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A transtibial prosthesis using a parallel spring mechanism

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Abstract— **Prosthetic legs have been used to restore function in the lower limbs lost due to amputation. Early designs including prosthetic legs with a passive joint or without any joint as well as the Energy Storing and Releasing (ESR) feet have shown deficiency in push-off torque, which results in asymmetric gait pattern, slower walking speed, and higher cost of transportation. Although powered prosthetic legs address the aforementioned problems, they suffer from lower energy efficiency, higher volume and weight. In this paper, a powered transtibial prosthesis using a Parallel Elastic Actuator (PEA) is proposed in order to generate the joint torque needed for walking with a lower-powered actuator for lighter and more compact design. A non-linear spring mechanism is proposed to generate the spring torque as needed. The implemented prosthetic leg is evaluated with three intact subjects. The experimental results shows that smaller torque is required for the motor with the spring mechanism. Therefore, less electrical power is consumed when the spring mechanism is used, which implies a lower-powered actuator is sufficient to generate the joint torque needed for walking.**

I. INTRODUCTION

Walking is one of the most important functions that are affected by the amputation in the daily lives of the lower limb amputees. In an effort to alleviate the adverse effects of amputation, prosthetic legs have been developed to restore the lost function of the amputees. Main focuses of early designs of the prosthetic legs with a passive or no ankle joint were partial restoration of walking function and better appearance [1]. An example is the Solid Ankle Cushioned Heel (SACH) foot, which has no ankle joint and therefore does not generate the torque needed to push the body forward when walking [2]. However, studies have shown that deficiency in ankle torque caused asymmetric gait, reduced walking speed, higher step-to-step transition cost [3], [4], [5], [6].

For better energy efficiency, Energy Storing and Returning (ESR) feet have been developed using elastic components and mechanical springs capable of storing energy [7]. Using the ESR feet, some of the mechanical energy during the double support phase and early stance phase is stored in the form of elastic potential energy and released to push the body forward at the toe-off. In contrast to the early designs of the ESR feet, which relied on the properties of the materials used for storing and releasing energy, advanced designs of the ESR feet have used clutches or latches to control the timing of the energy release [8], [9].

Since the ankle joint produces net positive work as shown in [10], however, deficiency in push-off at toe-off remains even with the ESR feet, which justifies the development of powered prosthetic legs. Studies have shown that powered prosthetic legs improve the walking speed and metabolic cost of transport [11], [12], [13]. Powered prosthetic legs have been developed using various types of actuators such as an Eletrohydrostatic Actuator (EHA) [14] and electrical motors [15]. Although these actuators provided the torque needed for walking, no mechanical energy was stored which lead to poor energy efficiency.

For more natural gait pattern and efficient energy consumption, there have been researches to include elastic components in the actuators to store energy, which is dissipated otherwise during walking. Series Elastic Actuators (SEA's), which are actuators with the elastic components attached in series, have been used for better energy consumption [16], [17], [18], for modulating the impedance of the powered prosthetic legs [19], [20], [21]. There have been researches on designing energy-efficient actuators using an elastic component attached to an electrical motor in parallel [20], [22], [23], which is called Parallel Elastic Actuators (PEA's). The PEA's have been used in a gait assisting exoskeleton [24].

When the PEA's are used for transtibial prosthetic legs, nonlinear spring torque are desirable since the ankle joint has nonlinear stiffness as shown in Fig. 1, which is re-produced using data in [25]. Nonlinear spring torques have been realized using a linear spring and a nonlinear transmission such as a cam mechanism [26], [27], linkage system [28], which could yield bulky, heavy, and complex designs.

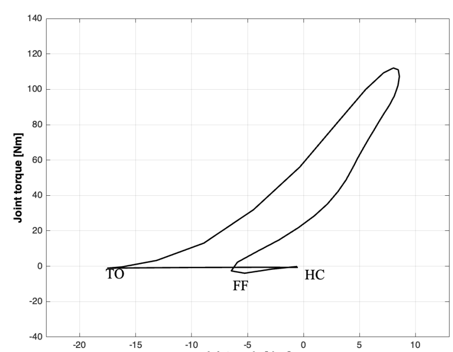

Fig. 1. Torque-vs-angle characteristics of a healthy human during level walking. Data is taken from [25].

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In this paper, a transtibial prosthetic leg actuated by a PEA using cable mechanism for the non-linear spring torque is introduced. The cable mechanism enables us to design a non-linear transmission with minimal increase of volume, weight, and complexity of the design. Since the spring mechanism generates torque as the angle of the joint changes during a gait cycle, the required motor torque is reduced. The experiment with three healthy subjects revealed that the required motor power when walking on a level surface was reduced due to the PEA, which implied lighter prosthetic leg is feasible with less-powered actuator.

In Section II, the working principle and the design of the prosthetic leg are explained and the implemented prosthetic leg is introduced in Section III. The experimental results in Section IV is followed by conclusion in Section V.

II. DESIGN

The transtibial prosthesis proposed in this paper is designed to generate ankle torque needed for walking using a PEA, which results in reduction of the required motor torque. In this Section, working principle and the concept of the prosthetic leg with PEA are explained.

A. Working principle

During one gait cycle, an ankle joint of an intact human produces torques needed for walking, which consists of the alternating stance phase and swing phase. The stance phase is when the prosthetic leg is supporting the rest of the body and the swing phase is when the prosthetic leg is swinging forward in the air. Torque-vs-angle characteristics of the ankle of an intact human walking on a level surface at a constant speed is shown in Fig. 1, which is a reproduction of the data provided in [25]. After heel contact (HC), plantarflexion occurs until the foot becomes flat on the ground (FF). After FF, dorsiflexion occurs, during which torque and angle of the ankle increase, and is followed by plantarflexion after reaching the maximum dorsiflexion until the toe leaves off the ground (TO), at which the stance phase ends and the swing phase begins. Note that the torque-vs-angle characteristics of the ankle joint reveals that the ankle joint largely acts as a "non-linear spring."

It is desirable for the transtibial prosthesis to have a torque-vs-angle characteristics similar to Fig. 1 for a natural gait pattern. However, the wide range of the joint torque requires a high-powered actuator, which causes higher volume and weight of the prosthetic leg.

B. Parallel Elastic Actuator

A Parallel Elastic Actuator (PEA) is an actuator consisting of an electrical motor with an elastic component attached in parallel to the motor, see Fig. 2. PEA's have been used for efficient energy consumption of exoskeletons and prosthetic legs in [22], [24]. The joint torque generated by the PEA consists of the torques by the electrical motor and the elastic component. Due to the elastic component, the torque by the electrical motor is reduced, which yields more energy efficient actuation while producing the joint torque needed for walking.

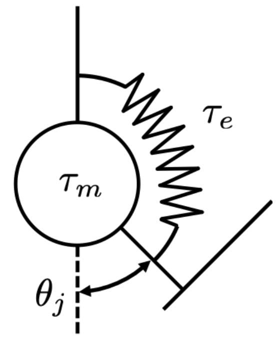


Fig. 2. Concept of a prosthetic leg using a PEA

However, having a linear spring attached to the ankle in parallel with the motor might results in higher motor torque required to compensate the undesired spring torque in some portion of the range of motion. As shown in Fig. 1, smaller stiffness at the joint for the angles less than $-5°$ and larger stiffness for the angles greater than $5°$ are desirable. Therefore, a mechanism to generate non-linear spring torque using a linear spring needs to be designed.

C. Spring mechanism

The prosthetic leg has shank and foot segments, which are connected at a revolute joint called "ankle." The ankle joint is actuated by the PEA whose elastic component of the PEA consists of a linear spring, idle pulleys, inner and outer acting pulleys, and cable. One end of the cable is attached to the linear spring, which is installed at the shank segment of the prosthesis. The cable goes through a number of pairs of the idle pulleys and inner and outer acting pulleys for the other end of the cable to be attached to the foot segment of the prosthesis. The idle pulleys are installed at the shank segment. The inner and outer acting pulleys are attached to the foot segment, see Fig. 3.

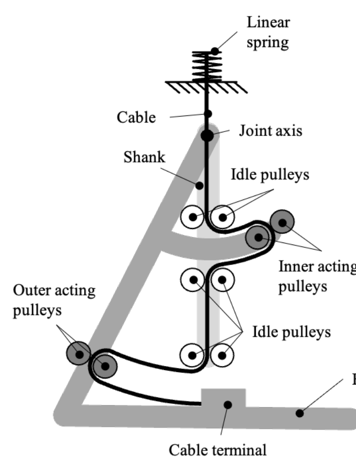
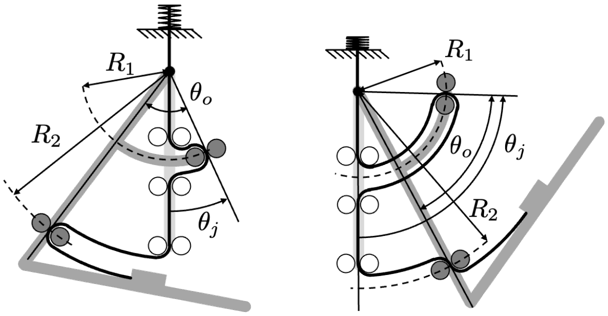


Fig. 3. Schematics of the proposed prosthetic leg with PEA. The inner and outer acting pulleys are installed at the foot segment whereas the idle pulleys are installed at the shank.



(a) Plantarflexion (b) Dorsiflexion

Fig. 4. The path of the cable when plantarflexion and dorsiflexion occur. When radii of the pulleys are 0, length of the cable path does not change during plantarflexion while length of cable path increases during dorsiflexion.

When the foot segment rotates about the ankle joint, the inner acting pulleys rotate around the joint axis at the distance of R_1 whereas the rotation radius of the outer acting pulleys is R_2. The inner and outer acting pulleys are apart from each other by an offset angle of θ_o, see Fig. 4. Assuming the radius of each pulley is negligible, the length of the cable path from the joint axis to the cable terminal at the foot segment is as follows.

$$L = \begin{cases} (2R_1 - R_2)\theta_j + R_2\theta_o + R_2 + C & 0 \leq \theta_j \leq \theta_o \\ (2R_1 + R_2)\theta_j - R_2\theta_o + R_2 + C & \theta_j > \theta_o \end{cases}, \tag{1}$$

where C is the length of the cable from the outer acting pulley to the cable terminal at the foot segment. Note that if $R_2 = 2R_1$, then the length of the cable path does not change for $0 \leq \theta_j \leq \theta_o$ and increases linearly with respect to θ_j when $\theta_j > \theta_o$.

III. Implemented Prosthetic Leg

A. Hardware

The implemented prosthetic leg is shown in Fig. 5. It consists of a shank and a foot segment, which are connected at the ankle joint. The ankle joint is actuated by a PEA that has an electric motor and the spring mechanism connected in parallel with each other.

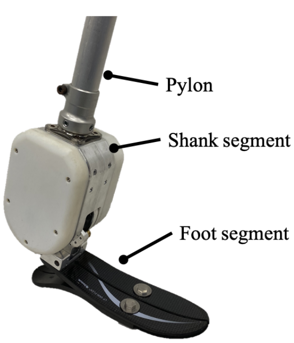


Fig. 5. Implemented prosthetic leg.

Fig. 6 shows cross-sectional views of the prosthetic leg using the spring mechanism parallel to the electrical motor. The shaft of the electrical BLDC motor (RBE(H)-00711, Kollmorgen) is connected to the input of the harmonic gear reducer (CSD-20-100, Harmonic drive) with a gear ratio of 100:1 via timing belts and pulleys with a gear ratio of 3:1, which yields in a total gear ratio of 300:1. The output of the harmonic gear reducer is attached to the foot segment. A rotary magnetic encoder (RMB28, Rotary Magnetic Encoder) is used to measure the position of the joint. For the cable, Dyneema® cable (DX core 99) with the diameter of 2.5mm is used for its capability to withstand the tension up to 725kg and easy handling. The linear spring is inside of the pylon, which is not shown in Fig. 6.

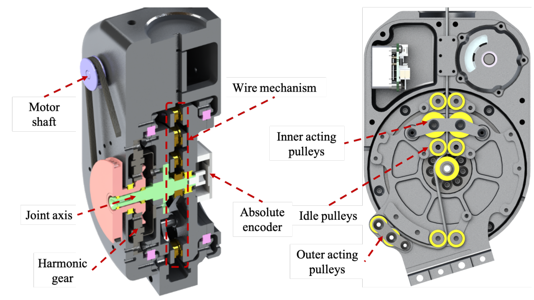


Fig. 6. Cross sectional views of the prosthetic leg. The motor torque and the spring torque are combined to actuate the ankle joint.

Off-the-shelf parts are used for the pylon (Ottoobok, 2R50), adapter (Ottoobok, 4R69), and the foot segment (OSSUR, LP-Varifelx®). The weight of the implemented prosthetic leg is 2.2 kg, excluding the weight of the battery and circuit.

The location and the size of the pulleys have been determined by optimization under constraints imposed by the physical and material limitations in order to minimize the maximum torque required by the motor during walking. Let τ_j be the joint torque needed for walking, which is taken from [25]. Then, the motor torque is $\tau_m = \tau_j - \tau_e$, where τ_e is the spring torque. Parameters to be determined are the distance of the inner and outer pulleys from the joint axis, radii of the idle pulleys, inner acting pulleys, and outer acting pulleys, and the offset angle between the acting pulleys.

As the foot segment rotates around the joint axis, the length of the cable path changes. Since the cable is attached to the spring, the linear spring is compressed, which increases cable tension. Therefore, the tension of the cable is obtained as follows.

$$t_w = K\delta L, \tag{2}$$

where K is the stiffness of the spring and δL is the changes in the cable path. For a given cable tension t_w, the torque generated by the tension is obtained as follows.

$$\tau_e = |t_w|(R_1(\cos(\theta_1) + \cos(\theta_2)) + 2R_2), \tag{3}$$

where θ_1 and θ_2 are the angles of the cable tangent to the inner acting pulleys. See Appendix for derivation.

The cost function for optimization is defined as follows.

$$J = \min \max_{0 \leq t \leq 1} \tau_m(t), \tag{4}$$

where t is a phase variable with values between 0 (0% of gait phase) and 1 (100% of gait phase). Using the cost function in (4), the size and location of the pulleys are optimized, which are listed in Table I. See Appendix for the definition of the variables.

TABLE I

DESIGN PARAMETERS FOR THE SPRING MECHANISM

Item	R_1	R_2	r_i	r_{ia}	r_{oa}	θ_o
Values	27.5mm	48mm	4.5mm	7mm	4.5mm	50°

B. Modeling of the Spring Torque

For modeling of the torque generated by the spring mechanism, tension of the cable in the spring mechanism with respect to the angle of the joint was measured as shown in Fig. 7. Note that there existed discrepancies between the measured tension and the calculated tension using equation (1) and (2) due to non-zero radii of the pulleys and diameter of the cable.

The measured cable tension exhibited hysteresis which caused about 20% higher tension during dorsiflexion than plantarflexion.

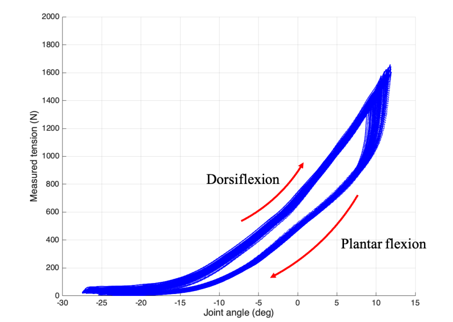


Fig. 7. Measured tension of the cable in the spring mechanism. It shows hysteresis with respect to the loading conditions.

Due to the hysteresis, data measured during dorsiflexion and plantarflexion were separately fitted with exponential functions shown in equation (5).

$$f(\theta_j) = ae^{b(\theta_j + c)} + d. \tag{5}$$

The fitted functions for loading and unloading conditions were averaged to be used for the controller. The estimated parameters for loading and unloading conditions are listed in Table II.

Using equation (3), the torque by the spring mechanism was calculated.

TABLE II

ESTIMATED PARAMETERS FOR SPRING TORQUE

Parameters	a	b	c	d
Loading	28.6921	0.1082	2.0124	-4.9512
Unloading	29.9921	0.1142	-0.1211	-3.5214

C. Control Design

The implemented prosthetic leg was controlled using a PID controller. The joint trajectory was taken from [25] and modified so that it matches to the estimated walking speed and the joint angle of the joint was maintained at 0 after 80% of the gait phase in the swing phase. The modification of the trajectory was intended to ensure the configuration of the joint unchanged even with the uncertainty of the timing of the heel contact. The gait phase was detected with pressure sensors using a Force Sensitive Resister (FSR). The pressure sensors were installed under the heel and toe of the feet. Using the measured pressure, the beginning and the end of the stance phase and swing phase were estimated, which were used to estimate the cadence and the speed, see Fig. 8.

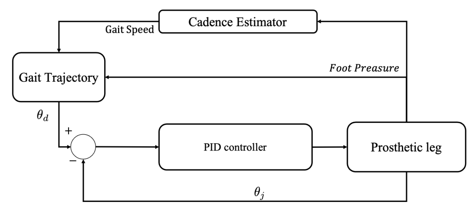


Fig. 8. Block diagram of the system

IV. EXPERIMENTAL RESULT

A. Setup for experiment

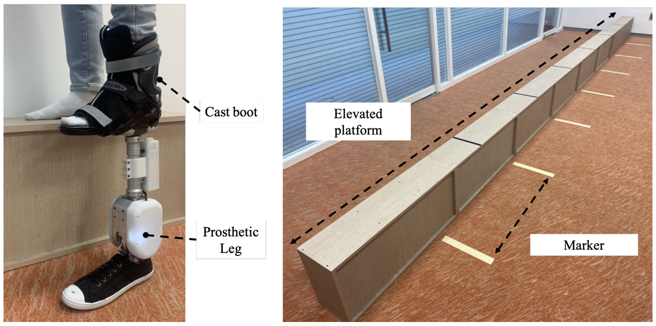
Three healthy subjects participated in the experiment. Detailed information on the subjects is listed in Table III. In order for the subjects to wear the prosthesis, it was rigidly attached to the sole of a ankle cast, see Fig. 9a. Note that the ankle cast prevented the ankle joint of the subjects from rotating. The subjects were instructed to walk with the implemented prosthesis on an elevated platform as shown in Fig. 9b. To maintain the walking speed and cadence constant, a metronome was set to 80 bpm and the ground was marked with markers 1-meter apart.

The subjects were asked to walk on the elevated platform under two conditions, which were walking with and without the spring of the prosthesis. The subjects were given a 20-minute adaptation period with the prosthesis before recording the experimental data. For each condition, the subject walked 7 steps on the platform 10 times while the joint positions and foot pressures were being recorded. Data for 50 steps were used for analysis since the first and last steps of each walk were excluded to remove acceleration and deceleration periods.

TABLE III

DETAILS OF SUBJECTS

Sub. No	Age(years)	Gender	Height(cm)	Weight (kg)
1	29	Male	175	80
2	31	Female	170	60
3	37	Male	174	68



(a) (b)

Fig. 9. Setup for experiment. (a) Prosthesis rigidly attached to the cast boot (b) The elevated platform and markers placed 1-meter apart.

B. Result

Fig. 10 shows the average motor torques with and without the spring mechanism. The red line represents the motor torque without the spring whereas the blue line shows the motor torque with the spring mechanism. The dashed green line is the joint torque produced by the prosthetic leg, which is calculated as the sum of the motor torque and the spring torque. Note that the joint torque is identical to the motor torque when the spring mechanism is not used. The joint torques with and without the spring mechanism are similar to each other, which implicates that the required joint torques when walking at a constant speed are identical regardless of the spring mechanism.

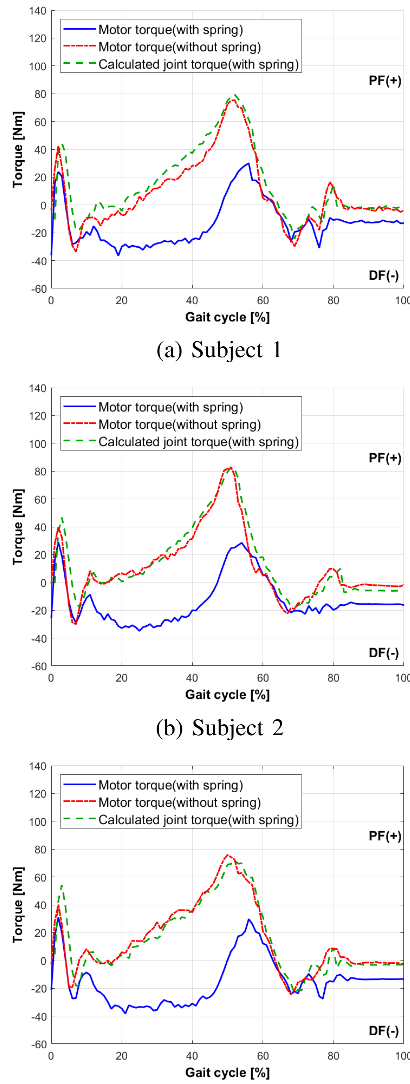
Compared to the motor torques without the spring mechanism, however, the motor torques with the spring mechanism are reduced while generating similar joint torques. Since the peak torques are reduced, low-torque motors are sufficient to actuate the prosthetic leg with the spring mechanism.

After 80% of the gait phase, where the joint angle is controlled to be 0 during the swing phase, non-zero motor torques are needed to compensate the undesired spring torque since the spring mechanism is designed to reduce the peak torque during entire gait cycle.

Powers of the motor, which is calculated as the product of the motor torque and joint velocity, are shown in Fig. 11. For each subject, the differences between the maximum mechanical motor power with the spring and without the spring are 74.97 W, 106.42 W, and 81.21 W, respectively. Note that the power reduced with the spring mechanism is different for each subject. Since the joint torques needed for walking are different with the different body masses, a fixed design of the spring mechanism would not result in an optimal result, which causes the motor to generate different powers for each subject.

V. CONCLUSIONS

A transtibial prosthetic leg using a parallel elastic actuator (PEA) was introduced. A spring mechanism was attached to the leg in parallel with the motor. The spring mechanism was composed of a linear spring, pulleys, and cable. The pulleys rotating around the joint axis changed the cable path of the spring mechanism and resulted in nonlinear spring



(a) Subject 1

(b) Subject 2

(c) Subject 3

Fig. 10. Averaged motor torque with and without the spring mechanism during one gait cycle. The red line represents the motor torque without the spring mechanism whereas the blue line shows the motor torque with the spring mechanism. The green dashed line is the joint torque.

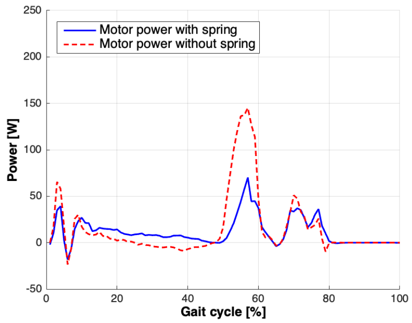
torques. Due to the spring mechanism attached to the leg in parallel with the electrical motor, less torque was needed to be generated by the motor, which reduced the required power of the motor. Experiment involving three healthy subjects wearing the prosthetic leg showed averaged motor torque during a gait cycle reduced, which resulted in less power required by the electrical motor.

APPENDIX

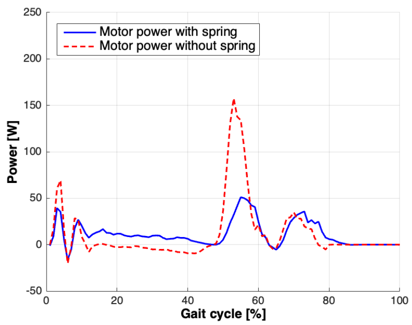
Let $P_0(x_0, y_0)$ denote the position of the joint axis. Let $P_i(x_i, y_i)$ represent the positions of the pulleys for $1 \leq i \leq 10$, see Fig. 12 for definition. Then, the positions of the pulleys are given as

$$x_1 = x_5 = x_7 = x_0 - r_i - d/2, \tag{6}$$

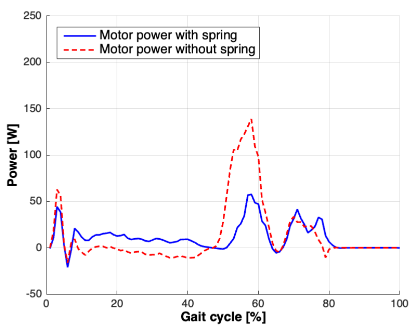
$$x_2 = x_6 = x_8 = x_0 + r_i + d/2, \tag{7}$$



(a) Subject 1



(b) Subject 2



(c) Subject 3

Fig. 11. Averaged required power for the electrical motor to generate during one gait cycle. Dotted red line shows the motor power without the spring mechanism and the blue line is the motor power with the spring mechanism

$$x_3 = x_0 + R_1 \sin(\theta_j - \sin^{-1}((r_{ia} + d/2)/R_1)), \quad (8)$$
$$x_4 = x_0 + R_1 \sin(\theta_j + \sin^{-1}((r_{ia} + d/2)/R_1)), \quad (9)$$
$$x_9 = x_0 + R_2 \sin(\theta_j - \sin^{-1}((r_{oa} + d/2)/R_2)), \quad (10)$$
$$x_{10} = x_0 + R_2 \sin(\theta_j + \sin^{-1}((r_{oa} + d/2)/R_2)), \quad (11)$$
$$y_1 = y_2 = y_0 - \sqrt{(R_1 - r_i - r_{ia} - d)^2 - (r_i + d/2)^2}, \quad (12)$$
$$y_3 = y_0 + R_1 \cos(\theta_j - \sin^{-1}((r_{ia} + d/2)/R_1)), \quad (13)$$
$$y_4 = y_0 + R_1 \cos(\theta_j + \sin^{-1}((r_{ia} + d/2)/R_1)), \quad (14)$$
$$y_5 = y_6 = y_0 - \sqrt{(R_1 + r_i + r_{ia} + d)^2 - (r_i + d/2)^2}, \quad (15)$$
$$y_7 = y_8 = y_0 - \sqrt{(R_2 - r_i - r_{oa} - d)^2 - (r_i + d/2)^2}, \quad (16)$$
$$y_9 = y_0 + R_2 \cos(\theta_j - \sin^{-1}((r_{oa} + d/2)/R_2)), \quad (17)$$
$$y_{10} = y_0 + R_2 \cos(\theta_j + \sin^{-1}((r_{oa} + d/2)/R_2)), \quad (18)$$

where R_1 is the distance of the inner acting pulleys from the joint axis, R_2 is the distance of the outer acting pulleys from the joint axis, r_i is the radius of the idle pulleys, r_{ia} is the radius of the inner acting pulleys, r_{oa} is the radius of the outer acting pulleys, and d is the diameter of the cable.

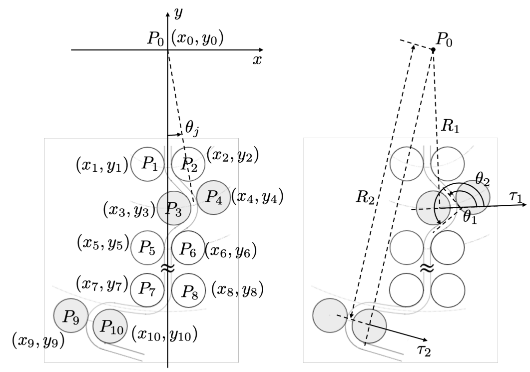


Fig. 12. Schematics of the spring mechanism for spring torque calculation

Let t_w be the cable tension. Let θ_j be the joint angle. Let θ_1, θ_2 be the angle of the cable tangent to the inner and outer acting pulleys, respectively. Then, the torque due to the tension is given as

$$\tau_1 = |t_w| R_1 (\cos(\theta_1) + \cos(\theta_2)) \quad (19)$$
$$\tau_2 = 2|t_w| R_2, \quad (20)$$

where τ_1 is the torque at the inner acting pulleys and τ_2 is the torque at the outer acting pulley. The torque by the tension is calculated as in (21)

$$\tau = |t_w| (R_1 (\cos(\theta_1) + \cos(\theta_2)) + 2R_2), \quad (21)$$

where

$$\begin{aligned}\theta_1 &= \frac{\pi}{2} - \tan^{-1}\left(\frac{x_2 - x_3}{y_2 - y_3}\right) + R_1 \sin^{-1}(r_a + d/2) + \\ &\cos^{-1}\left(\frac{r_i + r_a + d}{\sqrt{(x_2 - x_3)^2 + (y_2 - y_3)^2}}\right) - \theta_j, \quad (22)\\ \theta_2 &= \frac{\pi}{2} - \tan^{-1}\left(\frac{x_3 - x_6}{y_3 - y_6}\right) - R_1 \sin^{-1}(r_a + d/2) + \\ &\cos^{-1}\left(\frac{r_i + r_a + d}{\sqrt{(x_3 - x_6)^2 + (y_3 - y_6)^2}}\right) + \theta_j. \quad (23)\end{aligned}$$

when dorsiflexion occurs. Similarly θ_1 and θ_2 are calculated for the case of plantarflexion, which are given as follows.

$$\begin{aligned}\theta_1 &= \frac{\pi}{2} - \tan^{-1}\left(\frac{x_1 - x_4}{y_1 - y_4}\right) + R_1 \sin^{-1}(r_a + d/2) + \\ &\cos^{-1}\left(\frac{r_i + r_a + d}{\sqrt{(x_1 - x_4)^2 + (y_1 - y_4)^2}}\right) - \theta_j, \quad (24)\\ \theta_2 &= \frac{\pi}{2} - \tan^{-1}\left(\frac{x_4 - x_5}{y_4 - y_5}\right) - R_1 \sin^{-1}(r_a + d/2) + \\ &\cos^{-1}\left(\frac{r_i + r_a + d}{\sqrt{(x_4 - x_5)^2 + (y_4 - y_5)^2}}\right) + \theta_j. \quad (25)\end{aligned}$$

References

[1] A. J. Thurston, "Paré and prosthetics: the early history of artificial limbs," *ANZ Journal of Surgery*, vol. 77, no. 12, pp. 1114–1119, 2007.

[2] A. Staros, "The SACH (Solid-Ankle Cushion-Heel) foot," *Orthopedic & Prosthetic Appliance Journal*, pp. 23–31, 1957.

[3] H. Bateni and S. J. Olney, "Kinematic and kinetic variations of below-knee amputee gait," *JPO: Journal of Prosthetics and Orthotics*, vol. 14, no. 1, pp. 2–10, 2002.

[4] R. Waters, J. Perry, D. Antonelli, and H. Hislop, "Energy cost of walking of amputees: the influence of level of amputation," *The Journal of Bone & Joint Surgery*, vol. 58, no. 1, pp. 42–46, 1976.

[5] P. G. Adamczyk and A. D. Kuo, "Mechanisms of gait asymmetry due to push-off deficiency in unilateral amputees," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 23, no. 5, pp. 776–785, 2014.

[6] D. Wezenberg, A. G. Cutti, A. Bruno, and H. Houdijk, "Differentiation between solid-ankle cushioned heel and energy storage and return prosthetic foot based on step-to-step transition cost." *Journal of Rehabilitation Research & Development*, vol. 51, no. 10, 2014.

[7] D. C. Wing and D. A. Hittenberger, "Energy-storing prosthetic feet," *Archives of Physical Medicine and Rehabilitation*, vol. 70, no. 4, pp. 330–335, 1989.

[8] S. H. Collins and A. D. Kuo, "Recycling energy to restore impaired ankle function during human walking," *PLoS one*, vol. 5, no. 2, p. e9307, 2010.

[9] R. Holgate, T. Sugar, A. Nash, J. Kianpour, C. T. Johnson, and E. Santos, "A passive ankle-foot prosthesis with energy return to mimic able-bodied gait," in *ASME 2017 International Design Engineering Technical Conferences and Computers and Information in Engineering Conference*. American Society of Mechanical Engineers Digital Collection, 2017.

[10] D. A. Winter, "Energy generation and absorption at the ankle and knee during fast, natural, and slow cadences," *Clinical Orthopaedics and Related Research*, no. 175, pp. 147–154, 1983.

[11] A. E. Ferris, J. M. Aldridge, C. A. Raibago, and J. M. Wiken, "Evaluation of a powered ankle-foot prosthetic system during walking," *Archives of Physical Medicine and Rehabilitation*, vol. 93, no. 11, pp. 1911–1918, 2012.

[12] S. K. Au, J. Weber, and H. Herr, "Powered ankle-foot prosthesis improves walking metabolic economy," *IEEE Transactions on Robotics*, vol. 25, no. 1, pp. 51–66, 2009.

[13] H. M. Herr and A. M. Grabowski, "Bionic ankle–foot prosthesis normalizes walking gait for persons with leg amputation," *Proceedings of the Royal Society B: Biological Sciences*, vol. 279, no. 1728, pp. 457–464, 2012.

[14] T. Yu, A. R. Plummer, P. Iravani, J. Bhatti, S. Zahedi, and D. Mouser, "The design, control, and tesing of an integrated electrohydrostatic powered ankle prosthesis," *IEEE/ASME Transactions on Mechatronics*, vol. 24, no. 3, pp. 1011–1022, 2019.

[15] T. Elery, S. Rezazadeh, C. Nesler, J. Doan, H. Zhu, and R. D. Gregg, "Design and benchtop validation of a powered knee-ankle prosthesis with high-torque, low-impedance actuators," in *Proceedings of the IEEE International Conference on Robotics and Automation*. IEEE, 2018, pp. 2788–2795.

[16] X. Sun, F. Sugai, K. Okada, and M. Inaba, "Design, control and preliminary test of robotic ankle prosthesis," in *Proceedings of the IEEE/RSJ International Conference on Intelligent Systems and Robots*, October 2018, pp. 2787–2793.

[17] D. Dong, W. Ge, B. Convens, Y. Sun, T. Verstraten, and B. Vanderborght, "Design, optimization and energetic evaluation of an efficient fully powered ankle-foot prosthesis with a series elastic actuator," *IEEE Access*, vol. 8, pp. 461 491–61 503, 2020.

[18] M. Grimmer, M. Holgate, R. Holgate, A. Boehler, J. Ward, K. Hollander, T. Sugar, and A. Seyfarth, "A powered prosthetic ankle joint for walking and running," *Biomedical engineering online*, vol. 15, no. 3, pp. 37–52, 2016.

[19] J. Zhu, Q. Wang, and L. Wang, "On the design of a powered transtibial prosthetics with stiffness adaptable ankle and toe joints," *IEEE transactions on Industrial Electronics*, vol. 61, no. 9, pp. 4797–4807, 2014.

[20] B. Convens, D. Dong, R. Furnémont, T. Verstraten, P. Cherelle, D. Lefeber, and B. Vanderborght, "Modeling, design and test-bench validation of a semi-active propulsive ankle prosthesis with a clutched series elastic actuator," *IEEE Robotics and Automation Letters*, vol. 4, no. 2, pp. 1823–1830, 2019.

[21] A. H. Shultz, B. E. Lawson, and M. Goldfarb, "Variable cadence walking and ground adaptive standing with a powered ankle prosthesis," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 24, no. 4, pp. 495–505, 2015.

[22] D. F. B. Haeufle, M. D. Taylor, S. Schmitt, and H. Geyer, "A clutched parallel elastic actuator concept: towards energy efficient powered legs in prosthetics and robotics," in *The forth IEEE RAS/EMBS International Conference no Biomedical Robotics and Biomechatronics*, Rome, Italy, June 2012, pp. 1614–1619.

[23] M. Plooij, M. Wisse, and H. Vallery, "Reducing the energy consumption of robots using the bidirectional clutched parallel elastic actuator," *IEEE Transactions on Robotics*, vol. 32, no. 6, pp. 1512–1522, 2016.

[24] U. Mettin, P. X. La Hera, L. B. Freidovich, and A. S. Shiriaev, "Parallel elastic actuators as a control tool for preplanned trajectories of underactuated mechanical systems," *International Journal of Robotics Research*, vol. 29, no. 9, pp. 1186–1198, 2010.

[25] D. A. Winter, *Biomechanics and motor control of human movement*, 4th ed. John Wiley & Sons, 2009.

[26] F. Gao, Y. Liu, and W. Hsin Liao, "Implementation and testing of ankle-foot prosthesis with a new compensated controller," *IEEE/ASME Transactions on Mechatronics*, vol. 24, pp. 1775–1784, 2019.

[27] J. Realmuto, G. Klute, and S. Devasia, "Nonlinear passive cam-based spring for powered ankle prostheses," *Journal of Medical Devices*, vol. 9, no. 1, p. 011007, 2015.

[28] L. Flynn, J. Geeroms, R. Jimenez-Fabian, B. Vanderborght, N. Vitiello, and D. Lefeber, "Ankle-knee prosthesis with active ankle and energy transfer: Development of the CYBERLEGs Alpha-Prosthesis," *Robotics and Autonomous Systems*, vol. 74, pp. 4–15, 2015.