Design of Variable-Damping Control for Prosthetic Knee based on a Simulated Biped

Jie Zhao*, Karsten Berns*, Roberto de Souza Baptista†, Antônio Padilha L. Bó†
*Robotics Research Lab, Department of Computer Science, TU Kaiserslautern, Kaiserslautern, Germany
{zhao, berns}@cs.uni-kl.de
†Automation and Robotics Laboratory, University of Brasília, Brasília, Brazil
baptista@ieee.org, antonio.plb@lara.unb.br

Abstract—This paper presents a new variable-damping controller for a prosthetic knee. The development is also based on a novel approach: using a simulated biped robot for evaluating different prostheses designs and controllers before real tests are conducted on humans. The simulated biped incorporates several features of human walking, such as functional morphology, exploitation of inherent dynamics, hierarchical control network, combination of feed-forward and feedback controllers and phase-dependent modulation and so on. Based on this virtual model of human walking, we have studied biomechanical aspects of knee during walking and then developed a controller based on the damping profile developed by simulated bipedal robot throughout a gait cycle. This controller has been evaluated on a modified version of the simulated biped, in which a model of a prosthetic leg under construction has been incorporated. This simulated biped is 1.8m in height and 76kg in weight with 21 degrees of freedom. Results of such experiments for walking on flat and rough terrains have provided satisfactory outputs, including improved robustness.

I. INTRODUCTION

Amputation of lower limbs, either due to trauma, infections or other causes, may potentially decrease considerably the person’s quality of life. Walking is obviously of prior importance among the basic movements affected. This paper is related to the use of new tools in the development of control strategies for artificial legs, since current prosthetic legs are not yet capable of completely restoring functionalities.

To improve walking capability for handicapped people researchers examine various solutions. For instance, in the field of functional stimulation and neuroprostheses [1], [2], the control approach uses electromyographic signals captured from muscles and nerves as stimulations to activate prosthesis. Another class of solution is the use of powered joints like that presented in [3] in which an ankle is controlled during the toe-off to optimize energy efficiency and to increase stability. The powered knee prosthesis Genium™ [4] developed by Otto Bock is commercially available. This microprocessor-controlled knee can anticipate movements of users and adapt instantaneously in order to function as close to a natural leg as possible. Nevertheless, the performance of this artificial knee under different walking conditions is not clear [5]. Some researchers are looking into biological aspects of knee joint during walking and then trying to transfer the findings into robotics or prosthetic legs. Based on electromyograph analysis [6]–[8] proposed a central control unit generating commands for synergistic muscle primitives and reflex actions. In accordance with their work, human walking tends to be combined of five phases which can be affiliating to kinetic or kinematic events: weight acceptance, leg propulsion, trunk stabilization, leg swing, and heel strike [9]. Found by [10], maximum energy consumed in walking comes from the leg swing phase, which starts from the toe-off event and ends just when heel strike event is achieved. The knee joint is always keeping stiffen to provide enough support for human in stance phase while it is providing damping to prevent hyperflexion and hyperextension. Therefore, a variable-damping knee prosthesis with the capability of performing those functionalities is required. These knee prostheses are normally controlled by microcontroller or DSP to modulate the damping ratios and adopt to different walking scenarios. Variable-damping controlled knee prostheses have some advantages over passive knee prostheses [11]. These advantages covers enhanced knee stability, more smoothness of knee gait and adaptation to different walking velocities.

The suggested methodology results from a kinematic and kinetic analysis based on a simulated biped, as shown in Fig. 1. The bipedal robot has human-like features, including 21 DoFs, the weight up to 76kg and the height over to 1.8m. The Newton Game Dynamics™ is applied for dynamic calculation of rigid body. A biologically motivated control method is applied to control this biped and capture the kinematic and dynamical data set from different experiment scenarios [9]. The control architecture is designed as a hierarchical system of feed-forward and feedback control units. A central pattern generator coordinates the stimulation and synchronization of various control units. Instead of using a dynamic model of biped, reflex controllers and motor patterns play the most important roles in regulating locomotion of the biped. In next sections, we describes working principles of this prosthetic leg and its mechanical designs. We then study the kinematic and dynamical data captured from simulated biped, from which we extract a damping profile during leg swing phase within a gait cycle. Afterwards, a finite-state machine that regulates the switching of different phases is suggested. In accordance with this profile, the corresponding variable-damping controller for interesting walking phases is developed. At last, this controller is tested within a simulated biped with a leg prosthesis and results are demonstrated.

II. MECHANICAL DESIGN OF THE PROSTHESIS

The knee module of the ongoing prosthetic knee project is a polycentric knee mechanism with adjustable damping ratios.

*www.newtondynamics.com
A polycentric knee consists of any type of mechanism in which the center of rotation varies according to the angular position [12].

As in various polycentric knee designs, our polycentric mechanism is based on a four-bar linkage system [13]. In a four-bar linkage system, the length of the bars determines the trajectory of the instantaneous center of rotation [14]. The particular trajectory of a specific design brings many advantages to polycentric knees when compared to monocentric ones [12]. In the stance phase if the instantaneous center of rotation is positioned posterior the Trochanter-Knee-Ankle (TKA) line, the system is considered a stable weight bearing system. Furthermore, if the instantaneous center of rotation is located above the mechanical center of rotation of the mechanism, keeping the knee straight will require less effort from the user and will increase leverage for the residual limb. In the swing phase the angular motion of the four-bar linkage mechanism may shorten its total length compared to the full extension position. This characteristic is exploited to provide foot clearance during the swing phase. The trajectory of the instantaneous center of rotation also influences the dynamics of the mechanism, therefore the polycentric knee can be designed to ease acceleration during the flexion and deceleration during extension. In sum: the four-bar linkage mechanism applied to the design of prosthetic knees brings the following advantages when compared to monocentric: stability in stance phase, total length shortening, and acceleration and deceleration enhancement in swing phase.

Besides the advantages of the polycentric knee design, a magnetorheological piston is integrated to the prosthetic knee to provide damping adjustment of the mechanism. Magnetorheological fluids alter its viscosity in the presence of a magnetic field. A piston, or damper, filled with this type of fluid provides a continuously variable damping force controlled by a desired input [15]. In our project we used a commercially available magnetorheological piston from Lord Corporation. In this device, the magnetic field through the magnetorheological fluid is induced through an electric current, which is the control input to adjust the damping coefficient of the system.

The prosthetic knee prototype is currently in the finishing phase of production as shown in Fig. 1. The goal of this project is to investigate different control strategies taking into account human in the loop for above the knee amputees.

III. BIOMECHANICAL ANALYSIS OF KNEE JOINT

Understanding human walking and the dynamic properties in each phase is important for designing assistive devices that may improve gait robustness and performance. In this section, the knee biomechanics during different gait phases is studied for designing improved prosthetic knees. [6], [7] found that walking can be divided into five subdivisions, as shown in Fig. 2, in which the knee biomechanics can be detailedly studied.

A. Biomechanical Events at Knee joint

Through simulating a biped in a dynamic environment within scenario of flat ground walking, kinematic data of knee along gait cycle can be observed as in Fig. 3.

1) We consider that the first phase weight acceptance starts just after full contact with ground of the swing leg, namely after both heels and toes contact. In this phase the former leg supports the weight of upper trunk. The former knee begins to flex until around 20°, that performs like a compressed spring with stored potential energy. In this phase, the knee can be modeled as an angular spring.

2) When the maximum compression is achieved, the knee extends until maximum stance extension approaches. This extension phase is called leg propulsion, which makes the knee again act as an angular spring. The stiffness of knee keeps the same level as that in first phase.

3) The third phase is characterized by double support during which stabilization of the body posture is guaranteed. By analyzing kinematics of knee, we find that the knee begins to flex again for preparing leg swing.

4) After happening of the toe-off event, the swing phase starts, transferring the leg in front of the body, and the flexion ends when knee angle achieves at around 60°. It extends consequently until knee of swing leg is totally stretched. This phase includes both knee flexion and extension. Observing torque and angle of knee, we can theoretically calculate the work W and power P on knee joint as following:

\[ W = \tau \cdot \theta \]  \hspace{1cm} (1)

\[ P = \tau \cdot \dot{\theta} . \]  \hspace{1cm} (2)

Where \( \tau \) denotes the torque and \( \theta \) and \( \dot{\theta} \) are individually the joint revolution and velocity. The knee power consumption is generally negative (see Fig. 3) since it hinders knee angular velocity. Therefore in the swing phase, the knee can be modeled as a variable damper.

5) As soon as the swing leg’s knee is locked or its heel contacts with ground, the last phase begins. It manages the foot impact during heel strike and provides control concerning full contact of the foot. The knee should be again stiff to handle the impact of body weight.

\(^{2}\)The detailed architecture, i.e. biologically inspired control of a dynamically walking bipedal robot can be found in [9]
B. Extraction of Damping Profile

As shown in Fig. 3, the knee generates a positive, resistive moment during leg extension in swing phase. This negative power portion of the gait cycle can be effectively modeled as a variable damper, as shown by biomechanical analysis in Sec. III-A. Therefore, using Eq. 3, the effective damping coefficient of the knee throughout swing extension is calculated:

\[ B_k = \frac{\tau_k}{\dot{\theta}_k}. \]  (3)

The effective damping variable \( B_k \) is the ratio between the knee torque \( \tau_k \) and knee velocity \( \dot{\theta}_k \). By using the data set illustrated in Fig.3, one can calculate the damping coefficient directly as shown in 4. From Fig 4, we see the knee damping \( B_k \) decreases sharply when knee starts to flex from stance phase. Then \( B_k \) is mostly performing as a linear function of knee angle up to the maximum value of knee flexion. After extension of knee joint, \( B_k \) is nearly a linear function of knee angle between 20\(^\circ\) and 50\(^\circ\). The damping coefficient along increasing and decreasing knee angle have the same course during swing phase. Since in the stance phase, the knee joint is stiff to keep stability of the upper body, we then think its damping coefficient goes very high. Thus we only consider the damping coefficient in leg swing phase. According to Fig. 4, the damping coefficient can be represented as a function of knee angle:

\[ B_k(\theta_k) = \begin{cases} 
\frac{B_{k_{low}} - B_{k_{up}}}{\theta_{k_{up}} - \theta_{k_{low}}} \cdot (\theta_k - \theta_{k_{low}}), & \text{if } \theta_k > \theta_{k_{low}} \\
\infty, & \text{otherwise} 
\end{cases} \]  (4)

In Eq. 4 \( \theta_{k_{low}} \) and \( \theta_{k_{up}} \) mean a range in which knee joint can be modeled as a variable damper whereas \( B_{k_{low}} \) and \( B_{k_{up}} \) denote respectively the damping coefficient at \( \theta_{k_{low}} \) and \( \theta_{k_{up}} \).

IV. Finite-State Machine for Walking Phases

From a biomechanical point of view, it is known that walking can be divided into distinct phases. However, for controlling a prosthetic leg based on these different walking phases, those events and their features must be estimated in real-time. We have to use limited sensors mounted within prosthetic leg to decide the happening of critical events that walking phase is transferred from one state to another. The existed sensors on prosthetic leg are encoders on each joint.
are required: to successfully activate the state machine, following variables (ST), Stabilization can be actually arranged in three states, which are respectively knee and toe off gait phases are activated by three events, i.e., to present a health knee, as illustrated in Fig. 2. Transitions of as a central controller triggering state of walking phases.

A finite-state machine for cyclic walking is instrumented to present a health knee, as illustrated in Fig. 2. Transitions of gait phases are activated by three events, i.e., toe off, locked knee and full contact. Therefore, the whole five gait phases can be actually arranged in three states, which are respectively Stabilization (ST), Swing (SW) and Heel Strike (HS). In order to successfully activate the state machine, following variables are required:

- Knee angle ($\theta_k$) indicates the relative angle of knee joint. Fully extended knee angle denotes $\theta_k = 0$. Maximum knee angle is 2 rad/s.
- Ankle angle ($\theta_a$) denotes the relative angle of ankle joint in sagittal plane. Neutral position $\theta_{a0}$ is set at the position that human is standing still. Angle of ankle is set larger than neutral angle $\theta_{a0}$ when plantarflexion is happening, whereas is smaller when there is dorsiflexion.
- Foot load ($F_l$) means loaded force on the foot based on four force sensors respectively mounted on inner toe, outer toe, inner heel and outer heel. With

$$F_l = \frac{F_{force} - F_{low}}{F_{up} - F_{low}}, \text{ if } F_l \in [0, 1],$$

where $F_{force}$ is the measured force, whereas $F_{low}$ and $F_{up}$ indicating individually the lower and upper threshold of vertical force on feet. Besides, the value of $F_l$ is limited in $[0, 1]$. It can be divided into Heel Contact(H) and Toe Contact(T), in which 1 means full contact while 0 denotes no contact at all.

The elaborate descriptions are shown as follows:

1) When the heel strikes on the ground, the landing knee joint bends a little due strong impact of upper body. However it maintains around equilibrium position to support the weight of upper body, acting as a locked mechanism. After the forward transferring of the center of mass, the stored energy will be released. Meanwhile the opposite knee prepares to start the swing. State switches when following two kinematic events happened:

- the ground contact detected by force sensors mounted on feet is smaller than a predefined threshold value, e.g. $F_l < F_{threshold}$, and
- due to plantarflexion of ankle joint, it grows up till larger than neutral angle, e.g. $\theta_a > \theta_{a0}$.

2) Swing phase completely starts after toe-off. The knee is bended due to the inertia of lower body. The opposite knee is again extended to the neutral angle to support body. As the knee flexes beyond $\theta_{klow}$, damping control has been applied to resist hyperflexion. The position tracking in flexion phase is not necessary since it utilizes the passive dynamics of knee joint. The damping coefficient is slightly increased along knee angle until knee extension comes. The knee extension is caused by the gravity of leg and torque generated at hip joint which make leg extend and move forward. As in beginning of extension, knee acts as a passive joint and therefore no controller is required. Once knee angle achieves the $\theta_{kup}$, for resisting hyperextension, a damper controller is thus needed. Knee velocity is then gradually decreased due to resisted torques. When knee joint passes over $\theta_{kup}$ and approaches the equilibrium position, a lock mechanism will prevent the knee hyperextension. The next state Heel Strike happens when:

- the knee angle $\theta_k > \theta_{kthreshold}$; or
- the ground contact $F_l > F_{threshold}$, which means heel of swing leg starts to land on the ground.

3) Heel Strike is responsible for reducing the ground impact and for generating a lowering of toes after heel strike. Instead of modeling the knee joint as a variable damper, a locked mechanism within knee joint to support the impact of landing of upper body is suggested. It turns again to Stabilization phase, if following condition is fulfilled:

$$H > H_{threshold}.$$ It means the heel strike is finished and the foot has made full contact with the ground.

The elaborate descriptions are shown as follows:

\begin{center}
\begin{tikzpicture}
  \node (A) {Stabilization};
  \node (B) [below right of=A] {Swing};
  \node (C) [below left of=A] {Heel Strike};
  \node (D) [below right of=C] {\text{F}_l < \text{F}_{\text{up}} \text{ and } \theta_k > \theta_{a0} \quad \text{or} \quad \theta_k < \theta_{\text{bthreshold}} \text{ or} \quad F_l < F_{\text{threshold}}};
  \node (E) [below of=B] {\text{H} > H_{\text{up}}};

  \path[->] (A) edge node {\text{F}_l < \text{F}_{\text{up}} \text{ and } \theta_k > \theta_{a0}} (B);
  \path[->] (A) edge node {\theta_k < \theta_{\text{bthreshold}} \text{ or} \quad F_l < F_{\text{threshold}}} (C);
  \path[->] (C) edge node {\text{H} > H_{\text{up}}} (D);
\end{tikzpicture}
\end{center}

Fig. 5: Transitions of finite state machine regulating gait phases.

V. SIMULATION AND RESULTS

In this section, a simulated biped and the simulation environment are introduced firstly. A biologically motivated
control for a biped is briefly presented. Utilizing the gait phase analysis and variable-damping control, simulation results on a biped with a virtual leg prosthesis are then illustrated.

The leg prosthesis consists of one knee joint and ankle joints in both sagittal and frontal plane. Except for the variable-damping controlled knee joint, other joints are passive elements with fixed stiffness and damping. A sliding joint is placed in between top of the amputee and knee joint, whereas another is situated between knee and ankle joint. Those two sliding joints, which are shown in Fig. 1, provide a constant stiffness and lock mechanism that are applied to lock knee in late swing phase and stance phase.

In this paper, we have implemented the simulation of cyclic walking in a flat platform and then obtained the data from simulation which are used as kind of biomechanical data. To test adaptation of the suggested methodology, walking on rough terrain is performed as well.

A. Normal Walking on Even Terrain

The first tests conducted using the simulated biped were based on normal walking at the speed of 1.21 m/s. It provides a detailed evaluation of the calculated damping coefficient and the mechanism of the leg prosthesis. The kinetic and kinematic analysis give insight into joint trajectories and necessary joint torques. Angles trajectories of joints over the course of a gait cycle, which are respectively hip joint in sagittal plane, knee joint and ankle joints, are shown in Fig. 6. 15 consecutive steps of walking on flat ground are averaged by manually tagging the sampling data from one heel strike to next. Positive values indicates individually a joint flexion, abduction or dorsiflexion, while negative values denote extension, adduction or plantarflexion. The solid lines show the up-to-date average values of joints along gait cycle, where dashed lines illustrate the maximum and minimum values. The vertical dashed line around 69% is the location of the transition from stance to swing. The left column figures show the joints in prosthetic legs and figures on right one illustrate those joints in a healthy leg. The average duration of gait cycle is 1.35 s.

In Fig. 6, hip angles in frontal plane for prosthetic leg and healthy leg are generally similar, which means amputees with this prosthesis do not need to adjust the amplitude of hip swing in swing phase and postures during swing phase. We also found that the course of the knee angle at prosthetic leg has closely the same profile of that in a health leg. That means the variable damping control has fulfilled the functionality as required. But since we have only the passive joints for ankle, in which stiffness and torque control are both required, therefore, we see some differences between the prosthetic leg and the healthy leg. Since in healthy leg, the ankle joint in sagittal plane is actively controlled, meaning reflex controllers and motor patterns are applied at joint. The lateral stability is normally maintained by a stiffness control at ankle joint in frontal plane. However due to the missing of stiffness control, the lateral stability can not be guaranteed during the stance phase. Fig. 7 illustrates torques of joints in both legs. The torque here is a combination of pure motor torque and torque generated by a virtual spring and damper. Again, plots in the left column represent mean, minimum and maximum values of joint torques in the prosthetic leg, where the right ones denote those values of joint torques in the intact leg. Torques at hip frontal joint in both legs perform very close to each other. This means amputees do not need to generate more energy to swing the leg. Coming to knee joints, based on the calculated damping profile, the prosthetic knee produces enough torques
B. Walking on Rough Terrain

The following series of tests were conducted on rough terrain. The simulated terrain is built with roughness up to 3.3 cm, which is equivalent to randomly placing rocks or similar obstacles with this maximum height throughout the terrain. The kinematic and kinetic data from prosthetic leg are illustrated in Fig. 8, in which mean angles and torques for hip, knee, and ankle are shown in solid lines, where dashed lines denote the maximum and minimum values. The kinematic data shows that the average angle values are similar with those values in normal walking, but due to the rough ground, angle values can vary in a wider range than those in normal walking. However, with variable-damping control, the biped can still walk very smoothly on this uneven terrain. Looking into the course of knee angle and joint, we found that the torque generated by damper is less than that in normal walking. This is because protrusion on the ground curtails duration of swing phase and extends stance phase. Ankle vibration in frontal plane, that results from unevenness on the ground, has impact especially on lateral stability. In this case, a constant stiffness control limits its adaptation to various environments. Hence a variable-stiffness controller could be more comfortable for amputees in different walking scenarios.

VI. Conclusion

We have discussed a methodology of designing a variable-damping controller for a leg prosthesis in this paper. For controlling the phase-dependent prosthesis, we have firstly studied the biomechanical aspects of knee joint along walking cycle and then defined a finite-state machine to identify phases of interest. We have then performed simulations and obtained such data from a virtual model of human walking. Afterwards, we have analyzed the kinetic and kinematic data and extracted a damping profile along walking cycle. At last, we have tested this methodology within this simulation environment both on flat ground and rough terrain. This is one great advantage of the proposed method, since the control strategy may be evaluated without any risk for humans. The resulting gait was satisfactory and robust to different environments. Future works include testing the developed control strategy on the real prosthetic knee under development and expanding the proposed methodology for both active knee and ankle control.

Acknowledgment

We would like to thank DAAD and CAPEX for funding this work.

References


www.daad.de
www.capes.gov.br