ISSN 2176-5480

22nd International Congress of Mechanical Engineering (COBEM 2013)
November 3-7, 2013, Ribeirão Preto, SP, Brazil

DESIGN OF A KNEE ANKLE ROBOTIC EXOSKELETON TO INDUCE PERTURBATIONS DURING GAIT

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Abstract. *This work presents the mechanical project of an exoskeleton that will serve as a tool for the human control study. Different types of lower limb exoskeletons have been proposed in the recent literature. However, most of them remained in the laboratory while only a few have found real-life applications with strict exclusion criteria. Based on the current state of art, it is proposed the design if an exoskeleton that aims at perturbing gait at specific instants in order to analyze the mechanisms of gait control. The project begins with the definition of requirements as well as the construction and use restrictions for the exoskeleton. Some of these restrictions are due to the fact that it will be worn by a healthy person and it will apply controlled perturbations on the knee and ankle joints of the user during gait in order to analyze the human control response. These perturbations represent inertial, viscous and elastic loads. Therefore, the exoskeleton should be active in such a way that the actuators can replicate the different impedances. In addition, it is important to note that the perturbations will be applied during a few instants during the gait cycle and during the rest of the cycle the user should walk normally, as if there were no exoskeleton. Therefore, the controller should be able to measure the interaction force between the user and the exoskeleton in order to implement a control mode that makes the exoskeleton transparent to the user. This work will present the mechanical and electrical design of a knee-ankle exoskeleton along with preliminary results of a perturbation study using the exoskeleton. These results could be useful to propose and validate models for gait control. Moreover, this exoskeleton prototype will be useful to plan the conceptual design of an exoskeleton with functional compensation to be used by disabled people. This can be seen as the ultimate motivation of this work, although its current scope is to build a gait study tool.*

Keywords: *biomechatronics, exoskeleton, lower limb, gait control, gait perturbation*

1. INTRODUCTION

The control that the human body performs during human gait is still not fully understood by the scientific community. There is a hypothesis that there is a central pattern generator (CPG), located in the spinal cord, which provides control signals gait (ROSSIGNOL S., 2006), but the form of control applied, if is a force control or position control, is still unknown.

The leg performs different functions during a gait cycle. In support phase it should bear the body weight and provide an impulse to the ankle. In the swing phase the leg should bend and move forward with a distance between the foot and the ground enough to not touch it. At the end of the swing phase the leg should touch the ground with the heel, to start the next cycle.

A greater understanding of body control during gait will bring benefits both in the medical field, assisting the assessment, diagnosis and prognosis of neuromuscular diseases and the development of therapies, such as in the context of the control engineering and robotics, e.g. in project development based on biomimetic human motor control.

The term "exoskeleton" is used to define in biomechanics as a set of rigid segments and joints that are applied in parallel to biological members, conferring upon them, among other characteristics, an amplification of strength and structural stability.

According to the state of the art published in 2008 (HERR, 2008), the exoskeleton is defined as a mechanical device that is essentially anthropomorphic and it is "dressed" by the operator, being tailored to his body and working together with his movements. Therefore, the exoskeleton is in physical contact with the person, allowing transmission of power and various control signals (Kazerooni, 1990).

There are several classifications that can be applied to exoskeletons. The main ones are divided according to the type of actuation (electrical, hydraulic or pneumatic), to the region of the body (upper or lower), to the portability and to the field of application. There are not-portable exoskeletons that are fixed to an external base (AF RUIZ, 2009), in which there is the need to reduce weight and space occupied. The portable exoskeletons are uprightly supported by the user, exerting reaction forces on the person or on a mobile device such as a wheelchair. With respect to scope, they are divided into rehabilitation, functional support, research and increased capabilities of users.

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The first active exoskeleton was developed in the 1960s for the purpose of this last application (Zoss, 2006), named Hardiman. The University of Berkeley also developed an exoskeleton to amplify human power, called BLEEX.

The exoskeletons for functional support and rehabilitation are intended for people who have suffered diseases, neuromotor injuries and elderly. This type of equipment assists or amplifies the movements of the patient. The main bottleneck of these systems is the identification of the intention of the patient. For this, there are techniques to measure the residual motion, such as measuring the interaction forces with the device and electrophysiological measurements – electromyography (EMG), electroneurography (ENG) or electroencephalography (EEG). Furthermore, these patients may not load weight, limiting the mass of the exoskeleton (KONG and JEON, 2006).

The application of this type of exoskeleton is increasing in Brazil due to the fact that Brazilians are aging. In 2003 there were 17 million Brazilians over 60 years. In 2006, three years later, this number rose to 19 million (IBGE, 2009). This indicates the increase of people with functional incapacity in physical mobility. Other studies indicate that 14.5% of the population has some type of disability, with 27% of this value corresponds to the motor disability (IBGE, 2012). The motor disability has different causes. Among them, it may be cited: cerebral palsy, cerebral vascular accident (CVA) and spinal cord injury.

The therapy treatment for people with motor disability due to stroke is based on passive repetitive exercises (BÜTEFISCH, HUMMELSHEIM, et al., 1994) or active (KRAKAUER, CARMICHAEL, et al., 2012). These exercises can be supported and / or performed by active exoskeletons (exert force on the user through actuators) (KREBS, HOGAN, et al., 1998) and (LENZI, VITIELLO, et al., 2011).

According to (HERR, 2008), understanding the biomechanics of human gait is crucial for the development of exoskeletons and active orthoses for lower limb. Thus, as explained earlier, the objective of this work is to build a tool for this study.

The challenges encountered for the construction of an exoskeleton are mainly related to the human interface. May be mentioned especially: safety, comfort, stability and compliance (for the case where the user wishes to override its commands on the exoskeleton) (KREBS, HOGAN, et al. 1998). To supplement these needs, the mechanical design must be compatible with the controller to meet the requirements of strength and movement (impedance control) (HOGAN, 1985).

2. GOALS AND DESIGN REQUIREMENTS

The main goal of this work is to design and build an active exoskeleton that will be used to analyze the knee and ankle function during gait. This tool should be able to apply controlled torques to the joints of the leg during gait. The torques applied must simulate elastic, viscous and inertial disturbance torques on the leg.

2.1 Conceptual design of an exoskeleton lower limb

Ideally, the exoskeleton for this study should not interfere in any way with the progress of the march. In order to achieve this goal, the exoskeleton should have seven degrees of freedom of the leg: three rotations in the hip, one in the knee and three other in the ankle (Herr, 2008), and have much lower mass than the mass of the leg. The alignment of the device together with the biological joints is also an important factor.

Inevitably there will be physical contact between the user and the exoskeleton, and it must be done in a comfortable interface, which enables the measurement of muscle activity and allow the patient to carry out normal day-to-day activities as walk in plane surfaces, go up and down on stairs and ramps.

2.2 Design requirements

Considering the goal of the exoskeleton and the project constrains regarding the human-exoskeleton interaction, the following design specifications were considered:

- The target users of this device are healthy people. The device will be used during gait with the purpose of inducing perturbations during normal walking.
- The exoskeleton should be adjustable and may vary the length of the rods and the diameter of the couplings.
- The functional parts of the exoskeleton should be modular in order to allow independent analysis of knee and ankle functions.
- The structure should support the sensors and actuators needed.
- A mechanical stop should be included in the actuated joints to prevent any injury to the user.
- The total mass of the exoskeleton (including structure, actuators and power source) should not exceed 10% of the mass of the person so that the exoskeleton is comfortable and does not interfere significantly in the natural gait.
- There will be two actuated degrees of freedom: knee flexion-extension and ankle flexion-extension.
- The interface of the exoskeleton with the leg should provide the desired motion and torque without hurting the patient and avoid discomfort.

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- There were two conflicting requirements: On the one hand, the device should be as portable as possible, paving the way to a fully portable assistive exoskeleton design; therefore, a careful selection of light actuators has to be made. On the other hand, the device will be used in a laboratory environment and should be capable of emulating different types of perturbations; thus requiring very fast and powerful actuators. Taking these two conflicting requirements, electric DC motors were chosen because of their versatility.

2.3 Design methodology

The project was developed in phases, following the idea of the spiral design (Kaminski, 2000), i.e., the project must go through the steps several times, reducing errors to the extent of allowable tolerance (set by the specifications).

The sequence of the project will be given according to the following steps, listed in Tab. 1 below:

Table 1. Phases of the project

1. Literature review	4. Exoskeleton Modeling
2. Concept design proposal	5. Structure design of the exoskeleton
3. Study and design of the actuation module	6. Prototype fabrication

3. RESULTS

The results in this section are not presented in chronological order of work, as the concept of spiral design makes the steps continually reworked

3.1 Project and design of the actuation module

The proposed approach to analyze human gait control is based on measuring the recovery reactions of the body after a perturbation, such as a trip or a stumble (Forner-Cordero et al, 2003). In order to simulate this situation, the exoskeleton should apply a disturbance in the swing phase of the leg, which starts at 62% of the gait cycle. According recent data provided by Herr (Herr, 2008), the joint torques and powers during gait of a male individual, 28 years old, 82 kg and 0.99 m leg length are shown in the graphs below (see Figure 1).

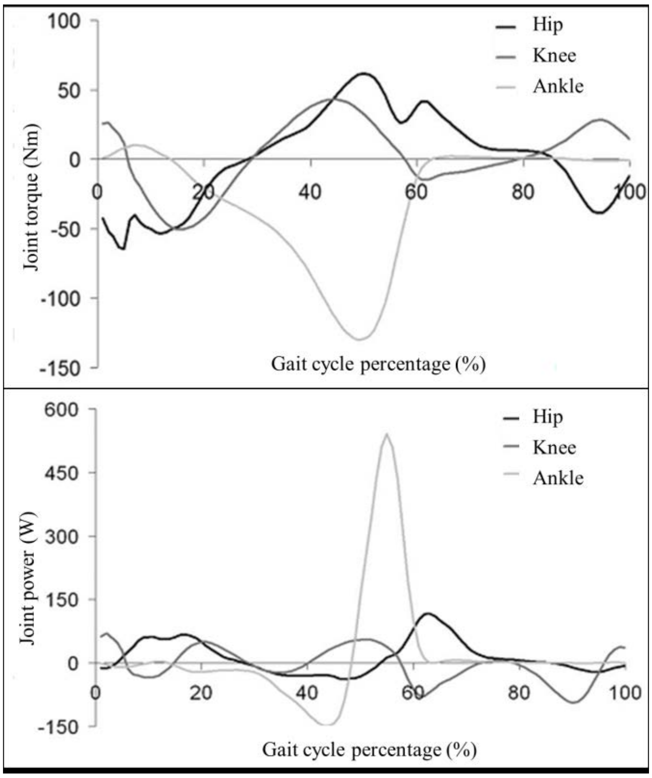

Figure 1. Torque (upper graph) and power (lower graph) in the sagittal plane on the lower limb joints: hip, knee and ankle, during gait. Adapted from (Herr, 2008).

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Initially we studied the actuation requirements for the knee in the sagittal plane. From the graphs above, the values corresponding to 62% of the gait cycle, which represent the beginning of the swing phase, are 80 W to the power and 20 Nm to the torque, approximately. The resulting angular velocity with those values is 4 rad/s, or 38.2 rpm. Since the goal is just to interfere or perturb gait and not replicate it, the minimum torque value should provide that the exoskeleton will be considered greater than that found in the study, so that the disturbance is more effective. Therefore, the minimum values adopted are those listed in Table (2) below:

Table 2 – Maximum values for actuation

Maximum torque (Nm)	Maximum power (W)
50	80

Once the operating conditions are defined, the alternatives for the knee joint actuation were considered. The choice of this activity will be done through a decision matrix, following the criteria below:

- Lower weight. A weight applied on the leg directly may alter the gait pattern, thus the lighter the better.
- Smaller size. An excessively large exoskeleton can be uncomfortable. In addition, the moment of inertia increases as the mass is placed further away from the body.
- Backdrivability. The larger the capability of the user to interfere with the movement of the actuator means a larger backdrivability the better. This criterion is related to the safety of the equipment.
- Better user safety
- Easier to adjust

3.2 Solution A: Flat brushless DC motor coupled to a strain wave speed reducer

This solution consists of a flat DC motor and a strain wave speed reducer, which can provide high reduction ratios with very narrow gearboxes. The benefits of this approach lies in the proximity between actuation module and actuated joint. Weight is minimized by the suppression of intermediate elements, also reducing the backlash and hence, the energy loss during transmission as well as positioning error. In addition, once the actuation is directly on the joint, this alternative allows the modularity between the knee and ankle joints as proposed in the desired requirements for the device.

For this purpose it was selected a flat DC brushless motor, (*Maxon EC 90 Flat)*. Despite the power of 90 watts provided by the motor, the maximum torque is up to 0.387 Nm, making it necessary a high gear ratio. A suitable choice due of its narrow geometry and high gear ratio is a strain wave gear, such as a harmonic drive.

The technical specifications of the components chosen for this solution are listed in the Tab. 3.

Table 3 - Technical specification of the motor and gear

MAXON EC90 *flat brushless*	
Maximum continuous torque (mNm)	387
Nominal speed (rpm)	2650
No load speed (rpm)	3190
Width (mm)	27.3
Mass (kg)	0,600
Harmonic Drive CSD-25-160	
Reduction	160
Maximum average torque (Nm)	75
Maximum input speed (rpm)	7500
Width (mm)	17
Mass (kg)	0,240

With this set of motor and reducer gear, the following torque and speed can be reached respectively, as shown in Eq. 1 and 2.

$$T_{output} = T_{motor} * reduction = 0.387 * 160 = 61.92Nm \quad (1)$$

$$N_{maximum\ output} = \frac{N_{motor}}{reduction} = \frac{3190}{160} \cong 20rpm \quad (2)$$

ISSN 2176-5480

3.3 Solution B: Knee joint actuation by a worm gear placed at the thigh

In this alternative, the torque of a motor placed near the hip, is amplified and driven to the knee joint by means of a worm gear. Thus, is possible to use a stronger motor that, although heavier, is placed closer the waist, reducing oscillating mass and allowing clearance to the motion of the leg.

For this solution the selected motor was the model HT03801 High Torque Series from Allied Motion, nominal power 241 W. The transmission set parameters are shown below, in the Tab. 4.

Table 4 – Design parameters of the worm gear

Crown speed = knee speed (rpm)	40	Crown outer diameter (mm)	89,13
Worm gear speed = motor speed (rpm)	1600	Distance between action (mm)	50
Number of threads	1	Worm drive pitch diameter (mm)	14,86
Number of teeth	40	Worm drive outer diameter (mm)	18,86
Module	2	Gear ratio	40
Pressure angle (degrees)	20	Worm drive torque = motor torque (N)	1,43
Crown pitch diameter (mm)	85,13	Crown torque (N)	57,2
Crown frontal circular pitch = Worm drive circular pitch	6,686	Motor mass (kg)	0,85
Crown width (mm)	21,91		

According the preliminary CAD model the weight of a worm gear set made in stainless steel would be of 1.7kg, resulting in 2.55kg plus the mass of the motor (without frames).

It is important to note that in this mechanism the backdrivability is very low.

3.4 Solution C: Knee joint actuation by cable transmission

In this alternative, the disturbance is induced by means of a bowden cable attached to the shin. The cable is driven by a pulley attached to the motor placed on the back of the user. When the motor is driven, the cable imposes a restriction to the movement. This method of power transmission does not increase the torque, so in order to reach the required torque, in addition is necessary a gearhead motor or a combination of pulleys.

The chosen actuator is a gearhead motor, (Crouzet), with 80W of power. Table 5 lists the parameters used in the analysis:

Table 5 - Design parameters for the actuation by cable transmission

Max distance between the cable and the bar thigh (mm)	150
Max angle between the cable and the bar (degrees)	60
Wire traction (N)	664
Pulley diameter (mm)	70
Pulley torque (Nm)	46,48
Motor mass (kg)	2,6

3.5 Solution D: Knee joint actuation by ball screw

In this solution, the motor would be fixed to the thigh rod, on a bearing that allow performing a rotation about an axis perpendicular to the rod. The spindle would be stuck to the engine at one end and the other end was attached to ball cage coupled to the rod of the shuttle, having also a rotation around an axis perpendicular to the bar. With decreasing length between the spindle motor and the ball cage, the rotation would be performed.

The engine selected for this application is an *Allied Motion* series *High Torque*, with 81 W of power. The spindle was selected from a catalog of NSK, using the maximum force applied on it as a choosing criterion. The parameters obtained in the analysis of this solution are listed in Table (6) below:

Table 6 – Project parameters for the cable transmitted actuation

Maximum spindle lenght (mm)	500
Maximum spindle force (N)	200
Drive torque (Nm)	0,25
Motor mass (kg)	0,33

The catalog does not provide the mass of the spindle, but for purposes of analysis it will be approximated by the mass of the steel cables of the previous solution. So the total mass, to drive on one leg, will be 1.33 kg. Regarding *backdrivability*, if the friction of the ball spindle is disregarded (good lubrication), the backdrivability is high.

3.6 Decision matrix

Table 7 shows the weighing between the proposed requirements and design criteria:

Table 7 – Weighing between criteria

Matrix for weighing							
Criteria	Mass	Size	Backdrivability	Safety	Set up	SUM	Normalization
Mass	1,00	1,00	5,00	0,33	5,00	12,33	0,50
Size	1,00	1,00	0,50	0,50	3,00	6,00	0,24
Backdrivability	0,20	2,00	1,00	1,00	2,00	6,20	0,25
Safety	3,00	2,00	1,00	1,00	3,00	10,00	0,41
Set up	0,20	0,33	0,50	0,33	1,00	2,36	0,10
					TOTAL	24,53	

Table 8 shows the decision matrix in order to choose the alternative that best fulfills the design requirements.

Table 8 – Decision matrix.

		Alternatives			
Criteria	Weight	A	B	C	D
Mass	0,5	9	1	5	7
Size	0,2	9	3	2	6
Backdrivability	0,3	1	1	9	3
Safety	0,4	3	1	0	9
Set up	0,1	9	3	1	1
FINAL		6,98	1,49	5,28	5,74

Therefore, the best alternative to acting on the knee joint is the direct coupling using the flat motor and Harmonic Drive reduction.

4. KNEE PROTOTYPE

After the selection of the motor dimensions and the *harmonic drive* up to four different modules were developed (following the spiral design concept) until the best was chosen, according to the criteria of mass, size and protection of components.

One of the project objectives was the creation of an actuation module which holds the assembly motor + harmonic drive securely, avoiding gaps that lead to unwanted efforts on the gearbox. It was considered to use bearings but the height of the actuation module would increase, increasing also the torque on the joint. Thus, a simpler model with just the bars and a low-friction element was chosen as a compromise solution for the project.

Figure 2 shows the CAD model of the fourth (and final) design of the exoskeleton actuator module, in cut view.

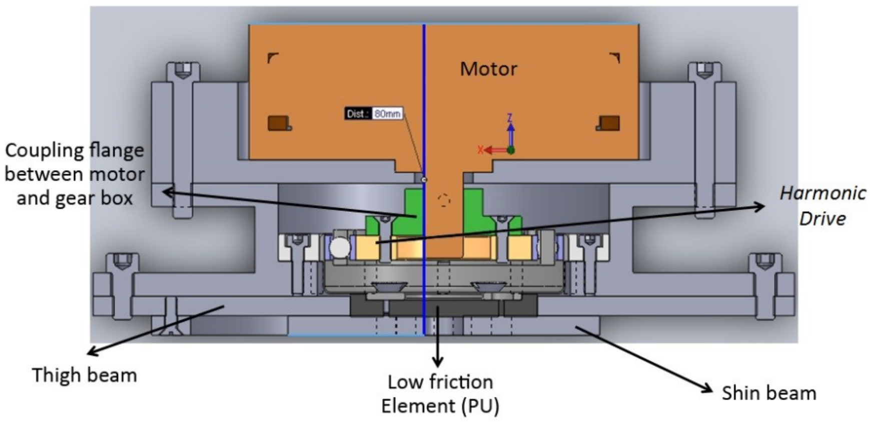

Figure 2. Exoskeleton designed in cut view, showing the parts involved.

ISSN 2176-5480

This solution was chosen because its height, 80 mm. However, the calculated mass of the exoskeleton was approximately 2.3 kilograms. In order to reduce this mass, some changes were made in the thigh beam and in the motor box and the gearbox. The mass was reduced by approximately 400g. Thus, the final set is 1.9 kilograms.

After fabrication of the parts and the couplings manufactured in carbon fiber, it was possible to assemble the knee actuation module, as shown in Figure 3.

Figure 3. Knee actuation module. On the left it can be seen the final prototype and on the right is worn by a user.

During the wearing of the first prototype some problems were observed. Due to the conical shape of the leg, with a few steps the exoskeleton moved down the leg, causing a misalignment between the human and the exoskeleton knee joints. This problem was solved with the use of a commercial hip orthosis that was attached to the thigh beam, fixed the exoskeleton on the hip and prevented the exoskeleton to move down. The *flex spline*, the portion of the harmonic drive in contact with the shin beam was causing excessive friction on it. There was also an undesired displacement of the shin beam, relative to the thigh beam and that would bring damage to the *flex spline*.

To solve these problems, rings were made on High Density Poliethylene (HDPE) and installed at the beams in order to avoid excessive abrasion and to prevent displacement of the shin rod. Carried out those changes, the construction of the prototype of the shin rod was completed, as shown in Figure 4.

Figure 4. Exoskeleton comprising the hip orthosis linked to the knee actuation module.

5. ANKLE PROTOTYPE

5.1 Design and construction of the ankle actuation module

Initially, the idea was to put the same module designed for operations in the knee joint of the ankle. But this option had two major drawbacks: the width of the module went below the foot, thus requiring the user to wear a platform to walk, increasing discomfort, and also increasing the moment of inertia of the extremities. The solution was to attach the module in the shin rod and make the transmission using a four bar mechanism.

The actuator module at the ankle used the same flat motor and *harmonic drive* chosen for the knee. Some of the problems encountered in the assembly he knee module were already solved in the design phase of this module. A support in the gearbox for the *flex spline* (an element of the harmonic drive) was designed. The fixing holes of the box were repositioned so that its outer diameter could be decreased. The material the box was made was also changed, from nylon to HDPE. Figure 5 shows the new actuation module for the ankle joint.

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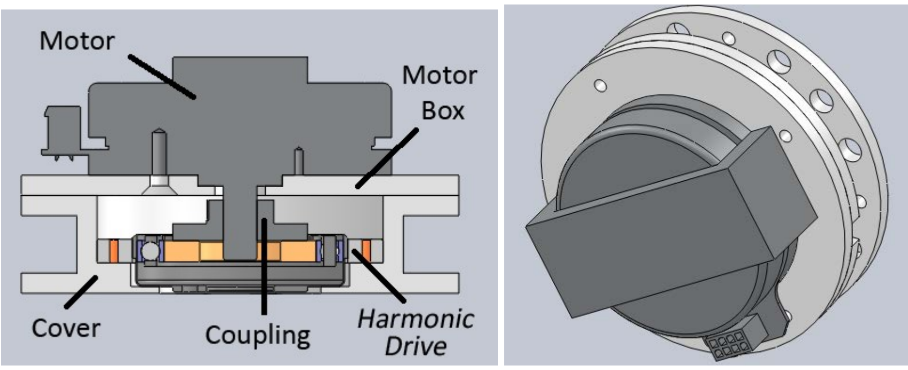

Figure 5. Ankle actuation module.

5.2 Ankle module structure

According to the constraints given in the project scope, the exoskeleton should be modular and adjustable. Therefore, it was thought in a structure that is adjustable relative to the knee module. To attach the module to the shin rod, a support was designed to allow an adjustment of the vertical position. In that support were screwed two stops, serving as mechanical constraints to movement. To integrate the ankle joint to the exoskeleton an ankle orthosis was bought (Dilepé), and coupled to the shin via Velcro straps. To transmit the movement of the module actuator to the ankle orthosis, it was designed a four-bar linkage, so as to convey the same angular variation for the ankle. Calculations were made to determine the dimensions of the rods and pins to resist the efforts in the joints. The result is shown in Figure 6.

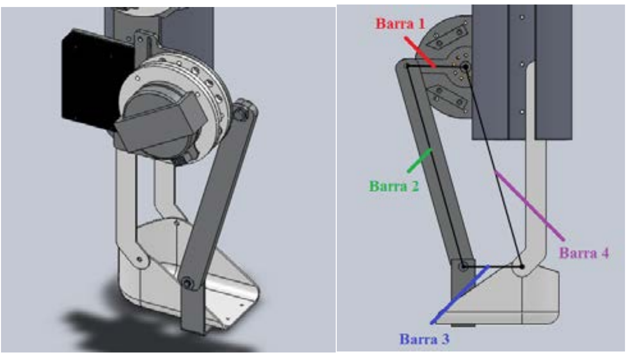

Figure 6. Complete assembly ankle module and the four-bar linkage.

5.4 Modeling of the exoskeleton

The exoskeleton is modeled after the built drawings for mechanical manufacture. The movement that the exoskeleton must accompany is flexion-extension in the joints of the hip, knee and ankle in the sagittal plane. Thus, the angular variations are disregarded in the frontal plane.

It will be considered three slashes mechanism coupled to an articulated quadrilateral. The parameters used in the model are the masses, lengths and moments of inertia of the bars, and the position of its center of mass. The modeling concepts will follow the Denavit-Hartenberg (Craig, 1989) to the kinematics of the movement, and the Lagrange method for obtaining the dynamic equations of motion. Figure 7 shows the modeled structures together with a diagram with the reference systems adopted.

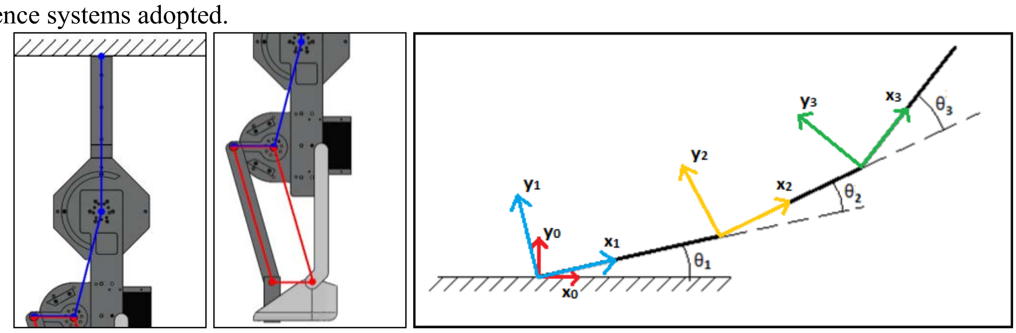

Figure 7. Representation of the three-bar mechanism and articulated quadrilateral, and the schema of references following the Denavit-Hartenberg notation.

ISSN 2176-5480

Considering only the torques carried by the engines, there are two torques: T2, applied at the knee joint, and T3, applied to the ankle joint. In this model, there is no generalized force involved in the hip joint. For each degree of freedom, represented by $\theta_1 \theta_2$ e $\theta_3$and the method of Lagrange generates a dynamic equation (3) as follows:

$$T_i = \frac{d}{dt}\left(\frac{\partial(K_1+K_2+K_3)}{\partial\dot{\theta}_i}\right) - \frac{\partial(K_1+K_2+K_3)}{\partial\theta_i} + \frac{\partial(U_1+U_2+U_3)}{\partial\theta_i} \tag{3}$$

Where K_i and U_i denote, respectively, the kinetic and potential energy of the i^{th} segment.

6. INTRUMENTATION OF THE EXOSKELETON

The measurement of the interaction forces exerted by the user on the exoskeleton, for both degrees of freedom, corresponding to the knee and the ankle angles, is needed to provide adequate motor control during human gait. The sensors selected were strain gages, attached to the shin rod and to the bar 1 of the four-bar linkage (Fig.8). To complete the required instrumentation for each measurement, a full Wheatstone bridge, an amplification circuit and an acquisition board were employed.

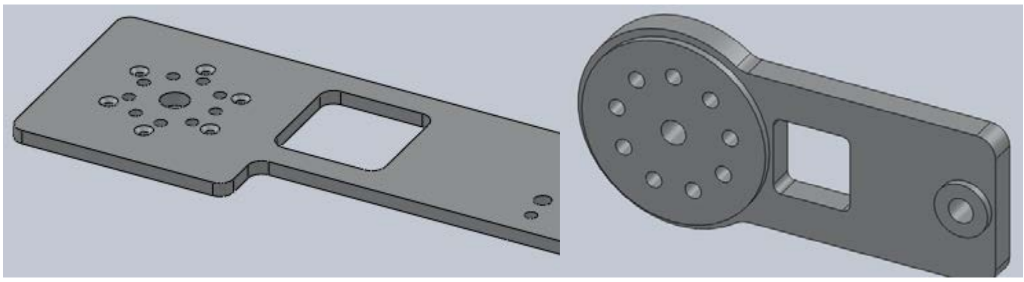

Figure 8. The shin bar (on the left) and bar 1 of the four-bar linkage (on the right).

The most suitable regions for the attachment of the strain gages on the shin rod and the bar 1 were determined through the FEM strain analysis (Fig.9). The simulations have demonstrated that higher values of deformation occur near the larger holes. Figure 10 show the location of the strain gages attached to the shin bar and to the bar 1 of the four-bar linkage.

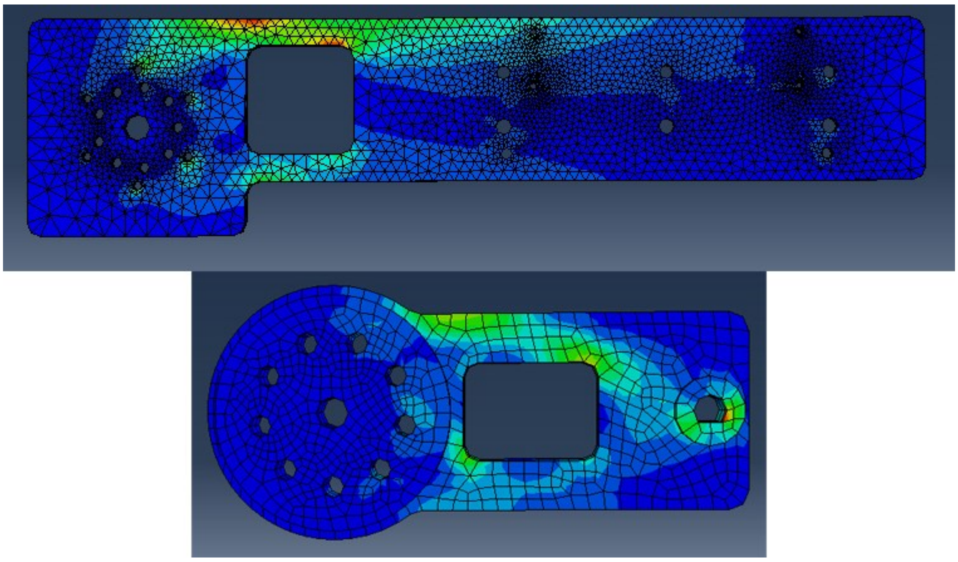

Figure 9. FEM strain analysis performed on the shin bar (top) and the bar 1 of the four-bar linkage (bottom).

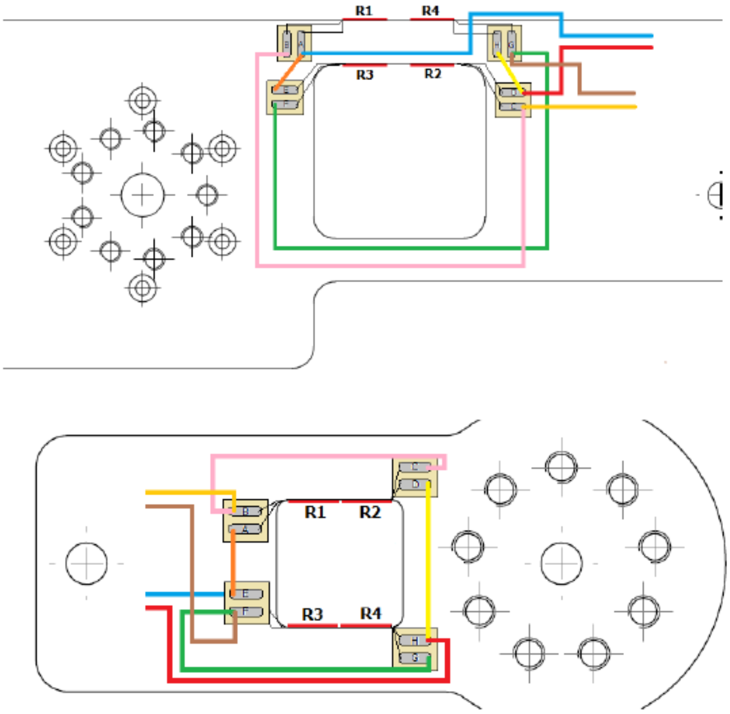

Figure 10. Location of the strain gages attached to the shin bar (top) and to the bar 1 of the four-bar linkage (bottom).

7. DISCUSSION

The design and the exoskeleton construction followed the modularity concept. Hence, not only the knee and ankle modules are quite similar, but also the two actuation modules.

When a robotic manipulator performs interaction with its environment, such as the case of the exoskeleton with the human body, it is necessary to control two types of variables simultaneously: the position and the force of interaction (GOLDENBERG, 1988). One feasible technique to achieve this goal is called impedance control.

To implement the impedance control, at least two types of input signals are required. The first type of input signal is acquired by the encoders coupled to the motor shafts. The other input signals are obtained from the strain gages attached to the shin bar and bar 1 of the linkage. The main advantages of using strain gages are the low weight and volume, easy installation, excellent linearity and excellent dynamic response.

At the current stage of development, the construction of the mechanical exoskeleton has been completed. However, in order to perform satisfactorily the function of gait analysis tool, it is still necessary to establish the actuators control, according to the modeling presented.

8. CONCLUSIONS

This work dealt with the mechanical design and the manufacturing of a knee-ankle exoskeleton, the dynamic modeling and the required instrumentation to be useful as a tool to propose and validate models for gait control. The mechanical prototype of the exoskeleton was completed in accordance with the technical specifications: it is intended to be worn by healthy subjects, it is adjustable for different users, it is modular, there is enough room for the installation of sensors without any impairment of structural parts, there are mechanical constraints that avoid possible injury to the user, and, finally, possesses two actuated degrees of freedom: flexion-extension of the knee and ankle.

9. ACKNOWLEDGEMENTS

The authors acknowledge the support provided by the Brazilian National Research Council (CNPq, grants 308244/2009-5 and 478863/2010-1) and the Advanced Rehabilitation Research Center (NAP – NARLE) at the University of São Paulo.

ISSN 2176-5480

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