International Journal of Engineering and Advanced Technology (IJEAT)
ISSN: 2249-8958 (Online), Volume-4 Issue-5, June 2015

# Implementation of a Novel Powered Ankle-Knee Prosthesis for Lower Limb Amputees and a Control Strategy Proposal

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***Abstract— In this work we present the second stage of a powered ankle-knee prosthesis design and construction for individuals with a lower limb amputation. The prosthetic leg is composed of two modules; the knee module and ankle module, they can operate independently or in conjunction. The ankle module is comprised of a unidirectional spring configured in parallel with a linear actuator. This spring is intended to store energy in dorsiflexion, and then released it to assist power plantar flexion. The knee module consists of a series elastic actuator and a linear solenoid actuator attached to the actuators transmission. Also we are presenting a control strategy of the prosthesis, using techniques of force, position and impedance.***

***Index Terms—Powered Prosthesis, Mechanical Design, Lower Limb Amputees.***

## I. INTRODUCTION

Individuals with lower limb amputation have shown to expend more metabolic energy than an individual with a healthy leg during normal walking. Transfemoral amputees expend up to 60% more metabolic energy compared with healthy subjects [1]. A transtibial amputee tends to expend 20-30% more metabolic energy in normal walking [2]. Thus, both cases of amputation tend to walk more slowly than an individual with healthy lower limbs. In addition amputees exhibit asymmetric gait patterns compared to non-amputees [3]. Currently most of the commercial prostheses available are passive prostheses. These are not able to bring positive work at phase stance, also have increased the risk of joint and back pains. Some researchers have shown that powered prostheses for lower limb are able of mimic human gait. They can provide negative and positive work in the stance phase, as wells as to improve amputees performance in a more natural gait and normal walking [4]. Ideally, a good design of prosthesis needs to have some characteristics described as: (1) be able to produce net power to the gait (2) the lowest possible energy consumption, and (3) should not exceed the weight and the height of the missing limb. In this work we are presenting the second stage of a prosthesis design developed previously [5] and in which we described the mechanical design of the overall prosthesis.

**Manuscript published on 30 June 2015.**
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*This work is supported by Technological Institute of Rehabilitation as part of the project Bio-Prosthesis.

Continuing with this research we present here a control strategy that can make possible the actuation of the powered ankle-knee prosthesis. In order to improve the performance of the prosthesis we made some changes in the original design. For example we replaced the clutch system for a linear solenoid actuator that makes possible to have a more reduced overall-length. These changes are described next in section 2. These new changes permit to have a complete prosthesis that include both modules in a lighter stiffer structure and a control unit that will permit to create leg healthy movements during normal gait.

## II. PROSTHESIS DESIGN

The Prosthetic leg proposed is divided in two sections, which are the knee and the ankle module. Both modules are able to work in conjunction and also operate independently. The purpose for which it was designed in a modular way is for a wider range of amputees have the opportunity to use this prosthetic device. Therefore it could be used by a transtibial amputee, as well as individual with a transfemoral amputation. The prosthesis was designed in the CAD software SolidWorks®. In the Figure 1 and Figure 2 we can appreciate all the mechanical components and parts of the sensors that comprise the prosthesis. The structure of both modules is made of Aluminum 6061-T6 with an anodized treatment in order to increase the resistance to abrasion and to protect it from chemical agents and environmental threats. On the other hand in order to avoid increase the weight of the prosthesis, we are considering using a 3D printer to make the structure that will store the electronics and the battery source, and will place at the sides and on the back of the both modules. The prosthesis parameter ranges was established as close as possible to the leg-human range motion during level-ground walking, as we can see in Table I.

**Table I. Physical parameters of the powered ankle and knee prosthesis modules.**

| Parameter | Ankle (Value) | Knee (Value) |
|---|---|---|
| Length | 24.5 cm | 24 cm |
| Width | 7.5 cm | 7 cm |
| Weight | 2.5 kg | 1.5 kg |
| Range of Motion | $40^0$ | $95^0$ |
| Output Torque (max.) | 135 Nm | 62 Nm |
| Peak Output Velocity | 5.88 rad/seg | 5.27 rad/seg |
| Spring Stiffness | 315 kN/m | 280 kN/m |

### *2.1 The Knee Module Implementation*

In this section we describe the mechanical components that comprise the knee prosthesis. In the normal level-ground walking and specifically in the phase stance that takes 40% of the gait cycle the torques generated by the knee presents almost linear behavior. The greatest demand of mechanical power during the gait cycle is presented in the phase stance [1]. The behavior of a spring is linear. Therefore choosing a spring with the correct stiffness the prosthesis can achieve the linear region in the phase stance and approximate to represent the torque behavior of a healthy knee in the gait cycle. The series elastic actuator (SEA) uses a brushless DC motor (mark 11) (*Maxon® EC-4pole 30,200 Watts*) with a high resolution encoder (mark 5) (Maxon® Encoder MR type ML) attached to the shaft. The actuator features transmissions comprises of timing belt pulley (mark 12), timing belt (mark 4) with a 3:1 ratio, coupled to a ball-screw (mark 3) (*NSK®,10 x 3 mm*) in series with a spring (mark 10) with a stiffness constant of 280 kN/m.

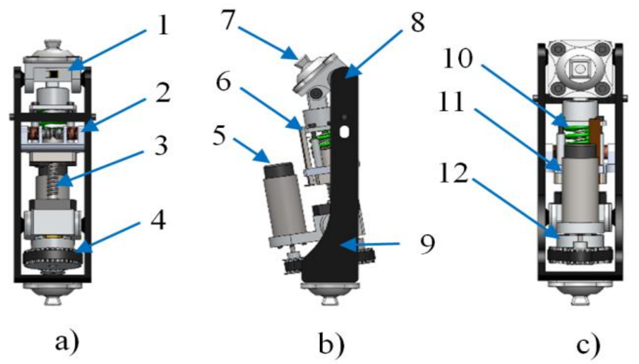


**Figure 1. CAD model of knee module with three different angle representations. a) Front, b) Side and c) Back view with a rotation angle of 0º, 45º and 90º respectively.**

A linear potentiometer (mark 6) is placed in parallel to the spring in order to measure the deflection of the spring and obtain the force applied to the prosthesis. In the first design we proposed to add a clutch in series to the SEA but then we realized that the overall-length and the weight of the prosthesis was increase significantly. Hence, we decide to redesign the model and replaced the clutch by a linear solenoid actuator (mark 2) that is attached to the transmission of the SEA. On the top of the chassis (mark 9) is place a moment load cell structure (mark 1) and also form part of the joint [mark 8] of the prosthesis. There is attached the pyramidal connector (mark 7) which will connect to the socket of the amputee as shown in Figure 1.

### *2.2 The Ankle Module Implementation*

The Ankle prosthesis is composed of an actuator in parallel with a unidirectional spring, best known as Parallel Elastic Actuator (PEA). For driving system, we used a brushless DC motor (mark 2) (*Maxon®, EC-4pole 30, 200 Watts*) that has attached to the shaft an encoder (mark 3) (*Maxon® Encoder MR type ML*). The motor drives a ball-screw (mark 10) (*NSK®, 10x3 mm*) via timing belt pulley (mark 1), and a timing belt (mark 11) with a 3:1 ratio. The ball-screw is coupled in series with low profile prosthetic foot [mark 9] (*Ossur®, Flex foot*). We used this elastic leaf spring to emulate the function of a human foot. It provides shock absorption and minimizes the ground reaction shock to the transmission. The ankle actuator incorporates two springs (mark 4) with an individual stiffness of 162 KN/m that are positioned in parallel with the actuator. The purpose to add the spring is to supplement the power output during plantar flexion. The parallel spring is unidirectional, and it is used only to provide stiffness when the ankle angle is greater than zero. As same as the knee module a linear potentiometer (mark 8) is attached to the spring to measure the spring deflection. At the top of the structure (mark 6) we placed a structure that has an array of strain gage (mark 5) an also this structure is able to attach the pyramidal connector (mark 12). Figure 2 shows a three different angle representation of the ankle module with legends.

### *2.3 Control Strategy Proposal*

We defined the control strategy as the architecture that governs the prosthesis and this is composed by different devices such as; sensors, drivers, actuators, microcontrollers and algorithms that are integrated in a single box and are able to recognize the gait patterns and commands demanded by the user. In Figure 3 we represented diagram that represents our control strategy. All information obtained by the sensors are acquired by a microcontroller and then processed in a computer algorithm. In this section we describe all devices included in this control system as well as the action related.

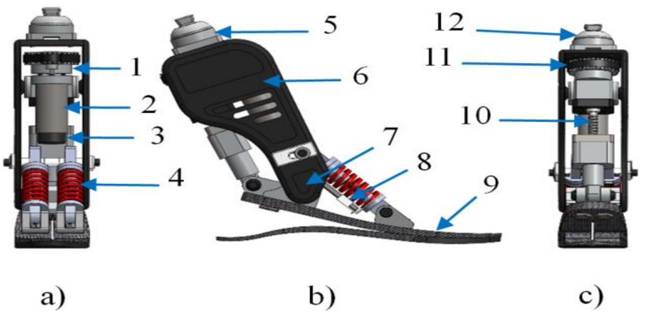


**Figure 2. CAD model of ankle module with three different angle representations. a) Front, b) Side and c) Back view with a rotation angle of 0º, 45º and 90º respectively.**

#### *2.3.1 Sensors System*

The sensing system or the recognition sensors system is comprised by devices that measure different physic magnitudes. We established The Inertial Measurement Unit (IMU) that is integrated by 3 axis gyroscopes, 3 axis accelerometers and 1 compass. By applying the mathematical Euler algorithm we calculated the yaw, pitch and roll angles. With these information it was possible to determinate the prosthesis orientation. Also we can to interpret these parameters and to estimate the action required by the individual. Electromyography (EMG) is an optional way to control the activities that performs the prosthetic device. It makes possible the interaction between the user and the prosthesis, permitting volunteer movements. Therefore it is possible to establish a more natural interface individual-prosthesis. The control of prosthesis by the EMG can be made establishing a series of commands that measure his pulse duration, amplitude and the sequence of the signal.


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The load cells are strategically positioned on the device, thus each module includes a uniaxial load cell (*Omega*®). Since impedance changes in a strain gages are very small, they were connected into a Wheatstone bridge circuit in order to increase the sensibility to voltage changes. The configuration of the bridge is a quarter-bridge with temperature compensation. We connected this output-bridge to high gain instrumentation amplifier and a 2nd order low pass Butterworth filter with a cutting frequency of 160 Hz that ensure a better signal to noise ratio. The information obtained by this sensors permits to obtain the applied force to the prosthesis and consequently to determinate the sagittal moment.

### *2.3.2 Recognition of Action*

We defined the recognition of action as daily activities of a person could make such as hiking stairs or walking over flat or inclined surfaces. Thus is essential that the prosthesis can performs at least these activities. The torque, angle movement, and cadence variations are parameters that change according to the activity developed for a healthy individual. Therefore knowing the action that an individual desires to do it will depend the type of control that will be executed.

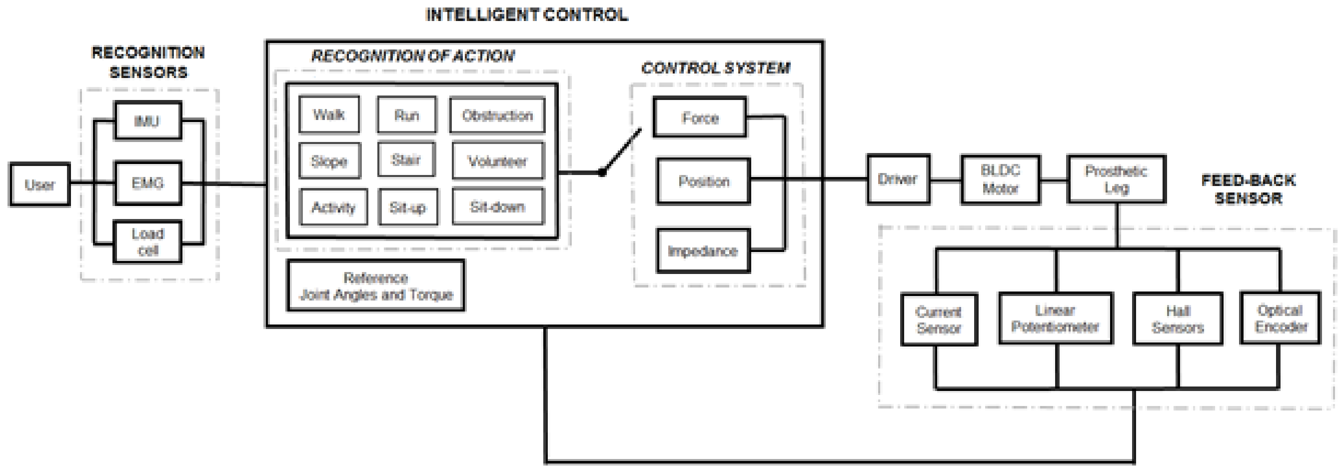


**Figure 3. Control Strategy proposed consists of three different sections.**

### *2.3.3 Control System*

The control system is comprised of a series of selectable control techniques that are activated according to the action required in the gait. We used a Maxon ® Controller ESCON 50/5 servo controller that includes a Current and Speed control. Using these controllers was possible to find out the force applying Equation 1. Where $\tau$ represents the applied torque in N/m, I represents the current through the motor in Amperes and Kt represents the constant torque of the motor (0.359 Nm/A) as shown en Equation 1. This control is applied in activities that demand higher efforts such as stance phase (push off), no matter whether is walking at level ground or running, this type of controller is always applied meanwhile the individual wants to take off the foot prosthesis from the ground. On the other hand position control ensures the moving joints with different angles. We used as position control the high resolution encoder Maxon® MR type ML 1000 CPT attached to the shaft of both motors. The pulses of the encoder signal are directly related to the joint position. The impedance control is a hybrid control that permits the interaction between the individual and the environment; it comprises a position and a force control. This control is essential and provides to the prosthesis a joints impedance variable. This variable impedance is related with motor parameters such as damping and inertial behavior.

$$\tau = K_{\tau} \times I \tag{1}$$

Once we know the torque we can calculate the mechanical power as:

$$P_m = \tau \times \omega \tag{2}$$

Where $P_m$ is the mechanical power produce by the motor is, $\tau$ is the torque and $\omega$ is the angular velocity.

### *2.3.4 Feedback Sensors System*

The feedback sensors are a group of sensors that are synchronized with the control system in order to manipulate the prosthesis movements. We used two shunt resistors as motor current sensors due to their low cost good stability and easy implementation. On the other hand a Hall Effect sensor was attached to shaft motor in order to control the speed.

## III. RESULTS

A real view of the prosthesis designed and new mechanical changes are shown in Figure 4. We can appreciate the complete prosthesis for individuals for complete amputations and separated modules for individuals with partial amputations. Complete results are considered will be presented after the implementation of control systems.


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**Figure 4. A real presentation of the complete prosthesis.**

## IV. CONCLUSION

We have designed a powered and human-like prosthesis similar in weight, size and functionality. The addition of elastic elements to the design of the prosthesis contributed to enhance the performance of the device by increasing the bandwidth force, and the tolerance to the impacts generated by the heel strike. The solenoid locking mechanism attached at the transmission of the knee allowed reducing the energy requirements by the actuator. Thus, it contributes to the motor life time. The overweight that presents the active prosthesis are a big challenge that has not been resolved yet. Therefore, we will need to replace the springs at the ankle by a composite leaf spring and a new design that considers reducing considerably the weight of the prosthesis.


## ACKNOWLEDGMENT

We want to thank to students Luis R. Fierro and Hector Mayagoitia who collaborated intensively with this project as research assistants.


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