Contributions of Knee Swing Initiation and Ankle Plantar Flexion to the Walking Mechanics of Amputees using a Powered Prosthesis

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***Abstract*— Recently developed powered prostheses are capable of producing near-physiological joint torque at the knee and/or ankle joints. Based on previous studies of biological joint impedance and the mechanics of able-bodied gait, an impedance-based controller has been developed for a powered knee and ankle prosthesis that integrates knee swing initiation and powered plantar flexion in late stance with increasing ankle stiffness throughout stance. In this study, five prosthesis configuration conditions were tested to investigate the individual contributions of each sub-strategy to the overall walking mechanics of four unilateral transfemoral amputees as they completed a clinical 10-m walk test using a powered knee and ankle prosthesis. The baseline condition featured constant ankle stiffness and no swing initiation or powered plantar flexion. The four remaining conditions featured knee swing initiation alone (SI) or in combination with powered plantar flexion (SI+PF), increasing ankle stiffness (SI+IK), or both (SI+PF+IK). Self-selected walking speed did not significantly change between conditions, although subjects tended to walk the slowest in the baseline condition compared to conditions with swing initiation. The addition of powered plantar flexion resulted in significantly higher ankle power generation in late stance irrespective of ankle stiffness. The inclusion of swing initiation resulted in a significantly more flexed knee at toe off and a significantly higher average extensor knee torque following toe off. Identifying individual contributions of intrinsic control strategies to prosthesis biomechanics could help inform the refinement of impedance-based prosthesis controllers and simplify future designs of prostheses and lower-limb assistive devices alike.**

I. INTRODUCTION

More than 600,000 individuals are currently living in the U.S. with a major lower limb amputation and this number is projected to increase dramatically [1]. Typically, amputees wear non-microprocessor-controlled mechanically passive knees that provide swing phase resistance through friction, pneumatic, or hydraulic mechanisms [2]. Microprocessor-controlled mechanically passive knees use on-board sensors to provide advanced dynamic control of the knee during gait. Some studies comparing microprocessor-controlled and non-microprocessor-controlled knees have found improvements in metabolic energy expenditure with these advanced devices [3, 4], while others found few biomechanical advantages [e.g., 2]. Amputees using microprocessor-controlled knees still exhibit gait asymmetries, such as reduced prosthetic side stance time and increased loading of the intact limb [5]. A known limitation of passive prosthetic devices is that they are unable to add net positive work to the user.

This work was supported by the US Army's Telemedicine and Advanced Technology Research Center, grant number 81XWH-09-2-0020.

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In able-bodied gait, muscles at the ankle produce net positive work over the course of the gait cycle [6]. Ankle plantar flexors have been shown to provide body support and forward propulsion as well as contribute to knee swing initiation [7]. Muscles at the knee, such as the quadriceps, provide body support and braking. Therefore, following transfemoral amputation, the loss of muscles spanning the knee and ankle and the functions they provide [6] may contribute to the asymmetries observed in amputee gait. It is unknown how various ways of adding energy at the knee and ankle affects the walking mechanics of transfemoral amputees.

Powered microprocessor-controlled prostheses are capable of providing net mechanical power to the user through motorized joints. Some powered devices that are currently on the market include the Ossur Power Knee [8] and the BiOM Ankle [9]. Other devices containing a motorized knee and/or ankle are still in development [10, 11]. Decreased metabolic cost of walking was shown in group of transtibial amputees using the BiOM compared to a conventional prosthesis [9]. With the rapid advances in powered prosthetic hardware and the encouraging preliminary results that demonstrate the potential benefits of these devices, there is a need to better understand how the control of these devices affects the biomechanics of the user.

Previous investigations of biological ankle impedance during walking have provided insight into potential improvements to impedance-based controllers of powered prostheses [12, 13]. Ankle stiffness was found to increase as a linear function of ankle angle over the mid-stance phase of walking at a self-selected speed in able-bodied subjects [12]. A subsequent study implemented this strategy in a powered knee and ankle prosthesis along with additional impedance-based control strategies that provided powered ankle plantar flexion and knee swing initiation in terminal stance as a function of decreasing axial force [14]. Using this integrated control strategy, transfemoral amputees were able to walk at varying speeds with kinematics and kinetics that closely resembled healthy gait, particularly at the ankle.

The individual contributions that knee swing initiation, powered plantar flexion, and increasing stance phase ankle stiffness provide to the overall mechanics and clinical outcomes of walking in transfemoral amputees are unknown.

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These findings could motivate the refinement of impedance-based prosthesis controllers and potentially aid in the simplification of future designs of prostheses and lower-limb assistive devices. The purpose of this study was to use a powered knee and ankle prosthesis to determine these contributions in transfemoral amputees. We tested three hypotheses: knee swing initiation would result in a more flexed knee at toe off and a more extensor knee torque following toe off; increasing ankle stiffness in stance would lead to higher plantar flexion torque in mid to late stance; powered plantar flexion would result in increased ankle power generation in late stance.

II. Methods

A. Powered Prosthesis

Subjects wore a powered knee and ankle prosthesis [11]. The device featured thirteen on-board mechanical sensors, including a vertical axis load cell, a 6-axis inertial measurement unit, and sensors that measured knee and ankle position, velocity, and motor current. The prosthesis was controlled using an impedance-based model to produce joint torque according to (1):

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

where i was an index corresponding to the knee or ankle, τ was the torque produced at the joint, θ was the joint angle and $\dot{\theta}$ was the joint velocity. The impedance parameters were joint stiffness, k, damping coefficient, b, and equilibrium angle, θ_e. Walking was controlled using a finite state machine and divided into four states: early to mid-stance, late stance, swing flexion, and swing extension. Transitions between states were controlled by setting thresholds on particular mechanical sensors (Fig. 1). Most impedance parameters within each state were set to constant values derived from existing literature [11] and previous empirical tuning sessions. Four modified intrinsic control strategies were used to modulate remaining impedance parameters during stance.

1) Increasing Ankle Stiffness throughout Stance

Ankle stiffness, k_{ankle}, was modulated during controlled dorsiflexion, spanning both early to mid-stance and late stance according to (2) [12]:

$$k_{ankle} = W\ x\ (0.237\,\theta_{ankle} + 0.028), \qquad (2)$$

where k represented ankle stiffness (Nm/deg) and W represented the user's body mass (kg). Ankle stiffness was set to a low constant value from heel strike through foot-flat, then followed (2) throughout the remainder of stance phase. Ankle stiffness was constrained to always increase and was capped at a value of 6 Nm/deg, as to not exceed the torque capability of the device.

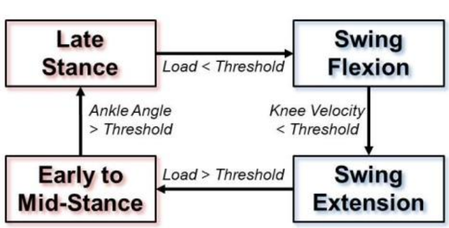

Figure 1. Diagram of the finite state machine used to control walking.

2) Knee Swing Initiation in Late Stance

Knee equilibrium angle, $\theta_{e\,knee}$ was modified in late stance as a function of decreasing prosthetic axial force, F, according to (3) [14]:

$$\theta_{ei} = C\left(\frac{F - F_{Initial}}{F_{Initial} - F_{Final}}\right)\left(\theta_{ei,\ Initial} - \theta_{ei,\ Final}\right) + \theta_{ei,\ Initial}, \qquad (3)$$

where i corresponds to the knee or ankle, $\theta_{ei,\ Initial}$ and $\theta_{ei,\ Final}$ were desired initial and final values of the equilibrium angle in late stance and C represented the rate at which the parameter changed as a function of decreasing load. For knee swing initiation, $\theta_{e\,knee,\ Initial}$ was set to 0° and $\theta_{e\,knee,\ Final}$ was set to 60-75° knee flexion (value determined from previous tuning sessions with each subject).

3) Powered Plantar Flexion in Late Stance

Ankle equilibrium angle, $\theta_{e\,ankle}$, was modified in late stance as a function of decreasing prosthetic axial load according to (3). For all subjects, $\theta_{e\,ankle,\ Initial}$ was set to 0° and $\theta_{e\,ankle,\ Final}$ was set to 12° plantar flexion.

4) Decreasing Knee Stiffness in Late Stance

Analogous to equilibrium angle, knee stiffness, k_{knee}, was modified in late stance as a function of decreasing prosthetic axial load according to (3). Across all conditions, $k_{knee,\ Initial}$ was set to 3-5 Nm/deg (value determined from previous tuning sessions with each subject), and $k_{knee,\ Final}$ was set to 0.4 Nm/deg.

B. Experimental Protocol

Four individuals with amputations (three transfemoral and one knee disarticulation, all male, 29-65 years old and 84-110 kg) participated in the study. All subjects had previous experience walking with the powered prosthesis (15 hours or more). Knee and ankle impedance parameters had been previously tuned for each subject to ensure adequate swing clearance and comfortable walking at their self-selected pace. The subjects gave informed consent to a Northwestern University Institutional Review Board approved protocol. A certified prosthetist and a licensed physical therapist were present for all sessions.

Subjects walked on five combinations of modified stance phase intrinsic control strategies (Table 1) that featured combinations of swing initiation (SI), increasing ankle stiffness (IK), and powered plantar flexion (PF) in a randomized order. Constant ankle stiffness was defined as the value of (2) evaluated at 8° dorsiflexion, which was 5 Nm/deg for three subjects and 6 Nm/deg for one subject. Subjects were given time to accommodate to each prosthesis configuration prior to testing. Three trials of a clinical 10m-

Table 1. Prosthesis Configuration Conditions

	Knee	Ankle	
	Late Stance	*Late Stance*	*Early to Mid-Stance and Late Stance*
Baseline	$\theta_{d\,knee}$=0°	$\theta_{d\,ankle}$=0°	Constant k_{ankle}
SI	Swing initiation	$\theta_{d\,ankle}$=0°	Constant k_{ankle}
SI+IK	Swing initiation	$\theta_{d\,ankle}$=0°	Increasing k_{ankle}
SI+PF	Swing initiation	Powered plantar flexion	Constant k_{ankle}
SI+IK+PF	Swing initiation	Powered plantar flexion	Increasing k_{ankle}

walk test were administered [15] to measure self-selected walking speed and to collect prosthetic joint kinematics and kinetics.

Baseline and SI conditions were compared to investigate the effects of knee swing initiation. SI vs. SI+IK and SI+PF vs. SI+PF+IK comparisons were made to investigate the effects of increasing ankle stiffness. Likewise, SI vs. SI+PF and SI+IK vs. SI+PF+IK comparisons were made to investigate the effect of powered plantar flexion. Prosthesis mechanical sensor data were sampled at 500 Hz. Joint kinematics were segmented from heel strike to heel strike using the load cell. Paired t-tests (α=0.05) were performed to objectively analyze differences in self-selected walking speed, knee angle at toe off, and average positive and negative knee and ankle torque and power between matched conditions.

III. Results

On average, subjects walked the slowest 0.89 (0.15 SD) m/s during the baseline condition than in any condition with swing initiation (SI: 0.96 (0.14) m/s; SI+IK: 0.98 (0.16) m/s; SI+PF: 1.03 (0.11) m/s; SI+PF+IK: 0.99 (0.18) m/s), although no significant differences were found. Providing knee swing initiation resulted in positive (flexion) knee torque and power in late stance, contrary to the negative (extension) torque and power observed in the baseline condition (Fig. 2). Swing initiation resulted in a significantly greater knee angle at toe-off compared to the baseline condition (Baseline: 21° (7.9°); SI: 42° (4.9°); p=0.002). In stance phase (Fig. 3), average knee extension torque was significantly higher in the baseline condition than in the SI condition (p=0.001). No difference was found in average negative knee power between baseline and SI conditions. In swing phase (Fig. 3), average knee extension torque was significantly greater in the swing initiation condition compared to baseline (p=0.005), as was average negative knee power (p=0.026). Conditions with increasing ankle stiffness (SI+IK, SI+PF+IK) showed lower dorsiflexion torque initially and a more gradual rate of plantar flexion

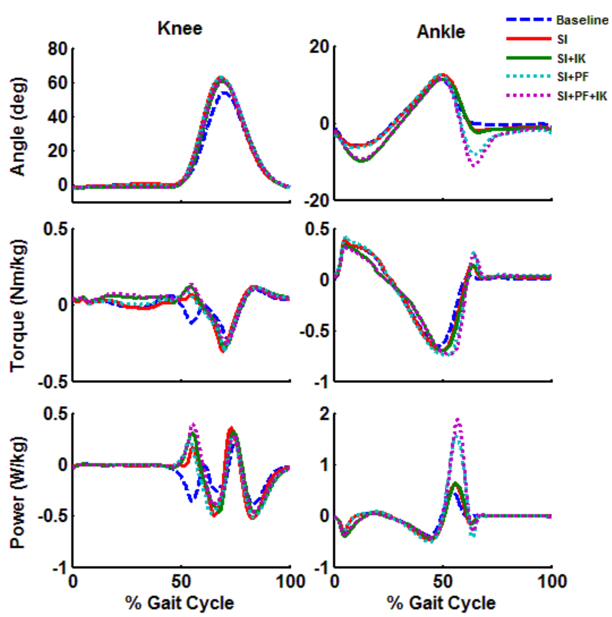

Figure 2. Group average kinematics and kinetics (n=4).

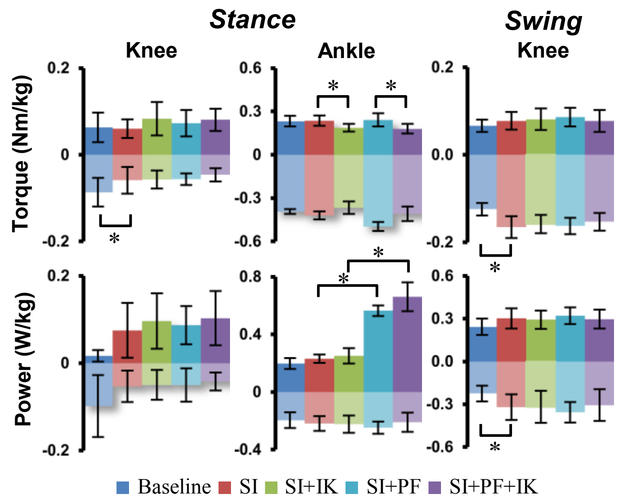

Figure 3. Average stance phase positive (dark solid) and negative (light solid) knee and ankle torque and power, and swing phase knee torque and power. Positive torque values indicate knee flexion and ankle dorsiflexion torque, negative values indicate knee extension and ankle plantar flexion torque. Positive power values indicate power generated, negative values indicate power absorbed. Statistically significant differences are denoted with an asterisk.

torque production throughout mid-stance compared to conditions with constant ankle stiffness (SI, SI+PF) (Fig. 2). Increasing ankle stiffness resulted in significantly lower average stance phase dorsiflexion torque compared to constant stiffness (Fig. 3) for both conditions without powered plantar flexion (p=0.006) and with powered plantar flexion (p=0.004). A lower average stance phase plantar flexion torque in conditions with increasing ankle stiffness approached significance for both conditions without powered plantar flexion (p=0.090) and with powered plantar flexion (p=0.059). Providing powered plantar flexion resulted in greater ankle plantar flexion angle and power generation in late stance (Fig. 2). Stance phase ankle power production was significantly higher with powered plantar flexion than without powered plantar flexion (Fig. 3) for both conditions with constant ankle stiffness (p<0.001) and with increasing ankle stiffness (p<0.001).

IV. Discussion

In this study, we measured prosthesis kinematic and kinetic differences attributed to providing knee swing initiation, increasing ankle stiffness, and powered plantar flexion. Providing powered plantar flexion led to significantly higher positive power at the ankle in late stance, with peak values (Fig. 2) that were comparable to previously reported values in able-bodied gait [16, 17]. These results suggest higher contributions of the ankle joint to the forward propulsion and support of the body [7] in these conditions relative to other conditions.

The addition of knee swing initiation resulted in a significantly greater knee flexion angle at toe off and a more extensor knee torque following toe off. These results indicate an earlier and more appropriately timed transition to swing. Previous work has shown that initial swing phase extensor torque (provided primarily by the rectus femoris muscle) increases with faster walking speed [18]. Therefore, the earlier transition to swing and increased initial swing extensor torque provided by knee swing initiation may help amputees walk faster with kinematics and kinetics that more closely resemble healthy gait.

A significantly lower stance phase dorsiflexion torque acting from initial contact to mid-stance (approximately 25% of the gait cycle) was produced in increasing ankle stiffness conditions compared to constant stiffness conditions (Fig. 2, Fig. 3). In a simulation study of able bodied walking [19], dorsiflexion muscles were found to negatively contribute to forward progression over the first 10% of stance phase, suggesting that lower ankle dorsiflexion torque early in the gait cycle would be more beneficial (i.e., provide less braking). Subjects also commented that they perceived a smoother progression from heel strike to foot flat using conditions with increasing ankle stiffness. No difference in plantar flexion torque was observed in late stance between conditions with constant and increasing ankle stiffness, contrasting our hypothesis. However, constant ankle stiffness values (5 or 6 Nm/deg) were nearly identical to the increasing ankle stiffness value (6 Nm/deg) for the last 30% of stance phase, so it is reasonable that we would not see significant differences in late stance plantar flexion torque between conditions. A lower, more conservative constant stiffness value may have been more appropriate.

Limitations of the study include the small subject pool. Testing more subjects could help prove or disprove results that approached significance such as the potential differences in plantar flexion torque and self-selected walking speed. In addition, the implemented increasing ankle stiffness strategy was based on work that estimated ankle impedance from 20%-70% of stance phase in walking at a self-selected pace [12, 13]. A recent study has investigated ankle impedance over the entire gait cycle using a wearable ankle robot, and found time-varying ankle impedance behavior during double support and swing phases of gait [20]. Our preliminary results in combination with these findings could improve the increasing ankle stiffness intrinsic control strategy. Future work may involve analyzing sound side walking mechanics to further determine the individual contributions of knee swing initiation, increasing ankle stiffness, and powered plantar flexion to the overall mechanics of walking with a powered knee and ankle prosthesis.

Acknowledgment

The authors thank Elizabeth Halsne, CPO, and Suzanne Finucane, PTA, for their assistance.

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