Optimization-based predictions of walking with a prosthesis: gait kinematics, optimal prosthesis actuation, and Pareto-optimal strategies

Matthew L. Handford and Manoj Srinivasan
Mechanical and Aerospace Engineering, The Ohio State University, Columbus, OH, USA
handford.4@osu.edu, *srinivasan.88@osu.edu*

1 Introduction

Robotic prostheses have the potential to improve the quality of life of amputees by reducing the amount of energy, improving comfort, and increasing speed while performing everyday tasks [1]. However, the current methods for designing such devices involve lengthy build cycles and hand tuning of parameters, which slows progress and possibly limits the effectiveness of the device. Here, we present our work towards developing a computer simulation that can accurately predict human gait kinematics while using prostheses with multiple architectures and control strategies. Such a predictive simulation could greatly reduce the time necessary to test theories and reduce the time wasted with lengthy build cycles when researching protheses. The predictive simulations are based on the hypothesis that humans, including amputee subjects, move in a manner that minimizes the metabolic energy cost or some similar effort-like objective function. We describe systematic differences between model-predicted optimal amputee gaits for different model assumptions and compare model-predictions (kinematics, ground reaction forces, and energetics) to data from human experiments. We comment on what modeling features and parameters are required to obtain our results and how they influence the match with the experimental data.

2 Method

We have developed a muscle-driven, planar model of a person with unilateral trans-tibial amputation (TTA) using a one-segment or two-segment torque-driven ankle prostheses within the MATLAB environment as shown in figure 1. A one-segment prosthesis architecture can be used to mimic devices that use a single ankle joint such as the BiOM [4, 5]; the two-segment architecture mimics devices with separated heel and toe components such as the ankle-foot prosthesis emulator developed by Caputo and Collins [2]. Our computer model allows us to test different types of prostheses and control strategies and observe how they will affect people's gaits.

The prediction of the person's movement and prosthesis' performance is accomplished by determining the body kinematics, muscle forces, and prosthesis torques (all functions of time) that minimize different cost functions based on muscle force, work, activation, or metabolic energy. In some optimizations, the prosthesis controller is fixed, and we simply seek the optimum human movement strategy for that fixed prosthesis controller. This computational optimization is performed using a 'multiple shooting method', which discretizes and solves for both body state and muscle forces simultaneously; this method promotes better optimization convergence without requiring a good initial seed.

When we perform simultaneous optimizations of human and prosthesis, we use a cost function that is a weighted sum of a human and a prosthesis cost to find the optimal tradeoff between the human and prosthesis energy costs.

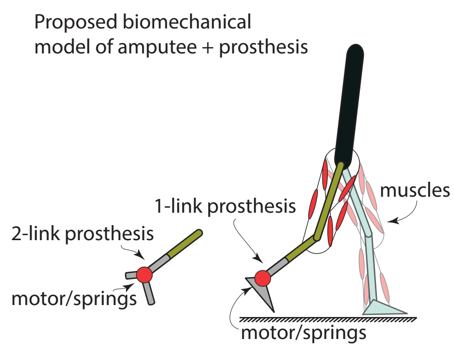

Figure 1: Schematic of computer model. Two different prosthesis designs are shown, one-segment and two-segment.

3 Selected Results

Even though we have used our model with a variety of cost functions, here, we present the results for a simple force squared cost function, defined by Cost $= \lambda \sum (F_m/F_{\text{iso}})^2 + (1-\lambda)(T_p/r)^2$, where λ is a weighting term selected at the beginning of the optimization between 0 and 1, F_m is the force in each muscle, F_{iso} is the max isometric force, T_p is the torque in the prosthesis, and r is the moment arm of the mechanism producing said torque. Our optimization is able to predict walking gaits which are kinematically similar to natural walking gaits (Figure 2) even when the optimization is started from standing (i.e., not walking) initial seeds. This is true for the modal using five different λ values (Figure 3).

This optimization can also produce a cost comparison between the cost of the person and the cost of the prosthesis based on different weights between the person and prosthesis cost functions (Figure 4).

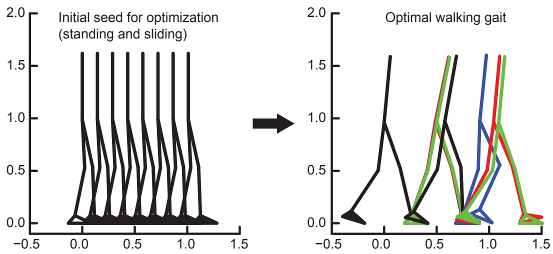

Figure 2: Basic walking motion obtained from muscle force and prosthesis torque squared optimization with single-link prosthesis model. The simulation does not require the initial seed to be close to the final solution. We used a non-physical initial seed with the person standing and sliding forward. Solid color foot = prosthesis.

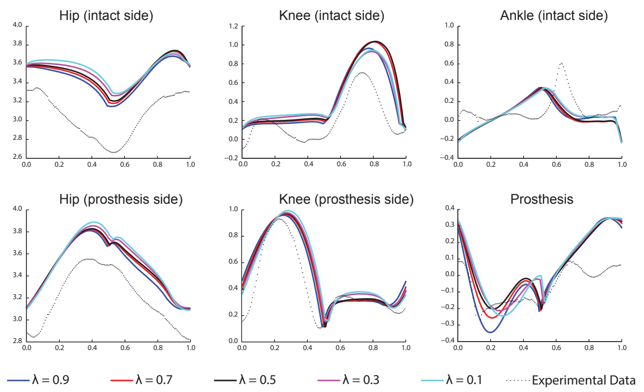

Figure 3: Model versus experiment. Joint angles of hips, knees, and ankles. Solid color lines represent the optimized simulation results for different weighting factors, λ, used. The dotted line shows experimental data from prosthesis testbed tests individuals without TTA [3].

4 Discussion

Our model produced gaits kinematically similar (at least qualitatively) to those found in experiments[3]. The differences between our simulation and experiments could be because the simulations have optimized prostheses whereas the experiments used a non-optimized controller. Other reasons could be modeling simplifications such as assuming rigid contact with the ground, modeling the prosthesis as a single link, and using a simplified cost function. These simplifications were used to ensure the results of the optimization were not heavily influenced by the hand tuning of ground contact mechanics, and to ensure the simulation would not crash due to numerical

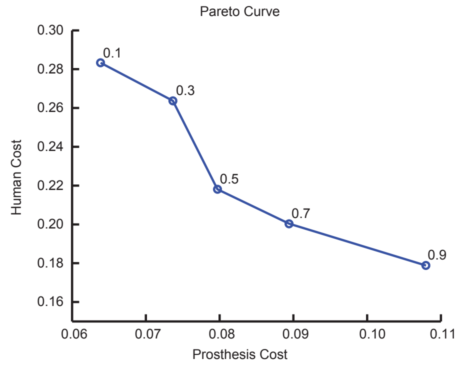

Figure 4: Pareto curve of human force squared cost per unit time and prosthesis torque squared cost per unit time (dimensionless). The points are labeled with the weighting factor, λ, used for that particular test.

issues inherent in optimization. We have also experimented with a two-link model that is capable of producing similar results but with more asymmetries present between the prosthesis and non-prosthesis sides of the model and less consistency in the minima found from the optimization. By fixing some of these consistency issues and making use of more biologically realistic cost functions, both the one and two-link models should be able to produce more natural gaits. Then, using these results, we would be able to comment on how these two prosthesis architectures affect the gait.

The Pareto curve obtained by optimization of the combined human and prosthesis cost shows that there is a trade-off between the human and prosthesis cost; that is, a decrease in the cost of one is accompanied by an increase in the other. This trade-off is similar to those in prior experiments [3] that show how powered prostheses can reduce the metabolic cost of the user by increasing prosthesis positive work.

References

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