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Rehap: A lower limb rehabilitation system for adults

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Abstract. During passive rehabilitation, the medical therapist performs specific routines on the patient's limbs who have suffered physiological damage and has partial lost or total mobility and muscle strength. Currently, there are commercial devices of manual nature that assist the doctor during the application of routines. However, the majority are closed kinematic chain which limits the number of angles that can be reached. In this article, it is detailed the design and development of an electro-mechanical system for an autonomous passive lower extremity rehabilitator. The mechanical system is developed considering the knee's biomechanics, the change in length of the limb during flexion and extension, and the anthropometry of the patients to be treated. The system is actuated by electromechanical elements to avoid costly pneumatic systems or hydraulic networks. The scheming and development of the transmission system for the electromotive actuator selection are also detailed in its corresponding section.

Keywords: Rehabilitation, Passive, Biomechanics, Knee

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1 Introduction

Rehabilitation, as defined by the WHO, is "a set of interventions designed to optimize functioning and reduce disability in individuals with health conditions in interaction with their environment" [1].

During rehabilitation, the therapist aims to reactivate and strengthen the limbs that have suffered from physiological damage. His goal is to return the person to the mobility and strength he had before the mishap as soon as possible. There are two types of rehabilitation, passive and active. In passive rehabilitation, the therapist mobilizes the limb by stimulating the nerves and muscles to reactivate them; this type of rehabilitation is suggested by therapists, in postoperative patients, and should be carried out as soon as possible to recover the mobility of the affected limb [2].

In active rehabilitation, the doctor proposes exercises or routines to the patient, who tries to strengthen and recover the movement of his limbs.

Nowadays there are different methods, techniques, as well as equipment for passive rehabilitation, that help the patient and the doctor to fulfill the patient's recovery. However, most of these methods are manual or have mechanical nature, which can lead the therapist to suffer from fatigue after repeated routines applied to different patients, and therefore reduce their performance.

In this article will focus on the design and development of a mechanical system for autonomous lower extremity rehabilitation, which will allow the automation of rehabilitation routines, relegating the therapist's responsibility to evaluation and assignment of sequences.

Unlike other commercial systems, the proposed device is smaller in size and works with electromechanical elements, which avoids expensive pneumatic and hydraulic installations, allowing easy installation, displacement, and use.

The presented work in this article is divided as follows: In the state of the art, it is detailed those experimental or comme rcial systems that currently exist and whose function or objective is like ours. In the development section we presented the limitations and objectives of the system, the workspace and expected patients. It is detailed the development of the mechanism and their calculation of the gearbox, and the test bench we used. In control law, we proposed a fuzzy PD for controlling the system, and detailed the proposed inference rules and gains. In desktop application we show the development of a user interface so the th erapist can manipulate the system and set routines to the patient. Finally, we provide our conclusions and future work.

2 State of the art

In the field of physical rehabilitation, there are some devices responsible for activating motor functions again, such as exoskeletons, orthoses, and therapeutic mechanisms. This section will detail the prototypes presented by various researchers.

The prototype published in [3], LOKOMAT is an exoskeleton that uses pneumatic pistons on a treadmill and a pulley system, which is responsible for supporting the patient's weight allowing the legs to have free movement. The hip and knee are also actuated by pneumatic pistons, allowing adjustment to different lengths of the lower and upper extremities for different patients' morphology. This product is currently marketed and intended for rehabilitation centers to treat patients with spinal cord injury problems and accidents. However, although LOKOMAT is considered by many to be the pinnacle of lower extremity rehabilitation treatment systems, its high cost and need for specialized facilities make it unfeasible to implement in small rehabilitation centers.

At MIT (Massachusetts Institute of Technology) a system dedicated to ankle rehabilitation was proposed [4]. The prototype is a robot with three degrees of freedom weighing 3.6 Kg. The robot has four types of movements: dorsiflexion, plantar flexion, inversion, and eversion, allowing 25° of flexion in dorsiflexion, 45° in plantar flexion, 25 ° in inversion, and 20° in evers ion. As well as a maximum of 15° of rotation.

The system proposed in [5] is 3 degrees of freedom system, which considers the biomechanics of the knee. The authors use linear motors, which avoid the use of pneumatic or hydraulic installations. The system was designed for walking rehabilitation, which requires the patient to be standing, since some patients, after undergoing surgery, cannot remain in such position.

In recent years, the system proposed by [6] proposes the use of a system with one degree of freedom for rehabilitation of the tibia. The system uses a pulley and a lanyard to mobilize the tibia and support the femur, the patient's leg is supported by Velcro cloth straps. The system uses torque control to ensure that the force applied to the system is adequate to move the patient. This s ystem is simple; however, it is only capable of mobilizing the tibia and therefore is not appropriate for a complete rehabilitation of the limb from the hip.

In [7] was proposed an exoskeleton that is used to give paraplegic people the opportunity to walk again. The exoskeleton is actuated by brushless motors, which allow the system to move freely. The control, logic, and voltage source are inside a backpack that holds the links that connect to the patient. However, the developers also ignored the biomechanics of the knee, considering it as a hinge. In turn, the layout of the system forces the patient to remain standing all the time, which again limits the impl ementation with those patients who cannot do so during their first phases of rehabilitation.

2 Development

2.1 The lower extremity and its physiology

The knee is one of the most used joints in the human body, being essential to move from one place to another. Contrary to popular belief, where the knee is considered to function as a hinge, the movement of the knee is more bio-mechanically complex. The knee can have a flexion that varies from 0° (measured from the femoral aspect) to 120° to 160°, depending on the hip position [8]. During open and closed chain movements of the knee, the ligaments undergo modifications due to the morphology. Figure 1 shown the main types of ligaments that alter their length due to limb movement.

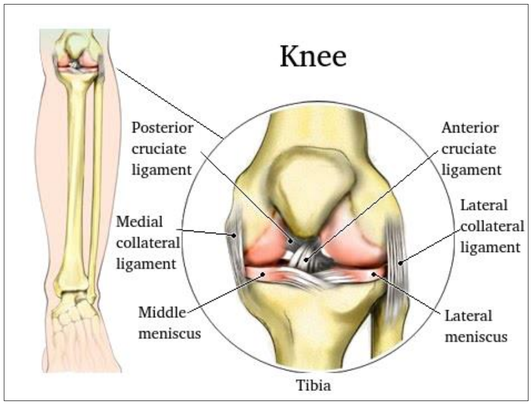

Fig. 1. Knee articulation.

The Posterior Cruciate Ligament (PCL) prevents the tibia and femur from sliding backward. The anterior cruciate ligament (ACL), like the posterior cruciate ligament, prevents the tibia and femur from sliding forward on each other. Both ligaments together offer stability to the knee when performing activities such as climbing stairs, running, or going up a very steep slope. The medial collateral ligament allows the union of the femur with the tibia. The lateral collateral ligament prevents the knee from moving laterally and genu varus between the femur and tibia. Finally, the menisci are fibrocartilages responsible for stabilizing the joints of the femur and tibia, preventing friction between the bones, and serving as a mechanical stop against sudden movements between the two segments.

During the closed chain kinetic exercises, when the knee is flexed, the posterior cruciate ligament changes its size, presenting a lengthening. When the ligament can no longer stretch, the head of the femur slides over the tibia to provide flexion, as seen in Figure 2. When the opposite movement is performed (stretching) the anterior cruciate ligament lengthens to the point where the head of the femur requires a recoil on the tibia to engage correctly, Figure 3 shows the effect of this movement [8].

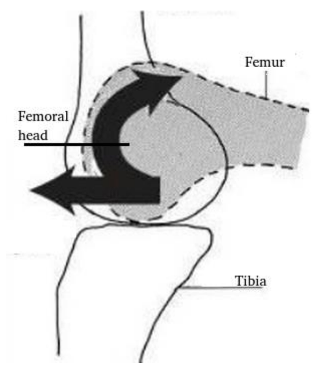

Fig. 2. Behavior of the femoral head (flexion)

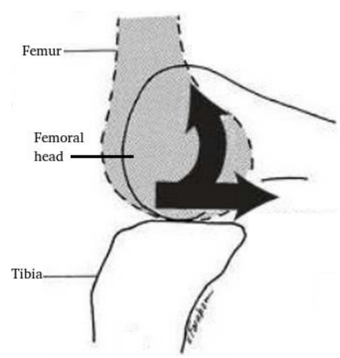

Fig. 3. Behavior of the femoral head (Stretching)

As can be seen in Figure 3, the movement of the knee causes displacement between the femur and the tibia, causing a lengthening of the tendons.

In Figure 2 and Figure 3, the movement generated in this joint can be observed in more detail. As the knee is flexed in the sagittal plane, the displacement causes a slight lengthening of the knee by approximately 2 cm as the patella engages in the new position [9]. The morphology of the knee is commonly overlooked by designers of rehabilitation systems, which could cause internal injury to the patient by presenting unnecessary compression. One of the main objectives during the design of the system is to offer a mechanism that avoids this.

2.2 Proposal

Based on the state of the art and the knee morphology, the construction of a system with two degrees of freedom has been suggested, that allows the therapist to mobilize the patient's limb in any desired degree of flexion. An open kinematic chain is chosen, as it provides mobilization to the member in any position that the therapist establishes. One advantage that an open kinematic chain offers advantages over a closed kinematic chain, it is no limitation by the actuator mechanism, however, it requires more attention to the required torque since there is no mechanical base to support the links.

The movements that the rehabilitator will perform are flexion and extension, which are used for the rehabilitation of the knee joint, as well as the hip joint. Figure 4 shown a proposed two-bar system to mobilize the patient's lower extremity. Link **L1** represents the femur link, which will be attached to the hip while the other end will support the tibia. Link **L2** is the link of the tibia, responsible for supporting this part of the member and in turn the foot.

Fig. 4. Link System

2.3 Patients

Regarding the design of the system's links, it is necessary to know the profile of the average Mexican patient, since the rehabilitator will be designed only for adult patients. Statistically, the patients over 18 years of age who present shorter height ar e Mexican women, with an average of 155.3±7.1 cm. While the tallest patients are men, with an average of 168.0±8.4 cm. Regarding the weight of the patients, the shorter women had an average of 57 Kg, while the taller men had an average of 66.4±13.1 Kg, as can be seen in Table 1.

Sexo	Peso promedio (kg)	Altura promedio (cm)
Mujer	57.1	155.3±7.1
Hombre	66.4±13.1	168.0±8.4

Table 1 Weight/height statistics for Mexicans over 18 years of age [10]

Based on these data, the range of patients varies from 1.5m to 1.8m, with a maximum of 80Kg.

Therefore, it is necessary to design the links considering that they can be expanded, and capable of adapting to any limb len gth within the range considered for patients. For this, we must consider the anthropometry of the body, which indicates the size proportions of each part according to the patient's height [11]. In this case, the femur has 24.5% of the patient's total height, while the tibia has 28.5%.

For the femur link, Equation 1 is used to determine the maximum and minimum lengths.

$$F_l = 0.245(H) \tag{1}$$

where F_l is the length of the femoral link and H is the height of the patient. Using Equation 1, the maximum height of the link is determined to be 0.441m and the minimum to be 0.365m.

To determine the measurements of the tibial link, Equation 2 is used.

$$T_l = 0.285(H) \tag{2}$$

where T_l is the length of the tibia. Given this equation, our maximum and minimum lengths for the tibial link are 0.513m and 0.427m. Rounding the measurements of both links, it is established that both can be extended within a range of 0.3m to 0.5m.

Once the height is obtained, it must be considered that, as a passive therapy will be carried out, the patient does not exert force to perform the routine, for this reason, the rehabilitator's transmission system must support the dead weight of the limb, plus the weight of the links on which it rests.

For this, we use table 2, which summarizes the study carried out by [12], where the masses of each limb of the body are determined. Each part of the lower limb is indicated as a percentage of the patient's weight.

Part	%Weight (Patient)
Femur	10.3
Tibia and foot	5.8
Lower limb (total)	16.1

Table 2 Percentages of the total weight in parts and sections of the human body

With these considerations into account, the link (femur and tibia support) and the transmission system for the lower extremity rehabilitator were developed and is explained more detailed in section *Gearbox*.

2.4 Links design

Figure 5 shows the link designed to support the patient's leg, along with its essential parts.

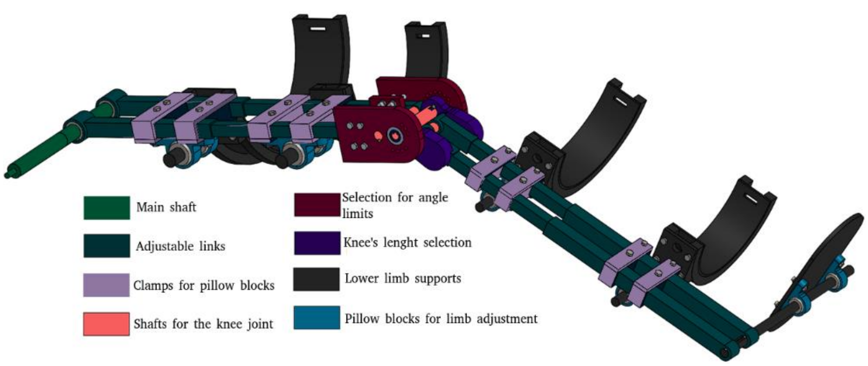

Fig. 5. System of links designed for the rehabilitator

The link has four supports for the patient's femur and tibia, at the end of the link is also the fifth support used for the sole. The pillow block bearings give mobility to the shafts, and these can better accommodate the morphology of the patient's leg. The pillow block bearings are attached to the lower extremity using clamps that can move throughout the entire link structure, and allow adaptation to the size of the patient's extremity. The main shaft is connected directly to the transmission box, which will deliver the torque to lift the femur and support the entire weight of the patient's limb and link; The knee support shaft is where the motor will actuate the tibia directly.

Regarding the union of the tibial link with the femur, a coupling system was developed that allows the knee to move and better fit to its biomechanical movement. Figure 6 shows the designed joint system in greater detail.

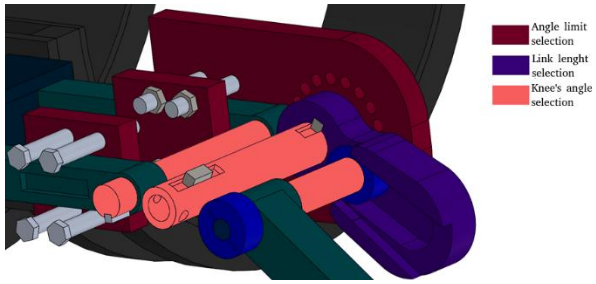

Fig. 6. Knee's joint mechanism in the system

The knee joint was developed to offer two degrees of mobility during therapy. The shaft that joins the femur link and the tibia can move through the slot of the knee-length adjustment, which allows a maximum displacement of 3cm, in turn, this adjustment can rotate up to a maximum angle of 27°. The rotation of the knee-length adjustment can be physically limited by a circular matrix of holes that have been implemented on the plate where it is held, so physical columns can be inserted in this part and prevent the rotation of this section.

This articulation added during the redesign phase presents an advantage of the system over some commercial therapeutic mechanisms where that are not considered. This mechanism allows the links to be better coupled to the structure of the leg, increasing its size and modifying its angle at the request of the limb. The sliding of the system will be carried out using pillow block bearings on the rails of the system with the part that resembles the number eight. Both are connected by a shaft that allows uniform movements throughout the mechanism.

To determine the stresses to which the system will be subjected, and whether there will be a rupture during its use, a static analysis was performed. For this, the maximum weight of the patient, 80kg is considered, the maximum height of 1.80m as was mentioned in the section *patients*, and the patient's limb is fully extended were considered the maximum torque. This configuration would require the greatest torque to move the leg and, in turn, would exert the greatest effort on the clamping axes. Figure 7 shows the Von Misses stress analysis, simulated in SolidWorks Simulation.

Initially, it was established that the five-link supports (two for the femur, two for the tibia, and one for the sole) would be made of 6063-t6 aluminum due to its easy commercialization and implementation in medical systems [13]. All the shafts were proposed to be made of AISI 1020 to avoid rupture by the cutting force during operation. With this, the density and mass of the system were obtained, plus that of the patient, we performed a static analysis.

As can be seen, the maximum stress in the shaft of the femur is equivalent to 163Mpa. Figure 8 shows the safety factor analysis, which allows us to know if the system will be able to support the load for which it is being designed.

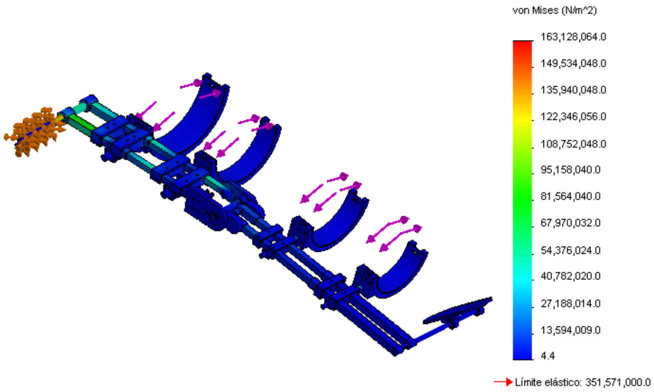

Fig. 7 a) Von-Misses stress

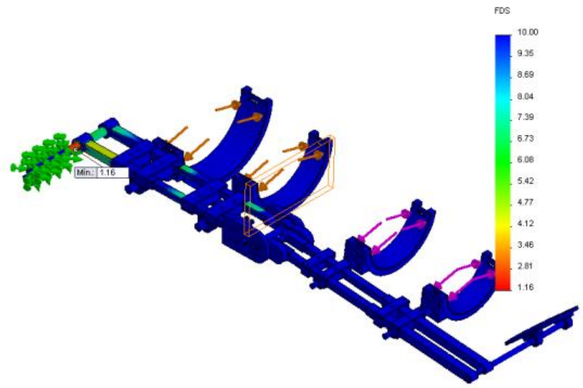

Fig. 7 b) Safety factor

In this case, the minimum factor of safety is 1.16, and it is again in the shaft of the femur. This means that the system supports up to 16% more than the maximum load (remembering that the maximum load is the weight of the patient's leg, plus the weight of the link) to which it is subjected, so it could be considered sufficient for the tests, with healthy volunteers.

2.5 Gearbox

The objective is to offer a rehabilitation system whose actuators are electromechanical, so the rehabilitator would not depend on pneumatic or hydraulic sources.

By using the dynamic model of an open chain with three degrees of freedom with friction, the torque surface required for the system was obtained according to its parameters (Table 3). With these surfaces, we were able to propose two motors.

Figure 8 and Figure 9 show the torque surfaces for the femur and tibia link, respectively. These indicate how much torque is required for each possible link configuration during a rehabilitation routine.

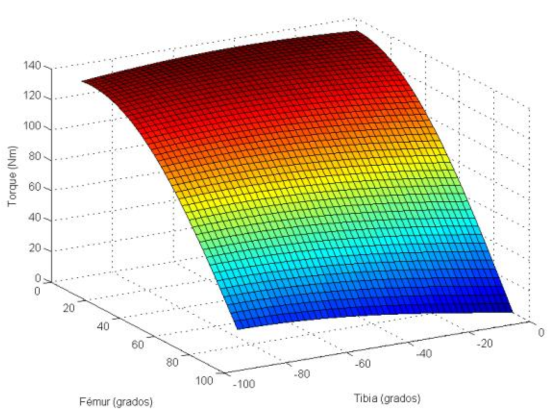

Fig. 8 Torque surface for femur link

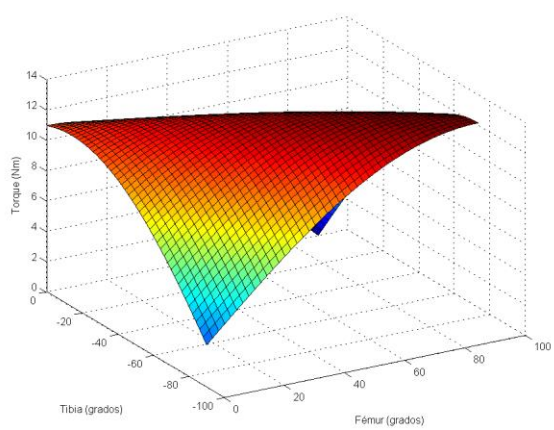

Fig. 9 Torque surface for tibia link

For the tibial motor, a maximum of 14Nm is required, while the femur link requires 133.68Nm. The actuators selected for their low cost and easy acquisition are the wiper motors that offer up to 18Nm (Figure 10).

Fig. 10 Wiper motor

For the tibial link, a single Wiper motor is enough to actuate it, but in the case of the femur, it is necessary to use a transmission box that increases the torque delivered by the motor. Equation 3 is used to determine the necessary ratio that the gearbox must deliver to achieve this.

$$r = \frac{T_{required}}{T_{delivered}} = \frac{133.68}{18} = 7.426 \approx 8 \quad (3)$$

In that case, it was decided to round the ratio to 1:8, due to the ease of obtaining elements with even ratios and in whole numbers. For the 8 ratios, it was also decided to divide it into 3 ratios of 1:2, to obtain the same torque gain. In this case, the transmission box was made using a sprocket chain system due to its ease of acquisition. Figure 11 shows the block diagram of the transmission box.

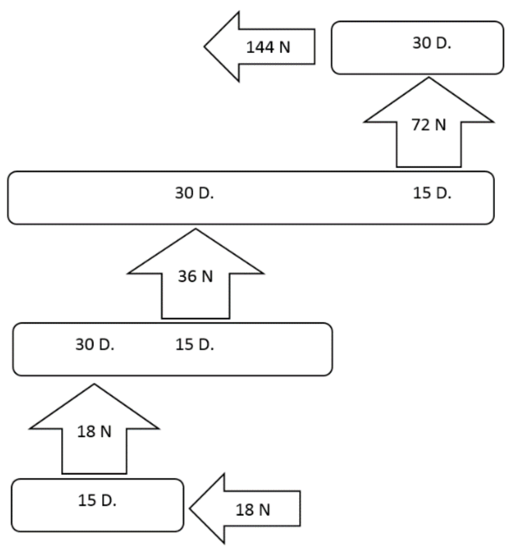

Fig. 11. Block diagram for the gearbox

With a ratio of 1:8, a final torque of 144Nm is obtained, which is enough to activate the femur link. Figure 12 shows the designed transmission box.

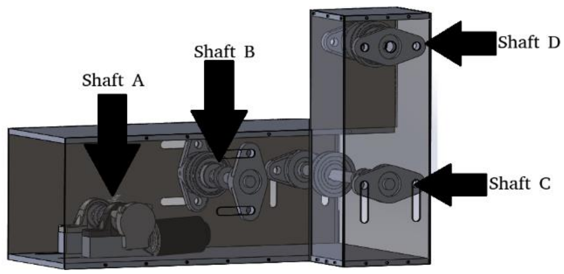

Figure 12 Gearbox design

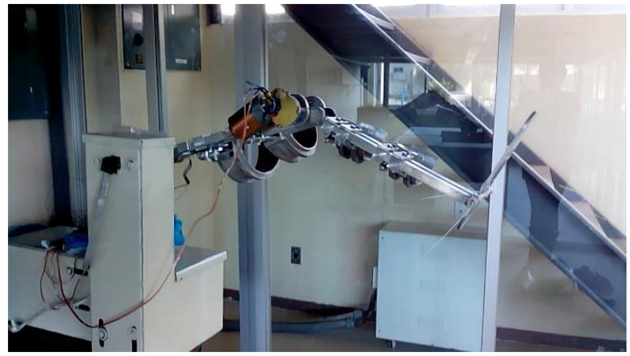

Figure 13 Links with gearbox

3 Control law

The problem faced setting a control law for the rehabilitation system is it's the non-linear nature. Because of this, a PD control cannot be used as the gains will not work at certain points of the trajectory. One advantage the system has, is the lack of a gravity compensator, as both gearboxes from the wiper motors support the femur and tibia links even without being powered.

3.1 Fuzzy PD

A fuzzy PD was chosen to solve the non-linear nature of the system [14]. The fuzzy PD allows setting a range of gains for the controller, which will be varying according to the position error. We use a decoupled control fuzzy PD control, as the femur and tibia do not have the same workspace, so we can tune the membership for each one individually. The control diagram is shown in Figure 13.

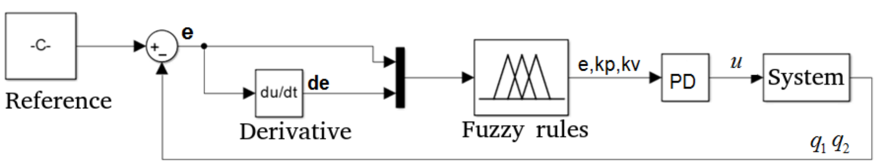

Figure 13 Control Diagram

For the fuzzy membership functions, we opted to use triangular functions that will vary the gains based on the current error from the position and the velocity of both links. Figure 14 shows the used membership functions, in this case, the maximum value for both PD, and Table 3 details the error range of the functions, where N is negative error, Nu is null error and P is positive error.

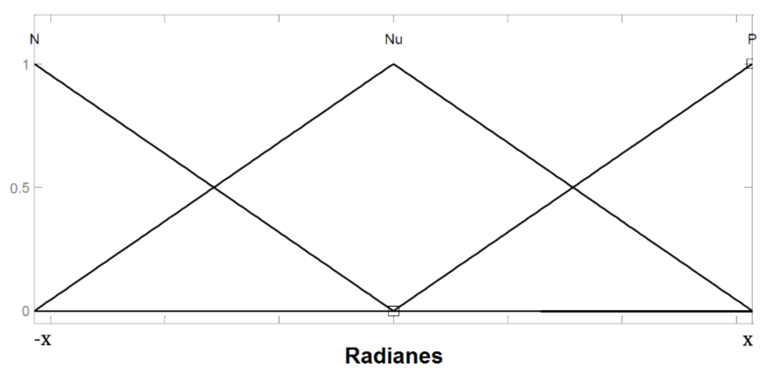

Fig. 14. Membership function

The established inference rules are shown in Table 5.

Link	Error type and gain	Range (radians)
Femur	Position (k_p)	-π, π
	Velocity (k_v)	-π/2, π/2
Tibia	Position (k_p)	-π/2, π/2
	Velocity (k_v)	0,3

Table 4 Trajectory result for the tibia link

e/de	N	Nu	P
N	Nu	N	N
Nu	P	Nu	P
P	P	P	Nu

Table 5. Inference rules

Although the fuzzy control allows the system to reach the reference, we also opted to implement a trajectory algorithm based on a Bézier curve, which was proposed in [15]for smooth motor control velocity. In this case, as the patient is in rehabilitation, it is imperative to avoid sudden changes of position or velocity to avoid injuries. This Bézier curve ensures a smooth beginning of the motion, a linear and constant movement, and a smooth end from one position to another. This also allows us to determine the total time of movement in seconds, which the medic can use to regulate the motion avoiding injuring the patient during the therapy. The equation that determines the Bézier curve is shown in (3).

$$r(t) = (rf - p)\left(\frac{t-T_1}{T_2-T_1}\right)^5 \left(r_1 - r_2\left(\frac{t-T_1}{T_2-T_1}\right) + r_3\left(\frac{t-T_1}{T_2-T_1}\right)^2 - r_4\left(\frac{t-T_1}{T_2-T_1}\right)^3 + r_5\left(\frac{t-T_1}{T_2-T_1}\right)^4 - r_6\left(\frac{t-T_1}{T_2-T_1}\right)^5\right) + p \qquad (3)$$

Where
r is the current position of the link
r_f is the reference or desired position for the link
p is the initial position of the link
t is the actual time of iteration
T_1 is the initial time
T_2 is the desired final time
$r_1 \dots r_6$ are constants whose values are 252, 1050, 1800, 1575, 700, and 126 respectively.

Figure 15 shows the Bézier curve create for the trajectory.

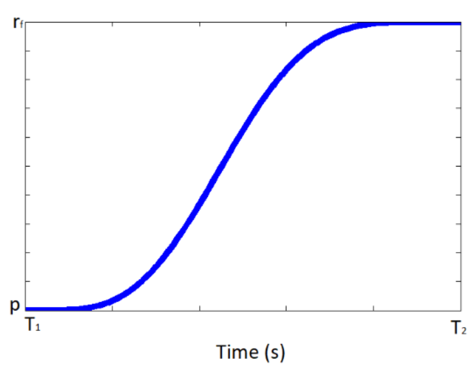

Fig. 15 Bézier curve for path creation

3.2 Test bench

Once the mechanical system was built, they were connected to the control and electrical system that had been previously designed for it. It was decided to make a fuzzy control implementing it in an Arduino DUE proposed by. Trajectory tests were carried out, without weight, to verify that the system reached the desired position in terms of the routines that were performed.

For the femur link, an ascending trajectory was made with a 90° reference. Figure 14 shows the results obtained from the test.

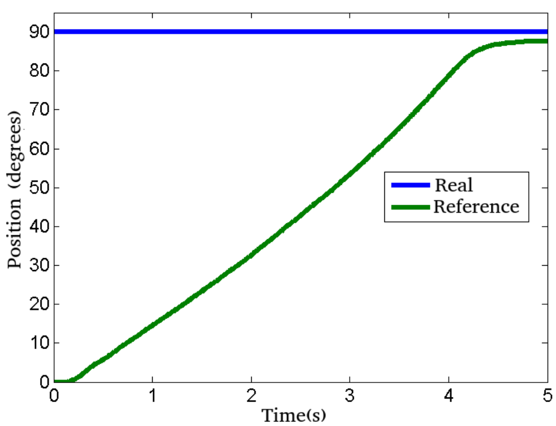

Fig. 14. Trajectory result for the femur link

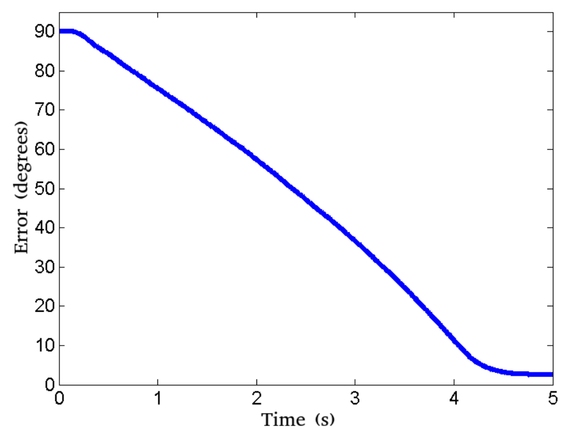

Fig. 15. Trajectory result for the tibia link

Using fuzzy control, the system has a stationary error of 2.46°, which represents a 2.7% error from the reference. Using the maximum length of a patient's limb would give us an arc of 3cm difference between the position of the foot concerning the hip. For the tibial link, a descending trajectory from 90° to 0° was performed. Figure 15 shows the results obtained by this trajectory.

As can be seen, the stationary error in the tibial link is 1.75°, which represents a 1.8% error concerning the reference. This would give an error of 1.3cm of difference between the position of the foot concerning the knee.

3.3 Framework

For implementing the system into an Arduino DUE, we opted for a methodology like Giotto [16] which implements a real-time system into embedded devices. In this case, priority levels and delay time are set for each task. A timer is set which executes the tasks when its delay wait is over. When two or more tasks are being to be performed during the same execution, the priority level is considered, so the lower priority tasks will be executed first. Table 6 details the tasks and, delay time and priority for each one.

Task	Delay	Priority
Reading	Interruption	0
Error	150µs	1
PD	2.5ms	2
Fuzzy	10ms	3
Bézier	25ms	4
Comm	100ms	5

Table 6 Tasks delays and priorities

The action of each task is detailed as follows:

- **Reading:** This task is the only one that is not timed and will be executed every time the encoder changes its value, as an interruption occurs, and the value is updated for both limbs.
- **Error:** This task calculates the error between the read positions of both links, so the PD control used them for the control signal calculation.

- **PD:** The PD control is calculated, and the control signal is applied in this task.
- **Fuzzy:** The membership functions are evaluated, and the gains are implemented in PD control.
- **Bézier:** The current position of the Bezier trajectory is calculated and used as a reference for the Error calculation.
- **Comm:** The serial communication with the desktop app is performed. The microcontroller updates the current position and signal control while checking if there is a new command to perform.

4 Desktop application

To operate the system a desktop application was developed in Visual Basic. The application allows to write the program in the proposed language, save and open routines. The application is divided into three modules: the routine editor, the teaching, and the capture.

4.1 The routine editor

In the routine editor, the therapist will propose the trajectory that each link must reach and the time that will take it. For the creation of the trajectories, we opted to develop an easy language, inspired by the G language used in the CNC machines and 3D printers. This language also allows to therapist to define the paths and times of the rehabilitation. In Table 7 the basic commands of the language and their use are detailed.

Command	Description
axx	Moves the femur link to the xx position (degrees)
bxx	Moves the tibia link to the xx position (degrees)
dxx	Waits xx seconds
txxayybzz	Moves the femur and tibia links to the yy and zz position, respectively, in xx seconds.

Table 7 Rehap language basic commands

In this case, the **a** and **b** command set all the speeds of the device to move both links, which is not recommended for treating patients. The **d** command allows waiting for certain seconds which is used to immobilize the limb in a certain position. The command **t** is the most useful as the therapist can mobilize the limb to reach a certain position in a specific amount of time. This command is also the one used for the microcontroller for calculating the Bézier function previously presented. Figure 16 shows the main window and description of the different widgets of Rehap application. In this window, the user can edit the routine, save, and open previously proposed routines in a text file. The routine will be executed from this window and allows the user to cancel it any moment if an error or emergency occurs. The interface provides a progress bar that reflects the current advance of the routine. Also, this main window provide access to the other established modes.

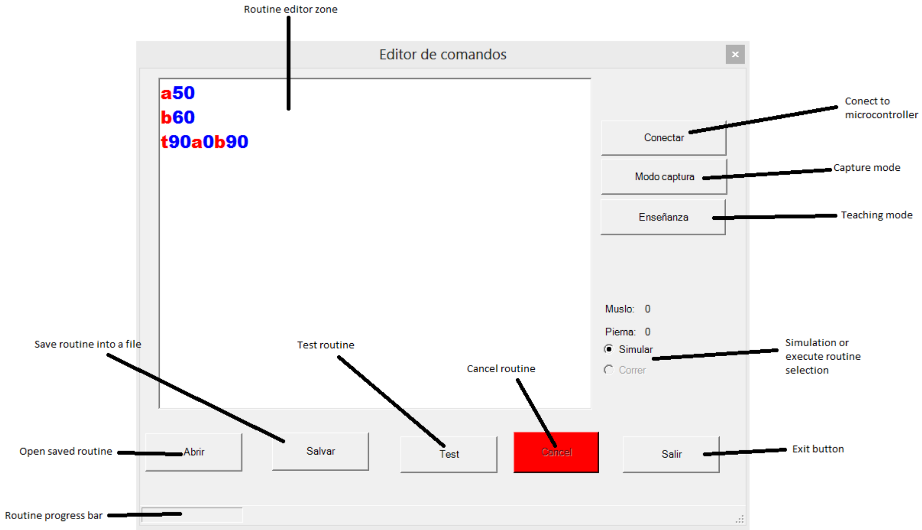

Fig. 16 Routine editor window

4.2 Teaching

In Figure 17, It is shown the teaching module developed for the Rehap application.

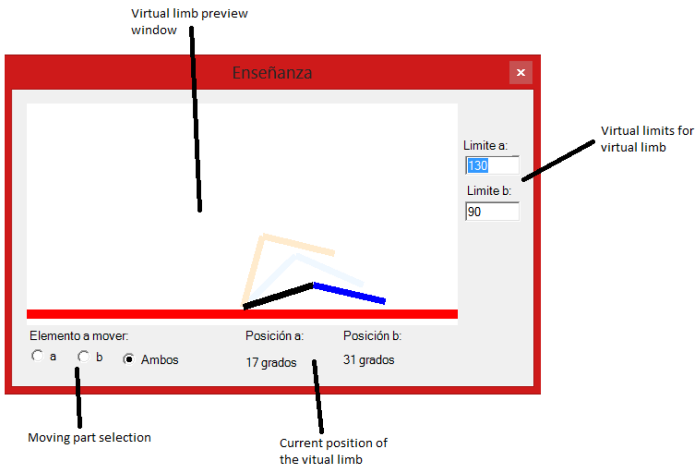

Fig. 17 Routine editor window

The teaching module has a virtual limb and two-position boxes that limits the maximum angle of each link, to prevent the therapist overpass a forbidden position that the patient is unable to reach. The teaching allows two principal functions: the first is to recreate the routine from the routine editor, which allows the therapist to evaluate if the ro utine is performed as desired. The second function is like the teach pendant from industrial robots, the therapist is allowed to mobilize the virtual limb by using the mouse and when the key enter is pressed, the position is saved into the editor, so the therapist can create a routine base more on qualitative positions than quantitative. The teaching keeps in carbon copy the last two positions of the virtual limb so the therapist know where was the latest position.

4.3 Capture mode

The last mode is used with the system turned on. In this case, a pair of sliders (shown in Figure 18) moves each link individually. The capture mode will record the code into the routine editor. This module is intended to evaluate the routine in real-time during execution.

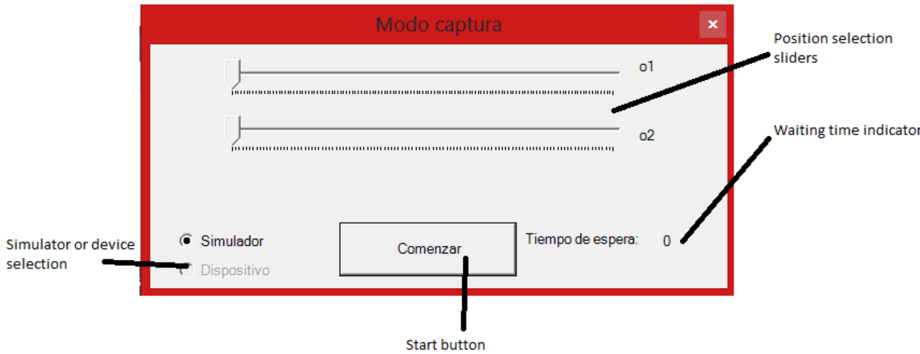

Fig. 18 Capture mode

5 Conclusion

The links of the rehabilitator were designed considering the biomechanical movement of the knee, which allows a better adaptation to the movement of the limb, avoiding hurting the patient during the routine.

The links were made from aluminum to lighten the effort and torque required. The gearbox allows increasing the torque of the selected motor with a ratio of 1:8, which powers the system electrically and does not depend on pneumatic or hydraulic systems for its operation.

In the tests that were carried out, the maximum position error was 3cm for the femur and 1.3cm for the tibia, they are considered acceptable in the range of passive therapies that are going to be established; since, although rehabilitating the patient's range of motion is a very important goal, most of the time therapists perform these routines without the help of measuring instruments, based only on the pain feedback given by the patient, and their experience.

6 Future work

The prototype will undergo two stages of evaluation. The first will be a test with mannequins which will help to evaluate different control laws. Subsequently, the second stage will be to work with patients and make a test bench to evaluate the efficiency of the rehabilitator.

For the control, more controls are being evaluated to compare the efficiency of each one and determine which fits the best into the therapy.

Bibliography

[1] World Health Organization, “Rehabilitation WHO,” *https://www.who.int/news-room/fact-sheets/detail/rehabilitation*, Aug. 2022.
[2] I. M. Dávila Castrodad *et al.*, “Rehabilitation protocols following total knee arthroplasty: a review of study designs and outcome measures,” *Ann Transl Med*, vol. 7, no. S7, pp. S255–S255, Oct. 2019, doi: 10.21037/atm.2019.08.15.
[3] J. Saso, G. Colombo, and M. Morari, “Automatic Gait-Pattern Adaptation Algorithms for Rehabilitation With a 4-DOF Robotic Orthosis,” *IEEE TRANSACTIONS ON ROBOTICS AND AUTOMATION*, 2004.
[4] A. Roy *et al.*, “Robot-Aided Neurorehabilitation: A Novel Robot for Ankle Rehabilitation,” *IEEE TRANSACTIONS ON ROBOTICS*, pp. 569–580, 2007.
[5] M. Lyu, W. Chen, X. Ding, J. Wang, S. Bai, and H. Ren, “Design of a biologically inspired lower limb exoskeleton for human gait rehabilitation,” *Review of Scientific Instruments*, vol. 87, no. 10, p. 104301, Oct. 2016, doi: 10.1063/1.4964136.
[6] H. Rifaï, S. Mohammed, K. Djouani, and Y. Amirat, “Toward Lower Limbs Functional Rehabilitation Through a Knee-Joint Exoskeleton,” *IEEE Transactions on Control Systems Technology*, vol. 25, no. 2, pp. 712–719, 2017, doi: 10.1109/TCST.2016.2565385.

[7] O. Harib *et al.*, “Feedback Control of an Exoskeleton for Paraplegics: Toward Robustly Stable, Hands-Free Dynamic Walking,” *IEEE Control Syst*, vol. 38, no. 6, pp. 61–87, Dec. 2018, doi: 10.1109/MCS.2018.2866604.
[8] G. McGinty, J. J. Irrgang, and D. Pezzullo, “Biomechanical considerations for rehabilitation of the knee,” *Clinical biomechanics*, vol. 15, no. 3, pp. 160–166, 2000.
[9] J. Blackburne and T. Peel, “A new method of measuring patellar height,” *J Bone Joint Surg Br*, vol. 59-B, no. 2, pp. 241–242, May 1977, doi: 10.1302/0301-620X.59B2.873986.
[10] B. E. Del-Rio-Navarro *et al.*, “Mexican anthropometric percentiles for ages 10--18,” *Eur J Clin Nutr*, vol. 61, no. 8, pp. 963–975, 2007.
[11] R. Drillis, R. Contini, and M. Bluestein, “Body segment parameters,” *Artif Limbs*, vol. 8, no. 1, pp. 44–66, 1964.
[12] C. E. Clauser, J. T. McConville, and J. W. Young, “Weight, volume, and center of mass of segments of the human body,” 1969.
[13] T. Newson, “Stainless steel--A family of medical device materials,” *This article first appeared in Business Briefing: Medical Device Manufacturing \& Technology*, 2002.
[14] Timothy J. Ross, *Fuzzy Logic with Engineering Applications*, 3rd ed. Wiley, 2010.
[15] Francisco Beltran-Carbajal, *Advances in Vibration Engineering and Structural Dynamics*. InTech, 2012. doi: 10.5772/3421.
[16] T. A. Henzinger, B. Horowitz, and C. M. Kirsch, “Giotto: a time-triggered language for embedded programming,” *Proceedings of the IEEE*, vol. 91, no. 1, pp. 84–99, Jan. 2003, doi: 10.1109/JPROC.2002.805825.