DEVELOPMENT OF A SIMPLE AND EFFICIENT ABOVE KNEE PROSTHESIS

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ABSTRACT

We present in this paper the state of the art of our research concerning the development of a new semi-active above knee prosthesis. The structure of the prosthesis is exposed. The design and components selection were done taking into account reliable functionality, low cost, low weight, low volume and availability. The control system is based on a micro-controller that ensures functions of locomotion modes recognition and brake control. A simple finite state approach is proposed in order to control the brake and in consequence to enhance patient acceptance and improve the overall functionality of the user.

1. INTRODUCTION

Lower limb prosthetics is aimed in restoring the mobility of an amputee by replacing the missing limb by an artificial one. Researchers efforts are focused on developing prosthetics that simulate, to the extent possible, the intricate and subtle biomechanics of human walking. The above knee prosthesis (AKP) is an artificial limb attached to the amputee’s body below his hip and its control system (if it exists) has to cooperate with amputee’s biological locomotion system. The main part of an AKP is obviously the artificial knee.

From a biomechanical point of view, the principal features of a natural knee in human locomotion are [1, 8, 9]:

- Bearing body weight,
- Absorbing the impact of the body with the ground,
- Maintaining posture stability,
- Contributing in initiating swing,
- Contributing in body lift off in slope and stair climbing,
- when the leg is hanged at the hip and swings forward, the knee flexes to keep clearance and then rigidly extends to accept the body weight.

The main differences between a natural limb and an artificial one are multiples. Indeed, instead of muscles around normal joints, the joint (knee) is generally passive. In addition, inertial and geometrical properties are quite different and the number of freedom degrees is one instead of three. In consequence, the motion of the prosthesis is defined by first, the nature of adopted mechanism then, mechanical characteristics of its components and finally by the remaining (healthy or partially healthy) joints and muscles. Although these differences, the prosthesis must ensure at least the following functionalities [1, 2, 3]:

- preserving sufficient stability to support the body weight,
absorbing the ground impact at heel contact,
allowing, during gait cycles, smooth forward progression with sufficient knee flexion.

The existing AKPs can be classified in three main categories. The first one regroups prostheses based on the passive type mechanism with constant mechanical properties. However such designs are enable to meet requirements of a natural gait due to the lack of knee joint control [2, 4, 6]. In fact, amputee uses the compensation actions like circumduction and vaulting to maintain a minimum mechanical energy consumption in each gait cycle [5]. The second category regroups active prostheses having hydraulic or electric actuators [2, 10, 12]. Generally electromyography (EMG) signals of the residual hip muscles are employed to recognize locomotion modes and to predict user’s intents [5, 6, 11, 12]. Although they are able to produce gait similar to that of normal persons but, they are too expensive, complex and heavy [4]. The last category concerns semi-active prosthesis SAP coupling best features of passive and active prosthesis [4, 11].

The potential of a SAP lies in the properties of simple automatic safety reaction by locking the flexion of the knee or to make it free during amputee’s activities. In fact, unlike a biological knee, the SAP must work without knowledge of its user’s intent or of the environment. Rather, the SAP must infer whether its user desires sit down, go down stairs, stance or swing behaviour and predict when future actions (for example stance/swing transitions) should occur. It must be robust towards various disturbances, such as lifting up a suitcase or moving across an irregular terrain.

The scope of the present paper is the description of last SAP prototype developed at the Robotic Laboratory of Politecnico di Milano. In fact, it represents the result of a long research activity at the lab [3, 7, 13, 14, 15, 16] and cooperation with industrial and medical partners (STM and INAIL corporations). Both mechanical and electrical parts are presented. A big attention is devoted to the control policy of the brake that has been made in order meet all the previous requirement while minimising the power consumption. The paper is organized as follows: section 2 is a description of the prosthesis from mechanical and electrical points of view; in section 3 we develop a simple mathematical model for the prosthesis; section 4 deals with the adopted control strategy and obtained results; and finally we conclude with a general discussion.

2. PROSTHESIS DESCRIPTION

The prosthesis is constituted of three main parts: a commercially available socket, a prosthetic foot (both furnished kindly by INAIL) and a knee that represents the fundamental element. From mechanical point of view, the knee can be assimilated to a revolute joint with one degree of freedom.

The basic components of the artificial knee (fig.1) are:

- a simple mechanical structure ensuring technological functionalities of a revolute joint and allowing the attachment of all others elements (fig. 2).
- a spring in ‘s’ shape that accumulates the energy expended by the amputee while bending his limb. Its particular shape makes it easy to control
- A miniaturized electro-magnetic brake that can be considered as the actuation element. It is coupled to a high speed reducer (harmonic drive) that increases delivered brake torque. The set is used to lock or partially locking the knee joint at desired instances.
- Strain gauges (fig. 3) in order to measure compression and flexion in the tibia. In fact, flexion and compression signals are the image of the ground force that is significant to detect weight bearing. They also signal the intention of the amputee and execution of stair ascent/descent.

- Rotary sensor for the measurement of relative angular position between tibia and femur. It measures the knee flexion which is useful to compute relative velocity and contributes to the identification of the prosthesis state.

- An accelerometer (fig. 4) supplied by STMicroelectronics to record radial acceleration of the tibia. It is used in order to identify several locomotion modes.

The measurements were selected to observe all possible prosthesis states during gait, sitting, standing and slope/stair ascent/descent.

3. PROSTHESIS MODEL

In the sagittal plane, the prosthesis is modeled as a two link rigid robot representing the thigh and the shank (fig.5). The subscript 1, 2 denotes thigh and shank. \( m_i \) are mass, \( d_i \) are distance from the mass center, \( I_i \) are moment of inertia in the mass center, \( l_i \) is the length of thigh, \( \Gamma_x \) and \( \Gamma_y \) are horizontal and vertical hip acceleration components, \( C_s \) is the spring torque.

All sensors data are preprocessed through the use of conditioning electronic, digitized and transferred to the control system. The control system is based on a micro-controller (supplied by STMicroelectronics) which offers simplicity, compactness of the system itself. It provides the proper execution speed for the control software implementation of the brake control algorithm. The control software is requested to track the input signals, to identify critical states and in consequence properly actuates the brake.

\[ \begin{pmatrix} M_{11} & M_{12} & \dot{\theta}_1 \\ M_{12} & M_{22} & \dot{\theta}_2 \end{pmatrix} \begin{pmatrix} \ddot{q}_1 \\ \ddot{q}_2 \end{pmatrix} + \begin{pmatrix} \kappa_1 \\ \kappa_2 \end{pmatrix} + \begin{pmatrix} G_1 \\ G_2 \end{pmatrix} = \begin{pmatrix} \tau_1 \\ \tau_2 \end{pmatrix} \] (1)

where :
\( (q_1, q_2) \) : the vector of the thigh and knee angles,
\( (\tau_1, \tau_2) \) : the vector of the thigh and knee torques,
\( M_{11} = I_1 + l_1^2 d_1^2 + m_1 (l_1^2 + d_1^2 + 2l_1 d_1 c_2) \)
\( M_{12} = I_2 + m_2 (d_2^2 + l_1 d_2 c_2) \)
\( M_{22} = I_2 + m_2 d_2^2 \)
\( \kappa_1 = -m_2 d_2 l_1 s_2 \dot{\theta}_2 (2 \dot{\theta}_1 + \dot{\theta}_2) \)
\( \kappa_2 = m_2 d_2 l_1 s_2 (\dot{\theta}_1^2 - \dot{\theta}_2^2) \)
\( G_1 = -g (m_1 d_1 c_1 + m_2 (l_1 c_1 + d_1 c_1 + 2l_2)) \)
\(-m_2 l_1 s_1 d_2 s_1) + m_1 d_1 s_1 \Gamma_x + (m_2 (l_1 c_1 + d_2 c_1) + m_1 d_1 c_1) \Gamma_y \)
It is clear that this model is valid only when the foot is not in contact with ground else ground efforts must be included as external forces.

This model is used for three main purposes:

a) **Prediction of joint torques**: for a predefined kinematics for both hip and knee joints, the torques can be deduced and in consequence executed if the knee joint was active. However, the prosthesis is semi-active. In this case, we propose to consider the brake as an actuator that can give only resistive torque (opposite to the motion direction) so a qualitative study is conducted to deduce opportune instances to use the brake.

b) **Prediction of the knee kinematics**: the prosthesis is considered as an under-actuated robot and only the hip kinematics is predefined. So, by integrating the second differential equation of (1) we can get temporal evolution of knee joint and in particular final conditions. It is very useful to know for example how the shank will reach the extended posture (with shock or not) so the brake action can be introduced appropriately.

c) **Optimizing the spring design**: spring characteristics are chosen in order to meet various kinematics and kinetics requirements. Obviously, once spring is redesigned the brake must be also reviewed.

### 4. PROSTHESIS CONTROL

To meet the requirement of a real time control and robustness, the controller must be built on a simple algorithm but an efficient one: simple means it doesn’t involve complex model such dynamic model, while efficient means that it deals with all locomotion modes.

The selected control strategy is based on a non-analytical control scheme [2, 12] that clones the biological control. The mechanism for implementation is a rule based system so it implements “if-then” relations. “if” part of a rule describes the sensory states, while a “then” part of a rule defines the corresponding action. In order to design such control a knowledge base has to be generated. This knowledge is based on existing experience and known facts about both normal and prosthetic gait. Indeed, two types of experts are involved in the construction of knowledge base, the biomechanical engineer and the clinical [2, 8, 9]. The first one is familiar with the technical structure and functionality of the system and he is involved in constructing the knowledge base for the locomotion modes recognition. The modes identification process relates the physical measurements to the heuristic amputee states that are well known to the clinical expert. The present and past measured signals from the prosthesis compose the date base.

The main modes to be handled by the control software are: standing (ST), sitting(SI), walking (WA) and going up/down stairs (GS). Possible transitions between these various situations are depicted in figure 6. This figure traduces the logic existing in human actions.

According to different methodologies, each mode can be divided into a finite number of sub-modes that can again be divided into other periods. For example a walking mode is the succession of two sub-modes: stance and swing. Stance sub-mode can also be divided into heel contact, foot flat and toes off periods [8, 9].

Real tests has been carried out at Budrio prosthetic’s center using the prosthesis without acting on the brake. The tests were made by an A/K amputee (age of 50’s, and weight of about 75kg) in various modes (sitting, walking, …). Typical
recorded data from different sensors are shown in figures 8, 9, 10 and 11.

In our control strategy, we have only kept three main periods reflecting the loading state of the prosthesis. The first one is the period of double support where the weight is distributed on the two legs (biological and artificial one). It is observed particularly in ST and WA modes. The second one is the period of single prosthesis support, the prosthesis supports alone at least the whole weight. It takes place for example during WA and GS modes. The last one is the free prosthesis period, it occurs when no load is applied on the prosthesis. It is the case for example in the swing phase of WA and GS. The periods are detected by observing when each of the measurements cross a preset threshold level and enter into one of the predefined ranges. The control algorithm relates the action on the brake to the different identified states.

Remind that the objective of the brake use is to make prosthesis motions as smooth as possible and to make its use safety. So, we decide to activate the brake progressively at the end of the third period in order to avoid shock when the prosthesis is completely extended. On the other hand, the brake will be completely activated (maximum excitation) during periods similar to the second one while it will be free during periods similar to the first one. The application of this strategy for the signals recorded in Budrio center leads to the brake control signal depicted in figure 12 the brake is excited during small periods in order to minimize energy consumption.

5. CONCLUSION

Through this paper we have discussed the design of a simple and efficient AKP with the most updated technologies while reducing the involved costs. The main functionalities such stability and shock absorption are guaranteed. A simple dynamic model was proposed for various purposes leading to an optimum prosthesis behavior. The mechanical and electrical AKP structure was simplified. The installed sensors allow the
identification of various locomotion’s modes and relevant periods that are the inputs of the control brake software. More complete validation tests are planned in order to make an objective evaluation of our prosthesis performances.

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REFERENCES


INTERNET LINKS

http://www.inail.it Centro Protesi INAIL.