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Design and Control of a Pneumatically Actuated Transtibial Prosthesis

Hao Zheng and Xiangrong Shen

Department of Mechanical Engineering The University of Alabama 359 H.M. Comer Hall, Box 870276 Tuscaloosa, AL 35487-0276

Abstract

This paper presents the design and control of a pneumatically actuated transtibial prosthesis, which utilizes a pneumatic cylinder-type actuator to power the prosthetic ankle joint to support the user's locomotion. The pneumatic actuator has multiple advantages over the traditional electric motor, such as light weight, low cost, and high power-to-weight ratio. The objective of this work is to develop a compact and lightweight transtibial prosthesis, leveraging the multiple advantages provided by this highly competitive actuator. In this paper, the design details of the prosthesis are described, including the determination of performance specifications, the layout of the actuation mechanism, and the calculation of the torque capacity. Through the authors' design calculation, the prosthesis is able to provide sufficient range of motion and torque capacity to support the locomotion of a 75 kg individual. The controller design is also described, including the underlying biomechanical analysis and the formulation of the finite-state impedance controller. Finally, the human subject testing results are presented, with the data indicating that the prosthesis is able to generate a natural walking gait and sufficient power output for its amputee user.

1. INTRODUCTION

For a lower-extremity prosthesis, the primary purpose is to restore the locomotive functions of lost limb sections and joints. Traditionally, such functions have been restored by energetically passive devices, i.e., devices that only dissipate energy, or store and reuse energy within a gait cycle. The passive nature of such devices is fundamentally different from the energetic role of the corresponding biological joints, and thus poses a significant limitation to their functionality and rehabilitation effects. For example, biomechanical studies on human locomotion highlight the important energetic role of the ankle joint. In level walking, the ankle produces substantially more work than the knee and hip [1]. Unlike the knee, the ankle's energetic behavior in level walking is clearly and significantly positive (i.e., integration over a cycle of power data is clearly and significantly positive) [2]. As such, for an amputee fitted with passive transtibial prosthesis, he or she has to expend more power on the unaffected biological joints to compensate for the lack of power generation in the prosthetic ankle, resulting in an asymmetric gait and greater energy consumption [3,4].

To address this important issue, a considerable amount of research has been conducted on the development of energetically active transtibial prostheses with powered ankle joints. In such efforts, the primary challenge is to generate sufficient power and torque output within a compact form factor. In the existing works, the major technical approach is electric

actuation, combining electromagnetic actuator (i.e., DC motor) with electrochemical batteries. Typical works adopting this approach include the powered ankle-foot prostheses developed by the Biomechatronics group at MIT [5-7], the two-degree-of-freedom SPARKy ankle prosthesis [8,9], and the powered transfemoral prostheses developed by the Center for Intelligent Mechatronics at Vanderbilt University (which include powered ankle joints) [10-12]. In spite of the improved gait quality provided by these active devices, they tend to suffer from multiple inherent weaknesses of an electric actuation system, primarily the heavy weight of the actuator and the short battery life that limits the duration of operation.

Unlike the aforementioned works, the research presented in this paper takes a different technical route to address this challenging issue. Instead of electric actuation, the transtibial prosthesis design in this paper utilizes pneumatic actuation, which is well known for its capability of generating large force and power output with light weight and compact volumetric profile [13]. Leveraging this unique advantage, Sup et al. developed a powered transfemoral prosthesis, in which both knee and ankle joints are powered with pneumatic cylinders [14]. Note that, in this design, the ankle actuator share the 'shank' space with the knee actuator, and thus cannot be isolated to form a standalone transtibial prosthesis. There have also been attempts of utilizing pneumatic muscle actuators in transtibial prosthesis design [15], and walking experiments have been conducted to demonstrate the feasibility of this new actuation approach [16]. However, a pneumatic muscle actuator expands radially during operation, which requires additional clearance from the supporting structure and enlarges the volumetric profile of the prosthesis.

Different from these earlier attempts, the work presented in this paper aims at developing a highly compact transtibial prosthesis with a potential for future practical use in amputees' daily life. To achieve this goal, the design of the prosthesis is based on the pneumatic cylinder actuator. A cylinder-type actuator does not expand in operation, which enables it to be placed horizontally under the rotational axis of the ankle. This spatial arrangement minimizes the height of the prosthesis, enlarging the population that can potentially benefit from this device. There are three specific objectives in this work, which will be detailed in the following sections: (1) Designing the pneumatically actuated transtibial prosthesis (Section 2); (2) Developing a prosthesis control algorithm for walking (Section 3); (3) Conducting preliminary testing on a transtibial amputee (Sections 4).

2. PROSTHESIS DESIGN

2.1 Design specifications

Ideally, a transtibial prosthesis should restore the lost limb functions in supporting the body weight and providing torque/power output for locomotion. Meanwhile, the weight and volumetric profile of the prosthesis should be kept under those of the corresponding limb segments for daily-use comfort and aesthetic reason. As such, the design specifications are primarily determined according to principles in human anatomy and biomechanics of locomotion. Furthermore, for the development of this first prototype, off-the-shelf commercial products are preferred for the choices of major components, with the purpose of reducing the cost and risk in the prosthesis design, fabrication, and testing.

The weight and volumetric profile of the prosthesis are determined for a male subject of 75 kg, approximately 50th percentile in weight. The height of the prosthesis should be comparable with or lower than that of a traditional high-profile transtibial prosthesis, which is approximately 180 mm from ground to the prosthetic adapter [17]. Taking the nominal height of a typical user at 1.75 m, the percentage of the missing limb mass under 180 mm from the ground surface (versus the entire body mass) is approximately 2.5% [18]. Multiplying the percentage to the nominal body weight, the mass of the missing limb segment is approximately 1.875 kg, which serves as the upper limit for the weight of the prosthesis.

The kinematic and kinetic specifications are determined according to the biomechanical data of ankle joint in walking. Note that the ankle motion is relatively slow in comparison with the knee, and thus the joint velocity is not considered as a major concern. Therefore, the design calculation is primarily focused on meeting the requirements in actuation torque and range of motion. According to the data by Winter [2], a joint angle-torque plot (for slow walking) is shown in Fig. 1. As indicated in this figure, the peak torque is approximately 115 Nm, which occurs at the joint angle of 8°. Note that the torque is a function of joint angle, and thus the prosthetic ankle should ideally cover the entire torque trajectory of walking. This serves as the design goal for the torque capacity of the prosthetic ankle joint. Additionally, the range of motion is determined to be at least -20°~10°, which covers the entire joint angle trajectory in walking.

2.2 Mechanical Design

The mechanical design of the prosthesis is largely driven by the need for a compact package of the device. Specifically, reducing the prosthesis height is considered as a major goal to fit the amputees with amputation sites close to the ankle. Existing powered transtibial prostheses usually cannot be fitted to such patients due to their large heights compared with unpowered devices. A schematic of the actuation mechanism is shown in Fig. 2. Unlike the existing designs of pneumatically-actuated ankles, the actuator is arranged in the horizontal direction, driving the ankle motion through an inverted crank-slider mechanism. With this unique design, the actuator is completely under the rotation axis of the ankle joint, minimizing the height of the device.

Figure 2 also defines the design parameters are directly related to the kinematic and kinetic characteristics of the prosthesis, including the lengths a and b and the angle ϕ (when the ankle angle θ is zero). These parameters, combined with the bore size and stroke length of the actuators, determine the range of motion and the torque capacity of the prosthetic ankle joint. Specifically, the maximum actuation force provided by the pneumatic cylinders F_{MAX} is determined according to the maximum air pressure in the actuator (usually the supply gauge pressure P_s), in combination with the corresponding piston area A_p :

$$F_{MAX} = P_s \cdot A_p \quad (1)$$

where

$$A_p = \begin{cases} \pi D^2 & \text{for the rodless chamber} \\ \pi (D^2 - d^2) & \text{for the chamber with rod} \end{cases} \quad (2)$$

In the above equation, D is the piston diameter, and d is the rod diameter. Note that the piston areas in the two directions are slightly different due to the existence of the piston rod. To obtain the corresponding maximum torque, the method of virtual work can be applied, which results in the following equation:

$$\tau = F \cdot \frac{dx}{d\phi} \quad (3)$$

where τ is the torque corresponding to a certain actuation force F . According to the geometric relationships shown in Fig. 2, actuator length x can be expressed as a function of the angle ϕ according to the following equations:

$$x = \sqrt{a^2 + b^2 - 2ab \cos \phi} \quad (4)$$

Substitute (4) into (3), the following equation can be obtained

$$\tau = F \left(\frac{ab \sin \phi}{\sqrt{a^2 + b^2 - 2ab \cos \phi}} \right) \quad (5)$$

Utilizing this equation, the torque capacity can be calculated according to the selected set of parameters. Note that, due to the prototype nature of the current device, off-the-shelf components are preferred. As such, the design parameters are determined by selecting a commercial pneumatic actuator and choosing the design parameters accordingly. In the current prototype, the actuator is a double-acting pneumatic cylinder (171.25-DP) in the Original Line from Bimba Manufacturing (University Park, IL, USA), with 38 mm (1.5 inch) bore size and 32 mm (1.25 inch) stroke. The other design parameters are listed in Table 1.

The design described above is able to provide a range of motion of $-25^\circ \sim 15^\circ$, exceeding the kinematic requirement. The torque capacity (obtained under the maximum pressure of 2 MPa) is shown in Fig. 3, which also displays the required torque curve associated with slow walking. The actuation system is able to provide sufficient torque within the majority of the range of motion, while only the peak torque at $3 \sim 8^\circ$ is beyond the torque capacity curve. Note that it would be possible to use a pneumatic cylinder with a greater bore size to meet the peak torque requirement. However, a bigger cylinder would significantly increase the size of the prosthesis. As such, the authors decided to use a relatively small cylinder to maintain the low volumetric profile of the prosthesis. It is worth mentioning that, as observed in the human subject experiments, the lowered torque capacity did not seem to affect the user's gait quality, as demonstrated by the results obtained in walking experiments.

Note that the design described above targets slow walking since it is the primary mode of motion for the majority of transtibial amputees, but the prosthesis can also be used for other locomotive modes. Within the same level walking mode, with the increase of speed, the

maximum physiological torque increases slightly (from 115 Nm for slow walking to 122 Nm for natural walking and 130 Nm for fast walking), and high torque output (>100 Nm) spans a wider angular range (from $\sim 3.5^\circ$ for slow walking to $\sim 5.5^\circ$ for natural walking and $\sim 7.5^\circ$ for fast walking), according to the data from Winter [2]. As such, an amputee user fitted with the prosthesis will experience more difficulty when walking at higher speed. For stair ascent and descent, the maximum physiological torques are lower than that for slow walking (96 Nm for stair ascent and 85 Nm for stair descent), according the data from Riener et al. [19]. The prosthesis is expected to provide sufficient torque for these modes. For sloped walking, biomechanical data by McIntosh et al. [20] indicate that there is an increase in peak torque for both upslope and downslope walking, with the only exception of 5° upslope walking (in which the peak torque decreases by $\sim 50\%$ compared with that for level walking). As such, an amputee with the prosthesis may also experience difficulty in sloped walking except in small-incline upslope walking. In summary, the prosthesis can provide sufficient torque capacity for slow walking and stair ascent/descent, but has difficulty in providing the high peak torques associated with faster walking and sloped walking.

After the design parameters are determined, the authors also conducted an analysis on the attainable speed of the prosthetic joint to verify that the prosthesis is able to provide the desired ankle speed in gait. According to the related biomechanical data [2], the maximum angular velocity of the ankle in slow walking is approximately 2.97 rad/s. On the other hand, for the pneumatic cylinder-type actuators, the maximum linear velocity can reach as high as 1000 mm/s with oil-free air, and may reach higher speed if lubricated air is used [21]. For a conservative estimation, take 1000 mm/s as the attainable linear velocity for the cylinders. According to the design parameters listed in Table 1, the corresponding attainable joint velocity is 21.9~28.9 rad/s (depending on the joint angle), far exceeding the maximum angular velocity in gait.

2.3 Instrumentation and the Complete Prosthesis Package

Multiple sensors are incorporated into the prosthesis design to provide the required information for the control prosthesis. The most important sensor signal required by the controller (to be described in the subsequent section) is the joint angle, which is provided by a string potentiometer mounted in parallel with the actuator (ZX-PA-1.5 analog position transducer, UniMeasure, Corvallis, OR, USA). This string potentiometer measures the displacement of the piston rod, which can be translated into joint rotation with simple trigonometric calculation. Additionally, for the modulation of the actuation force, a load cell (ELPF-T3E-500L, Measurement Specialties, Hampton, VA, USA) is mounted in line with the actuator. With the force signal measured with this load cell, a simple PID force control loop can be used to obtain the desired actuation force/torque to support the user's locomotion. With these sensors included, the total weight of the prosthesis is 0.9 Kg, and the height of the prosthesis is 98 mm, both far below their respective upper limits. The appearance of the prosthesis is shown in Fig. 4.

Currently, the control components of the prosthesis are implemented off-board, including the control valve (ZS-V-13000, Enfield Technologies, Shelton, CT, USA) and the control

calculation. In the walking experiments, the valve was attached to the pylon mounted between the prosthesis and the user's prosthetic socket, and the controller was implemented on the National Instruments LabVIEW platform, running on a desktop computer. In the future, the control valve, along with a microcontroller-based control system, will be integrated into the prosthesis itself. Furthermore, to form a completely self-contained prosthesis, a compact, portable pneumatic supply will also be incorporated. Two possible candidates are high-pressure carbon fiber compressed air tank and liquid propellant-based pneumatic supply [22]. A carbon fiber air tank stores compressed air under high pressure (up to 31 MPa, or 4,500 Psi), and there are multiple commercial products on the market, primarily serving the purpose of powering paintball guns. The availability of commercial products makes it an ideal short-term solution. The alternative solution, liquid propellant-based supply, generates high-pressure gas through the catalytic reaction of liquid propellant. According to a study by Goldfarb et al., a liquid-propellant-powered actuator offers an order of magnitude advantage over a comparable battery-powered DC motor actuated system in a system-level comparison of actuation performance [22]. As such, liquid-propellant-based pneumatic actuation is a highly competitive approach that can potentially lead to a practical transtibial prosthesis with long duration of operation. Compared with the traditional electromechanical actuation, the liquid-propellant-based pneumatic actuation does come with a few weaknesses, including: (1) the system configuration is more complex with more moving parts; (2) the noise level is higher; and (3) the system dynamics are highly nonlinear, which makes the motion control more challenging. However, in the authors' opinion, the gain in actuation performance outweighs these weaknesses, and pneumatic actuation can still be viable as a competitive actuation approach for future transtibial prostheses.

3. PROSTHESIS CONTROL

With the capability of supplying a significant amount of torque on the prosthetic ankle joint, this powered transtibial prosthesis requires a reliable and effective control approach to enable the natural and coordinated interaction with the user. To enable such interaction, the general impedance control framework is adopted for the development of the prosthesis controller in this work. The general impedance control theory was proposed by Hogan in 1980s for the control of robotic manipulators in the interactive tasks [23]. This control approach is especially useful in the robotic applications that involve interaction with human, considering the important role of impedance modulation in the human motion control [24,25]. The application of impedance control in lower-extremity prosthesis control is proposed by Sup et al., and its effectiveness has been demonstrated in the control of powered transfemoral prostheses with active knee and ankle joints [10].

For the application of impedance control in transtibial prostheses, there are two key steps in the process, including the proper segmentation of a walking cycle into a finite number of states (or phases), followed by the representation of the ankle biomechanical behavior with proper impedance parameters within each state. Note that this general method was used in the design of electric motor-actuated prostheses and guided the selection of spring elements [5,6]. Using this method for the controller design, the constraints associated with physical

springs no longer apply, resulting in more flexibility in the selection of impedance parameters.

3.1 Ankle Biomechanics in Walking

The impedance control approach in this work aims to simulate the biomechanical behavior of the ankle in human walking, which has been the topic of a large amount of studies. As the basis of the analysis, the ankle angle-torque trajectory of slow walking is plotted in Fig. 5a, utilizing the data in [2]. With the large variation in the shape of the curve, it can be clearly seen that it is impossible to describe the entire cycle with a single impedance representation, highlighting the importance of segmentation. The segmentation of the torque curve can be conducted with a few well-defined transition points (labeled as A~D in Fig. 5a), with which the gait cycle is divided into four distinct stages. The following is a brief summary of these events and phases:

Event A: Heel Strike: the heel touches the ground, marking the start of the gait cycle.

Phase #1 (A→B): Early Stance (ES). In this state the foot plantarflexes until it lays flat on the ground, and the ankle provides a small resistive torque that increases with the joint rotation.

Event B: Foot Flat: the foot starts to lie completely flat on the ground.

Phase #2 (B→C): Middle Stance (MS). In this state the foot stays flat on the ground while the shank moves forward (dorsiflexion), and the ankle provides a rapidly increasing torque.

Event C: Maximum Dorsiflexion: the joint angle reaches the maximum value.

Phase #3 (C→D): Late Stance (LS). In this state the foot pushes the ground and propels the body forward, with a torque that decreases with the plantarflexion of the ankle.

Event D: Toe Off: the toe leaves the ground.

Phase #4 (D→A): Swing (SW). In this state the foot is completely in the air without contact to the ground, and the ankle experiences fast dorsiflexion with low torque, creating sufficient ground clearance and getting ready for the heel strike of the next cycle.

3.2 Impedance Modeling

In general, the impedance applied to a robotic joint can be modeled as the resistance associated with a set of virtual springs and dampers. For a virtual spring, the corresponding torque is a function of the joint position; while for a virtual damper, the corresponding torque is a function of the joint velocity. The equation for the impedance torque is

$$\tau = \sum_{i=1}^n K_i (\theta - \theta_0)^{2i-1} + \sum_{j=1}^m P_j (\dot{\theta})^{2j-1} \quad (10)$$

where K_i 's and P_j 's are the stiffness and damping coefficients, respectively, and θ_0 is the equilibrium position of the virtual spring. Note that this equation includes the higher order terms for the general nonlinear spring and damper behaviors. Mathematically, it is possible to match a torque curve with infinite sets of parameters. However, to develop a practical prosthesis controller, its dynamic behavior should closely match the biomechanical behavior of the ankle with minimum complexity. To obtain a good understanding of such behavior, the original segmented torque curve (as shown in Fig. 5a) is fitted with simple linear-spring torque curve within each state, as shown in Fig. 5b. The impedance behavior within each phase is summarized below:

Phase #1 (Early Stance): The ankle functions like a spring with moderate stiffness (~ 1 Nm/deg) to provide shock absorption and appropriate plantarflexion resistance before foot flat.

Phase #2 (Middle Stance): The ankle functions like a very stiff spring (~ 8 Nm/deg) to absorb energy to get ready for the following push-off. Better matching to the nonlinear behavior can be obtained by adding higher order terms in the spring torque equation.

Phase #3 (Late Stance): The ankle functions like a stiff spring, with the stiffness lower than that in Phase #2 (~ 5 Nm/deg). However, with an equilibrium position significantly less than that in the Phase #2 ($\sim -16^\circ$), the ankle in this state generates higher torque output than in Phase #2, constituting the powered push-off in walking. As in Phase #2, better matching to the nonlinear behavior can be obtained by adding higher order terms in the spring torque equation.

Phase #4 (Swing): The ankle functions like a spring with very low stiffness, returning the ankle to a slightly dorsiflexed position ($\sim 2^\circ$) to get ready for the next heel strike.

Based on the impedance behavior summarized above, the actual control algorithm implemented for the prosthesis control takes the following form:

$$\tau = K(\theta - \theta_0) + P\dot{\theta} \quad (11)$$

in which a linear stiffness term (K is the linear stiffness coefficient) plays the primary role in the prosthesis control, and a damping term (P is the linear damping coefficient) is added to dampen the motion and improve the stability of the motion. The corresponding parameter values in each state are obtained through repeated tuning in actual walking experiments, as described in Section 4.

3.3 Controller Implementation

For the implementation of the controller, a finite-state machine is constructed and executed in real-time, as shown in Fig. 6. Four states are incorporated, as defined in the previous section.

Note that an important component of the finite-state machine is the switching conditions, which are usually associated with specific events in the walking gait (e.g., heel strike). A common approach to detect such events is through contact sensors, such as foot switch or force- sensing resistors under the foot. However, such sensors tend to unreliable and

susceptible to failure, according to the authors' experience. As such, the switching conditions in this work are constructed based solely on the ankle angle signal (θ) and its derivative ($\dot{\theta}$), as shown in Fig. 6. Specifically, the heel strike is detected according to the onset of plantarflexion ($\dot{\theta} < 0$). To avoid the false switching to Phase #3 (Late Stance), an additional condition ($\theta < \Theta_1$) is imposed, in which Θ_1 is a threshold value of approximately $2\sim 3^\circ$. The foot flat is detected according to the onset of dorsiflexion ($\dot{\theta} > 0$). The maximum dorsiflexion is detected when the ankle angle exceeds the second threshold value Θ_2 , with the specific value at approximately $7\sim 8^\circ$. The toe off is detected when the ankle angle decreases below the third threshold value Θ_3 , with the specific value at approximately -16° . With these switching conditions, the controller can be implemented reliably on the prosthesis in the repeated walking experiments.

The state machine (Fig. 6), combined with the impedance-based joint behavior representation (Eq. 11), generates the desired actuation torque command for the powered ankle joint. To obtain this desired actuation torque, the torque command is converted to the desired actuation force with the following equation:

$$F_d = \tau_d \left(\frac{\sqrt{a^2 + b^2 - 2ab \cos \phi}}{ab \sin \phi} \right) \quad (12)$$

The desired force, as calculated by the above equation, is compared with the measured actuation force F to generate the error signal $e = F_d - F$. Based on this error signal, the standard PID control is applied to obtain the valve command for the real-time implementation of the controller. The control gains were tuned in the experiments to generate a quick response without jeopardizing stability in operation.

4. PROSTHESIS TESTING

After the fabrication of the prosthesis prototype and the development of the prosthesis controller, treadmill walking experiments have been conducted to evaluate the prosthesis' performance in restoring an amputee user's locomotion functions. The experimental protocol is approved by the Institutional Review Board of the University of Alabama. One unilateral transtibial amputee participated in the study. The male participant was 22 years old, 175 cm in height, and weighed 56.7 kg. An extension pylon was used to fit the prosthesis to the subject. The pylon also served as the base for mounting the control valve in the experiments, such that the delay due to the connection tube could be minimized. Figure 7 shows the subject's lower limb when fitted with the prosthesis.

In the walking experiments, repeated tuning was conducted to obtain a satisfactory performance. In the tuning, the trajectory of the prosthetic joint was compared with the standard joint trajectory in biomechanical literature (such as [2]) to form an objective evaluation of the gait quality. Additionally, visual observation and the feedback from the test subject also played important roles for the adjustment of control parameters. The control parameters obtained from the tuning are summarized in Table 2. In the experiments, impedance values obtained from the physiological analysis (Section 3.2) were used as the

initial control parameters, and these parameters were modified according to the measured joint angle/torque trajectories and the subject's feedback. It is worth mentioning that, for the stiffness value in mid-stance, physiological analysis indicated a much higher value of 8 Nm/deg, representing a very stiff spring. The control parameter, however, was reduced to its current value (4 Nm/deg) after repeated tuning. The reason for this discrepancy, in our opinion, could be the subject's experience with his daily-use prosthesis, i.e., the subject has gotten used to the less stiff passive prosthesis, and thus his gait was different from the standard gait of healthy individuals. Although the stiffness and damping values are the same for mid stance and late stance, the equilibrium point for late stance (-25°) is much lower than that for mid stance (-3°), generating a much higher torque for the push-off behavior in the late stance.

The performance of the prosthesis and its controller is shown in Figures 8~10. Figure 8 displays a comparison between the angular trajectories of the prosthetic joint versus the biological joint, utilizing the standard biomechanical data in [2]. As shown in the figure, the prosthetic joint trajectory is smooth and close to the standard joint trajectory in the majority of the gait cycle. The prosthetic joint trajectory displays more obvious transition between adjacent phases, presumably because of the use of the finite-state impedance controller. Fig. 9 displays a comparison between the actuation torque of the prosthetic joint versus the measured torque in the biological joint in level walking. Again, the actuation torque curve resembles the biological joint torque curve, but displays more obvious transitions due to the use of the finite-state controller. Last but not least, Fig. 10 shows the power output within a gait cycle. As shown in this figure, the actuated ankle joint is able to generate over 120W peak power during the powered push-off, which is fundamentally different from the passive behavior of a traditional non-powered transtibial prosthesis. As such, a prosthesis user can leverage this active energetic behavior and enjoy a more natural and comfortable walking. This has been confirmed by the feedback of the test subject in the experiments.

5. CONCLUSIONS

This paper presents a unique robotic transtibial prosthesis that utilizes a pneumatic cylinder-type actuator to drive the prosthetic ankle joint. To reduce the prosthesis height, the pneumatic cylinder is arranged horizontally and drives the ankle joint through an inverted crank-slider mechanism. In the design process, the parameters were determined to provide the desired torque capacity while reducing the weight and volume of the system. The resulted prosthesis design is able to supply sufficient actuation torque for level walking within the majority of the range of motion. For the walking control of the prosthesis, a finite-state impedance controller has been developed based on the analysis of the biomechanical behavior of the biological ankle joint in walking. The angle-torque curve was segmented in to a number of individual phases, and within each phase the joint behavior is represented with a simple impedance behavior. For the implantation in the prosthesis control, the parameters were tuned in the treadmill walking experiments. After repeated tuning, the prosthesis was able to provide an improved gait compared with the traditional passive prosthesis, according to the data collected in the experiments and the feedback provided by the test subject.

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Figure 1.
The torque trajectory of the ankle in slow walking (data from [2]).

Figure 2.

The actuation mechanism of the transtibial prosthesis. Note that the shorter link (with length b) is rigidly connected to the base of the prosthetic connector, with the whole component rotating with the ankle axis as the center.

Figure 3.
Comparison of the actuation torque capacity versus the ankle torque trajectory in walking.

Figure 4.
Photo of the prosthesis prototype when removed from the foot shell.

Figure 5.

Analysis of the ankle dynamic behavior in walking: (a) Segmentation of the original torque trajectory, (b) Simple impedance representation to understand the dynamic behavior within each phase.

Figure 6.
Finite-state machine for the implementation of the prosthesis controller.

Figure 7.
The test subject fitted with the transtibial prosthesis.

Figure 8. Comparison of the trajectory of the prosthetic ankle joint versus the standard ankle trajectory of healthy subjects in level walking (data from [2]).

Figure 9.

Comparison of the torque trajectory of the prosthetic ankle joint versus the standard ankle torque trajectory of healthy subjects in level walking (data from [2]).

Figure 10.
Power output of the actuated ankle joint.

Table 1

The design parameters of the transtibial prosthesis prototype.

Parameter	Value	Unit
a	158	mm
b	46	mm
ϕ (when $\beta=0$)	98	$^{\circ}$

Table 2

Parameters of the prosthesis controller.

State	K (N-m/deg)	P (N-m-s/deg)	θ_0 (deg)
<i>ES</i>	<i>0.9</i>	<i>0.1</i>	<i>1</i>
<i>MS</i>	<i>4</i>	<i>0.1</i>	<i>-3</i>
<i>LS</i>	<i>4</i>	<i>0.1</i>	<i>-25</i>
<i>SW</i>	<i>0.9</i>	<i>0.01</i>	<i>1</i>