[DESIGN AND ANALYSIS OF POWERED ANKLE-FOOT PROSTHESIS]

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ABSTRACT

These days multiple types of research are going on several ankle-foot prosthetics, mostly concerned with powered ankle-foot prosthesis. The term “powered ankle-foot prosthesis" refers to a mechanical device that provides power to prosthesis during walking using different mechanisms. Every prosthesis design is made with an aspect to deliver expected torque and power while adhering to strict size and weight constraints. This paper presents the functionality of three different designs of power ankle-foot *prosthesis*: Powered polycentric ankle prosthesis, non-linear parallel spring mechanism ankle prosthesis, and ankle prosthesis with active alignment are the distinct designs that are displayed and discussed.

INTRODUCTION

Many people have lower-limb amputations as a result of the high frequency of peripheral vascular disease. People with limb loss and peripheral vascular disease are weaker and can only walk a short distance [1]. Considering these problems the research in the field of ankle- foot prosthesis begins. Ankle-foot prosthesis is made to match the lost physiological limb, to produce a substantial axial moment while walking transmitted to the residual limb via the socket joint [3].

The passive ankle prosthesis conducts energy absorption return functions by an actual ankle joint. To more accurately duplicate the biological ankle action, passive ankle prosthetics incorporate mechanisms through links and leverage coupled to dampers and springs. The range of motion (ROM) of these systems is greater than that of prosthetics without real ankle mechanics. The devices that actively adjust joint resistance using microprocessor-driven electrical motors or valves are known as quasi-passive. Since they are powered by a battery they can only switch their passive dynamic features while ambulatory. The limitation of the quasi-passive prosthesis is unable to infuse power in the gait cycle like natural ankles achieve while walking [1].

Then the power ankle prosthesis is made to provide a large extent of stiffness, speed, torque, and power to achieve gait. Power prosthesis were weighty and bulkier as compared to passive components; make them inappropriate for many amputated person. When prosthetic mass and inertia of intact limb matches it will contribute in increasing the prosthesis weight which increases metabolic energy consumption and hip effort. Additionally, the extra mass is pendent over the socket junction; which may cause instability of socket. As a result, new feasible design solutions for powered ankle prosthetics are required [1].

In this paper, three different designs of Powered ankle-foot prosthesis are discussed. The first is powered polycentric ankle prosthesis which is structured on a polycentric kinematic linkage with a completely powered actuation unit (P^2 Ankle) [1]. The second design non-linear parallel spring mechanism ankle prosthesis was created to reduce energy consumption and motor power requirements with simplifying prosthesis control [2]. The third design of an ankle prosthesis with active alignment is concerned with reducing strain on a residual limb by providing perfect alignment using a perfect robotic ankle prosthesis design. [3].

Powered Polycentric Ankle Prosthesis

Powered ankle prosthesis connects a prosthetic shank to a prosthetic ankle-foot joint, which is controlled using an actuator. These parts sustain ground reaction forces (GRF) and also generate the speed and torque required for walking. The objective of the p^2 ankle is to be a lightweight, powered ankle with an optimal polycentric mechanism that achieves a variable instantaneous center of rotation (ICR) along the prosthesis shank and foot. The ICR is directed to increase functionality across all ambulation activities by increasing electrical efficiency without the need for springs or clutches while preserving a wide range of motion and tiny dimensions [1].

Polycentric ankle prosthesis Design

A non-anthropoid polycentric Mechanism supports the outline of powered ankle/foot prostheses. The torque at the prosthesis shank was regulated using an open-loop control method based on motor voltage and polycentric mechanism. The suggested polycentric structure eliminates the necessity for a parallel dorsiflexion spring, which is prevalent in propelled ankle systems [1].

The design of polycentric foot and shank connects 3 parallel sets of links, two revolute-revolute chains (R1 R2 and R4 R3), and one revolute-prismatic-revolute chain (R5 P1 R2) with an active prismatic joint. The 4-bar chain r1 r2 r3 r4 has one single degree of freedom which is a combination of translation and a rotation motion in the (x;z) plane. In Figure 1, a red dot denotes the ICR of the shank corresponding to the foot at the junction of the paths connected with the two cranks R1 R2 and R2 R3. The DOF is varied by varying the inclination of the R1 R2 crank via the displacement of the joint P1 [1].

There is no actual ankle joint in the polycentric mechanism. The inclination of the shank in the vertical direction determines the ankle position. θ has a continuous variation concerning the inclination θb of the link R2 R3. The chain's kinetic parameters can be derived using ρ, the reduction ratio, which is a depiction of the output linkage as a function of the input linkage [1].

$$\theta = \theta_b - \theta_{offset};\ \theta_b = f_{4bar}(\theta_1) = \mathrm{atan2}(l_{34}s_4 - l_{12}s_1\ ;\ l_{14} + l_{34}c_4 - l_{12}c_1) \quad (1)$$

$$\theta_4 = \hat{\theta} + \mathrm{acos}\left(\frac{l_{23}^2 - \hat{l}^2 - l_{34}^2}{2l_{34}\hat{l}}\right) - \pi,$$
$$\text{with} \begin{cases} \hat{l} = \sqrt{l_{12}^2 + l_{14}^2 - 2l_{12}l_{14}c_1} \\ \hat{\theta} = \mathrm{atan2}(l_{12}s_1\ ;\ l_{12}c_1 - l_{14}) \end{cases} \quad (2)$$

$$\theta_1 = f_{act}(\delta) = \mathrm{acos}\left(\frac{l_{12}^2 + h^2 + v^2 - \delta^2}{2l_{12}\sqrt{h^2 + v^2}}\right) + \mathrm{atan}\left(\frac{v}{h}\right) \quad (3)$$

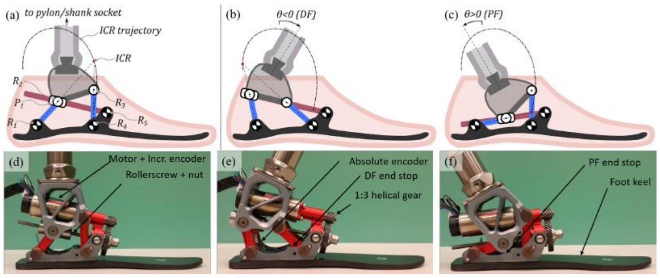

Figure 1: (a-b-c) Picture of p^2 ankle prototype with linkage mechanism. (d-e-f) Final design image in the same posture.

The GRF is balanced by force A applied by the actuator based on virtual work. The polycentric design allows the actuator's moment arm to be extended without increasing the prosthesis dimensions, as monocentric design structures require [1].

$$\begin{cases} \vec{A} + \vec{R} + \overrightarrow{B_{12}} + \overrightarrow{B_{34}} = 0 \\ A \cdot a_A = GRF \cdot a_R \Rightarrow A = \dfrac{GRF \cdot a_R}{a_A} = \rho \cdot T_a \end{cases}$$

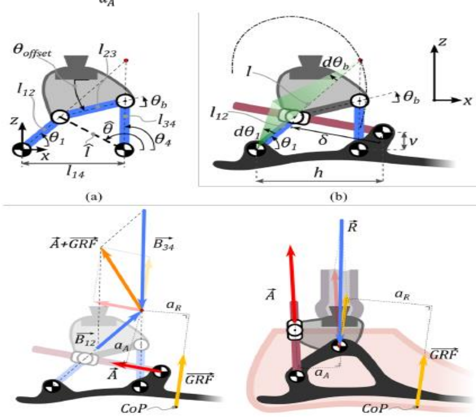

Figure 2: (a) Eqs. 1&2 4-bar linkage mechanism (b) Kinematics of the actuator chain for Eq.3. (c) The Free-body diagram of the foot. (d) Compared to a monocentric model using a linear actuator

The p2 Ankle was created using a repetitive modeling approach based on adaptive models, to enhance motor performance, increase range of motion (ROM) during the stance phase, reduce total mechanism size, and lower vertical translation movement. To determine a suitable motor, transmission, and linkages' dimensions, an electrical & mechanical model for power transmission was built by analyzing the kinematic model as mentioned in the design. A small absolute magnetic angular encoder measures the direction of the moving crank and an incremental optical encoder measures the motor's shaft position in the p2Ankle sensing system [1].

Experimental Results

Previously, polycentric designs were employed, but this is a new design technique that uses an optimal ICR track to improve the correlation among GRF prosthesis torque and motor torque to enhance electrical performance and minimize prosthesis size. We can get a large reduction ratio in a short prosthetic size without losing efficiency because of to the polycentric geometry, something we couldn't do with single-axis devices. Furthermore, the polycentric design offers the benefit of distributing mechanical stress throughout an identical structure, reducing the total size and weight [1].

A high transmission ratio is reported while performing a standing test on p2 revealed a positive impact of design. Identically, the ICR of the p2Ankle lies anterior to the anatomical ankle axis in a neutral posture (Figure 1). This position minimizes the bodyweight's moment arm, lowering the motor torque required to stand using the p2Ankle.

Similarly walking test results that p2 can make a person walk with close physiological kinematics by using less torque than required torque due to the shorter length of ICR and the COP (Figure3). According to the standing and walking test results, the p2 ankle doesn't need a parallel dorsiflexion spring to successfully transmit physical torque, which increases mobility, optimize design, and renders the system lightweight and more portable. The p2 design achieved 1.05 kg weight which is half of a similar type of prosthesis [1].

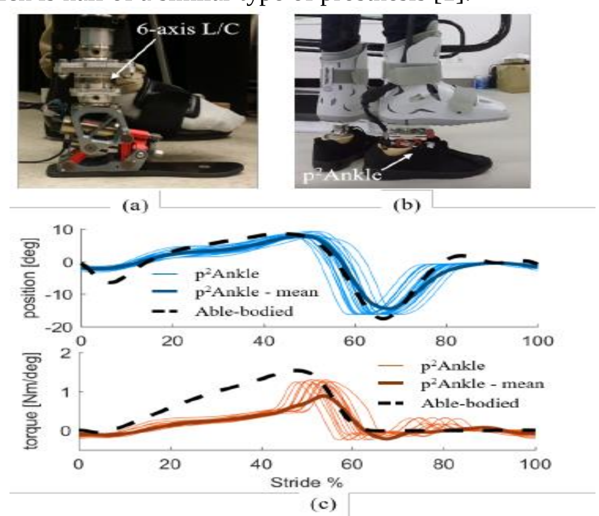

Figure 3: Testing on humans. Panels (a) and (b) Setup for standing and walking test, correspondingly. The experimental data for the walking test is shown in panel (c).

In conclusion, it is seen that standing up and moving around is fully possible for a healthy person using this device without the requirement of parallel dorsiflexion spring, a component which is otherwise essential to designs of a similar kind[1].

Non-linear parallel spring mechanism ankle prosthesis

The nonlinear parallel spring mechanism is designed to reduce energy consumption and power requirements while also simplifying prosthesis control. The design of a parallel spring mechanism is developed like a cam and spring system to replicate human ankle locomotion. This design is a modified form of its previous version which uses stored negative mechanical energy in parallel spring using cam and spring mechanism during controlled dorsiflexion (CD) phase. This design is a modified version of its previous version powered ankle prosthesis with parallel elastic actuators (PEA), where parallel elastic actuation unit is used [2].

Design of non-linear parallel spring mechanism

The idea behind this design arrived from analyzing concepts of gait and previous designs. A gait cycle is split up into two parts: the stance phase and the swing phase (SW). During the stance phase, the foot is planted on the ground to support the body and give the effort to push it forward. Three subparts of the stance phase are controlled plantar flexion/CD, controlled dorsiflexion/CD, and powered plantar flexion/PP. During the swing phase, the ankle gets reset to its desired stable position. The plot of the average ankle torque (Nm) vs. angle curve (rad) of a 70kg person is shown in Fig.4, based on data from Ref.[2] for building mechanism and concept of the parallel spring mechanism.

The rigidity of the human ankle is nonlinear during CD, and the ankle effort is negative, whereas positive work is essential in PP, and greater torque and power are required to drive a person forward than in CD. The best energy efficiency for powered ankle-foot prostheses powered by PEA can be acquired when the spring torque plot is bounded by the torque of the human ankle, as the negative mechanical energy in the CD can be saved entirely in the parallel spring and utilized in the PP. The same is illustrated in Fig. 4. This indicates that the motor must be turned on for the prosthesis to maintain its balance when standing. As a result, the majority of energy is wasted while standing, and the prosthesis' working duration is greatly decreased [2].

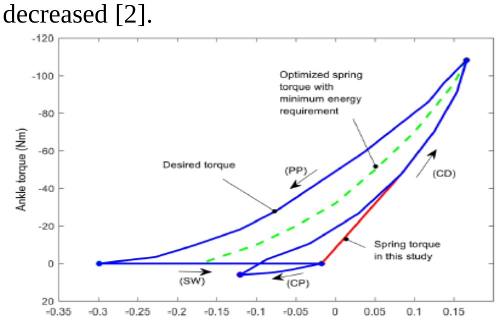

Figure 4: Design of parallel spring mechanism torque vs angle curve

The wasted energy is utilized in this design by setting the spring contact angle to zero. This proves that the parallel spring saves approx. 86 percent of the energy in the CD and uses it during the PP. There is an overall 31% and 56% decrease in the maximum motor power and torque respectively. Furthermore, no motor control is necessary throughout CD because the spring torque equals the required torque. During CD, the motor is shut off, and the parallel spring arrangement provides the appropriate torque. The motor in PP will provide the necessary torque by functioning with parallel springs.

The five major components of the design are the flexible foot, drive system unit, parallel spring mechanism, sensors, and batteries. The elastic foot is made up of carbon fiber that absorbs shock when the heel strikes. The sensors used are force sensor and angle sensor. A motor drive unit includes electric motors, a belt transmission, and a ball screw nut transmission. A cam-spring mechanism is developed and used in the system to regulate and utilize stored negative energy in a parallel spring system. As illustrated in Fig. 5, the parallel spring mechanism may imitate human ankle dorsiflexion rigidity with designing the cam profile. The parallel spring system is a cam-spring device designed to replicate the rigidity of human ankle movement**.** The cam profile is designed to achieve the desired aspect of using stored negative energy in spring [2].

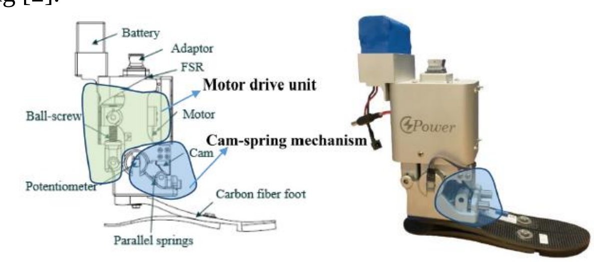

Figure 5: Design of powered ankle prosthesis driven by (PEA)

Experimental Results

Experiments reveal that with this design, the motor, which is in a previous similar type of design consumed energy continuously, is naturally switched off during the CD phase due to usage of the negative mechanical energy stored in the parallel spring mechanism for further actuation. Moreover, this results in a highly reduced working time of the motor, thus, in turn, decreasing overall power usage by significant levels. As seen on comparing the data from Fig.6 with prior designs, the motor's peak power and energy consumption during the PP phase are respectively reduced by the approximate values of 37.5% and 34.6% [2].

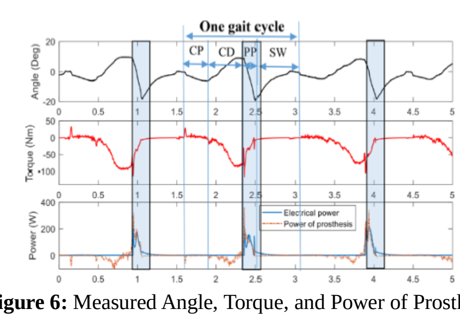

Figure 6: Measured Angle, Torque, and Power of Prosthesis

Ankle Prosthesis with Active Alignment

The huge moment is linked to high internal socket pressures while walking with an ankle prosthesis, which is frequently an origin of distress for amputation patients, restricts prosthesis use. The design of active alignment is developed with the motive to realign the damaged residual limb to the center of pressure (CoP) through active alignment without any discomfort to patients [3].

If the person who has had an amputation, the gastrocnemius and soleus muscles create most of the sagittal moment around the ankle, but they also counteract the moment transfer through bone tissue, maintaining the tibia primarily in contraction Fig.7 (a). High peak pressures are recorded on the patellar tendon and distal posterior portions of the residual limb when using a conventional prosthesis due to substantial moment transfer via the socket terminal Fig.7 (b). But the active alignment prosthesis *reallocates* the limb towards CoP to lower moment transmission and create net-positive work Fig.7 (c).

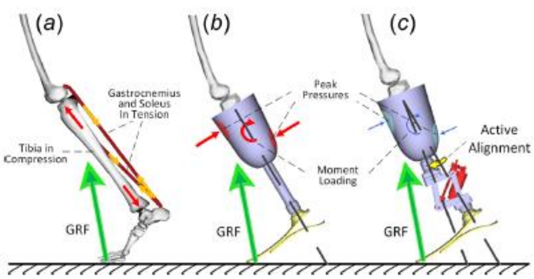

Figure 7: Gait analysis for active alignment

Active Alignment Design

The active alignment prosthesis is made out of a four-bar linkage mechanism that uses a single actuator to spin and move the foot relative to the shank shown in Fig.8. The ball screw actuator creates tensile stress among the joints that join the posterior links and the joint that connects the anterior links throughout midstance. The prosthesis extends as the actuator compresses, moving the foot center of rotation closer to the CoP. The prosthesis then swiftly returns to a stable position, allowing for ground clearance during swing [3].

Figure 8: Design of Active alignment prosthesis is displayed without coverings, with the main components highlighted (a) natural model of prosthetic (b) and full extension (c) demonstrating the altered kinematics.

Experimental Results

The past design mechanisms have achieved active alignment with multiple degrees of freedom prismatic joints in the chain but this design requires a single actuator with simple linkages. The four-bar linkage also works like a variable ratio transmission which is helpful as it permits extra force to be produced by the actuator as per requirement. For evaluating experimental results able-bodied adapters were fabricated to perform tests on the healthy person before tasting on those who had their lower limbs amputated. The results recorded during testing with extended shank able-bodied adapters show that the peak sagittal moment produced with active alignment is 33% less than with naturally aligned walking tests, as shown in Fig.9. These findings show that active alignment can be used in powered lower limb prosthetics to enhance mechanical stresses and provide a more relaxed stride for those who have had limbs amputated [3].

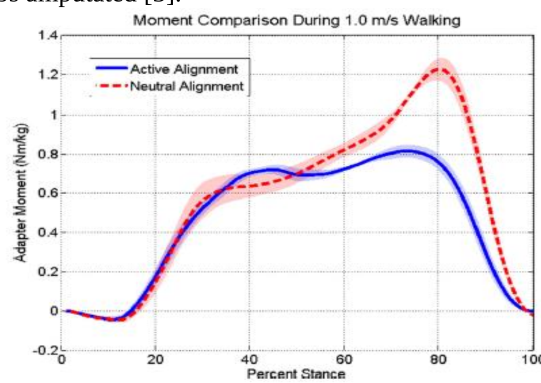

Figure 9: A comparison of moment data recorded while walking at 1.0m/s on a treadmill is shown while regulating at a neutral alignment and actively aligning is shown. The shaded areas represent standard deviations, while the lines represent average values for ten stages. According to the statistics, a peak moment was reduced by 33% during active alignment.

Conclusion

In sum, the three designs of the powered ankle-foot prosthesis are made to solve three different issues encountered in the field of ankle prosthesis. After analyzing three different design and their testing results it briefs that still there is some more work to be done on each. Although, all of the designs are successfully able to excel in their corresponding fields of expertise and a single ankle design that displays the positives of all 3 while possible is no easy task to fulfill.

REFERENCES

[1] M. Cempini, L. J. Hargrove and T. Lenzi, "Design, development, and bench-top testing of a powered polycentric ankle prosthesis," 2017 IEEE/RSJ International Conference on Intelligent Robots and Systems (IROS), 2017, pp. 1064-1069, doi: 10.1109/IROS.2017.8202276.

[2] Fei, Gao & Liu, Yannan & Liao, Wei-Hsin. (2018). Design of Powered Ankle-Foot Prosthesis With Nonlinear Parallel Spring Mechanism. Journal of Mechanical Design. 140. 10.1115/1.4039385.

[3] Kennedy LaPrè, A., Umberger, B. R., and Sup, F. C., IV (May 12, 2016). "A Robotic Ankle–Foot Prosthesis With Active

Alignment." ASME. *J. Med. Devices*. June 2016; 10(2): 025001. https://doi.org/10.1115/1.4032866