Strategies to reduce the configuration time for a powered knee and ankle prosthesis across multiple ambulation modes

Ann M. Simon, Nicholas P. Fey, Suzanne B. Finucane, Robert D. Lipschutz, Levi J. Hargrove

Abstract—**Recently developed powered lower limb prostheses allow users to more closely mimic the kinematics and kinetics of non-amputee gait. However, configuring such a device, in particular a combined powered knee and ankle, for individuals with a transfemoral amputation is challenging. Previous attempts have relied on empirical tuning of all control parameters. This paper describes modified stance phase control strategies—which mimic the behavior of biological joints or depend on the instantaneous loads within the prosthesis—developed to reduce the number of control parameters that require individual tuning. Three individuals with unilateral transfemoral amputations walked with a powered knee and ankle prosthesis across five ambulation modes (level ground walking, ramp ascent/descent, and stair ascent/descent). Starting with a nominal set of impedance parameters, the modified control strategies were applied and the devices were individually tuned such that all subjects achieved comfortable and safe ambulation. The control strategies drastically reduced the number of independent parameters that needed to be tuned for each subject (i.e., to 21 parameters instead of a possible 140 or approximately 4 parameters per mode) while relative amplitudes and timing of kinematic and kinetic data remained similar to those previously reported and to those of non-amputee subjects. Reducing the time necessary to configure a powered device across multiple ambulation modes may allow users to more quickly realize the benefits such powered devices can provide.**

Keywords—lower limb amputation; mechanically active prosthesis; prosthesis control; transfemoral amputee

I. INTRODUCTION

The vast majority of individuals with a transfemoral amputation use mechanically passive prosthetic legs. An advanced passive prosthesis, such as a microprocessor-controlled knee joint, uses intrinsic control in which onboard sensors detect gait phase and mechanical joint impedances are modified accordingly. While microprocessor-controlled knees allow transfemoral amputees to walk using less energy than when walking with non-microprocessor-controlled knees [1], they cannot provide the net positive mechanical power needed during many activities of daily living, such as ascending stairs or standing up from a seated position.

Powered lower limb prostheses capable of delivering near physiological power at the knee and/or ankle have been recently developed [2-5]. These devices often use finite state machine control in combination with either a kinematic or an impedance-based control model. Other state-of-the-art approaches use artificial reflexes or complementary motion estimation [6, 7]. Since motors provide virtual stiffness and damping characteristics, these prostheses are more configurable than their passive counterparts. After being configured for a given subject, such devices enable both transfemoral and transtibial amputees to more closely mimic the kinematics and kinetics of individuals without amputations during walking [4, 8] and across other ambulation modes [9-11].

The powered knee and ankle prosthesis designed by Vanderbilt University [2], is one of the first prostheses to incorporate two motorized joints. However, configuring both these joints to work together is challenging. Initial attempts, in one transfemoral amputee by Sup et al. [12], relied on empirical tuning (i.e. a combination of feedback from the user and visual inspection of the joint angle, torque, and power data) to configure all of the available impedance parameters. As a result, little is known on how these individually tuned parameters vary between different transfemoral amputees. In addition, the configuration process may benefit from improvements to the intrinsic control system. For example, physiologically-inspired control strategies that modulate joint impedance parameters in a natural and continuous manner are likely more effective and comfortable than step changes in parameter values, which have been used previously [13]. Furthermore, control strategies that are not tuned with respect to time may allow individuals to use the device at their own preferred pace, and to alter their pace as they become more familiar with the prosthesis.

Our goal was to develop and test modified intrinsic control tuning strategies for a powered knee and ankle prosthesis for five ambulation modes. We implemented stance phase control strategies that mimic biological joints or depend on the instantaneous loads within the prosthesis. We hypothesized that these strategies would (i) reduce the initial accommodation period across various modes and (ii) reduce the number of variables that must be empirically tuned for each user.

II. METHODS

A. Powered Prosthesis Control Strategies

The powered knee and ankle prosthesis used in this experiment was designed and fabricated at Vanderbilt University [2]. The prosthesis consisted of two brushless DC

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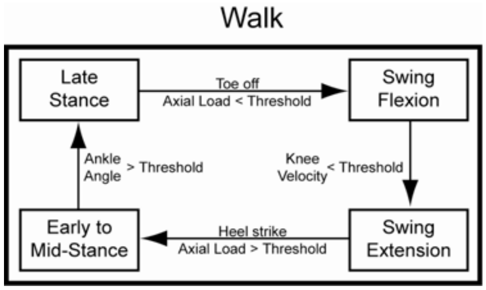

Fig. 1. Diagram of finite state machine for level-ground walking.

motors and was designed to provide up to 90 Nm of torque at the knee and 100 Nm at the ankle [2]. With a custom carbon fiber foot and shoe, total mass was 4.5 kg. The prosthesis was controlled using a finite state machine in which each state generated torque commands for the knee and ankle joints according to an impedance-based model (1):

$$\tau_i = -k_i(\theta_i - \theta_{di}) - b_i\dot{\theta}, \tag{1}$$

where i was an index corresponding to the knee or ankle, τ was the commanded torque, θ was the joint angle, and $\dot{\theta}$ was the joint angular velocity. The three impedance parameters were stiffness, k, equilibrium angle, θ_d, and damping coefficient, b. Five unique state machines were developed, corresponding to the following ambulation modes: level-ground walking, ramp ascent (10º slope), ramp descent (10º slope), stair ascent, and stair descent. The state machine architecture for each mode was similar to previous designs [2, 9, 10] but only included four states (Fig. 1). Stance was divided into two states: early through mid-stance and late stance. Swing was divided into two states: swing flexion and swing extension. Thus a total of 140 parameters could be tuned for each subject: 120 impedance parameters (2 joints, 3 impedance parameters per joint in each state, 4 states per mode, and 5 modes) and 20 event triggers (4 event triggers between states per mode for 5 modes).

The impedance control system differed from those in previous reports primarily in that parameters were not constrained to remain constant within a state. Instead, two stance phase control strategies modulated impedance parameters as a function of either joint position or axial load within the prosthesis. These strategies either mimicked biological joint responses or allowed subjects to control the rate of power generation or dissipation, with the intention of reducing the number of parameters that required tuning and smoothing the prosthesis response between states.

1) *Control Strategy 1: Impedance as a Function of Joint Angle*

Previously, ankle stiffness of non-amputees throughout the stance phase of walking (from 20% to 70% of stance phase) was measured as a function of ankle angle (2) [14]:

$$k_{ankle} = W\ x\ \left(13.6\frac{m^2}{s^2}\ \theta_{ankle} + 1.6\frac{rad\ m^2}{s^2}\right), \tag{2}$$

where k represented joint angular stiffness (Nm/rad), and W represented the subject's body mass (kg). Equation (2) was constrained to increase k_{ankle} throughout stance phase only as the ankle dorsiflexed. This strategy was applied during the following modes and states:

- Level-ground walking, ramp ascent, and ramp descent during early through late stance.

2) *Control Strategy 2: Impedance as a Function of Prosthesis Load*

A second strategy was to modulate joint impedance as a function of the axial load increase or decrease within the prosthesis. Joint impedance during a state was modulated based on load within the prosthesis according to the rate-based equation (3):

$$p_i = C_i\ \left(\frac{F - F_{Initial}}{F_{Initial} - F_{Final}}\right)\left(p_{i,Initial} - p_{i,Final}\right) + p_{i,Initial}, \tag{3}$$

where i was an index corresponding to the knee or ankle, p represented the impedance parameter to be modulated, and $p_{Initial}$ and p_{Final} were the desired initial and final values of the parameter within a state. F represented the instantaneous axial load in the prosthesis; $F_{Initial}$ and F_{Final} were the values of this parameter expected at the beginning and end of the state, respectively. C represented the rate at which the impedance parameter changed as a function of load. The value of C was constrained to be greater than or equal to 1 and p_i constrained to be between $p_{i,Initial}$ and $p_{i,Final}$. The faster the prosthesis user transferred weight onto or off their prosthesis the faster the corresponding impedance parameter would change. Likewise, slower transfer of weight resulted in slower changes in the impedance parameter.

This strategy was applied during the following:

- Level-ground walking, ramp ascent, and ramp descent during late stance; modulated k_{knee}, $\theta_{d\ knee}$, and $\theta_{d\ ankle}$ for reduced knee stiffness, powered plantar flexion and initiation of swing as force decreased.
- Reciprocal gait stair ascent during early through mid-stance; modulated $\theta_{d\ knee}$ for knee power generation as force increased.
- Stair ascent during late stance; modulated changes to $\theta_{d\ ankle}$ for powered plantar flexion as force decreased.
- Stair descent during early through mid-stance; modulated changes to $\theta_{d\ ankle}$ for controlled ankle dorsiflexion as force increased.

B. *Experimental Protocol*

Three individuals with transfemoral amputations (Table I) participated after providing informed consent. All subjects were community ambulators and had previous experience walking on the powered prosthesis. Subjects were fit with the powered device (Fig. 2) by a certified prosthetist, and a

TABLE I. SUBJECT DEMOGRAPHICS

Subject	**Gender**	**Age (years)**	**Time Post-Amputation (years)**	**Weight (kg)**	**Height (m)**
1	Male	56	43	185	1.80
2	Male	64	37	190	1.75
3	Female	22	4	115	1.60

Fig. 2 Transfemoral amputee using the powered knee and ankle prosthesis to ascend a ramp (left) and stairs (right).

therapist provided assistance at all sessions. Prior to testing state machine parameters that were dependent on subject weight were scaled. Other initial state machine impedance parameters and event triggers were set to nominal values determined during pilot tests across subjects.

Subjects began by walking between parallel bars to become comfortable using the device. When walking outside of the bars, subjects wore either a harness or gait belt for safety. Input from the subject, clinicians, and engineers was used to tune the prosthesis for all modes. For each patient, across all modes, impedance parameters were altered as necessary to achieve comfortable, safe ambulation.

For walking and ramp ascent, we focused on ensuring that the prosthetic knee and ankle were appropriately positioned for heel strike, the amount of powered plantar flexion during late stance was appropriate, and the prosthetic foot had adequate ground clearance during swing. We did not specifically tune the device for stance phase knee flexion during walking and ramp ascent. For ramp descent, subjects were directed to allow stance phase knee flexion by "riding" the prosthesis down. We focused on ensuring the prosthetic knee and ankle supported subjects throughout stance phase.

For stair ascent, subjects were initially instructed to only ascend one stair, starting with the prosthesis. This preliminary practice was performed because all subjects normally ascended stairs with their own passive prostheses in a step-to manner, starting each step with their sound side. We gradually increased prosthetic knee and ankle power until subjects were comfortable with both the task and the amount of power the prosthesis provided. Subjects then ascended a 4-step staircase using a reciprocal gait. We focused on ensuring that subjects used the prosthesis to assist them up the stairs rather than pulling themselves up with the handrails or using excessive vaulting strategies in their intact limb [15]. We also focused on providing appropriate stair clearance during swing phase. For stair descent, subjects were also instructed to descend the staircase using a reciprocal gait. We focused on ensuring that subjects had proper foot placement on each stair, were

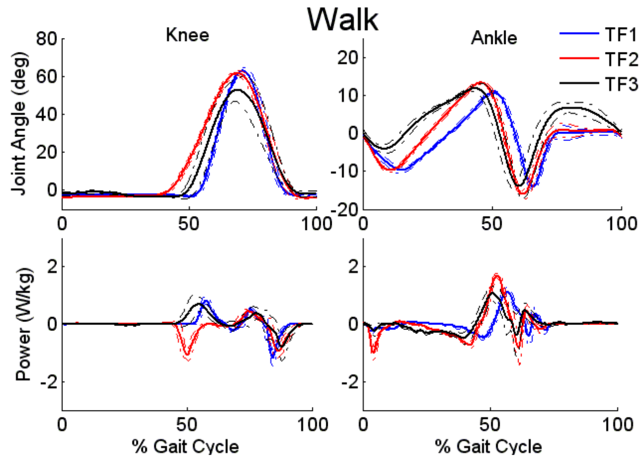

Fig. 3. Prosthesis joint angles and powers for level-ground walking. Individual subject averages and standard deviations are displayed. Joint power data are normalized by subject mass (kg).

supported by the prosthesis throughout stance phase, and were not relying on excessive use of the handrails.

An important outcome of the study included recording the total number and type of independent parameters that were modified for each subject in each mode. Once all ambulation modes were tuned, five trials of each ambulation mode were performed. For all trials, we collected mechanical sensor data from the powered prosthesis, sampled at 500 Hz, including axial load, knee and ankle joint position, joint velocity, and motor current. Data were segmented from heel strike to heel strike using the axial load in order to create average plots of prosthesis joint angles and powers. Joint powers were low pass filtered using a 20 Hz cutoff frequency and normalized to total body weight while wearing the powered prosthesis.

III. Results

A. Tuning Strategy

All subjects were able to walk over level ground using the initial nominal values of the state machine. Key changes which made subjects more comfortable walking included modifying stance phase knee stiffness and prosthesis swing timing. For two of the three subjects, we also modified the final knee equilibrium angle of equation (3) during late stance for improved swing clearance. Several parameters did not need modification across subjects, including variables from equation (3), such as the ankle equilibrium rate, C, during late stance phase and the initial and final desired knee and ankle equilibrium angles. Overall only 6 of the 28 available parameters (24 impedance parameters and 4 event triggers) were modified for walking mode. As a part of the tuning strategy, all parameter changes for level-ground walking were automatically populated for ramp ascent. Subjects were able to ascend ramps with the same parameters as walking, further reducing the overall configuration time as no additional tuning was necessary.

Not all subjects were able to descend ramps on their first attempt. It took time to tune stance phase knee stiffness and

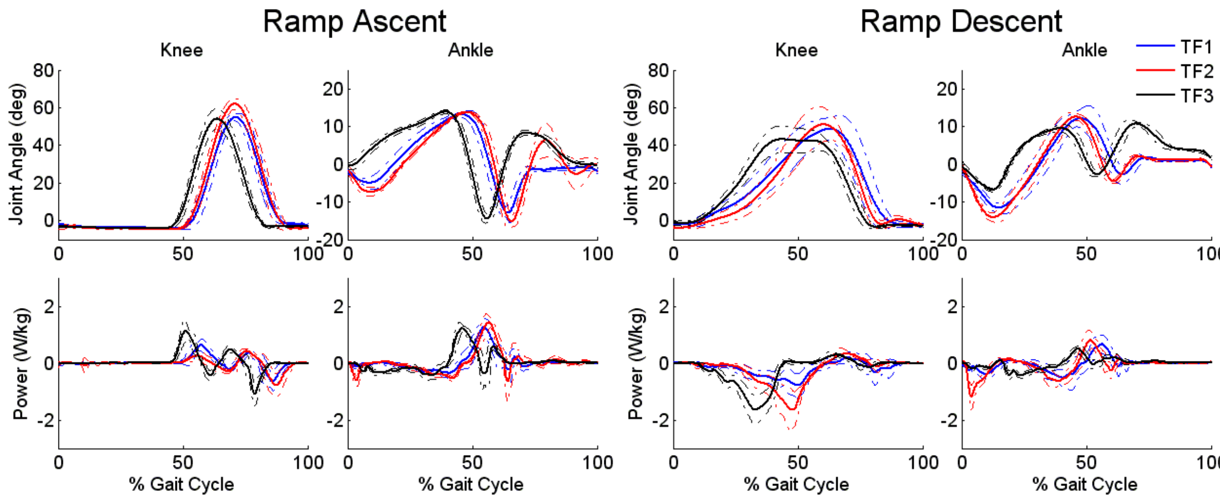

Fig. 4. Prosthesis joint angles and powers for (A) ramp ascent and (B) ramp descent. Individual subject averages and standard deviations are displayed. Joint power data are normalized by subject mass (kg).

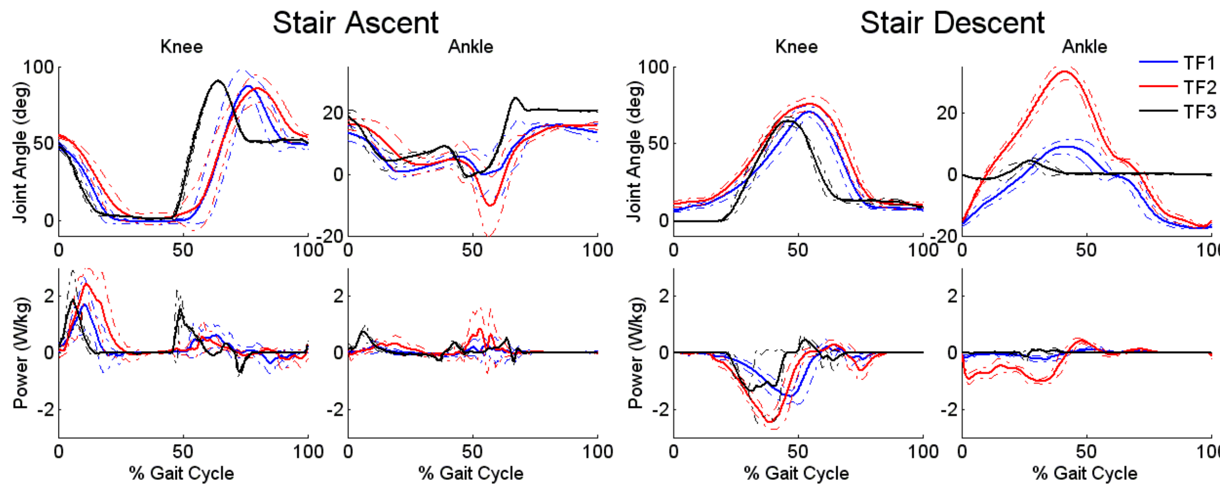

Fig. 5. Prosthesis joint angles and powers for (A) stair ascent and (B) stair descent. Individual subject averages and standard deviations are displayed. Joint power data are normalized by subject mass (kg).

damping and for each subject to practice stance phase knee flexion and to be able to "ride" the prosthesis down the incline. Initial state machine parameters for the knee in stance phase made it either too difficult to bend the knee (Subject 3) or too easy (Subject 2). Stance phase parameters were individually tuned for each subject, so that the powered knee and ankle prosthesis appropriately supported their weight and allowed the knee to flex in a controlled manner. Overall 6 of the 28 impedance parameters were tuned for ramp descent.

All subjects were able to ascend stairs using a reciprocal gait after knee power was increased to support their full weight. Several variables from equation (3) that were updated prior to testing (e.g. the final prosthesis load during stance phase of stair ascent was set to full body weight) and others that were automatically populated upon entering stance phase (e.g. the initial stance phase knee equilibrium angle during stair ascent was set to the current knee angle) did not need modification across subjects, Key changes for all subjects included modifying stance phase knee stiffness and the knee equilibrium rate, C, of equation (3) to provide a comfortable rate of power generation. Swing extension ankle equilibrium angles were modified such that subjects had appropriate placement of the prosthesis on the next stair. In total, 4 of the 28 parameters were modified for stair ascent. Stair descent with a reciprocal gait required 5 independent parameter changes. Key changes included stance phase knee and ankle stiffness and equilibrium angle during the swing extension phase.

B. Joint angles and powers

Prosthesis joint angle and power are shown for walking (Fig. 3), ramp ascent/descent (Fig. 4), and stair ascent/descent (Fig. 5). Walking and ramp ascent kinematics and kinetics demonstrate controlled plantar flexion, controlled dorsiflexion, and powered plantar flexion. While climbing stairs, subjects had similar knee kinematics to non-amputee subjects, a smooth development of knee power in early to mid-stance, and a burst of ankle power in late stance. Ramp descent and stair descent kinematics (Figs. 4 and 5) demonstrate subjects' ability to use stance phase knee flexion during these modes.

IV. Discussion

Using the collection of intrinsic control tuning strategies drastically reduced the number of impedance parameters that had to be tuned for each subject to achieve five distinct modes of ambulation (i.e., to 19 independent parameters out of a possible 140). Subjects were able to comfortably ambulate through all modes within the session, indicating relatively short configuration times. Starting with the same set of nominal state machine parameters for each subject and automatically propogating changes from one state or mode to another (e.g. all parameter changes for level-ground walking were also made for ramp ascent) contributed to the reduction in number of independent parameters that needed adjustment. Stance phase ankle stiffness mimicked physiological ankle stiffness [14]; ankle stiffness increased as the ankle dorsiflexed throughout stance phase during level walking, ramp ascent, and ramp descent. In all modes, impedance parameter changes based on axial load allowed subjects to control their rate of power generation or dissipation. During reciprocal stair ascent this feature was especially important. Prosthesis knee and ankle power was not generated until the user was ready for it; the timing of this stance phase power generation was dependent on when the subject shifted his or her weight onto the prosthesis.

Replacing a static control parameter with our novel control strategy (i.e. impedance as a function of prosthesis load) may have increased the number of possible parameters that had to be tuned for each subject. In practice, however, many of the variables were updated with expected values prior to testing (e.g. the final prosthesis load set to either body weight), automatically populated upon entering into the state (e.g. the initial stance phase knee equilibrium angle set to the current knee angle), or invariant across subjects (e.g. many of the impedance parameter rates, C, did not need subject-specific tuning). Furthermore, several of the initial and final values of the impedance parameters involved in equation (3) were set to equivalent values of the preceding and subsequent states, thereby further reducing the number of parameters that needed to be tuned.

Allowing for variable impedance in each state did not replace all necessary parameter changes for all modes. Many of the static parameter changes that were necessary were similar for all subjects and included several swing phase parameters, such as knee stiffness and damping during walking and ramp descent as well as ankle equilibrium angle during stair ascent. The ankle equilibrium angle during stair ascent required only slight changes across subjects to ensure that the prosthesis was appropriately positioned for the following heel strike. A potential reason for the differences in swing parameters is variability in the residual limb lengths across subjects, while the distance between the prosthetic knee and ankle was fixed (a constraint of the current device). Other parameters, such as knee stiffness during stair ascent, stair descent, and ramp descent required larger changes. The initial values of knee stiffness were not normalized by subject body weight. A larger group of subjects of various weights would provide more appropriate initial values of knee stiffness parameters, which may further reduce the number of independent variables requiring tuning. Knee stiffness could be normalized to weight in a similar manner to ankle stiffness.

The kinematic and kinetic data demonstrated relative amplitudes and timing that were similar to non-amputee subjects [16, 17]. The kinematic results are also similar to those published for one transfemoral amputee [2, 9, 10]. Our study shows that similar results can be achieved with multiple subjects and with a reduced amount of empirical impedance parameter tuning for each subject.

A potential limitation of this study is that subjects were not initially trained to reduce compensatory movements learned in response to their own passive prostheses (e.g. hip extension during heel strike to lock the prosthetic knee into extension) that are not necessary with the powered prosthesis. Additional training with a powered device would be required for subjects to eliminate these tendencies and to fully utilize the capabilities of powered devices. Another possible limitation of this study is that the subjects had some (albeit limited) previous experience using this device, which may have further reduced the number of necessary parameter modifications. In this study, all subjects started their tuning session from the same set of nominal impedance parameters. Future studies, in transfemoral amputees who do not have any experience ambulating with a powered device, will indicate what parameters need to be modified for completely naïve subjects.

V. Conclusion

Robotic lower limb prostheses can generate positive mechanical power at the knee and/or ankle joint. This greatly increases the number of ambulation modes that may be restored to amputees; however, empirically modifying the prosthesis response for individual users across all modes becomes challenging. We have developed strategies for tuning intrinsic control parameters together with a set of initial parameter values that reduce the amount of tuning needed to accommodate the device to the user. Reducing the configuration time necessary before transfemoral amputees can use a powered prosthesis may allow them to more quickly appreciate the benefits such devices can provide.

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