Pontifícia Universidade Católica
do Rio de Janeiro

Daniel Rivas Alonso

Advances towards an actuated orthosis for the rehabilitation of the motor function of the hand

Dissertação de Mestrado

Dissertation presented to the Programa de Pós-graduação em Engenharia Elétrica of PUC-Rio in partial fulfillment of the requirements for the degree of Mestre em Engenharia Elétrica.

Advisor: Prof. Marley Maria B. R. Vellasco
Co-advisor: Prof. Eduardo Costa da Silva

Rio de Janeiro
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To my parents, Rafael and Rosalinda,

You are my guide and my strength.

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Abstract

Rivas Alonso, Daniel; Vellasco, Marley Maria Bernardes Rebuzzi (Advisor); Costa da Silva, Eduardo (Co-Advisor). **Advances towards an actuated orthosis for the rehabilitation of the motor function of the hand**. Rio de Janeiro, 2017. 107p. Dissertação de Mestrado – Departamento de Engenharia Elétrica, Pontifícia Universidade Católica do Rio de Janeiro.

Strokes are a type of brain injury that affects over 750,000 people annually. Approximately half of the patients diagnosed with a stroke suffer chronic damage of the upper extremity function. Rehabilitation helps the patient to keep abilities and recover some of the lost ones, to become more independent. Hand rehabilitation exercises aim at assisting patients so they can regain finger mobility and strength. An actuated hand orthosis is a potential therapy tool for distal upper extremity weakness, since it can offer a promising approach to improve lost motor behaviors in stroke patients. Nowadays, hand movement assisting devices are developed for research applications, and even for commercial purposes. However, most of them are expensive, heavy and bulky or do not offer control over their whole operation. This work proposes and develops a new actuation system that can contribute to the construction of an easy to operate, portable, low cost and fully controlled hand movement assisting device. Several advances towards the creation of an actuated hand orthosis were achieved, leading to the creation of an electromechanical system capable of assisting finger movement along their full range of motion, while keeping low weight in the distal upper limb. The system is controlled by a computer software with a graphic user interface that allows the users to configure the system's parameters to their specific needs. The control software also allows the communication with a Brain-Computer Interface (BCI) in order to synchronize the system's movements with the user intentions, improving the recovery rates.

Keywords

Hand orthosis; Hand rehabilitation; Stroke; Brain-Computer Interface.

Resumo

Rivas Alonso, Daniel; Vellasco, Marley Maria Bernardes Rebuzzi (Orientador); Costa da Silva, Eduardo (Co-orientador). **Avanços em direção ao desenvolvimento de uma órtese automatizada para a reabilitação da função motora da mão**. Rio de Janeiro, 2017. 107p. Dissertação de Mestrado – Departamento de Engenharia Elétrica, Pontifícia Universidade Católica do Rio de Janeiro.

Os acidentes vasculares cerebrais (AVC) são um tipo de lesão cerebral que afeta mais de 750.000 pessoas anualmente. Aproximadamente metade dos pacientes com diagnóstico de AVC sofre dano crônico da função da extremidade superior. A reabilitação ajuda o paciente a manter as habilidades e recuperar algumas das perdidas. Uma órtese automatizada de mão é uma potencial ferramenta terapêutica para tratamento da debilidade da parte distal da extremidade superior, sendo uma abordagem promissora para melhorar os comportamentos motores perdidos em pacientes com AVC. Hoje em dia, a maioria dispositivos de assistência para o movimento de mão desenvolvidos são caros, pesados e volumosos, além de, muitas vezes, não oferecer controle sobre toda a sua operação. Este trabalho propõe e desenvolve um novo sistema de atuação que pode contribuir para a criação de um dispositivo de assistência a movimentação da mão, que seja fácil de operar, portátil, de baixo custo e totalmente controlado. O sistema é controlado por meio de um *software* com uma interface gráfica de usuário que permite que os usuários configurem os parâmetros do sistema de acordo com suas capacidades específicas. O *software* de controle também permite a comunicação com uma Interface Cérebro-Computador, que possibilita sincronizar os movimentos do sistema de acordo com as intenções do usuário, aumentando as taxas de recuperação dos pacientes.

Palavras chave

Órtese de mão; Reabilitação da mão; Acidente vascular cerebral; Interface Cérebro-Computador.

Summary

Acronyms

AC:	Alternating Current.
ADL:	Activities of daily living.
BCI:	Brain-Computer Interface.
CMC:	Carpometacarpal.
DC:	Direct Current.
DIP:	Distal interphalangeal.
EEG:	Electroencephalogram.
FES:	Functional electrical stimulation.
ICF:	International Classification of Functioning, Disability and Health.
IEC:	International Electrotechnical Commission
IP:	Interphalangeal.
ISO:	International Organization for Standardization.
MCP:	Metacarpophalangeal.
MI:	Motor Imagery.
MI-EEG:	Motor imagery electroencephalogram.
PIP:	Proximal interphalangeal.
ROM:	Range of motion.
sEMG:	Superficial electromyography.
SMR:	Sensorimotor rhythm.
USD:	United States Dollar.

Figures List

I do not think there is any thrill that can go through the human heart like that felt by the inventor as he sees some creation of the brain unfolding to success... such emotions make a man forget food, sleep, friends, love, everything.

Nikola Tesla

1.
Introduction

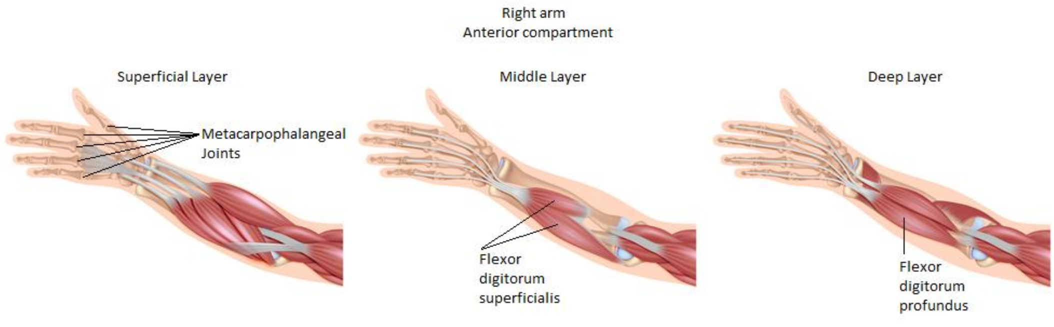
1.1.
Motivation

A stroke is a type of brain injury that affects over 750,000 people annually, and its symptoms depend on the brain area that is affected [1]. The most common type of stroke is caused by a reduction in the blood flow to the brain, when blood vessels are too narrow for blood to get through or are blocked by a clot. These are known as ischemic strokes. Due to the lack of oxygen, brain cells in the affected area die. Another type of stroke occurs when the blood vessels burst, leaking blood into the brain and damaging the surrounding area. Although strokes can affect people of all ages, they are more common in older people, with almost 75% of the registered cases occurring in people more than 65 years old [2].

After a stroke, physical disability is experienced by nearly half of all survivors [3–5], which is characterized by the loss of dexterity and strength in the affected side of the body or difficulties in walking, talking or thinking [2, 6]. The loss of strength is caused by the loss of motor function and muscular recruitment coordination. This means that the brain is injured, but the muscles and nerves are still functional [1]. About a third of all patients experience chronic hemiplegia or hemiparesis, which are paralysis or weakness, respectively, on one side of the body caused by an injury on the contralateral side of the brain [2–4, 7]. There were approximately ten million chronic stroke survivors with hemiparesis in developed countries around the world, in 2016 [8].

Chronic hemiparesis is especially prevalent in the distal upper extremity, the hand. For that reason, approximately half of the patients diagnosed with a stroke suffer from a unilateral motor deficit, which leads to chronic damage of the upper extremity function [9]. This disability, whether partial or total, affects the patient`s life quality by inhibiting activities of daily living (ADL) and reducing the engagement in community life [10–18]. The motor function most commonly affected is the ability to extend fingers. A few weeks after stroke occurrences, a stereotypical flexed finger posture is very usual [19, 20]. This is mainly caused by two factors: the weakening of the extrinsic finger extensors and the increased resistance to passive finger extension of the flexor digitorum superficialis and flexor digitorum profundus, which are shown in Figure 1 (b) and (c) [21, 22] (based on [23]). According to [20, 23] more than 70% of the severely impaired stroke subjects present a significant reduction in the metacarpophalangeal extension torque (Figure 1 (a)), that typically decreases to only 7% of the metacarpophalangeal flexion torque shown by normal subjects. Such impairment limits the ability of the stroke survivor to properly grasp or release an object [21]. Unfortunately, to perform ADL, a reasonably accurate hand motor function is necessary. Therefore, stroke survivors are frequently very dependent on compensatory strategies.

The stroke recovery process generally includes treatment, spontaneous recovery due to natural healing, rehabilitation to improve recovery and, finally, the return to community living. The first stage of treatment involves acute care to help the patient survive and prevent another stroke. Soon after the stroke, some lost abilities start to return naturally to most people, which is known as spontaneous recovery. On the other hand, rehabilitation is a part of the treatment that helps the patient to keep abilities and even recover, partially or totally, some of the lost ones, in order to become more independent. The last stage is the return to community living, which can last a lifetime as patients and their families learn to live with the remaining effects of the stroke [2].



(a) (b) (c)

Figure 1 - Some functional elements of an upper extremity. (a) Superficial muscle layer. (b) Middle muscle layer. (c) Deep muscle layer.

To this end, hand rehabilitation exercises aim at assisting patients so they can gradually regain finger mobility and strength. The repetitive controlled motion of the hand has demonstrated to be a helpful rehabilitation exercise for regaining motor function after a stroke [24–28]. Strength rehabilitation of the paralyzed hand can be improved by this method [27], which also benefits the recuperation, since motor functionality benefits from strength recovery [29]. Repetitive exercises forcing fingers to move from an open state to a closed one, moving the various finger joints through their full range of motion, are suggested for patients with surgically repaired fingers after traumatic injury. This type of exercises also helps to remap the brain motor function, allowing the patients to relearn how to move their own bodies again, in a very similar way that a child learn to walk for the first time [1]. This feature is known as brain plasticity.

Despite spontaneous recovery and intensive rehabilitation, more than a third of stroke survivors remain with an affected hand function, even one year after the stroke [30]. Actually, after the first 6 to 8 months, it is uncommon to experience significant functional recovery and there are just a few treatments for these chronic individuals [24, 31–33]. These recovery rates could be improved if patients were able to perform rehabilitation exercises more often and systematically. The problem is that these exercises are costly and labor intensive, since they require long hours of training with physical therapists due to the patients incapability to correctly perform these finger exercises without assistance [10]. A possible solution to this problem is the development of a functional, user-friendly, low cost and easily controlled assisting device that can allow patients to carry out the exercises on their own, either at their homes or in a clinic, making physical therapy more accessible and comfortable [34].

Certain areas of the brain are naturally and spontaneously activated when a person imagines a motor activity (body movement). Several studies have shown that the same specific areas of the brain are activated when a person effectively performs the same motor activity. This feature of the human brain is known as Motor Imagery (MI) [35, 36]. This behavior supports the idea that MI involves similar neural mechanisms to those operating during real movements [37]. This phenomenon provides a mean to access the motor system for rehabilitation at all stages of stroke recovery [38], since the ability to perform MI remains intact even after the onset of a stroke [39–41]. Nowadays, the electroencephalogram (EEG) is the most popular method to record the human brain electrical activity. The measurements are performed by electrodes placed on the scalp [42]. The EEG signals vary according to the type of MI performed. For example, the attempt to imagine a hand movement will imply in the attenuation of the EEG signals from the brain area responsible for the hand movement at the contralateral sensorimotor cortex [43, 44], and the increment of the EEG signals at the ipsilateral sensorimotor cortex [45], as shown in Figure 2 [46].

Motor imagery electroencephalogram (MI-EEG) is a promising therapeutic approach for the recovery of lost motor function. By using it, patients can learn how to modulate the amplitudes of their sensorimotor rhythms (SMRs), which may support functional recovery by stimulating cortical reorganization [47, 48]. Healthy subjects imagine several different movements and each one of their respective EEG patterns are registered and properly classified. On the other hand, paralyzed patients attempt to reproduce the same movements imagined by healthy subjects [49]. Based on the classification, the behavior of an assisting device can be controlled to help the patient to achieve the attempted movement, providing a biofeedback that can enhance the rehabilitation. The neural-rehabilitation training, however, is generally effortful and requires a lot of training over a prolonged period of time to facilitate adaptive cortical reorganization [50].

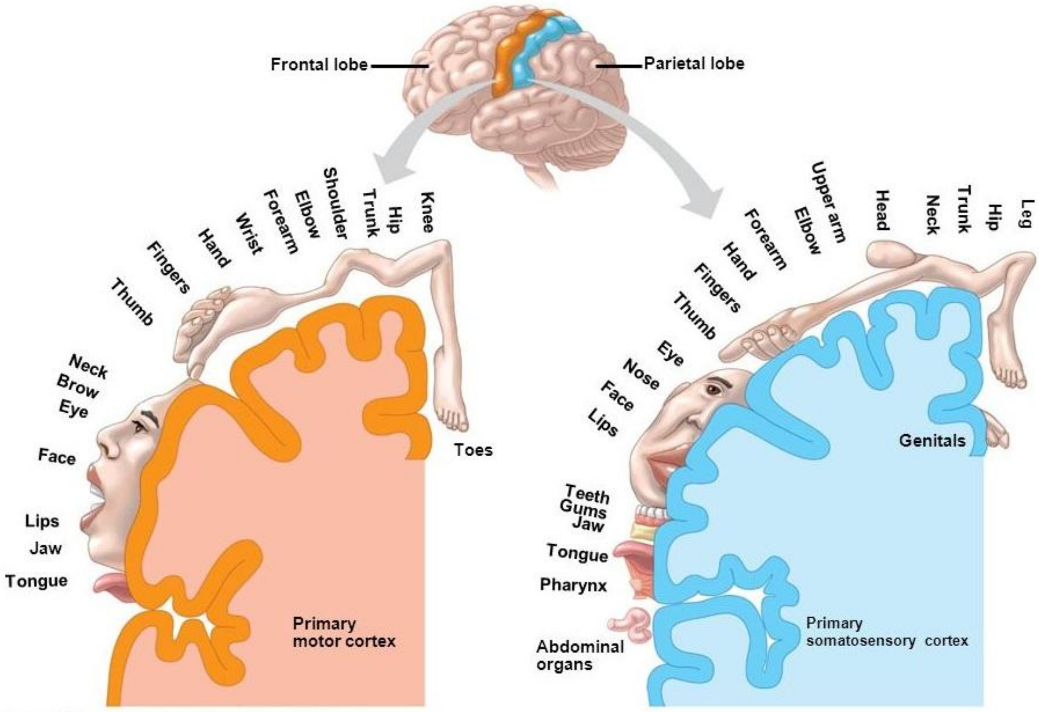


Figure 2 - Body parts representation in primary motor and primary somatosensory cortices.

A Brain-Computer Interface (BCI) is a communication and control system that translates human intentions into commands for external devices, such as computers, switches, wheelchair controls [51] or prostheses [52], by analyzing and recognizing the electrical signals from the brain [53]. It acts as a communication pathway between the mind and the environment without using the usual output pathways of peripheral neural system and muscles [53–59]. The brain activity can be accessed by taking measurements directly from the scalp, the cortical surface or cerebral subcortical areas. The EEG is the most popular recording technique of brain electrical activity used in BCI, because it is one of the less invasive techniques.

A hand orthosis driven by an MI-based BCI is a potential therapy for distal upper extremity weakness caused by strokes, aiding in the movement and coordination of gripping exercises, since it can offer a promising approach to restore completely or, at least, substantially improve lost motor behaviors in stroke patients [60–62]. A BCI based on the SMR modulation, which is highly correlated with voluntary movements [63–65], can be used in the neural-rehabilitation, promoting the reinforcement of new connections between the post-stroke motor cortex and spinal motor pools associated with the specific lost motor function [49, 66–68].

The aspects highlighted in this section indicate that the recovery process can be enhanced with the assistance of electro-mechanical devices for post-stroke hand rehabilitation. As a result, over the last years, many machines have been created to assist with hand movement and therapy, showing promising results when compared to traditional methods aimed at recovering the motor functionality [69–79]. However, current designs often present limitations that restrict their usage in long duration or home-based rehabilitation programs. The most common limitations are the high cost, the occlusion or restriction of some joints' movements, the assistance of only one type of motion, the non-portability due to their size and weight, the incapacity of being used in patients with totally paralyzed arms, and the impossibility to grasp real objects while using the device. Therefore, there is room for improvements focusing on the development of a more functional, low cost and portable BCI driven orthosis, flexible enough to be easily adapted to the particular characteristics of each patient, overcoming the limitations present in current designs.

1.2. Objectives

This work has the objective to achieve advances towards the development of a low cost and lightweight actuated orthosis, that can be used to improve the quality of current rehabilitation therapy techniques, conventionally applied to stroke survivors presenting hemiparesis in the upper distal extremity. The orthosis should be able to assist the users in performing repetitive exercises involving opening and closing movements of the hand, giving different levels of assistance according to the patient`s specific needs. An actuated orthosis with such characteristics can improve the quality of nowadays rehabilitation techniques and also contribute to the dissemination of this new kind of assistive devices through the health sector.

Some main concerns should be taken into account in the design of the idealized actuated orthosis, aiming at the development of a device capable of ensuring the users' safety and control at all times. The device must be controlled by computer software, aiming to assist the user in achieving the desired hand motion. On the other hand, the orthosis should be capable of establishing a direct communication with an EEG-based BCI to detect the user intended movements. Finally, a user-friendly graphic interface must be developed in order to adjust the orthosis parameters to the users' specific needs.

The ultimate goal is to develop a device that would allow the users to perform their rehabilitation programs with minimal physical therapist assistance. This dissertation contributes to the development of such a system, focusing on aspects related to testing the electromechanical system's performance and developing the computer software to control the entire device along with the graphic user interface and the connection to a BCI.

1.3. Project Overview

This work was developed according to the following stages:

- A study of the human hand`s anatomy;

- A study about neural rehabilitation therapies for hand motor function recovery;
- A study on actuated hand orthoses;
- A study of robotic control systems;
- Conceptual definition and development of a test bench; and
- Evaluation and testing.

In the first stages, a bibliographic research was performed about the anatomy of the human hand and its functional aspects. After that, it was conducted a detailed review on how a stroke affects the normal performance of the motor function. Furthermore, a study of the current exercises and therapeutic strategies used in patients to regain hand functionality was accomplished.

Research on the state of the art regarding actuated hand orthoses and devices used in motor hand function recovery was also carried out. This included an analysis of the advantages and disadvantages of existing systems. Additionally, different control systems were studied and evaluated, to find the most suitable to the project`s needs.

The stage of conceptual definition had the objective to design a system capable of overcoming the limitations of the studied rehabilitation devices. Based on the proposed design, an experimental test bench was developed for evaluating the system performance and its capabilities.

1.4. Dissertation Structure

This thesis is organized in four additional chapters.

- Chapter 2, Conceptualization, contains the main concepts of the present work. The sections cover neural rehabilitation bases and the anatomical elements of the hand. A review on the existing actuated orthoses and assisting devices is included, as well as a discussion about the components used in the proposed design.
- Chapter 3, Orthosis design, presents the characteristics and concepts of the proposed model, describing the actual

components of its design, the controlling software, and the graphic user interface developed.

- Chapter 4, Case study, includes the results of the tests executed over the model to evaluate its performance.
- Chapter 5, the conclusions and foreseen future work are presented.

2. Contextualization

This chapter introduces the main concepts and definitions related to the design of an assistive device for rehabilitation exercises of the motor function of the hand. A review of rehabilitation devices developed in recent years is also presented, highlighting the state of the art in the area. Finally, some of the main components considered in the design of the device idealized in the present work are evaluated according to the project's requirements.

2.1. Neural rehabilitation

The first step of the research was a bibliographical review aiming at learning about hand motor function recovery procedures and rehabilitation, to understand the functions that the proposed device would have to perform.

Rehabilitation is a part of the post-stroke recovery treatment, its main goal is to improve physical, mental and emotional functions for restoring as much independence as possible to the patient [2]. Rehabilitation procedures must pursue that goal preserving the patients dignity and motivating them to re-learn basic skills that may have been affected by the stroke, such as eating, dressing and walking [2, 80].

Neural rehabilitation, as a concept, comes from the theories that support neural plasticity, which is defined as any change in neuron function or structure observed either directly from measurements of individual neurons or inferred from measurements taken across populations of neurons [81]. The Hebbian learning theory is the origin of the use of neural plasticity to induce behavioral change [82]. Hebbian learning strengthens synaptic connections when pre- and post-synaptic neurons are co-active, which gives it an important role in neural plasticity [82]. Such co-activation may be accomplished when two or more neurons, that were disconnected by an injury, are activated at the same time [83]. If both neurons are separately connected to a fully interconnected circuit of neurons, then the two neurons are activated simultaneously when the circuit of neurons is activated [83]. This means that Hebbian learning may help to explain how cortical functions may be regained through the recovery of the original connections pattern in partially injured neural circuits. This concept is important for neural rehabilitation and neural plasticity, since it helps to define the framework for therapies attempting to optimize recovery mechanisms.

The lack of use of the affected arm-hand in daily life situations, because of inactivity, spasticity, and/or stiffness, can lead to secondary complications, including pain, problems in performing basic ADL and hygienic issues [84, 85]. About 40% of stroke survivors have serious injuries caused by falls, within a year of their strokes, because of balance issues and arm or leg paralysis. However, with rehabilitation and therapy, these patients may improve their balance and ability to move [80]. Generally, the factors that improve post-stroke recovery include early intervention, movement repetition and motivation. Those patients who are more persistent and active in their rehabilitation, usually, have a better chance to regain more of their functions [29, 86].

Over the last decades, the treatments to motor rehabilitation, focused at post-stroke arm-hand performance recovery, have changed considerably. Nowadays, the treatment approaches are focused on the International Classification of Functioning, Disability and Health (ICF) activity and participation level, while, previously, they mainly focused on ICF function level [17]. In recent years, well-explored training approaches have emerged [84, 87–89] addressing impaired motor control and paresis [90–92]. These programs include exercises that the patients can use in therapeutic or home-based situations [93–98]. Furthermore, they make use of training features such as: specificity, challenge, feasibility, meaningfulness, and high-intensity and task-oriented training [99]. Constraint-induced movement therapy [100] and task-oriented training [101–103] focus on the ICF participation level and activity level.

Some rehabilitation goals must be established based on the effects of the stroke in the patient capabilities and on the patient`s wishes. Most patients consider rehabilitation a hard work, since the recovery procedures focus on maintaining abilities and, at the same time, on regaining others. It is normal for patients to feel discouraged and tired sometimes, because activities that used to be easy can become very difficult to perform after the stroke [2]. For that reason, it is important to establish realistic goals. If they are too ambitious, the patient will not be able to reach them, which could lead to loss of motivation. On the other side, if they are too easy, the patient may not achieve its full recovery potential [2].

The most common rehabilitation goals aim at regaining the capacity to perform some daily living activities, focusing on being able to: walk, at least with a cane or walker, take care of oneself, with some special equipment, drive a car and have a job. However, to reach the treatment goals does not mean that the recovery is complete. It just points out that the patient and the family are ready to continue recovery on their own [2].

The rehabilitation process should start as soon as possible after the stroke, even while the patient is still in the hospital. The rehabilitation options will depend on several factors, including the ability to tolerate the intensity of the rehabilitation program and the degree of disability [2, 80]. The most important variable is the absence or presence of dexterity in the affected arm-hand. Therefore, it is recommended to stratify patients with an impaired arm-hand into a number of dexterity levels, in order to define the potentially most effective treatment approach [89, 104].

There are many kinds of rehabilitation programs for post-stroke recovery. Hospital programs are usually the most intense, requiring a greater effort from the patient. In these programs, complete rehabilitation services are available since they can be provided by rehabilitation units in acute care hospitals or by special rehabilitation hospitals. An organized team of specially trained professionals provides the therapy, while the patient stays in the hospital during the whole program [2].

On the other hand, the so-called nursing facility programs also require that the patient stays at the facility during the rehabilitation. However, some of these programs provide more limited rehabilitation services than those performed in hospitals [2].

In turn, the outpatient programs allow patients to get all the rehabilitation services while living at home, by visiting an outpatient hospital department or rehabilitation facility, where the patient will spend many hours a day in physical therapy, occupational therapy, speech therapy, recreational therapy, group activities, and patient and family education [2].

Finally, the home-based programs have the advantage that patients learn skills in the same place where they will use them, since they receive the rehabilitation services at home from visiting professionals [2].

Due to the brain damage caused by the stroke, the electrical signals that flow from the brain to the muscles often do not work right. This damage can cause paralysis or spasticity of any limb. The first one is characterized by the inability of a muscle or group of muscles to move on their own. The latter is a condition where muscles are stiff and resist being stretched. There are several treatment options for spasticity, which are generally a combination of therapy and medicine. The therapy can include range-of-motion exercises, gentle stretching, and splinting or casting. The medicine can act on multiple muscle groups in the body and treat the general effects of spasticity. These treatments may include [80]:

- injections of botulinum toxin to prevent the release of chemicals that cause muscle contraction;
- direct delivery of a drug into the spinal fluid using a surgically placed pump; and, as the last resource;
- surgery on the brain, muscles and joints to block pain and restore some movement.

A good way to improve the recovery is through physical exercise. With the objective of keeping the patient's body strong and flexible, it is recommended for them to walk, bend and stretch. Moving a weakened or paralyzed body part is a good exercise, that can be done while the patient is sitting or lying down [80].

Task-oriented approaches train the patients in specific functional tasks, skill-related, normally using objects commonly found on ADL [102], thereby teaching the stroke survivors how to solve specific problems related to issues, such as cognitive processing or anticipatory locomotion adjustments by using efficient goal-oriented movement strategies [105]. The transfer of learned skills to non-trained skills occurs when there are similarities between them. This kind of training strategy has a higher functional benefit than the muscle strength training [99].

The rehabilitation of the strength and motor function of the hand can be improved by a repetitive controlled motion of the paretic hand [27]. This is done by individualizing the movements associated with a given task and practicing repetitively each one of these movements to improve hand accuracy, strength, and range of motion [10, 76]. The repetition of ADL tasks is often involved in occupational therapy for post-stroke rehabilitation. To build up strength and dexterity, the patient is involved with several games and tasks during the recovery treatment, including exercises like picking up objects and placing them elsewhere, eating, dressing, and many similar tasks that require coordinated manipulation of objects, and opening and closing the hand. Moreover, the patient's level of functionality and the occupational therapist assessment are the elements that determine the level of difficulty of each task [106], that is gradually increased as the patient's ability and functionality improves. This kind of therapy is more common in hospital settings. However, it can migrate toward home therapy, incorporating the recovery of ADL functions as well as environmental adjustments at home, helping to improve efficacy. Home-based rehabilitation procedures benefit the recovery of the psychological and functional performance, and the patient's independence [107].

As part of the rehabilitation, patients may use different kinds of therapeutic equipment. In the early stages of recovery, assistive intervention allows the patient to regain functions. Patients at a higher functional level can strengthen their bodies through resistance exercises. In the past, it was thought that strengthening spastic muscles might do more harm than good. However, a recent review of 13 studies including 517 stroke survivors with mild to moderate arm impairment indicates that strengthening spastic arm-hand muscles with small weights, resistance bands, and pulley weights, can even reduce spasticity [108]. Also, it has been demonstrated that building strength increases function, proving that exercising the affected area benefits the recovery [29]. To develop various motor functions, a combination of assistance and resistance can be used in some patients' rehabilitation sessions [86].

Several research laboratories have explored the use of BCI systems as therapies for motor recovery of individuals affected by strokes [41, 47, 67, 68, 109, 110]. These BCI-driven devices provide real-time sensory-motor feedback during simultaneously attempted movement, or motor imagery, by coupling them with an output device capable of reproducing the same movement, such as an orthosis [47, 109], a robotic manipulator [41, 67, 110] or a functional electrical stimulation (FES) device [68].

Studies have shown that the size and location of the affected area of the brain can be very heterogeneous in patients that suffered strokes [59]. Thus, it is difficult to delineate how BCI systems promote neural plasticity. The study of how a BCI-based rehabilitation therapy reorganizes the brain can contribute to explain the brain mechanisms for motor improvement and, consequently, may facilitate restorative treatments for post-stroke recovery [90]. The studies carried out by [75, 111] proved the effectiveness of BCI-controlled devices. In [111], a 2 week-long BCI-based post-stroke rehabilitation therapy proved to be generally effective. However, the subjects had different degrees of improvement. A detailed analysis of the collected data indicates that the degree of improvement was directly associated with the level of activation in the damaged hemisphere, during the BCI therapy. Such results are compatible with the ones previously presented in [112, 113]. In [75], robotic and standard hand therapies for recovering stroke patients were compared. The achieved results have shown that those subjects using the robotic system recovered more effectively and with fewer injuries.

2.2. Anatomic analysis of the hand

The robotic devices for hand rehabilitation and assistance have a level of complexity and functionality similar to the human hand. For that reason, it is necessary to understand what elements in the human hand intervene in its motor functions and how they do it. This analysis contributes directly to the design of hand rehabilitation systems and avoids any potential damage that it could cause to the user's hand.

All the inter-person biological variations found in length of bones, branching of tendons and insertion of muscles [114], even when there is no consensus about the human hand dexterity definition, suggest that dexterity is a highly personal property, not only shaped by individual’s motor control ability, but also bonded to the owner's biomechanical characteristics. The unique biomechanics of the human hand, including complicated shapes of bones, varied rotational axes, and other biomechanical aspects, makes the hand's anatomy extremely complex. This complexity is related to the enormous range of applications the human hand has to cover. Therefore, the human hand represents the most differentiated and sophisticated musculoskeletal tool in human beings, demanding the largest capacity of the nervous system in relation to its size. Such applications include gestural communication, holding objects forcefully or with precision, and recording of sensory information. When performing grasping tasks, the fingers and the palm follow a specific pattern in order to hold or release an object. The palmar grab, or power grab, is one of the most important and widely-used types of grasp [115]. It is characterized by the four fingers and the thumb closing homogeneously around an object, providing a very versatile and firm grip for objects with different shapes. It can be used in situations requiring a strong and stable grasp, or when holding large objects. In [116], authors achieved this kind of grabby synchronizing the four medial finger movements, while stabilizing the thumb.

The 27 bones that compose the human hand can be seen in Figure 3 [117]. Eight of those bones are tightly packed in the wrist, each finger is composed of three phalanges and a metacarpal bone, excepting the thumb that only has two phalanges and a metacarpal bone. The thumb, unlike other fingers, is opposable, which means that it is the only finger on the human hand which can oppose or turn back against the other four fingers. This characteristic is crucial to the motor function of the hand, since it allows to grasp objects [118].

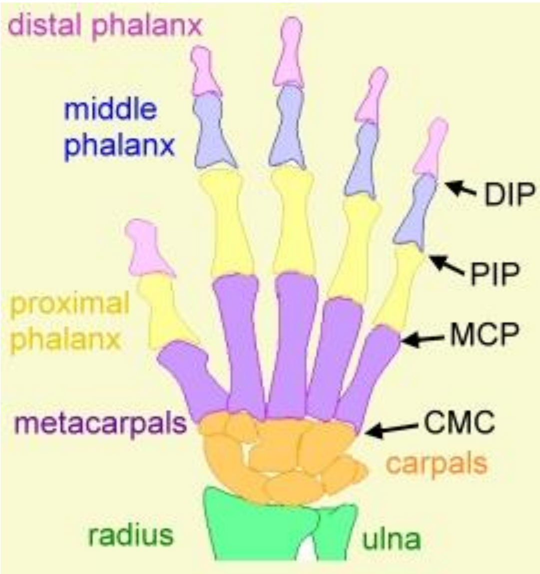


Figure 3 - The definition of the bones and joints of the human hand.

The connection between two adjacent bones is known as a joint and the bones' contacting surfaces determine the joint's possible motions. The finger's different set of motions, also known as range of motion (ROM), are determined by the different types of joints. The connection of the phalanges to the metacarpals form the metacarpophalangeal (MCP) joints. Besides the MCP joint, the thumb also has the so-called interphalangeal (IP) joint, located between the two thumb phalanges. On the contrary, in the four other fingers, the MCP joint is followed by the proximal interphalangeal (PIP) joint and, in the sequence, by the distal interphalangeal (DIP) joint. The finger joints work as mechanical hinges during the bending motion. Additionally, the MCP joints have an extra set of active ROM that allows the fingers abduction and adduction motions, which is the capacity to move from side to side. Also, the MCP joints also have one passive ROM that permits twisting motion around the axis of the finger phalanges. The thumb has a set of complicated movements resulting from the irregular contact surface between its corresponding carpal bone and metacarpal bones at the carpometacarpal (CMC) joint. This allows the thumb to have a wide ROM: up (adduction) and down (abduction), bent (flexion) and straightened (extension), and the ability to move across the palm (opposition) [118]. All of the human hand joints are also shown in Figure 3.

The ligaments are tough bands of fibrous tissues inserted on both sides of two adjacent bones that restrict the ROM at each finger joint. Two important branches, called collateral ligaments, can be found in all finger joints with variations in length and thickness. Their function is to stabilize the joint, shape the ROM, and prevent abnormal sideways bending of each joint. The volar plate is the thick ligament formed on the finger's palmar side. It also has insertions on both sides of the bones and prevents the occurrence of finger deformity from hyper-extension. Together with other accessory ligaments and soft tissues, collateral ligaments and volar plate form an important structure that is known as joint capsule [118].

There are two groups of tendons in the human hand between the bones and muscles. The extensor tendons straighten the fingers, while the flexor tendons bend the fingers. The tendons motions are originated from the corresponding muscle groups located in the forearm. In an electrical analogy, the muscles can be modeled as actuators, which deliver output contraction forces and the tendons of the hand as transmission wires, that smartly partitions the forces and smoothly delivers torques to each finger joint. The extensor tendons start from the wrist, branch out and have multiple insertion sites on the dorsal side of the finger bones. The flexor tendons on the palmar side pass through the carpal tunnel, travel through a series of pulley-like tendon sheaths grown onto the palmar side of the bones and eventually insert at the base of the DIP and PIP joints. The two tendon groups collaborative motions make possible a fluent hand movement. Besides being connected to the main muscle, some fingers tendons are cross-linked more closely to each other. The cross-linking degree varies per person, but the index, middle, and ring fingers are more prominently cross-linked [119]. This anatomical structure implies in a coupled actuation that avoids an independent movement of some fingers, making them move together to some degree.

The extrinsic muscles, which predominantly control the gripping motion, are large muscle groups directly connected to the central branch of the flexor and extensor tendons. Most of them originate from the elbow and have muscle bellies located in the forearm (Figure 4, adapted from [120]). There also are the intrinsic muscles that are composed of several small muscle groups slim enough to reside in the gap between two adjacent metacarpal bones. Most of these small muscles connect the wrist and the thin branches of the extensor tendons of each finger, near the MCP joint. These intrinsic muscles provide passive reflex-mediated stiffness at finger joints during various hand activities [118]. To close the hand, the flexor digitorum profundus (Figure 4 (a)) provides much of the necessary tension to flex the fingers. When opening the fist, the extensor digitorum, located on the dorsal side of the forearm (Figure 4 (b)), extends the fingers by pulling them back.

Most of the hand motions tasks require the contraction of strong muscles connected to the flexor tendons. This leaves to the extensor tendons the task of working as a breaking system that constantly regulates the torques exerted on finger joints. This breaking system functionality relies on the extensor hood, that is a region composed by a fibrous, thin, complex, and collagen-based web structure that directly wraps around the finger phalanges from the dorsal side and smartly transmits muscle forces to finger joints [118].

The tendon sheaths are fibrous tissues wrapped around the flexor tendons and have multiple insertions on the dorsal side of finger bones. They work as a series of elastic pulleys that enhance the efficiency of the transmission of flexion forces from the muscles to the joints, by flattening down when the finger straightens and bulging out when the finger bends [121].

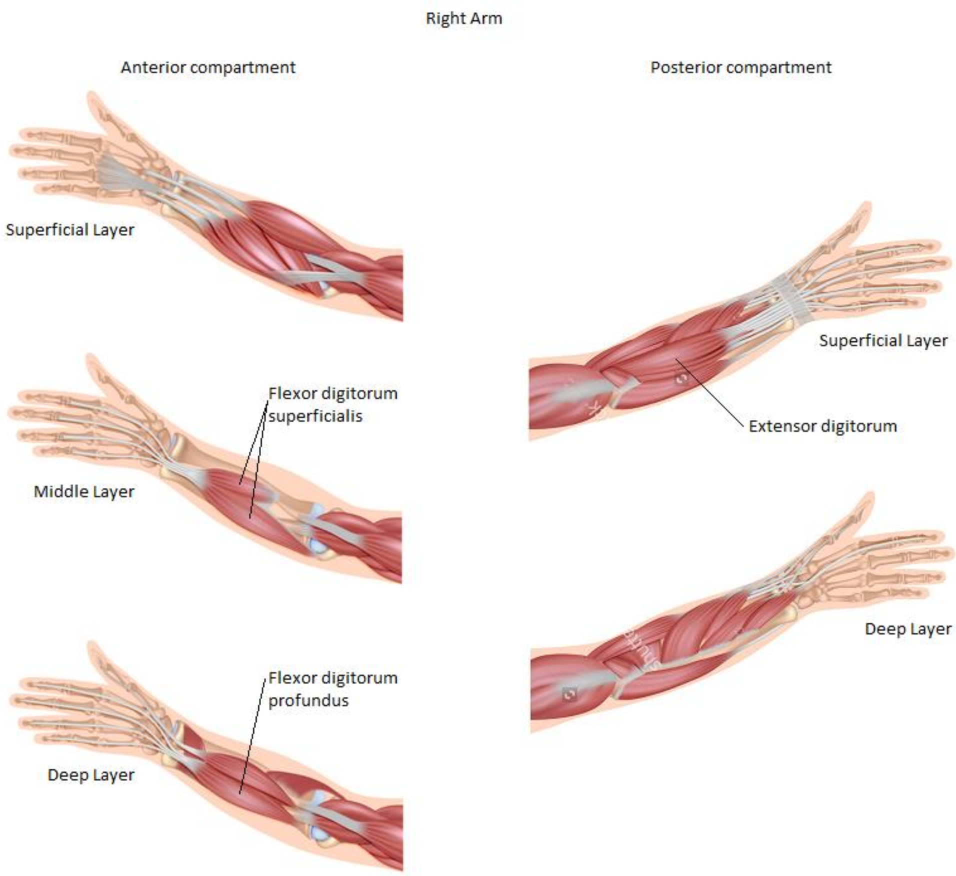


Figure 4 - Main muscles of the forearm. (a) Anterior compartment. (b) Posterior compartment.

The strength of the handgrip can be quantified by using a dynamometer to measure the static force that a hand can squeeze, as shown in Figure 5 [122]. The maximum forces generated by the hand during a grip depend on gender and age, ranging from an average of 200 to 400N [123]. Since hand grip strength is a measure of maximum hand voluntary force, it is the simplest method to evaluate muscle function [124]. Depending on the task and the force applied, the forces in the individual finger joints can vary. For example, 110N are enough to overcome spasticity and tone imbalance in a patient's hand [125], while 130 N are needed to open a hypertonic stroke hand [11]. All these values are critical for establishing limits and defining some of the characteristics of the rehabilitation devices.

Figure 5 - Jamar dynamometer for measuring handgrip strength.

2.3. Orthoses

The state of the art of rehabilitation devices for motor function recovery was meticulously analyzed, focusing on hand rehabilitation systems. It was noticed that many mechatronic devices were developed in recent years to assist the repetitive practice of hand exercises [23, 126–130]. It was also observed a substantial growth in the use of general robotic systems in physical assistance and rehabilitation therapies, due to recent developments of biomedical technologies based on robotic devices.

An example of such developments is the use of oscillatory EEG signals, recorded from the scalp, as inputs for BCI systems [56, 58, 131]. MI-based BCI is considered as one of the most effective ways to provide a control mechanism for people suffering from motor disabilities [53], using imagined movements as inputs for a BCI and then translating the MI-EEG into commands to an actuator. By analyzing the changes in the brain activity and their spatial distributions, it is possible to classify different imagery activities [132] or motor intentions [133]. A clear example is a robot controller that transforms mental commands (decoded from EEG signals) into appropriate motor actions (for example: left, right and forward), while the user imagines different limb movements [134, 135]. It must be pointed out that these BCI systems require a fair amount of computing power, since, in order to correctly detect and classify the user's intentions, the EEG signals must be preprocessed. The preprocessing of EEG signals seeks to separate the cerebral electrical activity of interest from the rest of the brain activity. Nowadays, many methods and combinations are used to achieve high classification rates combining different classifiers [136, 137] with different spatial filters [138, 139], different spectral filters [132, 140], and so on.

More particularly, [74] highlights a study developed by the author of the present dissertation aimed at evaluating and comparing different spatial filtering methods on EEG data, in order to improve the classification of Motor Imagery electrical brain information. Multiple spatial filtering methods, such as Laplacian filter, Common Average Reference method, Weighted Average Filter, Spatial Smoothing Filter, Common Spatial Patterns method and the Non-Homogeneous filter, have been analyzed. A Multilayer Perceptron Neural Network was used as a classifier and its input features are the EEG signals extracted with Wavelet Packet Decomposition. Best results were obtained using the Laplacian Filter, attaining a correct classification average increment of 10% over no spatially filtered data.

This technique has allowed patients in a late stage of amyotrophic lateral sclerosis (ALS), also known as locked-in syndrome, to interact with their environment. Such interaction has been done by using a simple binary output signal to select letters, symbols or words on a computer monitor [58, 141–143]. It was also possible to use BCI-based systems in applications related to: controlling a Functional Electrical Stimulation (FES) device for patients with spinal cord lesions [68, 144–146]; controlling a robotic device like a prosthesis [147–149]; or controlling a hand orthosis for stroke rehabilitation [47, 67, 150, 151]. An orthosis (plural: orthoses), according to the International Organization for Standardization (ISO), is "an externally applied device used to modify the structural and functional characteristics of the neuromuscular and skeletal system" [152].

2.3.1. Hand orthoses

Traditional robotic devices for hand rehabilitation still have some ergonomic and functional issues. They tend to be bulky and difficult to operate [79, 153]. These issues limit their dissemination to a wide range of users, especially when thinking about the personal use of these devices. Also, the majority of them assist only in finger extension movements. Most of them are powered by electric or pneumatic motors, which substantially increase the device weight due to the motors and their respective power supplies. Consequently, their weight becomes considerably higher than the weight of the human hand. These factors limit the usage of such systems in long duration or home-based rehabilitation programs, since stroke survivors commonly present proximal arm weakness. Besides, some of them make use of internal grasp structures that limit the range of motion and diminishes the possibility of use with real world objects [130, 154, 155]. There is a complete review about this theme in [156], but most powered external grasp devices limit the type of grasp and hand orientation in practical tasks.

Nowadays devices are usually based on large robotic machines or glove-like orthoses. The amount of feedback, interaction, assistance and complexity of these systems varies from design to design. Usually, the robotic machines have sensors and motors for feedback and assistance. However, their size and weight usually makes them non-portable devices. Very few devices are both actuated and portable.

Therefore, research groups currently designing robotic systems for hand rehabilitation are addressing this issue and focusing on the development of a more compact and ergonomic structure. Devices that resemble a glove are more intuitive to use and are more likely to be portable, which would benefit the patients, allowing them to fully explore their environment and the use of the device outside of the hospital. In spite of the fact that glove-like approaches are generally portable, it should be mentioned that the majority of the current designs tend to be unpowered orthoses, also known as passive, that only provide support and coordination.

The field of passive hand rehabilitation devices is relatively small, and passive devices usually compensate arm weight by using overhead pulley systems [157], passive exoskeleton rehabilitation devices [158, 159], or spring-based arm orthoses attached to wheelchairs [160, 161]. These aspects limit the usage of such devices in home-based programs.

A passive device consisting in a cable driven orthosis is presented in [162]. In order to assist finger extension with this device, the patient must use shoulder and elbow movements, which decrease its capacity to aid the patient in the reproduction of normal movements' kinematics in reach and grasp training. In

[21], this device was extended to an actuated device called the J-glove, which can be controlled by manual inputs, voice commands or electromyography. The cables are driven by motors and run through tension sensors. It only provides extension assistance, but the finger extension motion, using cable tension, could be utilized for finger flexion. However, the increased complexity and weight reduces the portability and limit its potential use at home-based programs.

The Hand Spring Operated Movement Enabler (HandSOME) is a passive lightweight hand rehabilitation device. It provides a large region of movement and allows even severely impaired patients to grasp large and small objects. A linkage between the finger and thumb is used as an actuating component to ensure proper inter-joint coordination in grasping movements. Unfortunately, this device does not offer flexion assistance, since most stroke survivors are able to flex the fingers voluntarily [163].

There are several commercial devices which provide finger extension assistance. However, these devices commonly provide only one control mechanism. The device called "HandTutor" is a $2000 USD glove that allows the users to play games during hand exercises, while it tracks the hand motions. This provides feedback to the patients and allows the improvement of their hand motor function [79]. A design like this, suited for use in ADL, allows the rehabilitation to be performed simultaneously with the patient's daily tasks. In turn, the SaeboFlex, Figure 6, is an unpowered wrist-hand-finger orthosis developed by Saebo Inc., which costs about $1690 USD. It is composed of adjustable springs that provide resistance and stability to the fingers during rehabilitation exercises. It is used in tone management therapy for patients that need to regain hand muscle tone [129]. However, this device cannot provide a functional grasp of diverse objects, as it was designed only for picking up objects with a diameter from 7.5 to 10 cm approximately, so smaller or bigger objects cannot be grasped. The range of motion is also limited, since the springs connected to the distal phalanx of each finger provide an increasing force with finger flexion, making difficult to reach and maintain full flexion [163]. Also, the movement of MCP joints is limited and DIP joints motion is occluded [22, 129]. Among this family of devices, the Hand Mentor Pro, from Motus Nova, should also be highlighted. It creates combined flexion/extension of the wrist and fingers, but restricts the movement of PIP and

DIP joints [164]. Finally, the Cybergrasp, from CyberGlove Systems, provides independent extension forces to each digit, but restricts arm movement [165].

Many approaches for the implementation of rehabilitation devices are based on the use of soft actuation methods. Some of them [153, 166, 167] employ the fluidic McKibben actuator [168, 169], which is a soft linear actuator made of a rubber tube surrounded by a braided shell, or similar types of pneumatic muscle-like actuators. Other approaches use soft pneumatic networks (PneuNets) embedded in elastomers [170–172], for bending or curling the material by air pressurization [173]. Generally, air is preferred over fluid due to its ease of storage, low weight, compressibility, low viscosity, environmentally benign nature and fast response times.

A pneumatic rehabilitation glove, proposed in [34], can achieve bending motions similar to the human fingers flexing motion. This behavior is achieved by integrating soft elastomeric actuators into a neoprene glove. Despite been a low weight device, it only assists the fingers flexion motion, since the soft actuators are located on the dorsum of the hand. To assist in finger extension motions, it would be necessary to use actuators on the palm of the hand. However, this would compromise the capacity of grasping real objects, since such soft actuators tend to be bulky.

Figure 6 - SaeboFlex by Saebo Inc., a spring-loaded orthotic device which holds the hand in an opened position.

Many exoskeletons for hand manipulation have been developed, but due to technical issues their development have been limited or they have not been allowed to be applied in rehabilitation [23, 126, 127]. For example, a device proposed by [174] use pneumatic cylinders to control wrist and metacarpophalangeal (MCP) joints flexion and extension, but it does not actuate on proximal interphalangeal (PIP) and distal interphalangeal (DIP) joints.

The Rutgers Master II-ND (RMII) is a device used in clinical trials with stroke survivors [128]. It uses custom-made pneumatic cylinders located on the palmar side of the hand to push the fingertips into extension. Therefore, it is not possible to grasp real objects when wearing the device [130]. Otherwise, the orthosis designed in [150] is a wearable exoskeleton that lets the palm free, by fixing the entire structure on the hand dorsum. This design keeps the palm functional, but adds an undesired extra weight on the distal side of the generally weak patient's arm.

However, many of these devices require bulky and expensive equipment [47], extensive training for days [109], or even weeks [68], do not offer BCI control during all of their operation states [47, 109], or need a previous selection of brain areas based on able-bodied anatomy in order to work properly [68, 109]. Besides, it is well-known that FES systems introduce electrical and neurological interference into EEGs, which increases the complexity of systems like [68, 116, 175], making them more difficult to be properly developed.

Recent works report the development of systems based on superficial electromyography (sEMG), that present a compact platform that provides assistance and resistance in repetitive motion therapy, like the one developed in [86]. However, this signal acquisition approach makes them unsuitable for patients with totally paralyzed arms. Some of these systems are limited by the latency they show between the onset of an attempted movement and the device mediated movement, that is generally caused by the signal processing procedures [7, 176, 177].

There are several devices being developed that have a good performance in the rehabilitation procedures, but have not achieved the required size and weight to be portable [125, 178]. Some of them require that the user carries a backpack [86].

2.4. Actuators

Passive orthoses only provide coordination and support. Besides, they cannot be directly controlled and, normally, require the use of arm weight compensation systems that are heavy and complex [157, 160, 161]. This work aims at surpassing these issues by focusing on the design of an actuated orthosis.

There are two main types of actuated orthoses: the mechanically actuated orthoses and the FES orthoses. The current work intends to contribute to the development of a hand rehabilitation device controlled by an EEG-based BCI. In this way, it was decided to focus on the design a mechanically actuated orthosis, because FES systems introduce electrical and neurological interference into the EEG. For that matter, there were considered many types of actuators for the design of the idealized orthosis, including: (1) rotary actuators, such as AC motors, DC brushed motors, DC brushless motors, servo motors and stepper motors; (2) linear actuators and (3) pneumatic soft actuators. These actuators could be used to directly control the motion of the system's joints or to regulate the mechanical tension over a cable system in order to provide a tensile force that can be used to control the fingers motion. The selection of the optimal actuators for the proposed prototype takes into account several aspects, such as: torque, size, weight, input voltage, capability to be controlled and price.

2.4.1. Rotary actuators

A rotary actuator is an electromechanical component that transforms electrical energy into mechanical energy, more precisely into rotary motion or torque. There are several types of rotary actuators, generally named motors.

2.4.1.1. AC motors

An AC Motor is an electric motor driven by alternating current. The alternating current that flows through the motor coils produces rotating magnetic fields that are used to generate mechanical energy. Two major components integrate an AC Motor: on the outside, there is a stationary stator with two permanent magnets, and on the inside, there is a rotor with coils supplied by AC current attached to the output shaft [179]. Figure 7 [180], shows an AC Motor diagram.

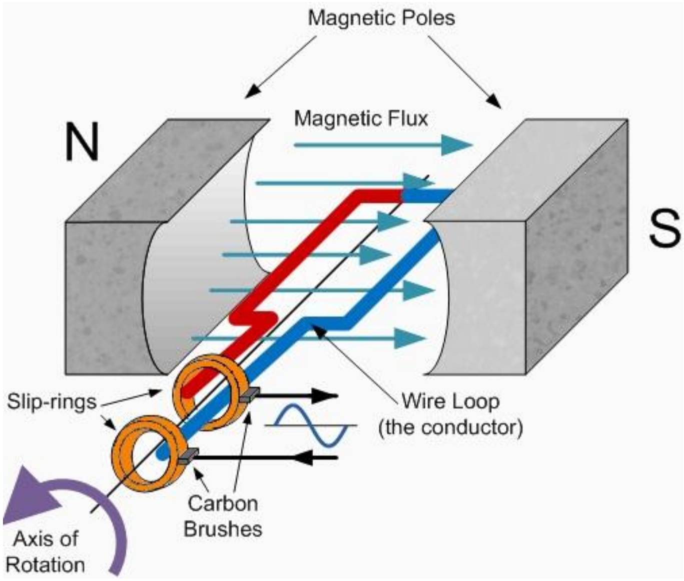


Figure 7 - AC motor diagram.

The fundamental operation of an AC Motor relies on principles of magnetism. The simple AC Motor contains a coil of wire and two fixed magnets surrounding a shaft. When an alternating electric current is applied to the coil, it becomes an electromagnet, generating a magnetic field. In turn, this magnetic field interacts with the field generated by the permanent magnets, disposed over the rotor, and that interaction induces a mechanical force capable of rotating the shaft [179], as seen in Figure 7.

2.4.1.2.
Brushed DC motors

A brushed DC motor is driven by a direct current that can be used to control its rotation speed with precision. Besides, they also have a particularly high torque to inertia ratio. Brushed DC motors are composed by six main components: the axle, armature/rotor, commutator, stator, magnets and brushes. Some of these elements are shown in Figure 8 [181]. The motor offers a stable and continuous current to a magnetic drive that controls the rotation of the motor's armature. These motors can vary the speed-torque ratio in almost any possible way [182].

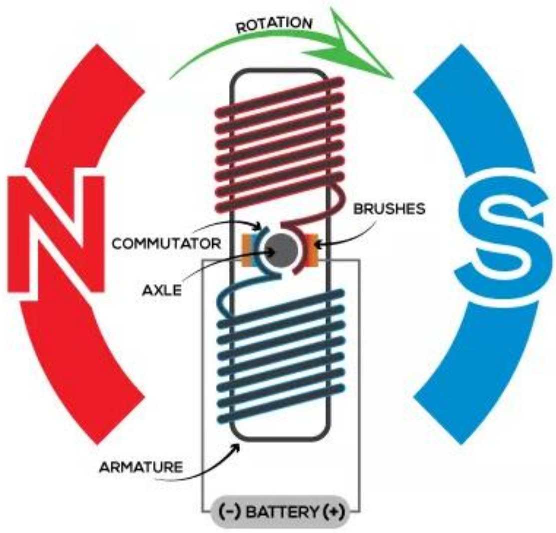


Figure 8 - Brushed DC motor's main components.

The component that gives the name to brushed DC motors is a device known as a carbon brush, which conducts current between stationary wires and moving parts. In order to work, the coils of the rotor must be connected to a closed loop electrical circuit. To that end, slip rings are attached to the shaft of the motor, and brushes are attached to the rings, in order to conduct the current [182].

A brushed DC motor consists of two magnets facing the same direction, which surround two coils of wire disposed in the middle of the motor, around the rotor. The coils are excited by a direct current that generates an associated magnetic field around the armature, capable of rotating it. When the armature becomes horizontally aligned, the torque becomes zero. At this point, the commutator reverses the direction of current flowing through the coil, reversing the magnetic field in order to let the rotational movement to continue. The commutator is a split-ring device used to reverse the current every time the coil moves through the plane perpendicular to the magnetic field. The electrical contacts to the rotating ring are called "brushes" [182].

2.4.1.3.
Brushless DC motors

A brushless DC motor is a synchronous electric motor powered by a direct current. It is composed by a rotating rotor, permanent magnets and a stator, as shown in Figure 9 [181]. Unlike brushed DC motors, these motors do not use carbon brushes. In a brushless DC motor, the electromagnets within the motor remain stationary along with the armature, while the encased permanent magnets rotate, generating torque. This type of motor is synchronous, meaning that the rotation of the shaft is synchronized with the frequency of the current used to power the electromagnets, avoiding any type of "slip", that most induction motors exhibit [183].

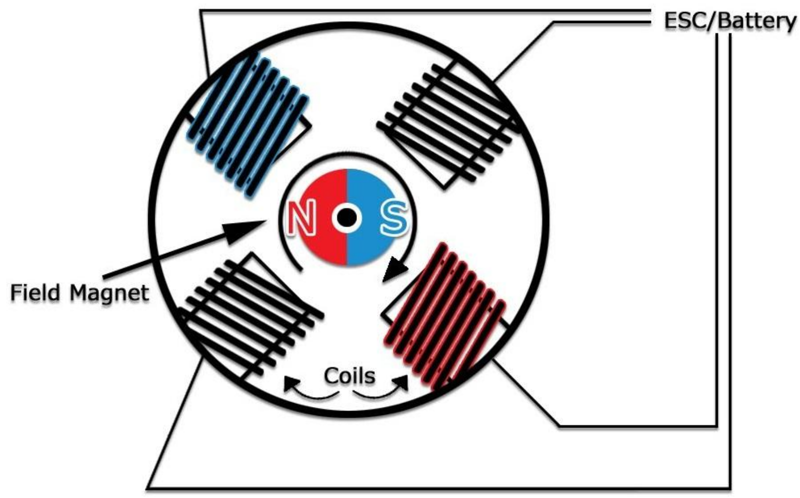


Figure 9 - Brushless DC motor's main components.

2.4.1.4. Servo motors

A servo motor is defined as an automatic device that corrects its motion using an error-correction routine. The term servo can be applied to any system that uses a feedback mechanism, such as an encoder or other feedback devices, to control the motion parameters [184].

A servo motor can be a DC, AC, or brushless DC motor, combined with a position sensor, usually, a digital encoder. Servo motors are frequently the option of choice in critical applications, due to their extremely high precision level in positioning tasks. Normally, a servo motor system presents a higher cost than a stepper motor system, due to the servo motor's feedback sensor and processing electronics [184].

A servo motor can be divided in three basic modules: a motor, a control board, and a potentiometer (variable resistor) connected to the output shaft. The motor is connected to a potentiometer and to an output shaft by a set of gears. A diagram of a servo motor system is shown in Figure 10 [185]. As the motor rotates, the resistance of the potentiometer changes and, consequently, can be used as a control parameter in order to define the rotation angle of the servo motor. When the control circuit detects that the correct position was achieved, it stops the motion of the servo motor. On the contrary, if the control circuit detects that the correct position was not achieved yet, it will keep rotating the motor in the correct direction until the reach of the desired position. The motor speed is proportional to the difference between its actual position and desired position [184].

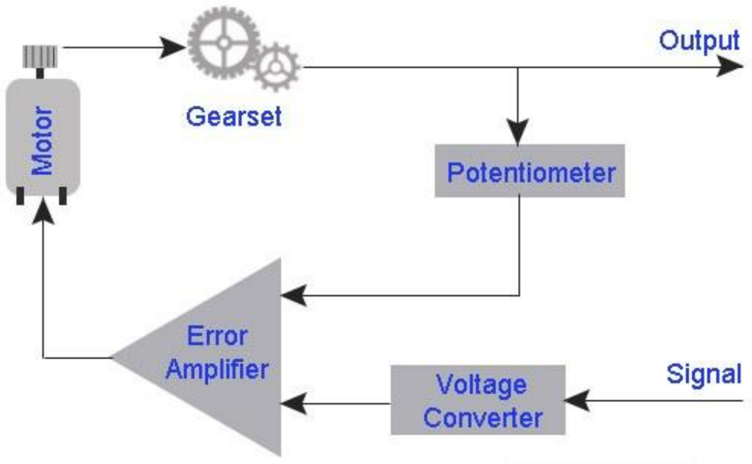


Figure 10 - Servo motor system.

2.4.1.5.
Stepper motors

A stepper motor is an electromechanical device that converts electrical pulses into mechanical movements. Stepper motors are driven by digital pulses rather than by continuously applied voltage or current. Unlike conventional electric motors, which rotate continuously, stepper motors rotate or step in fixed angular increments. Since this kind of motors lack feedback to maintain control of position, they are classified as open-loop systems [186].

The motion of stepper motors is controlled by a sequence of digital pulses that make the shaft rotates in discrete step increments. The pulses used to control the rotation of a stepper motor system can be produced by microprocessors, timing logic, a toggle switch, relays, etc. A diagram of the control block of a stepper motor is shown in Figure 11 [187]. A train of digital pulses is translated into shaft revolutions. Each revolution requires a given number of pulses and each pulse equals one rotary increment or step. With proper sequential pulses being delivered to the device, the rotation of the shaft will undergo a clockwise or counterclockwise rotation. Also, the speed of the motor shaft rotation is increased with the frequency of the input pulses [186].

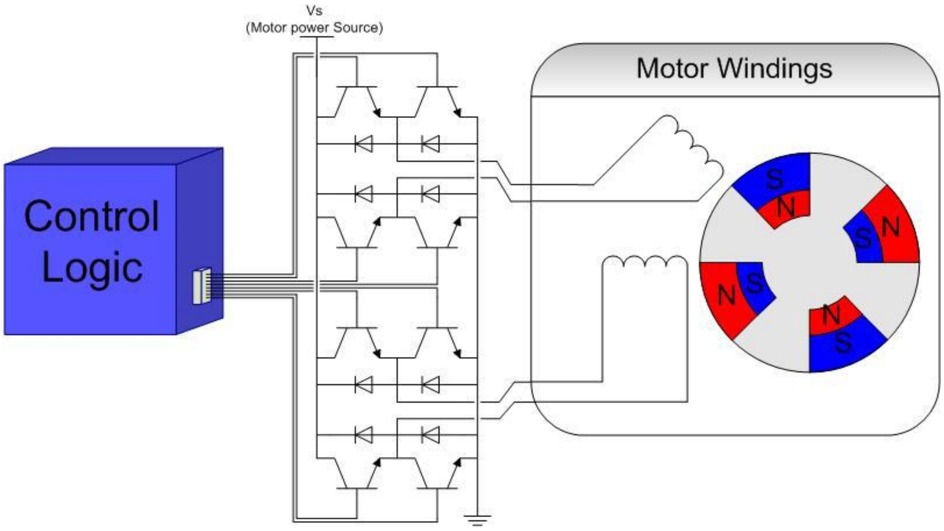


Figure 11 - Stepper motor control block diagram.

2.4.2.
Linear actuators

A linear actuator is a mechanical device that converts energy to create motion or to apply force in a straight line. The energy source can be extracted from electricity, air or fluids. In most cases, the basic principle of operation is inspired on an inclined plane (ramp). The threads of a lead screw act as a continuous ramp that allows a small rotational force to be used over a long distance to accomplish movement of a large load, over a short distance [188], Figure 12 [189].

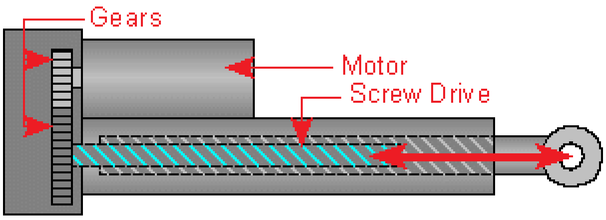


Figure 12 - Linear actuator diagram.

2.4.3.
Soft pneumatic actuators

The actuators that possess a bendable and flexible case rather than having a metal housing as their main structure are known as soft actuators. Usually these actuators are made of elastomers and use gases or fluids to perform their motions. Air is preferred over many other gases and fluids due to its ease of storage, low weight, compressibility, low viscosity, environmentally benign nature and fast response times. One type of soft pneumatic actuators is based on the PneuNet principle of operation [173]. The enclosed PneuNets are parallel and run along the entire length of the actuator. They combine an extension with a strain limiting layer along their length. As they are pressurized internally, part of them grows while their strain-limited portion does not. This causes the actuators to bend or even completely curl. The air flow between the PneuNets is achieved with an air channel located at the middle-bottom of the structure [34].

2.5.
Sensors and Transducers

In order to properly control the orthosis behavior, it is necessary to have access to some important feedback parameters about its position, speed, exerted torque, etc. This can be achieved by using a set of transducers to measure the desired physical variables, converting them into electrical signals that can be processed by the device's control unit. The control strategies implemented to control the motion performed by the actuators attached to a glove-like device require some critical information related to a set of physical variables. In that way, a good approach to the design of a glove-like device must include the use of sensors capable of measuring the deformation in the device areas corresponding to the hand joints. So, this section discusses the characteristics of some of these sensors.

2.5.1.
Strain Gauges

A strain gauge is a device with an electrical resistance that varies proportionally to the amount of induced stress. A metal strain gauge is a thin wire, commonly implemented by metal strips in a grid pattern. The grid pattern maximizes the number of strips deformed in one direction. The grid is printed on a thin substrate, the carrier, which is directly attached to the point of measurement. Therefore, deformations suffered in the neighborhood of this specific point will be directly transferred to the strain gauge, that will respond with a linear variation in its electrical resistance [190].

2.5.2. Flex Sensors

The Spectra Symbol Company developed a flex sensor capable of sensing a bend in any device under test, by changing its electrical resistance. It is available in different sizes and it can also be uni-directional or bi-directional. Despite the fabricant does not offer a detailed explanation on the functioning principle of these sensors, they have an operation mode similar to the strain gauges. However, these sensors offer several advantages over the strain gauges [191], for example:

- Strain gauges are generally uni-directional in operation while the flex sensors are not.
- Strain gauges need to be designed and located so that they are exposed to regions of maximal strain (to maximize response sensitivity or magnitude). The flex sensors do not have such restrictions.
- The flex sensors offer an easy way to measure resistance variation, while the actual magnitude of the signal in strain gauges is not large, requiring the use of amplification circuits.
- Strain-gauges saturate easily under large deflections, while the flex sensors do not have saturation issues.
- The flex sensors show dependence between path or local deflection profile and actual measurements. Strain gauges only depend on the magnitude of the local strain.
- The flex-sensor can measure large strains, while strain gauges do not support this kind of measuring.

2.5.3. Piezoelectric film sensors

Piezoelectric sensors are used to measure flexions, touches, vibrations and pressure. Whenever a structure moves, it experiences acceleration. In turn, piezoelectric sensors generate a voltage potential when physically accelerated [192].

Piezoelectric devices are based on minerals like tourmaline and quartz or synthetic materials, that can transform mechanical energy into an electrical output, generating a voltage proportional to the applied pressure (Greek: *piezo*). These devices can be used to detect single-pressure events or repetitive events, up to a certain frequency. Sensors based on the piezoelectric effect are able to sense transverse, longitudinal or shear forces, and they are not very sensitive to electric fields and electromagnetic radiation. Their behavior is also very linear over a wide range of temperature, making them ideal for applications in different weather conditions [192].

The physical design of piezoelectric sensors is very versatile. It can be manufactured in different shapes and sizes, depending on the application requirements. Piezoelectric film sensors can be used to measure the pressure exerted on a surface, while piezoelectric coaxial cable sensors are useful to measure the tension over a cable system, since their electrical output is proportional to the cable stretching deformation [192].

3.
Design and Development of the Proposed System

This chapter describes the main features of the proposed actuation system, that pursued two main goals: to create a device suitable for hand motor function rehabilitation and to avoid the limitations of current devices. Considering these goals, the final device must be:

- Safe,
- Durable,
- Configurable,
- Actuated,
- Fully controllable,
- Able to communicate with a BCI,
- Intuitive,
- Comfortable, and
- Affordable.

The features highlighted above were listed in order of importance, being the users' safety the most important aspect of the design. Safety was understood as the guarantee that the user's physical integrity must not be compromised, at any moment, by the device operation. The durability of the device takes into account that it must endure the realization of repetitive exercises for long periods of time. The device reconfigurability ensures that it provides different assistance levels to fit the widest possible range of patients' dexterity levels and capacities. An actuated device is required since the objective is to provide assistance to the user in the performance of basic hand movements and not only in support and coordination. A fully controllable configuration gives the users the certainty that they are able to control the device at any time, being able to stop a training session or motion in case of discomfort or weariness. The communication with a BCI makes possible a higher recovery rate through the synchronization of the patient's attempted movements and the orthosis actions. In turn, an intuitive and comfortable design encourages the use of the device and allows the patients to withstand frequent usage. Finally, the proposed design also aims at developing an affordable device, in order to reach a higher percentage of the people in need of rehabilitation or movement assistance.

The first step of the design process was to identify the main elements required for the proposed system to accomplish the project goals. The Control Unit was designed to control the whole system's behavior. It is connected to a group of actuators used to move a Mechanical Structure attached to the user's hand, using a Mechanical Subsystem. In this way, the system can assist hand movements. Besides, sensors are used to provide some sort of feedback to the Control Unit, returning relevant parameters measured from the Mechanical Subsystem and/or the Mechanical Structure, so they can be properly controlled. The Control Unit also has a User Interface that allows patients to interact with it and a BCI Interface for communicating with such devices. Figure 13 presents a block diagram showing the basic elements of the proposed system. The ones with a solid red contour are the principal elements of the system.

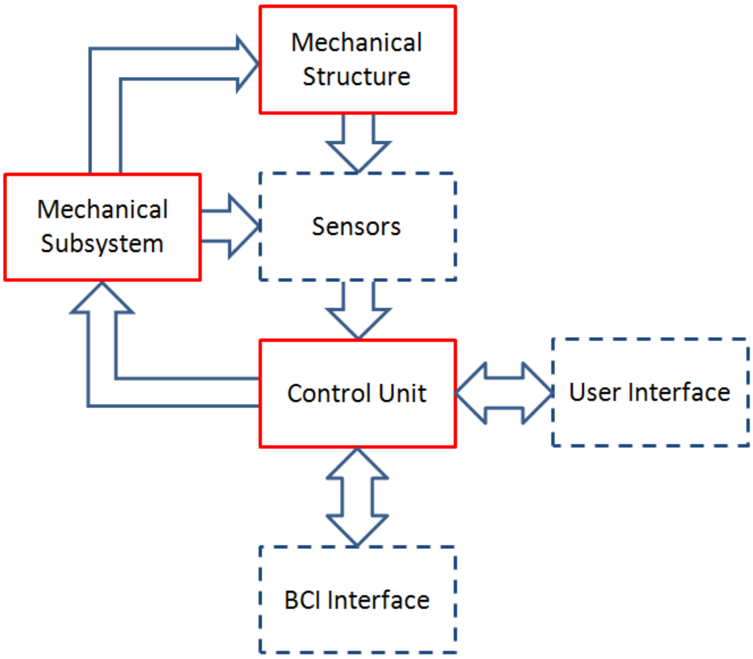


Figure 13 - Diagram of the interaction of the basic elements of the proposed actuation system.

Once the basic elements were defined, the conceptual design was stated as a wearable glove-like device, Figure 14. The glove itself is the Mechanical Structure, which will be actuated by motors, located on the upper arm or on the forearm, which will move the fingers through a cable system, which would act as the Mechanical Subsystem. The cables would be attached to the glove's fingers in a similar pattern to tendon insertions into the bones. This configuration was idealized to result in a lightweight device, especially on the distal end of the arm, which allows patients suffering from weakened arm to be able to use the orthosis comfortably. In addition, the glove design was meant to let the user's hand reach its maximum movement capacity, so the inner part of the glove is completely free of blocking elements. Also, a glove-like design gives the device a more friendly aspect, which would encourage the patients to use it.

The system is controlled by a versatile computer software, that was developed to act as the Control Unit and to communicate with the User Interface and the BCI Interface. For this very reason, the software offers two control modes. The first one is a user-friendly graphic interface that allows the configuration of the device according to the patient's specific needs, enabling the control of the device's motion by a set of buttons. In the second control mode, the device is connected to a BCI that sends motion commands according to the classification of the user's intended movements.

Due to time limitations, only the main electromechanical concepts and ideas were experimentally evaluated on the developed test bench. For that reason, the stages of materials selection for the glove and the cable system were postponed for future work.

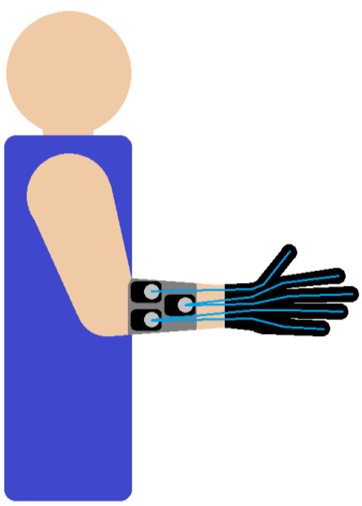


Figure 14 - Device's conceptual idea: the glove is connected by cables to servomotors in the forearm. The motors are controlled by computer software that receives commands from a BCI or a graphic user interface.

3.1. Mechanical Structure

The components preselected to integrate the proposed design needed to be tested and evaluated in order to determine which ones are better suited to the project requirements. However, in the first experimental tests, the device could not be directly connected to a user's hand, for safety matters. At the first development stages, there is a high probability of a malfunction or a design error, which could compromise the user's physical integrity. For that reason and the lack of a more complex hand mannequin, a simple model of the hand's joints and structures was developed and used as a mechanical structure for the project. In future works, this hand model will be replaced by the user's hand with a glove-like structure directly attached to it.

The hand model was designed to be lightweight, resilient and easy to use. These requirements discarded materials like metal or stone due to their weight. On the contrary, paper, paperboard, and cardboard are lightweight materials, but lack the durability required for the model. For these reasons, wood and plastic were chosen as the most suited and easy to find materials for the model.

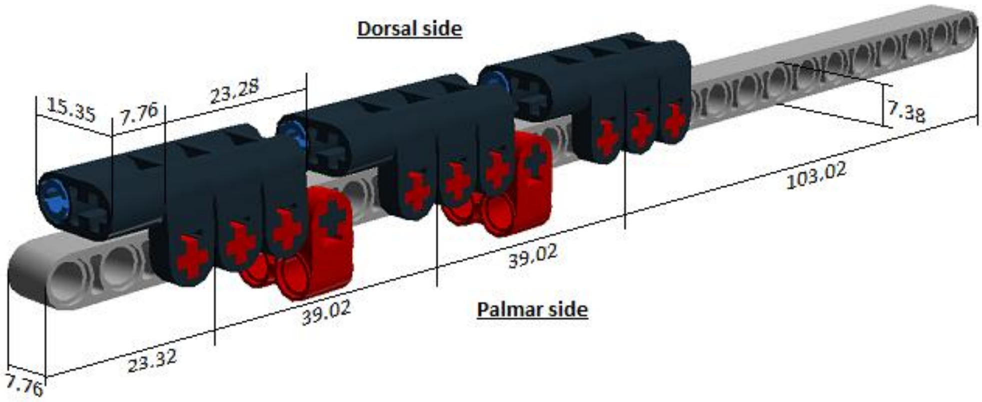
The use of LEGO kits seemed like a good approach for the creation of the model. The LEGO Technic line, or Expert Builder Technical LEGO line, includes several plastic rods and parts, shown in Figure 15, which can be assembled to create complex mechanical systems, like joints and cable guides.

Figure 15 - Some LEGO Technic rods and other parts.

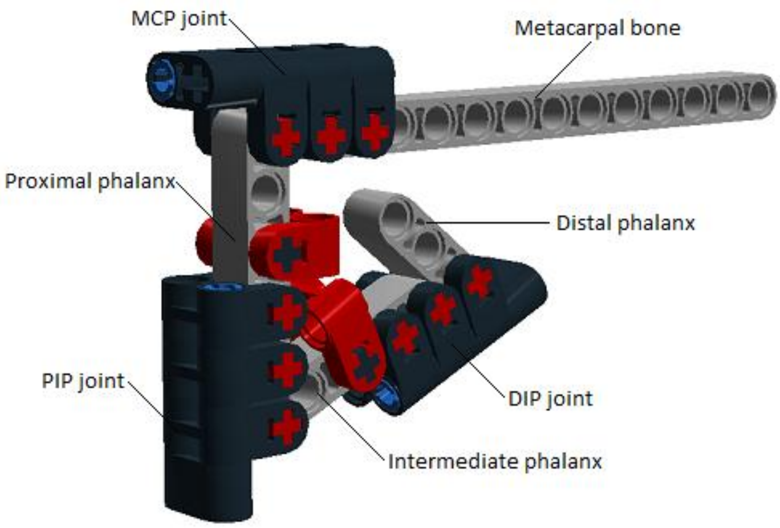
In that way, a hand model prototype was built with LEGO kits. The fingers were mounted as shown in Figure 16. All fingers had 3 rods corresponding to the phalanges and one for their metacarpal bone, except the thumb that had only 2 rods for the phalanges and one for its metacarpal bone. All joints (MCP, IP, PIP and DIP) were assembled in a configuration that mimics the ROM of such joints in a healthy hand from the flexed position, Figure 16 (a), to the extended position, Figure 16 (b). The dimensions of the plastic parts used in the hand model were selected to mimic the regular size of an adult’s hand.

The metacarpal and carpal areas were built as a single rigid element along with the forearm. This configuration is equivalent to the immobilization of the wrist joint, which simplifies the model and the actuation of the system. The adoption of this simplification does not interfere with the capability to evaluate the movement of the fingers by the electromechanical system. Also, this limitation will be eliminated in future works, since it is intended to implement the final glove design independent from the wrist, elbow and shoulder joints' positions. This will be done by using a Bowden cable system between the motors and the cable guides in the hand. A Bowden cable is a "spring steel wire enclosed in a spiral wire casing for transmitting longitudinal motion at a distance especially (as in a hand brake) around curves" [193].

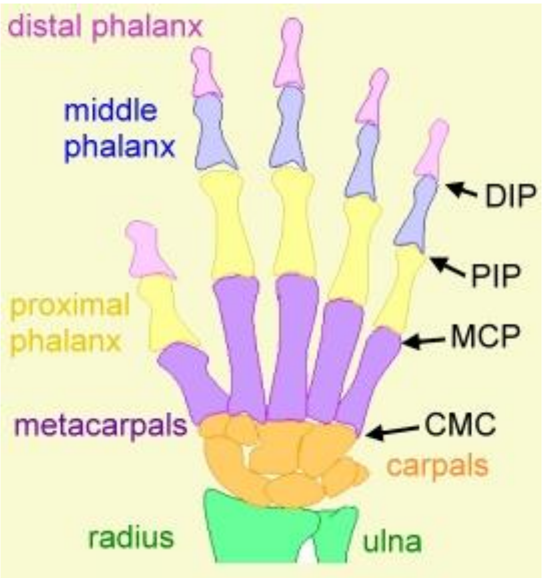
The forearm of the model is screwed to a wooden platform that holds the motors and acts as the model's base. The forearm was originally designed as a beam composed by 3 rods, but this configuration proved to be too unstable when the motors were actuating. In order to fix this problem, two additional lateral beams were mounted, from the model's wrist to the base, improving the stability of the test bench. Figure 17 shows the forearm-carpal-metacarpal structure.



(a)

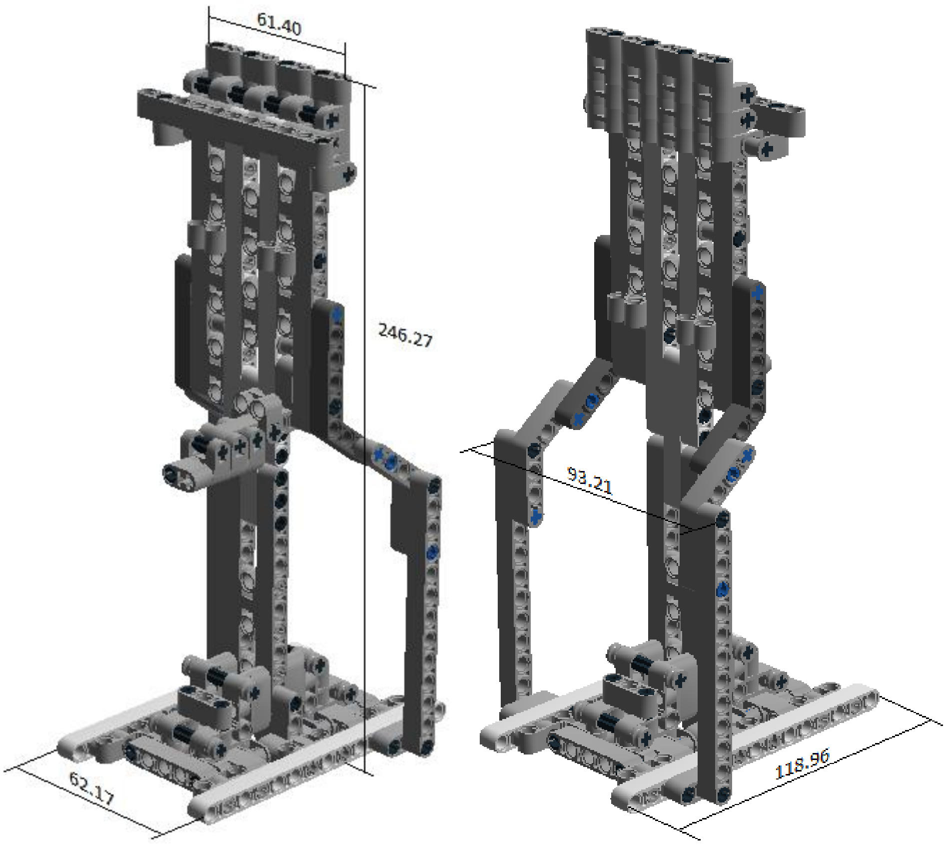


(b)



(c)

Figure 16 - Finger model: (a) Full extension and (b) Full flexion. Dimensions are in millimeters. (c) Hand bones and joints.



(a) (b)

Figure 17 - Forearm-carpal-metacarpal structure: (a) Palm and (b) Dorsum. Dimensions are in millimeters.

3.2. Mechanical Subsystem

The mechanical subsystem of the developed test bench includes: the hand model, the cables, and the actuators, as seen in Figure 18. This mechanical system implements a simple, but effective, representation of the mechanical behavior of the proposed robotic rehabilitation glove during its operation. The nylon cables are attached to each fingertip of the model, by a knot, and run through the cable guides from the fingers to the base of the model's forearm. The cables are wrapped around six pulleys attached to the shafts of six actuators, 3 for flexion and 3 for extension. One pair of actuators controls the thumb, while the other pairs control 2 fingers each. This configuration was selected over a pair of actuators for each single finger in order to reduce the size and weight of the final proposed device. It must be pointed out that grouping the fingers in pairs does not affect the rehabilitation, since the device will move all fingers homogeneously during its operation for assisting the user in palmar grasp movements. Also, this configuration resembles the cross-linked tendons present between some fingers. The radius size of the pulleys was selected taking into account the minimum force required to be exerted on the user’s hand. Besides, the cable is long enough to allow the full extension and flexion of each finger. The actuators are fixed to a wooden base, at 100 mm from the forearm structure.

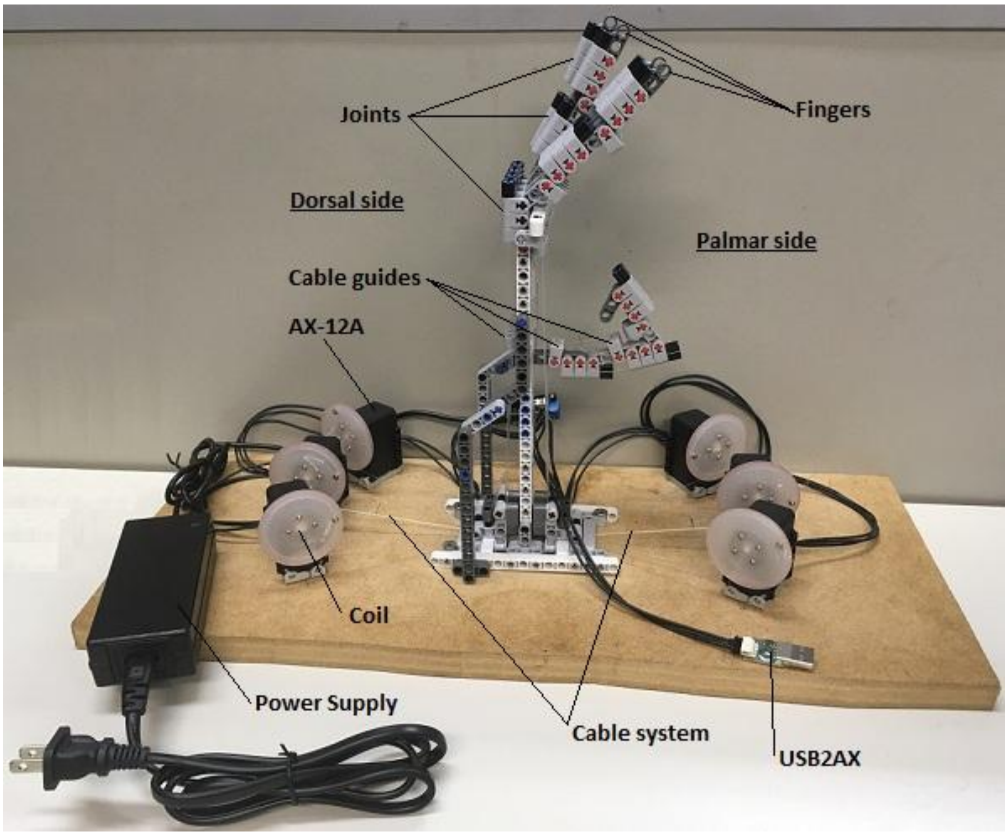
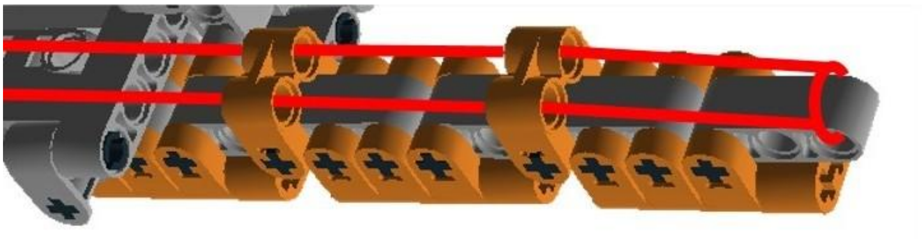


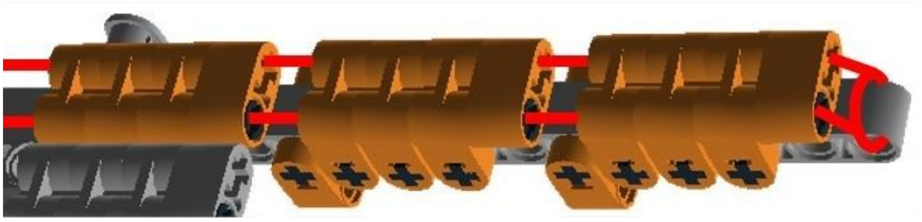
Figure 18 - Mechanical structure of the developed test bench: Hand model, cable guides, cables, pulleys and actuators.

3.2.1.
Evaluation of the conceptual idea

The first step in the development of the idealized test bench was to define an effective way to flex the fingers by running a cable along the fingers and the palm of the proposed hand model. To this end, 3 cable guides were placed on the palmar side: two of them on the center of the proximal and intermediate phalanges and the other one on the metacarpal bone, near the metacarpophalangeal joint. Figure 19 shows the configuration used to move each finger with the cable system, which was implemented by two cables running by the sides of the finger's palmar face with a separation around 7.9 mm between them. Both cables are pulled by the same actuator in order to prevent adduction or abduction of the fingers. This configuration leaves the internal part of the finger able to touch, providing sensory feedback to the patient, which may lead to greater recovery [108]. The experimental evaluation of the developed mechanical system had shown that it served its purpose, since it was possible to satisfactorily flex and extend all fingers by acting on the cable system.



(a)



(b)

Figure 19 - Cables, in red, running in pairs through the finger's cable guides, in yellow: (a) Palm cables and (b) Dorsum cables.

3.2.2.
Actuator selection

After analyzing the actuators detailed in section 2.4, a selection process was carried out to choose the most suitable actuator for the designed device. Each actuation method has its pros and cons, which will be highlighted in the next subsections.

3.2.2.1.
Soft pneumatic actuators

Actuating the glove with soft PneuNet based actuators would provide the advantage of a device motion that closely mimics the fingers movement. These actuators can be designed for different bending rates on each of their transversal sections, making possible the reproduction of joints movements [34]. The actuators disposed on each side of the hand would push the fingers, bending or straitening them. However, this solution would interfere with the ROM of the fingers, since the allocation of actuators on the palm of the hand would not allow the proper closing of the fingers and would make impossible to grasp objects while wearing the device.

3.2.2.2. Linear actuators

Linear actuators are suited for applications where it is intended to provide motion collinear to the motor’s shaft [188]. So, to drive the glove motion, the motor shafts of these actuators must be placed collinear to the motion required for applying tension to the cable system. This leads to a motor shaft's length proportional to the cable displacement required. However, focusing on the proposed glove-like system, this characteristic become an inconvenience, because it usually implies in motor’s shafts too long for their allocation on the patient's arm.

3.2.2.3. Rotary actuators

Rotary actuation can provide tensile force through a cable system attached to the fingers. The motors could reel in the cables pulling each digit. This solution overcomes the limitation of linear actuators, since the actuator's size is not related to its reeling capacity. There are several types of rotary actuators, with different features that were taken into consideration on the selection process. These aspects are briefly discussed in the next subsections.

3.2.2.3.1. AC motors

The AC motors considered for the device can reach more than 24 Nm of torque. However, as the name states, they require the use of AC power supplies, of either 110 V or 220 V [179]. These power supplies can be easily obtained from the electrical grid outlets, which are common in most buildings. However, this solution compromises the portability of the system. Also, with frames sizes over 45 mm, those AC motors are too big to be properly attached to a human arm.

3.2.2.3.2. Brushed DC motors

Brushed DC motors may provide a good solution for the glove actuation system. They can be found in a wide range of sizes, and the ones considered for the proposed device have less than 50 mm in their frames, and weights ranging from 20 g to 100 g. However, these characteristics are directly proportional to the torque exerted by the motor [182], being lesser or equal to 0.3 Nm for all the brushed DC motors considered. The torque levels required for the fingers movement can be reached by using gearboxes, without requiring too large motors. Furthermore, the operation control of these motors can be easily done by using an encoder.

3.2.2.3.3. Brushless DC motors

Brushless DC motors are another plausible solution for actuating the glove. The ones assessed for the proposed device are small, typically less than 30 mm of diameter, and lightweight, ranging from 40 g to 80 g. Their main inconvenience is that they normally provide many revolutions per minute (rpm) and low torque levels [183]. Then, to compose the designed glove-like system, it would be necessary to use gearboxes with high gear ratios in order to provide the required torque and shaft rotation speed. The friction in the system increases with the gear ratio, diminishing its efficiency. Besides, gearboxes with high gear ratios are also less resilient than the ones with low gear ratios, compromising the system durability.

3.2.2.3.4. Stepper motors

The stepper motors considered for the project have more than 200 g, which make them too heavy to be placed on a patient's arm. On the other side, the smaller motors had frame sizes of 28 mm, which is desirable, and a torque of 0.098 Nm, too small for the torque requirements of the project. Also, the greatest advantage of the stepper motors may become its greatest disadvantage, when considered as actuators for the glove. This type of motor always moves to fixed positions, that only depend on the amount and order of the steps they receive. This behavior provides an excellent solution from the control point of view [186]. However, this may not provide the operation smoothness required by the proposed design. If the steps are not small enough, the device may move the patient's hand too abruptly, causing discomfort and maybe injuring the patient.

3.2.2.3.5. Servo motors

Servo motors, based on DC motors, were considered as the best solution for the glove-like device. They provide torque levels that meet the device specifications and can be easily controlled by using their control boards [184]. The specific servo motor selected to be used as the orthosis actuator was the Dynamixel AX-12A, from ROBOTIS [194]. The system has 6 actuators: three for extension and three for flexion. The thumb was controlled by a pair of actuators, the index and middle fingers by another pair, and the ring and little fingers by the last pair.

3.2.3. Dynamixel AX-12Aactuator

Figure 20 shows the Dynamixel AX-12A actuator, a smart and modular actuator that incorporates, in a single package, a gearbox, a precision DC motor and a control circuitry with networking functionality. Despite its compact size, it can produce high torque and it is made with high-quality materials to provide the necessary strength and structural resilience to withstand large external forces. It also has the ability to detect and act upon internal conditions, such as changes in internal temperature or supply voltage. Summarizing, the Dynamixel AX-12A robot actuator shows better features than other similar products analyzed in this work [194].

Figure 20 - Dynamixel AX-12A actuator front and back sides.

These actuators are compact (see Figure 21 for dimensions) and lightweight (55 g). They are made with high-quality engineering plastic which enables it to handle high torque loads. A bearing is used at the final axis to ensure that high external loads do not degrade the motor efficiency. There is a Status LED on top of the back side of the actuator, which can indicate error status to the user. The motor body has a rectangular shape with several screw holes that allow for ease of mounting in different mechanical structures. On top of that, they are relatively cheap, costing $44.90 USD each motor.

From the electromechanical point of view, the actuator requires a DC power supply from 7 V to 12 V. It has a gear reduction ratio of 1:254 and it is capable of developing up to 1.62 Nm of torque. The highlighted characteristics made it fit into the requirements of the orthosis.

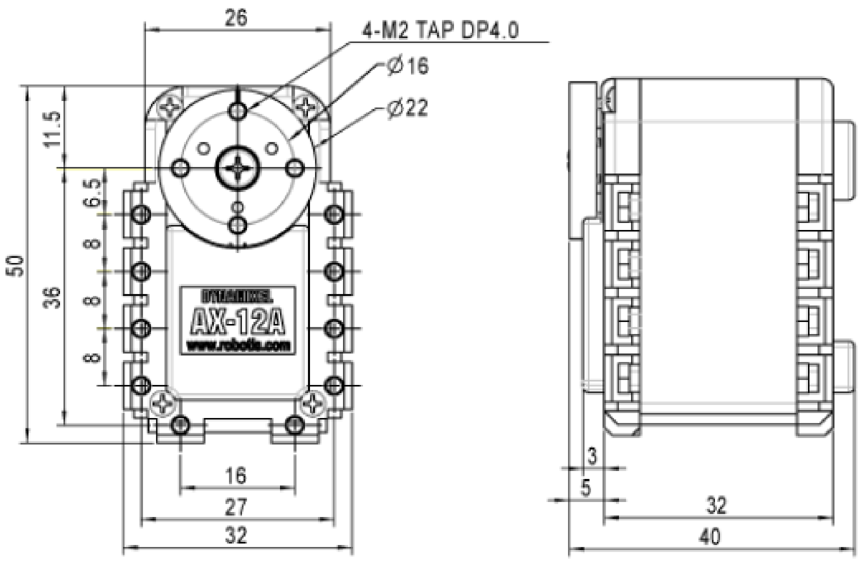


Figure 21 - Dynamixel AX-12A actuator's dimensions are in millimeters.

3.2.3.1.
Internal sensors and controls

The AX-12A possess a precise control over several of its features, being able to feedback the shaft position, its speed, the load torque exerted on the shaft, the motor internal temperature and the input voltage. Such feature satisfies the need for sensor elements expressed in the basic system diagram to provide the necessary feedback to the Control Subsystem. This motor has two modes of operation: wheel mode and joint mode. In the wheel operation mode, the motor runs continuously in the direction and speed established by the user. In the joint operation mode, it can only move along 300°, as depicted in Figure 22, but this mode allows to control the shaft's position. The angular velocity can be set to values between 0 rpm and 114 rpm. The user is able to set the goal position and the speed of operation, with a resolution of 1024 positions. The angular position resolution is 0.35°and the angular velocity resolution is 0.111 rpm, as given in the actuator's datasheet [195]. The torque, temperature and voltage feedback are used in an alarm system that alerts the user when any of them deviate from the imposed operation limits. Based on the measurement of these parameters, the motor is also capable to act automatically in order to prevent the occurrence of malfunctions. For example, it could disable the torque if the torque required to achieve a goal position surpass a pre-established limit. This behavior protects the motor from a permanent damage, which could occur, for example, if there is an obstacle preventing it from achieving the desired position. All these parameters can be set with a single command packet, enabling the main processor to control many Dynamixel units, using only one communication port and one timer.

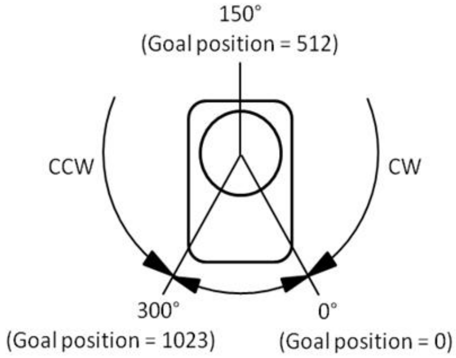


Figure 22 - Dynamixel AX-12A actuator's operation angles in joint mode.

3.2.3.2.
Communication interface

The Dynamixel AX-12A's communication interface is very easy and based on a daisy chain configuration (see Figure 23), that allows the sequential connection of several motors, resulting in a less messy connection and requiring only one USB port of a PC. The communication is made by using digital packets at speeds ranging from 7343 bps to 1 Mbps. Each packet carries an ID number corresponding to the servomotor the packet is addressed to. The ID numbers can go from 0 to 253, making possible the control of up to 254 actuators with a single daisy chain connection.

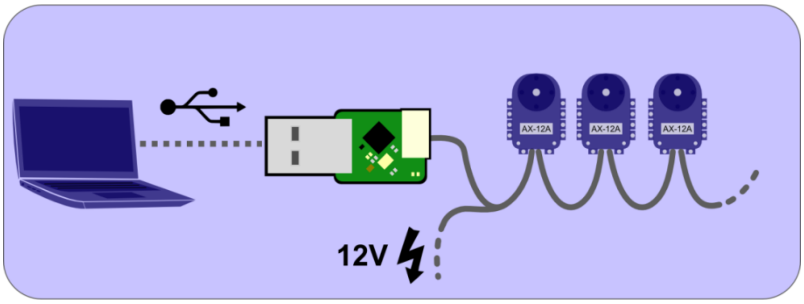


Figure 23 - Diagram of a daisy chain connection between several AX-12As and a PC through an USB2AX interface.

The interface between the motors and the USB port can be done with a USB2Dynamixel interface or a USB2AX interface. The USB2Dynamixel interface was developed by ROBOTIS and is able to directly control any of their Dynamixel servomotors. On the other hand, Xevelabs [196] developed the USB2AX interface, shown in Figure 24, which is able to control only the AX and MX Dynamixel servomotors series. The USB2AX interface is smaller and possesses an optimized latency, when compared to the USB2Dynamixel interface. It is also compatible with all the platforms and libraries that are able to interact with the USB2Dynamixel interface. Also, the USB2Dynamixel costs $49.90 USD, while the USB2AX costs only $39.95 USD

Figure 24 - USB2AX interface.

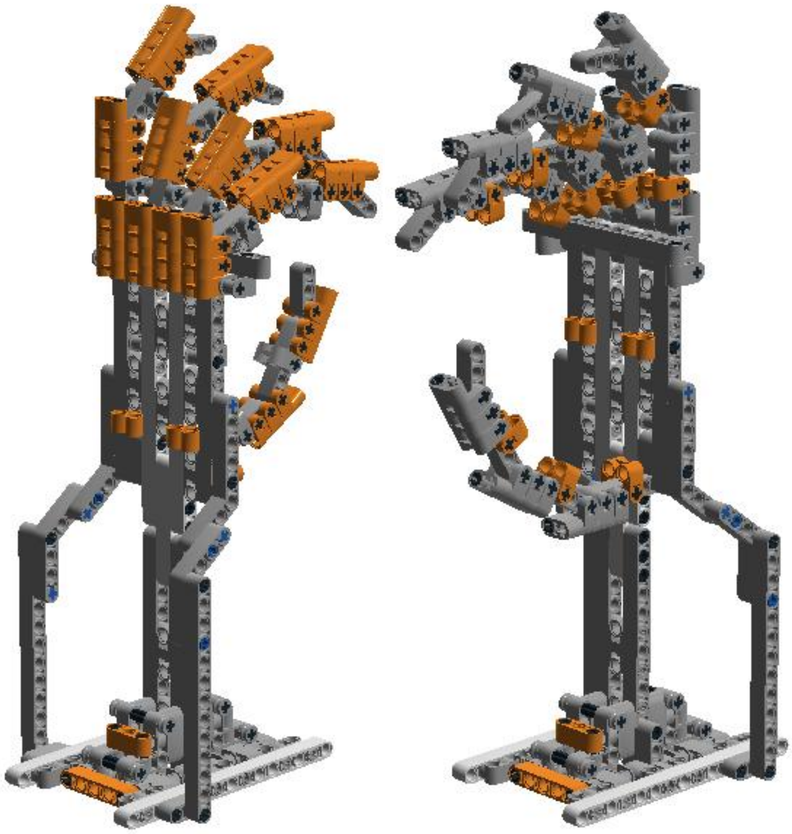
3.2.4. Cable system

A cable driven system is a soft robotics solution that is similar, in function, to natural tendons being pulled by muscles. The coupling of the cables with the actuators provides tension and movement to the whole system. Two sets of cables are used: one on the dorsal side, to assist finger extension, and one on the palmar side, to assist finger flexion. The mechanical structure used to mimic the glove behavior has cable guides, along with each finger long axis, to support the cable and create a linear path for actuation. The cables are fixed to the fingertips of the structure on both dorsal and palmar sides.

Several pieces, acting like cable guides, were placed along the dorsal and palmar sides of the fingers, thumb, metacarpus and base of the hand model, as presented in Figure 25. All of the cables have independent guides on the fingers, the thumb, and the metacarpus. On the palmar face, the guides are fixed at the midpoint of each phalanx distributing the forces along the finger. On the dorsal face, the structure of the IP and MCP joints allows to use them as cable guides as well, since they present a similar structure to the cable guides on the palmar face. At the base of the model, there are only two guides, one on each face, that act as concentration points before the cables go to their respective actuators. The cables run through the holes of the LEGO parts acting as cable guides, which provide a low friction and wide guide for the cables to pass through. The fingertips are the only points on the model where the cables are rigidly attached, to maximize the leverage on the distal phalanges. The extensor cables were reeled into the pulleys clockwise and the flexor cables, counterclockwise. This distribution ensures that all motors move in the same direction for a given command: clockwise for opening the hand and counterclockwise for closing the hand.

A strong but flexible cable was needed since the cables have to be able to bend along with the fingers and pull them to the desired position. The cables used on the test bench were made of nylon copolymer lines. Such lines have a 0.25 mm diameter and can be subjected to tensile strengths over 37 N, which made them a good choice for performing the evaluations. However, this material is not suitable for the final glove design, since nylon has a high elongation of about 30% to 100% depending on the specific type of material. This means that the line could stretch over time, becoming considerably longer than its initial size. The significant variability of its length over time makes this type of line not ideal for the proposed system. A possible solution could be the use of Kevlar cables, which may have elongations under 5%.

Once the whole cable system was installed on the model, some tests were performed in order to determine the cable displacement required to fully flex and fully extend the fingers of the model. The cable displacement is the same in all of the model's fingers since all fingers were built with the same length (101.36 mm). This simplification was adopted because the main focus of the model was to assess the performance of the mechanical systems. For that reason, specific details, like fingers' lengths, which can vary from user to user, were overlooked in the design process. However, this issue should be taken into account in the future design of the glove-like mechanical structure. The test consisted of measuring the displacement of the nylon lines from the fully extended to the fully flexed finger positions. During these tests, the flexor cable displacement was around 85mm, while the extensor cable displacement was only 39 mm. These values were taken as the maximum required cable displacements, since the developed hand model is larger than an average hand by more than 20 mm. From the tests results, it was concluded that, in order to reach the full range of motion, the flexor actuators have to reel in more cable, when flexing the hand, than the extensor actuators, when extending it; while having a maximum rotation arc of 300°.



(a) (b)

Figure 25 - Cable guides highlighted in yellow: (a) Dorsal guides and (b) Palmar guides.

3.2.5. Pulleys

The nylon lines for the flexion and extension of the same finger require different amounts of displacement. There are several configurations that can provide different amounts of cable length reel in for the flexor and extensor lines. One possible solution is the coupling of two pulleys with different sizes to the same actuator, with the cables winded in opposing directions. This allows the use of a single actuator for each finger, instead of two, but requires the creation of one pair of pulleys with specific dimensions, for each finger of each user. Instead, focusing on generality, the proposed solution is that different actuators manage the movement of the flexor and extensor cables, with pulleys large enough to provide sufficient cable displacement for the majority of the users' hands dimensions. This configuration, however, requires the actuators to be synchronized in motion, which is not a problem since the operation of AX-12A actuators can be synchronized.

The minimum radius of the pulleys was determined by the maximum cable displacement (85 mm), highlighted in the previous section. The pulleys radius calculation was done following the arc length equation (1),

$$s = r\alpha, \tag{1}$$

where s is the displacement of the line, r is the radius of the pulley, and α is the maximum angular displacement of the actuators, which is 300° or, equivalently, 5.24 rad. The minimum pulley radius required to fully cover the cable displacement of the model is16.3 mm.

Besides the cable displacement, there is another constraint that should be taken into account in the definition of the radius of the pulleys. Since the proposed rehabilitation device must provide assistance and resistance therapies, it is necessary that it could achieve the forces exerted by the fingers of a healthy person. In turn, for a given motor torque, the force decreases with the pulley radius, according to

$$\|F\| = \frac{\tau}{\|r\| \sin \theta}, \tag{2}$$

where$\|F\|$ is the force exerted,τ is the torque, $\|r\|$ is the absolute value of the radius of the pulley, and θ is the angle between the radius and the force (90° in this case).

The mean force exerted by an index finger, when fully extended and trying to press an object, perpendicularly to its long axis, is 35.20 N. Also, an index finger bent 90° is capable of exerting a pulling force over 49.33 N on average [197]. Taking the maximum of those values as the minimum required force to move a finger and knowing that the torque exerted by the AX-12A is 1.62 Nm, the maximum possible radius of the pulleys is 32.8 mm.

Since the AX-12A actuators only revolve 300°, the lines were attached to the pulleys at the middle point of such displacement (150^{o}). This configuration allows the cables to be tensioned at all times while reeled into the pulleys and also that they come out of the pulleys tangent allowing a maximum force transfer from the actuators to the cables.

Taking into account all of the considerations highlighted in this section, it was decided to make six pulleys, one for each motor. The raw material for the pulleys was nylon cylinder with 51 mm of diameter. Using a lathe, 6 wheels were made with 20 mm radius, providing an arc length of 104.72 mm to the 300° that the AX-12A motors can revolve. This arc length can supply the required amount of cable displacement (85 mm) without reaching the motor limit positions. With a 20 mm radius and the cable leaving the pulley tangentially, the maximum force exerted by the AX-12A motor with these pulleys is 81 N, which is more than enough to meet the system requirements. The wheels present a 5 mm band guard that prevents the cable from slipping out of the pulleys during actuation. The fixation point between the cables and the pulleys was a hole on the side of the wheel, where the lines pass through and are secured with a bolt and a nut. Figure 26 shows the dimensions of the wheels. The pulleys have 4 holes around the center of the circumference for coupling them to the motors' shafts.

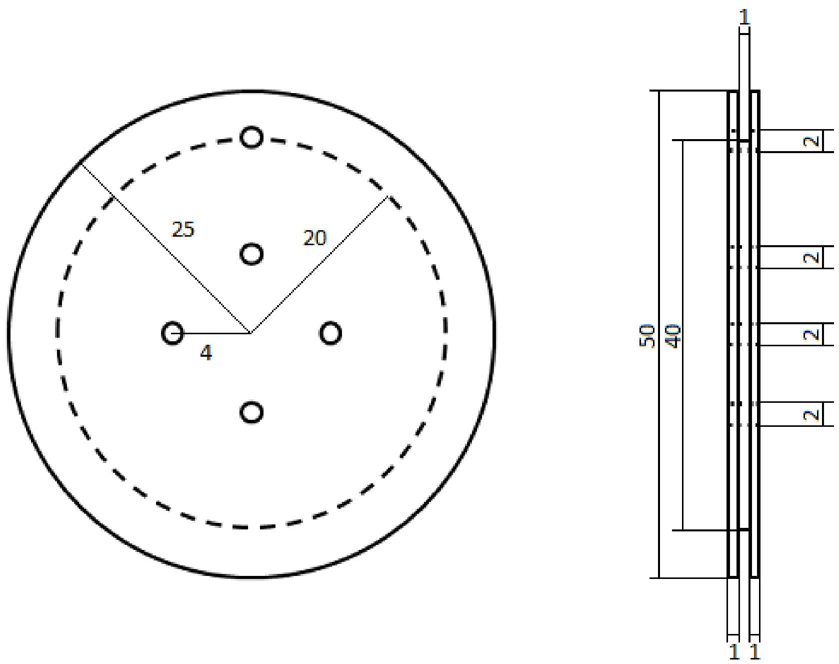


Figure 26 - Pulleys dimensions in millimeters.

3.3. Control Unit

3.3.1. Microprocessor

The initial idea for the project was to use a microcontroller as the control unit of the BCI driven orthosis, providing more portability to the solution. Unfortunately, the signal processing and classifying of the BCI requires the use of a computer as processing unit. Due to this requirement and to the existence of libraries in LabVIEW and Matlab for controlling the Dynamixel actuators, the final control design for the orthosis was based on a PC.

3.3.2. Software

One of the many features of the drivers for the Dynamixel actuators is the possibility to control them using libraries created for popular programming tools, like Visual C++, Visual Basic, Visual C#, Matlab and LabVIEW. LabVIEW was chosen to develop the orthosis control software due to its powerful Dynamixel SDK library and its capacity to create graphic user interfaces with ease. Dynamixel SDK is a standard programming library to develop software for controlling Dynamixel actuators. It is highly portable to several platforms since it is written in C language.

Figure 27 shows the control software functional diagram. When the system starts all the actuators are driven to their default positions, in which all the cables are loosed, and all of their parameters are reset to their default values. After the initialization process, the calibration stage begins. In this stage, the system is adapted to the user's motion capacity to ensure full grasping and extension of the hand. For doing this, the user must be asked, or helped, to reach the fully extended fingers' position. Then, by using the dials on the graphical interface shown in Figure 28, each actuator is moved to a position where all cables are tensed. When all motors are in the right position, the "Minimum (Opened)" button is pushed. This ensures that the individual motor positions are recorded as the minimum values each motor can achieve when opening the hand. A similar procedure is performed for the flexed fingers' position. A calibration procedure can be seen in

[198]. Once the orthosis is calibrated, the "OK" button must be pressed to continue to the Operation Stage.

On the beginning of the Operation Stage, the system calculates the operation ranges from all the motors by subtracting each minimum position value from the maximum position. To achieve a synchronized movement of all the fingers, each range is divided into a fixed number of steps. All the motors make a step when certain conditions, which depend on the support level and direction, are achieved. This ensures that all fingers move together, guaranteeing a natural hand motion. Since different fingers may have different ROM and even for the same finger the dorsal and palmar motors have different operation ranges, the steps' size has to be calculated for each individual range. Each range is divided by the number200, which provides200 steps with a different size for each finger. This amount of steps was obtained empirically and it provided a step size small enough to provide a smooth transition from step to step and large enough to allow a good operation speed for the device. When all these calculations are done, the system checks which operation mode is selected by the respective switch's position on the interface.

If the BCI is selected as the operation mode, the system establishes communication with a BCI and waits for a command. The command could be to open or close the hand. When a command is received, the system supports the user in the performance of the motion. On the contrary, if the Button operation mode is selected, the system waits for the operator to press one of the buttons on the interface (Open or Close) and then executes the associated action.

Whatever the chosen mode is, the orthosis movements are performed according to the type of support and support level selected on the interface. The support could be assistive or resistive, and the support level goes from 0 to 1023. For example, in the assistive support mode, the palmar motors help to close the hand, but, in the resistive support mode, these same motors oppose the opening hand movement. On the other side, the dorsal motors help opening the hand, while operating in the assistive support mode, and are the ones that offer resistance during closing hand movements, while operating in the resistive support mode.

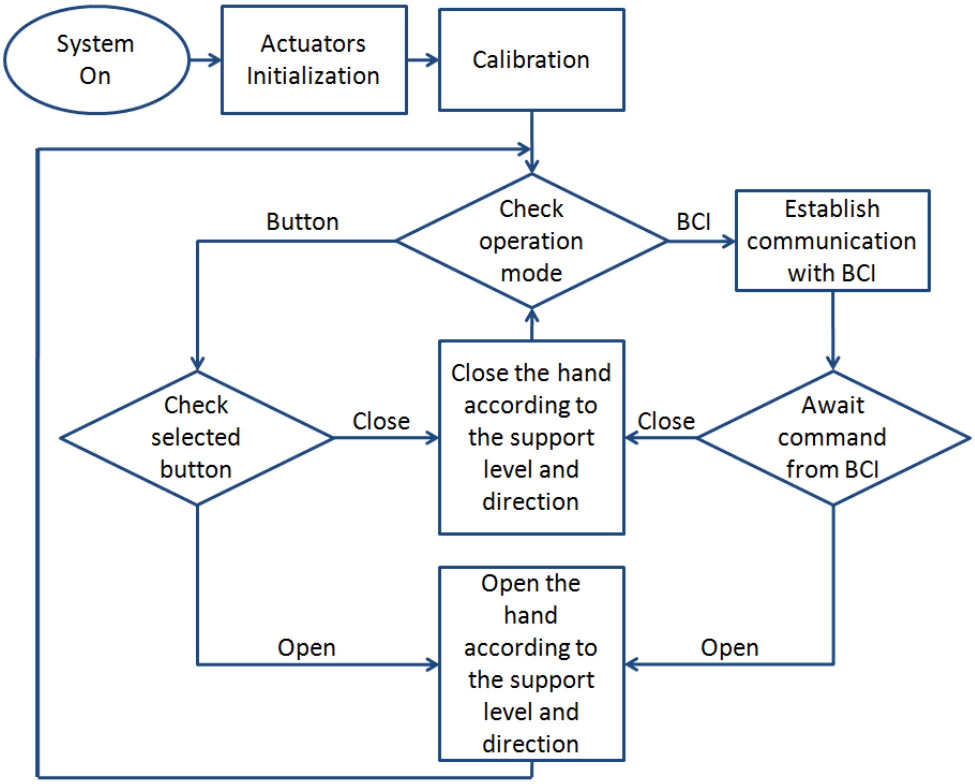


Figure 27 - Functional diagram of the computer software developed to control the system.

If the assistive support is selected, the orthosis will help the user to perform the chosen movement, giving a greater assistance for higher support levels. This is done by checking the load readings, which vary from 0 to 1023, from the motors performing the hand motion. For example, when closing the hand, the system checks the load readings of the palmar motors, a high load means that the patient's hand is lingering (it is more opened that it should be). If the load values are higher than the support level, it means that the patient is doing less effort than the required for that level. When this happens, the motors stop and wait for the load value to go under the support level, which means that the patient achieved the desired hand position by himself; then the motors move to the next step position, where they again check the load readings. On full support level, the system will never get a load reading higher than the support level and will complete the intended movement even if the patient is not making any effort. This type of support is intended for augmentation in patients with limited or no mobility on their hands. In [199] it is shown the motion of the model when on assistive support and a full support level.

For the resistive support, the orthosis resists to the movement performed by the user, giving a higher resistance for higher support values. This can be done by checking the load values from the motors that are not performing the hand motion. For example, when the close hand command is received and the resistive support is selected, the system checks the loads perceived by the dorsal motors. These load values are proportional to the force exerted by the patient, on the device, when closing the hand. If the load values are lower than the support level it means that the patient is not applying enough force to perform the movement for the selected support level. When this happens, the motors stop and only make a step if the load values exceed the support level, meaning that the patient is applying enough force in the performance of the movement. This last type of support is aimed at patients trying to gain strength in their hands.

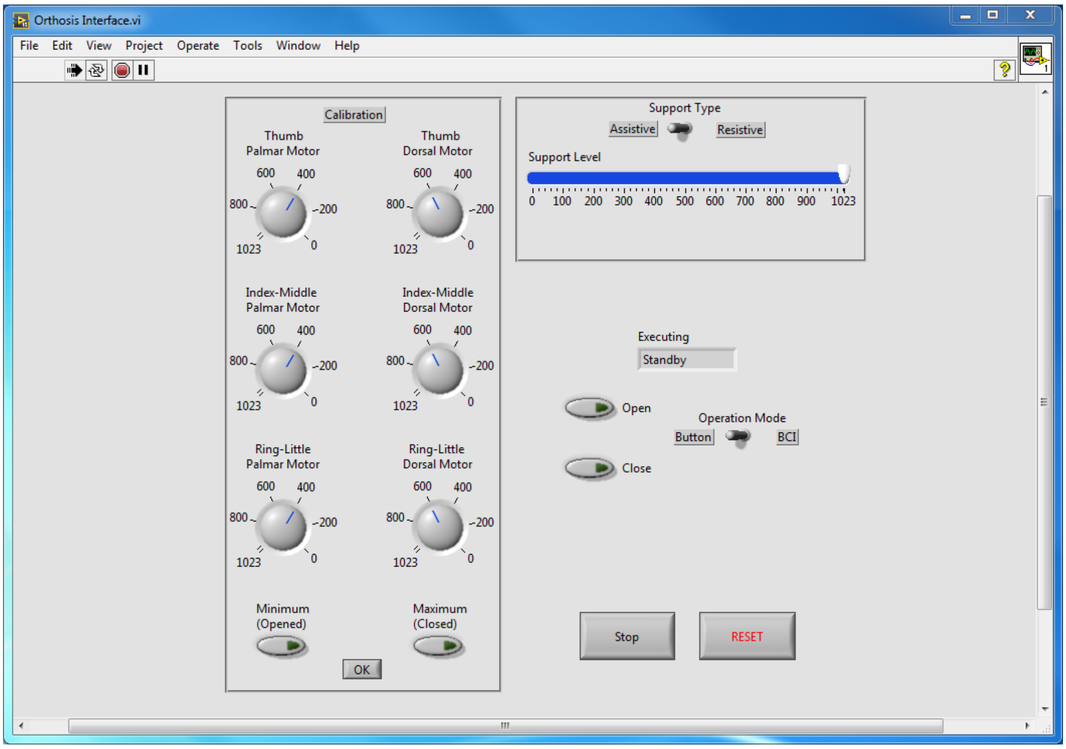


Figure 28 - Graphic User Interface for controlling the orthosis.

There are two more buttons on the interface, the Stop and Reset buttons, that can be activated during both operation modes. These buttons are capable of stopping the current movement and resetting the whole system parameters, respectively. In this way, a training session can be paused or stopped at any time. Besides, they can be used as safety measures to protect the hyperextension or hyperflexion of the user's fingers.

In order to behave as described, the control software was implemented following a linear sequence of events, which are divided into three main stages: Initialization, Configuration and Operation. The program begins with the Initialization Stage, which is divided in 4 consecutive sections, as shown in Figure 29. The first section establishes a connection with the actuators and initializes some variables like the minimum and maximum limit positions and the step sizes for all actuators. The second section makes invisible all elements that cannot be altered directly by the user. In section three, all motors receive the speed of operation and their respective default minimum and maximum limit positions. Finally, the last section sets all motors positions to their default values.

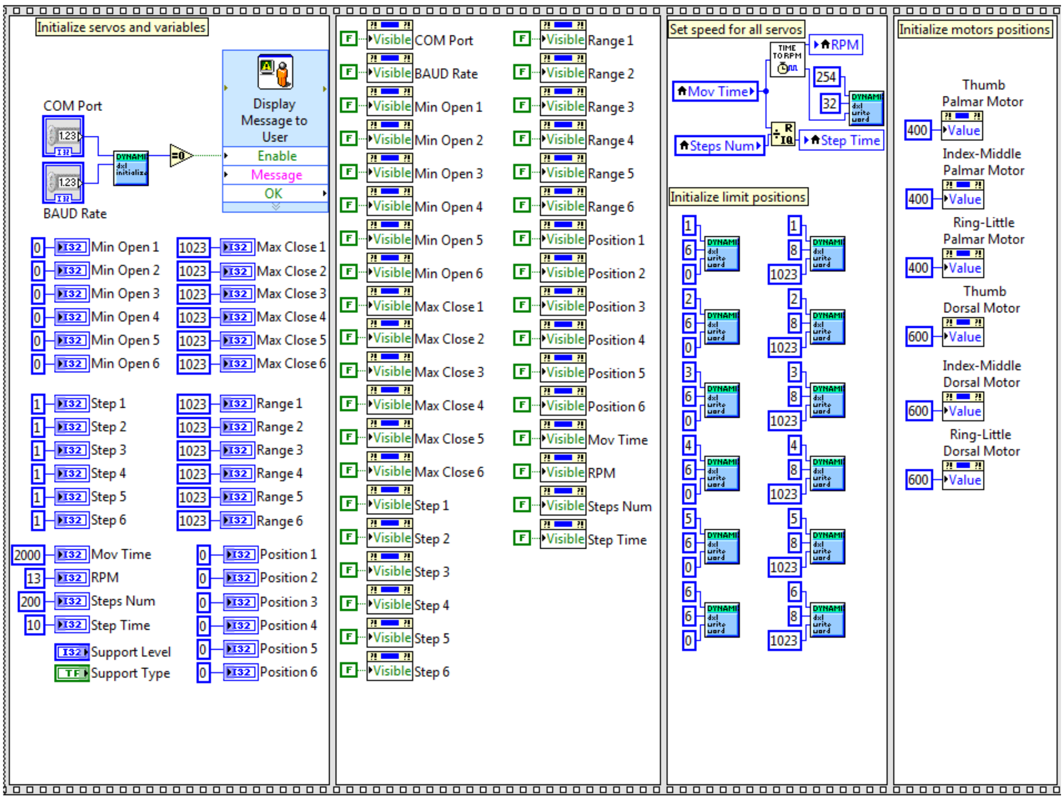


Figure 29 - Initialization Stage.

After the Initialization Stage comes the Configuration Stage, highlighted in Figure 30. This stage is divided into 3 consecutive sections. In the first section the motor positions are controlled by their respective dials and the limit positions of each actuator are recorded, by pushing the "Minimum (Opened)" and "Maximum (Closed)" buttons. After that, the limit positions are sent to the actuators and, in the third section, the ranges of operation and step sizes are calculated.

The final stage is the Operation Stage, presented in Figure 31, which contains a loop that can only be interrupted by pressing the "Reset" button. At the beginning of each cycle, the software checks which operation mode is selected, as described in Figure 32. If the Button Mode is selected, it creates a string containing the action to be performed according to the pressed buttons, Figure 32 (a). If the BCI Mode is selected, the software executes a Matlab script, Figure 32 (b), which executes the BCI processing and returns a string with the action to be performed, according to the classification of the BCI.

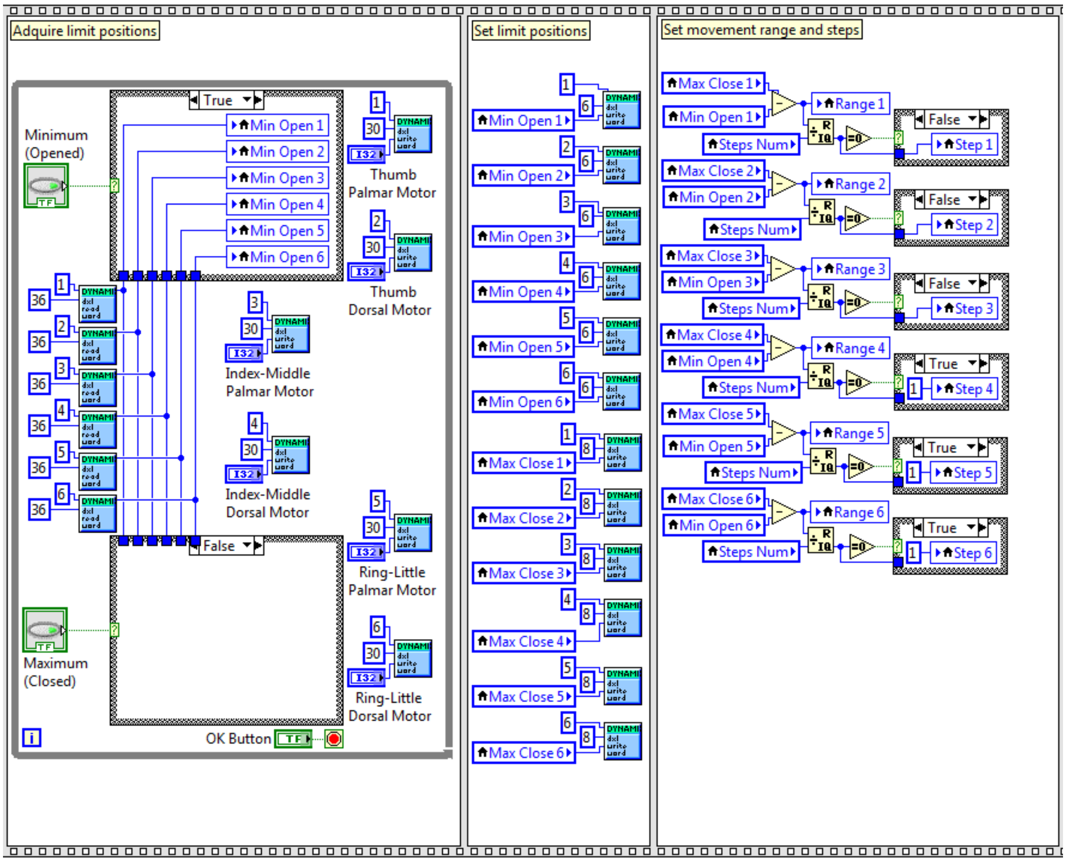


Figure 30 - Configuration Stage.

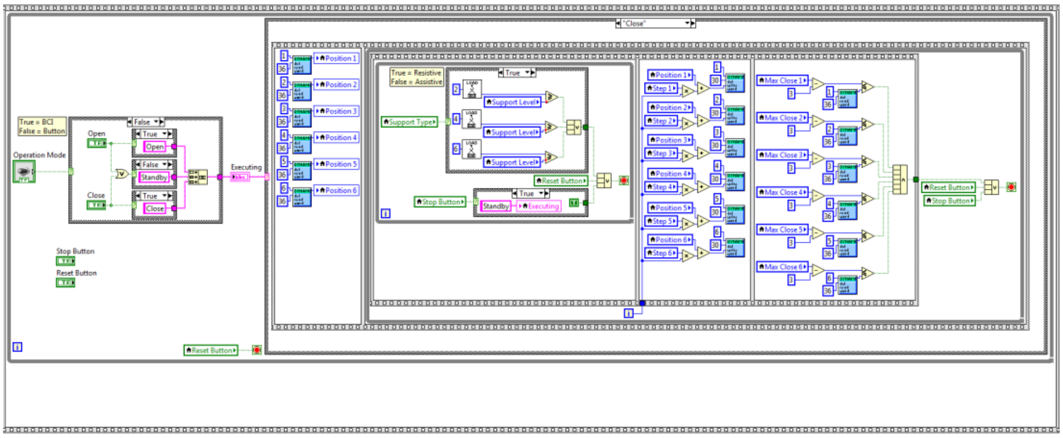
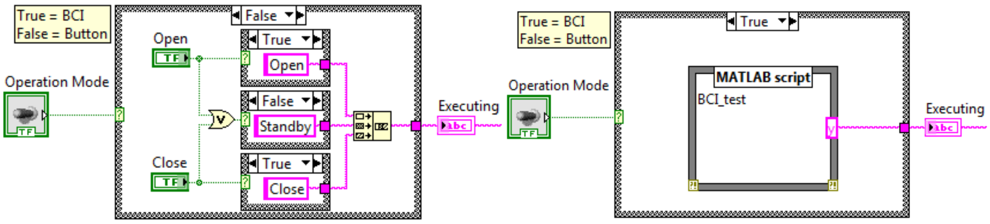


Figure 31 - Operation Stage.



(a) (b)

Figure 32 - Operation mode selection: (a) Button Mode and (b) BCI Mode.

Once the action to be performed is determined, the software enters in the second part of the Operation Stage, highlighted in Figure 33. In this section, the current positions of each motor are recorded. Then, the software enters a loop that only ends when the "Reset" or "Stop" buttons are pressed or when the desired position is achieved (opened or closed hand). At the beginning of this loop, the software compares the load readings from the motors with the Support Level selected. If the requirements are met, it goes one step towards the desired hand position. At the end of the loop, the software checks all motors positions. If the desired positions are reached, it exits the loop and starts the Operation Stage again by checking which Operation Mode is selected. If not, it compares again the load readings with the Support Level to determine if it can make another step towards the desired positions.

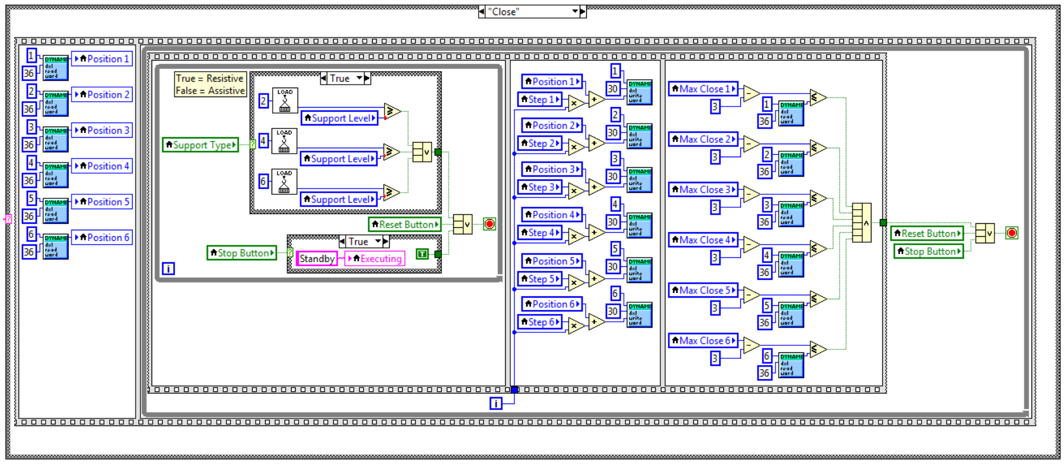


Figure 33 - System's motion control.

4. Results and discussion

The developed prototype demonstrates the functionality of the proposed design: a cable driven electromechanical system for hand motion assistance and rehabilitation. The system was able to effectively articulate all the fingers of the hand model, performing a grip-like movement. The actuation carried out by the AX-12A actuators, the pulleys and the cable system provided the necessary force to open and close the prototype developed to mimic a human hand. It must be pointed out that the system was not tested on real human hands, but, the obtained results suggest that the application of the model in an orthotic device would provide a feasible method for hand motor function rehabilitation. The cable guides have proven to be an elegant and effective solution for holding the cables in place without compromising the fingers' ROM. However, the cable system may be improved by using more durable and less stretchable materials, maintaining the system parameters. Regarding the control, the software and the graphical user interface reached the desired level of functionality. The overall cost of the prototype did not surpass the $350 USD, which is way cheaper than some commercial hand orthoses which may cost over $1500 USD.

A set of tests was performed to assess the functionalities of the developed system's hardware and software, as well as the operational models that were created. The tests aim at evaluating the performance of the whole system under different activities that it was designed to perform.

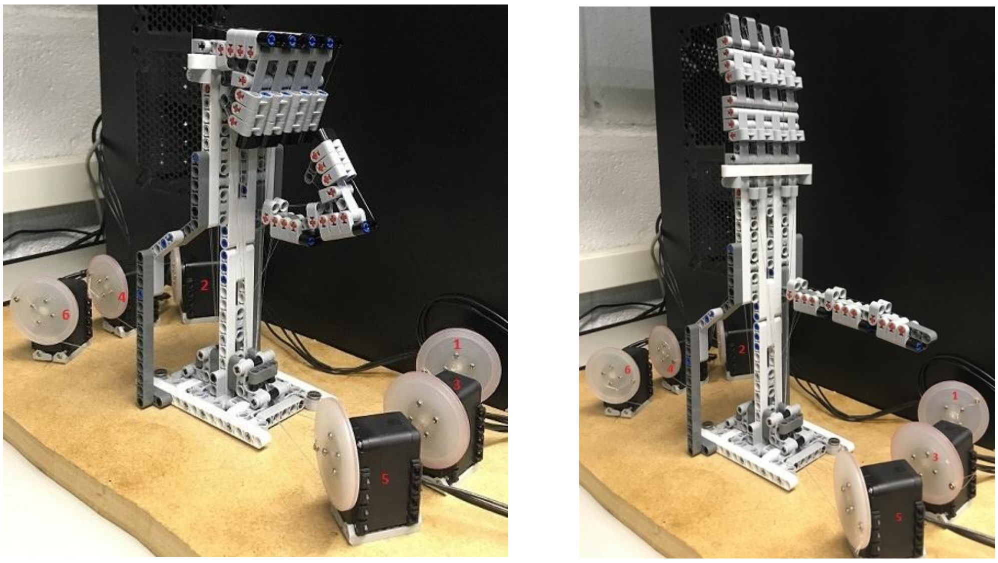
4.1. Safety tests

The first functionality to be evaluated was the user's safety, understood as the prevention of fingers hyperextension or hyperflexion. At the beginning of each training session, the device is calibrated taking into account the anatomic aspects of each specific user. The calibration consists in recording two motor positions for each one of the six motors presented in the developed system. One of the positions corresponds to the fully flexed hand and the other to the fully extended hand, as shown in Figure 34. After that, the motors only operate between their pair of limit values, avoiding finger positions that are naturally unreachable for the user. The objective of the safety test is to verify if those limits are never surpassed. The test consisted in 20 repetitions of opening and closing hand movements. Each motor had a different set of limits due to the different length of the cables connected to each finger. Table 1 shows the predefined limit values that were established for each one of the six motors present in the system, as well as the minimum and maximum values experimentally reached by them, among the 20 repetitions, the average of the reached values and their standard deviation. As can be seen in Table 1, each motor has a different set of minimum and maximum limit positions. This is caused by two main facts: the different distances between the motors and the fingers, and the different amounts of cable displacement required by each finger. The results show that the minimum and maximum positions of each motor, achieved for an opened and closed hand respectively (see Figure 34), remain almost constant and near to the predefined limit values, even under different operational conditions. In the tests where the limit values were surpassed, the difference between the established limit value and the achieved one was equivalent to 0.35°, in the worst case, which corresponds to 0.12 mm of cable displacement. Such difference does not compromise the user's safety.

Table 1 - Maximum and minimum reached positions for pre-established limit positions.

Motor	Established limit positions	Reached limit positions	Average of the reached limit positions	Standard deviation of the reached limit positions
1	270-800	269-798	269.00-798.00	0.00-0.00
2	100-550	100-548	100.00-548.00	0.00-0.00
3	120-850	119-849	119.00-848.20	0.00-0.41
4	200-800	200-799	200.00-799.00	0.00-0.00
5	180-900	179-899	179.00-898.90	0.00-0.31
6	150-740	150-737	150.05-737.00	0.22-0.00

It must be highlighted that the calibration process proved to be more complicated than expected. Since the model does not have any elements to maintain a certain position beside the motors and cable system, the user must hold the fingers in the desired position while using the dials, in the user interface, to tense the cable system. The cable system tension on the limit positions should be maintained low to avoid opposite motors from applying a load to each other. For this reason, achieving the equilibrium point at the desired position was tricky. Also, while calibrating the motors responsible for controlling the movement from one side of the hand, the motors responsible for controlling the movement from the other side of the hand must be set in positions that let loose their respective cables. This allowed to freely calibrate the motors, since opposite motors do not move synchronically during this stage. However, this methodology often let the cables so loose that they came out of the pulleys, and the user had to manually put them back while rolling them in.



(a) (b)

Figure 34 - Picture of the hand model. (a) Closed position (maximum motor positions from Table 1). (b) Opened position (minimum motor positions from Table 1).

4.2. Resistive mode tests

Several tests were performed to characterize the resistive support mode by determining how much resistance the device could offer to the user's desired motion and its relationship with the support level. These tests used a Z3H2R load cell from HBM, which has a load capacity of up to 1961.33 N and a resolution of 0.001 N. The cell was plugged to an HBM MGC Plus reader to register the measurements. An INSTRON 5500R universal testing machine was used to apply tension to the cables, with a maximum load capacity of 10 tons.

The first experimental test was performed by attaching the cables coming from an AX-12A motor to the load cell and then to the INSTRON, as seen in Figure 35. The motor's position was maintained while the INSTRON started to ascend approximately at 2 mm/min. In this configuration, the load cell measures the rising cable tension in newtons and this value can be associated with the increasing load readings obtained from the AX-12A. Table 2 shows the association between the discrete load levels of the motor and their respective forces, measured by the load cell.

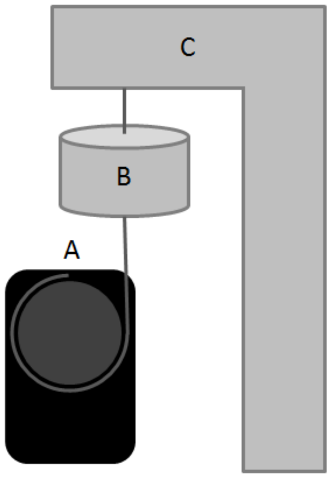


Figure 35 - Diagram for the direct motor force to load value association test. A: AX-12A actuator. B: Z3H2R load cell. C: INSTRON 5500R universal testing machine.

The load readings are discrete integer numbers, ranging from 0 to 1023, that increase with the external force, but they are not consecutive values. The readings start at 0, the next value is 64 and then increase in steps of 32 units. Also, the transition between one level and the next was not sharp, even using extremely low tension growing rates of around 0.01 N/s. The load readings returned by the motor were noisy near the transition between successive load levels. As the force increases, it was noticed that at a certain point the load readings started to oscillate between the present value and the next in the scale, or sometimes going further to an even higher value, before stabilizing on the new value. The values in Table 2 were taken once the new value was stable. The test was stopped at level 544, because the AX-12A raised an alarm corresponding to an overload error caused by having an increasing tension for almost an hour without any pause. However, the mean force applied by a healthy subject when pulling with one finger is about 60.09 N [197]. Therefore, if at the middle of the scale an AX-12A can provide 51.68 N, it can be assured that these motors are capable of providing enough force support to strength building training devices.

Table 2 -Equivalence between AX-12A load values and tensile forces for a 40 mm diameter pulley.

Load	Associated force	Force increment
0	0 N	-
64	5.55 N	5.55 N
96	11.94 N	6.39 N
128	14.44 N	2.50 N
160	16.17 N	1.73 N
192	22.55 N	6.38 N
224	24.95 N	2.40 N
256	31.00 N	6.05 N
288	31.80 N	0.80 N
320	36.80 N	5.00 N
352	40.07 N	3.27 N
384	43.01 N	2.94 N
416	45.76 N	2.75 N
448	46.95 N	1.19 N
480	49.67 N	2.72 N
512	51.68 N	2.01 N
544	53.18 N	1.50 N

It was expected that for every 32 units increment on the scale, the force increment in newtons would be linearly proportional. However, the obtained results show a nonlinear dependency between the two set of values. It was assumed that this behavior was caused by the noisy values obtained near the transition between successive load levels. The average value for the force increments is around 3.13 N with a standard deviation of 1.76 N. The obtained standard deviation is considerably large when compared with the average force value, which demonstrate the high variability of the readings when compared to the expected force values. This variability may be related to the fact that the values in Table 2 were registered only when the load readings stabilized, after the transition between the previous level and the actual one.

A second test consisted into attaching the cables coming from one of the fingers of the hand model to the load cell and then to the INSTRON, as seen in Figure 36. The motor A was set to maintain the finger at a fixed position. In this configuration, as the crane goes up, the rising cable tension is measured in newtons by the load cell. These values can be associated with the load readings from the AX-12A motor as the force needed to be applied by the user in order to overcome each resistive support level. Unfortunately, the hand model was too flexible, due to the plastic pieces and the assembly method, this caused the model to bend rather than effectively transfer the tensile force from the INSTRON to the motor. The motor load value changed from 0 to 64 only at a load cell reading of 9.7 N. Besides, even increasing the load cell readings up to 60 N, the motor load level never reached any value above 64, which means that during the whole test it received less than 11.94 N. This result, however, do not affects the usage of the proposed electromechanical system in a finger motion assistive device, since the structural base model was the element that compromised the results, and it will not be part of the final orthosis design which will have the user's hand as structural base.

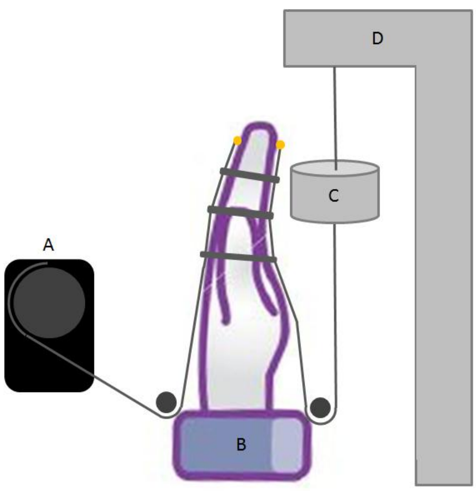


Figure 36 - Diagram for the finger force to load value association test. A: AX-12A actuator. B: Hand model. C: Z3H2R load cell. D: INSTRON 5500R universal testing machine.

4.3. Assistive mode tests

Another test was performed to assess the load levels that make the motors to hold their positions for different support levels in assistive support operation. This was done by holding the fingers in their position with plastic pieces, as seen in Figure 37, and registering the load values at which the motors stopped their motions. Figure 38shows the test results in a Support Level vs. Stop Load graph.

Ideally, the corresponding stop load value for a given support level should be equal to the higher stop load value closest to the respective support level. Then, for example, for any support level under 64, the stopping load would be 64, and so on. However, the test results showed that when the support value was near, but still beneath, the stop load value threshold, it already jumped up to the next stop load level. This behavior can be caused by oscillations at load value readings, that occur when they are close to transition values (64, 96, 128, 160, etc.), causing the system to register a higher load value than the one been actually exerted on the AX-12A motor. However, the results confirm a directly proportional relationship between the required force needed to make the motors stop and assistive support values. This behavior makes the system suitable for a configurable finger motion assisting device with different levels of support.

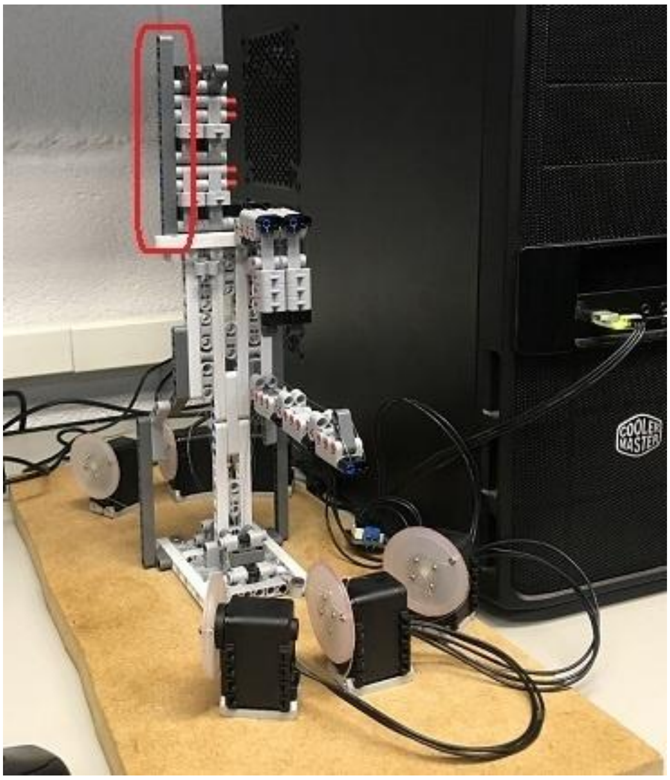


Figure 37 - Picture of the experimental setup for the assistive mode tests. The element marked in red are the pieces that hold the fingers in a fixed position.

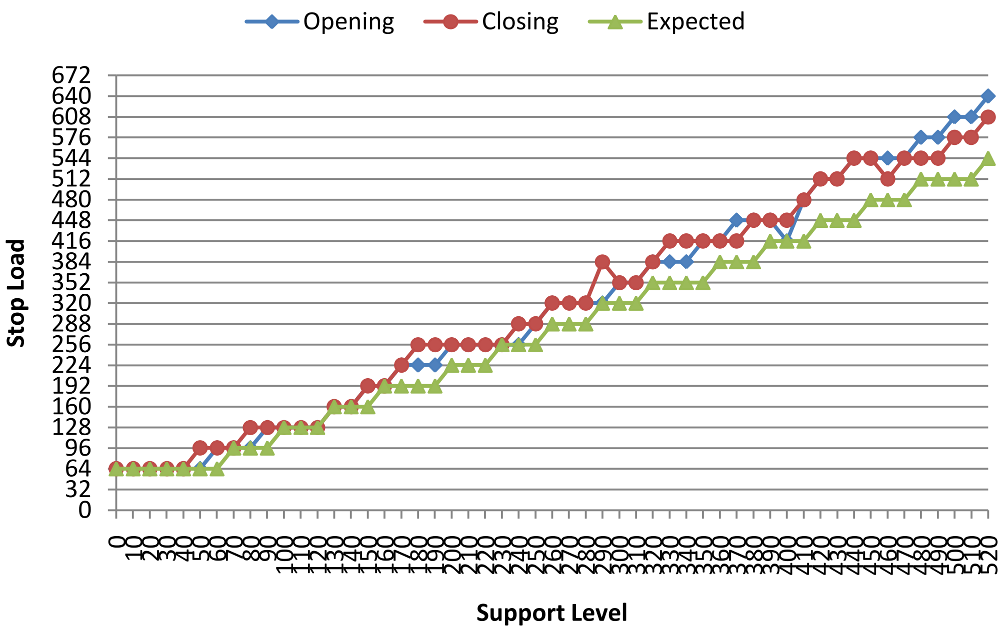


Figure 38 - Load values of motors holding their positions for different Support Levels.

5.
Conclusions and Future Works

5.1.
Conclusions

Even though there are several hand motion assisting devices in the present day, the electromechanical system developed in this project, for a rehabilitative actuated glove, includes features that most systems do not possess. The system comprises a lightweight, low cost, computer controlled, adjustable cable driven actuation system that can be used in actuated orthoses to assist patients in performing repetitive hand opening and closing exercises. These characteristics were achieved through a cable and guides system that proved to be an elegant and effective solution for actuating the fingers while minimizing the weight on the distal upper limb. The result is a more compact system that does not interfere with the fingers' ranges of motion or the grasping capabilities of the patients, allowing them to interact with their environment. Such features, although desired, are not common in many of the existent designs, which are usually expensive, bulky, heavy or do not offer control over the whole system operation.

The pulleys and actuators of the developed system allowed sufficient force to move the model fingers along their full range of motion and with enough resolution for multiple positions. This would allow an orthosis based on this system to assist in opening and closing hand movements comprising all of the fingers joints. This provides an advantage over most systems developed to date, since most of them do not assist both types of hand movements or comprise the motion of only a few of the fingers' joints in their movements. Besides being able to actively move the developed hand model, the system is capable of resisting intended movements with enough force to aid in strength building training sessions. Such characteristic can provide a future orthosis the capability to be used in movement assistance and hand strengthening therapies, which is not a very common capability in the devices reviewed in the present research. Different tensile forces may be supplied according to the level of support selected in the developed graphic user interface. The system was capable of resisting more than 60 N, for support levels set near the middle of the full scale, which is the average force exerted by a healthy person's finger when pulling.

The graphic user interface created for controlling the system and the future orthosis enables the user to fully control the device motion at any time. The interface offers two control options, Buttons and Brain-Computer Interface, allowing different rehabilitation approaches in recovery therapies for stroke survivors. The Buttons option offers a more classic approach for the system control, while the BCI control allows the communication with such device to increase the patients' recovery rates by synchronizing the patients' intentions with their hand movements. The interface also allows the configuration of the system parameters to adjust it to the patients' specific needs. Regarding the physical characteristics of the patient's hands, the system can be adjusted by controlling the minimum and maximum positions the system can reach. This physical adjustment proved to be capable of ensuring the patients' safety by limiting the system movements only to positions that can be naturally reached by the patients. The limit positions were reached with little to no variability during the tests. Such variations were always around 0.12 mm of cable displacement, which does not compromise the patients' safety or the system performance. Another configurable feature of the developed system is the type of support (assistive or resistive) and the level of support it provides. Such adaptability enables the use of the system by patients in different stages of recovery, by allowing them to easily adjust the degree of assistance provided by the cable system. Last but not least, the system grants the users the capacity to stop or reset the system motion at any time. This feature, although needed at the end of any training session, is also provided as a safety functionality, by giving the patient the capacity to easily stop movement, in case any discomfort arises, or even set the system loose.

The future of this model is the development of a universal device that can be used by people recovering from stroke or that just need rehabilitation therapy or motion assistance.

5.2. Future Works

The project goals of creating an electromechanical system capable of assisting hand motion for a hand motor function rehabilitation device were met. However, this model is not an actuated orthosis yet. To achieve the actual creation of a hand rehabilitation device there are some works to be performed:

- Design the glove-like structure that the users are going to wear. Such glove must be adjustable to different hand sizes in order to make it as versatile as possible. This design must follow the safety standards for medical equipment defined by the International Electrotechnical Commission (IEC) and the International Organization for Standardization (ISO), such as the IEC 60601-1:2005+AMD1:2012: Medical electrical equipment – Part 1: General requirements for basic safety and essential performance.
- Use a Bowden cable system from the motors to the glove in order to transfer cable movement over a distance independently from the positions of the shoulder, elbow and wrist joints.
- Create a more compact system, using a microcontroller, like a PIC or an STM32, instead of a computer as the control unit, in order to make the device more portable.
- Improve the calibration method to avoid that the cables get out of the pulleys during this stage.
- Add communication capabilities to the software in order to control the device from BCIs developed in environments like Python.
- Analyze the effect of gravity over the developed system.

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