A Low Cost Design of Powered Ankle-Knee Prosthesis for Lower Limb Amputees
Preliminary Results

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Abstract: In this paper we described a new kind of a powered knee and powered ankle prosthesis for individuals who have suffered a complete or partial lower limb amputation. Our prototype prosthesis consist of two modules, the ankle module and the knee module. The first contains a unidirectional spring configured in parallel with a force-controlled actuator. This spring is intended to store energy in dorsiflexion, and then released it to assist power plantar flexion. The knee modules consist of a series of elastic clutch actuators and a unidirectional spring positioned in parallel to the motor. Preliminary results show two modules designs and ankle module prosthesis prototype almost complete. Two modules working together will enhance the performance of amputee individual, producing more natural gait and reducing the metabolic cost at walking in level-ground.

1 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 (Waters, 1976). A transtibial amputee tends to expend 20-30% more metabolic energy in normal walking (Colborne, 1992). 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 (Winter, 1991).

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 (Au, 2009; Martinez-Villalpando).

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.

Improvements in elastic elements to prosthetic devices have been shown several advantages. These include, increasing tolerance to the load impact, stored and released energy, as wells as reducing energy requirements to the actuation and increasing peak power output (Grimmer, 2012). Therefore many applications have been developed as result of these advantages such as it has been implemented successfully in many applications, for example in exoskeletons devices, active ortheses and robotic legs (Dollar, 2008; Zoss, 2005).

Currently most of the developed power prostheses are still on development stage. One of them is the MIT powered ankle-foot prosthesis that has been developed using series elastic actuators and parallel springs. This has shown to reduce the metabolic cost of walking in transtibial amputees (Au, 2009). *F. Sup et al*. have developed a powered transfemoral prosthesis which has incorporate a spring in parallel to the ankle joint to reduce peak motor torque requirements and to increase the bandwidth (F. Sup, 2009). Another example is the power knee developed Also *Bellman et al* proposed a prosthetic foot that it used successfully elasticity

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elements, they called it The AMP-Foot prosthesis and it has a mechanism that permit storage and release of the energy spring and later releasing it when they needed (Bellman, 2008). This will permit the use of smaller actuators and less energy requirements. Therefore this will reduce the overall prostheses weight, a crucial issue in the development of a powered prosthesis.

In this paper we proposed to develop a power knee and ankle prosthesis divided in two modules that can work together or separately to be used in individuals with a complete or partial amputation. This prosthesis should be able to produce net positive and net negative work during the stance phase of normal walking, thus it will satisfy the requirements of normal gait at level-ground walking. We described design procedures, as well as preliminary results, presenting power ankle prototype module completely.

2 MATERIALS AND METHODS

Prosthesis proposed has two modules, the knee and the ankle module. The purpose of designing the prosthesis in two modules is for a wider range of amputees have the opportunity to use this prosthetic device. It would be used by transtibial amputee individuals, as well as transfemoral amputee individuals. Therefore a person with a transtibial amputation should use only the ankle module, and in the case of a transfemoral amputee it can use both modules or choose only one.

Both modules are able to work together in coordination and each one can work independently on of the other. Both designs were made in SolidWorks ® (see Figure 3 and Figure 4).A simple description of two modules is described in Figure 1, where we can appreciate springs and clutch needed for both modules that are considered in design as crucial elements working as storage-release energy jointly with actuators.

The chassis was designed to contain the actuators, on the other hand, the electronic and the battery sections are placed on the right side of the prosthesis. The structure of the prosthesis except to the foot was made with aluminum 6061 T6.

The prosthesis range motion measured in degrees was established as close as possible to the leg-human range motion during level-ground walking, as we can see in Table 1.

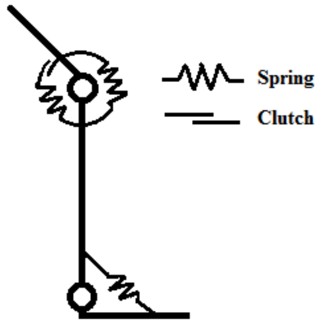

Figure 1: A simple description of springs and clutch used in prosthesis proposed. Two modules are presented separately by ankle module and knee module.

Table 1: Maximum joint motion ranges used in power prosthesis design.

Phase / Condition	Human Walking Max.	Powered Prosthesis Max.
Ankle dorsiflexion	14.1^0	15^0
Ankle plantar flexion	20.6^0	25^0
Knee flexion	73.5^0	120^0

2.1 Knee Design

The actuator features transmissions comprises of a 3:1 timing belt drive coupled to a ball-screw (NSK 10x3 mm) in series with a spring. The translational movement of the ballscrew converts an angular rotation motion of the knee via the series spring with a moment arm $r = 0.045$ m.

During the leg-human motion as described in Figure 2 we can appreciate early stance phase of the knee flexion and extension, the knee torque-angle relationship behaves like a spring and it takes approximately 40% of the gait cycle. Thus this period is linear and consists of the greatest positive and negative mechanical power phase required during the gait cycle. The series elasticity was chosen to mimic this linear region. Thus, choosing the correct spring knee prosthesis can mimics the same behaviour of a biological knee at stance phase.

Spring stiffness that we have chosen satisfies requirements of 74 kN/m and provides a rotational stiffness of 150 Nm/rad to the powered knee module. The series elastic actuator (SEA) uses a brushless DC motor (Maxon® *EC-Max 30*). A unidirectional spring was placed in parallel to the actuator to assist to motor when the knee is more than 74^0 degrees in flexion. The stiffness of the tension spring is 86 kN/m.

Adding a clutch in parallel to the SEA can be reduced the electrical energy consumption in the actuator; therefore we opted to add a clutch to our

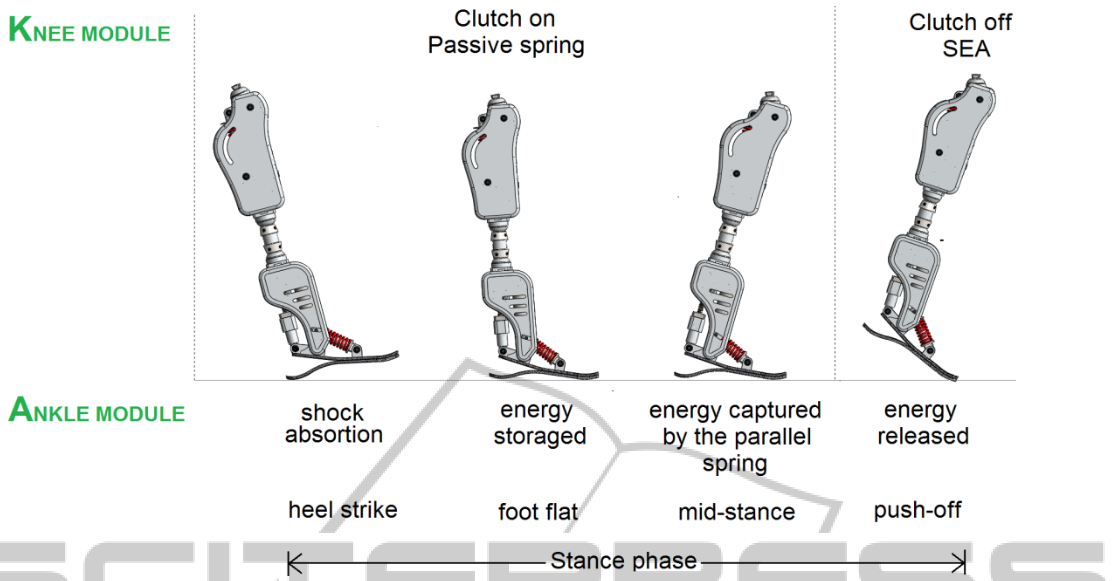

Figure 2: Walking phases described by prosthesis. Clutch is activated during heel strike to mid-stance phase emulating a passive spring.

design (Rouse, 2013). When the clutch is disengage of the transmission the powered knee it behaves like a SEA, and when the clutch is engage the prosthesis behaves like a passive spring. The clutch is disengaged automatically when the spring release all stored energy.

On the top and in the bottom of the chassis was placed a pyramidal connector which is connected to the socket and the extension tube respectively of the amputee (See Figure 1 and Figure 1).

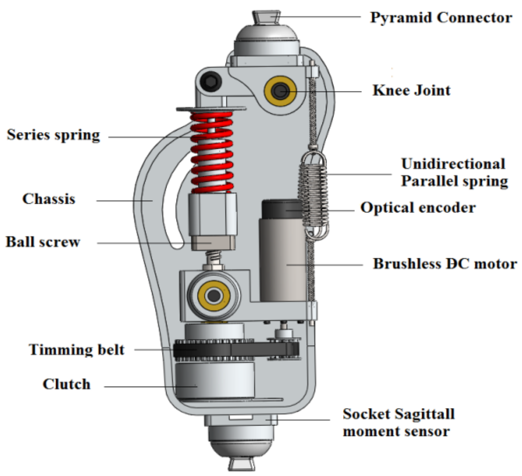

Figure 3: Internal schematic of power knee prosthesis and its corresponding associate sections.

2.2 Ankle Design

The Ankle prosthesis is composed of an actuator in parallel with a unidirectional spring. For driving train system, we used a brushless DC motor (Maxon®, *EC-max 30, 60 Watts*) that drives a ball-screw (NSK®, 10x3 mm) via timing belt that drives the transmission with a 3:1 ratio. The ball-screw is coupled in series with low profile prosthetic foot, (Ossur®, *Flex foot*). This elastic leaf spring was used to emulate the function of a human foot. It provides shock absorption, stores energy during early stance, delivers energy during late stance, as well as to minimize the ground reaction shock to the transmission. At the top of the structure we placed a pyramidal connector (See Figure 3 y Figure 4).

During the stance period the human ankle can reach a velocity of 5 rad/s. In the table 2 it can be seen that the ankle module can achieve a velocity of 3.45 rad/s, therefore it cannot satisfy that requirement. However this problem can be solved changing the motor for another one that be more faster. A good solution is the brushless DC motor Maxon EC-Powermax 30 of 200 W. This motor is similar in size but with almost than thrice the nominal speed and torque, compared with the motor that we are using. Therefore doesn't affect the design of the actuator. This change will enhance the performance of the actuation, increase nominal velocity and the output torque of the actuator.

The ankle actuator incorporates two springs with stiffness of 162 KN/m in parallel with the motor. The purpose is to supplement power output during plantar flexion. The parallel spring is unidirectional, and it is used only to provide a rotational stiffness value of 518 Nm/rad when the ankle angle is greater than zero.

Table 2: Physical parameter specifications of the powered ankle prosthesis.

Variable	Value
Max. dorsiflexion	15^0
Max. plantar flexion	25^0
Height	27 cm
Weight	2.5 kg
Peak output torque	135 Nm
Peak velocity	3.45 rad/s
Parallel offset stiffness	324 kN/m

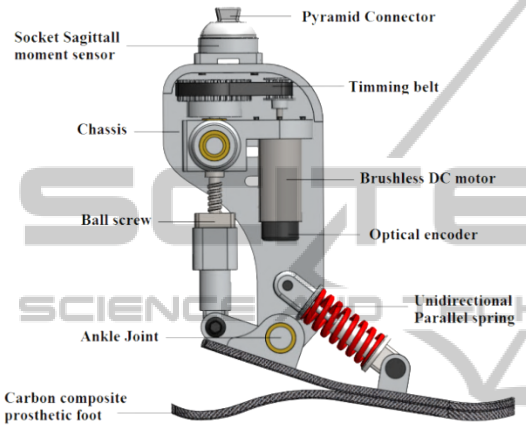

Figure 4: Internal schematic of power ankle prosthesis and its corresponding associate sections.

2.3 Sensing System

Sensing system is a crucial section required to get kinematic and kinetic information related with prosthesis movements that should be processed on real-time by controller algorithm.

There are several methods for measuring force and torque. We choose three different types of sensors for this application. For measuring the ground reaction force and the sagittal plane moment we opted to use load cells, due to their simplicity in design, long lasting, and easy to implement in the control architecture.

The load cells are strategically placed on prosthesis, and each module includes a uniaxial load cell (Omega®). Since impedance changes in a strain gages are very small, strain gages were connected using a Wheatstone bridge circuit configuration.

The sagittal plane moment was measured above the knee joint at the socket Interface. For measuring the force ground reaction we placed the load cell at the top of the ankle module inside the socket of the pyramidal connector (see Figure 3 and Figure 4).

Once we have measured force ground reaction in the load cell we proceed to calculate the torque that would be on the ankle in a static situation using equation 1:

$$\vec{\tau} = \vec{R}\, x\, \vec{F} \quad (1)$$

Where $\vec{F}$ is the force ground reaction vector, $\vec{R}$ is the force ground reaction to ankle vector, and $\vec{\tau}$ is the ankle torque calculated.

In this prosthesis we used a linear potentiometer on the series spring of the knee and on the parallel spring of the ankle to measure the spring deflection and then estimate the joint torque.

We used a position sensor to determine the angle between knee and ankle joint. Since each joint is actuated along a single axis in the sagittal plane, a single position sensor can be placed on each joint. A rotary encoder was attached directly on each joint; it was possible to have a more accurate measurement of the angle for each joint. In order to have a better control for each actuator, we placed an optical encoder on the shaft of both actuators.

For reading inclination angle of the residual limb of the amputee it needs a system that provide us measurements in real-time that can be later utilized by the control system to execute movement commands of the prosthesis. Therefore we opted to utilize the combination of two sensors; a 3-axis gyroscope (L3GD20) and a 3-axis accelerometer (MMA7341L).

One gyroscope and one accelerometer were used for each joint. This method provides flexion and extension angles by estimating acceleration of the joint centre of rotation. Another application in which we used the accelerometer is in the estimation of the ground slope. In order to estimate the ground slope, we used the accelerometer in tangential direction. Assuming that the foot is positioned in parallel with the ground, the only component of acceleration present is the gravity; the gravity direction vector was calculated and computed by monitoring the variation in the acceleration component orthogonal to the long axis of the foot.

3 RESULTS

In this paper, we presented preliminary results of a design and manufacturing of the powered knee-ankle prosthesis prototype. We made some trials of prototype functionality of the powered ankle separately as indicated in Figure 5.

It has been performed studies of finite element analysis by computer for all the structural elements that are part of the prosthesis, this with the purpose of ensure that the device can withstand dynamic loads for subjects with a weight less than 85 kg. These studies were not addressed in this paper. The mechanism of the ankle module has demonstrated to provide a full range of motion of 40 degrees, more than necessary for represent a normal walking. On the other hand the ankle module is capable to generate a torque of 135 Nm, enough to mimic the biological ankle torque at level-ground walking. The springs that are attached in parallel with the actuator on the ankle module can be replaced by composite materials and this change will permit reduce the overall weight of the ankle module.

The addition of elastic elements and a clutch mechanism to our design has been demonstrated theoretically to reduce significantly the electrical energy requirements required by the motor. Simultaneously we amplified the force bandwidth of the actuator. Thus it was permitted that the prosthesis performs more complex tasks where we need more power on the actuation, such as walking upstairs and down stairs, as well as standing from a seated posture.

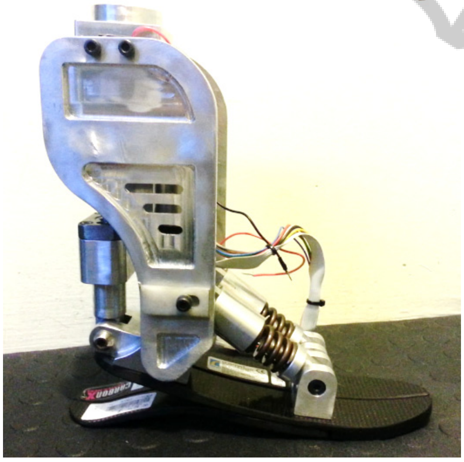

Figure 5: Physical prototype of the powered ankle module.

4 CONCLUSIONS

In this paper we describe the design of a prototype of a powered ankle- knee prosthesis that is human-like in weight, size and functionality. The architecture that comprises this prosthetic device permits to mimic the behaviour of human leg in normal walking. Consequently it would be capable to satisfy the necessary requirements such as, torque and movements range, performing these tasks with economical electrical-energy consumption. The prototype of the ankle module is fully assembled as can be seen in Figure 5. The weight and the size of the ankle module can be reduced optimizing the design and replace some elements that it would permit enhance the torque bandwidth. The knee module is on manufacturing process, once completed this process it will start to test the control system algorithm that will control the actuation of the prosthesis

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