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Design and Control of an Active Electrical Knee and Ankle Prosthesis

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Abstract

This paper presents an overview of the design and control of an electrically powered knee and ankle prosthesis. The prosthesis design incorporates two motor-driven ball screw units to drive the knee and ankle joints. A spring in parallel with the ankle motor unit is employed to decrease the power consumption and increase the torque output for a given motor size. The device's sensor package includes a custom load cell to measure the sagittal socket interface moment above the knee joint, a custom sensorized foot to measure the ground reaction force at the heel and ball of the foot, and commercial potentiometers and load cells to measure joint positions and torques. A finite-state based impedance control approach, previously developed by the authors, is used and experimental results on level treadmill walking are presented that demonstrate the potential of the device to restore normal gait. The experimental power consumption of the device projects a walking distance of 5.0 km at a speed of 2.8 km/hr with a lithium polymer battery pack.

I. Introduction

The native limb generates significant net power over a gait cycle in many locomotive functions including walking, walking up stairs and slopes, running and jumping [1-8]. In the absence of net power generation, transfemoral amputees with passive prosthesis have been shown to expend 60% more metabolic energy [9] and exert three times the affected-side hip power and torque [1] when compared to healthy subjects during level walking.

To the authors' knowledge, the earliest powered transfemoral prosthesis was developed at MIT during 1970's and 1980's [10-16]. This prosthesis consisted of an electro-hydraulically actuated knee joint tethered to a hydraulic power source and utilized off-board electronics and computation. As described in [13], an "echo control" scheme was developed for gait control. In echo control, the modified knee trajectory from the sound leg is played back on the contralateral side. Popovic and Schwirtlich reported the development of an active knee joint actuated by DC motors [17]. They utilized a finite state knee controller with robust position tracking control for gait control. In [18], the design of an active ankle joint using McKibben pneumatic actuators is described. The feasibility of electromyography based position control approach for transtibial prosthesis is assessed in [19]. Although no scientific literature is available, Ossur, a prosthetics company, has recently made available a powered knee and a self-adjusting ankle. The "Power Knee" uses a control approach similar to echo control, which utilizes sensors on the sound leg. The Ossur powered ankle prosthesis, called the "Proprio Foot", does not contribute net power to gait, but rather quasi-statically adjusts the ankle angle to optimize gait.

The authors have developed a pneumatically powered knee and ankle prosthesis prototype, in which they used finite state-based impedance control that only utilizes sensors on the prosthesis itself [20]. This prototype was built to take advantage of the recent advances in monopropellant based pneumatic actuation described by [21-24]. The authors believe that the monopropellant technology in its current state is not ready for commercialization in the near-term. Current lithium-polymer batteries have an energy density approaching 200 W·h/kg [25], which enable the development of a transfemoral prosthesis with a reasonable weight and an acceptable, although limited, range of locomotion. The energy density of such batteries is expected to nearly double in the next decade (driven largely by the automotive industry's needs for electrical vehicles) [25], which will provide a more generous range of locomotion.

In this paper, the authors describe progress toward the development of an electrically powered active knee and ankle prosthesis. This prosthesis will be able to generate human-scale power at the joints and incorporate a torque-based control framework for stable and coordinated interaction between the prosthesis and the user. The paper describes the mechanical design of the prosthesis, provides an overview of the finite-state based impedance control framework, presents experimental results on a healthy subject using an able-bodied adapter, and discusses the electrical power requirements in different gait modes.

II. Design Specifications

A. Prosthesis Prototype

The active joint torque specifications were based on an 85 kg user for fast walking and stair climbing, as derived from body-mass-normalized data [1,3]. The prosthesis is capable of 90° of flexion at the knee and 45° of planterflexion and 20° of dorsiflexion at the ankle. The electric powered prosthesis prototype is presented in a labeled photograph, Fig. 1. The prosthesis is actuated by two motor-driven ball screw assemblies that drive the knee and ankle joints, respectively, through a slider-crank linkage. Each actuation unit consists of a Maxon motor (Model 148867) capable of producing 150 W of continuous power connected to a 2mm lead ball screw of 10 mm and 12 mm diameters for the knee and ankle, respectively, via Oldham couplings. The ankle actuation unit additionally incorporates a spring of stiffness 1100 N/cm in parallel with the ball screw, the purpose of which is to bias the motor's axial force output toward ankle plantarflexion, and to supplement power output during ankle push off. The resulting axial actuation unit's force versus ankle angle plot, Fig. 2, graphically demonstrates for fast walking the reduction in linear force output supplied by

the motor at the ankle through the addition of the spring. Note that the compression spring does not engage until approximately five degrees of ankle plantarflexion. Each actuation unit additionally includes a uniaxial load cell (Measurement Specialties ELPF-500L), positioned in series with the motor for force control. Both the knee and ankle joints incorporate composite plain bearings (Garlock model DU) and, for joint angle measurement, integrated precision potentiometers (ALPS RDC503013). A strain based sagittal plane moment sensor, Fig. 3, is located between the knee joint and the socket connector, which measures the moment between the socket and prosthesis. The ankle joint connects to a custom foot design, Fig. 4, which incorporates strain gages to measure the ground reaction forces on the ball of the foot and on the heel. The present prototype houses onboard signal conditioning electronics and relies on a tether for power, computation and power electronics.

The knee height of device is varied by changing the main structural tube and the clamping supports for the knee actuation unit and ankle spring. Additionally, the ankle joint and the sagittal moment load cell incorporate standard pyramid connectors for coupling the prosthesis to the foot and socket, thus enabling a high degree of adjustment in the knee and ankle alignment, as is standard in transfemoral prostheses. Combined with the custom sensorized foot (0.35 kg) and foot shell (0.24 kg) the total weight of the tethered transfemoral prosthesis is 3.8 kg, which is within an acceptable range for transfemoral prostheses, and less than a comparable normal limb segment [26]. An untethered version incorporating batteries and on-board electronics is expected to weigh less than 4.5 kg with further structural weight savings.

B. Moment and Force Sensing

The load between the user and prosthesis, and between the prosthesis and ground, is sensed in order to infer user intent and enable prosthesis control. Based on the data presented in [1] and [4], the required range of measurements was determined to be 100 Nm of sagittal plane moment and a ground reaction force of 1000 N. The sagittal plane moment is measured above the knee joint at the socket interface and the ground reaction force is measured by the sensorized foot. The location of the sensors was chosen to avoid coupling the desired measured ground reaction force and sagittal moment with the joint torques. In addition, incorporating the ground reaction load cell into the structure of a custom foot itself eliminates the added weight of a separate load cell.

The sagittal plane moment sensor shown in Fig. 3 is a low profile design to allow for the longest residual limb of the user. The mechanics of the design are based on a single beam in bending with a moment and force applied at the center. The design is a flat plate mounted on two supports on the edges parallel to the frontal plane. A central ridge along the top concentrates the load and adds rigidity to resist the frontal plane moments. On top of the central ridge is a platform that allows for a pyramid connector to be attached. Finite element analysis, using ProEngineer Mechanical, was used to minimize the overall design height and to achieve the desired strains in the load cell. The sensor was fabricated from 7075 aluminum and has an assembled weight of 120 grams including the stainless steel pyramid connector. The overall height of the sensor including the pyramid connector is 35 mm and the base is a 50 mm square. The device was calibrated for 100 Nm with $\pm 5\%$ error at full state output.

A sensorized foot, Fig. 4, is used to measure the ground reaction force and is comprised of toe and heel beams rigidly attached to a central fixture. The toe and heel portions of the foot are arranged as cantilever beams with an arch that allows for the load to be localized at the ends. The device is fabricated of 7075 aluminum and weighs 350 grams. The foot fabricated for the test subject measures 220 mm long, 56 mm wide and is 35 mm tall to the top of the

central fixture and approximates a US size 12 or EU size 46 foot. The overall dimensions and weight are similar to commercial low-profile carbon-fiber prosthetic feet, such as the Otto Bock Lo-Rider. The prosthetic foot was designed to be housed in a soft prosthetic foot shell (see Fig. 1). The device was calibrated for 1000 N with $\pm 4\%$ error at full state output.

C. Control

The control approach developed by the authors utilizes an impedance-based approach to generate joint torques [20]. Generating torques rather than positions enables the user to interact with the prosthesis by leveraging its dynamics in a manner similar to normal gait, and also generates stable and predictable behavior. In the algorithm, the knee and ankle torques, τ_i , are characterized by Eqn. 1, with a series of finite states, Fig. 5, consisting of passive spring and damper behaviors.

$$\tau_i = -k_i(\theta - \theta_{ki}) - b_i\dot{\theta} \quad (1)$$

Where k_i and b_i denote the linear stiffness and damping coefficient for an i^{th} state, respectively. Energy is delivered to the user by switching between appropriate equilibrium positions, θ_k , (of the virtual springs) during state transitions.

In this manner, the prosthesis is guaranteed to be passive within each gait mode, and thus generates power simply by switching between modes. Since the user initiates mode switching, the result is a predictable controller that, barring mode switching input from the user, will always default to passive behavior.

III. Experimental Setup

The main components of the experimental setup consist of a tethered powered prosthesis and a treadmill. The powered prosthesis is tethered to two Kepco BOP 36-12D servo-amplifiers, and a laptop computer running MATLAB Real Time Workshop for controller implementation. In addition, current and voltage sensing circuitry is employed to measure the currents and voltages passing through the knee and ankle motors. The prosthesis is tested using an able-bodied adapter on a healthy male subject, who is 1.93 m tall and weighs 86 kg, as shown in Fig. 6.

The prosthesis is tuned for the subject using the finite-state impedance approach for standing and for walking at three different walking speeds – slow, normal and fast (2.2, 2.8 and 3.4 km/hr). The parameters of the tuned modal impedance functions are presented in Tables I and II. The current and voltage to each motor and prosthesis sensor data are collected for each walking speed and standing on two separate trials lasting 100 seconds. During the standing mode, the test subject alternately shifted his weight between limbs, turned in place, and stood still. The prosthesis sensor data consists of joint positions, velocities and torques, socket sagittal plane moment and heel and ball of foot loads.

IV. Results and Discussion

A. Performance

Measured joint angles from the prosthesis' onboard sensors during level treadmill walking at 2.2, 2.8, and 3.4 km/hr are presented in Fig. 7. In comparing the knee and ankle angles to the prototypical data [1], one can observe that the powered prosthesis and controller provide behavior quite similar to normal gait. The knee and ankle torque trajectories with references

are presented in Fig. 8, demonstrating capabilities of the electric prosthesis to produce the requisite forces.

B. Power Consumption

One of the primary constraints of the electrical powered knee and ankle prosthesis design is the power source. As such, the power consumption of the prosthesis was assessed to characterize the feasibility of such as device in terms of mass and range. The power delivered at each joint during a slow walking trial is shown in Fig. 9. Fig. 10 shows the average mechanical power of the prosthesis for the various modes, along with the average electrical input power. As can be observed from the figure, the maximum average power consumption in the normal walking mode is about 65 Watts. It is assumed that high frequency switching servo-amplifiers with 90% efficiency will be used to drive the motors. Moreover, an additional 5 Watts is added for on-board signal conditioning and computation. Thus, it is estimated that normal walking will require approximately 77 Watts of average power. Finally, it is assumed that the prosthesis can accommodate approximately 600-700 grams of battery. Given these assumptions, Table III shows several lithium polymer battery options for the prosthesis, and the estimated walking distance in normal walking mode for the healthy experimental subject with the able-bodied adapter. As seen in the table, such an arrangement provides approximately 4.4-5.0 km of walking range. Given current projections of battery technology, this range should roughly double (to approximately 10 km) within the next decade.

V. Conclusion and Further Works

The paper describes the design of an electrically powered knee and ankle prosthesis capable of producing human-scale power. Utilizing finite-state impedance based control and an able-bodied testing adapter, experimental results were obtained to demonstrate the merit of the device and measure its power consumption. With current battery technology the prosthesis range is 5.0 km at a normal walking speed, which is projected to double within the next decade. Future work includes testing the powered prosthesis on amputee subjects and implementing a self-contained version with on-board servo-amplifiers, batteries and computational capabilities.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

Acknowledgments

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Fig. 1.
The power tethered prototype.

Fig. 2. The reduction of linear force output required by the ankle motor unit by the addition of a spring in parallel for fast walking, taken from averaged normal biomechanical data [1].

Fig. 3.
Sagittal moment load cell, top and bottom views.

Fig. 4.
Sensorized prosthetic foot.

Fig. 5.
The finite state switching model for impedance control. Blocks represent a state and arrows represent the corresponding transitions.

Fig. 6.
Able-bodied testing adapter used in the development of the prosthesis and controllers prior to transfemoral amputee participation.

Fig. 7.
Measured joint angles of the powered prosthesis for ten consecutive gait cycles of treadmill walking at three different speeds - 2.2, 2.8, 3.4 km/hr.

Fig. 8. References and actual knee and ankle joint torques of the powered prosthesis for one stride at normal speed.

Fig. 9. Measured mechanical power of the prosthesis for ten consecutive gait cycles of treadmill walking at slow speed.

Fig. 10.

Average electrical power consumption and mechanical power generation at the ankle and knee of the powered prosthesis for standing and different walking speeds.

TABLE I
Impedance parameters for walking from experimental tuning. Highlighted parameters vary with walking speed.

Mode	Knee Impedance					Ankle Impedance			
	Speed km h ⁻¹	<i>k</i> Nm deg ⁻¹	<i>b</i> N s m ⁻¹	<i>θ_k</i> deg	<i>k</i> Nm deg ⁻¹	<i>b</i> N s m ⁻¹	<i>θ_a</i> deg	<i>k</i> Nm deg ⁻¹	<i>θ_a</i> deg
1	2.2	1.5	0	10	3.5	0	-4.4 [*]		
	2.8	1.5	0	13	3.5	0	-4.4 [*]		
	3.4	2.0	0	13	4.0	0	-4.4 [*]		
2	2.2	1.5	0.05	16	3.5	0	-18		
	2.8	1.5	0.05	16	4.0	0	-19		
	3.4	2.0	0.05	16	4.5	0	-20		
3	2.2	0	0.01	50	0.3	0.02	0		
	2.8	0	0.01	50	0.3	0.02	0		
	3.4	0	0.01	50	0.3	0.02	0		
4	2.2	0	0.045	45	0.3	0	0		
	2.8	0	0.035	45	0.3	0	0		
	3.4	0	0.04	45	0.3	0	0		

^{*} Equilibrium angle is a nonlinear function of knee and ankle angles. The listed value is the initial equilibrium set point.

TABLE II

Impedance parameters for standing from experimental tuning.

Mode	Knee Impedance				Ankle Impedance		
	k Nm deg ⁻¹	b N s m ⁻¹	θ_k deg	θ_k deg	k Nm deg ⁻¹	b N s m ⁻¹	θ_k deg
Ground Contact	2.0	0.05	5	5	5.0	0	-4
No Ground Contact	0	0.02	5	5	2.0	0	-4

TABLE III

Selected commercially available lithium polymer battery options and the walking distance at normal walking speed.

Option	Voltage (V)	Capacity (mA·h)	Energy (W·h)	Weight (gr)	Walking Distance (km)
1	29.6	4000	118.4	622	4.4
2	29.6	2000	118.4	640	4.4
3	33.3	2000	133.2	720	5.0