Cable-Driven Two Degrees-of-Freedom Ankle–Foot Prosthesis[1]

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1 Background

The ankle is of fundamental importance during locomotion as it is the first major joint to transfer the ground reaction torques to the rest of the body. Similar to the role of muscles and tendons in the musculoskeletal system, a prosthesis may benefit from cable-driven systems, specifically Bowden cables, to actuate joints. For an ankle–foot prosthesis, Bowden cables allow the placement of the motors and gearboxes away from the distal parts of the limb and near the center of gravity of the user, reducing the metabolic cost. A mass at the feet increases the user metabolic cost by 8–9%/kg during walk, while a mass carried at the waist increases the metabolic cost by only 1–2%/kg [1]. Bowden cables also allow for flexibility on the customization of the prosthesis, especially when long residual limb would limit the amount of space available for the active components.

Currently available ankle–foot prostheses focus on improving gait in the sagittal plane. However, turning steps may represent up to 50% of all the steps depending on the activity [2]. Turning steps require the ankle functions in inversion–eversion (IE) and in dorsiflexion–plantarflexion (DP). In IE, ankle torques are required for walking in a straight line, and larger torques are required for turning [3,4]; this requirement can be addressed by designing two degrees-of-freedom (DOF) ankle–foot prostheses. Bellman et al. designed an ankle–foot robot with 2DOF [5]; however, a prototype had not been developed.

Impedance controllers have been used in the control of prostheses due to their ability to adjust the stiffness of the prosthetic joints [6,7]. The impedance of a system is defined as the evoked force due to an input motion perturbation. The human ankle shows a time-varying impedance through the stance phase of gait [8,9]. This motivated the use of impedance controllers for the ankle–foot prosthesis controllable in the frontal and sagittal planes presented here. In this paper, the prosthesis design is presented, and the controllers for the prosthesis are explained.

2 Methods

The ankle–foot prosthesis with 2DOF in DP and IE is shown in Figs. 1 and 2. The components were chosen based on the power and energy requirements of an 80 kg person. Such individual would require 36 J (250 W peak power) at each step and torque as high as 140 N · m [10]. Amputees wearing transtibial prosthesis require up to 35% more power than nonamputees during self-selected free-walk speed [11]. The prosthesis, Bowden cables, and gearboxes are expected to have losses up to 40%. These result in an anticipated power requirement of 470 W and maximum torque of 264 N · m.

The prosthesis is composed of two brushless motors and gearboxes (A), which transfer the torque to four Bowden cables (B) using two cable drums (C). The cables are connected to a carbon fiber plate spring (D) at two points in the rear side of the plate and pass through a pulley (E) in front of the plate. The carbon fiber plate is securely attached to the foot (I) (Össur Flex-Foot) to assure the cables are always under tension. A universal joint (F) constrains the foot to the pylon (G) and allows the foot rotations in DP and IE. The mechanism works because three points are sufficient to define a plane in the space. By constraining the plane in all the translations and in one rotation (using the universal joint), these three points can be used to control the remaining 2DOF, i.e., rotations in DP and IE. Rotational encoders are mounted at the motors and are used to calculate the angular rotation of the foot in both DOF. Strain gauges (H) are mounted to the foot and are used to measure the foot deflection due to external ground reaction torques during walking. The passive components of the prosthesis, including the pylon and foot, weigh 1.13 kg. The active

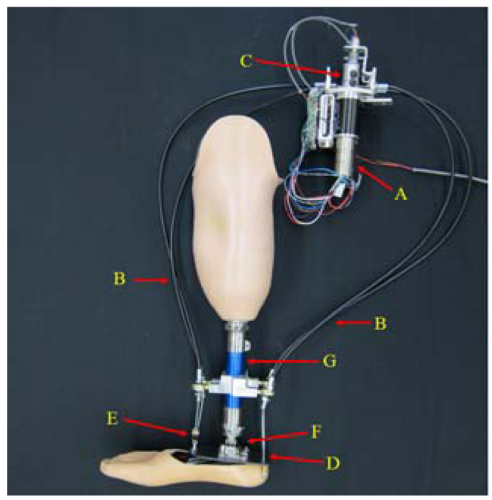

Fig. 1 A 2DOF ankle–foot prosthesis

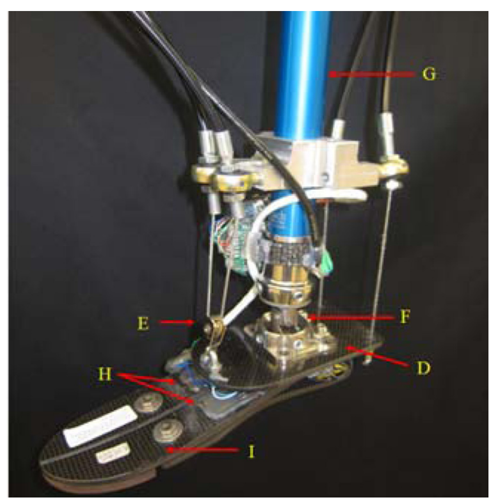

Fig. 2 Detail of the 2DOF ankle–foot prosthesis

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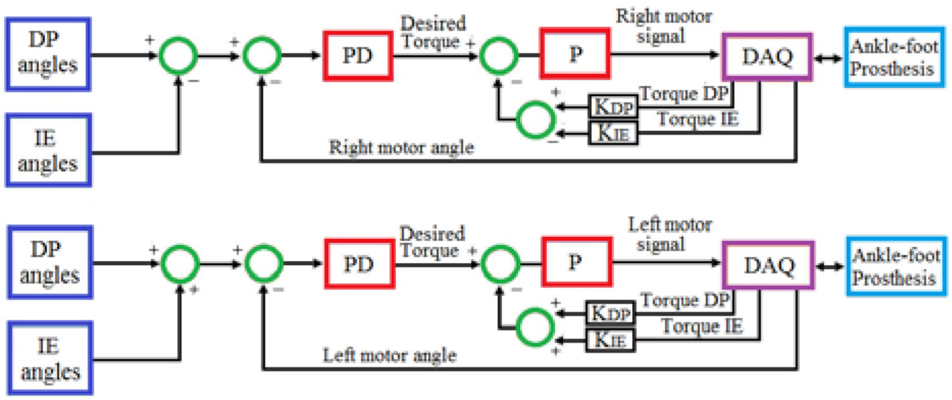

Fig. 3 Impedance controllers for the ankle–foot prosthesis

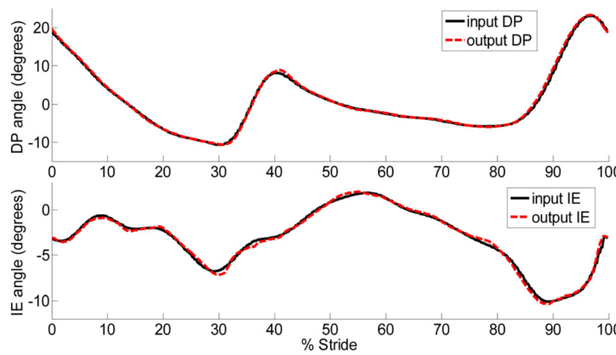

Fig. 4 Trajectories of the ankle–foot prosthesis (output) following the recorded human ankle kinematics (input) with a time shift of 40 ms for easier visualization

components of the prosthesis, including the motors and gearboxes (excluding battery), weigh 2.2 kg.

To control the prosthesis, independent impedance controllers were developed (Fig. 3) for the left and right motors. The difference between DP and IE angles is used in the right motor controller, and the sum of DP and IE angles is used in the left motor controller. This difference in the controllers for the right and left motors is required as the sum of the two motors' rotations generated DP and the difference in motors' rotations generated IE. The DP and IE angles were recorded from a human gait experiment using a motion capture camera system, while the person was performing a turning step. The reference angle is compared to the angle feedback from the motor encoders, and the error is used in a proportional plus derivative (PD) controller to estimate the appropriate prosthesis torque. The estimated torque is compared to the feedback torque, and the error is used in a proportional (P) controller to estimate the motor input voltage. Similar to the angles, the controller used the difference between DP and IE torque feedbacks in the right motor controller, and the sum of DP and IE torque feedbacks in the left motor controller. The torque feedback gains KDP and KIE can be used to change the impedance of the prosthesis.

3 Results

The ankle–foot prosthesis was capable of mimicking the recorded human ankle motion during walking as shown in Fig. 4. In this experiment, there was no load on the ankle, thus the torque feedbacks were zero, and the impedance controllers behaved as position controllers. A second experiment consisted of loading the ankle with $23\,\mathrm{N\cdot m}$ of torque in DP and $15\,\mathrm{N\cdot m}$ of torque in IE and measuring the ankle deflections. This experiment was performed with different torque feedback gains KDP and KIE. This allowed for the estimation of the ankle quasi-static impedance (stiffness) at different torque feedbacks. In DP, KDP ranged from -0.05 to 0.4 and the stiffness ranged from $282\,\mathrm{N\cdot m/rad}$ to $23\,\mathrm{N\cdot m/rad}$, respectively. In IE, KIE ranged from -0.1 to 0.4 and the stiffness ranged from $36\,\mathrm{N\cdot m/rad}$ to $6\,\mathrm{N\cdot m/rad}$, respectively.

4 Interpretation

The proposed ankle–foot mechanism was designed to mimic the human ankle kinetics and kinematics. This includes weight, power, speed, and range of motion. Preliminary evaluation of the prosthesis showed that it was capable of mimicking the human kinematics during a turning step, indicating a plausible kinematics design. Loading the ankle with known amounts of torque and measuring the deflection with different torque feedbacks showed that the controllers have desirable capability of impedance modulation. Moreover, it showed that the prosthesis could potentially mimic the kinetics and kinematics of the human ankle during walk, including the time-varying impedance of the human ankle.

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