An above-knee prosthesis with magnetorheological variable-damping

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***Abstract*— This work presents the design of a variable-damping prosthesis for above-knee amputees. The proposed low cost system is self-contained and based on a four-bar polycentric mechanism, in which a magnetorheological linear damper is integrated to enable variable-damping control. The paper also describes a control strategy based on a Finite State Machine, which will modulate the damping level according to the actual state of the gait cycle. Preliminary tests on an amputee subject provided a satisfactory performance of the system while operating in the passive mode, *i.e.*, simulating situations when the battery runs out, and also enabled correct identification of gait events.**

I. Introduction

Knee prostheses have evolved quite a lot in the past few years. Today, different options of commercial prostheses that enable users to perform daily activities and multiple motor tasks are available. Controlled variable-damping knees offer smoother walking at different speeds, decreasing the metabolic cost of ambulation, if compared with non-actuated prostheses [1]. Examples of this kind of systems are the C-Leg (Otto Bock) and the Rheo Knee (Össur). The former is a knee prosthesis with hydraulic actuation whose control strategy contemplates the stance and the swing phases, while the Rheo Knee uses a magnetorheological brake on its knee joint. Both systems also feature an electronic embedded system for damping control.

Academic research groups also have came out with new prostheses designs. Some works have focused on the development of powered prostheses, which are able of providing positive work to the user, thus replicating human motor control actions in different activities, such as level walking and sit-to-stand. In [2], the authors present a complete transfemoral prosthesis with powered knee and ankle joints. The electrical actuators that drive the system are controlled for walking and standing using an impedance-based strategy. Another systems is proposed in [3], in which a powered knee prosthesis with an agonist-antagonist configuration is described. The control strategy is based on a variable-impedance behavior for improving gait behavior during stance and swing phases in level-ground walking.

Although powered systems developed recently have been designed to provide an improved functional benefit to user, unfortunately the only powered prosthesis available to the end user today, the Power Knee (Össur), has not yet achieved unquestionable commercial success. The precise reasons for such fact are unclear, but some possibilities are the reduced autonomy, lack of actuators that are silent and present muscle-like behavior, inadequate response when battery runs out, high cost, unfriendly or unnatural user interface.

Within this scenario, in this paper we present the development of a variable-damping transfemural prosthesis that potentially presents lower manufacturing cost and better response when operating in passive mode. The mechanical design is based on a polycentric configuration for passive knee prostheses, the four-bar linkage mechanism. This type of joint provides better foot clearance compared to its single-axis counterparts. It's also more stable, resulting in a safer ambulation for the user. The damper is a magnetorheological (MR) piston whose damping properties can be modified using a controlled electric current. The onboard electronics are composed by a high capacity microprocessor, one inertial measurement unit (IMU) and an absolute encoder, along with other components for communication and data storage. The mechanical and electronic design of the whole system, as well as the control strategy and the experimental results, are further detailed in the following sections.

II. Prosthesis Design

A transfemoral prosthesis with embedded electronics is composed of a knee mechanism, a foot, a tibial extension, a prosthetic socket, and the correponding sensors, actuators and devices that comprise the electronic system. The design of the prototype developed in this work consisted mainly on the develpment of the knee mechanism, the embedded system, and their integration carbon, as well the software development. With respect to the remaining components, the system features a 1C30 Trias (Ottobock) foot, and prosthetic socket was built by a prosthetist. The assembled prototype is presented in Figure 1.

A. Knee mechanism

The developed knee has been designed to perform movements in a polycentric way, which means a combination of both rotational and translational movements between the prosthetic tibia and femur, describing an instantaneous center of rotation (ICR) that moves along a well defined path according to angular position [4]. There are several arrangements that create this polycentric behavior in a knee prosthesis. In this work a four-bar mechanism has been implemented [5]. The bar lengths determine the path of the ICR, situated at the intersection between the virtual extensions of the bars that attach the knee upper part to its lower section. Figure 2 shows the knee mechanism in a CAD

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drawing and the actual prototype, while in figure 3 the four theoretical bars and the path of the ICR according to the knee angle is illustrated.

This polycentric configuration brings functional advantages to the user [6]. Stability in the stance phase is guaranteed by the position of the ICR, which is always anterior to the load line. This effect may be observed in Figure 4, where the user is bearing his weight on the prosthetic limb. Within the loading response event, indeed the load line remains posterior to the ICR, guaranteeing mechanical support to the body weight. This additional safety is not guaranteed in monocentric designs, such as the C-Leg. In this type of system, the load line may be positioned anterior to the ICR, thus providing no support to the user. During normal operation, this feature causes no problems to the user, since the active controller modulates the hydraulic system to provide the necessary support. However, if the battery runs out or in the case of system failure, the will either be unstable to the user or completely blocked, thus requiring additional effort from the user.

Another advantage of the polycentric configuration is the shortening effect in the prosthesis length during the swing phase [7], which allows a greater foot clearance. Without an appropriate foot clearance, the patient would be forced to performed an extra knee flexion during swing to avoid collision between the prosthetic foot and the ground. This represent an additional effort of the muscles groups from the non-affected side, which leads to gait asymmetries [8]. These characteristics are important if it is considered that an actuated prosthesis should have a safe mechanism in the off-mode operation.

Concerning its kinematic range, the knee prototype rotates only in the sagittal plane with maximum flexion angle of 94^{o}. Its total weight is 4 kg, which is within the range of human limbs for subjects whose weight is at least 68 kg [9].

Fig. 1. The transfemoral prosthesis described in this work.

Fig. 2. CAD drawing (left) of the knee prosthesis and the real mechanism (right). The green lines show the four-bar linkage mechanism.

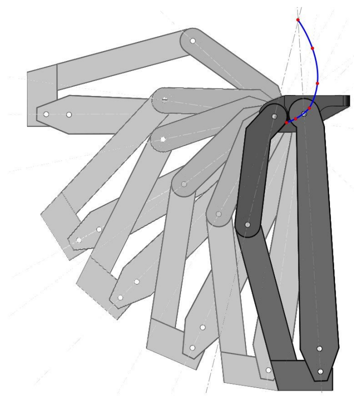


Fig. 3. Polycentric knee motion. The ICR trajectory changes for each knee angle.

B. Actuation

Variable-damping pistons with MR fluids have already been used in rehabilitation devices for restoring motor function [10]. MR dampers exhibit a rapid response, controllable damping force and lower power consumption [11]. In this project, a MR damper manufactured by LORD® is used. The RD-8040-1 piston[1] is a linear stroke damper which is capable of changing its damping constant with a controlled electric current [12]. Besides the compartment that holds the magnetorheological damping fluid, there exists a second chamber holding a pressurized gas that is compressed together with the fluid [13]. The gas compression adds a spring feature to the piston in series with the damping and a pressure offset [12].

Besides presenting compatible specifications with respect to the required performance, the selected damper is also an off-the-shelf product whose value is reduced when compared to candidate systems.

C. Embedded System

The prosthesis has an embedded system which mainly consists of a Teensy 3.0 board, an ARM-based microcontroller with high-processing capacity ($48MHz$). It contains three UART ports, one I^2C port and one SPI port, all of

[1]The MR damper was gently provided by the Brazilian division of LORD® Corporation.

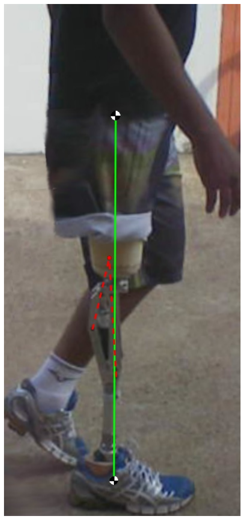


Fig. 4. Amputee subject using the prototype illustrating the load line (green line) position with respect to the ICR (intersection between the red dash lines).

them used in this project. Figure 5 illustrates the embedded system with all its main components.

Besides the Teensy board, the electronic system integrated to the prosthesis features two sensors that are interface with the microcontroller. An inertial measurement unit (IMU) is used for detecting gait events, such as the step initiation and the beginning of the swing phase. This sensor contains a three-axis accelerometer (ADXL345), a gyroscope (ITG3200) and a magnetometer (HMC5883). The angular information is directly obtained from an absolute encoder (AMT203). Both sensors provide digital outputs with high resolution (13 bits and 12 bits, respectively).

All the measurements are sent to an external computer using a wireless communication protocol by means of XBee modules. This information is saved on-board using a SD card. These modules were designed to be detachable from the main setup. This was decided knowing that their functionality would be only needed during the research and the development process, *i.e.*, for a everyday operation there is no need of wireless communication or information storage.

The actuator driver module is composed of a PWM driver (TB6612FNG). This driver is controlled by a PWM signal generated by the microcontroller. This module is responsible for delivering the appropriate current level for the MR piston in the active operation of the prosthesis.

The whole system is powered using a lithium polymer battery with a capacity of $1450mAh$ and $11.1V$ of nominal voltage. The total consumption of the system is $194.22mA$ in passive mode. In active mode, the consumption values can continuously change according to the variable-damping control strategy. Since different control strategies are still under evaluation, at this moment there is still no calculation of the system autonomy for active operation.

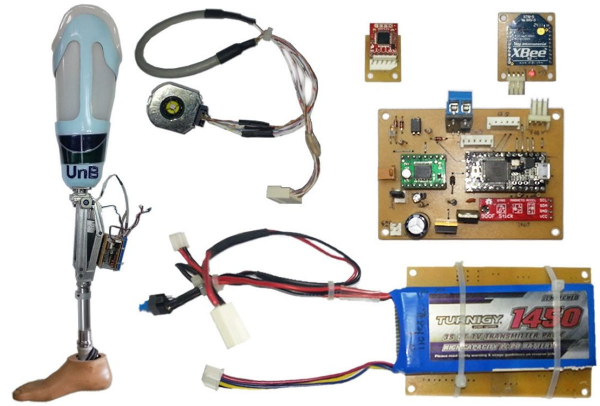


Fig. 5. Embedded system of the prosthesis. The main board has the microcontroller (Teensy), the IMU and the current driver. The other modules are the SD card, the XBee transmitter and the battery. The encoder is placed on the prosthesis, but is shown separately for illustrative purposes.

III. Control Strategy

A. Knee motion

In the sagittal plane, the knee joint flexes and extends during the stance and the swing phase, with a greater range of motion in the latter. The first *flexion-extension* moment occurs during the stance phase to absorb the impact as the leg hits the ground (flexion), and to stabilize the limb when the body weight is borne to that side (extension). When swing begins, the knee flexes to allow limb advancement and then it extends to be prepared for the next step [14].

The beginning and the end of each moment of flexion or extension are delimited by events that occur along the stride. In this work, four events are used to identify the movements of the knee joint in a gait cycle. Figure 6 shows the knee profile for a single gait cycle, as well as the events that mark each flexion-extension sequence. The beginning of the stance is known as *heel strike (HS),* this is the moment when the knee begins to flex until the whole body weight is transferred at the *foot flat* (FF) event. Then, the knee starts to extend slower in comparison to its flexion during stance, approximately half of the speed. At the beginning of the swing phase, when the *toe off* (TO) occurs, the knee flexes again until it reaches its maximum angle at *mid-swing* (MS). Finally, the knee begins to extent as fast as it flexes during the first half of the swing phase, preparing for the next contact at heel strike.

Referring to the control of the knee prosthesis, these events represent the transitions between the different behaviors of the system, which are switched using the Finite State Machine (FSM) described in the next subsection. Also, it is definitely important to identify the transitions correctly in order to provide a reliable operation of the prosthesis during walking. This topic is discussed in Section 3.3.

B. Finite State Machine Control

As mentioned earlier, the knee motion during walking can be considered as the sequence of two states of flexion and extension, whose amplitudes and durations may vary

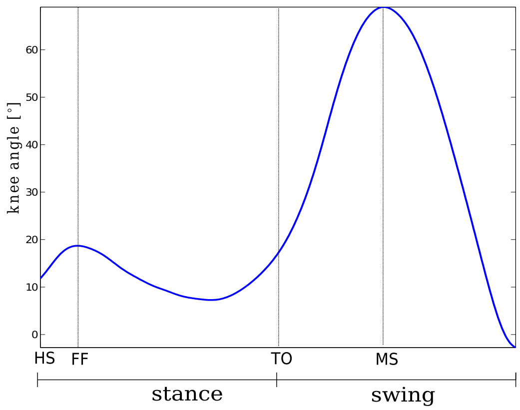


Fig. 6. Knee motion during a gait cycle in the sagittal plane. The four gait events correspond to: *heel strike* (HS), *flat foot* (FF), toe off (TO) and mid-swing (MS).

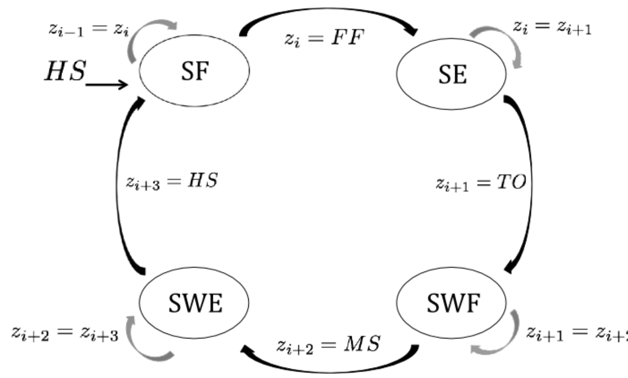


Fig. 7. Finite state machine for the prosthesis control. z_i represents the measurements from the IMU and the encoder, along with the angular velocity calculated from the angle information. Each state represents the flexion and extensions during stance (SF/SE) and swing (SWF/SWE) phase. The four gait events (HS, FF, TO and MS) delimits the states corresponding durations.

according to the gait phase. The transition between those states are given by the four events specified in the previous section. Hence, the knee motion during walking can be thought as a Finite State Machine (FSM). Many *FSM-based* controllers have been implemented for different types of lower limb prostheses [15],[16],[17]. Its easy implementation and adaptation make it a good choice as a first attempt for controlling this kind of systems.

The FSM proposed for this work have four states, stance flexion (SF), stance extension (SE), swing flexion (SWF) and swing extension (SWE). The transitions between the stance flexion and extension is given by the FF event, while the MS event denotes the transition between the swing flexion and extension. Every new step is delimited by HS, which also corresponds to the beginning of the stance phase, while the swing phase initiates at TO. Figure 7 illustrates the proposed FSM for controlling the knee prosthesis.

As a part of the overall control strategy, the FSM will detect the actual phase of gait as the subject is walking with the prosthesis. Then, a MR driver will produce a proportional damping level, actuating the MR piston.

C. Event detection

The detection of the gait events (HS, FF, TO and MS) can be performed by means of observing certain patterns on kinematic data, like acceleration, velocity or angle information [18]. In this work, the detection of HS and TO events were performed using the acceleration measurements extracted from the IMU, along with the angular velocity of the knee joint calculated from the absolute encoder readings. The MS event is detected by simply observing the angle from the encoder measurements, since the mid-swing corresponds to the moment when the knee presents its maximum angle.

IV. Preliminary Results

The purpose of the preliminary experiment described in this work is to identify the knee angle trajectories and velocities, as well as the dynamic acceleration measured by the IMU, while a person is walking using the prototype knee prosthesis on a normal speed. The mechanical performance of the prototype was also evaluated.

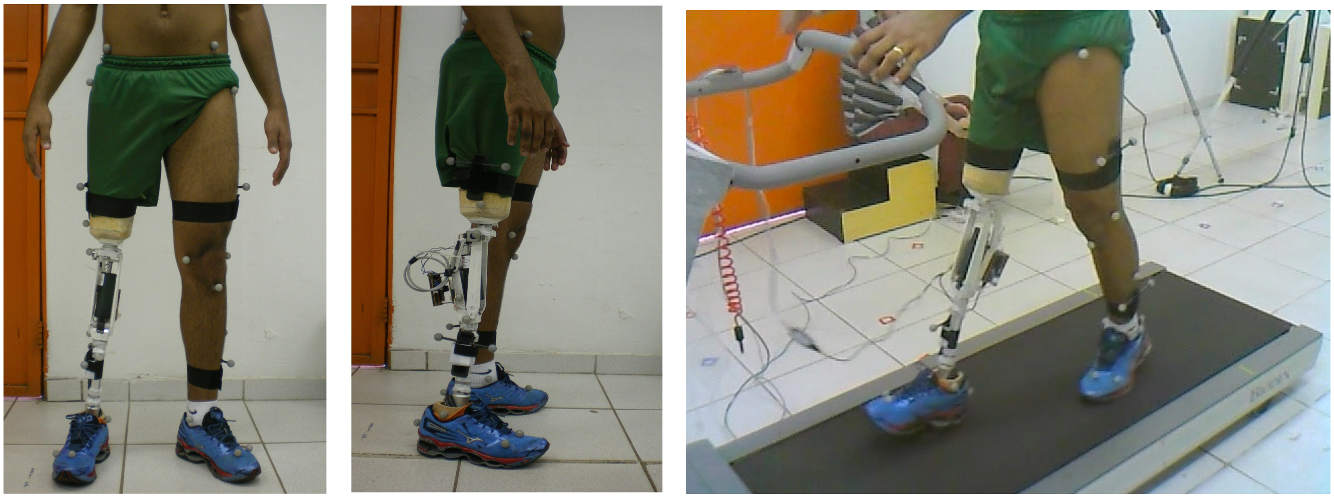
The voluntary user who is participating in this study is a 35 year-old male subject ($1.85m$, $75kg$), who has been an unilateral amputee for 12 years. The subject uses a 3R80 knee prosthesis (Otto Bock) for his daily activities. Before the experiment, the subject wore the knee prosthesis aided by a certified prothetist, who performed a laser alignment process to assure a good fitting. The study was conducted in accordance with the Declaration of Helsinki and the participant has signed informed consent, which was approved by the local ethical committee.

A. Experimental Setup

The experiments consisted on several walking trials performed by the subject wearing the prosthesis on a treadmill (Figure 8). Before testing, the subject's comfortable speed was subjectivelly determined, so the treadmill could be adjusted to this value. A motion capture (MOCAP) system (Qualisys) was used for extracting the kinematics characteristics of the lower limbs by tracking the trajectory of 33 reflective markers placed along both legs. Concurrently, the embedded system installed on the prosthesis recorded the information from the IMU and the absolute encoder. The frequency rate for both systems was set to $250Hz$. Ten trials were conducted with a duration of 60 seconds each.

B. Data collection

Figure 9(a) shows the knee angle measured by the encoder while the subject is walking on the treadmill. The angle raw data was post-processed with a simple smoothing algorithm based on a moving average filter. It is possible to see that the maximum swing flexion angle is around 55^{o}, which corresponds to a typical value for walking on an even terrain. Using this angle information, the MS event has been identified by simply calculating the maximum values of the data set. From this measurement it can be observed that the knee mechanism performs relatively well in passive mode.



(a) (b) (c)

Fig. 8. The amputee subject wearing the knee prosthesis during the experiments.

For the HS and TO detection, the three-axis accelerometer from the IMU was used, along with the angular velocity calculated from the encoder data. In order to obtain a reference measurement for comparing the identification of the HS and TO events from the prosthesis sensors, the information of the MOCAP system was also used to calculate these events.

From the motion capture system, the HS and TO events were detected analyzing the vertical trajectory of a marker placed on the heel of the prosthetic side, noting that minimum and maximum vertical displacements occurred at HS and TO, respectively. Considering that the sampling frequency of both systems was set to the same value, a simple interpolation method was used for fitting both data sets (motion capture system and the prosthesis) to the same size.

Figure 9(b) shows the z-component of the acceleration detected by the IMU and the angular velocity of the knee joint during three consecutive steps. The HS and TO events, detected from the heel marker trajectory, are illustrated as a reference. However, this information was not used in the identification process. As it was mentioned at the beginning of this section, the stance flexion-extension cycle is not present in the subject walking, so no angular velocity information is expected during this phase. This identification process was performed for all trials, showing the same behavior that is presented here.

V. Discussion

The knee motion presented in Figure 9(a) validates the mechanical response of the prosthesis, due to the fact that without any active operation, the prosthesis is able to provide a normal knee motion during the swing phase, which can be improved once the variable-damping control, based on the FSM, will be implemented. Although further validation is required, this data indicates that the proposed prosthesis may present better performance when the battery runs out as compared to other variable-damping systems.

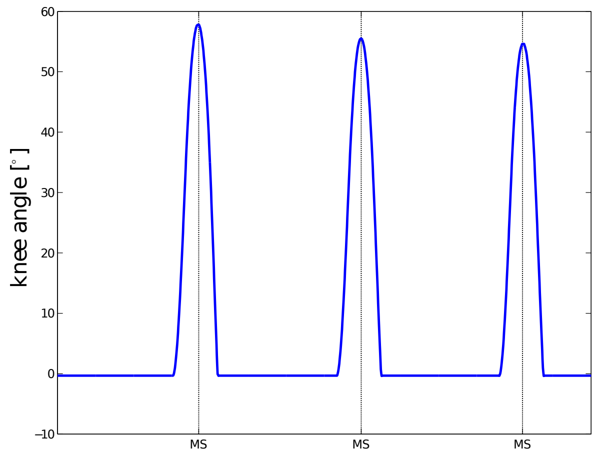
Another issue that is evident looking at the knee motion curve is the absence of the stance flexion-extension cycle. This can be associated with the passive mode of the prosthesis during the experiments, but also it is related to the subject's inability to really bear the body weight to the prosthetic side, as a way to prevent himself from a possible fall. This *self-preservation* mechanism has to be trained in upcoming experiments, so the subject can feel secure using the prosthesis and be able to perform a more "natural" gait, including the flexion-extension cycle during stance.

Observing the acceleration signal in figure 9(b), it is possible to identify a pattern along the three steps, which comes as no surprise because of the cyclic nature of walking. It's interesting to see that at HS, the z-component of acceleration presents a negative peak of approximately $-0.7\,g$ with an average duration of $16\,ms$. On the other side, when TO takes place the acceleration crosses to zero, while the angular velocity presents a high positive value, $140^{\circ}/s$ approximately. These values were calculated considering the whole data set (15000 data points) and the mean values are the ones presented here.

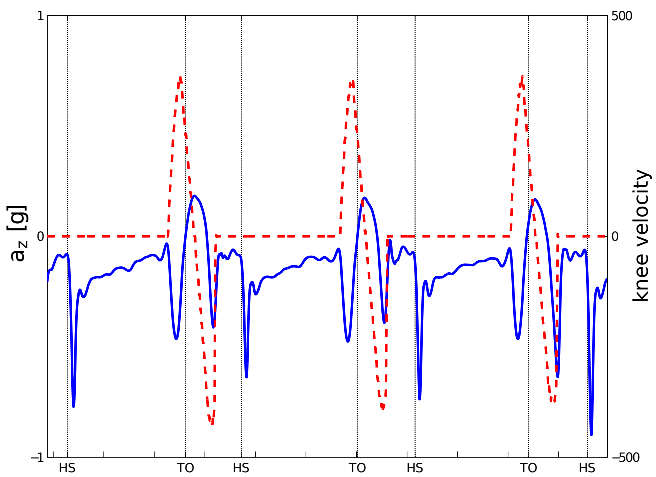
From these results one may infer that it is possible to calculate proper thresholds levels from the sensors information, which may be related to the gait events for level-ground walking in a self-selected speed. Even though the FF event could not be detected in these initial tests, the other events identification will enable the implementation of a variable-damping control for the swing phase.

VI. Conclusions

In this paper, a prototype of a variable-damping prosthesis is presented, including descriptions of the mechanical design, the embedded system and the proposal of a control strategy. Experimental tests were conducted validate the mechanical design of the prosthesis and its response to situations where the battery runs out (passive mode). Furthermore, the tests



(a)



(b)

Fig. 9. Gait event identification: (a) MS event from angle information. (b) HS and TO detection using acceleration and angular velocity. The vertical lines are the reference information from the MOCAP system. It can be observed that the angular velocity and the acceleration present a clear pattern along the steps.

enabled identifying the gait events using the information provided by the onboard sensors. Three of the four events were identified successfully. The foot-flat event (FF) could not be identified due to the absence of stance *flexion-extension* cycle, and thus no non-zero accelerations or angular velocities could be retrieved during the stance phase. The performance of the prototype in the passive mode was evaluated as satisfactory, particularly taking into account the angular excursion of the knee joint during swing, which is similar to the values encountered on a non-amputee. These initial tests led us to identify the proper levels for the transition conditions for the Finite State Machine. Future tests will be conducted towards the implementation of the control strategy to use the prosthesis in the active mode.

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