Does the Impedance of Above-knee Powered Prostheses Need to be Adjusted for Load-carrying Conditions?*

Andrea Brandt, Ming Liu, and He (Helen) Huang, *Senior Member, IEEE*

Abstract— **Powered knee prostheses provide substantial advantages for amputees compared to traditional passive devices during basic walking tasks (i.e. level-ground, stairs, ramps), but the impedance control parameters are fixed. For environments that differ from the well-controlled setting of the clinic, amputees must compensate their gait patterns because fixed control parameters ideal for walking on level ground in the clinic do not meet real-life task demands. Load carriage is one instance where fixed control parameters may lead to undesired gait patterns and potentially result in injury. To evaluate the importance of impedance control parameters for different walking tasks, we tested one above-knee amputee walking using an experimental powered prosthesis under four walking conditions. The amputee walked with and without added mass with both load-specific and non-specific impedance control parameters. The load-specific parameters significantly reduced the amputee's intact-leg compensations, asymmetry, and perceived exertion compared to the non-specific control parameters. Powered lower limb prostheses that modulate impedance control parameters for load-carrying tasks may improve the gait performance, safety, and comfort of amputees.**

I. Introduction

The number of individuals with major lower extremity amputations in the US is estimated at over 600,000 and is expected to more than double by 2050 [1], increasing the demand for reliable prostheses. Recently developed powered lower limb prostheses are a considerable improvement from energetically passive prostheses because they enable more natural and smooth gait while reducing the amputee's metabolic cost and functional limitations during locomotion [2, 3]. In attempt to mimic the biological system, most powered knee prostheses modulate the stiffness of the prosthetic knee joint within predefined gait phases using finite state impedance control [3, 4]. The performance of these powered prostheses rely heavily on impedance control parameters and must be fine-tuned for individual amputee users by an expert (e.g. clinician). Traditionally, the impedance control parameters are only tuned during level-ground walking in the clinic, where the environment is highly controlled [5]. This tuning procedure has an inherent and significant limitation because the prosthesis control parameters are fixed once the amputee leaves the clinic. These fixed impedance parameters may not be suitable for real-world environments that differ from the clinic. As a result, the inability for current powered knee prostheses to adapt their control parameters to various environments may contribute to the amputee user's limited mobility, stability, and independence [6, 7].

A widely-investigated mobility concern that alters the dynamics of the human-prosthesis system is load carriage. Daily tasks such as carrying school/work bags, groceries, toddlers, and substantial weight gain (e.g. pregnancy) compromise amputees' gait and may lead to slip and fall injuries and/or device abandonment. Even carrying 10 pounds of groceries is markedly difficult for above-knee amputees due to functional limitations of their prostheses [8]. Yet, current powered knee controllers rely on one fixed impedance parameter set to control prosthesis knee mechanics for all level ground walking, despite variability in the dynamics of the human-prosthesis system during varied locomotor tasks. Numerous biomechanical analyses of able-bodied people report *symmetric* load carriage adaptations (e.g. increased joint work) to transport increased mass [9, 10]. However, unilateral below-knee amputees walking with energetically passive ankle prostheses adapt *asymmetrically* to load carriage by mainly and significantly increasing power generation and absorption in the contralateral (i.e. intact) knee [11]. This increased reliance on the intact leg can lead to numerous issues in lower limb amputees, most notably osteoarthritis and scoliosis [12]. In above-knee amputees, amputation also correlates with an increased risk of falling [6, 13], and intact joint dynamics are more difficult to replicate with two missing joints (i.e. knee and ankle) versus one joint in below-knee amputees. Hence, we expect even greater compensatory behavior on the intact leg in above-knee amputees during load carriage. If this behavior is in fact observed in above-knee amputees walking with powered knee prostheses, we may be able to reduce this undesired behavior through impedance parameter tuning.

The purpose of this preliminary study was to examine (1) the gait changes that occur when an above-knee amputee walks with added mass using an experimental powered knee prosthesis and (2) the effects of impedance parameter tuning on these gait changes. The results from this study may lead to improved adaptive tuning systems [14] and/or tuning procedures used in the clinic.

II. Methods

A. Participant and Equipment

One above-knee amputee gave informed consent to participate in our 2-day protocol approved by the Institutional Review Board of University of North Carolina

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A. Brandt, M. Liu and H. Huang are with the NCSU/UNC Department of Biomedical Engineering, NC State University, Raleigh, NC, 27695; University of North Carolina at Chapel Hill, Chapel Hill, NC 27599.

Corresponding author: Andrea Brandt (e-mail: ambrand3@ncsu.edu, phone: 919-515-8541).

at Chapel Hill. The subject (height: 183 cm, weight: 66 kg) walked on a split-belt force-measuring treadmill (1000 Hz, Bertec Corp., Columbus, OH, USA) with our experimental powered knee [21] and carbon high-performance foot (1E56 Axtion, Otto Bock, Germany). We recorded full-body kinematics using an 8-camera motion capture system (42 markers, 100 Hz, VICON, Oxford, UK). We then calculated sagittal plane kinematics, kinetics, joint work, and spatiotemporal parameters using Visual 3D (C-Motion, Inc., Germantown, MD, USA). We manually adjusted inertial parameters of the body and prosthesis segments in Visual 3D to correct for the mass of the backpack and powered prosthesis. Prior to this study, the subject trained with our experimental powered prosthesis for more than 10 hours and until he felt comfortable walking on the treadmill at 0.6 m/s without relying on the treadmill handrails or a harness for body-weight support. During testing, we placed weight in a backpack and secured it flush against his back to induce an inertial change similar to daily backpack-carrying tasks.

B. Experimental Design

Day 1: The expert tuned the impedance control parameters as the subject walked with the backpack containing no load (i.e. original impedance) and 20% of his body weight (i.e. load-specific impedance). For each backpack load, the subject walked for 3 minutes with a standard set of impedance parameters to get accustomed to the testing condition, the expert adjusted the parameters, and then we set the parameters to original levels and loaded/unloaded the backpack for the second weight condition. Similar to in-clinic tuning, the expert tuned each condition based on both visual observations of the subject's gait performance (including the prosthesis knee angle) and subjective feedback. The powered knee prosthesis impedance parameters (i.e. equilibrium position, stiffness, damping) remained constant during each gait phase.

Day 2: We tested both impedance parameter sets (i.e. original and load-specific) at each backpack load (i.e. no load and 20% body weight). We recorded 3 2-minute trials at each of the 4 conditions. We blinded the subject to testing objectives and conditions. The subject rested for 3 minutes, or longer at the subject's request, between each trial. To minimize the potential of fatigue/training time confounding our results, we randomized the sequence of the 12 trials. To assess relative subjective preference, we recorded the user's rate of perceived exertion (RPE) after each trial using the modified Borg CR10 scale [15]. RPE incorporates psychological as well as physiological feedback from locations such as joints and muscles; however, RPE scores are not reflective of metabolic cost.

C. Data Analysis

We analyzed 20 consecutive gait cycles at the end of each trial to account for the subject acclimating to experimental conditions. We examined the significance of using load-specific impedance control parameters using two-way ANOVA ($p<0.05$) and post-hoc paired t-tests ($p>0.05$) to examine within-factor significance. We treated the 3 repeated trials at each testing condition as independent measurements. To evaluate symmetry, we calculated the subject's stance time asymmetry index: $\frac{(Stance_{Intact} - Stance_{Amputated})}{(Stance_{Intact} + Stance_{Amputated})*0.5} \times 100$ [16].

III. Results

A. Powered Knee Prosthesis

The prosthesis knee angle profiles were similar to a normative knee profile when the tuned impedance parameter sets matched the backpack load. However, when the amputee walked with load-specific parameters without the backpack (i.e. mismatched), we observed a flexed peak stance extension angle (6±0.3, Fig.1). When he walked with the original parameters and the backpack, we observed a hyperextended peak stance extension angle (-5±2.9, Fig. 1). Both load ($F(1,8)=12.91$, $p=0.001$) and impedance ($F(1,8)=38.46$, $p<0.001$) significantly affected the peak knee extension angle during stance extension.

Compared with the original impedance control parameters, the expert tuning resulted in load-specific parameters with an increased equilibrium position (14%) in initial double support and equilibrium position (22%) and stiffness (20%) in single support. The effects of this tuning is evident in the prosthesis knee angle profile (Fig. 1).

B. Whole body (Temporal Parameters, Subjective Feedback)

The amputee walked most symmetrically with the two sets of impedance control parameters that were tuned specifically for the no-load and load condition (Fig. 2, filled point characters). A significant interaction between the two factors (i.e. load condition and impedance parameter set) ($F(1,8)=30.1$, $p<0.001$) supports the importance of having impedance control parameters appropriate for different load-carriage conditions. The control parameters had a significant main effect on temporal asymmetry ($F(1,8)=5.5$, $p=0.047$), but load did not. One reason the subject walked asymmetrically with mismatched load and impedance parameters is because with the original parameters, the intact-leg stance time was longer with the added load. With load-specific parameters, the prosthesis stance time was shorter without the added load. Additionally, while the matched control parameters resulted in comparable intact-

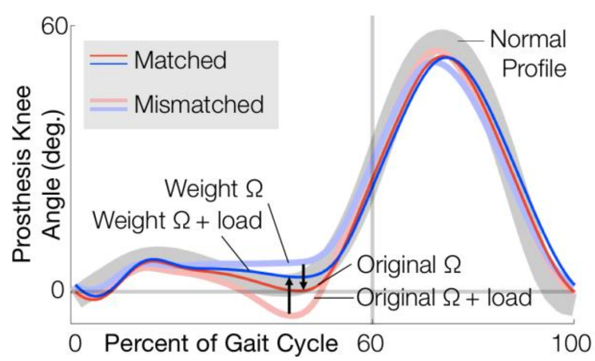

Figure 1. Powered prosthesis knee profile for each condition. Line style shows whether the impedance control parameters matched the load being carried by the subject. Arrows show favorable changes. Grey line indicates normative profile.

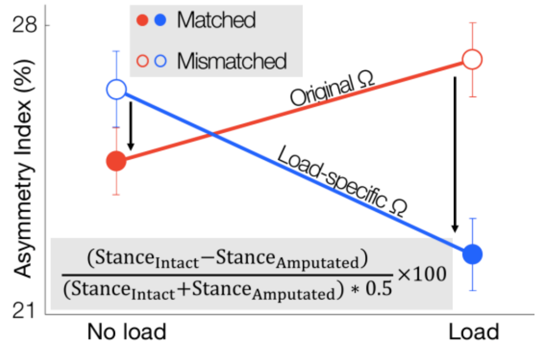

Figure 2. Interaction between backpack load (x-axis) and impedance control parameters (colors) on stance time asymmetry. Error bars show 1 standard deviation. Arrows show favorable changes. Asymmetry index equation shown on plot.

leg initial double support time, the subject spent significantly more time in this phase when carrying load (versus no load) with the original parameters (p=0.001) and significantly less time when carrying no load (versus with load) and load-specific parameters (p=0.002).

With load, the subject gave lower RPE scores for the matched load condition with load-specific parameters (5±1.5, versus 7±1.2 with original parameters). Without load, RPE was lowest with the mismatched no load condition with load-specific parameters (3±0.6, versus 4±0.6 with original parameters).

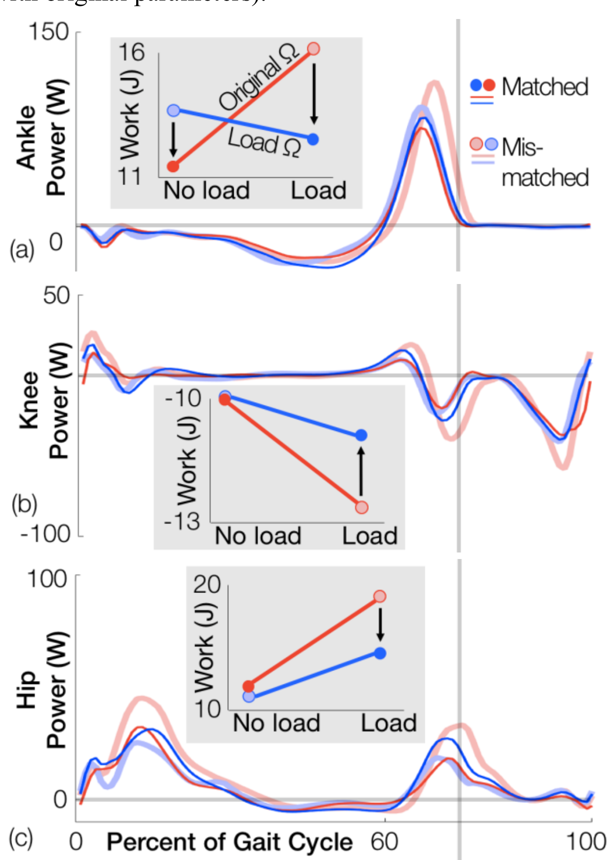

Figure 3. Joint power and work for each intact-side joint. Insets show interaction between backpack load (x-axis) and impedance control parameters (colors) on joint work. Arrows show favorable changes.

C. *Intact leg (Joint Work)*

Total mechanical power generated on the intact side was greater than the prosthesis side for all four conditions. The greatest changes on the intact side are visible in the positive ankle power, negative knee power, and positive hip power (Fig. 3). For the ankle, total positive work decreased when the load and impedance parameters were matched (Fig. 1a). For the knee and hip, there were observable changes only when the subject was carrying the load– negative knee work decreased and positive hip work decreased (p=0.008). The interaction between the load and impedance parameters was only significant at the hip ($F(1,8)=5.52$, p=0.047).

IV. Discussion

In this study, we examined the effect of impedance parameters on gait mechanics during load-carriage and found that impedance parameters tuned to specific load conditions is important. These specific, tuned parameters, versus original parameters, effectively allowed the amputee to walk more symmetrically and load the intact leg less.

The change in the prosthesis knee angle can be expected because the momentum of the trunk contributes to the passive progression of the knee joint from flexion to extension. As the mass of the body increased, it created more torque about the knee. Posterior muscle action typically maintains the knee in a neutral (0°) position for maximum stability during weight bearing [17]. However, with the added load, the prosthesis torque with the original impedance parameters was not sufficient to prevent hyperextension similar to the biological knee joint. For each load-carriage condition, the expert tuned the impedance control parameters to "optimize" gait performance. One factor the expert (and clinicians) may prioritize during tuning is matching the prosthetic knee angle to a normative knee angle profile. This is reflected in the final tuned knee profiles (Fig. 1, dark lines).

Unilateral amputees' existing asymmetric gait [18] likely attributes to higher-than-average energetic and mechanical costs for walking [19] and intact-joint overuse issues. Increasing gait asymmetry may result in higher metabolic and mechanical costs and a faster progression of overuse comorbidities (e.g. osteoarthritis). The mismatched prosthetic knee impedance and backpack load conditions negatively affected symmetry, supporting the idea that load-specific parameters are critical. Many factors such as the subject's typical asymmetry and walking speed may have contributed to the discrepancy between the asymmetry index and RPE of both matched conditions (Fig. 2, filled point characters).

At the mismatched no-load condition, the prosthetic knee joint stiffness was unfavorably high with the impedance parameters tuned with the load. As seen in Fig. 1, the prosthetic knee joint remained flexed during stance extension because the subject's body mass alone was not sufficient to fully extend the knee. This smaller prosthetic knee joint range of motion coincided with the subject landing on his intact foot more quickly and transferring his body weight from the prosthetic to intact leg more quickly. This quicker transition may have caused the prosthetic-leg stance time to shorten, which is already and commonly shorter relative to the intact leg [20]. Moreover, the subject may have needed the extra propulsive ankle force (Fig. 3c, light blue) to overcome the higher stiffness during initial double support and achieve the preferred amount of knee stance flexion.

To meet the energy demands of the backpack load, the subject compensated more with his intact side when the impedance parameters were mismatched to the extra load (i.e. original parameters). For example, the subject transferred his body weight to his intact leg more quickly potentially because he felt unstable on the powered knee. Loading of the intact leg immediately follows prosthetic-leg single support phase, when the impedance parameters had the greatest effect. Therefore, is seems that the impedance parameters had the most immediate and greatest effects in hip power on the intact leg during loading response. Additionally, the subject may have generated more total power on the intact side during the propulsion phase to "catch up" his intact leg on the fixed-speed treadmill to compensate for longer stance time on the intact leg.

V. Limitations

To improve the impact of this case study and confirm our observed trends in the data, the number of subjects studied should be expanded. Trial randomization is only effective with more than one subject, so this study may carry the consequences of secondary factors such as fatigue. Also, the sources of load-related compensations can only be inferred from our current data set. Further, walking over ground may reveal other adaptive and/or compensatory gait changes with load carriage. In addition to perceived effort, a fear of falling questionnaire (e.g. Falls Efficacy Scale commonly used by physical therapists [21]) modified for walking may be more indicative of subjective preferences.

VI. Conclusion

We have shown that load-specific powered knee impedance control parameters improve the gait mechanics of an above-knee amputee while carrying extra mass. The subject relied less on the intact leg with the load-specific impedance parameters and consequently exerted less perceived effort. This suggests that the fixed impedance parameters used in current powered knee prostheses may not be sufficient for daily use outside of the clinic for load carrying tasks. Above-knee amputees may experience improved gait performance, safety, and comfort with load-specific tuning in addition to traditional tuning methods and adaptive tuning alone.

Acknowledgments

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