Optimal Control of an Electromechanical Above-Knee Active Prostheses with Energy Regeneration

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INTRODUCTION

Walking with a transfemoral prosthesis requires up to 65% more energy than able-bodied walking [1], possibly due to compensatory movement strategies. Powered prostheses have been developed to address this problem, but performance is limited by the available battery power. During walking, there are periods of net negative work in the knee and ankle, and the net work done by the knee is negative [3]. This suggests the possibility of controlled storage and release of energy, and transfer of energy from knee to ankle.

In this paper, we describe an electromechanical above-knee active prosthesis with energy storage and regeneration. The goal is to design a prosthesis that can control the trajectory of the knee and ankle with minimal or possibly zero use of external energy. An optimal control approach will be used to determine how the energy use depends on how closely the device tracks able-bodied gait kinematics.

METHODS

The electromechanical system is schematically shown in Fig. 1. Two DC motors control movement in the knee and ankle through ideal mechanical gear systems. Springs parallel to the motors were included in the model but not used in the present study. A single ultracapacitor is used to supply and store energy for the two-joint system. The voltage supplied to each motor is controlled by a four-quadrant power converter between capacitor and motor. The inputs of the system are the knee torque (M_K) and the ankle torque (M_A) which were obtained from normal gait data [3]. The dynamic model of the system has five state variables: position and velocity in each joint, and the capacitor charge. The two control inputs are the transformer ratios u_K and u_A.

The model parameters included the torque constant, electrical resistance, and moment of inertia of a 24V DC motor (Pittman, 14201 series) with a gear ratio of 350. The capacitance of the ultracapacitor was 100F.

Open loop optimal control was used to determine the controls that minimize a two-part cost function, consisting of tracking error and energy loss over one gait cycle:

$$F = W_1 \int_0^T \left[\left(\theta_K(t) - \theta_{K,0}(t)\right)^2 + \left(\theta_K(t) - \theta_{K,0}(t)\right)^2 \right] dt + W_2 \frac{Q(0)^2 - Q(T)^2}{2C}$$

Tracking error was the difference between simulated joint angles $\theta(t)$ and able-bodied joint angles $\theta_0(t)$ from [3]. Energy loss was computed from the capacitor charge Q. Similar to [2], the optimal control problem was transcribed using direct collocation, using the midpoint Euler discretization. Grid refinement showed that 50 time nodes per gait cycle was sufficient. The resulting nonlinear program was solved by IPOPT. Pareto-optimal solutions were obtained by performing optimizations with different cost function weights W_1 and W_2.

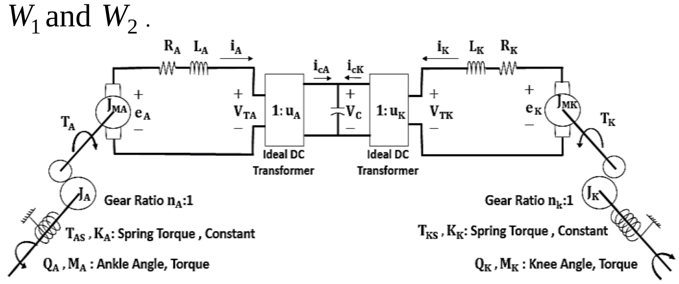

Figure 1: Schematic diagram of the electromechanical prosthetic system with motors at ankle (left) and knee (right).

RESULTS AND DISCUSSION

When the cost function is weighted appropriately, the optimal solution produces a good compromise between tracking error and energy loss. Joint angles track able-bodied data well, with a root-mean-square (RMS) of 4.78° (Figs. 2a and 2b). The energy stored in the capacitor fluctuates during the gait cycle, resulting in a net loss of 24.4 J (Fig. 2c).

Figure 2d shows the results of Pareto optimizations, indicating the trade-off between the tracking objective and the capacitor energy. At one extreme, the tracking error was nearly zero, and the associated energy loss was 94.2 J. At the other extreme, enough energy is harvested to operate the system without external energy, but the tracking error of 30° is unacceptably large. The middle solution is the compromise shown in Figures 2abc.

Table 1 shows the energy balance of the system in each of the three solutions, separated into the work delivered by the prosthetic system, the heat generated in the motors, and the change in stored energy.

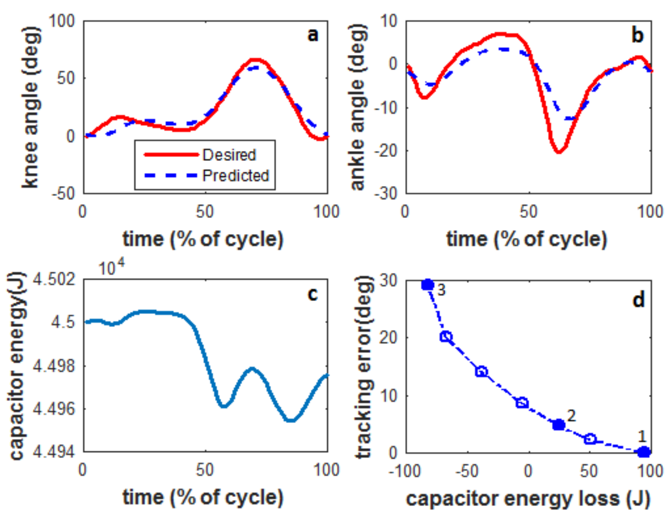

Figure 2: Joint angles (a,b) and capacitor energy (c) for a typical simulation. The entire range of possible performances is shown in (d) as a Pareto front.

Table 1: Work-energy values for the three results.

	W_{knee} (J)	W_{ankle} (J)	*Heat* (J)	ΔE_{cap} (J)
1	-14.4	18.7	90.1	- 94.2
2	-20.0	6.5	38.0	-24.4
3	-31.9	-85.2	34.5	82.5

In the solution with perfect tracking, energy is harvested from the knee, but not sufficient to power the ankle and compensate for energy dissipated as heat. The system uses about 94 J of external energy in each gait cycle. In the intermediate solution, tracking error is 4.78°, which is an acceptable performance, and external energy use is reduced to 24.4 J. In this solution, the joint movements were subtly altered, resulting in large energy savings.

A substantial energy loss occurs in the motor resistances. The optimal control approach found solutions where this loss is reduced, but it remained too large. We hope to reduce the energy dissipation by optimizing the gear ratios and by adding optimally designed passive springs and dampers to assist the motors. It may also be possible to find alternative motors with lower resistance.

The model has some limitations. Damping and friction in the gear system was neglected, and the power converter was assumed to be ideal with no energy loss. These refinements will be added in future work.

CONCLUSIONS

A model of a regenerative electromechanical above-knee prosthesis showed that energy cost could be reduced by about 74% if a well-controlled deviation from able-bodied movement is allowed. Further optimization of the design may reduce the energy cost sufficiently to operate the system without energy loss.

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