KNEE SWING-INITIATION AND ANKLE PLANTAR FLEXION CONTROL USING AN ACTIVE PROSTHESIS ACROSS WALKING SPEEDS AND USERS

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INTRODUCTION

Lower-limb amputees commonly experience abnormal gait characteristics, chronic pain and joint disorders [for review, see 1]. These behaviors are attributed to the absence of muscles and use of prostheses that inadequately restore muscular functions. Moreover, the prevalence and/or intensity of these behaviors generally increase as the level of leg amputation increases [1]. Therefore, advancing prostheses to better emulate the biological system is important, in particular for individuals with a high level of amputation such as above-knee amputees.

The large majority of available prostheses are mechanically-passive devices. Conversely, muscles produce, dissipate or transfer power about the hip, knee and ankle to satisfy energetic requirements of locomotor tasks [e.g., 2]. Recently developed mechanically-active (i.e., powered) prostheses [e.g., 3, 4] have the capacity to more closely approximate the biological system. However, more effective and generalizable control strategies governing the power transfer to each user are needed. Furthermore, strategies that are biomimetic and generalizable across users as well as user-initiated modulations of ambulation speed would be beneficial.

The purpose of this study was to develop a generalizable control strategy of an active knee and ankle prosthesis and demonstrate its performance across various walking speeds and amputee users.

METHODS

An active prosthesis was controlled using a finite state machine and impedance models of the knee and ankle [4]. The models consisted of an angular stiffness, damping and equilibrium position in each state. Within this framework, we implemented four modifications to control stance.

Based on experiments of non-amputee walking [5], we facilitated increasing ankle stiffness during controlled dorsiflexion as a linear function of ankle angle, normalized to user weight. Also based on data from non-amputees [6], we decreased knee stiffness during terminal stance as a linear function of decreasing axial shank force. Similarly, knee swing initiation and powered ankle plantar flexion were controlled by changes of their equilibrium positions as linear functions of shank force.

A rate-based equation, containing straightforward tuning constants, was developed to modify a given impedance parameter, \( p_i \), as a function of decreasing axial shank force, \( F \) (i.e., a load cell measurement).

\[
p_i = C_i \times \left( \frac{F - F_{\text{initial}}}{F_{\text{initial}} - F_{\text{final}}} \right) x (p_{\text{initial}} - p_{\text{final}}) + p_{\text{initial}}
\]

Two tuning constants (in bold) were proportionality constants, \( C_i \), which scaled each rate-of-change, and final “desired” impedance values, \( p_{\text{final}} \). Other constants were either detected at state changes or constrained by the state machine.

We tested 4 amputees (3 traumatic, 1 sarcoma; 43±20 years age; 23±21 years post-operation). Amputation level varied from knee disarticulation to the proximal third of the femur. All had Medicare functional classification levels 3 or higher and a minimum of 10 hours experience using the device. For each user, the prosthesis was tuned at their comfortable speed. Subjects walked at 3 self-selected speeds (comfortable, slow and hurried). No additional adjustments were made across speeds. An average of 20 strides was collected per condition. The onboard inertial measurement unit and potentiometers were used to estimate walking speed [7] and joint angles. Knee and ankle power were computed as the product of commanded joint torque and joint velocity.
RESULTS AND DISCUSSION

Seamless modulations of walking speed relative to the comfortable condition were facilitated (i.e., increases and decreases of $30\pm8\%$). Slow, comfortable and hurried speeds were $0.58, 0.82$ and $1.1\pm0.09$ m/s, respectively. As speed increased, increases of peak ankle dorsiflexion and plantar flexion were observed (Fig. 1). Also for increasing speed, negative and positive ankle power increased during stance (Fig. 2). These trends and magnitudes are comparable to non-amputees at these speeds [e.g., 8], especially from mid to late stance. The plantar flexors provide body support and forward propulsion in the second half of stance, and increase their contributions as speed increases [2]. These data suggest the prosthetic ankle behaved similarly.

The stance to swing transition was timed appropriately and performed smoothly since it was a function of a decreasing axial force. This was supported by knee flexion angles initiated earlier as speed increased (Fig. 1). However, net knee power over this transition was positive at the two slowest speeds (Fig. 2), contrasting non-amputee data [6, 8]. These differences suggest the prosthetic knee (i.e., with one actuator per degree-of-freedom) may not consistently replace all knee muscle functions (e.g., gastrocnemius and rectus femoris functions, described in [2]). However, these data suggest this control strategy provided appropriate knee flexion kinematics across a range of speeds and users.

These results were produced with little variation of stance phase tuning constants across subjects. Proportionality constants governing decreasing knee stiffness, swing initiation and powered plantar flexion did not vary (1.0, 1.0 and 1.5, respectively). In addition, final plantar flexion equilibrium angles were $10-12\%$. Final knee flexion equilibrium angles were an exception as they had larger variations (45-70\%). These differences were needed to achieve ground clearance for the various amputation levels while the distance between the knee and ankle was fixed (i.e., a current device constraint) [4].

CONCLUSIONS

These rate-based control strategies of an active prosthetic knee and ankle enable amputees to alter their speeds, while displaying kinematic and kinetic profiles supportive of scalable prosthesis function. The generalizability of these strategies was further demonstrated by minimal tuning across users.

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REFERENCES


Figure 1: Group averaged prosthetic joint angles and velocities. Comfortable speed standard deviations are shaded.

Figure 2: Group averaged prosthetic joint powers. Comfortable speed standard deviations are shaded.