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CABLE-DRIVEN ACTIVE ARM ORTHOSIS FOR REHABILITATION OF THE UPPER LIMB

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Abstract. *Stroke alone is the leading cause of disability worldwide. Consequently, with the aging of the population, it is expected an increase in people dependent on assistive technologies due to neurological disorders. These conditions lead to enormous impacts on the quality of life by the individual's capacity to accomplish basic activities of daily living, causing life-long dependency. In this context, the movements of the arm play a special role, allowing the individual to interact with the ambient and relating to the performance of several of our daily tasks. Therefore, a special attention must be given for the rehabilitation of the upper limb. For that, the robotic rehabilitation has emerged as a novel treatment for the neuromotor rehabilitation of impaired limbs, due to relevant evidence of its effectiveness and reduction of therapist active assistance. While several robotic devices have been proposed to mimic and rehabilitate the human arm, few have the purpose of being lightweight, especially due to the heavy motors and gearboxes composing these equipments. In this perspective, we propose a cable-driven active arm orthosis for the rehabilitation of the arm and accomplishment of activities of daily living, with the goal of being lightweight and portable. The cable-driven actuator allows for the relocation of the motor unit from the users joint by means of a bowden cable-driven transmission, reducing the overall weight, volume and inertia of the system. That is especially relevant in this scenario, while lower limb exoskeletons can compensate for the weight of the structure, upper limb exoskeletons require the user to carry and sustain the device's weight. Therefore, the optimization of mass and volume are one of the main requirements for such equipments. The resulting mechanism allows for the movements of flexion and extension of the elbow and pronation and supination of the forearm, while providing adequate torques, speeds and ranges of motion for performing the basic activities of daily living.*

Keywords: *Active Orthosis, Robotic Rehabilitation, Wearable, Assistive Technology, Upper Limb*

1. INTRODUCTION

Stroke is a recurrent neurological disorder worldwide, and the main cause of adult disability (Ward and Cohen, 2004). Patients with those conditions suffer from severe movement limitations, reducing the individual's capacity of accomplishing basic activities of daily living (ADLs), consequently causing life-long dependencies (Stinear, 2010). Particularly, the movements of the upper limb play a special role, allowing the interaction with the environment. With the predicted global aging of the population (Kinsella and Phillips, 2005) and the close relationship between neuromuscular disfunctions and aging (Béjot et al., 2016), an increase in the number of people dependent on solutions enabling the rehabilitation of impaired limbs is expected (Béjot et al., 2019). In this context, robotic rehabilitation has emerged as a novel therapeutic approach for the neuromotor rehabilitation of impaired individuals, due to relevant evidence of its effectiveness and reduction of therapist's active assistance (Andrade et al., 2019) (Klamroth-Marganska et al., 2014). In cases where therapist induced movement is needed, especially in cases of severe spastic symptoms, which fatigue sets on the therapist, the manual therapy shows its's limitations and robotic therapy comes into play.

When considering therapy efficiency, it has been shown that recovery is significantly influenced by repetition and extension of the therapy session. In this case, robotic wearable and portable devices show greater advantage by allowing the patient to address rehabilitation routines outside of the clinic. Overall, while several devices have been proposed to mimic and rehabilitate the human upper limb (Gopura et al., 2016), most still lack in key aspects that could lead to better results in patient development. Due to its application background, such devices must meet requirements of safety, comfort and effectiveness. Existing designs of rehabilitation robots often cause discomfort to the user as they are not designed for long time usage. Considering that these equipments are essentially designed to share common space with the user, comfort becomes a key issue (Yakub et al., 2014). In this perspective, the construction aspects of the system turn out to be a relevant factor to achieve adequate usability and, consequently, better efficacy. When considering the overall weight and volume of the system this matter becomes even more relevant, while lower limb exoskeletons can compensate for the weight of the structure (Andrade and Bonato, 2021), upper limb exoskeletons require the user to carry and sustain the equipment's weight, significantly affecting its functional efficacy (Hyun et al., 2019). This issue is hardly addressed because of the naturally heavy components of these exoskeletons, such as electric motors and gearboxes.

In this context, bowden cable-driven actuators allow for great advantages over conventional transmission systems, allowing for the relocation of the actuation unit away from the user's joint, reducing the overall weight, volume and inertia of the equipment. This is relevant specially when considering wearable devices for long treatment sessions or to be used as a daily driver in the assessment of the activities of daily living, to improve usability. To address this situation, in this work, a wearable cable-driven active arm orthosis is proposed, for the rehabilitation of the arm and accomplishment of activities of daily living, with the goal of being lightweight and portable. A control architecture, based on electroencephalography biological signal, to be further implemented, is presented. The resulting mechanism covers 2 degrees of freedom (DoF), allowing for the flexion and extension of the elbow ranging up to 80% of the full range of motion and up to 18 *N.m*, and pronation and supination of the forearm addressing the full range of motion for the ADLs, while providing up to 3 *N.m* of torque. The device actuated unit of the device weighted only 0.835 *kg* while providing adequate movement and safety measures.

2. MECHANICAL AND DESIGN REQUIREMENTS

Designing robotic devices for rehabilitation is a process that relies on several different factors: the system's weight, stiffness, volume, safety, and multiple other parameters. Most importantly, to provide adequate conditions for patients exercising the equipment must rely on the natural conditions of the human body, replicating the operational workspace and dynamic capabilities of the healthy human upper limb. Therefore, to comprehend the design methodology applied to the development of the mechanism responsible for the movement of the user's joint, one must first understand the functional aspects of the movements of elbow flexion and forearm rotation. To define the required kinematic and dynamic parameters required for the system, the activities of daily living must be analyzed. While for the lower limb the kinematic aspects of the movements are usually analyzed for a specific pattern, as for the gait, the upper limb is responsible for different nonstandard tasks. In this case, the kinematic and dynamic characteristics must be defined based in range obtained from the aspects of a set of activities, normally being basic activities of daily living, such as eating, reaching for an object or hygiene-related tasks.

2.1 Elbow Flexion and Extension

The elbow joint (ulnoumeral joint) is primarily considered a hinge type, although studies have shown that variations in the center of rotation exist during the movements of flexion and extension (Ishizuki, 1979), from a practical point of view, the deviation is minimal, thus the articulation can be assumed to be fixed and uniaxial, except at the extremes of flexion and extension (An and Morrey, 2018). Moreover, it is hard to define the exact position of the joint's axis of rotation as it is covered in biological tissue, adding up, no fixation of the exoskeleton to the body is ever perfectly rigid, thus, commonly, slips will generate eventual misalignments.

In normal conditions, human elbow flexion will range from 0º, when fully extended, to 150º degrees in flexion. A study of Perry et al. (2019), on the other hand, analyzing 19 different tasks related to the activities of daily living has shown that elbow flexion is mostly limited to angles below 120º with a mean value of 92.1º. Previous studies from Morrey et al. (1981) had shown similar results, with a range of motion of approximately 100º, ranging from 30 degrees in flexion up to 130º. Another relevant aspect of the kinematics of the elbow is the velocity expected for the accomplishment of relevant tasks. Murray (1999) evaluated several different activities of daily living, finding a maximum elbow angular velocity at 3.6 *rad/s* for the tasks of reaching the back of the head and lifting an object to shoulder height.

Considering the dynamic aspects of the motion of the forearm it is relevant to evaluate the torques expected at the joint. Defining the adequate torques for human elbow flexion is a complex task, as this is a highly individual strength aspect. In the previously presented study from Perry et al. (2019), the torques were observed with a mean value of 0.45 *N.m*, but reaching maximums of 3.5 *N.m*. Another study by Murray and Johnson (2004) analyzing healthy subjects while performing 10 different ADLs observed elbow moments peaking at 5.8 *N.m*. In fact, as previously discussed, the torques will range differently according to the proposed task or the weight being carried. Another important aspect to be considered is that one of the most recurrent conditions of a poststroke patient is spasticity, which produces high undesired resisting torques at the elbow joint during extension. Studies have shown that spastic torque varies greatly for each subject, but values from 4 *N.m* to 8 *N.m* have been observed (Schmit et al., 1999).

2.2 Forearm Pronation and Supination

The movements of pronation and supination consist of a complex joint configuration in which, during the distal portion of the radius rotates around the head of the ulna, such displacement is minimal in the proximal portion of the bone when compared to the distal region. This leads to conclude that the axis of rotation is not parallel to the bones. Another relevant aspect of the forearm is that the distal portion has a greater amount of soft tissue when compared to regions closer to the wrist, this makes any physical human-robot interface stiffer when occurring in the distal portion of the forearm. That is mainly because soft tissue will suffer excessive deformation due to interaction forces, whereas the bone will act as a rigid interface (Jarrasse, 2012).

These movements are extremely important for properly performing several tasks. Studies by Morrey et al. (1981) and later revisited by Sardelli et al. (2011) evaluated that most activities of daily living can be performed with a 100º range of motion, reaching from 50º to -50º measured from the neutral position. It has also been observed that angular velocities at this joint reach values of over 7 *rad/s* (Rahman et al., 2014).

The dynamic aspects of the rotation of the forearm show great divergence when comparing different activities and conditions. Although maximum isometric torques for pronation and supination reach a high torque of up to 16 *N.m* (O'Sullivan and Galway, 2002), during tests for ADLs different studies from Perry et al. (2007) and Murray and Johnson (2004) observed torques below 1 *N.m*.

2.3 System requirements

Based on the previous review for the kinematics and dynamics of the arm joints during the performance of the activities of daily living the requirements for the actuators comprising those joints can be defined. One important aspect to be taken into consideration when defining a requirement in a mechanical system is the power transmitted. A study by Danoff (1978) evaluated the power produced during elbow motion in 20 healthy subjects, concluding that the maximum output power occurs at 50% of the maximum load. Although higher loads could be sustained, from this point on the lower velocities resulted in a lower power. In this regard, maximum power during elbow flexion can be calculated by means of 50% of the maximum expected joint torque. Therefore, Table 1 presents the kinematic and dynamic requirements for each joint. One important notice is that those values should be expected at the joint, and do not consider the transmission efficiency, which will be later discussed.

Table 1. Kinematic and dynamic requirements of each joint for the accomplishment of ADLs

Requirement	Flexion/Extension	Pronation/Supination
Range of motion (ROM), º	120	100
Angular velocity, rad/s	3.6	7.0
Mean Torque, N.m	5.8	1.0
Power, W	10.5	7.0

Other previously discussed aspects of rehabilitation robots must also be taken into consideration. Primarily, the goal is to produce a portable active arm orthosis, consequently, usability characteristics must be a primary goal. Therefore, the design must take into consideration the biological mechanical properties of the joints while considering aspects of overall mechanism weight and svelteness. Considering this, it is proposed that the mechanism is as light as possible, preferably below 1 *kg* and as slim as possible, to reduce overall inertia supported by the user's limb.

3. MECHANICAL DESCRIPTION

3.1 The Bowden Cable-Driven Transmission

To achieve the desired ergonomic goals, a bowden cable-driven actuator is proposed. One major advantage of this configuration is that it allows the heavy motors and gearboxes to be relocated from the user's limb by means of flexible sheaths through which cables are guided transmitting forces and power, allowing for a remote actuation. Specifically, in the case of single-axis rotary joints, as defined by the two degrees of freedom addressed by the proposed exoskeleton, a pull-pull configuration is used. In this case, a motor with a driving pulley is attached to two cables which in the opposite end are attached to a driven pulley acting as a rotary actuator, as presented in Figure 1. Rotation of the motor in one direction will produce a retraction of one of the cables, promoting motion by means of the mechanical displacement between the cable and the flexible sheath. To prevent slacks during motion, the cables are commonly pre-tensioned.

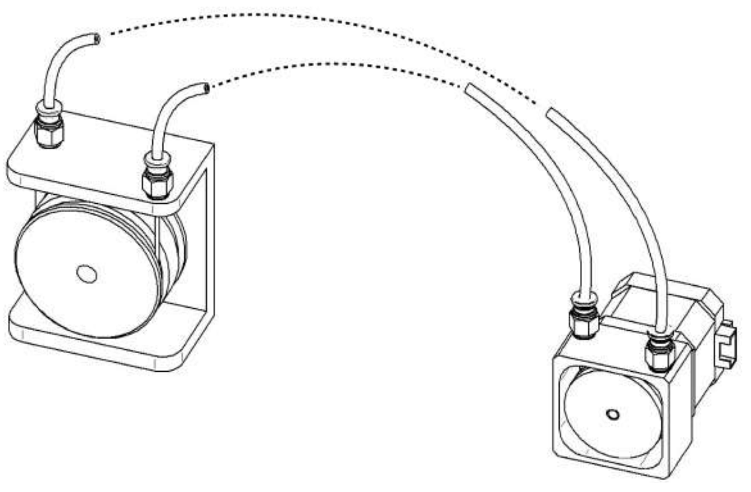

Figure 1. Schematics of a bowden cable-driven rotary actuator, in the right is the moving pulley attached to a motor, and in the right is the moved pulley.

One relevant aspect to be taken into consideration when designing cable-driven actuators are the inefficiencies. Losses occur due to complex non-linear friction produced by the sliding friction between the cable and its housing. A relevant factor to be taken into consideration is the material pair used, most commonly, the sheath will be coated with or made of polytetrafluoroethylene (PTFE), due to its low friction coefficient (Carlson et al., 1995). The force transmission in this configuration can be, theoretically, modeled as a cable sliding over a pulley, as presented in Eq. 1, where μ is the friction coefficient for the material pair and θ is the sum of bending angles (Schiele et al., 2006).

$$F_{input}/F_{output} = e^{-\mu\theta} \tag{1}$$

The kinematical characterization of a bowden cable actuator is that of a cable attached to a pulley, consequently, no slacks are expected if the cables are properly tensioned. Since the cable is rigidly attached to both the driving and driven pulley, the motion transmission can be characterized as such for a pulley transmission and the reduction ratio can be defined as the ratio between the pulley's diameters. One relevant aspect is that since motion of the cable in the driving pulley is entirely transmitted to the driven pulley, the end position of the actuator can be defined from the position of the motor. Consequently, since DC motors are easily instrumented with encoders for position control, there is no need for instrumentation in the driven portion of the actuator. In this context, the angular position of the actuator can be defined in terms of the angular position of the motor by Eq. 2, where the term in parenthesis represents the reduction ratio provided by the pulleys (being D_P and D_A, the diameter of the driving and driven pulley, respectively), i_r corresponds to the gear ratio of any reducer attached to the motor and θ_A and θ_M, correspond to the position of the actuator and of the motor, respectively. Other kinematical modellings for velocity and acceleration can be obtained from the differentiation of Eq. 2.

$$\theta_A = \left(\frac{D_P}{D_A}\right) i_r \theta_M \tag{2}$$

3.2 Mechanism

Previous studies have analyzed in depth the actuator of pronation and supination of the forearm taking into consideration the different possible configurations for the execution of the movements (Dias and Andrade, 2020). The most basic approach for an exoskeleton to perform the rotation of the forearm consists of having a circular bearing over the forearm. In this configuration, the arm is attached to the internal ring while the external ring of the bearing is connected to the proximal part of the orthosis, consequently, the angular displacement between the rings produces the forearm rotation. This approach has a major downside of demanding the user to put his arm through the equipment, which makes it difficult to execute a fast removal of the orthosis in emergencies and produces heavy and bulky structures.

Based on this a C-shaped mechanism is proposed. In this configuration, a moving rail attached to the user's wrist allows for rotation relative to a stationary guide. Although this usually still results in heavy and bulky mechanisms, the use of anthropometric data in the design stage allows the development of an optimized mechanism, improving usability (Dias and Andrade, 2020). Figure 2 presents the proposed configuration for the actuator of pronation and supination.

In this design, a 150º range of motion is accessible by the user, while weighting only 0.145 *kg* and fitting the 85th percentile of the population according to anthropometry data

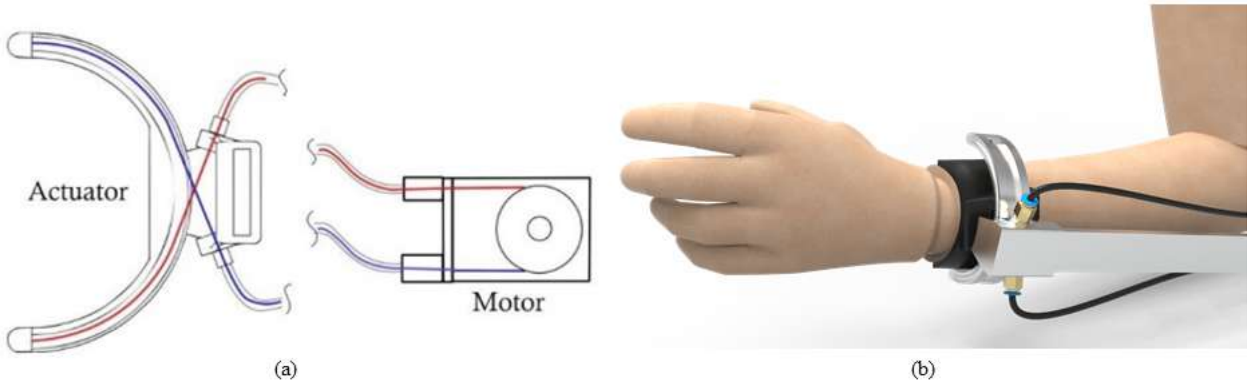

Figure 2. (a) Schematics for the cable-driven actuator of pronation and supination of the forearm. (b) Rendering of the actuator.

For the flexion and extension exoskeleton a simpler approach was taken, transmitting the motion by means of a standard driven pulley, to which the cables are attached. Considering that the needed range of motion is of 120º, a single slot, slim pulley to which both cables are attached following a single guide has been designed in order to reduce the overall size and weight. The pulley is then attached to a link which comprises a cuff to be attached to the user's proximal portion of the forearm, and the pronation and supination actuator, which is attached to the wrist. This link is then responsible for transmitting the torque from the pulley to the wearer's limb for actuation. A pair of bearings were used to support the pulley to transmit only tangential forces to the link preventing undesired forces in the elbow joint. One important aspect is that the actuator is angled relative to the sagittal plane in order to address the inclination of the axis of rotation of the elbow according to its carrying angle while also considering it a stationary axis. This inclination is produced relative to the link that is attached to the arm. Figure 3 presents the setup of the actuator for flexion and extension with the link responsible for transmitting the loads to the user's forearm.

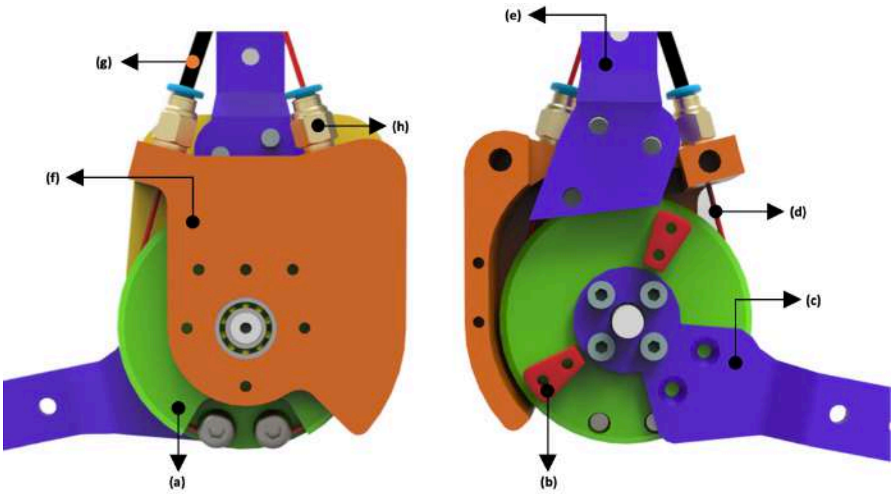

Figure 3. Front and back of the cable-driven actuator of extension and flexion of the elbow (illustrating colors). (a) actuator driven pulley, (b) flexion and extension limiters, (c) distal link, (d) driving cables, (e) proximal link, (f) pulley housing, (g) bowden housing, (h) pneumatic couplings.

Adjustable mechanical limiters are located behind the link and can be used to adequately change the range of motion according to the user's necessity. This characteristic allows for safe operation reducing the chances of hyperflexion or hyperextension of the joint, optionally, the adjustable limiter can be unused, in this case the actuator will be intrinsically limited to a range of motion of 120º by the pulley housing.

To achieve the needed torques and velocities previously discussed, sets of motors and gearboxes were used. Commonly, DC motors produce low torque and high velocities, therefore, some type of gearbox is needed in order to produce adequate dynamic and kinematic condition. To select the appropriate motor for each degree of freedom, the expected output power was taken into consideration, for that, the expected system inefficiencies must be taken into consideration. A study by Schiele et al. (2006) evaluated force transmission efficiency for different bowden cable transmission set-ups, for a PTFE-Steel pair an efficiency of, as low as, 70% can be expected. Since the pulley reduction ratio is not enough to achieve adequate torques, some other reducer is needed coupled to the motor, commonly, planetary reducers or harmonic drive reducers will operate in an 80% to 70% efficiency range, depending on its conditions. Consequently, an overall minimum efficiency can be expected at 50% for the transmission. Minimum expected power, therefore, for the motors to achieve adequate torques must be of 21 W and 14 W, for the flexion/extension and pronation/supination degrees of freedom, respectively.

Brushless DC (BLDC) motors were then selected according to the needed power, higher torque BLDC motors were prioritized in order to reduce the overall needed reduction ratio, consequently minimizing the overall weight of the system by reducing the number of stages needed in a planetary gearbox. The selected motors for each degree of freedom are presented in Table 2, where the reduction is calculated in order to achieve the appropriate velocities.

Table 2. Selected motors and reduction ratios at each stage for each DoF, Maxon® motors and gearheads were selected. Flexion/Extension: EC 45 Flat + GP 32 A. Pronation/Supination: EC-i 30 + GP 32 C.

Requirement	**Flexion/Extension**	**Pronation/Supination**
Motor power, W	30	20
Motor nominal speed, rpm	3290	6000
Motor nominal torque, mN.m	66.0	32.6
Needed reduction,	97:1	91:1
Selected Planetary gearbox reduction,	66:1	66:1
Designed cable-driven transmission reduction	1.5:1	1.4:1

Controlling BLDC motors requires adequate drivers, being field oriented control (FOC) the most common strategy for position and torque control (John et al., 2011). Maxon® provides proprietary controllers with native position, velocity and torque control for its brushless motors, thus, the EPOS4 (Maxon, Switzerland) digital positioning controller will be used to produce the primary experiments with the presented design.

4. CONTROL STRATEGY

Control strategies for rehabilitation robotics differ greatly from common robotics control, that's mainly due to the presence of the human operator who integrates as part of the control loop (Yang et al., 2008), since it shares common space with the mechanism and is directly related to the system's dynamics. In this condition, the mechanism's operation is defined by the operator's decision, for that, the system must be capable of understanding the user's motion intention (Huang et al., 2015). For that goal, control methodology of exoskeletons can be classified according to the control input, being biological signal based, non-biological signal based and platform independent (Gunasekara et al., 2012).

Control methods based on human biological signals are mostly based on electromyography (EMG) (Balasubramanian et al., 2018) or electroencephalography (EEG) (Bhagat et al., 2014), to detect muscle activity levels or classify motion intention. Biological signal based on EEG has been shown to be effective for rehabilitation purposes, inducing neuroplasticity (Ang et al., 2015). The control architecture for this cable-driven active arm orthosis, presented in Figure 4 is based in a state machine that switches between transparent operation mode (Andrade et al., 2019), in which the movement of the mechanism is transparent to external forces, and torque control according to motion intention detection by means of EEG sensing, in which flexion/extension and pronation/supination occur based on motion intention classification and a torque gain is defined based on activity level.

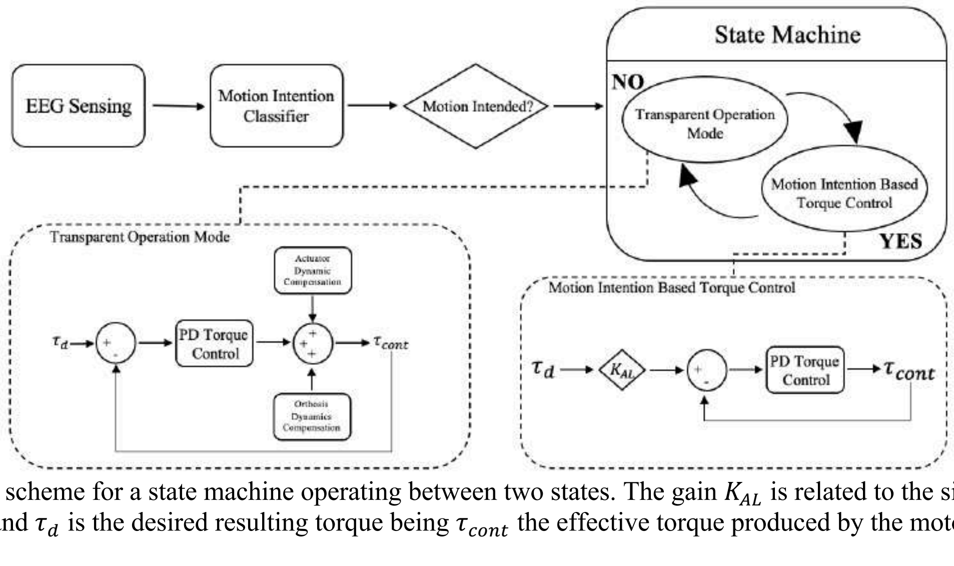

Figure 5. Control scheme for a state machine operating between two states. The gain K_{AL} is related to the signal activity level and τ_d is the desired resulting torque being τ_{cont} the effective torque produced by the motor

5. RESULTS

Although no topology optimization algorithm was applied to the system the overall mass obtained for the actuation mechanism was lower than the proposed requirement, for that all parts of the actuator were designed in aluminum alloy 7075-T6. The actuator for flexion and extension weighted 0.341 *kg* and the actuator for pronation and supination weighted only 0.145 *kg*. By considering the links and cuffs needed for the force transmission to the user's limb, the final weight of the structure is estimated in 0.835 *kg*. Both the links and cuffs used were adapted from an elbow immobilization orthosis (Brace Pauher - OrthoPauher) Even at a low weight, by means of the bowden cable-driven transmission it was possible to obtain a mechanism that achieves the adequate kinematic and dynamic requirements of the joints for activities of daily living and matching safety and usability requirements.

With the configurations previously proposed in Table 2, the system achieves a nominal torque of 6 *N.m* and a maximum torque of 18 *N.m*, considering the 50% efficiency conditions in the second case, for flexion and extension. For the actuator of pronation and supination a nominal torque is expected at 1.5 *N.m* and 3.2 *N.m* of maximum torque, at 50% efficiency. The proposed design also allows each degree of freedom to achieve its full functional range of motion for activities of daily living, 120º for flexion/extension of the elbow and 150º for pronation and supination of the forearm. The final design containing both actuators is presented in Figure 4.

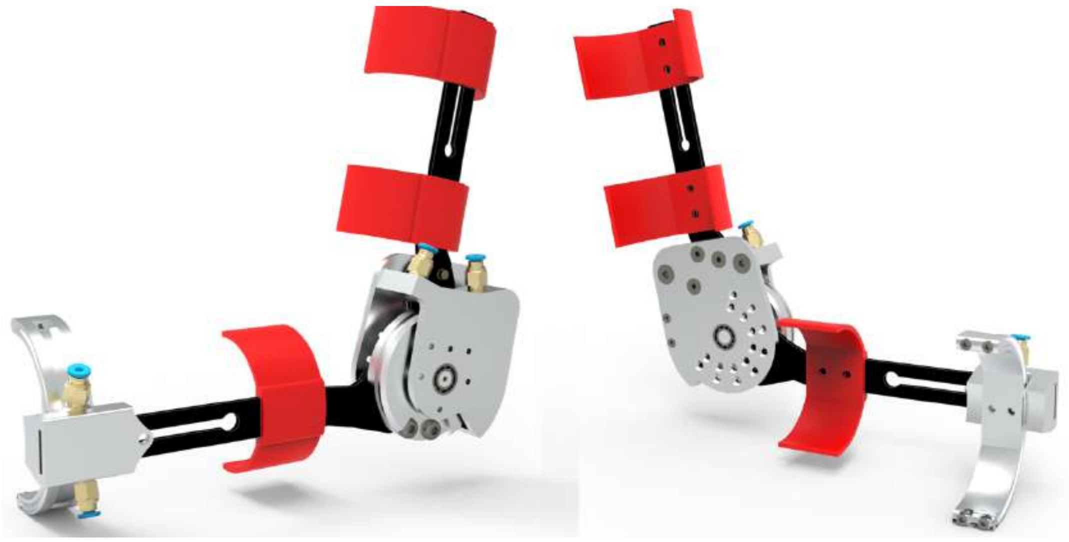

Figure 5. Cable-driven active arm orthosis for elbow flexion and forearm rotation. In red is a representation of the cuffs that attach to the user's arm.

The proposed control strategy should allow for safe and transparent operation, avoiding unnecessary constraint of the arm during unoperated moments, requiring the user's engagement in the activity to produce movement. This combination of conditions should allow for the equipment to be used in daily living, while still promoting rehabilitation at home and during daily activities, by exercising producing feedback to motion intention.

6. CONCLUSION AND FUTURE WORK

The main goal of this is study has been to propose and develop an active arm orthosis for the rehabilitation of the arm in impaired patients. To appropriately achieve this goal, aspects of performance and usability were defined for an adequate mechanism which is both capable of accomplishing proposed tasks safely. With the increasing tendency in robotic rehabilitation technology, and mostly with equipments that are user focused, the usability properties of a wearable robot become greatly relevant. In this regard, the cable-driven actuator has been presented as an alternative to achieve the design requirements.

The actuator resultant can achieve a range of motion of 150º in forearm rotation and 120º in elbow flexion and achieving adequate torques for all the degrees of freedom. The nominal torques at minimum efficiency conditions are expected at 6 *N.m* and 1.5 *N.m*, respectively, for flexion/extension of the elbow and pronation/supination of the forearm, while the moments during the accomplishment of ADLs reach maximums of 5.8 *N.m* and 1 *N.m*. These dynamic and kinematic requirements are met while weighting a total of 0.835 *kg*, lower than the defined requirement.

Future works will analyze dynamic aspects of the bowden cable-transmission and its reliability for medical purposes. Important future studies also must on the implementation and improvement of the control strategy for rehabilitation purposes. One important aspect for the transparent operation control is the dynamic modelling of the actuators, for that friction must be modelled and such mechanism includes complex non-linear friction phenomenon, that can be better understood and mapped experimentally. After manufacturing, the prototype must also undergo usability and safety tests in order to verify that no parasitic forces and moments are produced in users joint.

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8. RESPONSIBILITY NOTICE

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