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

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***Abstract*—This paper presents the development of a variable-damping controller for a prosthetic knee using a simulated biped in a virtual environment 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. Based on this virtual model of human walking, we have studied biomechanical aspects of the knee joint during walking. Observing the damping profile developed by the simulated biped throughout a gait cycle, we designed a controller for the knee joint. This controller has been evaluated on a modified version of the simulated biped, in which the model of a real prosthetic leg was incorporated. 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, decreases 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.

Recent improvements were made in the development of artificial limb devices, specially the so called intelligent, or micro-controlled, prosthesis [1], [2]. Microprocessor-controlled prosthesis can anticipate movements of users and adapt instantaneously in order to function as close to a natural leg as possible. Nevertheless, in many cases, the performance of these devices under different walking conditions is still not clear [2]. Moreover, in the context of lower limbs prosthesis, the design and function of the prosthetic knee is of great importance because it is the most proximal artificial joint that the amputee must stabilize and control to effectively ambulate [3].

To enhance the performance of prosthetic knee devices, researchers are looking into biological aspects of knee joint during walking and then trying to transfer the findings into robotics and prosthetic legs. Based on electromyograph analysis [4]–[6] 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 associated to kinetic or kinematic events: *weight acceptance*, *leg propulsion*, *trunk stabilization*,*leg swing*, and *heel strike* [7]. According to [8], maximum energy consumed in walking comes from the leg swing phase, which starts on the toe-off event and ends just when heel strike event occurs. Meanwhile, in the stance phase the knee joint is kept stiff to provide enough support for human while providing a certain damping to prevent hyperflexion and hyperextension. Therefore, a variable-damping knee prosthesis with those functionalities is required. Variable-damping controlled knee prostheses have some advantages over passive knee prostheses, which includes: enhanced knee stability, more smoothness of knee gait and adaptation to different walking velocities [9].

The development and adjustment of control strategies for prosthetic limbs are usually based on trial-and-error approach and/or rely deeply in a specialist's intuition. These approaches are time consuming, difficult to apply in larger scale, and not applicable to limbs under development with more anthropomorphic motion and actuation [10]. To overcome these challenges, model-based and simulation-based design approaches have been explored.

A model-based approach is used in [11] to investigate the kinematic adaptation of an ankle prosthesis to sudden changes in ground slope. This model, however, represents only the mechanical device and does not consider the amputee's body. The work developed in [12] explores the use of a hybrid dynamical model to represent a human with a transfemoral prosthesis and to tune a PD control. This physical model assumes five point-masses for simplicity: one for the hip, one for each thigh and one for each calf. In [13], a dynamic model to represent an above-the knee prosthesis during a complete gait cycle is proposed. Using optimization procedures, the author was able to design a controller to achieve a knee flexion pattern close to that of the normal gait. In this case, the physical model used is a two-dimensional dynamic model composed of three rigid segments connected via revolute joints. Another interesting employment of simulation tools is shown in [14], where a model-based simulation environment is used to analyze simple passive devices and also test control algorithms for active prostheses. Although more complete than the before mentioned models, this one simplifies the representation of the upper body in a point-mass and a spring-damper coupling with the lower body. As observed, the methods developed so far for simulation and controller development are usually subjected to model simplification and do not embrace the complete dynamics of a whole human body in a 3D space.

In this paper we use a simulated biped in a virtual environment to design and evaluate a variable-damping knee prosthesis controller before real tests are conducted on humans. The virtual environment, introduced in Section II, is used

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for two main purposes. First we use the simulated biped to generate and study kinematic and dynamical data. Second, a modified version of the simulated biped, in which a model of a prosthetic leg has been incorporated, is used to evaluate the proposed controller. Using the kinematic and dynamical data captured from the simulated biped, we extract a damping profile during leg swing phase within a gait cycle, which is shown in Section III. In Section IV, a finite-state machine that regulates the switching of different phases is suggested. In accordance with the previously extracted profile, a variable-damping controller for certain walking phases is proposed. Section V presents the performance results of this controller when tested within a simulated biped with a leg prosthesis in two different walking scenarios: flat surface and rough terrain. Finally the conclusions are presented in Section VI.

II. Simulation Platform

A. Mechanical Knee-Motivation

The knee module of the ongoing prosthetic knee project is a polycentric knee mechanism with adjustable damping ratios. The prosthetic knee prototype is currently in the finishing phase of production. The overall goal of this project is to investigate different control strategies taking into account human in the loop for above the knee amputees. With this in mind, the simulated biped virtual environment provides human-like behaviors which are considered in the first steps in the design of the controller for the prosthetic knee.

B. Simulated Biped

The bipedal simulator has human-like features, including 21 DoFs to represent the different joints in the human body. It is 1.8 m high and contrary to point-mass models, its weight is distributed based on average human data, with the total weight adding up to 76 kg.

The *Newton Game Dynamics*[1] 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, such as: flat surface, positive and negative slopes, and rough terrain [7]. 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 the biped, reflex controllers and motor patterns play the most important roles in regulating locomotion of the biped. The similarities to human walking based on biomechanical kinematic data comparisons is shown in [7].

In order to develop and test the proposed control strategy for the prosthetic knee, the simulated biped was altered to have one of its lower limb behave like a prosthetic device, as shown in Fig. 1. The variable-damping controlled leg prosthesis consists of one knee joint and ankle joints in both sagittal and frontal plane. Except for the variable-damping controlled knee joint, others joints are passive elements with fixed stiffness and damping.

[1]www.newtondynamics.com

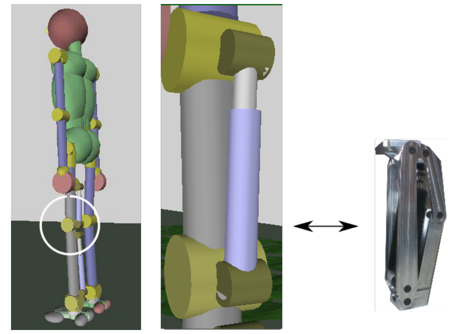

Fig. 1: Mechanical configuration of prosthesis in a simulated biped and prototype.

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 to enhance the controller design for the prosthetic knee. According to [4], [5], a gait cycle can be divided into five phases, as shown in Fig. 2, in which the knee biomechanics can be studied in detail.

A. Biomechanical Events at Knee joint

Using the simulated biped in a dynamic environment [2] in a flat surface walking scenario, kinematic data of the knee can be extracted, as shown in Fig. 3, along the different phases of a gait cycle:

1) *Weight Acceptance*-this phase starts just after full contact of the swing leg with the ground, namely after both heels and toes contact. In this phase the support leg holds the weight of upper trunk. The support knee begins to flex until around 20° and behaves like a compressed spring with stored potential energy. Hence, the knee can be modeled here as an angular spring.
2) *Propulsion*-when the maximum compression is achieved, the knee extends until maximum stance extension approaches and the knee acts again as an angular spring. The stiffness of knee is kept at the same level as in the first phase.
3) *Stabilization*-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, preparing for the leg swing.
4) *Leg Swing*-after the toe-off event occurs, the swing phase starts and the leg is projected in front of the body. This phase includes both knee flexion and extension. Knee flexion occurs first, until around 60°, followed by the extension, until knee of swing leg is totally stretched. Observing torque and angle of the knee, we can theoretically calculate the work W and

[2]The detailed architecture, i.e. biologically inspired control of a dynamically walking bipedal robot can be found in [7]

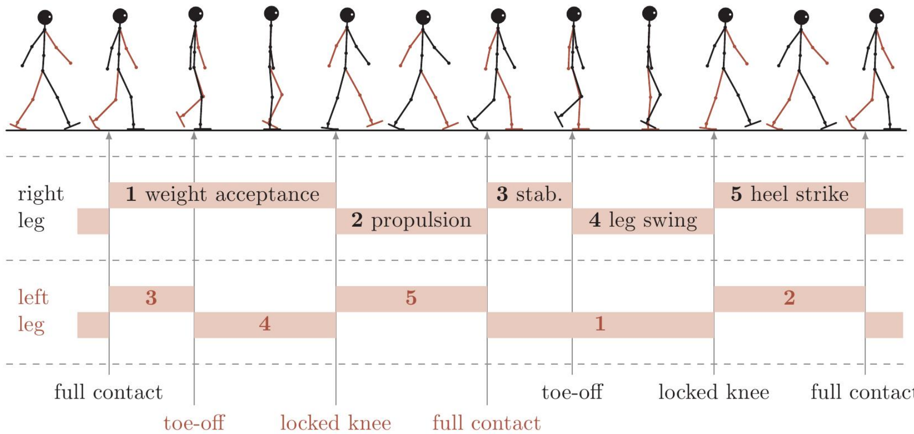

Fig. 2: Walking phases for both legs and sensor events for switching phases [7].

power P on the knee joint as following:

$$W = \tau \cdot \theta \quad (1)$$
$$P = \tau \cdot \dot{\theta}. \quad (2)$$

Where τ denotes the torque and θ 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) *Heel Strike*-as soon as the swing leg's knee is locked or its heel contacts with the 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.

B. Extraction of Damping Profile

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

$$B_k = \frac{\tau_k}{\dot{\theta}_k}. \quad (3)$$

The effective damping variable B_k is the ratio between the knee torque τ_k and knee velocity $\dot{\theta}_k$.

By using the data set illustrated in Fig.3, we can calculate the damping coefficient directly as shown in Eq. 3. 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 the knee joint, B_k is nearly a linear function of knee angle between 20° and 50°. The damping coefficient along the increase and decrease of the knee angle display similar courses during swing phase. Since in the stance phase, the knee joint is stiff to keep stability of the upper body, we observe that its damping coefficient highly increases. Thus we only consider the damping coefficient in the leg swing phase. According to Fig. 4, the damping coefficient

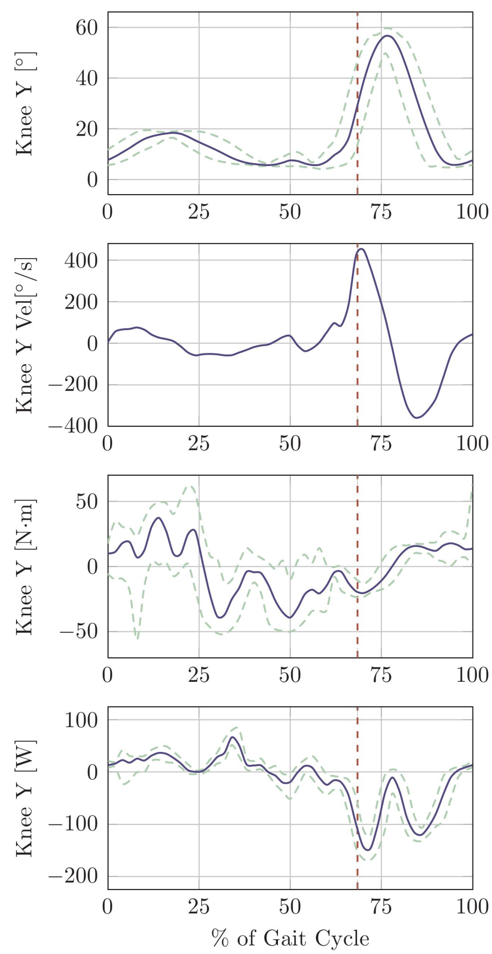

Fig. 3: Angle, velocity, torque and power consumption at knee joint. The solid lines indicate mean values while dashed lines denotes maximum and minimum values. The vertical dashed line denotes the toe-off event.

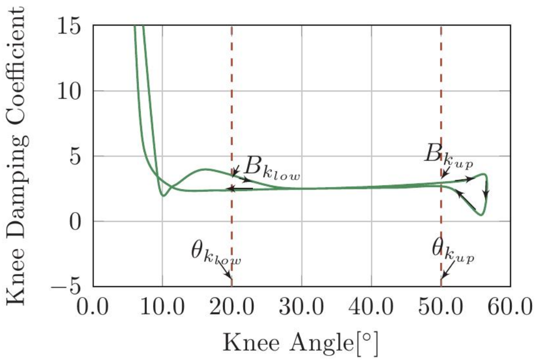

Fig. 4: Knee damping coefficient along the knee angle during gait cycle.

can be represented as a function of the knee angle:

$$B_k(\theta_k) = \begin{cases} B_{k_{low}} + \dfrac{B_{k_{up}} - B_{k_{low}}}{\theta_{k_{up}} - \theta_{k_{low}}} \cdot (\theta_k - \theta_{k_{low}}), & \text{if } \theta_k > \theta_{k_{low}} \\ \infty, & \text{otherwise} \end{cases} \quad (4)$$

In Eq. 4 $\theta_{k_{low}}$ and $\theta_{k_{up}}$ represent 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, walking can be divided into 5 distinct phases. However, in order to control 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 the prosthetic leg to decide the occurrence of critical events that indicates switching among the walking phases. The existing sensors on the prosthetic leg are encoders on each joint and four load cells on each foot. As existence of a hierarchical control architecture, sensor information are introduced into the Central Pattern Generators, i.e. CPG, which plays the role as a central controller triggering the state of the walking phases.

A finite-state machine for cyclic walking is proposed to represent a healthy 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 actually be arranged in three states, which are *Stabilization* (ST), *Swing* (SW) and *Heel Strike* (HS), respectively. In order to successfully activate the finite-state machine, the following variables are required:

- Knee angle (θ_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 (θ_a) denotes the relative angle of ankle joint in sagittal plane. Neutral position θ_{a_0} is set at the position that human is standing still. Positive angles denote plantarflexion while negative denote 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

$$F_l = \frac{F_{force} - F_{low}}{F_{up} - F_{low}}, \text{if } F_l \in [0, 1], \quad (5)$$

where F_{force} is the measured force, F_{low} and F_{up} are respectively the lower and upper threshold of the vertical force on the feet. 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.

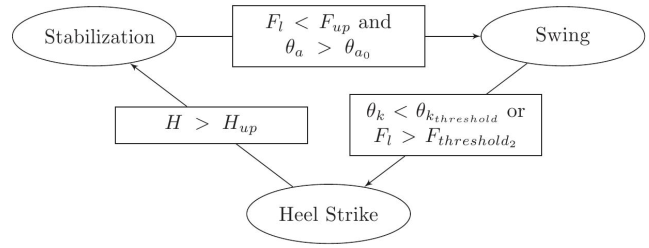

Fig. 5: Transitions of finite state machine regulating gait phases.

Figure 5 shows the proposed Finite State Machine and we can now present an elaborate description:

1) *Stabilization*-when the heel strikes on the ground, the landing knee joint bends a little due to the strong impact. However it maintains around an equilibrium position to support the weight of the upper body, acting as a locked mechanism. After the forward transferring of the center of mass, the stored energy is released. Meanwhile the opposite knee prepares to start the swing. State switches when the following two kinematic events occur:
   - the ground contact detected by force sensors mounted on feet is smaller than a predefined threshold value, e.g. $F_l < F_{threshold_1}$ and
   - due to plantarflexion of ankle joint, it grows up till larger than neutral angle, e.g. $\theta_a > \theta_{a_0}$.
2) *Swing*-the swing phase starts after toe-off. The knee is bent due to the inertia of the knee. The opposite knee is again extended to the neutral angle to support the body. As the knee flexes beyond $\theta_{k_{low}}$, the damping control is applied to resist hyperflexion. The position tracking in flexion phase is not necessary since it utilizes the passive dynamics of the knee joint. The damping coefficient is slightly increased coupled to the knee angle until knee extension occurs. The knee extension is caused by the gravity acting on the leg and the torque generated at hip joint which make leg extend and move forward. As in the beginning of extension, the knee acts as a passive joint and therefore no controller is required. Once the knee angle achieves the $\theta_{k_{up}}$, a damper controller is needed to prevent hyperflexion. Knee velocity is then gradually decreased due to resista torques. When the knee joint passes over $\theta_{k_{low}}$ and approaches the equilibrium position, a lock mechanism will prevent the knee hyperextension. The transition to the next state, *Heel Strike*, happens when:

- the knee angle $\theta_k > \theta_{k_{threshold_2}}$; or
- the ground contact $F_l > F_{threshold_2}$, 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 the toes after heel strike. Instead of modeling the knee joint as a variable damper, a locked mechanism within the knee joint is suggested to support the impact of landing. The finite-state machine turns again to the *Stabilization* phase, if the following condition is fulfilled:

    - $H > H_{threshold}$,

    which means the heel strike is finished and the foot has made full contact with the ground.

V. Simulation and Results

This section presents the simulation results of the proposed controller on the simulated biped containing the leg prosthesis on two distinct walking scenarios: flat surface and rough terrain.

A. Normal Walking on Flat Surface

The first tests conducted using the simulated biped were based on normal walking at the speed of 1.21 m/s. It allows a detailed evaluation of the proposed controller for the prosthesis compared to a simulated healthy subject. The kinetic and kinematic analysis give insight into joint trajectories and necessary joint torques. Figure 6 shows the angle trajectories

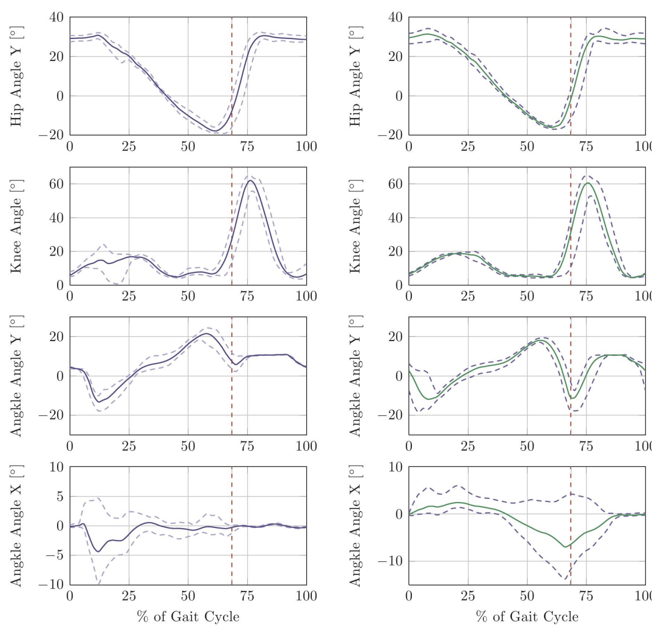

Fig. 6: Joint angles over the course of gait cycle. Solid lines represent mean values, dashed lines mark the minimum and maximum values, the vertical dashed line denotes the transition from stance to swing. Left column healthy leg, right prosthesis.

of hip, knee and ankle joint in the sagittal plane, and ankle in the frontal plane over the course of a gait cycle. Fifteen consecutive steps of walking on flat ground are averaged by manually tagging the sampling data from one heel strike to the next. Positive values indicate, respectively, a joint flexion,abduction and dorsiflexion, while negative values denote extension, adduction and plantarflexion. The solid lines show the up-to-date average values of joint angles along gait cycle and dashed lines illustrate the maximum and minimum values. The vertical dashed line around 69% is the location of the transition from stance to swing.

In Fig. 6, hip angles in the frontal plane of the 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 the swing phase and postures during the swing phase. We also found that the course of the knee angle at the prosthetic leg has closely the same profile of that in the healthy leg. That means the variable damping control has fulfilled the functionality as required. As for the ankle joint, there are some differences between the prosthetic leg and the healthy leg. In a healthy leg, the ankle is actively controlled in the sagittal plane, meaning reflex controllers and motor patterns are applied at this joint. However, due to the lack of stiffness control in the prosthetic leg, the lateral stability can not be guaranteed during the stance phase.

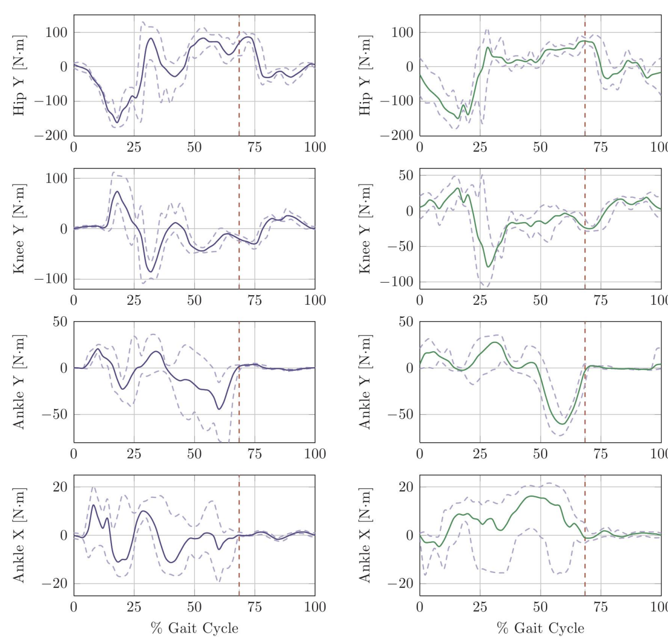

Fig. 7: Joint torques over the course of gait cycle. Solid lines represent mean values, dashed lines mark the minimum and maximum values, the vertical dashed line denotes the transition from stance to swing. Left column healthy leg, right prosthesis.

Figure 7 illustrates the torques in the joints in both cases. In the simulated biped the torque is a combination of pure motor torque and torque generated by a virtual spring or/and damper. Torques at the 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. Looking at the knee joints, based on the calculated damping profile, the prosthetic knee produces enough torque to restrain its locomotion and therefore achieve a very human-like walking gait.

B. Walking on Rough Terrain

The second series of tests were conducted on rough terrain. The simulated terrain is built with roughness of up to 33 mm, which is equivalent to randomly placing rocks or similar obstacles with this maximum height throughout the terrain. The kinematic and kinetic data from the prosthetic leg are illustrated in Fig. 8. The kinematic data show that the average angle values are similar with those values in flat surface walking, but due to the rough terrain, angle values can vary in a wider range. However, with the variable-damping control, the biped can still walk very smoothly on this uneven terrain. Looking into the course of knee joint, we found that the torque generated by the damper is less than that in normal walking. This is because protuberances on the ground shorten the duration of the swing phase and extend the 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 in this joint seems to be more comfortable for amputees in different walking scenarios.

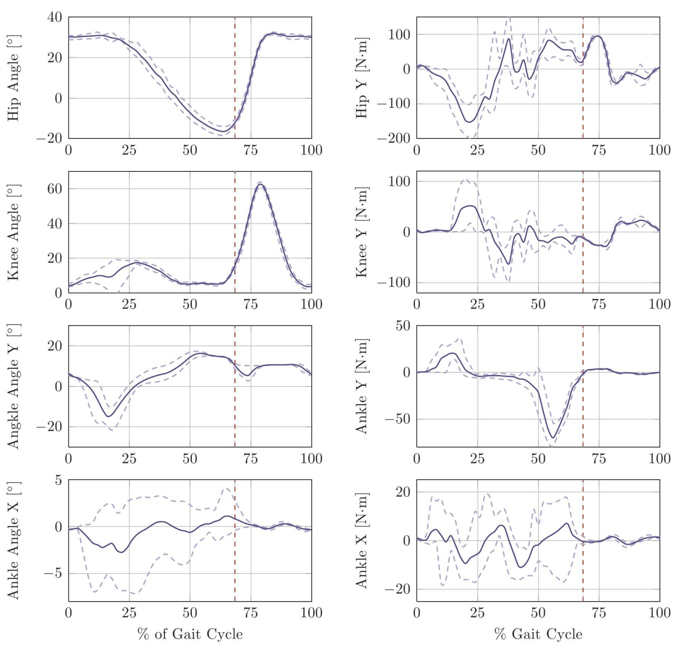

Fig. 8: Prosthetic joint angles, left, and torques, right, over the course of gait cycle.

VI. Conclusion

In this paper, we presented a methodology for designing a variable-damping controller for a leg prosthesis using a simulated biped in a virtual environment. To control the phase-dependent prosthesis, we first studied the biomechanical aspects of knee joint along walking cycle. We then performed simulations and obtained such data from a virtual model of human walking. Afterwards, we analyzed the kinetic and kinematic data and extracted a damping profile along walking cycle. Based on this, we defined a finite-state machine and the corresponding damping control for each state. At last, we 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.

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