Variable Series Elasticity Control of a Pneumatically Actuated Transtibial Prosthesis[1]

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1 Background

The ankle plays an important role in human walking. Biomechanical studies show that the ankle generates more power than hip and knee in walking [1]. Without the power generated by the ankle, transtibial (TT, also known as below knee) amputees fitted with the traditional passive prostheses suffer from multiple problems, e.g., asymmetric gait and greater metabolic energy expenditure.

Motivated by this reality, researchers have expended a significant amount of efforts in creating energetically active (i.e., powered) TT prostheses. Typical examples include the multiple prototypes developed by the MIT Biomechatronics Group (e.g., Ref. [2]), the SPARKy prosthesis, and the Vanderbilt transtibial prosthesis. These prostheses are all powered with electric motors, augmented with mechanical springs to simulate the elastic characteristics of the biological ankle. For example, the prosthesis design by Au et al. uses springs in series with the electric motor to form a series elastic actuator, and an additional spring is used to simulate the energy storage-and-return feature in the ankle loading and push-off [2].

Compared with the actuation approaches in the aforementioned works, pneumatic actuation is a strong competitor with its multiple advantages. Pneumatic actuators have higher power density than electric motors in general. More importantly, a pneumatic actuator features a physically existing elasticity that can be adjusted via pressure regulation. As such, a pneumatically actuated prosthesis has a potential of simulating the elasticity of the biological ankle without using mechanical springs, significantly simplifying the system design. However, such potential has not been utilized in existing pneumatic TT prostheses (e.g., Ref. [3]). In this paper, a new control approach, namely, *variable series elasticity control* (VSEC), is presented, which provides effective control of a pneumatically actuated TT prosthesis, leveraging the aforementioned elasticity in pneumatic actuators.

2 Methods

The powered TT prosthesis that serves as the basis of this work is the Alabama Powered Prosthetic Limb–Ankle (APPL-A) [4],

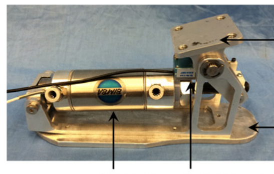

Fig. 1 The pneumatically actuated TT prosthesis APPL-A

which is powered with a double-acting pneumatic cylinder placing in the horizontal direction (Fig. 1). The actuator drives the prosthetic ankle through a crank–sliding mechanism, converting the linear motion of the cylinder piston into the rotation of the prosthetic joint.

To model the joint elasticity, the linear stiffness of the pneumatic cylinder can be expressed as [5]

$$K = \frac{m_a RT}{\left(\frac{L}{2} + L_{da} + x\right)^2} + \frac{m_b RT}{\left(\frac{L}{2} + L_{db} - x\right)^2} \tag{1}$$

where K is the stiffness, R is the universal gas constant, T is the air temperature, L is the stroke, x is the displacement with respect to the middle point of the stroke, L_{da} and L_{db} are the dead lengths of chambers a (rodless chamber) and b (chamber with rod), respectively, and m_a and m_b are the air masses in chambers a and b, respectively. It can be clearly observed from this equation that the actuator stiffness is a function of chamber air masses m_a and m_b. Similar to the stiffness, the equilibrium point x_e can also be expressed as a function of the chamber air masses

$$\frac{m_a RT}{\frac{L}{2} + L_{da} + x_e} - \frac{m_b RT}{\frac{L}{2} + L_{db} + x_e} - P_{\text{atm}} A_r = 0 \tag{2}$$

where A_r is the rod cross-sectional area. Combining Eqs. (1) and (2), the air masses can be calculated based on a set of desired stiffness-equilibrium point [5]. Such method can be combined with the biologically inspired finite-state impedance control [4] to obtain effective prosthesis control with simple chamber pressure control using inexpensive on–off valve. Specifically, the desired air masses, as calculated by the desired stiffness-equilibrium point, can be converted to the desired pressure commands through the ideal gas equation

$$P_a = \frac{m_a RT}{V_a} = \frac{m_a RT}{A_a\left(\frac{L}{2} + L_{da} + x\right)} \tag{3}$$

$$P_b = \frac{m_b RT}{V_b} = \frac{m_b RT}{A_b\left(\frac{L}{2} + L_{db} + x\right)} \tag{4}$$

Subsequently, the desired pressure is compared with the measured chamber pressure to determine the initial valve commands: open the chamber to pneumatic supply if the measured pressure is lower than desired, otherwise open the chamber to exhaust. When the desired pressure is reached, close the chamber until the state machine transitions to the next phase. Note that the attainable chamber pressure is limited to the range as defined by the supply pressure (150 psi, or 1034 kPa, as dictated by the maximum pressure rating of the control valve, VQ1300K-5B1, SMC Corporation, Tokyo, Japan) and atmosphere pressure. As such, special

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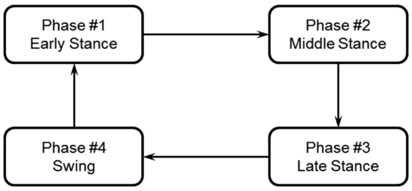

Fig. 2 Finite-state machine in the prosthesis controller

treatment has to be taken for certain phases and phase transitions, as detailed in the subsequent section.

3 Results

To implement the VSEC, a gait cycle is divided into four distinct states, as shown in Fig. 2.

(1) Early stance (ES): The ankle functions like a spring with moderate stiffness to provide shock absorption and appropriate plantarflexion resistance before foot flat. The initial stiffness is set at 0.8 N·m/deg with the equilibrium point of 1.72 deg, translating into the initial (absolute) chamber pressures at 214 kPa and 185 kPa for chambers a and *b*, respectively.

(2) Middle stance (MS): The ankle functions like a very stiff spring to absorb energy to get ready for the push-off. The initial stiffness is set at 4.3 N·m/deg with the equilibrium point of −5 deg, translating into initial chamber pressures at 1135 kPa and 1096 kPa. Note that the initial stiffness is limited by the valve pressure rating. However, with the considerable displacement in this phase, the stiffness increases significantly when the piston approaches the ends of the cylinder, generating a closer match to the biomechanical data.

(3) Late stance (LS). The ankle functions like a stiff spring, with the stiffness lower than that in MS. However, with the equilibrium point shifting, the ankle in this state generates higher torque output than in the MS, constituting the powered push-off. Specifically for the chamber pressures, due to the compression in the MS, chamber b pressure is expected to rise to 2657 kPa. As such, the valve for this chamber should stay closed to maintain the pressure. Chamber a pressure, expected to be 731 kPa at the end of MS, needs to be lowered to ∼264 kPa to increase the actuation torque for a powered push-off.

(4) Swing (SW): The ankle functions like a spring with very low stiffness, returning the ankle to a slightly dorsiflexed position to get ready for the next heel strike. In this phase, chamber b should be exhausted over the entire period to avoid unintended pressure increase, while chamber a should be pressurized to 197 kPa at the beginning to generate a smooth SW motion.

The values discussed above are obtained through a dynamic simulation using the biomechanical motion data by Winter [1]. Figure 3 shows the joint torque by the control approach above in comparison with the biomechanical torque trajectory in Ref. [1]. In this figure, points A–D mark the transitions between phases: A → B is ES, B → C is MS, C → D is LS, and D → A is SW. It can be clearly observed that the control approach in this paper is able to provide a torque trajectory similar to the biomechanical trajectory, and replicate the key features in ankle biomechanics, including ankle loading in MS and powered push-off in LS.

4 Interpretation

Presented in this paper is a new TT prosthesis controller, namely, VSEC, which simulates the elastic characteristics of the ankle in human walking by modulating the series elasticity in the pneumatic actuator. A two-input-two-output dynamic model of the pneumatic actuator was presented, which converts the desired stiffness-equilibrium point into the desired chamber masses. As such, the finite-state impedance control of TT prostheses can be implemented through simple pressure control of each chamber in the actuator, significantly simplifying the implementation of the prosthesis controller. Simulation has been conducted, with results indicating that such control approach is able to replicate the desired dynamics of human ankle in walking. Currently, human testing is being conducted to test the performance of this controller (Fig. 4), which will be compared with that of the more traditional force-controlled implementation of prosthesis impedance controller in the authors' prior work [4].

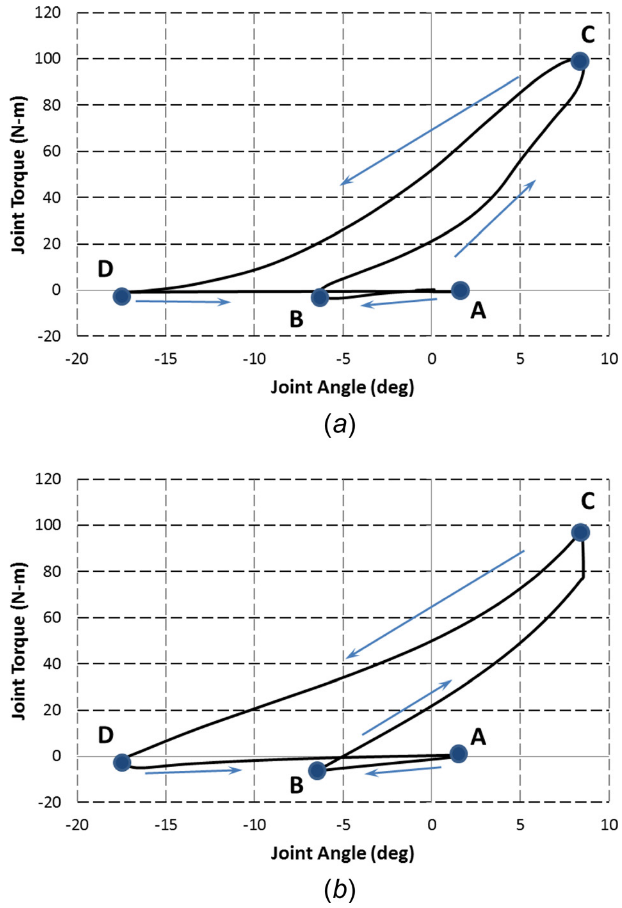

Fig. 3 Comparison of the typical torque trajectory for a 65 kg person in walking (*a*) versus the trajectory generated by the VSEC controller (*b*)

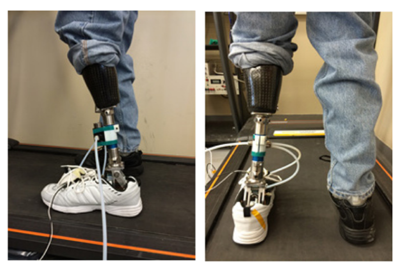

Fig. 4 A test subject fitted with the APPL-A prosthesis

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