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Simulation of a Slope Adapting Ankle Prosthesis Provided by Semi-Active Damping

Andrew K. LaPrè, *Student Member, IEEE,* and Frank Sup, *Member, IEEE*

Abstract— **Modern passive prosthetic foot/ankles cannot adapt to variations in ground slope. The lack of active adaptation significantly compromises an amputee's balance and stability on uneven terrains. To address this deficit, this paper proposes an ankle prosthesis that uses semi-active damping as a mechanism to provide active slope adaptation. The conceptual ankle prosthesis consists of a modulated damper in series with a spring foot that allows the foot to conform to the angle of the surface in the sagittal plane. In support of this approach, biomechanics data is presented showing unilateral transtibial amputees stepping on a wedge with their daily-use passive prosthesis. Based on this data, a simulation of the ankle prosthesis with semi-active damping is developed. The model shows the kinematic adaptation of the prosthesis to sudden changes in ground slope. The results show the potential of an ankle prosthesis with semi-active damping to actively adapt to the ground slope at each step.**

I. Introduction

Commerical passive prostheses currently available to lower limb amputees are optimized for level ground walking [1]. In these devices, the foot and ankle are replaced with a fixed carbon fiber spring that cannot adapt to variations in terrain or the user's activity (such as walking on slopes, descending and ascending stairs, and while standing up from a seated position). A fixed, stiff ankle can create potentially unsafe conditions for amputees, especially if there is a sudden or unexpected disturbance such as stepping on a rock or curb [2-4]. Studies have shown that the falling incidence rate in the amputee population is equal to that of institutionalized elderly, and about one out of ten lower limb amputees have reported requiring medical attention for a fall that has occurred within the year [5]. A significant limitation of passive ankle prostheses is that their range of motion is directly coupled to the ankle moment through the spring foot. This hinders an amputee's ability to maintain stable (foot flat) ground contact on uneven terrains. Resulting in a significant decrease in the walking stability of an amputee compared to an able-bodied person while walking on slopes [2]. In order to increase the stability of the amputee, an ideal prosthetic ankle would adapt to the terrain at each and every step.

Manuscript received April 17, 2011. Revised June 17, 2011. A. K. LaPrè and F. Sup are with the Department of Mechanical Engineering Department, University of Massachusetts, Amherst, MA 01003 USA (corresponding author F. Sup: phone: 413-545-2946; fax: 413-545-1027; e-mail: sup@ecs.umass.edu, alapree@ student.umass.edu).

To study the first step response to a new ground slope, able-bodied persons were studied during their initial step onto an inclined surface (called "wedge stepping") [6]. This wedge stepping study focused on the biomechanics of the lower limbs and postural changes during the adaptation phase of gait. While no similar amputee wedge-stepping study was found in the literature, a study was conducted that focused on steady-state inclined walking of lower limb amputees [3]. The study clearly shows that an amputee's hip, knee, and sound ankle trajectories significantly compensate for both the kinematic and kinetic limitations of the ankle when walking up and down slopes [3].

The first attempt to address the adaptive deficiencies in prosthetic ankles was by Hans Mauch from the 1950s through the 1970s [7]. His passive ankle incorporated a hydraulic damper that allowed the ankle to adapt to the ground slope. It also featured a spring to return the foot to a neutral position after toe-off. The device demonstrated the potential of ankle adaptation, but mechanical failures prevented the device from being commercialized. Inspired by Mauch, a friction-based device that adapts to sloped surfaces at heel strike and then locks as the device bears weight was developed at Northwestern University [8]. Commercially available, the Endolite Echelon foot/ankle is a passive hydraulic prosthesis with independent plantarflexion and dorsiflexion damping values that must be manually set. The prosthesis aims to passively mimic the visco-elastic response of human muscle with fixed damping and spring constants to provide ankle adaptation on sloped surfaces. Another commercial device, the Ossur Proprio Foot, is an electromechanical prosthesis. Using the onboard sensors, its control algorithm adapts incrementally by estimating the ground angle of the current step and then adjusts the ankle during swing assuming the next step will be on the same ground slope. The design of the Proprio Foot only allows for ankle adjustments while not bearing weight.

Powered active prostheses have been developed to deliver human-scale torque and power with an actuator [9-11]. Collectively these robotic devices demonstrate the potential to create intelligent prostheses using robotic technology that has only recently become available. The ability for these devices to enable step-by-step adaption is complicated by the advantage they offer (i.e. their ability to act in a forceful manner). This requires more detailed knowledge of the terrain and user's activity to coordinate the prosthesis with the user so as not to destabilize the amputee.

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Of the above passive prostheses, none modulate the ankle's impedance in real-time to improve the adaptive behavior of the ankle. The semi-active damping approach presented in this paper modulates the behavior of the ankle in real-time based on the characteristics of the forceful interaction between the ground, prosthesis, and amputee that encode the changes in terrain and user's activity. In this manner, a semi-active approach provides the advantages of a fully active system, but relies on the user-prosthesis interaction to enhance an amputee's stability throughout the gait cycle. In this paper, the biomechanics of transtibial amputees' first step onto ten-degree inclined and declined slopes from a level surface are studied. Based on this data, a model of an adaptive ankle comprising of a spring foot in series with an adjustable damper is presented. Simulation results show that such a device would be capable of effectively adapting to changes in ground slope in the sagittal plane. This work will provide a basis for future development of an adaptive ankle prosthesis that uses a semi-active damper to adapt to the ground slope at each step.

II. Model

The adaptive ankle model, Fig. 1, places the stiff spring foot, k_f, in series with a semi-active damper, b. In this model, the effective ankle angle is the summation of the deflection of the spring foot given by

$$\theta_A = \theta_f + \theta_d \tag{1}$$

θ_f, θ_d, and θ_A are the carbon foot angular flexion, damper angle, and total ankle angle, respectively. In this configuration the net motion of the ankle is augmented by the damper element that is driven by the force transmitted by the spring foot. To modulate the damping coefficient in real-time, the system leverages semi-active damping to optimize the adaptation throughout the gait cycle. In hydraulic semi-active damping, the impedance is modulated by controlling the flow-rate of hydraulic fluid from one side of a double acting cylinder to the other by means of an electronic servo valve (as conceptually depicted in Fig. 1). A parallel spring is used to return the ankle to the neutral position after toe-off when the modulated damper is set to a low value during swing flexion.

The simulation model is obtained by combining (1) and (2) that provide the solution for the total ankle angle as a function of ankle torque and a modulated damping coefficient, (3). The total ankle moment, τ_A, is transferred

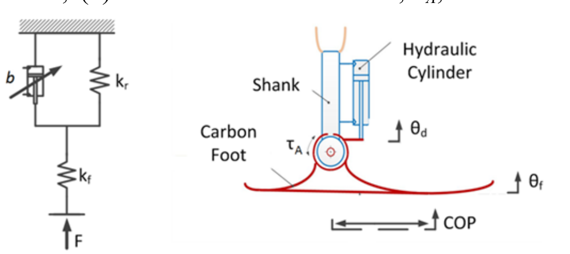

Fig. 1. Spring-damper model representing a carbon-fiber foot, k_f, in series with a modulated damper, b, and return spring, k_r in the linear domain (left) and rotational domain (right).

through the parallel damper and return spring that is in series with the spring foot, (2).

$$\tau_A = k_f\theta_f = b\dot{\theta}_d + k_r\theta_d \tag{2}$$

$$\theta_A = \frac{\tau_A}{k_f} + \int \frac{\tau_A - k_r\theta_d}{b}\,dt \tag{3}$$

III. Data Collection

The input of the adaptive ankle prosthesis model is ankle torque. To obtain the real-world input to the system, biomechanics data was collected with unilateral transtibial amputee subjects traversing both a ten degree incline and decline in a single step. A ten degree slope was chosen to represent an extreme that might be encountered in an outdoor setting without placing significant demands on the test subject. Further, it has been shown in able-bodied persons that if the ground slope of the initial step is greater than fifteen degrees, changes gait posture may occur [6]. For this study, all data collected was on level ground or ten degree slopes to avoid changes in gait form.

While walking at their self-selected pace on a level surface, subjects approached either the inclined or declined surface measuring 0.75 m in length. They took a single step with their prosthetic-side landing heel first or foot flat on the sloped surface and then continued on a level surface at the new elevation. The subjects repeated the trial until 10 successful trials for each test were completed for both sound and prosthetic leg. A successful trial was defined as walking at a steady speed and stepping on incline or decline with only the appendage of interest without visibly altering their gait. It should be noted that no difference in the body mechanics is observed in healthy biomechanics when landing heel first or foot flat on an incline [6]. The biomechanics data was recorded as the subject traversed the sloped surface.

The experiments took place in the Biomechanics Laboratory at the University of Massachusetts, Amherst. The analysis used an eight-camera Qualisys Oqus 3-Series optical motion capture system operated by Qualisys Track Manager software (Qualisys, Inc., Gothenberg, Sweden) to sample test subjects fitted with reflective infrared tracking sensors, at 240Hz. Ground reaction forces were recorded with a floor mounted strain gauge force platform (OR6-5, AMTI, Inc. Watertown, MA, USA), on which the ramps were placed. The calibration markers were placed at the following anatomical features to reconstruct the bone structure during data processing: 1st and 5th metatarsals, medial and lateral knee joint as well as ankle malleoli, and the greater trochanters in order to reconstruct points of rotation. Four tracking markers were fixed to each foot, shank, and thigh, as well as the hip segment throughout all testing to track the trajectories of each segment. On the prosthesis, virtual malleoli markers were created during the post-processing of the data by mirroring the able side to represent the prosthesis center of rotation. It should be noted that this assumption for the prosthetic-side can only provide approximations of the actual joint motions and moments.

The data was processed using Visual 3D v4 software (C-Motion, Inc, Rockville, MD, USA) to calculate all joint positions, velocities, moments, and power.

Two unilateral amputee subjects were recruited through a local prosthetist. Note that all aspects of the study described herein were approved by the University of Massachusetts-Amherst Institutional Review Board, and all subjects signed informed consent forms prior to the participation. Exclusion criteria included bilateral transtibial amputees, younger than 18 years of age or older than 55 years of age, overweight, physically unfit, confounding medical condition that would place the subject at risk, and a poor sense of balance with a history of falling. Subject #1 was a 34-year-old female (1.78 m, 70 kg) more than 15 years post amputation. Her daily-use prosthesis was an Ossur LP Vari-Flex. Subject #2 was a 42-year-old male (1.88 m, 102 kg) more than 20 years post amputation. His daily-use prosthesis was an Ossur Modular III Category-7.

IV. Methodology

Using MATLAB Simulink, (3) was modeled. The prosthetic-side ankle torques for each test subject were entered as the system input. In this approach, the power loss of the modulated damper is ignored. The effect of this assumption is that more energy will be stored in the foot spring that would be returned during push-off. Since this simulation focuses on the kinematic response, the simulation remains valid for studying the effects of adding a damper in series with a spring foot.

To tune the damping coefficient b, eight recognizable events were chosen. The damping constant for each phase was tuned manually to closely mimic the healthy biomechanics during slope walking [12]. For example, at heel strike the initial damping value is set very low to allow the ankle to adapt to the ground slope. The second event occurs just after heel strike and increases impedance of the system to provide support and prevent toe-slap. Subsequent events occur at inflection points in the torque profile (where the ankle acceleration would be zero). Through an iterative tuning approach, it was found that altering the damping at these points allows for smooth simulated angular accelerations. The events are shown graphically in Fig. 2. The stiffness of the foot spring was found by independently linearly fitting angle-torque response during controlled plantarflexion and controlled dorsiflexion to model the heel and toe stiffness, respectively. The stiffness of the return spring (1.0 N-m/deg) was chosen to provide the appropriate response during swing to provide toe clearance.

V. Results and Discussion

Figure 3 presents a kinematic comparison of the amputee wedge stepping study and the adaptive ankle prosthesis simulation to able-bodied sloped walking data. The tuned damping coefficients corresponding to each gait event from Fig. 2 are presented in Table 1. Using the biomechanics data, the stiffness of the foot and heel of the subjects' daily-use prostheses were found to be 2.9 and 12 N-m/deg for Subject #1 and 3.7 and 10 N-m/deg for Subject #2.

TABLE I
Damping Coefficient (N-m-s/deg) updates at different events

Event	Subject 1			Subject 2		
	Level	Up	Down	Level	Up	Down
1	0.1	0.1	0.1	0.1	0.1	0.1
2	0.12	0.12	0.4	0.15	0.2	0.1
3	0.5	0.8	0.8	0.5	0.8	0.2
4	1000	1000	1000	1000	1000	1000
5	1000	1000	3.0	1000	1000	8.0
6	100	100	2.0	100	100	4.0
7	0.5	10	1.0	1.0	10	1.0
8	0.1	0.1	0.1	0.1	0.1	0.1

-For each simulation, the damping coefficient b is updated once immediately following heel strike before foot flat, and at the inflection points of the input torque.
-A damping coefficient of 1000 simulates a very high stiffness that would occur if the damper were locked by closing the valve completely. Practically, the damping coefficient would be limited by the leakage across the cylinder and valve seals.

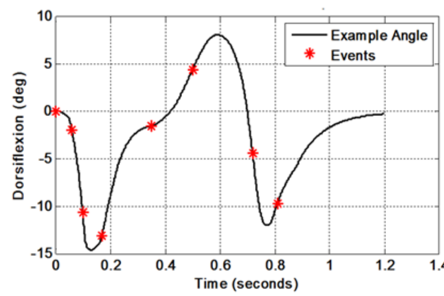

Fig. 2. Identifiable gait events that mark the changes in the damping coefficient shown on an example adaptive ankle angle plot.

In comparing the kinematic data of each transtibial amputee's passive prosthesis to able-bodied data, it is evident that key adaption characteristics are missing. Specifically, after heel strike during controlled planatarflexion response proportional to the ground slope is observed in the able-bodied and adaptive prosthesis, but not in the passive prosthesis. The able-bodied data of the ankle angle increases throughout controlled dorsiflexion to a maximum ankle angle. Through the plantarflexive torque the adaptive prosthesis is able to leverage the summation of foot spring and damper to closely match the able-bodied data. During this same phase, the spring foot only exhibits nearly the same behavior on each slope. Lastly, the adaptive foot/ankle prosthesis returns to the neutral position after toe-off, by means of the return spring in parallel with the modulated damper. This characteristic is not required in the passive spring foot because its neutral position does not vary.

In reviewing the tuned damping coefficients in Table 1, a ramp-type function is observed. The initial damping coefficient is low to allow for ground slope adaptation, but preventing the toe from slapping the ground. After initial adaptation, the damping coefficient increases to effectively lock the position towards the end of controlled dorsiflexion. Just before toe-off, the coefficient decreases to allow for increased dorsiflexion (comparable to able-bodied data). On declines, the coefficient decreases earlier in order to allow

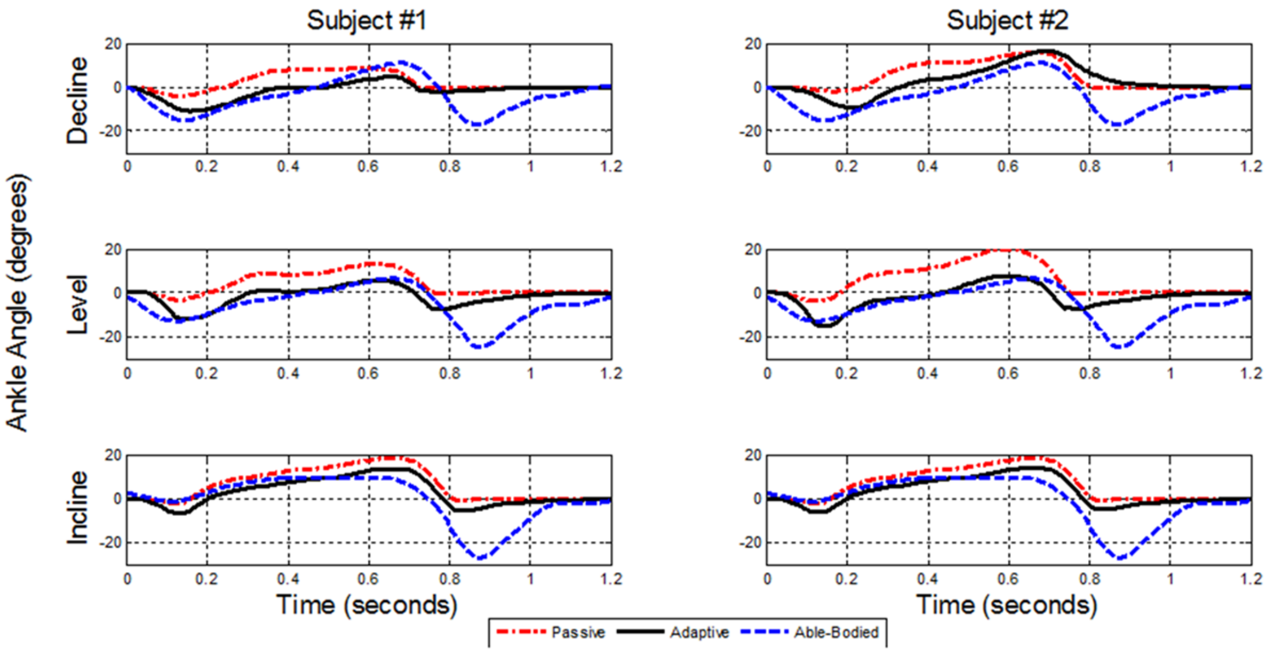

Fig. 3. Adaptive ankle simulation compared with the subject's passive spring foot/ankle and able bodied data [12] on level ground and ten-degree inclines and declines.

for increased dorsiflexion, accommodating the slope.

It is important to note that the overall stiffness of the semi-active ankle prosthesis cannot be larger than the stiffness of the carbon fiber foot. If the ankle angle is greater than that of the desired trajectory of a given slope, the angle could not be corrected. For this reason, it was found that the effective ankle stiffness could be lowered via a low damping coefficient on heel strike, to identify the ground slope and correctly modify the gait profile during the rest of the stance phase. In this manner, it is shown that the simulated ankle angle better represents the ankle angle of an able-bodied individual than a passive prosthesis with no damping.

Implementing this approach would leverage a supervisory controller to monitor activities and terrain conditions. The position/velocity/torque response of the impedance-based ankle prosthesis would encode whether the user is walking on level, declined, or inclined surface. A state controller could then be used to switch the damping coefficients based on the identifiable gait events. These controllers would use on-board sensors that monitor the position and pressure of the hydraulic chamber (ankle torque and position) as well inertial measurements.

This study presents a starting point for the development of an intelligent passive prosthesis that has the ability to modulate its internal damping characteristics to conform to the ground slope at each step.

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