BIOMECHANICS-CENTERED DESIGN OF ROBOTIC LOWER-LIMB PROSTHESES

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INTRODUCTION

Robotic ankle-foot prostheses can improve mobility for individuals with amputation, yet we do not know by how much, nor what designs are optimal in general or for a given individual. Improvements in metabolic energy consumption and preferred speed have been demonstrated [1], often using specialized mechanisms to conserve electrical energy [2]. Development has been centered on robotic designs themselves, however, with years of refinement required before the more interesting questions of human biomechanical response can be answered.

Systematic explorations of prosthesis design space, with a focus on human response, would generate a more rational framework for design. Ankle push-off seems a promising place to begin; increased work may improve human economy, but implies heavier motors and batteries. A quantitative characterization of this trade-off could reveal optimal characteristics.

METHODS

We used a versatile ankle-foot prosthesis testbed to explore human responses to changes in push-off work. This testbed, described in detail in [3], was actuated by a powerful motor, connected through a Bowden-cable tether to an instrumented prosthesis end-effector worn by the subject (Fig. 1).

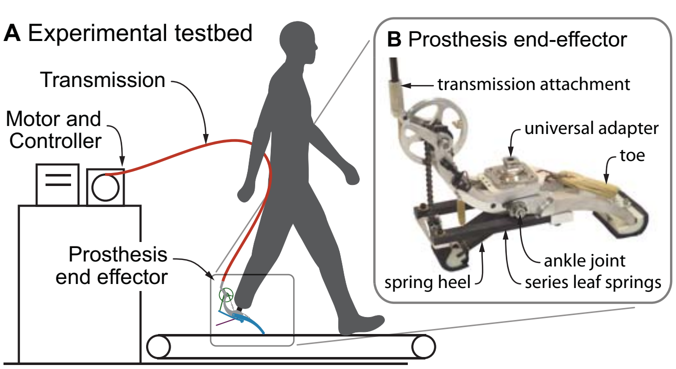

Figure 1: Testbed decouples function from embodiment.

We performed tests on a single healthy subject (N=1, 70 kg, 0.94 m leg length, 22 yrs.) walking on a treadmill at 1.25 $m \cdot s^{-1}$ wearing the prosthesis on one leg using a simulator boot [2]. The prosthesis behaved like a stiffening spring with separate constants during dorsiflexion and plantarflexion [3], resulting in an ankle-joint work loop. Plantarflexion settings were varied across conditions to generate different amounts of net mechanical work per step. We selected values that roughly corresponded to -0.5, 0, 1, 2, 3, and 4 times the net work performed by the biological ankle during walking, and called this scaling term C_w. The subject trained on all conditions one day prior to collection, all conditions lasted 10 minutes, and all were presented in random order. Procedures were approved by CMU IRB.

We measured average metabolic rate during the final 3 minutes of each condition using indirect respirometry, with quiet standing as a baseline. We measured prosthetic ankle torque and position using onboard sensors and used these to calculate prosthetic ankle power. We also calculated average net prosthesis power, defined as positive minus negative work per stride divided by average stride time. We then performed linear regression to obtain a relationship between average net prosthesis power and human metabolic rate across conditions.

RESULTS AND DISCUSSION

Increasing C_w led to increased prosthetic ankle power and work per stride (Fig. 2) as intended. This led to a decrease in metabolic energy expenditure (Fig. 3) with linear coefficient of -1.78 (R^2 = 0.97) and coefficient of performance [4] of 0.45.

This relationship may have interesting implications for the design of robotic ankle-foot prostheses. Parallel and series elasticity and escapements could minimize motor requirements, allowing ideal power and energy densities of about 0.075 W per gram and

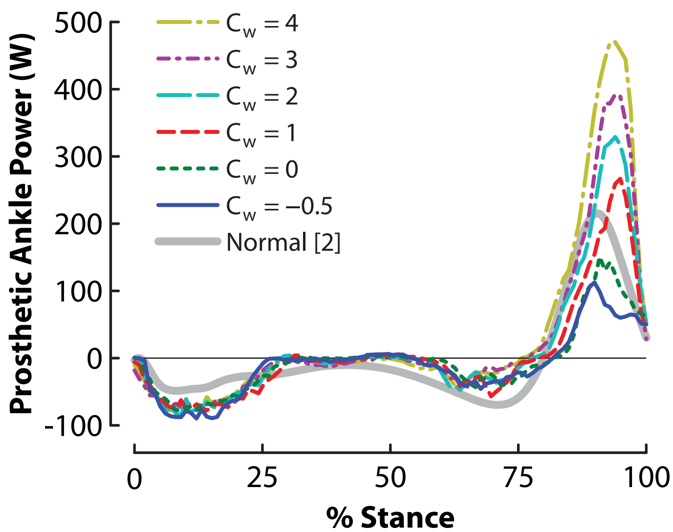

Figure 2: Robotic ankle prosthesis power during the stance period for each condition. Ankle power during normal gait from [2] provided for reference.

250 J per gram, respectively, including transmission efficiency [5]. An individual who takes 3000 steps per day would then need 13 + 11 = 24 grams of motor and battery per 1 W push-off assistance. For each gram added at the ankle, we expect a 0.015 W increase in metabolic rate [6], leading to an expected 0.34 W increase per 1 W assistance. Each 1 W of assistance would thus reduce metabolic rate by 0.34 – 1.78 = -1.54 W. In other words, for this individual, a bigger robotic prosthesis is always better. For extreme values this fit will likely break down, but apparently not for the wide range tested here, which far exceeds values for normal walking and commercial robotic prostheses. Benefits would be even greater for batteries stored at the hip or with high nominal prosthesis mass. For lower power or energy density, or more steps per day, a tipping point would be reached at which a passive device would instead be preferable. Similar implications might also be derived for robotic orthoses [7].

Our results must be taken as preliminary, however, due to the small sample size, lack of human kinetics and kinematics measurements, short training period, and use of a simulator boot. We are currently collecting a more complete data set on individuals with amputation. The low coefficient of performance may indicate the possibility of improved delivery of prosthesis work, and a more detailed study of control parameter space is warranted.

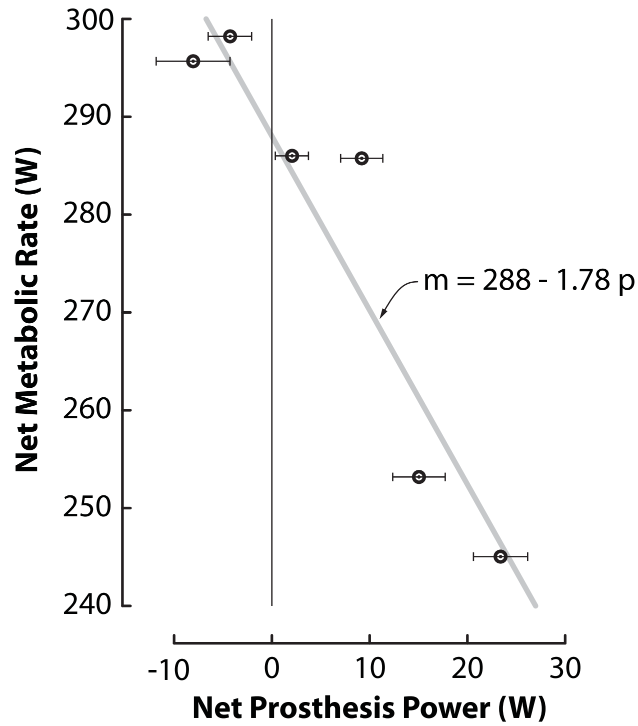

Figure 3: Net human metabolic rate vs. average net prosthesis power. Axis scales are equal. Average net prosthesis power presented with ± st. dev. Net metabolic rate during Normal walking was 223 W.

CONCLUSIONS

These results illustrate the potential for thorough exploration of biomechatronic design spaces using experimental testbeds and the types of quantitative design frameworks that can be derived with this approach. In particular, our findings suggest that ankle-foot prostheses could provide even more benefit with increased push-off work production. We think this approach could facilitate systematic, rational design of improved robotic prostheses.

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