of nominal torque and 24.8Nm of peak torque, allowing it to comfortably lift the entire forearm and end effector assembly at the elbow joint, in the most load critical scenarios. This motor, similarly to the AK60 also features a planetary gear reduction, with a 10:1 reduction with 0.12° of backlash, allowing a higher torque output while maintaining a low backlash angle and smooth back drivability[\[108\]](#).

However, one of the standout features of the CubeMars AK60 Motor is its compact and lightweight construction, weighing only 368 g and having a diameter of under 80 mm and depth under 40mm. This provides a significant boost to the weight reduction effort carried out during the development of this end effector, as well as due to its compact volume, it facilitates integration into the end effector structure. The lightweight nature of the motor helps ensure compliance with the Kuka manipulator's payload capacity while also decreasing the required torque to maintain posture or perform active movement of the patient's arm. With a maximum torque-to-weight ratio of 24.46 Nm/kg, the motor offers an ideal balance of power and efficiency for this kind of rehabilitation device [\[107\]](#).

This motor also features a low back drive torque of 0.2 Nm as well as a minimal backlash of 0.55°, this ensures smooth and compliant interactions with the user, particularly for interactions that require accurate feedback or impedance based control. Furthermore, the AK60 motor is also equipped with an internal 14bit magnetic encoder, providing detailed, high resolution position feedback for closed loop control. This precision is essential for rehabilitation protocols that require precise motion [\[107\]](#).

Similarly, the CubeMars AK70-10 KV100 motor also features a compact, lightweight construction, with a weight of 521g and a diameter of 89mm and a height of 50.25 mm. This is a significant aspect to consider in the aspect of creating a lightweight and compact end effector, creating the conditions for an easier integration to the end effector. Accordingly, it also helps integration into the Kuka manipulator, with their lightweight nature leading to a more comfortable gap to the payload ceiling of the Kuka iiWa capacity of 14kg [\[108\]\[29\]](#).

The AK70-10 motor features low backdrive torque and a small back drive of 0.48 Nm and a



Figure 4.5: CubeMars AK60 KV80 Brushless DC Motor [107]

backdrive of 0.12° . This ensures smooth and compliant interactions with the patient, especially for interactions that require accurate feedback or impedance based control. Furthermore, the AK70-10 motor is also equipped with an internal 14-bit magnetic ring encoder providing detailed, high resolution position feedback for closed loop control [108].

The motors are also designed with a wide operating range of conditions, being able to operate at nominal ambient temperature and within a -20°C to 50°C range. It also emits acceptable levels of noise, with it being specified that at a distance of 65 cm the perceivable noise is at most 60 dB. This provides a safe environment, being below the threshold for hearing damage while at the same time providing a non intimidating and friendly environment [109]. Finally, this motor supports CAN Bus communication, with CAN being a well known, reliable communication protocol used in many environments, allowing for straightforward integration into a control system, as well as UART communication [107][108].

The selection of the CubeMars AK60 KV80 and CubeMars AK70-10 KV60 motors was based on a comparative analysis of several commercially available actuators that are commonly used in assistive and rehabilitation robotics. Key criteria were:

- High torque-to-weight ratio, to ensure safe actuation of the human arm without placing excessive load on the KUKA manipulator's payload constraints;

- Compact and lightweight construction, to allow direct integration at the elbow and wrist joints;
- Integrated sensing and control support, namely through a high-resolution magnetic encoder and CAN Bus communication;
- Backdrivability and minimal backlash, critical for smooth, user-compliant interaction and safety in physical human-robot interaction.

Among other candidate actuators, the AK60 and AK70-10 offered the most balanced trade-off between torque capability, mechanical integration, sensory feedback, and cost. Its moderate reduction ratio (6:1) and (10:1) for the AK60 and AK70-10 respectively, also enabled backdrivable control while providing sufficient torque amplification. These characteristics made them particularly well-suited for a system designed to switch between passive and active rehabilitation modes, where both forceful assistance and user-initiated motion must be supported. Additionally, both featured a welcome weight balance, being among the lightest of the evaluated candidates while providing ample documentation for operation and communication, along with a competitive price to performance ratio [107][108][110].

4.4.2 Motor Driver and Communication

Each of the CubeMars motors is controlled by a CubeMars AK Series motor driver, which supports real-time torque, speed, and position control modes. These drivers have two default control modes: Servo mode or MIT mode. Each offering different features for different purposes.

Servo mode enables the motor to operate with 6 different control loops, including duty cycle, current (torque), speed, position, and position-speed-acceleration loop control. Each mode operates using internal PID control logic, with commands sent over CAN specifying exact set points. This mode is best suited for tasks that demand rigid control, repeatable positioning, or constant-speed execution, such as in rehabilitation applications where repeatability of movements is a key factor.

The PID control performed in this mode is available for the speed, position and current loop control modes, with the speed and current loops utilising PI control methods while the position loop utilises a full PID controller. However, the tuning of the PID is limited to CubeMars upper computer proprietary software, where the values for K_p , K_i and K_d need to be set and saved to the motor, these values are not able to be changed through any other method. However, the manufacturer's motor configuration tool already provides an automatic calibration method that sets up these values.

MIT Mode is designed for simpler control. Instead of sending discrete set points, the controller specifies a desired position, along with stiffness (K_p), dampening (K_d), and optional feed-forward torque. These parameters allow the end user to determine how "rigid" or "soft" the actuator behaves, simulating a spring-damper system. This allows for low-impedance behaviour. These characteristics are very basic, and while they could be adequate for safe human-robot interaction,

they fail to provide all the necessary control variables to perform assist-as-needed rehabilitation, which is critical to the purposes of rehabilitation in stroke patients[110].

Given the focus of this project on rehabilitation and patient safety, Servo mode was selected as the primary control method. Unlike MIT Mode, which can only utilise one control loop at one single moment, Servo Mode allows for multiple control loops (e.g. speed, position, and acceleration) to be simultaneously utilised. This enables more precise and adaptable behaviour, as it is required in compliant assistance. This is essential for stroke rehabilitation, where the system must not only assist with motion but also allow voluntary user input and adapt to variable resistance or user effort. The ability to utilise different control loops simultaneously in real time enables the system to modulate support based on clinical needs, from active to passive to assist as needed [110].

In both MIT and Servo modes the driver communicates with the microcontroller through a CAN Bus. This provides a robust and reliable way to communicate with the motor in real time. Each driver is assigned an unique CAN ID in order to allow for multiple motors to be connected to the same CAN Bus, and listens for 8 byte standard data frames containing the command values. In MIT Mode, the CAN message structure includes fields for target position, stiffness (Kp), dampening (Kd), and feed-forward torque, all encoded as 16-bit integers. These values are interpreted internally by the driver to generate actuator responses. The driver also transmits feedback messages at a rate of 1-500Hz that include current position, velocity, applied torque, and temperature, allowing the controller to monitor system status and close the control loop. The Servo mode CAN structure utilises extended CAN frames, where the different control loops are selected through a CAN identifier; this pack encodes both the motor's CAN ID and the selected control mode. Then the data payload is read for the corresponding control mode. For single control loop, the CAN frame contains a 4 byte integer representing the commanded variable. However, for the speed-position-acceleration mode the full 8 byte payload is used to send all 3 command variables in a single frame. Servo mode also enables the transmission of feedback frames containing the encoder position, speed, output current, temperature and the error state of the driver.

The CAN bus operates by default at a baud rate of 1 Mbps, although this can be changed to lower values, and communication is initiated by the controller. This configuration ensures timely command delivery and reliable two-way communication, which are essential for executing real-time control in patient-facing rehabilitation systems [110].

4.4.3 Control Hardware

The control hardware of the system is designed to serve as a compact and reliable embedded platform capable of issuing real-time motor commands, receiving feedback, and managing communication between the rehabilitation device and the user interface. It is composed of a Raspberry Pi Pico 2 microcontroller paired with an MCP2515 SPI-CAN controller and TJA1050 CAN transceiver, which together handle the core responsibilities of CAN communication with the Cube-Mars motor drivers. The microcontroller operates in a continuous loop, receiving serial commands

from the computer, sending position and torque commands in Servo to the motor while simultaneously sending back to the user incoming feedback frames. The wiring of this system can be observed in [Figure 4.6](#).

The hardware architecture is centred around a Raspberry Pi Pico 2 microcontroller running the Pico 2 SDK framework, selected due to its cost-effectiveness, small form factor, and extensive support for embedded C development, diverse peripheral interfaces (i2c, SPI, UART), low latency real time processing, and extensive community that allows for easier debugging of issues. In order to enable CAN communication, which is required for interfacing with the CubeMars AK Series motor drivers, the microcontroller is paired with an MCP2515 SPI-CAN control module. Due to the Pico's lack of native CAN controller, this external module handles the framing and transmission of CAN messages over the physical bus, with communication established via the SPI bus on the Raspberry Pi Pico 2. The CAN network operates at a baud rate of 1 Mbps, and the MCP2515 is configured with standard 11-bit identifiers to enable compatibility with the motor drivers' protocol.

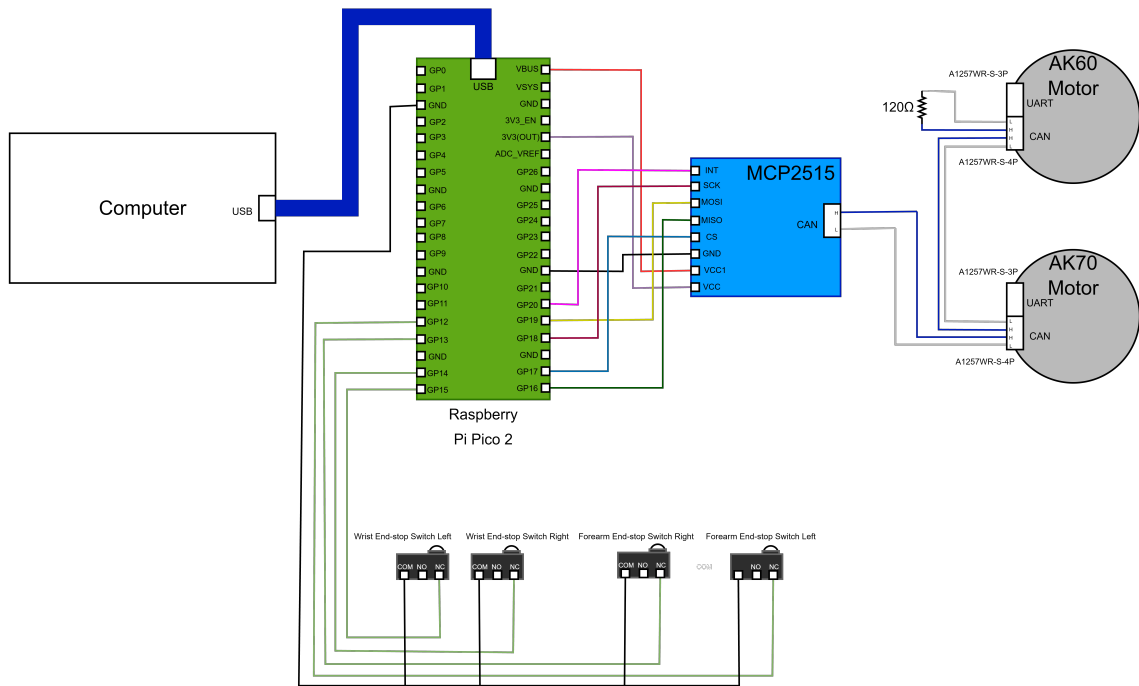


Figure 4.6: Wiring Diagram of the microcontroller and peripherals

In this configuration, the microcontroller executes a real-time control loop that transmits motor commands in Servo Mode and concurrently listens at a fixed frequency for feedback messages from each driver. These messages contain position, velocity, temperature, current, and error status, allowing the controller to check actuator performance and trigger safety commands if required. The command packets are constructed within the firmware by encoding control values (e.g., position, current, speed, torque) into an 8-byte CAN message, formatted according to CubeMars' specification.

In order to facilitate human interaction, the control hardware is connected to a host computer via a USB-based UART serial connection. This link supports a custom-designed UI, through which the operator can issue commands such as movement profiles, mode selection, or calibration routines. These commands are read by the Pico and converted into appropriately scaled Servo Mode frames for transmission to the motor drivers.

Additionally, the microcontroller interfaces with a small set of physical inputs, specifically the D2HW-A201D Omron Electronics microswitch (Figure 4.7), with the task of calibrating encoders and invoking end of line stop conditions. This microswitch was selected due to its compact dimensions ($12.8 \times 6 \times 5$ mm) and reliability, making it particularly useful for the compact and lightweight nature of the end effector. These inputs allow for physical intervention of the mechanical components of the system during system activity without relying solely on software commands.



Figure 4.7: Omron Electronics D2HW-A201D microswitch [111]

In summary, the control hardware was designed to balance simplicity and real-time responsiveness, enabling reliable communication and precise actuation within a safety-critical environment. By combining the communication capabilities of the Raspberry Pi Pico 2 with the CAN communication features of the MCP2515 module, the system achieves low-latency bidirectional control over the CubeMars motors while maintaining a compact footprint. The utilisation of serial communication with the host computer and hardware based inputs ensures flexibility, robustness and safety during operation, facilitating both automated routines and safety critical physical interven-

tions. Overall, the control hardware architecture provides a solid and adaptable foundation for the system's closed-loop motor control and user interaction requirements.

4.4.4 Software Architecture

The software architecture implemented on the Raspberry Pi Pico 2 microcontroller is responsible for executing the low-level control loop of the system, coordinating real-time communication with the motor drivers, and managing interaction with the user interface hosted on an external computer. The control loop is represented as a fluxogram in [Figure 4.8](#). It is structured to operate reliably under time-constrained conditions, ensuring reliability in tasks such as command transmission, feedback parsing, and motor coordination.

The main control loop when executed performs three main tasks:

1. Read local (hardware) inputs, such as the end stop micro switches;
2. Reading incoming serial commands from the PC;
3. Formatting CAN messages to the CubeMars motor drivers in Servo Mode;
4. Transmitting the CAN frame to the motor driver;
5. Receiving, decoding, and relaying motor feedback;

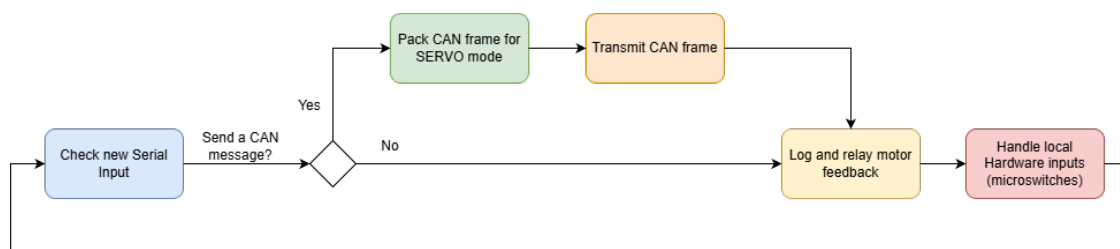


Figure 4.8: The firmware control loop for the Raspberry Pi Pico 2 microcontroller

High-level commands are sent from the user interface over a USB-serial UART connection and follow a simple text-based structure. These commands are read by the microcontroller and are used to update control parameters such as desired position, Torque or Assist-as-needed parameters. The structure of each command follows the same logic: Starting with a string related to the command type and being followed by three space separated floating point values with two decimal spaces. These floating point values transmit the relevant parameters for each command. This format allows the firmware to reliably interpret and read the incoming strings using standard functions (i.e., `sscanf`), allowing for effective decoding and flexibility in command handling.

In Servo Mode, communication between the control system and the CubeMars motor drivers is performed via the CAN bus using a 29-bit extended CAN identifier. Each command sent to the motor is associated with a specific packet type defined in the protocol (e.g., `CAN_PACKET_SET_RPM`, `CAN_PACKET_SET_POS`, `CAN_PACKET_SET_CURRENT_LOOP`, etc.), which is

combined with the motor's unique ID to form the final CAN ID. These CAN Frame modes can be seen in greater detail in Table 4.2. These identifiers are constructed in the format `(PacketType << 8) | MotorID`, enabling targeted communication with multiple devices on the same bus. The control parameters, such as target RPM, position, current, duty cycle, or acceleration, are encoded as signed integers into data frames, typically 4 or 8 bytes in length. For example, position commands are scaled by a factor of 10,000 and transmitted as 4-byte signed integers, while more complex commands like `POSSPD` include an 8-byte payload consisting of position, velocity, and acceleration. These values are formatted into byte arrays and transmitted using the `mcp2515_send_extended()` function, which manages low-level SPI communication with the MCP2515 module. The controller in turn handles the transmission of the data frame onto the CAN bus. The CAN Frame extender format is also displayed in Table 4.3.

The speed values are read in ERPM, electronic rotations per minute. This value cannot be directly translated to RPM as such a conversion is necessary. ERPMs are calculated from multiplying the rotor's physical RPM by the number of pole pairs in the motor and the gear reduction value. Solving this equation in reverse, we can correctly calculate the RPM for each motor. Both equations are displayed below. In the AK60 motor 1 RPM is equivalent to 86 ERPM and for the AK70 motor 1 RPM is equivalent to 210 ERPM.

$$RPM = \frac{ERPM}{Pole\ pairs * Gear\ reduction} \quad (4.5)$$

$$ERPM = RPM * Pole\ pairs * Gear\ reduction \quad (4.6)$$

In addition to command transmission, telemetry reading from the CubeMars motor drivers is handled by continuously polling the CAN bus for incoming messages using the MCP2515 module. Within the firmware, this is achieved through the `mcp2515_receive_frame()` function, which checks the CAN interrupt flags, retrieves pending frames from the receive buffers, and clears the corresponding status bits. In Servo Mode, telemetry is periodically broadcast by the motor driver and includes joint position, velocity, motor current, temperature, and fault status. The application extracts the full data from these raw data frames, and then the `servo_decode_and_print()` function decodes the data and assigns, based on position, each of the data points to its relevant position in a string, which is then sent over serial to the UI. In parallel, the `servo_decode_to_feedback()` function stores the decoded position, speed, and current values in internal state variables associated with each motor. This allows for continuous monitoring of joint position and motor status monitoring during operation. The feedback loop supports real-time monitoring while also enabling more advanced features like assist-as-needed logic, where actual position is compared to a desired trajectory and motion is applied if a deviation or delay is detected. By leveraging this feedback architecture, the system ensures responsive and safe closed-loop control of the rehabilitation end effector in real-world use cases.

The controller software also supports physical input events, such as end-stop detection through a microswitch input. These events are checked during each control loop iteration and can override

Table 4.2: CubeMars Servo Mode CAN Frame Modes [110]

Mode ID	Mode Name	CAN Frame Data Format (8 bytes)
0	Duty Cycle	[0-3]: Duty Cycle $\times 100000$ (float $\rightarrow$ int32) Range: -1.0 to 1.0
1	Current Loop	[0-3]: Current [A] $\times 1000$ (float $\rightarrow$ int32) Range: ± 60 A (This Value can be lower depending on the motor's maximum rated current)
2	Current Brake	[0-3]: Brake Current [A] $\times 1000$ (float $\rightarrow$ int32)
3	Speed Loop	[0-3]: Electrical RPM (int32) Range: $\pm 100,000$ ERPM
4	Position Loop	[0-3]: Position [deg] $\times 10000$ (float $\rightarrow$ int32) Range: $\pm 36000^\circ$
5	Set Origin	[0]: 0 = temporary origin, 1 = permanent (dual encoder only) Remaining bytes unused
6	Position-Speed Loop	[0-3]: Position [deg] $\times 10000$ (int32) [4-5]: Speed [ERPM] (int16) [6-7]: Acceleration [ERPM/s ²] $\times 0.1$ (int16)

Table 4.3: CAN Frame Format Servo Mode (Extended Frame Format) [110]

CAN ID Bits	Field Name
[28] – [8]	Control Mode
[7] – [0]	Source Node ID

motor commands or interrupt system operation if safety conditions are violated.

In addition to the microcontroller's software, a custom-built UI, displayed in [Figure 4.9](#), was developed using Python and Qt to serve as the main operator-machine interface for the end effector. This interface communicates with the Raspberry Pi Pico over a USB serial connection using Serial text read protocol. The UI allows the user to issue commands such as setting the encoder's origin, position, resistance toggling, and emergency stop routines, among others, as described in [Table 4.4](#). Internally, the software encodes command strings based on user input and transmits them via the pyserial library. On the microcontroller side, the embedded software reads the input and constructs appropriate CAN frames for motor actuation, as stated above. The UI also provides access to a set of settings, where the desired values for actuation can be inputted. Motor responses

and system feedback are displayed in a window, with the encoder position being sent to a visual dial, where the position of each rotor can be visualized. Incoming messages are read and printed through a Serial terminal window as well. For testing and development purposes, the software also includes a fallback mode using a simulated serial port (DummySerial) in the absence of connected hardware. This interface provides a flexible and user-friendly platform for clinical testing, parameter tuning, and system debugging.

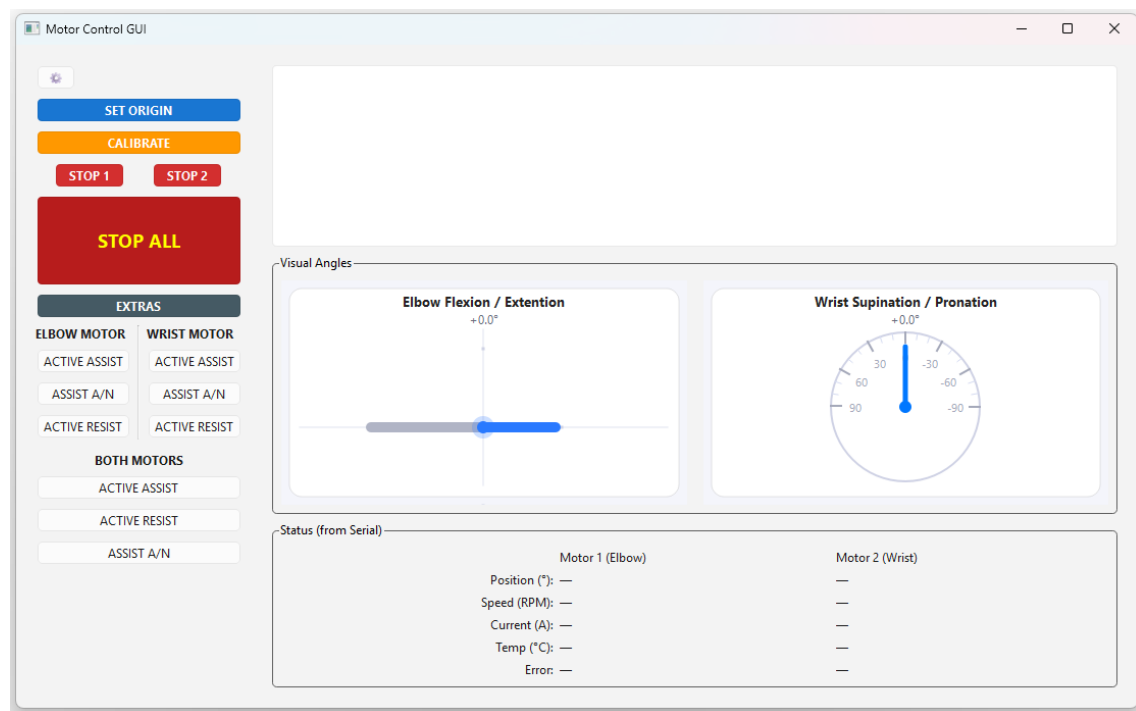


Figure 4.9: Computer User Interface

In order to accommodate these commands on the microcontroller side, a command state base control loop was built. A state diagram was created in order to allow for a better understanding of this loop, as seen in Figure 4.10. For each motor, the most recent command is assigned to the respective control mode and the main execution loop allows the system to continually perform actuation of any given loop until said loop either reaches a finish condition, is manually or automatically stopped by the user or end switches, or the system reaches the command timeout time. This continuous actuation is performed by the main loop continuously re-issues the corresponding CAN command (speed, position, position–speed–acceleration, current, or brake current) at run-time. Mode transitions occur when the microcontroller receives a new UI command (e.g., switching from speed to position control), when a STOP/STOPALL command is received, when a safety condition is detected (end-stop activation), or when a keepalive timeout expires; in these cases the controller forces the motor to a safe state by commanding zero current/brake current and returning the motor control mode to IDLE. Encoder origin setting is implemented as a one-shot command that is transmitted once when requested and does not persist as a continuous mode. In addition, the assist-as-needed routine maintains its own internal state and, when enabled, temporarily overrides

Table 4.4: Full serial command list and description

Command	Arguments	Description
STOP ALL	none	Sends STOPALL to stop all motors
STOP	<motor ID>	Sends STOP to stop the motor with the respective CAN id
POS	<motor ID> <POS>	Sends a position command in degrees to the respective motor
RPM	<motor ID> <RPM>	Sends an electrical RPM speed command to the respective motor
CURRENT	<motor ID> <I>	Sends a current command (in amperes) to the respective motor
BRK	<motor ID> <brake>	Applies braking current (in amperes) to the respective motor enabling the resist mode
AAN	<motor ID> <start_p> <end_p> <dur>	Executes Assist-as-Needed with start position, end position, and duration to the respective motor
PSA	<motor ID> <p> <v> <a>	Sends a Position-Speed command with position, velocity, and acceleration to the respective motor
TORQUE	<motor ID> <torque>	Sends a torque command in Nm to the respective motor
DUTY	<motor ID> <duty>	Sends a duty cycle command (0–100%) to the respective motor

the nominal motor mode until its completion criterion is met. This structure allows for an efficient and reliable operation of the motor, simplifying its operation on the firmware side.

The Assist-as-needed (AAN) routine is the most complex routine implemented. This routine enables the motor to support patient-initiated movement by dynamically responding to delays relative to a predefined linear trajectory. The routine receives a target motion defined by a start angle, end angle, and an expected movement speed in RPM, all inputted by the system's operator. During execution, servo feedback frames are continuously decoded, and the measured joint position is stored for each motor. At each control cycle, the expected position is computed assuming a linear progression over time, and the measured position is compared against the expected position shifted by 5 s. If the motor is detected to be at least 5 s behind the reference position, assistance is enabled and a Position–Speed–Acceleration command (PSA) is issued. The commanded target position is then moved gradually from the current measured position toward the final position using a ramping assistance factor, producing a smooth increase in assistance rather than an abrupt correction. While AAN is active, it has priority over other command modes for that motor, and it terminates when the end position is reached within a tolerance, when a STOP command is issued

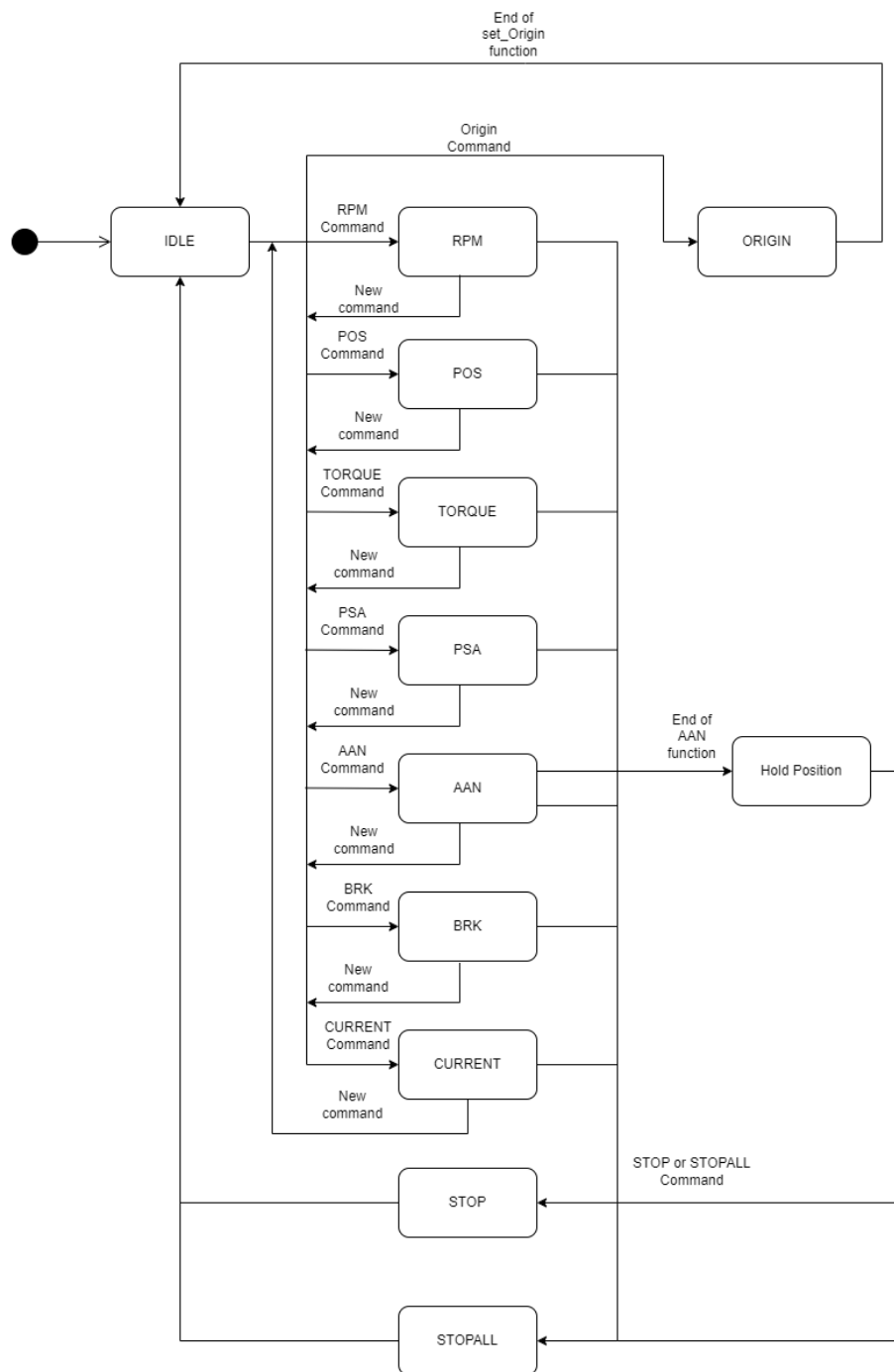


Figure 4.10: State based Control Loop Diagram

or an endstop switch is pressed. On the control UI level, this routine is complemented by a repetition factor. Where the system's operator can introduce the number of repetitions that are desired for the given exercise or enable unlimited repetitions until the STOP button is pressed.

Despite the AAN routine being the most complex routine implemented. Two other routines were implemented. One completely active routine, where the motor performs the full movement through its actuation, and another, fully passive, resistive routine, where the motor induces a brake

current to add resistance to the movement. The fully active routine on the controller side is implemented as a command to the motor in the triple control loop mode of acceleration, position and speed. These values are inputted by the system's operator. However, on the control UI level, this routine is complemented by a repetition factor. Where the system's operator can introduce the number of repetitions that are desired for the given exercise or enable unlimited repetitions until the STOP button is pressed, similarly to the operation mode of the AAN mode. The system operator is able to input the starting and ending positions of the movement and its speed and acceleration alongside the number of repetitions. In this context a repetition is a full "back and forth" movement, as in the motor moves from the start to the end position and back to the start position after 1 s at the end position. The passive resist mode is implemented as a brake current command. This command, as seen in Table 4.4 is sent as a current value in amps to the motor. However, on a practical level the objective is to select the resistance value in Nm. To achieve this, the system's operator inputs a value in Nm, and before this command is sent to the microcontroller, it's converted to a current value appropriate for the respective motor. In order to perform this calculation, the torque constant of each motor (K_t) is inputted in the following equation: $I = \frac{\tau}{K_t * N_g}$, where N_g is the motor's gearbox reduction value. The K_t constant values used were 0.123 and 0.078 for the AK70-10 and AK60 CubeMars motors, respectively.

The software architecture was designed to support safe, flexible, and real-time control of the rehabilitation system through a modular approach. The embedded software manages time-critical tasks such as CAN communication and feedback handling, while the external Python-based UI provides an accessible and intuitive control environment for the end effector operator. This separation of functions enhances maintainability and usability and allows for reliable user interaction without compromising system responsiveness. Together, the software components form a cohesive platform capable of supporting both experimental development and practical deployment in upper-limb rehabilitation scenarios.

4.5 Limitations

The developed system is able to perform all the goals set in Section 4.1. However it still is a limited system. With its limitation impacting several characteristics, such as clinical approval, performance and results.

In the first place, the system has not undergone clinical testing with actual stroke patients, the target demographic for such a system. While it was designed based on physiotherapy guidelines and expert input, its performance under real therapeutic conditions, particularly with regard to long-term use and therapist interaction, remains unvalidated. As a consequence, its medical efficacy and safety are assumed but not clinically approved.

Another limitation is the lack of direct closed-loop torque force feedback. While motor torque can be inferred from driver telemetry, by measuring the motor's current and utilizing the motor's

torque constant ($\tau = K_t * I$), the system does not include dedicated force sensors at the limb interface. This limits the platform's ability to perform true impedance control or detect unintentional resistance by the patient—features that are valuable in assist-as-needed rehabilitation systems.

Mechanically, the modularity and adaptability of the system are constrained by its fixed geometry and reliance on manually adjustable components. While the length of the hand support is adjustable, the system does not automatically adapt to radically different limb lengths or joint positions. This may hinder quick reconfiguration between patients or across therapy sessions. Furthermore, the use of common screws further hinders the ease of use across a varying range of patients.

From a software perspective, the user interface, although functional, lacks patient tracking, or parameter auto-tuning features. These would be highly desirable for rehabilitation tracking and therapist-driven adjustment during use.

Finally, the lack of a full protective enclosure or safety interlock mechanisms makes the system unsuitable for unsupervised use. Although motor parameters are adjustable in order to ensure soft actuation, there is a residual risk of unintended motion if commands are misinterpreted or improperly issued.

Chapter 5

Tests & Results

This chapter will dive into the experimental validation of the end effector. A series of tests were conducted to assess and verify the capabilities of the developed system. Ensuring this system is able to achieve the required goals.

This chapter is structured into four sections where the testing methodology, the tests performed, the obtained results and their analysis will be reviewed and developed.

5.1 Assembly

This section will display the process performed in order to assemble the entire end effector, from the mechanical and electronics side.

With the components gathered, assembly can be started. Once again the assembly process can be divided into the electrical and electronics assembly and the mechanical assembly. The mechanical assembly is the most complex part of this process, requiring the most amount of time and effort. In order to minimise the necessary tools for assembly, all the screws were chosen with a hexagonal socket head, requiring only 3 distinct screwdrivers in order to assemble all screws. A 4 mm hexagonal screwdriver for all M5 and M6 screws, one 2.5 mm hexagonal screwdriver for the M3 screws and a 6 mm hexagonal screwdriver for the M8 screw.

The mechanical assembly was performed in 3 sections that were then joined to for the full end effector. These were the Handle and AK60 wrist motor/front section (Figure 5.1). The wrist coupling/middle section (Figure 5.2) and the Arm structure/rear section (Figure 5.3). The following figures show the 3 sections assembled before being joined.

The assembly of the front section was performed by firstly connecting the 2 end-stop switches to the AK60 housing coupler. Secondly, the housing coupler was attached to the AK60 motor, utilising 6 M3x6 screws, followed by the attachment of the AK60 rotor connector, also utilising 6 M3x6 screws. The two handle connectors are connected to the rotor connector utilising 2 M3x6 screws each and finally the handle is pressed between the two handle connectors, finalising the assembly of this part.



Figure 5.1: End Effector
Front section



Figure 5.2: End Effector
Middle section

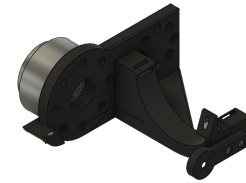


Figure 5.3: End Effector
Rear section

The middle section started its assembly by press-fitting the 61828-2rs1 bearing into the interior and exterior wrist structures. Once fitted, these should rotate independently and freely. Followed by the insertion and coupling of the 5 aluminium profiles. These utilise 1 T-slot nut each to ensure the profile gets secured to the structure. The 2x 20x20 120 mm profiles are inserted into the 2 parallel 20x20 holes in the exterior wrist coupling and secured with an M5x10 screw. The 350 mm 20x20 profile is fitted to the remaining 20x20 hole in the front of the same piece utilising the same method. The last 2 profiles, the 11x20 profiles, are attached to the interior wrist structure, also utilising a T-slot nut and M5x10 screw. Finally, the elbow structure is secured between the two parallel 20x20 profiles, and the AK70 rotor connector and elbow rotation front structure are connected to the outside of these profiles utilising 2x T-slot nuts and 2x M5x10 screws. To finalise the assembly of this middle section, 2 velcro straps are inserted into the interior wrist structure where the wrist will be placed and in the elbow structure.

The final section, the rear section, is the simplest section, consisting only of 5 parts. Their assembly is started by inserting the Kuka disc into its hole on the outside wall rearwards of the arm coupling. Secondly, the end-stop pieces are screwed into place. The top end-stop couples in the vertical M5 hole close to the elbow rotation structure, using the M5x10 screw, while the bottom end-stop is screwed down utilising the same 8 M3x12 screws that couple the AK70 motor housing. This piece needs to be aligned so that the end stop is horizontal in relation to the Arm structure. The 2 end-stop switches are also inserted into their places, and the elbow rear rotation structure is also screwed in with 2 M5x10 screws. Finally, the AK70 motor is secured to the Arm structure utilising the partially screwed M3x12 screws that hold the bottom end-stop. The arm coupling piece also utilises a velcro strap that needs to be placed in the piece.

To complete the assembly, all 3 parts need to be joined together. In order to perform this, the 91903-2rs1 ball bearing is press-fitted between the 2 elbow rotation pieces, and the M8x25 screw and nut are fitted and tightened afterwards in the hole in the middle of these pieces. Finally the AK70 rotor connector is attached to the AK70 rotor utilising 6 M3x6 screws. This joins the rear and middle parts of the end effector together. In order to connect the front and middle parts together, the handle must be inserted in the two 11x10 profiles and the AK60 housing coupler in the front 350 mm 20x20 profile. Once in the desired position, 3 M5x10 screws and M5 T-slot nuts are tightened to hold these pieces in place. This completes the assembly of the mechanical part of the end effector. In order to assemble the electronics and electrical part of the end effector, the

wiring diagram shown in [Figure 4.6](#) must be followed, and the two motors are powered by the 2 power cables.

5.2 Testing Methodology

This section will further explain the methodology utilised in the testing of the end effector. Particularly regarding the testing considerations, procedures, conditions, environment and execution.

5.2.1 Configuration and setup

In order to carry out any test, firstly there was the need to perform the necessary calibration and initialisation tests on the motors utilised in this project. In order to properly execute this, without any issues, a checklist was developed to ensure proper operation and initialisation of the CubeMars motors. This checklist can be viewed in [Appendix A](#). Along with the motor's initialisation, the mechanical system was also assembled separately.

As soon as the motors were properly configured and set up, and after the software was fully developed, the first tests could be performed. However, in order to ensure the safety of the user and the safeguarding of the equipment, firstly the motors were limited to parameters that ensured reliable, repeatable operation while not utilising more than the required force or speed for operation. In order to create these limits, two tool methods were used. Current and speed limitation. As seen in the previous chapter, with the knowledge of the motor's torque constant (K_t), the inputted torque can be calculated through $\tau = I * K_t$, and based on the calculations performed, in [4.4.1 Actuation Requirements and Motor selection](#), the required torque is already known. Based on this, the AK70 motor was limited to 8.5 A. The AK60 motor, due to its low nominal torque, did not require any current limiting. As for the speed limitation, both motors were limited to 30 RPM, or 1 revolution per 2 seconds. This limit was viewed as the maximum required speed needed for any given exercise.

On the software side, the configuration is able to be changed easily through the settings menu. This settings menu can be accessed from the main page of UI as seen in [Figure 5.4](#). This can be done at any moment before operation of the motors. These settings simply set the values that are sent in the serial commands to the microcontroller. In this menu it's possible to configure all the parameters of the 3 implemented work modes for each motor individually, as seen in [Figure 5.5](#). For the **ACTIVE ASSIST**, it is possible to configure repetitions, either until STOP is pressed (forever) or a limited number of repetitions. Afterwards, it's possible to configure the actuation parameters of the control mode. In these sections it's possible to configure the start and end positions, along with the desired movement speed. The position units are degrees, and the speed units are RPM. Finally, it is possible to configure the acceleration, however the value inserted by default already ensures a safe and soft operation and was chosen as a compromise between responsiveness and comfort. The **ACTIVE RESIST** only has one setting. The resistance torque setting. This setting allows the operator to choose the torque required to overcome the motor's resistance. The input is inserted in Nm. Lastly, the **ASSIST-AS-NEEDED** features setting is

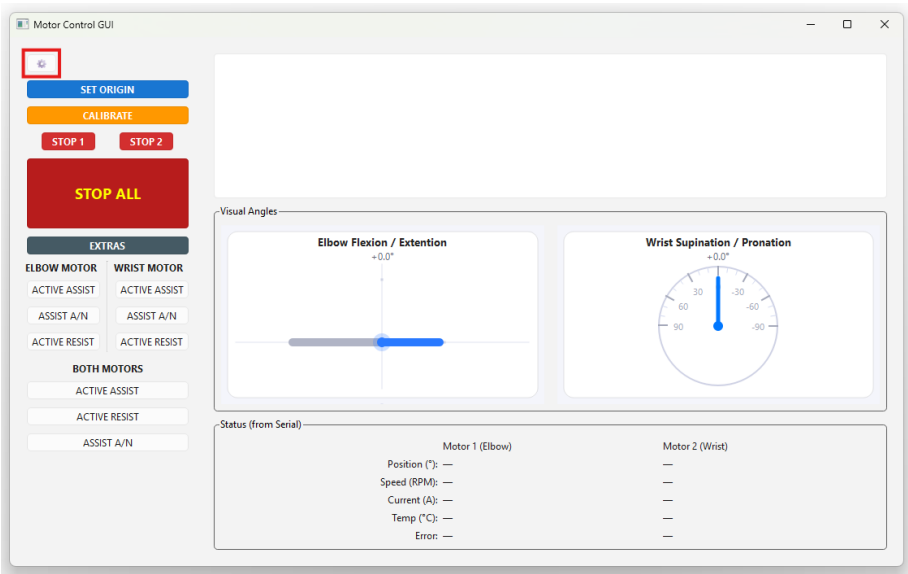


Figure 5.4: UI with settings button highlighted

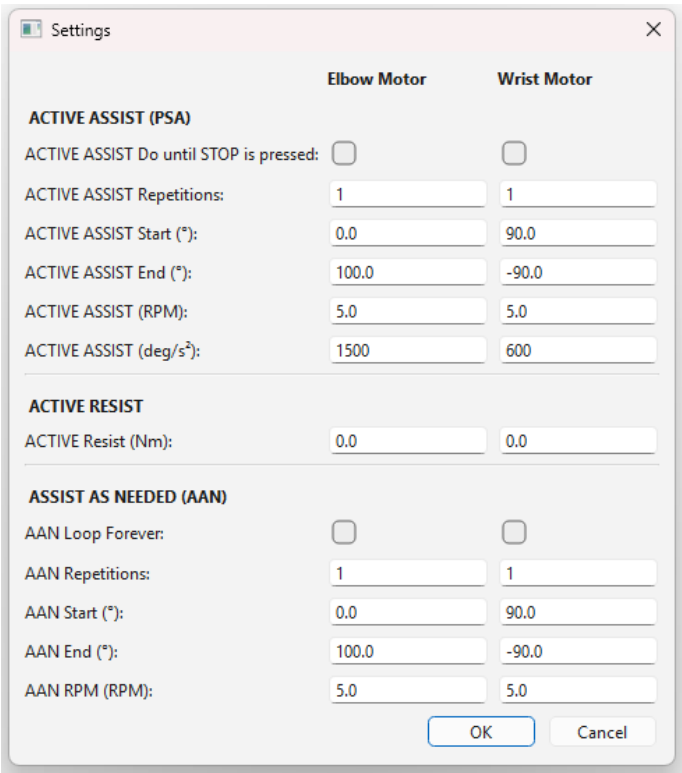


Figure 5.5: UI software settings

similar to **ACTIVE ASSIST**, the repetitions that are chosen in the same form as the **ACTIVE ASSIST** mode and the start and end positions are set in degrees. While the AAN mode is not mandatorily active, it can become active. As the name implies, it only provides assistance when the patient can no longer perform the motion to the end of the course, activating the motor and finishing the motion at the speed set in the speed setting, this input is also set in RPMs. The

settings are only saved once the OK button is pressed. Table 5.1 provides a more concise display of the settings here described. The settings are only saved once the OK button is pressed.

Table 5.1: UI Settings options list

Setting	Units	Description
ACTIVE ASSIST Do until STOP is pressed	N/A	Sets the number of repetitions as infinite or finite
ACTIVE ASSIST Repetitions	Integer number (0 or more)	In case the checkbox in the above line is unpressed, it set the number of repetitions to execute
ACTIVE ASSIST Start ($^{\circ}$)	degrees	Sets the start position for the movement
ACTIVE ASSIST End ($^{\circ}$)	degrees	Sets the end position for the movement
ACTIVE ASSIST (<i>RPM</i>)	Rotations per minute	Sets the motor's desired speed
ACTIVE ASSIST Start (deg/s^2)	degrees per second squared (deg/s^2)	Sets the motor's acceleration parameter
ACTIVE Resist (Nm)	Newton-meter	Sets the desired torque necessary to perform in order to overcome motor resistance
AAN Loop Forever	N/A	Sets the number of repetitions as infinite or finite
AAN Repetitions	Integer number (0 or more)	In case the checkbox in the above line is unpressed, it set the number of repetitions to execute
AAN Start ($^{\circ}$)	degrees	Sets the start position for the movement
AAN End ($^{\circ}$)	degrees	Sets the end position for the movement
AAN RPM (<i>RPM</i>)	Rotations per minute	Sets the motor's desired speed

Once these parameters are all set, the system is ready for operation while maintaining a safe and user friendly experience.

5.2.2 Operation

The operation of this system involves two distinct users: The patient and the system operator. The patient is the person who's the focus of the rehabilitation exercises. They have a passive role in the system's operation, only being required to perform the given exercises. The operator is the one who is responsible for setting the work parameters, enabling the work modes and monitoring the patient. Acting when needed on the physical end effector and interacting with the UI.

The operator is responsible for the active operation of the end effector. As such, most of this subsection will be directed towards the operation of the end effector from the operator's point of view.

Once the system is powered up and connected, the UI can be connected to the microcontroller. Once the system is live, telemetry can be observed in the telemetry fields, and a data stream is visible in the serial window in the top section of the UI window. The first step necessary for the operation of the system is origin calibration. Due to a limitation of the motor's encoder driver, the origin is not persistent. In order to mitigate this limitation, it is necessary to perform origin calibration. This procedure can be performed utilising 2 special pieces developed for this purpose. The wrist motor's pieces are designed to align the origin along the vertical axis of the motor rotation. This piece should be inserted between the vertical bar on top of the AK60 housing coupler and the AK60 rotor connector, holding both vertically in the same alignment as seen in [Figure 5.6](#). The calibration of the elbow motor is performed by placing a small rectangular piece between the bottom end-stop mount and the AK70 rotor connector and pressing both together, as seen in [Figure 5.7](#). Once both motors are align the operator should press the blue **SET ORIGIN** button in the UI and verify both position values are at 0.

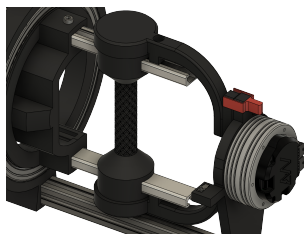


Figure 5.6: End Effector Wrist Motor's Origin Calibration

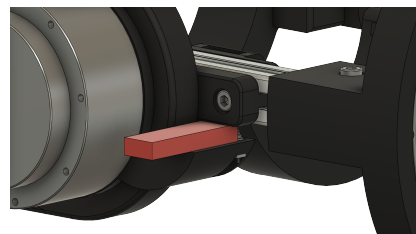


Figure 5.7: End Effector Elbow Motor's Origin Calibration

Following the completion of the origin calibration, the system can be operated to perform the rehabilitation exercises. The operation of these exercises is fairly simple. The operator has the three modes of operation listed on the left side of the UI and can choose to perform the exercise in a single joint and, therefore, a single motor, either wrist or elbow, or to perform it in both joints at once. These sections are composed of 3 buttons for each of the 3 combinations of joint actuation as seen in [Figure 5.8](#). The parameters for these actions are chosen in the setting menu as described in [5.2.1 Configuration and setup](#).

Besides the nominal operation modes and the **SET ORIGIN** mode, as is observed in [Figure 5.8](#), the operator also has at their disposal 3 **STOP** buttons. These buttons instantly stop any

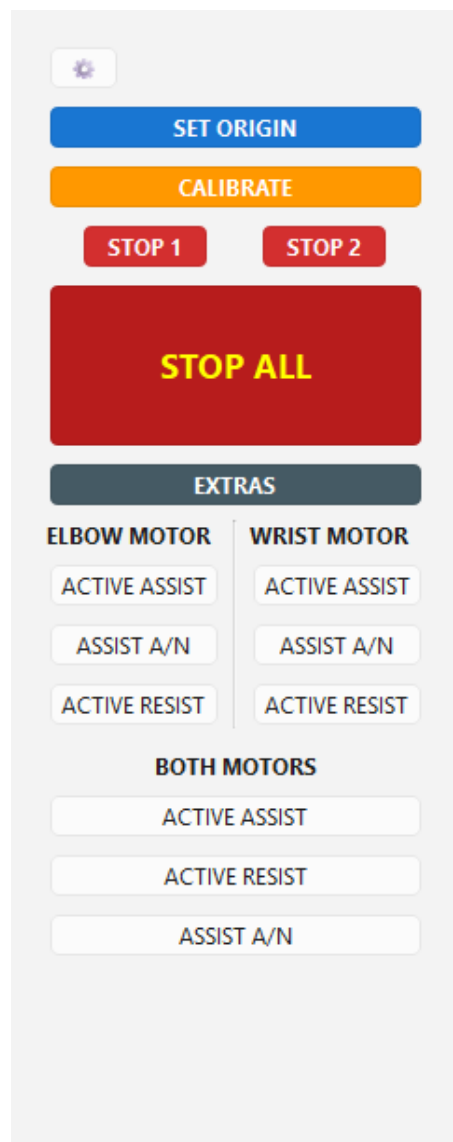


Figure 5.8: Operation modes in the UI

actuation the motors are performing. The operator can choose to stop each motor individually with **STOP 1** and **STOP 2** or both at once with the **STOP ALL** button. These buttons are one more layer of safety besides the end-stop switches and physical barriers. They allow the operator to intervene in a simple and responsive way to any observed malfunction or discomfort reported by the patient.

In terms of patient-side operation, as stated above, this is only passive. As in the patient is only required to do the exercises mandated by the operator. Having no direct intervention on the operational choices of the system.

5.3 Tests

This section is dedicated to describing the tests performed to evaluate the behaviour of the developed system in controlled conditions. Each test was performed to assess the performance of the end effector. The tests were structured so that their difficulty was gradually increased up to nominal end effector operation. These tests will be **No Load tests**, **Structure Tests** and **User Tests**. For each of these tests the 3 work modes (ACTIVE Assit, ACTIVE Resist and ASSIST-AS-NEEDED) will be analysed, for the No Load test 2 further limitation tests will also be performed.

Due to an encoder limitation, the position values on the Elbow (AK70) motor are inverted in the telemetry. This is due to the motor performing its actuation on the reverse direction. This has no impact on results. However it inverts the telemetry values in relation to the expected values

5.3.1 No load Tests

The first step in the testing programme here developed was the testing of the firmware, the UI, and its commands. In order to assess this, the most basic of tests was performed. A no load test would enable the extraction of basic data that would allow the assessment of the basic performance of the work functions. In this test, it would be possible to test the limits set in the firmware and basic motor operation, such as correct movement to position and function timing.

For these no load tests the motors were placed on their back and had the motor rotor connectors attached. In the case of the AK60 wrist motor, the AK60 housing coupler and handle were connected as well to enable end switch and handle integrity testing. The testing setup can be observed in **Figure 5.9** and **Figure 5.10**.

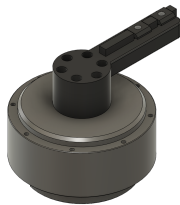


Figure 5.9: No load AK70 Motor test setup



Figure 5.10: No load AK60 Motor test setup

For these tests, the tests would be realised while performing the 3 previously described work modes. There were 3 main goals for the no load tests: Work mode conformity, End-stop switch testing, firmware limitation testing. As such several different parameters were chosen for the tests. For the first round of tests where the work modes were tested the following parameters, shown in **Table 5.2** were used.

The first test would allow the test of work modes, their conformity and performance. Eliminating the mechanical loads of the structure and gravity, this test would allow for the establishment of a baseline for the functionality of the implemented control systems for the actuators, independently of the structural load. The three following images (**Figure 5.11**, **Figure 5.12** and **Figure 5.13**) display the telemetry output of the motors during operation in each of the 3 modes. The

Table 5.2: Control parameters for elbow and wrist motors in no load test 1

Parameter	Elbow Motor	Wrist Motor
Active Assist (PSA)		
Active Assist repetitions	2	2
Active Assist start (°)	0.0	90.0
Active Assist end (°)	130.0	−90.0
Active Assist speed (RPM)	5.0	5.0
Active Assist acceleration (deg/s ²)	1500	600
Active Resist		
Active Resist torque (Nm)	2.0	1.0
Assist-As-Needed (AAN)		
AAN repetitions	-	-
AAN start (°)	0.0	90.0
AAN end (°)	100.0	−90.0
AAN speed (RPM)	5.0	5.0

motors were operated individually, however the data was joined into a parallel graph system for better organisation. They display the position in degrees, the speed in RMP and ERPM the current in Amps and the motor's internal temperature. The motor's internal temperature showed no significant data to be analysed and as such will be excluded from the displayed graphics.

For the two remaining modes, human interaction was required to verify their functionality. In ACTIVE Resist the motor is applied a brake current and requires force to be applied in order to display increased current application. As for the ASSIST-AS-NEEDED this mode can work without any human interaction; however, for further and more robust testing, the mode was utilised as it was designed. A movement was performed up to a certain point, and then the assist function was intended to take over. For the AK60 motor, the motor was rotated from the 90° position to approximately the -20° position and then the assist finished the movement, and for the AK70 motor it started at the 0° position and was moved manually to approximately the -100° position, and the assist functionality took over the actuation.

These results will be reviewed in the next section. [Section 5.4 Results Analysis](#)

Following the testing of the work modes. A second set of tests was performed to evaluate and validate the systems' set operational limits and safety features. In order to achieve this, 3 sets of tests were designed to be performed. A motor driver speed limit test, a motor driver current limit test and a controller firmware speed limit test. However, due to the low load. The driver current limit was not able to be achieved. Therefore, this test didn't yield any useful results and it won't be displayed. The parameters displayed in [Table 5.3](#) were used in these tests.

The first test to be performed was the driver speed limitation, this was set at 1900 ERPM (around 9 RPM) and performed in the AK70 elbow motor, and a ACTIVE Assist command was

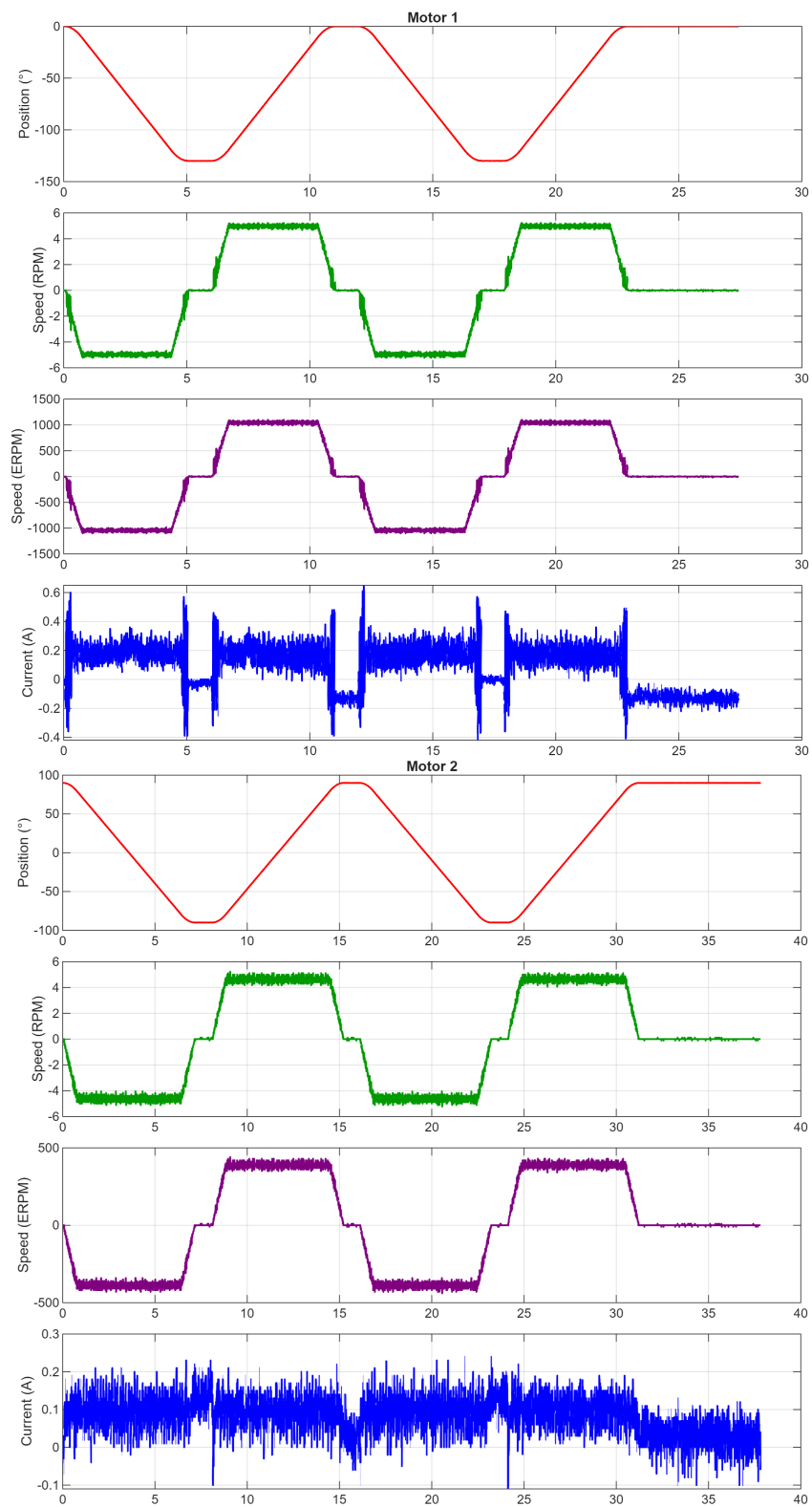


Figure 5.11: Telemetry results for ACTIVE Assist on the AK60 and AK70 motors without any load

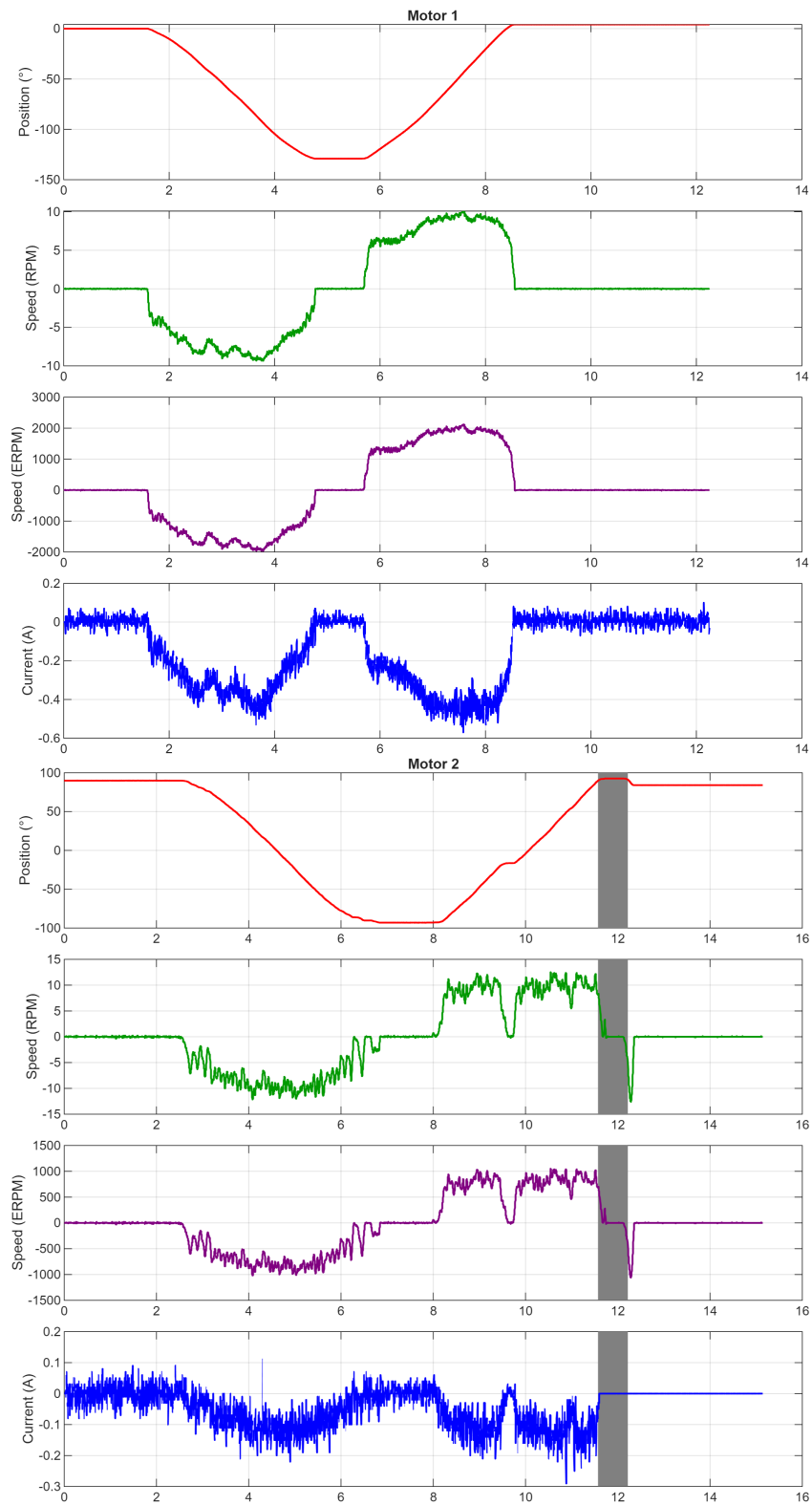


Figure 5.12: Telemetry results for ACTIVE Resist on the AK60 and AK70 motors without any load

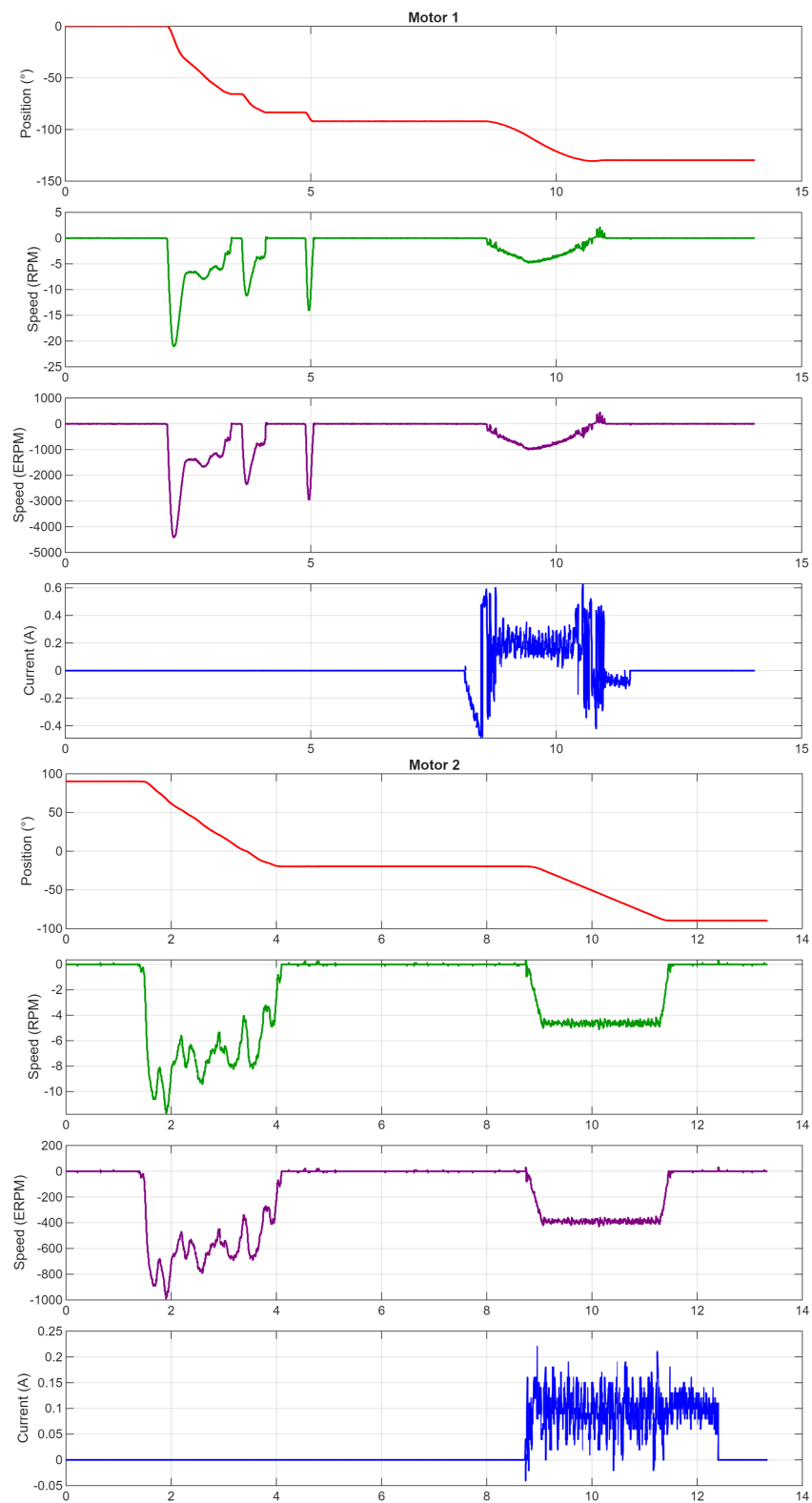


Figure 5.13: Telemetry results for ASSIST-AS-NEEDED on the AK60 and AK70 motors without any load

Table 5.3: Control parameters for elbow and wrist motors in no load test 2

Parameter	Elbow Motor	Wrist Motor
Active Assist (PSA)		
Active Assist repetitions	-	-
Active Assist start (°)	0.0	90.0
Active Assist end (°)	130.0	−90.0
Active Assist speed (RPM)	10.0	50.0
Active Assist acceleration (deg/s ²)	1500	2000
Active Resist		
Active Resist torque (Nm)	0.0	0.0
Assist-As-Needed (AAN)		
AAN repetitions	-	-
AAN start (°)	0.0	0.0
AAN end (°)	0.0	0.0
AAN speed (RPM)	0.0	0.0

sent with a speed value of 10 RPM. These results can be observed in [Figure 5.14](#). The second test, the controller firmware speed limitation (30 RPM) was performed in the AK60 wrist motor, and in order to achieve the required 30 RPM speed the required speed a 50 RPM speed command was input. The results of the second test can be observed in [Figure 5.15](#).

These results will also be reviewed in [Section 5.4.1 No load Tests](#).

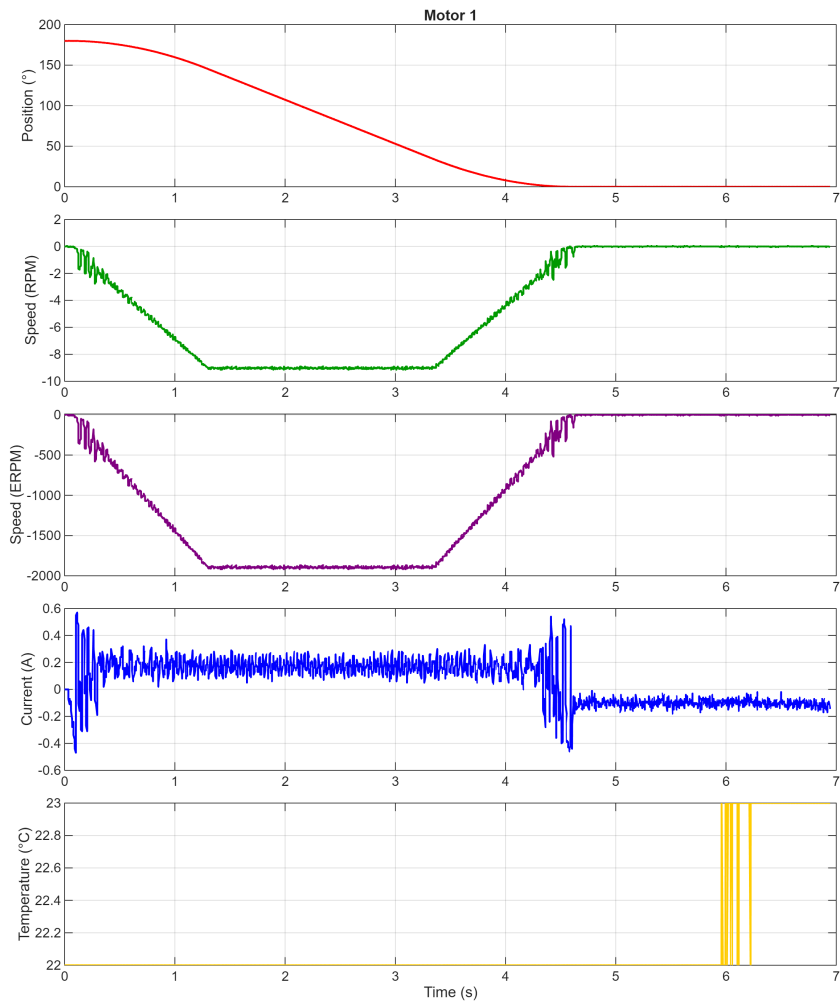


Figure 5.14: Telemetry results for ERPM driver set speed limit in the AK70 motor without any load

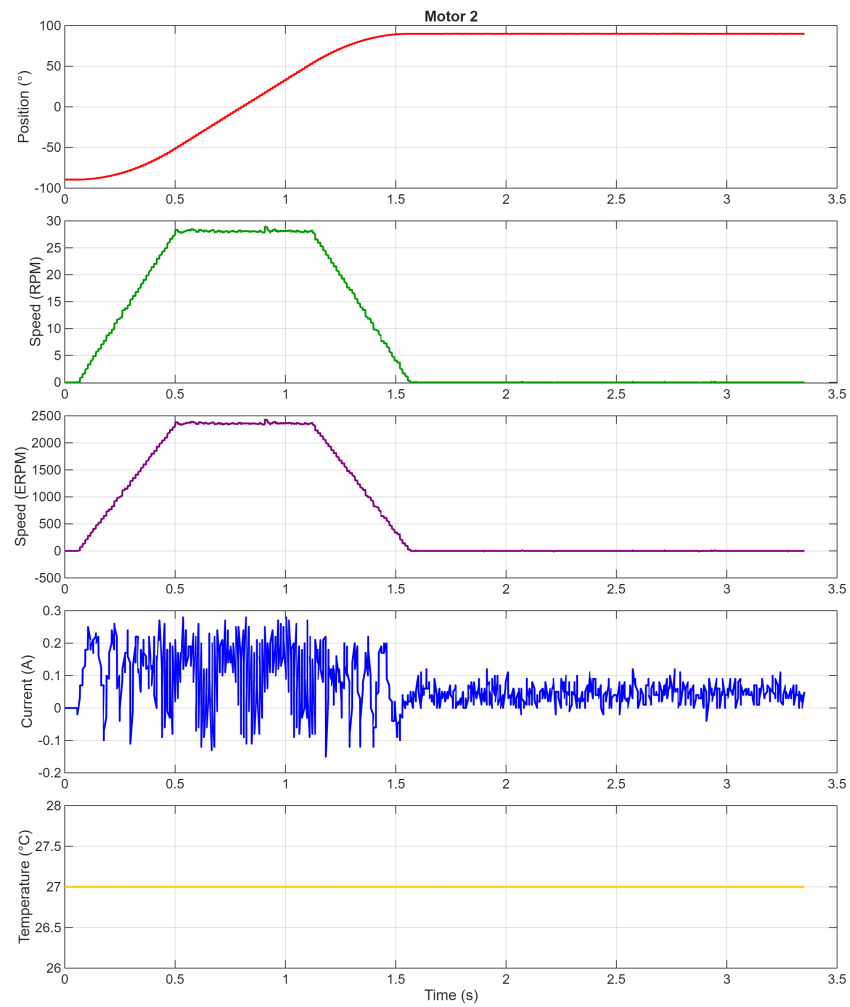


Figure 5.15: Telemetry results for RPM controller firmware set speed limit in the AK60 motor without any load

5.3.2 Structure Tests

Following the verification and validation of the work modes and limitations in isolation, a new set of tests was conducted with the actuators integrated into the end effector structure. This set of tests was designed to evaluate and validate the behaviour of the mechanical structure and the actuation on it.

In this configuration the motors were attached to their respective positions in the end effector structure, with this structure being fully assembled. With the actuation forces now being transmitted through the corresponding mechanical elements. The actuators were also submitted to the end effector's structure's mass and its gravity load. This would allow the tests to be safely performed, with focus on their actuation performance and feasibility. The structure test setup can be observed in [Figure 5.16](#).

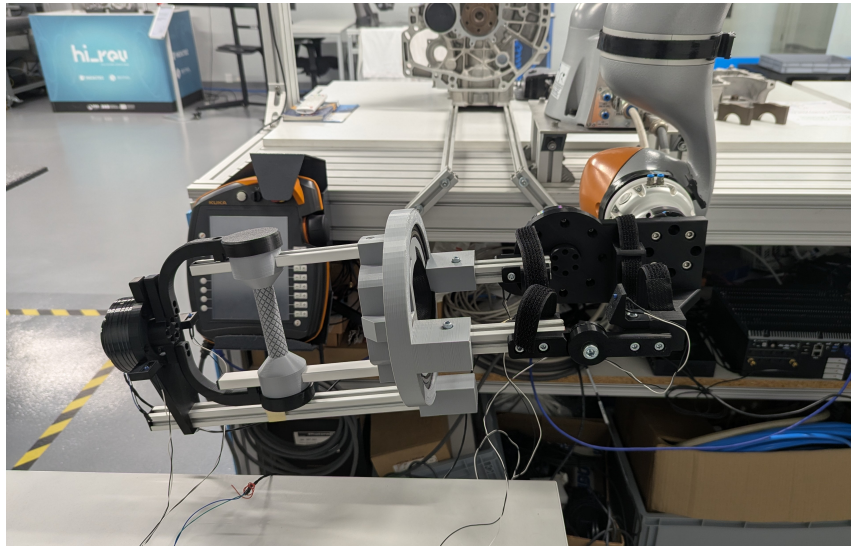


Figure 5.16: Structure Test Setup

In order to be able to create comparable tests. A similar set of parameters was utilised in the actuation of this test. [Table 5.4](#) displays the chosen parameters.

The structure tests were performed utilising the 3 existing work control modes, as such, the ACTIVE Resist and ASSIST-AS-NEEDED modes are subject to the same limitation described in the previous subsection. The ACTIVE Resist mode had to be manually actuated in order to detect any current, and the AAN mode was performed with manual actuation up to a certain point, and then the assist routine would finish the movement. However, due to this being a structure only test, these manual actuations were performed with the actuation being performed on the outside. That is without any arm inserted in the structure and performing the actuation in the final arm position inside the structure. These structural tests yielded the following results([Figure 5.17](#) for ACTIVE Assist, [Figure 5.18](#) for ACTIVE Resist and [Figure 5.19](#) for ASSIST-AS-NEEDED):

These results will be reviewed in [subsection 5.4.2 Structure Tests](#)

Table 5.4: Control parameters for elbow and wrist motors in the structure test

Parameter	Elbow Motor	Wrist Motor
Active Assist (PSA)		
Active Assist repetitions	2	2
Active Assist start ($^{\circ}$)	0.0	90.0
Active Assist end ($^{\circ}$)	100.0	-90.0
Active Assist speed (RPM)	5.0	5.0
Active Assist acceleration (deg/s^2)	1500	600
Active Resist		
Active Resist torque (Nm)	2.0	2.0
Assist-As-Needed (AAN)		
AAN repetitions	1	1
AAN start ($^{\circ}$)	0.0	90.0
AAN end ($^{\circ}$)	100.0	-90.0
AAN speed (RPM)	5.0	5.0

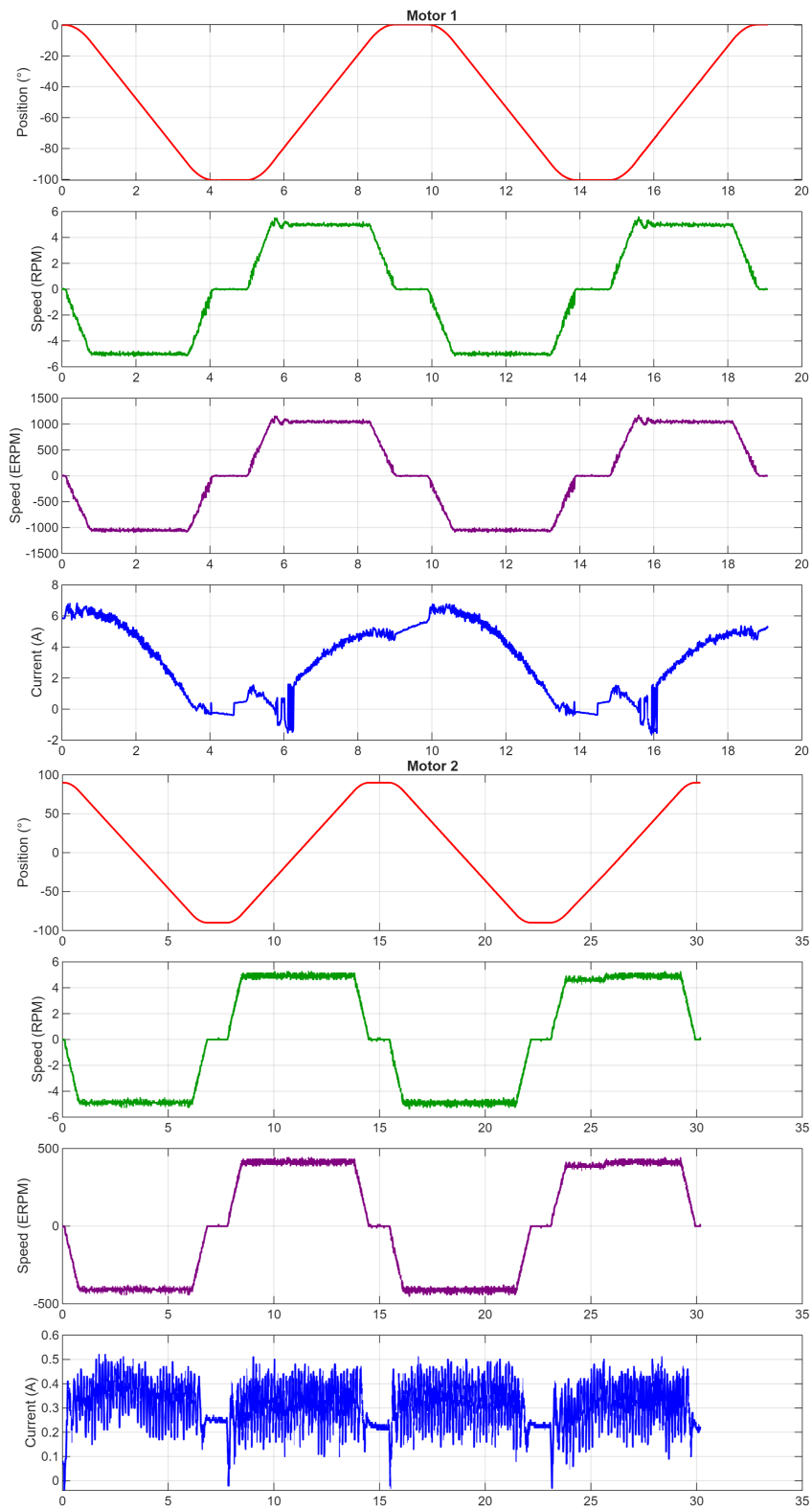


Figure 5.17: Telemetry results for ACTIVE Assist on the AK60 and AK70 motors attached to an empty structure

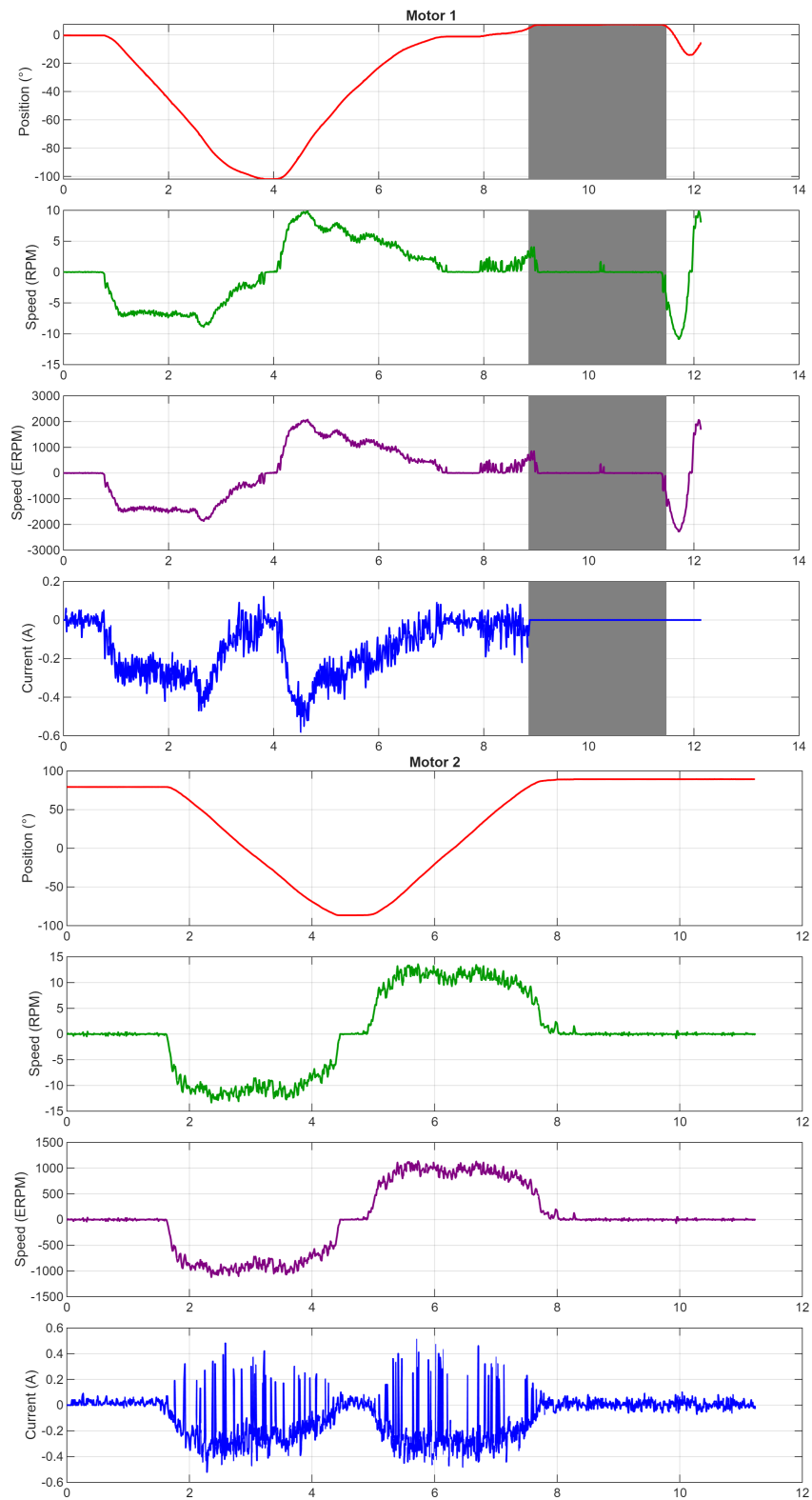


Figure 5.18: Telemetry results for ACTIVE Resist on the AK60 and AK70 motors attached to an empty structure

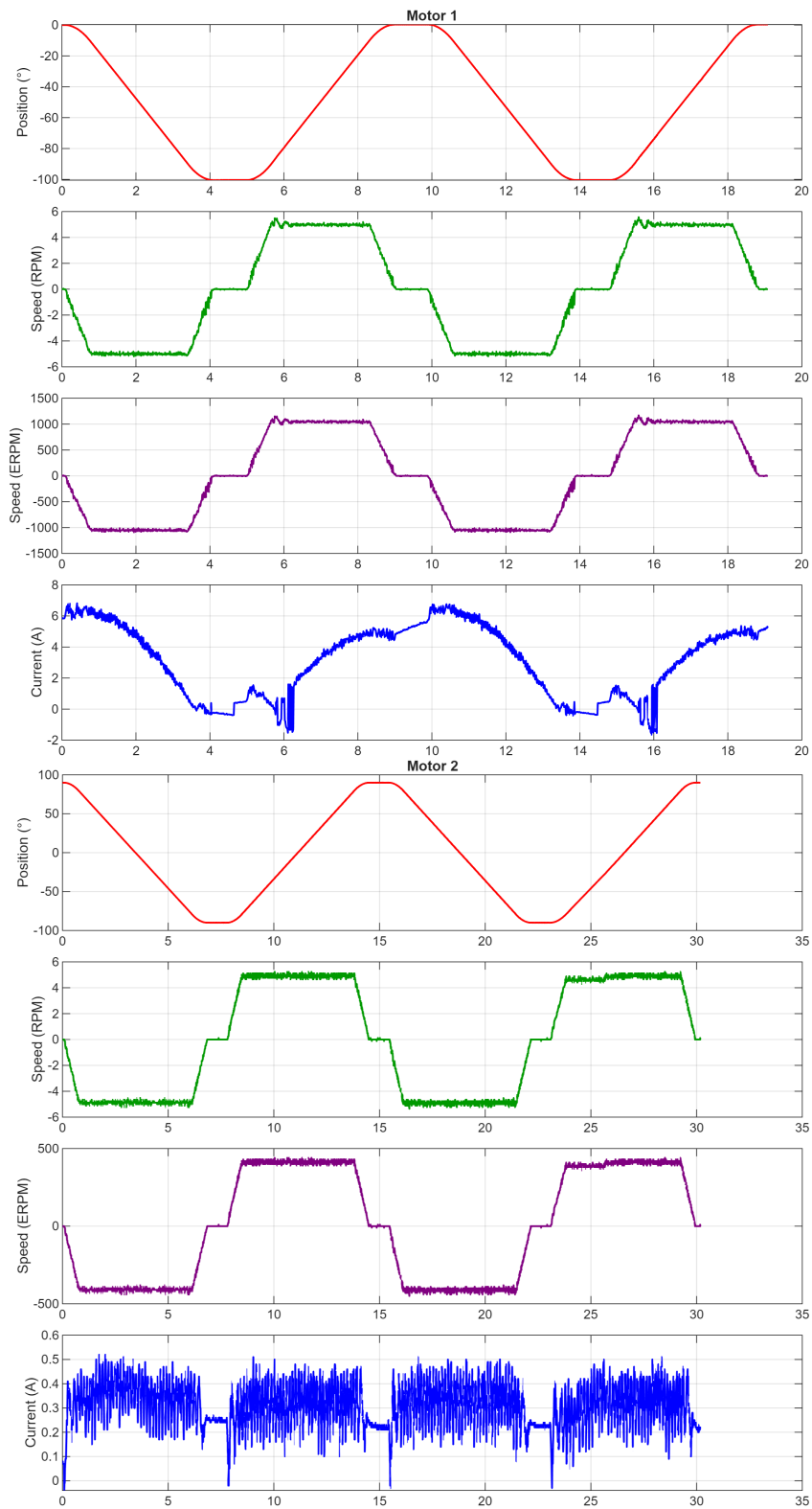


Figure 5.19: Telemetry results for ASSIST-AS-NEEDED on the AK60 and AK70 motors attached to an empty structure

5.3.3 User Tests

The final set of tests performed were the user tests. These tests were the final tests required for system validation. After verification and validation of correct system function with the structure tests, user tests were required in order to validate the system with real upper limb loads and the variable forces performed by the human arm during movement.

This test was performed on a "healthy patient", that is, a person who doesn't face any motor function impairments. The user who performed the test performed it voluntarily and under supervised conditions. These tests were limited to basic functional validation and did not involve clinical assessment or therapeutic intervention. This test setup can be observed in [Figure 5.20](#).



Figure 5.20: User 1 performing System Testing

Due to GDPR laws, the test subject will be referred to as "User 1". And all images shown are of the tests performed on myself. These tests were performed on 1 person only, due to a technical

issue found on one of the motors during the testing phase. For this test all 3 work modes were utilised with the parameters displayed in [Table 5.5](#).

Table 5.5: Control parameters for elbow and wrist motors in the User tests

Parameter	Elbow Motor	Wrist Motor
Active Assist (PSA)		
Active Assist repetitions	1	1
Active Assist start (°)	0.0	90.0
Active Assist end (°)	100.0	−90.0
Active Assist speed (RPM)	5.0	5.0
Active Assist acceleration (deg/s ²)	1500	600
Active Resist		
Active Resist torque (Nm)	2.0	2.0
Assist-As-Needed (AAN)		
AAN repetitions	1	1
AAN start (°)	0.0	90.0
AAN end (°)	100.0	−90.0
AAN speed (RPM)	5.0	5.0

In order to properly log the user's information, a form was created to catalogue the length of the patient's arm, forearm and hand palm, along with their height, sex and age. In this form the used parameters were also set, and at the end a set of questions was asked. These questions had the purpose of assessing the user's comfort and safety during operation. This form can be seen in [Appendix B Test Form](#).

Due to a technical malfunction, only one set of tests were able to be performed. These tests were performed on myself (User 1), and as such, the ending questionnaire had no effect, as it would constitute a conflict of interest to utilise this data as unbiased data. However, from the perspective of this test subject, the test was performed without any safety or comfort issues.

These tests were performed with the parameters described above and in the same order, firstly ACTIVE Assist ([Figure 5.21](#)), then ACTIVE Resist ([Figure 5.22](#)) and finally ASSIST-AS-NEEDED ([Figure 5.23](#)). In order to properly test the AAN mode, the movement was only partially conducted to require the system to assist the final stage of movement. These tests yielded the following results:

The user's feedback can be found in [Table 5.6](#) along with the user's information. These results will be reviewed in [subsection 5.4.3 User Tests](#).

Table 5.6: User information and usability assessment

User	User 1
Test Date	10/02/2026
Age	24
Sex	Male
Height	165 cm
Palm length	10 cm
Forearm length	22 cm
Arm length	26 cm
Comfort score (1–10)	7
Safety (1–10)	8
Ease (1–10)	7
Support (1–10)	7
Pain or Discomfort	No

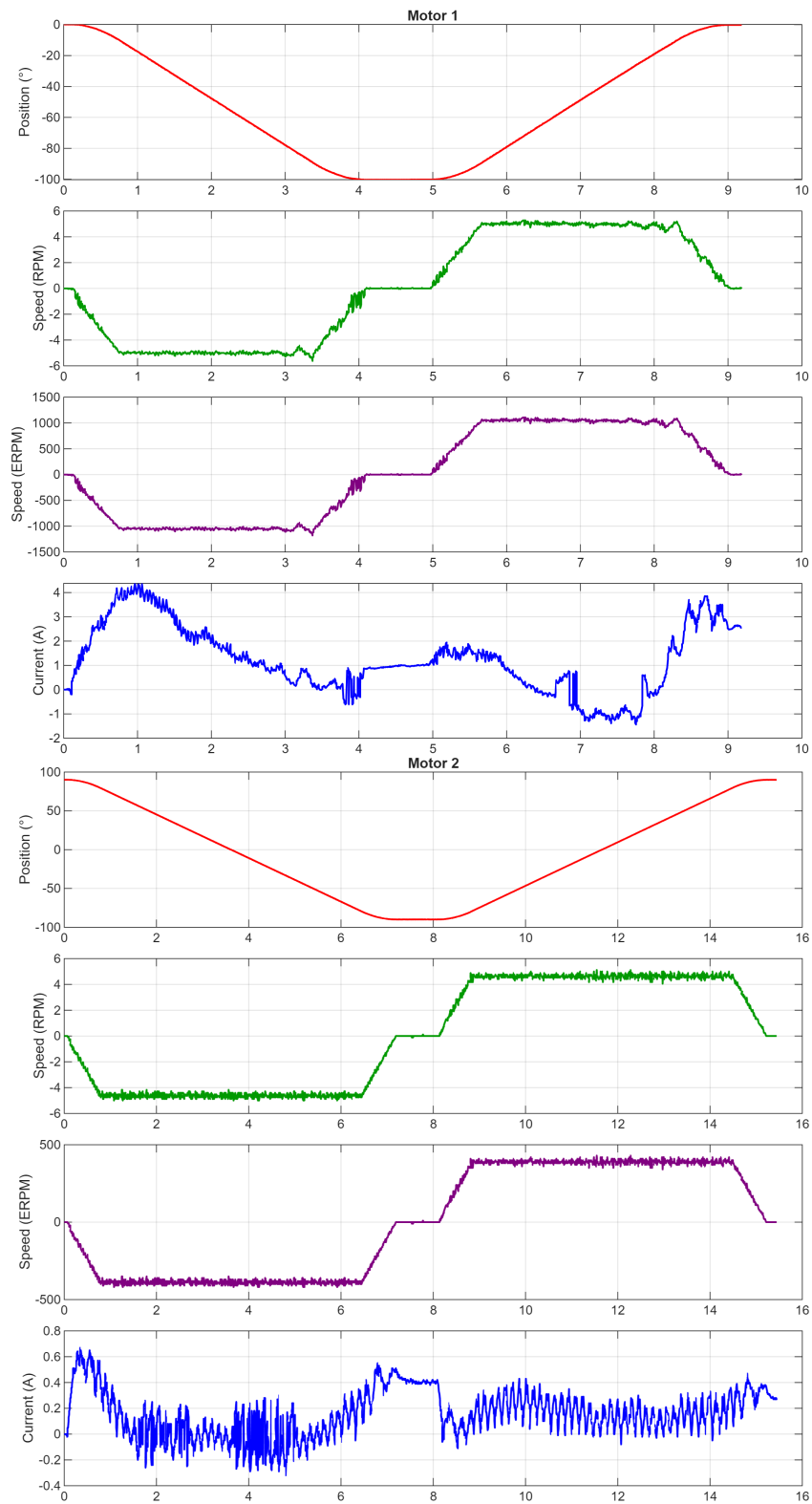


Figure 5.21: Telemetry results for ACTIVE Assist on the AK60 and AK70 motors with User 1

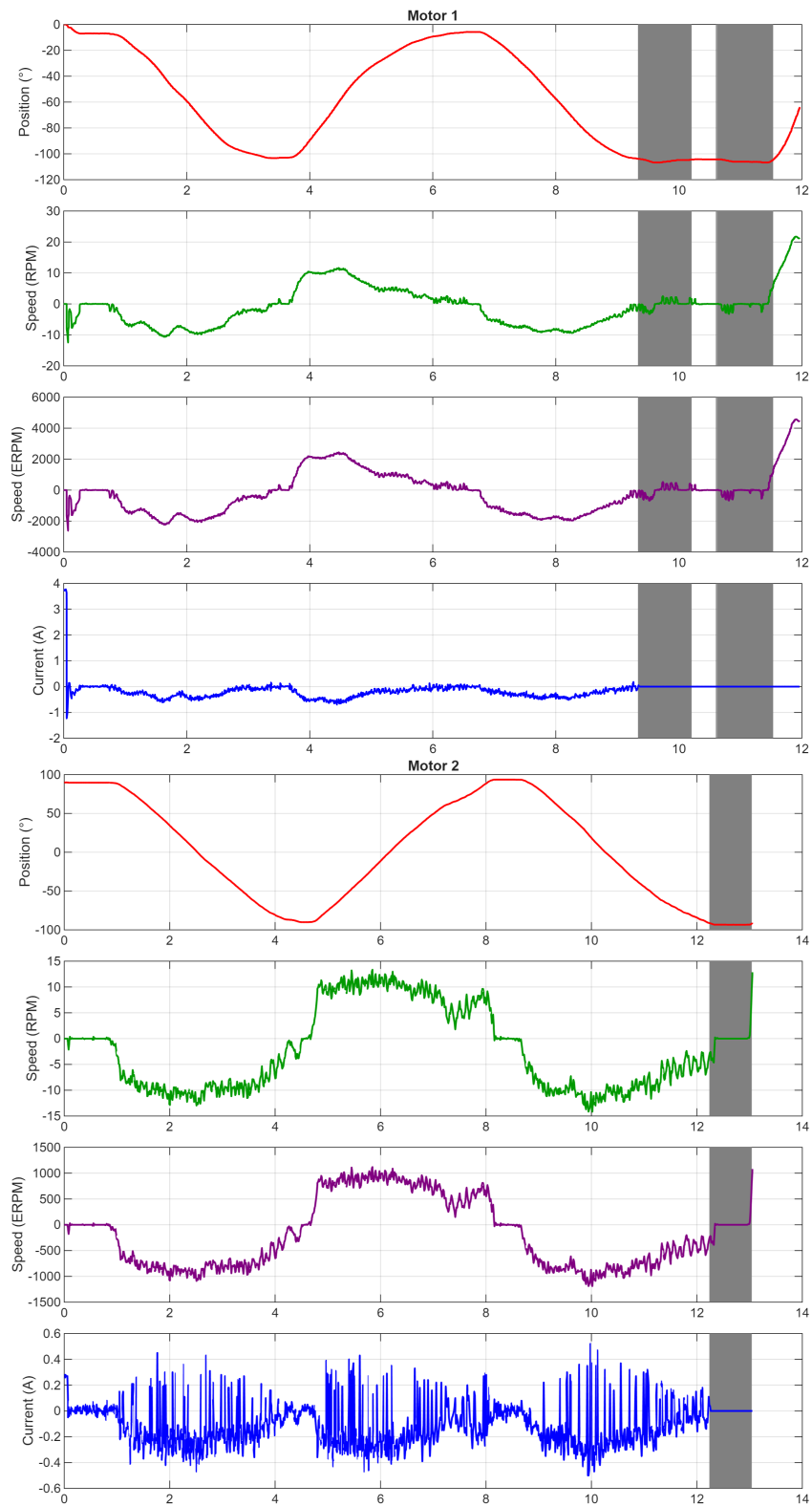


Figure 5.22: Telemetry results for ACTIVE Resist on the AK60 and AK70 motors with User 1

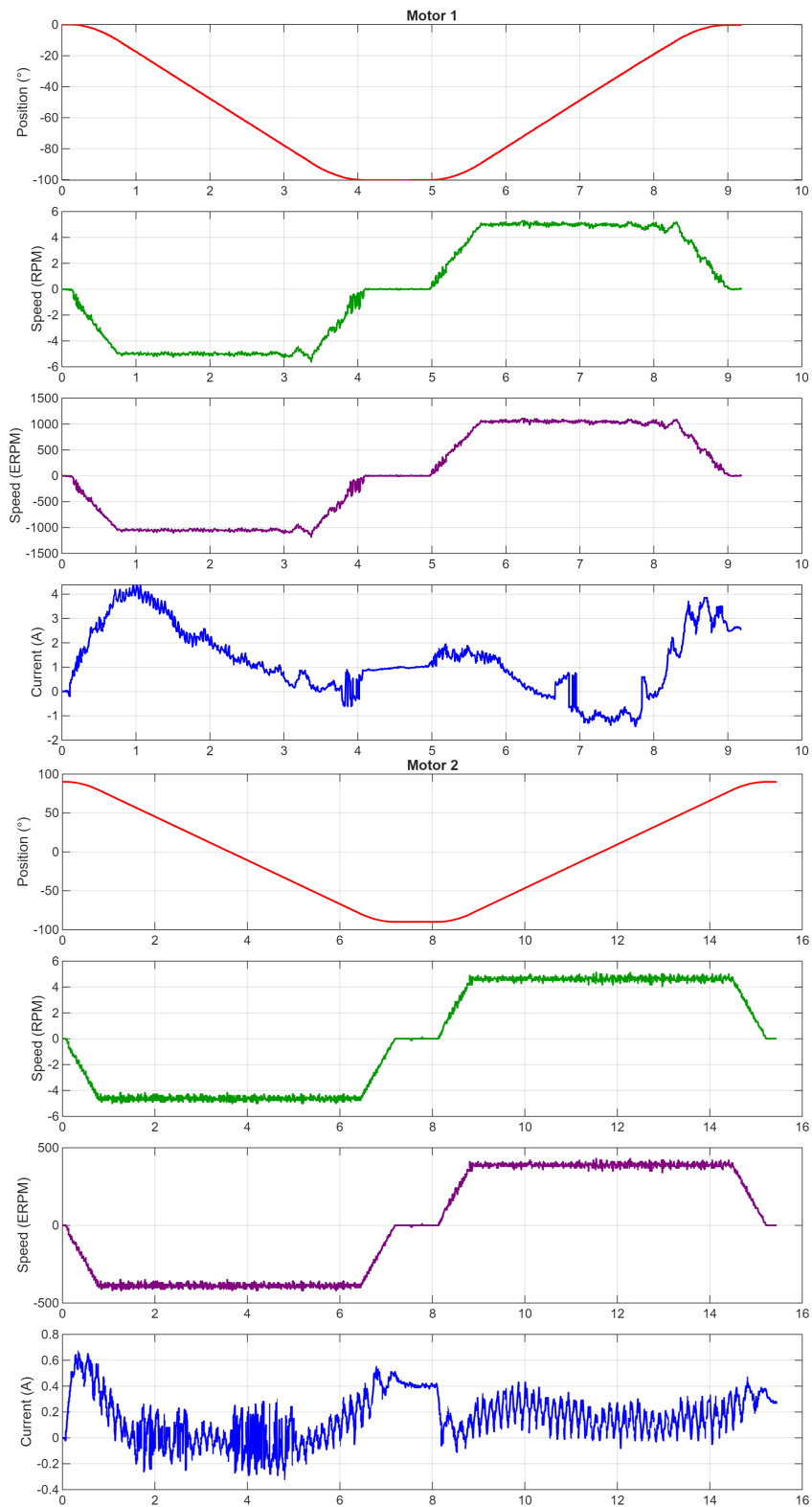


Figure 5.23: Telemetry results for ASSIST-AS-NEEDED on the AK60 and AK70 motors with User 1

5.4 Results Analysis

This section analyses the experimental results obtained from the conducted tests, with the objective of assessing system behaviour, control response, and compliance with the predefined operational requirements. The analysis focuses on the observed actuator performance and joint motion characteristics under the tested conditions based on the data obtained from motor telemetry.

5.4.1 No load Tests

The no load tests were able to confirm the expected behaviour and performance of the actuators. As stated before. Five tests were conducted, three for each work mode: ACTIVE Assist, ACTIVE Resist and ASSIST-AS-NEEDED. And 2 other limitation testing tests where the firmware and driver speed limits were tested. In the following paragraph these results will be analysed.

The first test to be analysed will be the ACTIVE Assist (Figure 5.24). This mode implements a fully actuated movement, where the motor should rotate at a speed of 5 RPM and perform a movement from 0° to -130° in Motor 1 and 90° to -90° in Motor 2. In the presented graph it's possible to observe in the red squares that the motor achieves the required speed approximately, registering a maximum speed of 5.24 RPM for Motor 1 and Motor 2. However, these maximum speeds are only achieved at specific instances, with the trace line showing a clear horizontal line around 5 RPM for both motors. In yellow it is also possible to observe the acceleration curve, with this being mostly linear and respecting the expected acceleration for the given parameters. In the current graph it's possible to observe a current rise in the acceleration phase, shown in orange, and a more constant current value once the speed is constant, as observed in the green squares. The current rises when accelerating to the required speed and falls when decelerating to a stop. It's also observable that in this mode of operation the motor always induces a small amount of current to hold the expected stopped position. This behaviour is nominal and expected. Finally it's possible to observe that in both motors the position movement is performed according to expectation, respecting the set values to a 0.1° tolerance.

Secondly, the motors' behaviour in the ACTIVE Resist mode was analysed in Figure 5.25. In this test it should be expected that when the motor is moved in any direction, a current is applied to increase resistance to that movement. As observed in the figure, when any speed is added to the motor's rotor, the current increases directly proportionally to the rotor's speed. This can be observed in the red and green rectangles. The movements performed were as similar as possible to the movements expected to be performed in the final end effector. Furthermore, end stop switch activation and actuation can also be observed in the Motor 2 graphics column. The grey vertical lines indicate end switch activation. Due to the end switch being activated before its end of course stop, a small movement in position and speed can still be observed after activation. However, it's also observable that as soon as the end switch is triggered, the motor shuts down any current input, becoming idle and free to move, this can be observed in the yellow box in the current graphic for Motor 2.

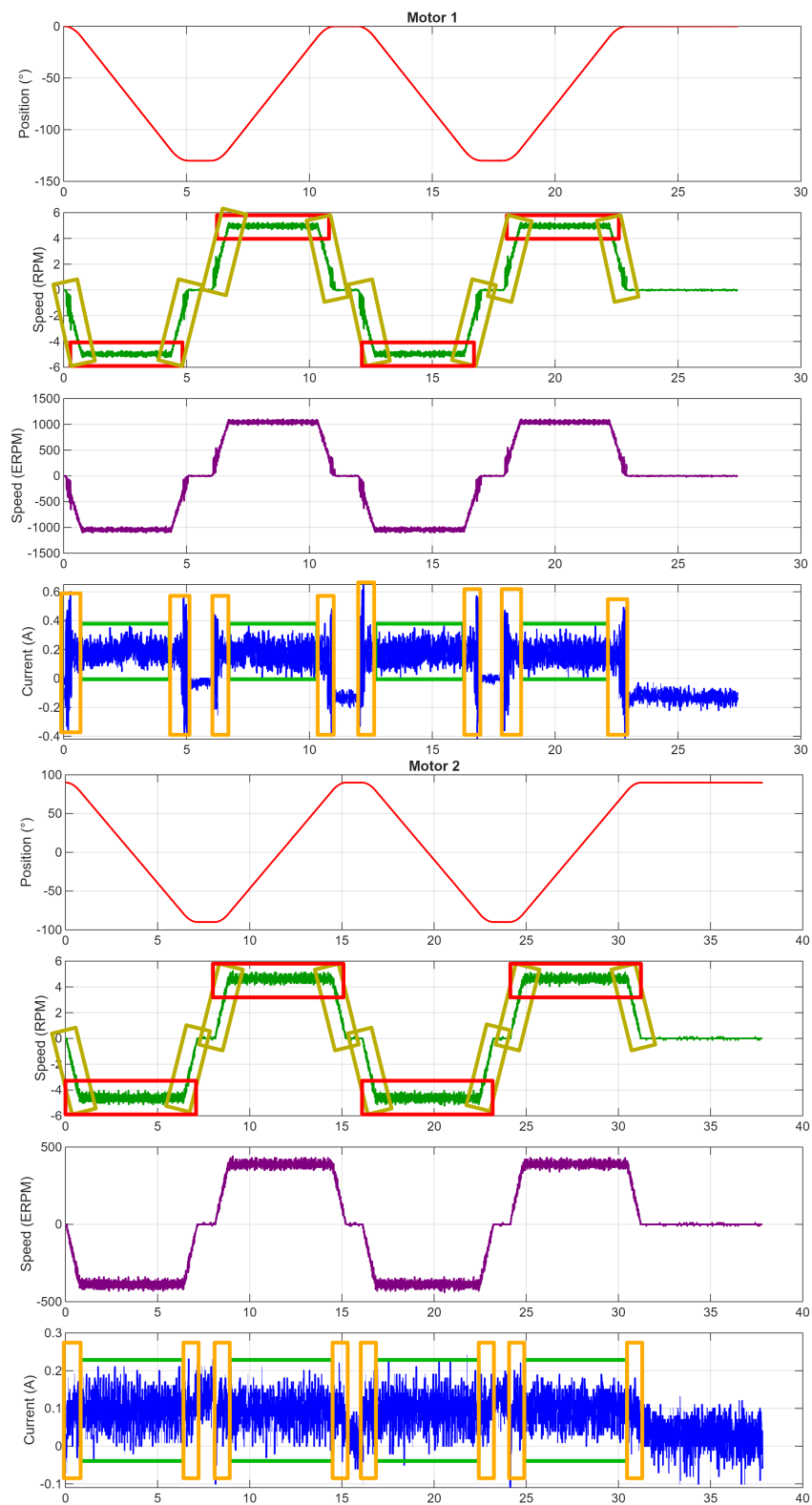


Figure 5.24: Telemetry results for ACTIVE Assist on the AK60 and AK70 motors without any load, analysed

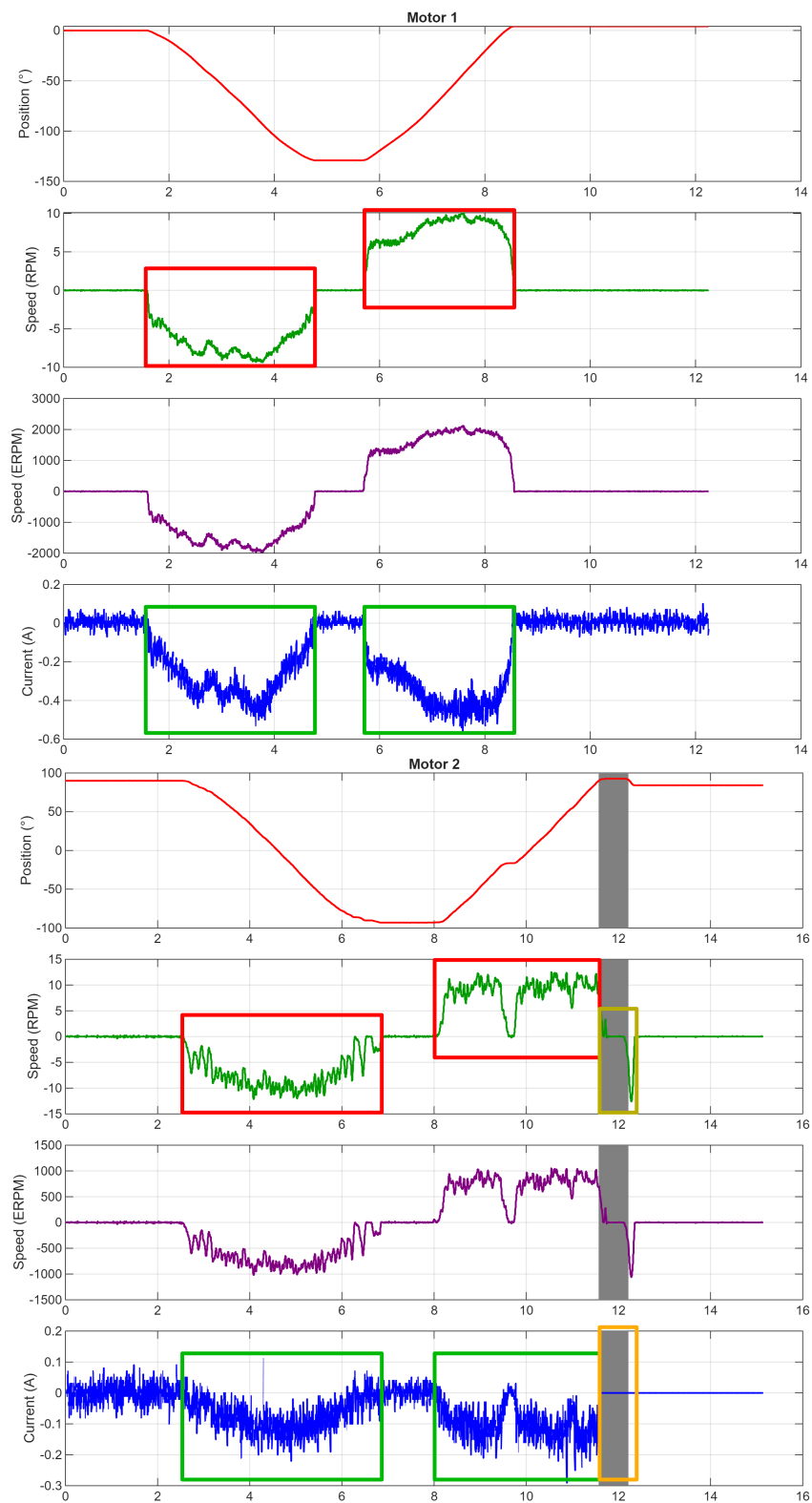


Figure 5.25: Telemetry results for ACTIVE Resist on the AK60 and AK70 motors without any load, analysed

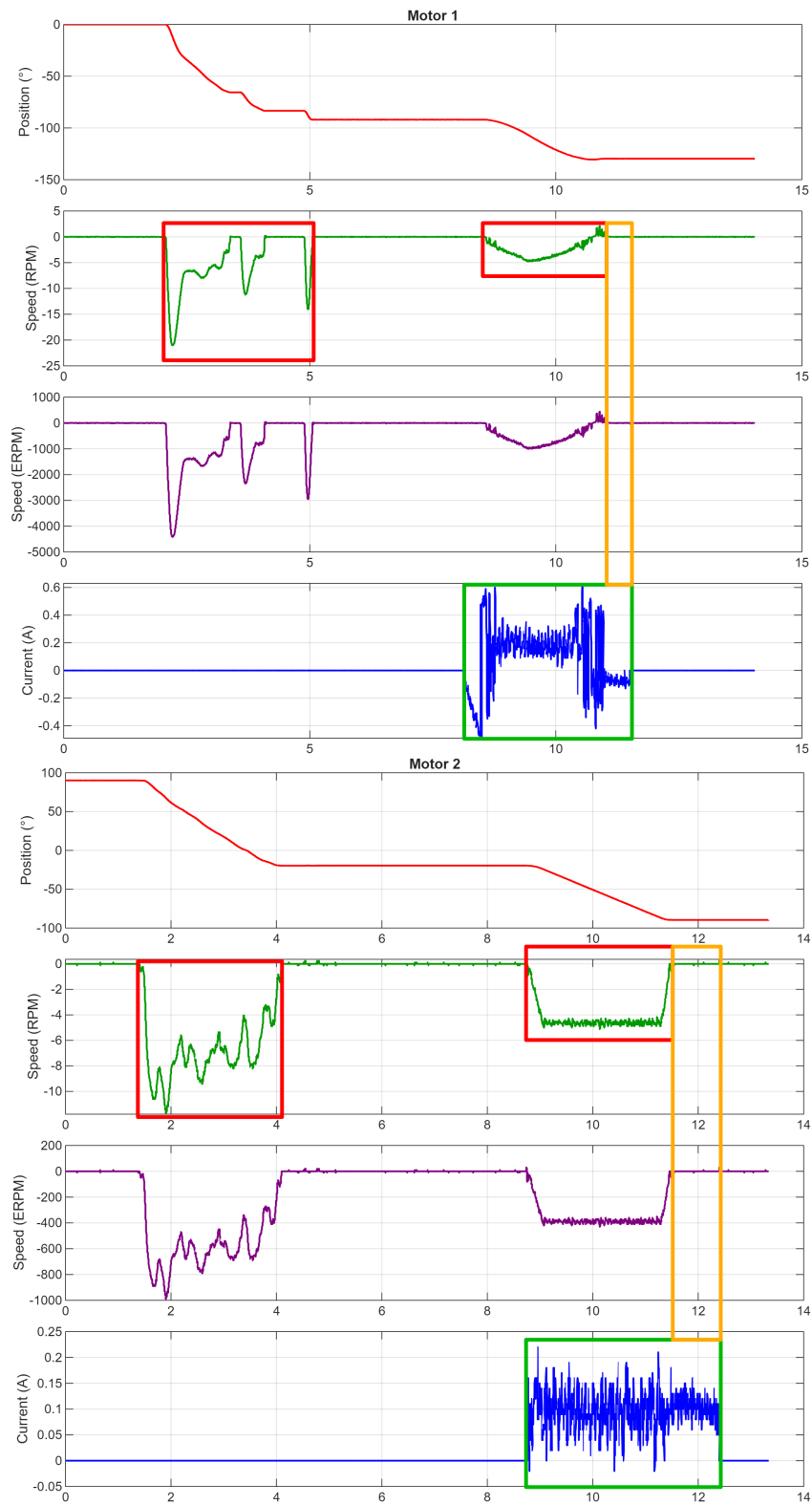


Figure 5.26: Telemetry results for ASSIST-AS-NEEDED on the AK60 and AK70 motors without any load, analysed

The last work mode test conducted in a no load configuration was the ASSIST-AS-NEEDED mode test. In this test, as described before, a partial movement was conducted, and then the actuator finished the movement, assisting it. In both motors the behaviour is relatively similar as seen in [Figure 5.26](#). In the first red rectangle for each motor, it's observable the free movement of the motor's rotor followed by an interruption in movement between the 2 red rectangles. This is the moment where the human actuation was stopped and the system was waiting for the assist function to begin. For motor 1 the movement stops at 5s into the AAN mode start time and the assisting functionality starts actuating at 8.1 s. This is performed when the motor is in the 92° position and the movement is stopped for 3.1 s. At the defined speed of 5 RPM the motor is expected to be at the 92° position in roughly 3.1 s. For Motor 2 the manual movement stops 4.1 s into the AAN routine, and the automated assist starts at 8.6 s. The motor's rotor also stops in the -19.6° position, that is, 109.6° into the movement, for a duration of 4.5 s. At the given speed of 5 RPM, the expected time to reach the 109.6° position is 3.6 s. This validates the operational functionality of the implemented ASSIST-AS-NEEDED routine, given the 5 s delay for the assist actuation. Furthermore, it can be observed that Motor 1 is never able to achieve a constant speed of 5 RPM during the assist phase, due to the need to decelerate at a given rate to stop in the final position. Motor 2, however, performs the assist phase nominally and in a similar manner to the ACTIVE Assist Routine, as expected. It can also be observed that once the motor reaches the final position, it holds the position for a brief time before shutting off, this is the gap shown in the orange box on the graphics.

After testing all 3 work modes, the no load testing programme performed the limitation testing.

In this test the motor driver was limited to 1900 ERPM for the AK70 motor, then an ACTIVE Assist command was issued with a speed of 10 RPM (2100 ERPM for the AK70 motor), as shown in the red rectangle in [Figure 5.27](#), the motor achieves a maximum speed of 1920 ERPM at 1.41 s. This represents an overshoot of 1.05%, that isn't ideal but an acceptable value, as it would still present a safe value for operation.

Finally, the last test conducted without any load was the microcontroller firmware speed limitation. Similarly to the previous test this test was executed with an ACTIVE Assist command. The speed value introduced in the parameters was 50 RPM with the microcontroller limiting the speed to 30 RPM. As it can be observed in [Figure 5.28](#), the motor's speed seems to be constant below the 30 RPM line, with a maximum registered value of 28.93 RPM. This limitation mode, contrary to the one in the previous test, appears to be underachieving compared to the given command. However, in safety critical fields such as rehabilitation, this is a more desirable outcome compared to an overshoot.

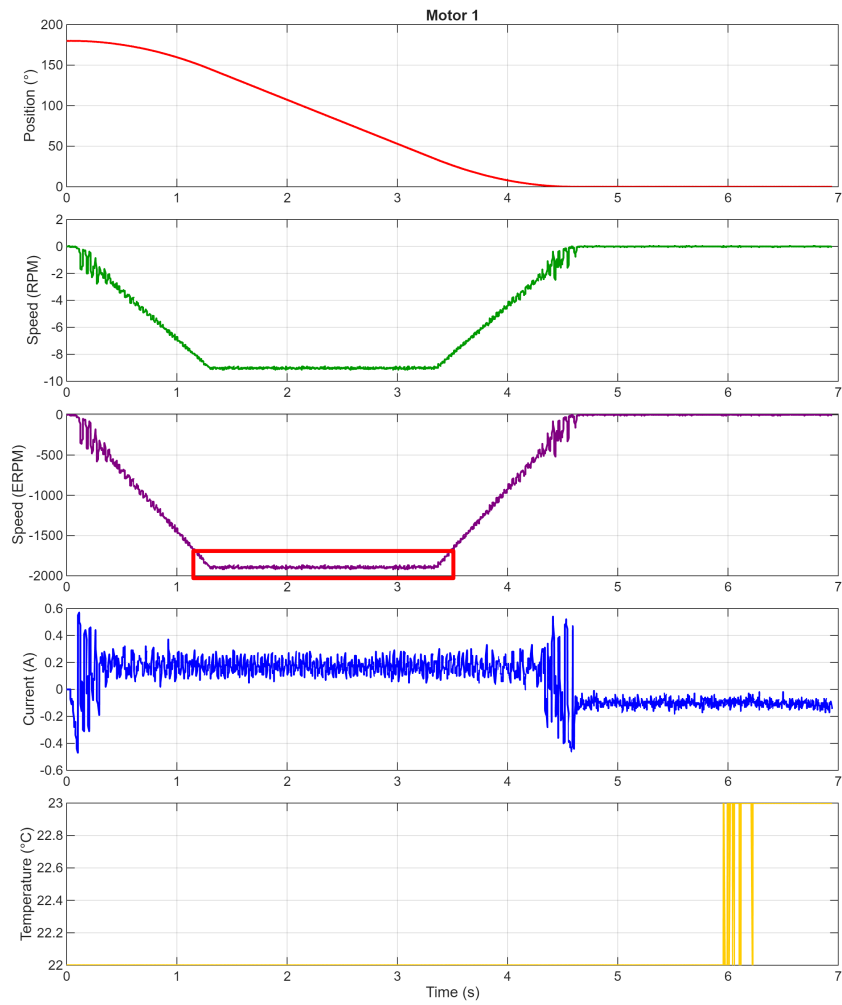


Figure 5.27: Telemetry results for ERPM driver set speed limit in the AK70 motor without any load, analysis

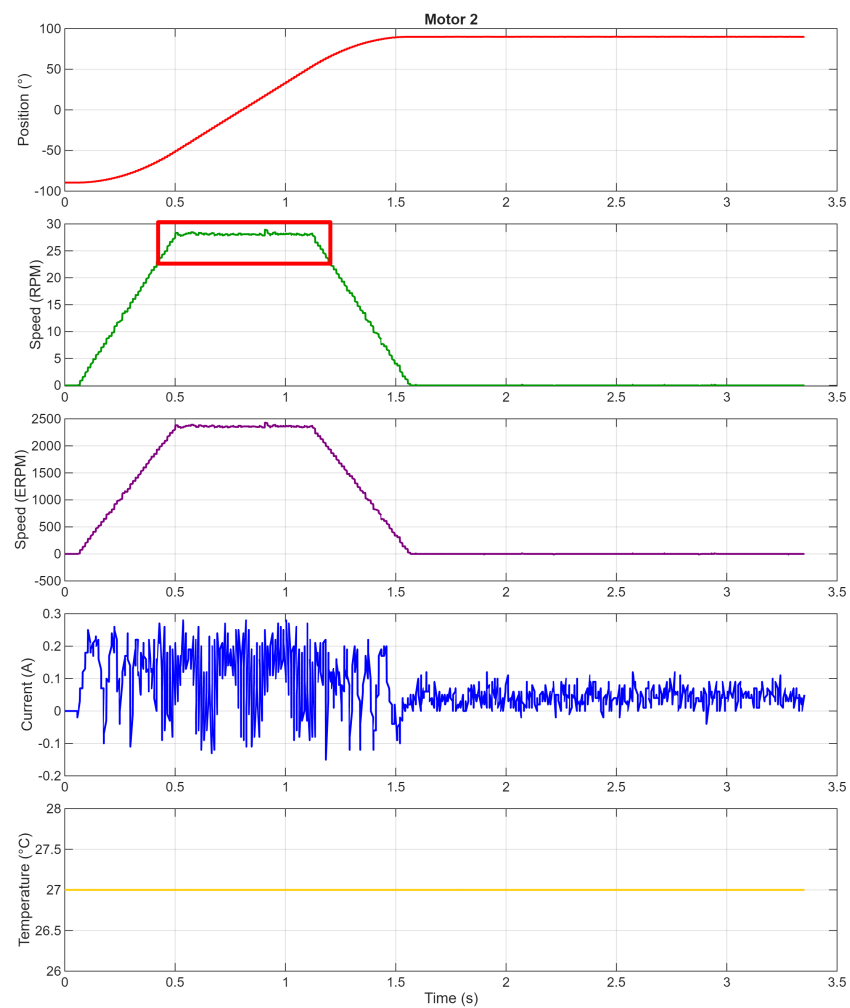


Figure 5.28: Telemetry results for RPM controller firmware set speed limit in the AK60 motor without any load, analysed

5.4.2 Structure Tests

Once the basic actuator validation was performed, the testing of the fully assembled system was performed. This allowed the testing of the actuators in their intended positions and with the minimum possible mechanical loads. The previous set of tests verified the capabilities and capacity of the actuators. This set of tests would allow the assessment of the real performance and behaviour of the actuators. In order to ensure the capability of the actuators to perform the given tasks, firstly, the entire system was tested unloaded, with only the structure of the end effector being the mechanical load.

As stated in [subchapter 5.3.2 Structure Tests](#), the structure was fully assembled, and the test consisted of executing all three work modes on an empty structure.

The first test performed was the ACTIVE Assist test. Similarly to the ACTIVE Assist test conducted in the previous subsection, this test utilises similar parameters. The only meaningful changes were the amplitude of movement for Motor 1 and the utilisation of 2 Nm of brake torque in the resist mode. The amplitude of movement was determined to be a maximum of 100°. Due to the more complex nature of this test, each motor will be evaluated individually.

The results obtained for Motor 1 in [Figure 5.29](#) show linear position movement between extension and flexion of the forearm, with the speed data behaving similarly to the no load tests, as seen in the position graphic and the red and yellow rectangles in the speed graphic. This displays the motor's capability, as expected, to perform consistently against the gravity load placed on it by the end effector's weight. The current data log is the most significantly changed data output compared to the no load tests. The previous 2 data fields (speed and position) allowed the assessment of correct motor function. The current field allows for the assessment of the motor's load bearing capacity. The CubeMars AK70-10 motor used in this joint has a rated current of 7.2 A and a peak current of 23.2 A [108]. The maximum registered current value in this test was 6.82 A, a value lower than the rated current. It can also be observed in the green rectangles in the current plot, this peak is only achieved at the start/end of the movement when the end effector is close to normal to the force of gravity, with the load diminishing as the forearm is moved to perpendicular to the force of gravity. It can be concluded that the motor utilized in the elbow joint is completely capable of performing the ACTIVE Assist task in these circumstances, with a significant margin to the peak allowed current of the actuator.

The results obtained for motor 2 are largely similar to the ones obtained in the no load test. As stated in [subsection 4.4.1 Actuation Requirements and Motor selection](#) this motor has a negligible gravity load factor. However, this test is able to display the repeatability and consistency of its operation. Similarly to the situation analysed in the no load test, the speed, position and acceleration expectations are fulfilled, and in the current graph we can observe an increased current load due to the added resistance of the bearing that this motor is now required to actuate on. During the acceleration phase an increase in current can be observed, and during the deceleration phase a contrary decrease in current is observed, as shown in the orange rectangles, with the constant speed operation showing a higher current draw compared to idle, horizontal operation, as shown

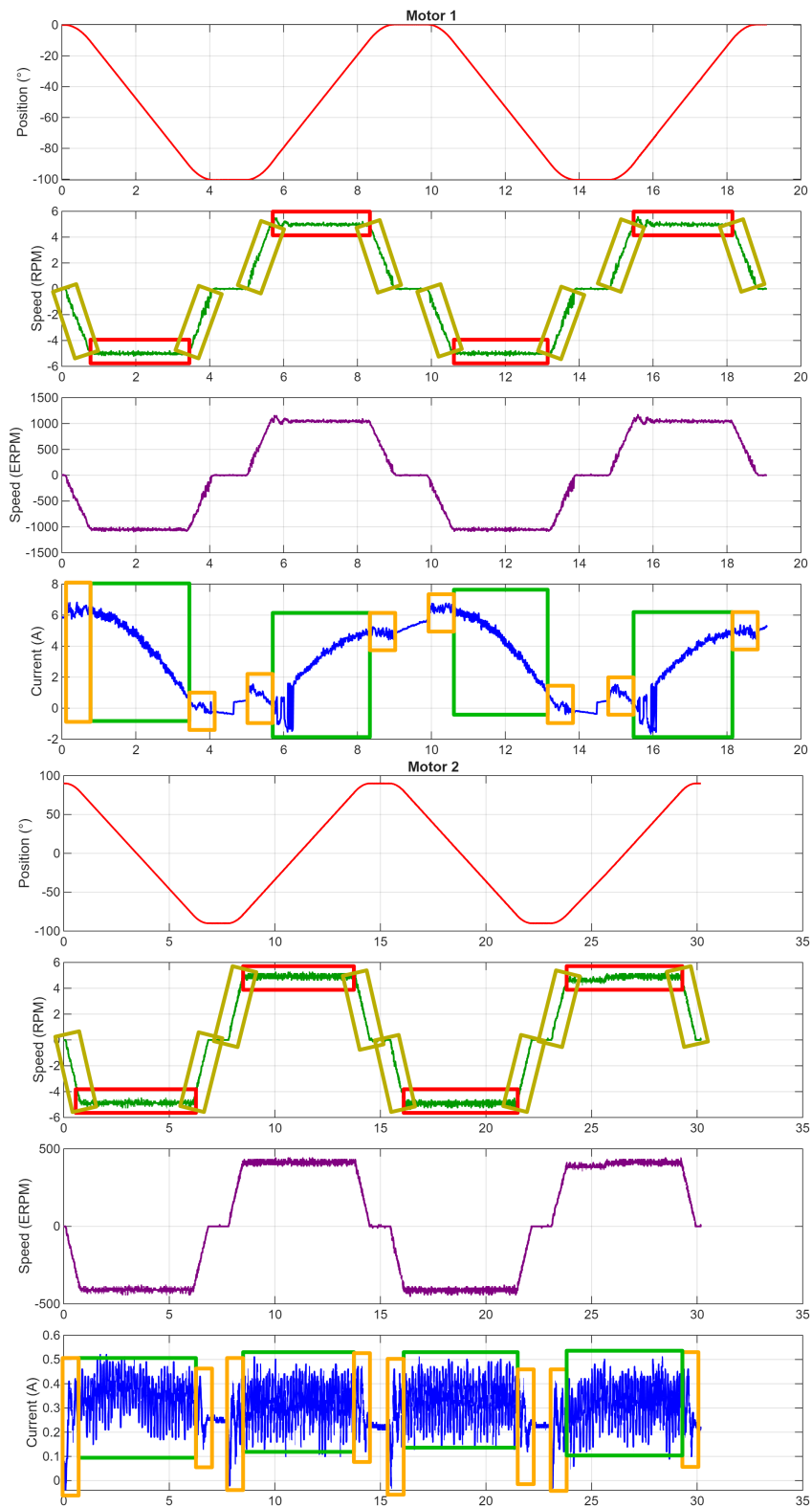


Figure 5.29: Telemetry results for ACTIVE Assist on the AK60 and AK70 motors attached to an empty structure, analysed

in the green rectangles.

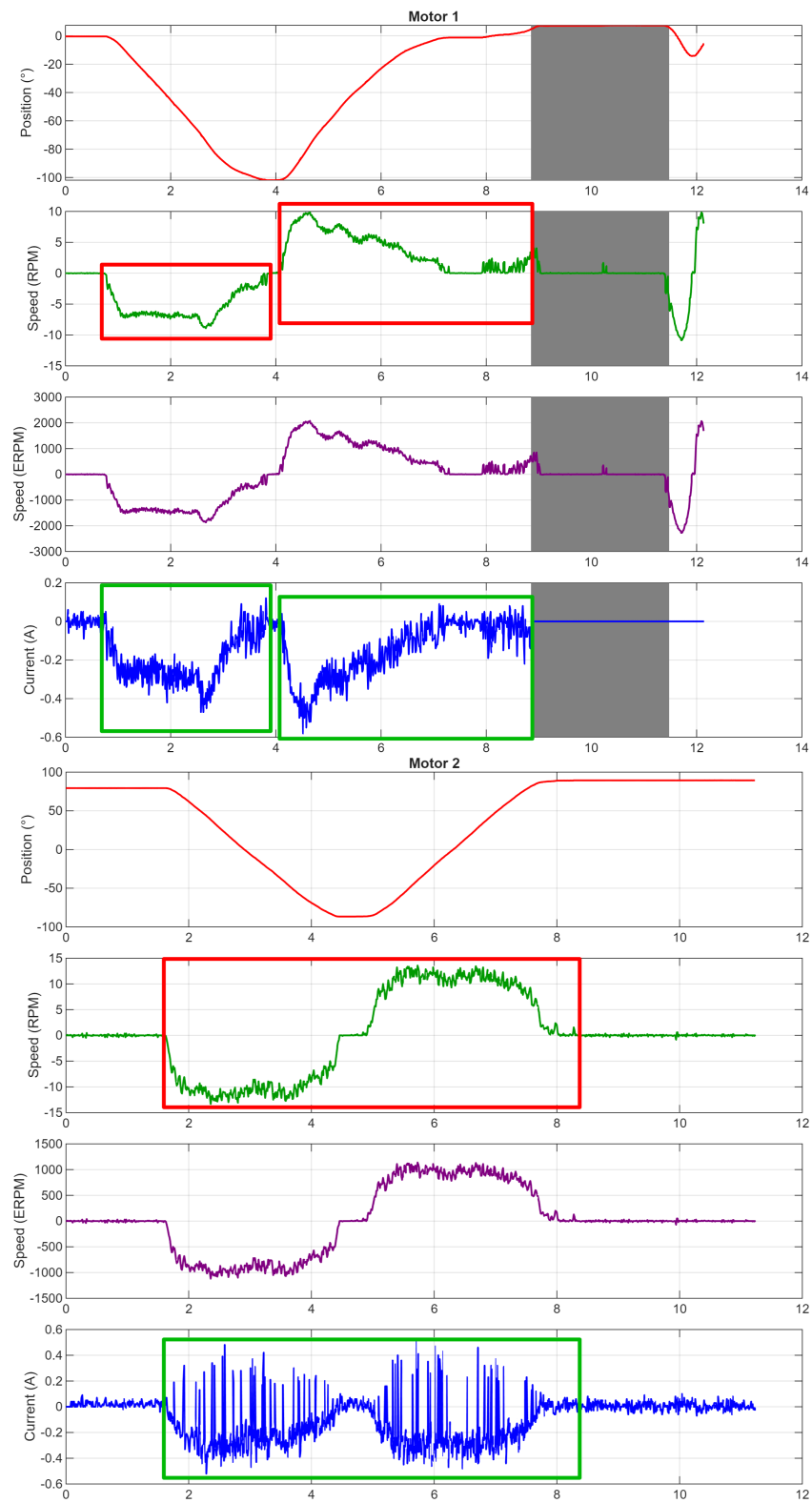


Figure 5.30: Telemetry results for ACTIVE Resist on the AK60 and AK70 motors attached to an empty structure, analysed

In the second work mode, ACTIVE Resist (Figure 5.30), the results were similar to the ones obtained in the no load test, in this mode, the motor simply increases the current proportionally to the rotor speed and up to a certain level that requires the set amount of torque to move the rotor of the motor. Due to the low load on motor 2, the current required for the 2 Nm with the handle and bearing load is smaller and more affected by the current variations. Both motors behave in the expected manner.

Finally, the last mode tested in the structure only test was the ASSIST-AS-NEEDED mode test (Figure 5.31). In this test, as described before, a partial movement was conducted, and then the actuator finished the movement, assisting it. Similarly to the test performed in subsection 5.4.1 No load test in this test, a movement will be performed manually, completing partial movement, with the assisting actuation finishing the remaining movement. Different from the test conducted without any load, in this test the ASSIST-AS-NEEDED mode would perform a full extension and flexion movement for the elbow joint and a full pronation and supination movement of the wrist in sequence. Similarly to the ACTIVE Assist structure test, both motors will be analysed separately.

In motor 1, the elbow motor, in order to compensate for gravity, the motor begins with a command to hold the 0° position before the AAN command is given. Once this command is given, any induced current in the motor is removed, and the flexion movement can begin. A movement is performed up to the 47.4° position. Once this position is deemed to be 5 s behind the expected position, the motor starts actuating to assist in finishing the movement, at 6.6 s since the start of the movement, shown in the green rectangle, up to the specified 100° position. Each pair of the red rectangles displays, firstly, the manual movement and secondly the motor actuated movement. Once the final position is reached, the extension movement is started. Similarly, this movement was done partially manually, and the assisting feature would complete the movement. For the extension movement, the structure was moved to the 37.5° position, and the actuated assist movement began actuating at 7.1 s from the start of the flexion movement correctly. The actuation times, once again, match the expected times. The current in this exercise shows a smaller current draw in the first movement, compared to the ACTIVE Assist mode, this is due to the less demanding actuation movement from 47.4° to 100°. In the second movement, after an initial current spike, the current can be seen increasing up to the horizontal position. Once the movement is finished, the motor holds the 0° position as shown in the final part of the first green rectangle.

For motor 2, the wrist motor, the behaviour is very similar to the behaviour observed in the no load test. The first movement is performed from a 90° to a 1.4° position and the assist mode finishes the movement up to the -90° position. The assist started 8 s after the AAN routine was started. In the opposite movement the motor was not moved and stayed at the -90° position and actuation started at 5 s after the AAN routine for the second movement started and finished the movement in the 90° position. The time for actuation in these tests is also validated. For these assist movements, the current displays a similar behaviour to the one already analysed in the ACTIVE Assist mode, as seen in the green rectangles, with the motor holding position for a short time after movement completion, this can be seen in the yellow rectangle.

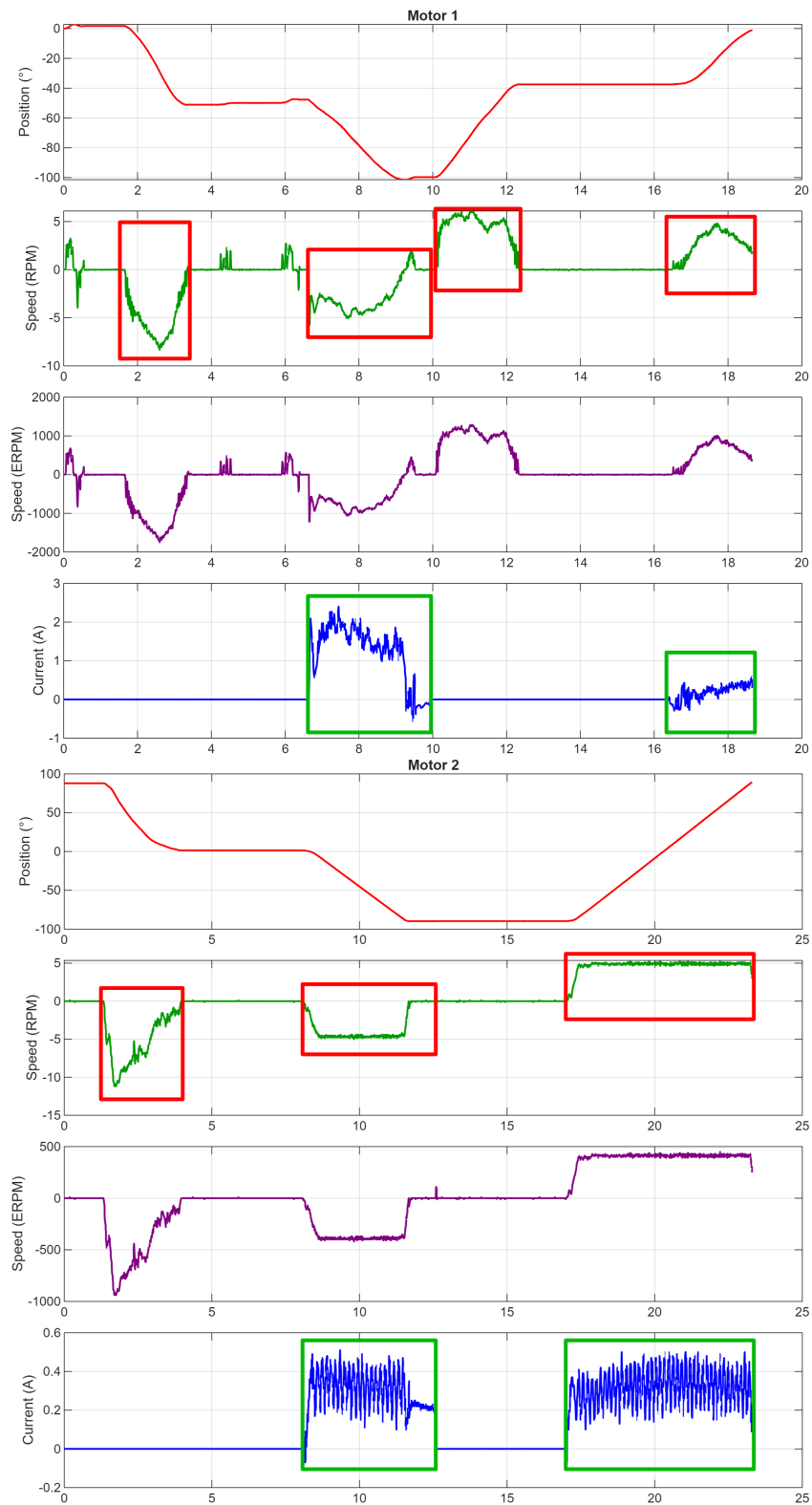


Figure 5.31: Telemetry results for ASSIST-AS-NEEDED on the AK60 and AK70 motors attached to an empty structure, analysed

5.4.3 User tests

After motor and structure validation, the final set of tests required to ensure correct system performance were the user tests. These tests would allow the testing of the system for higher load conditions, variable load and the system's response to these factors. Besides this, it allowed the evaluation of the mechanical efficacy, safety and comfort for the end user.

The first test performed was with the ACTIVE Assist mode, in this mode the user isn't required to perform any actuation on the structure. However, unlike the structure test in this test, the system was only programmed to do 1 repetition, that is, one flexion/extension and pronation/supination.

The results obtained for the ACTIVE Assist mode, shown in [Figure 5.32](#), display a clear capacity of the system to keep linear and smooth actuation under increased load, as seen in the position profile for both the wrist joint and the elbow joint. No noticeable overshoot or drift is observed either. In the speed profile it's possible to observe linear acceleration (yellow rectangles) with some punctual fluctuations until the defined speed, and both motors are shown to be capable of holding this speed during the entire movement (red rectangles). However, it can also be observed that motor 1, the elbow motor, displays more fluctuations due to the higher load compared to the wrist motor. Nevertheless, the motor still performs the required movement in an expected manner.

In the current field clear differences can be observed compared to the unloaded structure test. Particularly for motor 1, the elbow motor. In this motor, during the acceleration phase of the flexion movement, the first movement, a rising current, is observed. This happens during the highest load moment, when the forearm is moving from the horizontal to the vertical position. This corresponds to the additional torque required to overcome system inertia, gravitational loading, and static friction during the motion, as well as to actively brake and dissipate kinetic energy near the end of the trajectory (first and last orange boxes). During constant velocity intervals, acceleration is negligible, and the required torque is mainly associated with gravity compensation, resulting in lower current levels (green boxes). The acceleration at the end of the flexion movement and the start of the extension movement, when the arm reaches and leaves the 100° position, has very low gravity load, which greatly reduces the torque requirements and induces a lower current as a consequence. During the extension movement (second movement from 100° to 0°), a different current shape is observed when compared to the structure only test. This can be attributed to the user's arm adding resistance to the movement, requiring the system at a certain point to apply torque in the direction of movement, while for the most part in the structure test the system purely performs gravity compensation. The shape of the flexion movement is similar to that of the structure tests but has lower current values due to the user's arm providing some force during the movement, diminishing the load on the motor.

For motor 2 the current behaves in a similar manner to the structure only test. During the first movement (supination), it can be observed that a very low current is applied to the motor due to the user's input aiding the movement and unloading the motor. Sometimes even being required to add a resistive force to the wrist. This can be observed particularly in the middle of the movement. The second movement (pronation) displays a more similar shape and value to the structure test, with

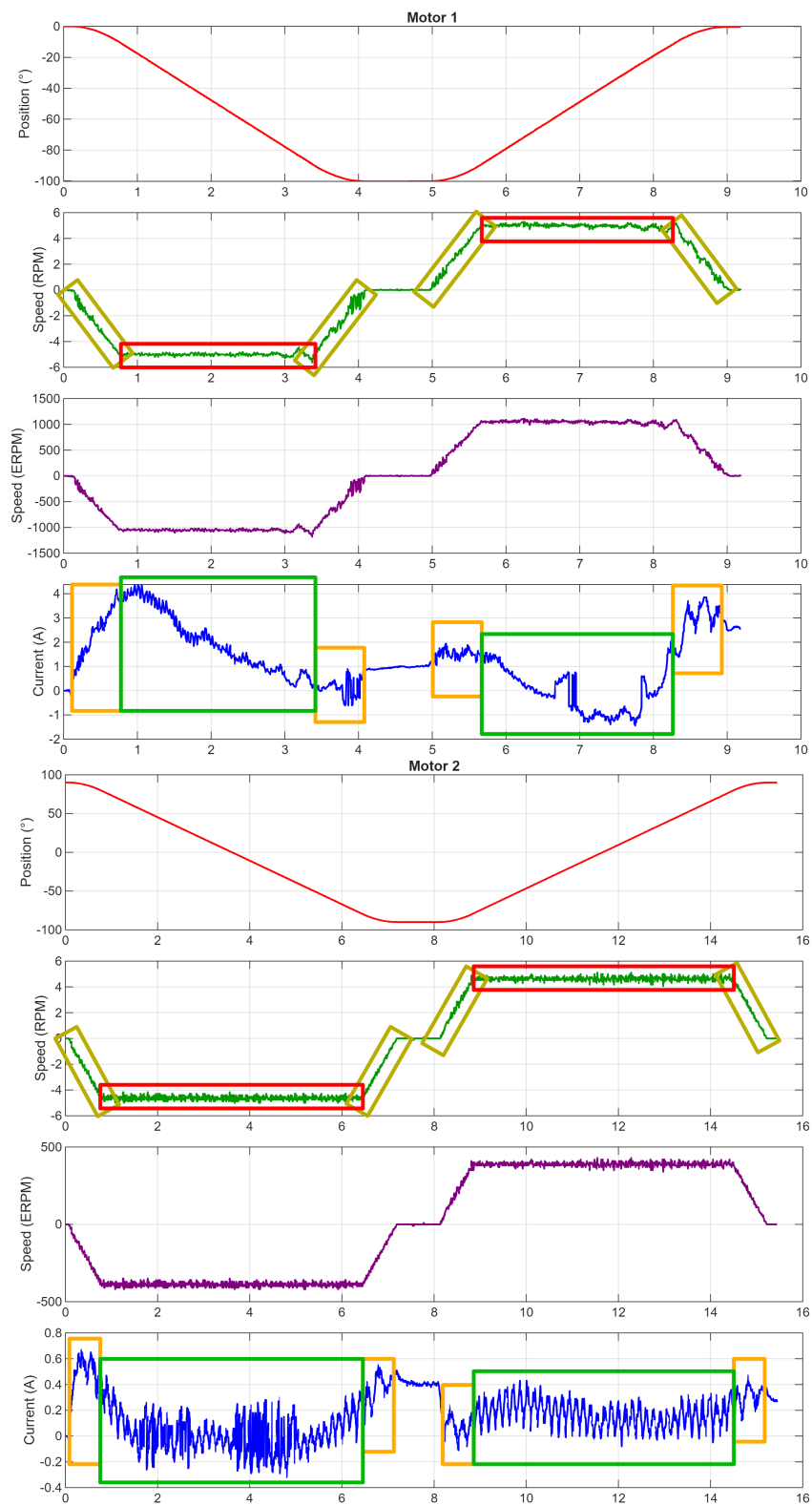


Figure 5.32: Telemetry results for ACTIVE Assist on the AK60 and AK70 motors with User 1 analysed

the motor pushing the wrist with a consistent force during the movement. This can be observed in the green boxes. During the acceleration phases of the movements, current spikes can be observed due to the increased load, resistance and momentum added by the human wrist, as observed in the orange boxes.

Overall, the results demonstrate that in ACTIVE Assist mode the developed system is capable of providing smooth and stable assistance while preserving accurate position and velocity control. The increased current demand observed in loaded conditions reflects appropriate torque generation to compensate for limb weight and passive resistance, without compromising stability or safety. These findings validate the suitability of the proposed actuation and control architecture.

For the ACTIVE Resist, the results obtained in [Figure 5.33](#) are once again similar to the ones previously obtained. As in the structure only tests, the motor current increases proportionally with rotor speed in order to generate the required resistive torque. Compared to the unloaded case, higher current levels are observed, particularly in motor 1, reflecting the combined effects of limb inertia and increased gravitational loading from the extra load of the arm. Furthermore, small fluctuations in both speed and current are present throughout the movement, resulting from voluntary user interaction. End switch activation can also be observed in both motors in the grey bars, displaying the same behaviour previously analysed. In this instance it is also possible to observe the switch is activated twice. This is due to the user performing a small movement that depresses the switch but soon after lowering the structure and pressing the button again. This variation is only of 2.4° . However it is sufficient to be displayed as 2 different instances. This has no effect on the system due to the first end switch activation disabling the motor.

Despite these disturbances, the controller keeps a stable resistive behaviour, effectively limiting joint motion and preserving the proportional relationship between speed and current. No signs of instability, saturation, or excessive current draw are observed.

The actuation remains consistent with the increased load during the interaction. This provides consistent and controllable resistance during rehabilitation exercises.

Finally, the last performed test was the ASSIST-AS-NEEDED mode test ([Figure 5.34](#)). In the ASSIST-AS-NEEDED test performed with the user's arm coupled to the end effector, similarly to the previously done AAN test, a partial movement is performed, and then it is finished by the active assistance from the motor. With the only meaningful difference in this test compared to the structure only test being the increased load provided by the arm's mass during motor actuation.

For motor 1 a movement is performed under no load until the 72.7° position during flexion, and the assistive movement starts 7.3 s after the AAN sequence began. The motor actuated assistance finishes the movement, with a similar current curve to the one observed in the ACTIVE Assist test, as expected due to the movement being similar. This is highlighted by the first green box. The second movement is manually performed up to the 30.5° mark, and assistive actuation starts at 2.3 s after the beginning of the AAN sequence. The assist actuation current curve is also similar to the one observed in the ACTIVE Assist mode, as seen in the second green box. The assistance timing algorithm is also preserved, with the assistance starting with a 5 s delay to the expected position in this motor. As observed in the second and fourth red rectangles, the 5 RPM is never achieved

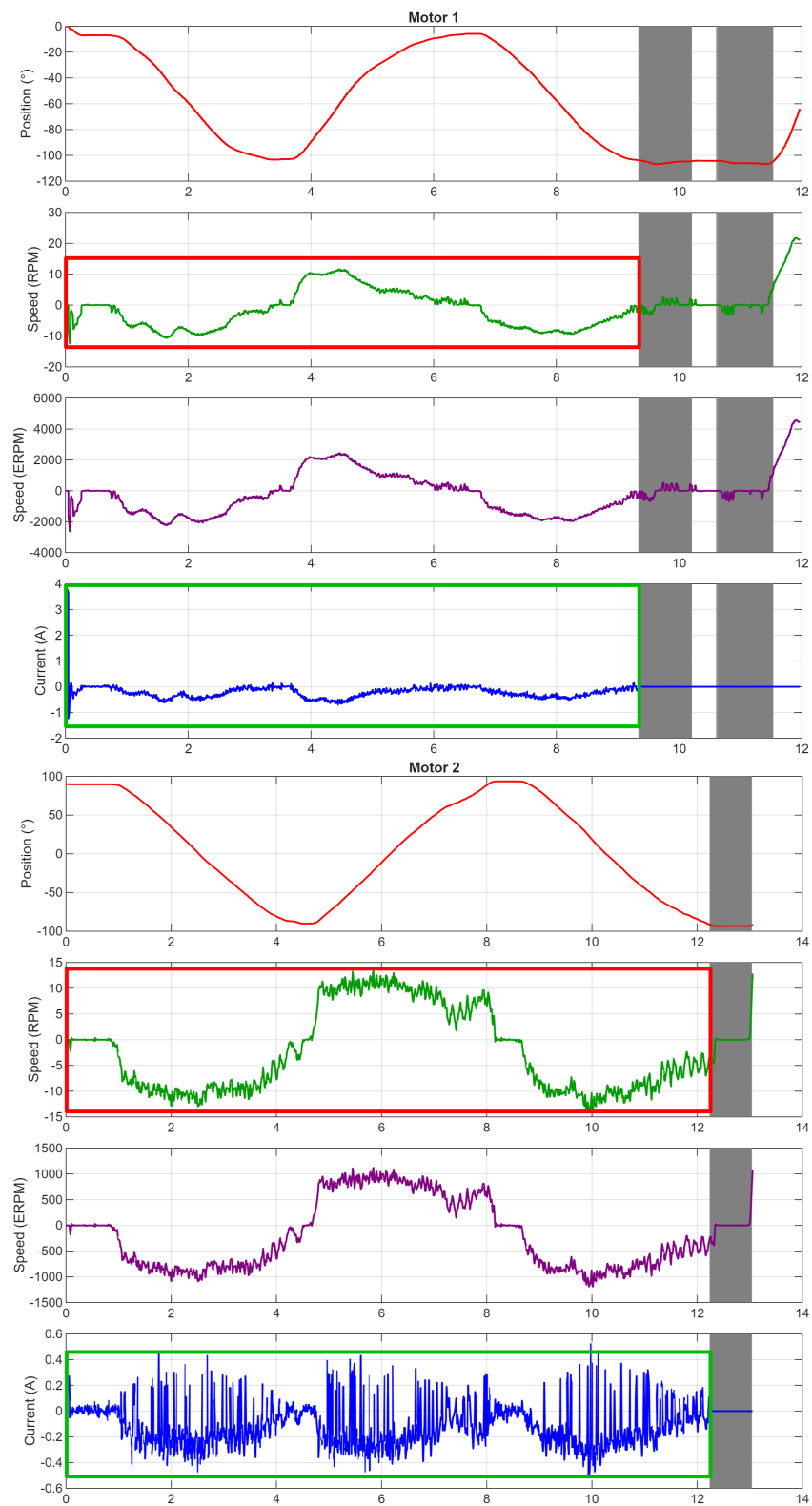


Figure 5.33: Telemetry results for ACTIVE Resist on the AK60 and AK70 motors with User 1 analysed

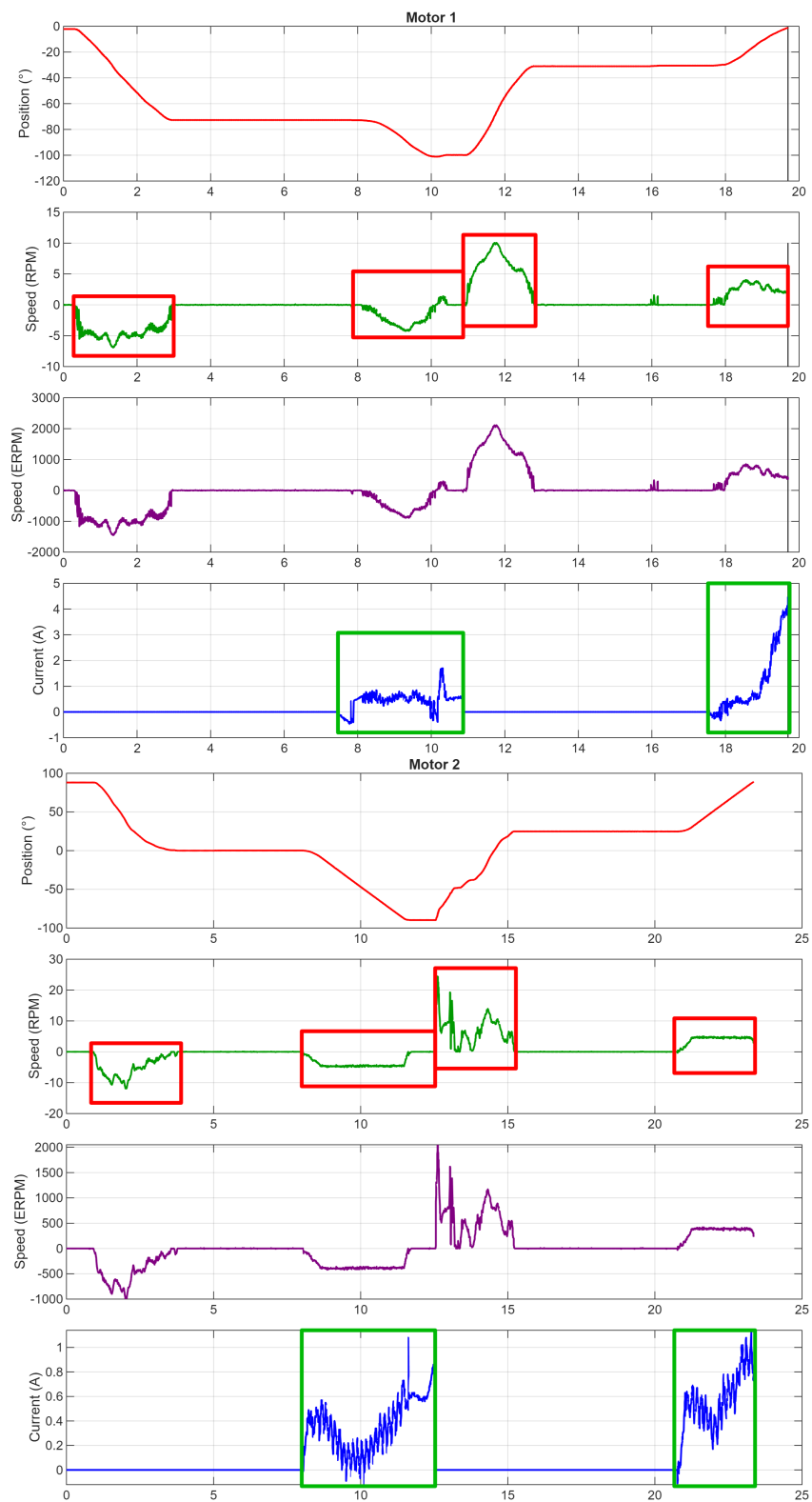


Figure 5.34: Telemetry results for ASSIST-AS-NEED on the AK60 and AK70 motors with User 1 analysed

during both assistive movements. This is due to the required movement amplitude being small, and thus never reaching the defined speed with the defined acceleration parameters. Acceleration is also shown to be less linear. This can be attributed to the force applied by the user's arm on the structure during the hold phase creating increased resistance at the start of the movement. For the second motor a similar behaviour is observed. Timing is respected, and the current curves during the assistive part of the movement are similar to the ones observed in the ACTIVE Assist mode. Some current spikes and variations are observed, and those can be attributed to resistance induced by the wrist. In this motor the 5 RPM is achieved and maintained effectively during the assist phase of the movement.

These results display the correct and effective behaviour of the assisting phase of the AAN control, performing the actuation in the correct timing and performing its given function.

5.5 SWOT Analysis

In order to complement the technical analysis performed above, a SWOT analysis was performed in order to contextualise this project within the existing solutions in the rehabilitation robotics domain. **Figure 5.35** displays the Strengths, Weaknesses, Opportunities and Threats (SWOT) analysis created.

Strengths	Weaknesses
Multiple control modes (ACTIVE Assist, Resist and AAN)	2 DoF only
Real Time telemetry	No Clinical Testing
Adjustable for different physionomies	No Certification
Real time adpatative assist-as-needed mode	No impedance based AAN mode
Low manufacturing cost	Manual Calibration dependancy
Redundant safety features (electrical and mechanical end stops)	No intent detection
Ease of use and operation	Low test volume

Opportunities	Threats
Growing market of rehabilitation robotics	Regulatory approval requirements and costs
Increased operation range when coupled with Cobot	Medical certification requiriements and costs
EMG-based adaptive assistance with intent detection	Already existing, clinically approved solutions
Machine learning-based difficulty adaption alqorithms	Funding constraints for clinical trials
Expansion to multi-joint upper-limb system	Rapid evolution toward soft robotics and wearable exoskeletons
	Reliability concerns under long-term clinical use

Figure 5.35: SWOT Analysis

5.6 Testing Limitations

Although the conducted tests provide an expected validation of the actuator behaviour, control modes, and safety mechanisms, the experimental evaluation is subject to several limitations that must be acknowledged. These limitations are mainly related to the extent of testing. The lack of user based testing and the lack of "real-world" testing, as in testing with proper stroke patients.

All experimental tests were performed, either without a human user or with a healthy user utilising the device on their arm. As a result, the observed system behaviour reflects only the mechanical, structural, and control characteristics of the end effector, without accounting for any particular situation that can be encountered with disabled users. As a consequence, the results cannot be directly extrapolated to real rehabilitation scenarios involving patients or therapists.

The user testing data in this project was limited to only one user due to technical issues. Although this allowed a greater analysis of the system function for human interfacing, it also

removed extra analysis, particularly regarding system comfort and operation under different arm typologies and loads.

The duration of individual tests was limited, and long-term operation was not evaluated. Potential effects related to thermal accumulation, mechanical wear, and control drift over extended usage periods were therefore not assessed. These factors may become relevant during prolonged rehabilitation sessions.

Despite these limitations, the conducted tests provide a robust validation of the system's fundamental operation, safety enforcement, and control behaviour. The obtained results establish a solid foundation for future work involving increased mechanical loading, human-in-the-loop testing, and extended evaluation under clinically relevant conditions.

5.7 Testing Conclusions

This chapter presented the assembly process and experimental validation of the design. Covering the assembly process steps and bill of materials and tested actuators, mechanical structure, software and safety features. The conducted tests were able to demonstrate compliance in the operation of the system when compared to design requirements.

The system validation was performed in several steps, firstly with no load tests followed by full system tests. These first tests allowed for basic actuator, controller firmware and UI software validation. All three work modes, ACTIVE Assist, ACTIVE Resist and ASSIST-AS-NEEDED were validated and behaved nominally, reliably and repeatably in these isolated conditions. Correctly responding to all commanding inputs. The second full system test allowed the full validation of the end effector system. In these tests the actuators behaved normally, being able to compensate for the increased load of the end effector's weight.

Overall, the experimental results confirm that the developed system fulfils its intended operational role as a controllable, safe, and adaptable actuation platform for upper-limb rehabilitation tasks. While the evaluation was limited to bench-top testing without human interaction, the validated functionality provides a solid foundation for future extensions involving increased loading conditions and user-in-the-loop testing.

Chapter 6

Conclusion

The aim of this project was the development of an upper limb human robotic end effector interface for rehabilitation in stroke patients. This goal was successfully achieved. The main objective defined in the initial stage of this work was the creation of an actively actuated interface capable of assisting elbow flexion–extension and wrist pronation–supination movements while ensuring controllability, safety, and adaptability

The developed system integrates mechanical, electrical, and software subsystems into a unified architecture. The mechanical structure demonstrated rigidity and stability under expected load conditions. The actuation system, based on CubeMars brushless DC motors controlled through a CAN based architecture, achieved reliable position, speed, and current control. The firmware implementation enabled modular control routines and stable communication between the user interface and the embedded system, while the software allowed for easy to use, reliable operation through a simple and intuitive user interface.

All implemented rehabilitation modes, ACTIVE Assist, ACTIVE Resist, and ASSIST-AS-NEEDED were experimentally validated. The system responded correctly to commanded inputs, performing movement linearly and softly, it exhibited the expected proportional current behaviour in resistive mode, adding the necessary torque to overcome the resistance, and demonstrated consistent assistive timing actuation in assistive modes. Full-system tests confirmed the actuators' ability to compensate for structural load and validated safety features such as end-stop activation.

Overall, the experimental results confirm that the developed solution fulfils its intended role as a controllable active end effector platform for upper-limb rehabilitation tasks. The work establishes a functional technological foundation for further development toward clinically developed robotic rehabilitation systems.

6.1 Limitations

Despite the successful validation of the mechanical and control subsystems, several limitations must be acknowledged.

The experimental validation was performed under controlled laboratory conditions. Human testing was limited and involved healthy users rather than stroke patients. As a result, the observed system behaviour reflects mechanical and control performance without accounting for any neuromuscular impairments, spasticity, or variability typically present in clinical populations. Consequently, the results cannot be directly extrapolated to real rehabilitation scenarios.

The number of evaluated users was limited due to technical constraints, which restricted the assessment of comfort, adaptability across different anthropometric profiles, and robustness under varying arm typology. Additionally, long-term operation was not evaluated. Thermal accumulation, mechanical wear, backlash evolution, and potential control drift during extended rehabilitation sessions remain untested.

These limitations define the current level of the system as a validated prototype rather than a clinically ready solution.

6.2 Future Work

Future developments should prioritise full system integration with the collaborative manipulator to enable synchronised kinematic coordination between the robotic arm and the end-effector. This integration would allow more advanced movement trajectories and enable advanced control strategies combining manipulator functionality with the developed end effector's modes.

Clinical validation also would constitute a crucial step. Trials involving stroke patients are required to assess therapeutic effectiveness, usability, safety perception, and adaptability to pathological movement patterns. Such evaluation should include quantitative outcome metrics to compare robotic-assisted therapy with conventional rehabilitation approaches.

From a control perspective, the implementation of impedance or admittance control strategies, along with machine learning adaptive difficulty algorithms, would improve interaction safety and compliance. Adaptive assist-as-needed algorithms based on real-time performance metrics or physiological feedback, such as electromyography (sEMG), could further enhance rehabilitation specificity. Further safety sensorisation, such as obstacle detection pinching detection would greatly improve the user's safety during operation

Mechanical refinements should include fatigue testing under dynamic load conditions, extended use analysis, and ergonomic optimisation to improve comfort and long-duration usability. Reduction of overall weight and simplification of assembly could further increase clinical practicality. The use of medically approved materials would also improve the system's readiness for deployment. Currently the system is assembled with 3D printed components and industrial aluminium profiles.

Through these developments, the presented prototype may evolve into a fully integrated and clinically validated robotic rehabilitation platform capable of contributing meaningfully to upper-limb stroke recovery.

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Appendix A

Motor Checkist

The CubeMars motor drivers allow for 2 different modes of operation, Servo Mode and MIT mode, however, for this project only Servo Mode will be utilized. Servo mode encompasses 6 different Control loops, of which several will be utilized in this project. The 6 Control loops are the following:

- Torque Control loop
- Position Control loop
- Current Control loop
- Brake Current Control loop
- Speed Control loop
- Duty cycle Control loop

Servo mode also allows for dual loop control, enabling to utilize the speed and position control loops simultaneously, in this dual loop mode acceleration can also be defined.

In order to test the motor actuators that will be attached to the end effector. A testing check list was created in order to ensure proper operation.

In order to develop this checklist the following video from the manufacturer was used as source for these steps. https://www.youtube.com/watch?v=Y6BMy1lSnvA&list=PL5IcNN8-SrWJU0_3r6SwsieHIdqw1NJTk&index=2

1. Connect the power supply to the motor, the AK70-10 needs to be powered at 24V and a maximum of 23.2A and the AK60 needs to be powered at 24V and a maximum of 22.7A, however the CubeMars driver can be powered at a Maximum of 48V and 60A.
2. Connect motor to the computer through CubeMars' RubiLink (as described in [Figure A.1](#));
3. Start up the CubeMars Upper Computer software, set it to English utilizing the sliding bar on the bottom left ([Figure A.2](#));

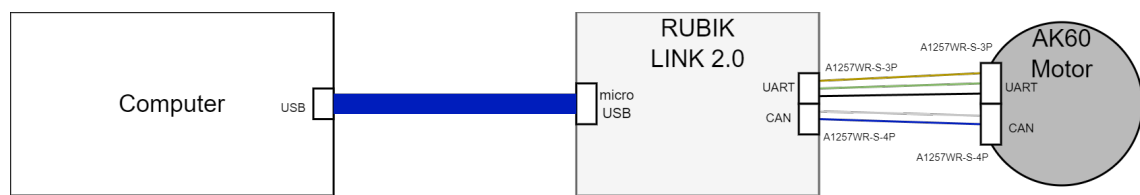


Figure A.1: Wiring diagram for step 1 of the checklist

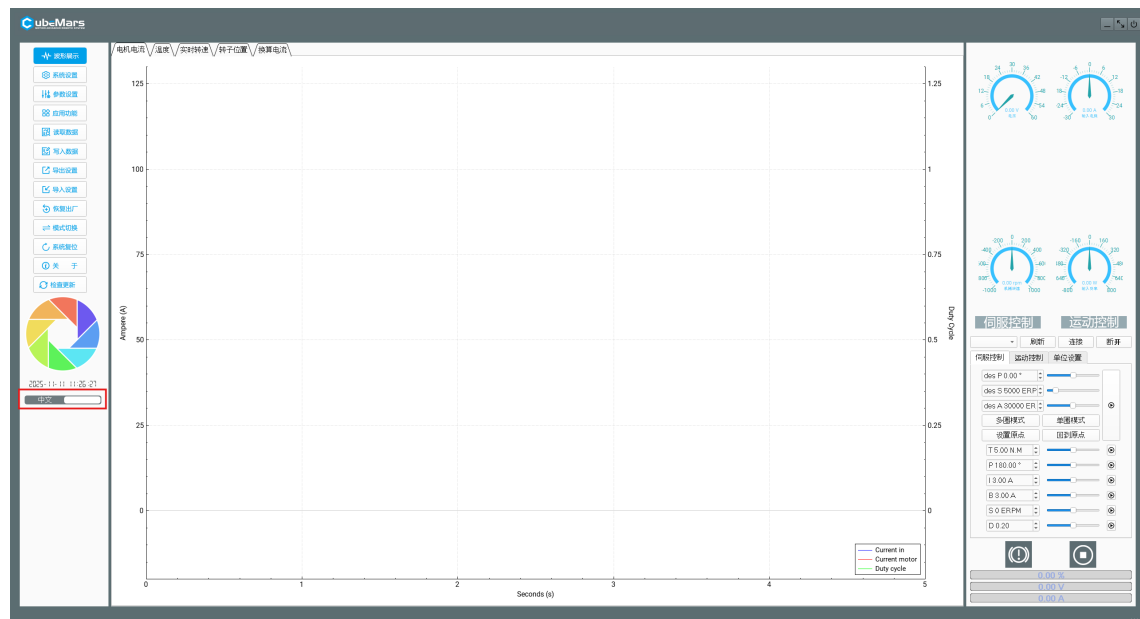


Figure A.2: Upper Computer Language toggle

4. Check the COM port that the motor is utilizing and check the correct driver (CH340) in the Windows Device Manager;
5. Connect the motor in the Upper Computer software utilizing the correct COM port (Check motor light, should be green);
6. Go to the *Mode Switch* side menu and select Servo App (Figure A.3);
7. Set the correct CAN ID (1 for the AK70-10, 2 for the AK60) and check the *send status over CAN* box in the **Application Functions** menu, then proceed with Writing the parameters to the driver, with the **Write Parameters** button on the left side bar (Figure A.4);
8. Calibrate the Servo mode in the **Parameter Settings** menu, then proceed with Writing the parameters to the driver, the tuning parameters will be kept as default. In order to perform the calibration the following steps must be followed (Check Figure A.5);
 - (a) In the **Parameter Settings** at the bottom of the page the **Detect and Calculate Parameter** click **Measure R/L**, and press OK in the pop up window;
 - (b) When the R, L, Kp and Ki boxes bellow the button turn blue, press **Measure Lambda**, and press OK in the pop up window (The motor will rotate in this step, it is nominal);

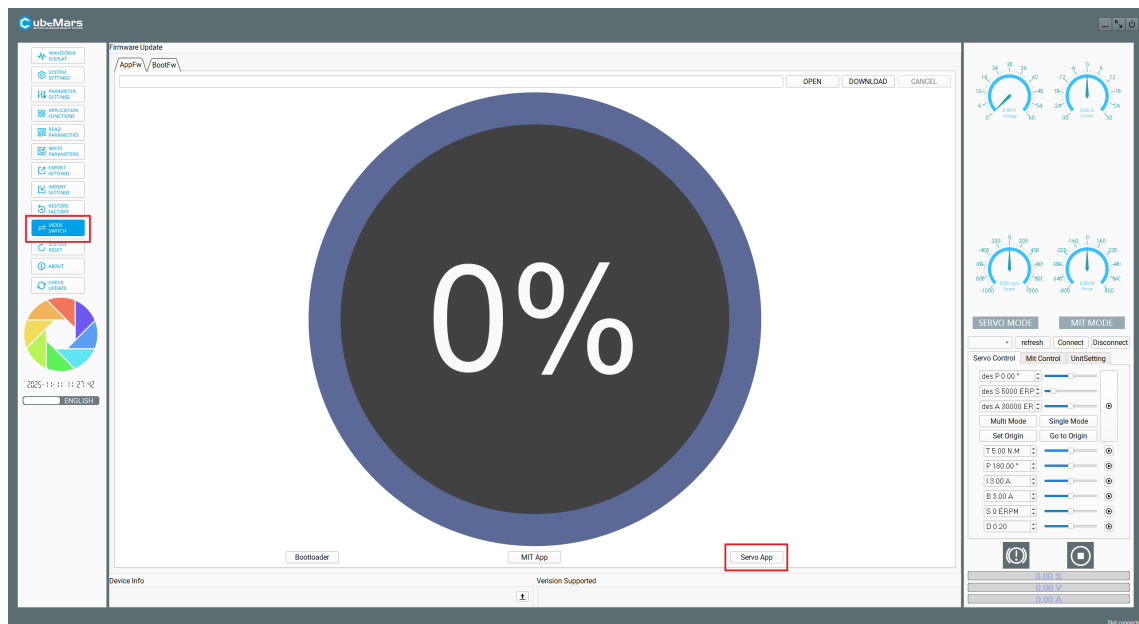


Figure A.3: Servo Mode Switch

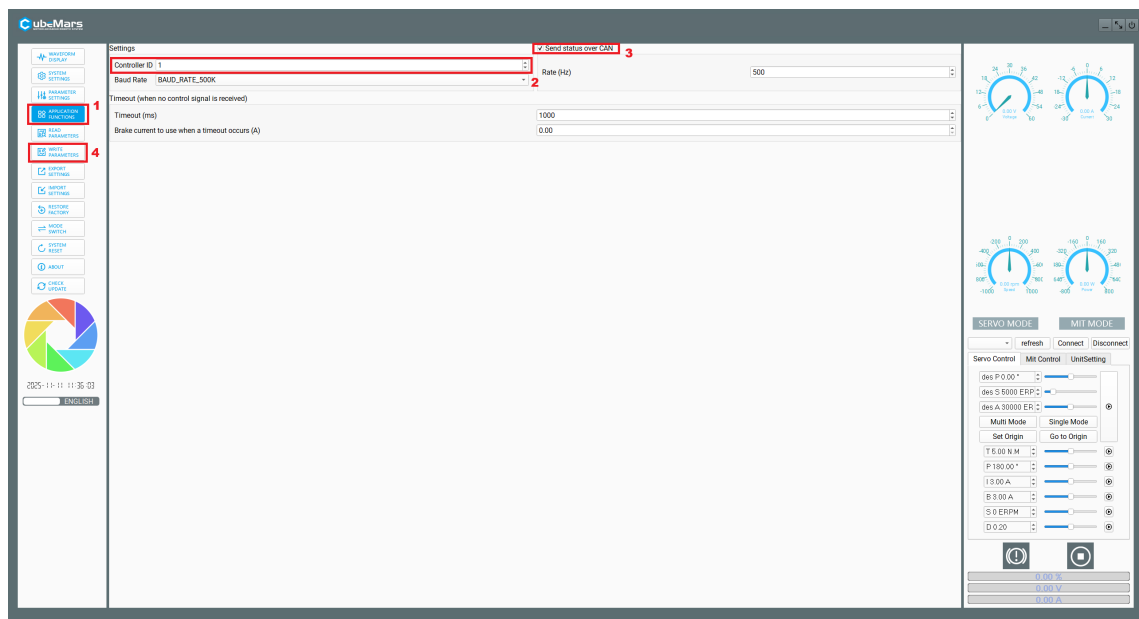


Figure A.4: Motor Application Functions Configuration

- (c) After all 6 boxes bellow this button turn blue, press **Update** and check for a green message in the bottom right of the screen with the **KP, KI and Observer gain applied** text;
- (d) At the top of the same page, in the Detect Encoder submenu, press the **Start**, and click OK in the pop up window (The motor will rotate in this step, it is nominal);
- (e) Once the motor stops rotating and making any movement or noise, press the **Update** button;

(f) Press the **Write Parameters** button on the left side of the screen.

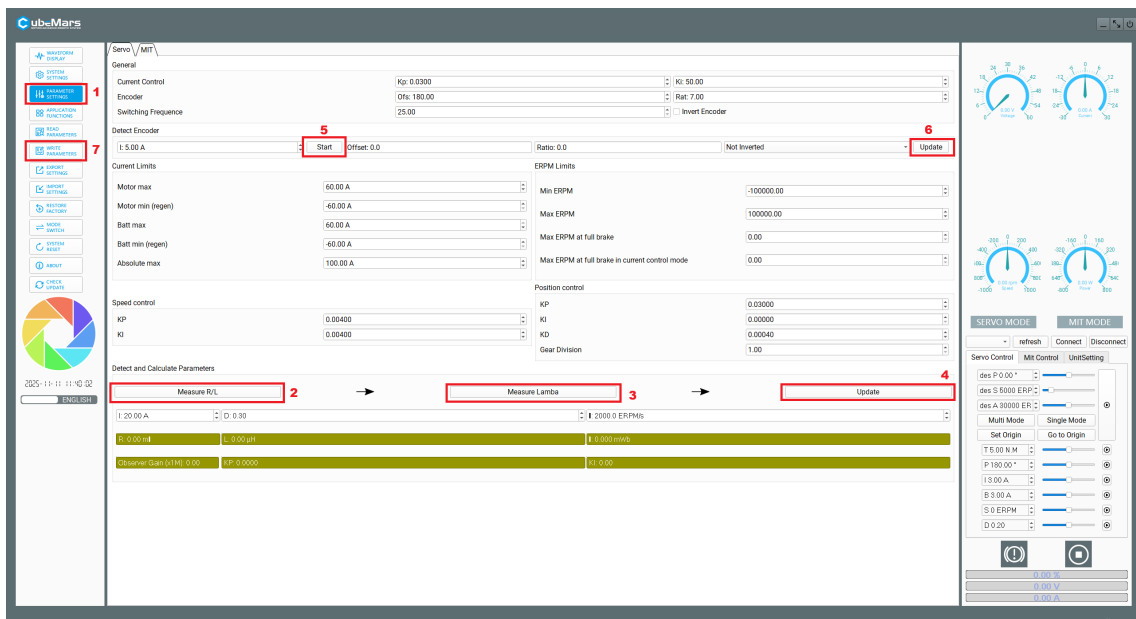


Figure A.5: Servo Mode Calibration

9. Test motor movement with the existing control loops in the Upper Computer control panel (right side of the screen) ensuring the Servo Control page is selected (Figure A.6);

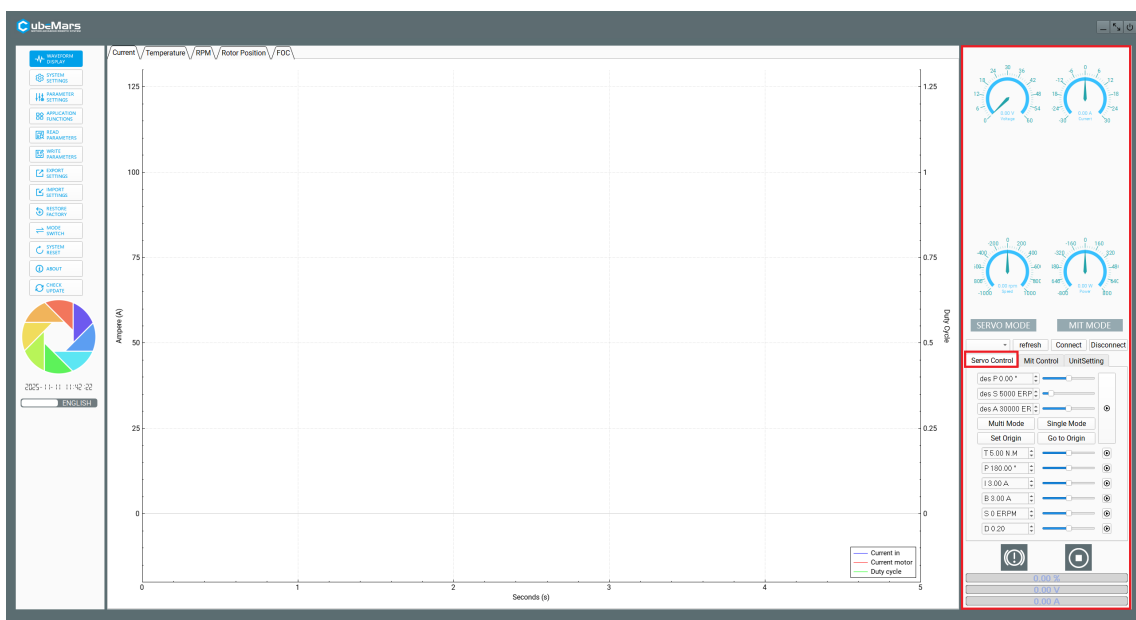


Figure A.6: Test Setup

- Test Speed Control loop. Start with low values and increase gradually. Starting at 10rpm speed;

- Test Position Control loop. With $+90^\circ$ movement and then -180° movement;
- Test Current Control loop starting at 0.5A and increasing by 0.5A up to 5A;
- Test Torque Control loop starting at 0.5Nm and increasing by 0.5Nm up to 5Nm
- Test MultiMode Control loop with the combination of the above values.

10. Connect the motor through CAN with the microcontroller (**Figure A.7**) to a test code executing simple functions emulating the test setup performed with the Upper Computer Software;

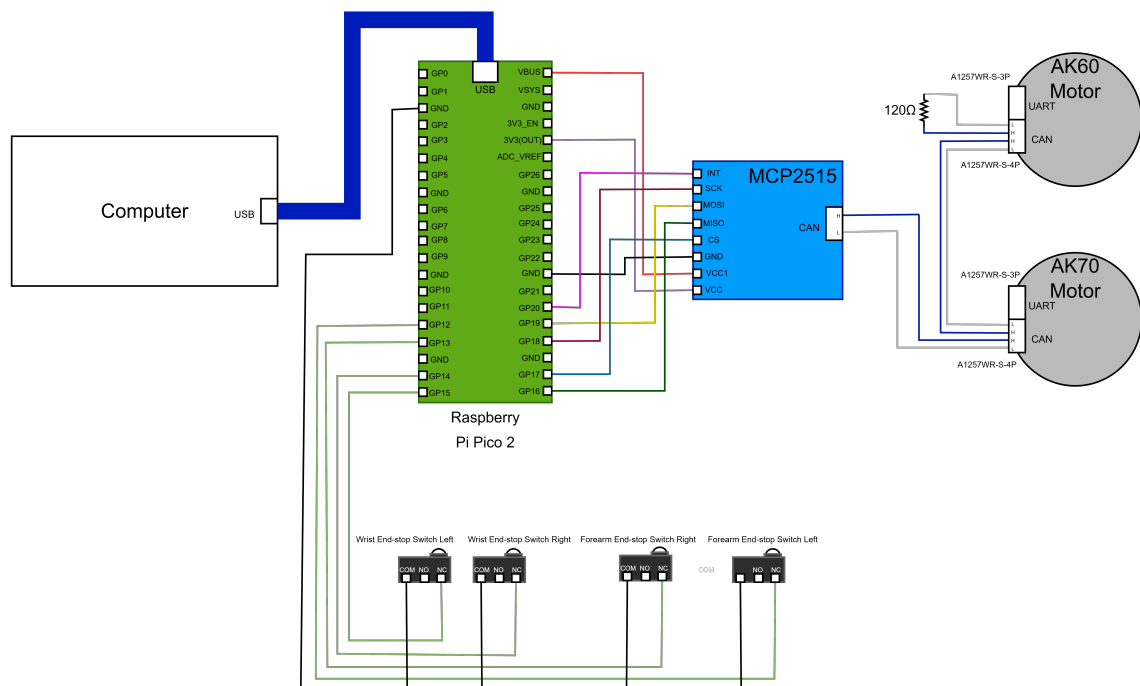


Figure A.7: Wiring diagram for step 10

11. Connect the motor to the UI application ensuring correct connection and transmission reception;
12. Full code testing alongside UI application transmission reception and output.

Appendix B

User Test Evaluation Form

Name _____ Date _____

Age _____ Sex _____ Height _____

Palm length	
Forearm length	
Arm length	

Mode	Motor	Rep.	Start	End	Speed
Active Assist	M1				
	M2				

Mode	Motor	Torque
Active Resist	M1	
	M2	

Mode	Motor	Rep.	Start	End	Speed
Assist-as-Needed (AAN)	M1				
	M2				

Subjective Evaluation

Comfort Score (1–10) _____

How safe did you feel using the system? (1–10) _____

How easy did you find the system to use? (1–10) _____

How supported did your arm feel during operation? (1–10) _____

Did the device at any moment cause pain or discomfort?

Yes _____ No _____

If yes, please describe:

Extra Comments:
