Abstract: Current finite-state control strategies for powered below-knee prosthesis, though effective to the normal gait, can not eliminate the disturbance of abnormal gaits such as slip and stamp. In addition, toe joint is not taken into consideration. This paper presents a finite-state control strategy for a powered below-knee prosthesis with ankle and toe. We first introduce the concerned prosthesis prototype in detail. By dividing the walking gait with toe and joint into more states and setting stricter transition conditions, the gait identification becomes more accurate and gait disturbance such as slip and stamp can be eliminated. Experimental results show that the proposed method improves the accuracy of gait identification. It makes the motion of the prosthesis more natural and provides better bases for control.

Keywords: Finite-state control, gait identification, prosthesis, ankle, toe.

1. INTRODUCTION

There are many kinds of commercial below-knee prostheses, few of them, however, can provide net positive work, which causes the transtibial amputees spend 20%–30% more energy when walking at the same speed as able-bodied individuals (Molen. (1973); Colborne et al. (1992)). Those existing prostheses are called passive prostheses as there is no power source to provide positive work. In 1998, Klute et al. (1998) first built a pneumatically powered ankle-foot prosthesis capable of providing net positive work and improving the energy efficiency of the amputee. Afterwards, Sup et al. (2007) and Au et al. (2009) developed hydraulically and electrically powered ankle-foot prosthesis, respectively. Different from the passive prostheses, the powered ones can both act as well as react, to actively mimic the dynamical and kinematic characteristics of the normal limb. Hence, it is necessary to develop an effective and reliable control strategy for the prostheses to ensure stable interaction with the amputee and the environment.

Flowers (1973) developed an echo control scheme for gait control based on an electrohydraulic transfemoral prosthesis. This strategy, however, requires installing sensors on the sound-side leg, brings inconvenience to the amputee and presents a problem for odd numbers of steps. EMG signal-based control strategy was also used to control the prosthesis, but its accuracy and instantaneity is far from satisfactory. Varol (2007) developed the decomposition based control strategy, which divided the normal gait into four distinct phases and could generate real-time trajectories of the knee joint angle and interaction forces. Zlatnik et al. (2002) and Au et al. (2009) developed finite-state control strategy to control knee and ankle’s movement, respectively. Though effective to the normal gait, their control strategy becomes inappropriate for abnormal gaits such as slip and stamp. What’s more, their mechanical design as well as control strategy did not take the toe into consideration, which is believed to play an important role in stable foot lifting, less energy consumption walking and stepping upstairs (Nishiwaki et al. (2002)). The control strategy presented in this paper is based on a powered below-knee prosthesis with toe as well as ankle (Zhu et al. (2010)) and the finite states become more detailed. By collecting more motion information and setting stricter state transition conditions, the control strategy can not only identify the states accurately, but also eliminate the disturbance of some abnormal gaits.

The rest of this paper is organized as follows. Section 2 introduces the concerned prosthesis with ankle and toe in detail. Section 3 presents the finite-state control method for the proposed prosthesis. Section 4 shows the realization of the control method in sensors and circuits. Experimental results are shown in Section 5. We conclude in Section 6.

2. POWERED BELOW-KNEE PROSTHESIS WITH ANKLE AND TOE

2.1 Biomechanical Behaviors of Human Walking

Human walking is a cyclic pattern of bodily movements that is repeated over and over, step after step. Every gait cycle starts with heel strike (HS) when the heel initially touched the ground and ends with the next HS of the same leg. Each cycle can be divided into two main phases: stance phase and swing phase (Inman et al. (1981)). The stance phase begins at the moment of HS and ends at the moment of toe off (TO) when the forefoot pushes off the ground. The swing phase begins at the moment TO and ends at
the next HS. The stance phase takes up 60% of the gait cycle and includes four subphases, which are:
(1) HS to Foot Flat (FF);
(2) FF to Midstance (MS);
(3) MS to Heel Off (HO);
(4) HO to TO.

At the moment of HS, the ankle has to endure the impact force when the heel initially contacts the ground. From HS to FF, the ankle stores elastic energy to plantarflexor muscles. During FF to MS, the energy stored during the last period is released to help the body change the center of gravity from the support leg to the foreleg. At MS, the ankle begins to store elastic energy to dorsiflexor muscles and at HO the ankle reaches a state of maximum dorsiflexion. Then in the period HO to TO, ankle releases the energy stored in the last period to propel the body. However, the stored energy is much less than the energy needed. Then the ankle has to output much more net positive work. Meanwhile, the forefoot is bent to store energy. At the last of the period HO to TO, the toe joint releases the stored energy and supplements some net work to propel the body. During swing phase, the main function of the ankle and toe joints is to adjust the position of the foot to prepare for the next stance phase.

2.2 Powered Below-Knee Prosthesis with Ankle and Toe

The powered prosthesis with powered compliant joints and segmented foot (PANTOE 1) is designed for below-knee amputees for level-ground walking (Zhu et al. (2010)). Fig. 1 and Fig. 2 show the CAD model and the prototype of PANTOE 1 respectively. As far as we know, this is a first attempt to add segmented foot with toe joint to the prosthesis. The segmented foot decreases the torque of the ankle and makes the amputees effort-saving. Both the ankle and toe joints are driven by two series-elastic actuators (SEA), which not only provide enough torque, but also tolerance shocks. The detailed design of PANTOE 1 is described in Zhu et al. (2010).

3. FINITE-STATE CONTROL

As mentioned above, we divide one walking gait cycle into seven phases. Each phase represents a unique state where the prosthesis performs specific behavior and will be controlled by specific control strategy. Besides, one state can transit to another if the triggering transition requirements are meet. This method is called finite-state control and can be described as follows (Zlatnik et al. (2002)):

\[ A_i = f_a(S_i) \]  \hspace{1cm} (1)

Where \( f_a \) is the action function indicating the specific output behavior \( A_i \) of the specific state \( S_i \).

\[ S_{i+1} = f_s(S_i, I_i) \]  \hspace{1cm} (2)

Where \( f_s \) is the transition function indicating the transition between two adjacent states and \( I_i \) represents the input triggering information.

As defined, a level ground walking gait cycle begins with the heel strike of one foot and ends with the next heel strike of the same foot (Au et al. (2009)). It is generally divided into the stance phase when the foot is on the ground and the swing phase when the foot is off the ground. To get a more accurate description of the prosthesis state, we divide each phase into different sub-phases (see Fig. 4):

**Stance Phase:**

1) **Controlled plantar flexion of ankle:** CPA begins at heel-strike when the ankle joint begins to plantarly flex and ends at foot-flat when toe-strike occurs. The behavior of the ankle joint at this state is consistent with a linear spring response that the output joint torque is...
proportional to the joint angle (Au et al. (2009); Palmer (2002); Hansen et al. (2004); Gates (2004)).

2) Controlled dorsiflexion of ankle: CDA begins at foot-flat when the ankle joint begins to dorsiflex and ends when the ankle reaches the maximum dorsiflexion angle. The behavior of the ankle joint at this state can be described as a nonlinear spring (Au et al. (2009); Palmer (2002); Hansen et al. (2004); Gates (2004)).

3) Powered plantar flexion of ankle: PPA begins after CDA when the ankle joint begins to plantarly flex again and ends at the instant of toe-off. The function of the ankle joint at this state is the superposition of a nonlinear spring and a torque source (Au et al. (2009); Palmer (2002); Hansen et al. (2004); Gates (2004)). This phase is the main phase that can reflect the dynamical characteristics of the ankle joint.

4) Controlled dorsiflexion of toe: CDT begins after heel-off when the toe joint is compressed and begins to plantar flex. It ends when the toe joint is compressed to specific extent. Similarly, the behavior of the toe joint at this state can be modeled as a linear spring.

5) Powered plantar flexion of toe: PPT begins when the toe joint angle reaches a specific value and the toe joint begins to dorsiflex to push the body forward and
upward together with the ankle joint. It ends at toe-off. The toe joint’s behavior at this state can be modeled as the superposition of a linear spring and a torque source.

Swing Phase:

6) Early swing: ESW begins at toe-off when the ankle joint and the toe joint begin to restore to the equilibrium position and ends after a predefined time period when the two joints move back to the equilibrium position. Both of the two joints play the role of position source to reset the prosthesis to equilibrium position.

7) Late swing: LSW begins just after the ESW and ends at the next heel-strike. The toe joint and the ankle joint just keep the balanced state and get ready for the beginning of the next gait cycle. The function of the two joints can also be modeled as the position source.

We define the ankle joint angle to be zero when the shank is perpendicular to the foot. From the zero position, the angle will be negative if the ankle plantarly flex and positive if the ankle dorsiflex. The toe joint angle is defined to be zero when the joint is not compressed and to be positive we compressed to dorsiflex, as shown in Fig. 5. To accurately identify each gait state and decide the transition between different states, we collect the information below:

a) Heel contact \( H=0 \) indicates that the heel is off the ground and vice versa.

b) Heel pressure \( F_H \). \( F_H \) indicates the pressure that the ground exerts to the heel and \( F_{PH} \) indicates the predefined heel pressure which is obtained experimentally.

c) Toe contact \( T=0 \) indicates that the toe is off the ground and vice versa.

d) Toe pressure \( F_T \). \( F_T \) indicates the pressure that the ground exerts to the toe and \( F_{PT} \) indicates the predefined heel pressure which is also obtained experimentally.

e) Ankle joint angle \( \theta_a \).

f) Ankle joint’s angular velocity \( \dot{\theta}_a \). \( \dot{\theta}_a \) indicates the rotatory direction of the ankle joint.

g) Toe joint angle \( \theta_t \).

h) Ankle torque \( T_a \).

i) Stance phase time period \( \Delta t_{\text{stance}} \). \( \Delta t_{\text{stance}} \) can be used as an indicator of the walking speed.

j) Swing phase time period \( \Delta t_{\text{swing}} \). \( \Delta t_{\text{swing}} \) can be used as an indicator of the walking speed together with \( \Delta t_{\text{stance}} \).

We hope that the finite-state control strategy can not only identify the specific gait state, but also eliminate the disturbance of some abnormal gait such as slip, stamp, standing on tiptoe and standing on heel.

1) If the wearer just stands on heel and do not intend to walk forward, the transition condition \( H=1 \) and \( \theta_a<0 \) before between LSW and CPA will not be sufficient enough. By adding the condition \( F_H>F_{PPH} \) and \( \theta_a<0 \) to constrain the minimum heel pressure and rotatory direction respectively, the disturbance can be eliminated. So is the case with standing on tiptoe. If we only constrain the toe angle to be larger than a predefined value as the transition condition between the toe joint’s CDA state and PPA state, standing on tiptoe may trigger the transition and prevent the toe joint from rotating. This disturbance can be eliminated by adding the condition \( F_T>F_{PT} \) to constrain the minimum toe pressure.

2) If the wearer just stamps on the ground and does not intend to walk forward, the transition between ankle’s CPA and CDA state is not expected. We can eliminate the disturbance by constraining the time interval between heel-strike and toe-strike \( \Delta t_{pp} \). For the heel-strike and toe-strike of the stamping gait happen almost at the same time and the time interval is quite small, while the interval of the walking gait is relatively large. The predefined parameters such as \( F_{PH}, F_{PT}, \tau_{pp}, \tau_p \) and \( \theta_p \) are tuned experimentally.

3) If the wearer slips during walking, it is expected that the prosthesis keep actionless until the wearer stands up again. For example, if the wearer slips backward at ankle’s CPA state, it will be difficult for the wearer to stand up again if the ankle dorsiflex to CDA state so we hope the prosthesis just stop to keep the CPA state. Similarly, if the wearer slips forward at ankle’s CDA state, the prosthesis is hoped to keep in CDA state and do not plantar flex to PPA state. Hence we set a stop state and any abnormal gait like slip which leads to \( H=0 \) and \( T=0 \) during stance can trigger the transition to the stop mode. The wearer can also choose to transit to the stop state by lifting the foot \( H=0 \) and \( T=0 \) intentionally. If the wearer stands up again \( H=1 \) and \( T=1 \), the prosthesis will reset and get ready for the next gait cycle.

Besides being able to eliminate abnormal gait disturbance, the control strategy can also adapt to speed variation by measuring the stance and swing phase time period. For the time percentage of every phase is relatively changeless. Knowing the stance time or the swing time, we can estimate the time period of the gait cycle and adjust the prosthesis’ movement velocity of each phase.

4. SENSORS AND CONTROL CIRCUITS

As shown in Fig. 1, we install different kinds of sensors on the prosthesis to collect the triggering information and decide the prosthesis’ states:
Fig. 6. The control platform.

a) Two contact switches, one beneath the heel and the other beneath the toe, to detect the heel-strike $H=1$ and toe-strike $T=1$, respectively.
b) Two force sensors with the 1000N range, one beneath the heel and the other beneath the toe, to measure the pressure $F_H$ and $F_T$ exerted by the ground.
c) Two 4.7kΩ circular-shaped potentiometers, one placed at the ankle joint and the other at the toe joint to measure the ankle angle $\theta_a$ and toe angle $\theta_t$, respectively.
d) A 5kΩ linear potentiometer placed parallel with the ankle spring to measure the displacement of the spring and estimate the ankle torque $T_a$.

The ankle angle velocity is the differential of the angle and can be easily calculated. The stance time period, swing time period and the other time intervals can be measured by the timer in the controller chip.

To provide power, we install a 83W Faulhaber DC motor at the ankle joint and a 45W Faulhaber dc motor at the toe joint. Each motor is actuated by a specific actuator.

We also design a PCB with 5 AD ports and 7 IO ports, and choose chip STM32F103C8 as the controller, as is shown in Fig.6

5. EXPERIMENTAL RESULTS

Based on the prosthesis and the control platform, we conduct preliminary experiments to verify the effectiveness of the finite-state control strategy. It is expected that the control strategy can accurately identify all the gait states, eliminate the disturbance of the abnormal gait and adapt to speed variation. As there is no amputee to wear the prosthesis at present, we simulate the wearer by pressing the prosthesis with hand. Because the pressure generated by hand press is smaller than that generated by the amputee, the predefined parameters such as $F_{PH}$, $F_{PT}$ and $\tau_{pp}$ should also be set smaller.

The control flow chart is shown in Fig. 7. When the prosthesis is energized, it will reset to the equilibrium state, namely set the ankle angle and the toe angle to be zero. Then the prosthesis speed is initialized, whose value is first predefined and will be adjusted according to the wearer’s walking speed estimated from the measured stance or swing time. The system then scans the sensors to check if the transition conditions are met. If so, the prosthesis will transmit to a new state and generate specific action, or it will keep in the state before. Time intervals between every two states are measured to calculate the stance and swing time.

Gait identification results at different walking speed with the finite-state control strategy is shown in Fig. 8. Seven prosthesis states can be identified accurately. The angle variation of ankle and toe joint is shown in Fig. 9, which is a satisfactory mimic of the normal limb.

6. CONCLUSION AND FUTURE WORK

This paper has demonstrated the design, implementation and preliminary experiments of the finite-state control strategy based on a powered prosthesis with ankle and toe.
Taking the toe joint into consideration, the walking gait cycle is divided into more phases to get a better description of the gait. Different kinds of sensors are installed on the prosthesis to collect the state information and decide the prosthesis’ movement state. Besides, by adding more limiting conditions to the state transition, the control strategy is more robust than that of Zlatnik et al. (2002) and Varol (2007) and can eliminate the disturbance of some abnormal gaits. Lastly, the control strategy can adjust the prosthesis’ speed based on the estimation of the wearer’s walking speed, and can adapt to speed variation. Preliminary experimental results can verify the control strategy’s effectiveness in gait identification and speed adaptation, but the experiment is not conducted with the amputee. Besides, the control strategy is not universal enough as some parameters of it must be tuned experimentally according to the wearer. In the future, more experiments need to be conducted with the amputee to test the effectiveness of the finite-state control strategy. We should also go one step further to extend the strategy to more applications such as walking upstairs and downstairs.

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