Engineering Solid Mechanics 11 (2023) 339-352

Contents lists available at GrowingScience

Engineering Solid Mechanics

homepage: www.GrowingScience.com/esm

Design of a novel above-knee prosthetic leg with a passive energy-saving mechanism

Amer Imran[a], Mohammad Reza Haghjoo[b] and Borhan Beigzadeh[a*]

[a]*Biomechatronics and Cognitive Engineering Research Lab, School of Mechanical Engineering, Iran University of Science and Technology, Tehran, Iran*
[b]*Faculty of Mechanical and Energy Engineering, Shahid Beheshti University (SBU), Tehran, Iran*

A R T I C L E I N F O

Article history:
Received 6 April 2023
Accepted 27 May 2023
Available online
27 May 2023

Keywords:
Above-knee amputees
Prosthetic Leg
Energy saving
Elastic energy storage and return
Walking
Passive mechanisms

A B S T R A C T

The push-off phase is a critical part of initiating movement during walking, and it requires a significant amount of energy. Recent research has shown that the passive use of springs in parallel with the leg can harvest the push-off energy and reduce the total metabolic energy of walking for healthy subjects. In this study, we present the design of a prosthetic leg with a passive-based mechanism to reduce walking energy consumption for above-knee amputees. The mechanism stores energy during the stance phase of the gait cycle and releases it to support the prosthetic leg during locomotion. The known polycentric knee joint 3R36 and the ankle-foot joint ESAR were chosen and adopted for this study. We also utilized a ratchet clutch that connects with a spring and rope from the pylon to the foot which regulates movement and saves energy. Our simulations demonstrate that the spring stores elastic energy from approximately 22% of the gait cycle and reaches its maximum energy storage at approximately 50% of the walking cycle. The energy is then released at approximately 58% of the stride cycle during the push-off phase. The motion of the proposed prosthetic leg for individuals with transfemoral amputations mimics the normal walking pattern of healthy individuals well.

1. Introduction

Walking is the only way to get locomotion in the harsh topography of the land because of the difficulty of transportation by wheeled vehicles. The legs are basic and necessary parts of the body that move and travel the body from one place to another in all our routine life. The activities of daily living are being greatly expanded by perpetual walking and minimal energy consumption. Therefore, improving gait is of economic benefit to society and individuals. In spite of the fact that the human walking manner is naturally well-tuned (Collins et al., 2015). Nevertheless, it has space for upgrading principally when devices and mechanisms of the exoskeleton are improved for serving human locomotion.

As it's known, the ankle produces half of the demanded amount for mechanical energy during push-off (Farris & Sawicki, 2012 b), therefore, some researchers took up the study of ankle-foot exoskeleton and mechanisms during the last two decades. Various proposed devices have been designed, which are powered and unpowered, to help the walkers in the push-off phase to reduce energy consumption (Sawicki & Ferris, 2008).

Despite the ankle-foot orthoses having the potential for assisting during gait, these devices overwhelmingly are used for patients who are suffering from leg diseases for augmentation of their mobility capability (Blaya & Herr, 2004). Ankle-foot exoskeleton increases ankle movement efficiency by wearers, dissimilar to ankle-foot orthoses (Dollar & Herr, 2008).

* Corresponding author. Tel.: +98 21 7724 0094
E-mail addresses: b_beigzadeh@iust.ac.ir (B. Beigzadeh)

ISSN 2291-8752 (Online) - ISSN 2291-8744 (Print)

doi: 10.5267/j.esm.2023.5.009

Ankle-foot exoskeleton devices are classified into two groups depending on the actuation method: passive and active ankle-foot exoskeleton. All devices of active ankle-foot exoskeleton used outer energy sources for supplying bioequivalent levels and decreasing the demanded energy from muscles during walking (Malcolm et al., 2013). Some of the smart mechanisms are modified to transform the power of the motor into a torque about the ankle-foot joint. An autonomous exoskeleton was designed in (Mooney et al., 2014) and (Mooney & Herr, 2016), which consists of struts arranged in a triangular frame. It was actuated by a winch. Also, it obtains the torque for the ankle-foot joint by the pulling force. Some researchers used pneumatic muscles that have a distinct advantage over electric motors due to their lightweight nature and structural resemblance to a muscle-tendon unit. This makes them highly suitable for use in the design of active devices, including exoskeletons created for elderly individuals by different researchers such as (Ferris et al., 2005, 2006; Kao et al., 2010; Sawicki et al., 2005;Galle et al., 2014, 2017).

Although the design of active devices has the potential to provide the necessary energy required, they do have some drawbacks. One of the significant concerns is that their force profiles and onset timing are dependent on the individual's gait pattern. Thus, it is crucial to detect gait stages accurately, which requires the use of electronic sensors to identify all types of gait events. Furthermore, the exoskeleton must be compliant enough to accommodate the full range of motion of the ankle joint during the swing phase without hindering normal human movement. Not being able to move the foot upwards (dorsiflex) without hindrance can cause a significant metabolic cost (Wiggin et al., 2011). The distal mass presents another challenge. Circuit-based sensor-control systems and power sources can add extra weight to the system, making it less comfortable to wear. Additionally, the circuit may be damaged by shocks while walking. As a result of these concerns, passive exoskeletons have been developed as a substitute.

In contrast to powered exoskeletons, some researchers have recently focused on studying passive exoskeletons because they do not require a power source and use a control system based on circuits. To enhance walking efficiency, elastic elements such as springs and elastic cables are typically utilized to store energy from the human body and release it to help users move faster (Grabowski & Herr, 2009). Farris and colleagues (Farris & Sawicki, 2012a) created an exoskeleton that included a suspended spring running parallel to the calf muscles. However, because there were no means of identifying the gait pattern, the resting length of the spring had to be customized for each individual. Smart clutches (Collins et al., 2015; Diller et al., 2018; Diller et al., 2016; Yandell et al., 2019) have been developed to mechanically control the engagement of elastic components and provide assistance at the appropriate time in order to detect the gait cycle without the use of electronic sensors. Collins and colleagues (Collins et al., 2015) developed a clutch that utilizes the ratchet-pawl mechanism to control the spring linkage by adjusting the timing of the pawl latch and release.

Transfemoral leg amputation is one of the most frequently occurring types of amputation in the human population (Sarvestani & Azam, 2013). This condition disrupts the natural coordination of the legs and considerably impairs the mobility of amputees (Gailey et al., 2008). They may experience a less efficient, less stable, slower gait that requires more effort and energy to complete it (Schmalz et al., 2002; Lamoth et al., 2010). In cases of an above-knee amputation, the biological ankle and knee joints are typically replaced with prosthetic joints that cannot perform all of the functions of the missing leg joints (Goldfarb, 2013). The limited functional mobility provided by currently available prosthetic limbs significantly impacts the quality of life for millions of individuals worldwide (Moxey et al., 2011). Any issues that arise with the musculoskeletal system or its coordination pattern can increase the metabolic rate (Westerterp, 2013). To ensure the satisfaction of amputees, it is crucial to consider not only the development of new active prostheses but also the use and improvement of well-established passive prostheses, which can offer a viable solution to these limitations. The availability and affordability of passive devices are also compelling reasons to continue their use and advancement (Major & Fey, 2018).

As a solution, a passive mechanism is required for the transfemoral prosthetic leg to meet the basic energy consumption requirements during walking for above-knee amputees. The ideas of clutches developed with spring in exoskeletons can be used and incorporated in the field of prosthetic legs; because amputees who have had an amputation above the knee rely on their remaining leg, back, and hip muscles to walk, which requires a considerable amount of metabolic energy.

We introduced a novel prosthetic leg design incorporating a passive-based mechanism aimed at reducing energy consumption for above-knee amputees. The mechanism effectively stores energy during the stance phase of the gait cycle and subsequently releases it to provide support during locomotion. In this study, we opted for the well-known polycentric knee joint 3R36 and the ankle-foot joint ESAR, which were carefully selected for their compatibility. To optimize energy usage, we incorporated a ratchet clutch that connects a spring and rope from the pylon to the foot. Our design offers two distinct modes: one for energy storage during the stance phase and another for unrestricted rotation during the swing phase. An exceptional characteristic of our design is its reliance on purely passive components, eliminating the need for motors or electronic control. This not only removes the necessity for an external power source but also ensures a lightweight and easily portable prosthetic. Furthermore, this design accurately replicates the inherent mechanism of elastic energy storage and subsequent release that takes place during the normal walking motion of the human body.

Thus, this paper is organized into the following sections: It deals with the energy transfer systems of transfemoral amputees, and the conceptual design of polycentric knee, spring-loaded five-bar foot, and energy-saving mechanism. Then, the detailed design of the new transfemoral prosthesis mechanism, including its components and features, is described. This encloses a discussion of the ratchet clutch mechanism. Following that, the results of the containing energy storage and return, and typical motion patterns.

2. Energy systems for Prosthesis

The study of the walking energy system of transfemoral amputees is important in order to harvest the most efficient energy during motion and to continue it. The mechanical system for steady-state walking comprises energy inputs, outputs, storage, and transfers within the system. In all instances, any chemical or electrical energy input is ultimately released as heat because the average mechanical energy of the system remains constant and no useful work is carried out on the body or the surroundings. As per the strict definition, energy efficiency is zero in every case since all the input energy is eventually released as heat. Thus, the measure of energy effectiveness or energy economy is expressed in terms of 'cost of transport', which refers to the amount of energy utilized per unit weight for each unit of distance traveled.

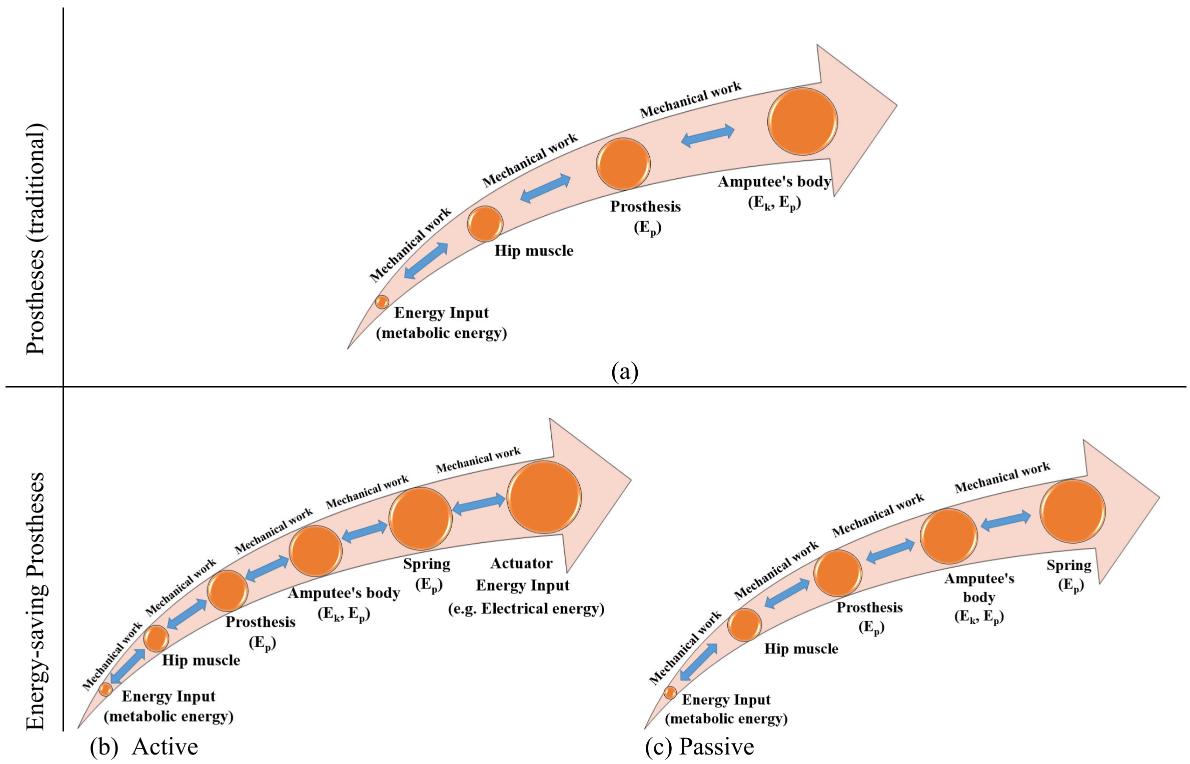


Fig. 1. Energy system graph for transfemoral amputees during walking. (a) Prostheses (traditional), (b) Active energy-saving prostheses, (c) Passive energy-saving prostheses.

2.1 Prostheses (traditional)

The metabolic energy used by hip muscles serves multiple purposes, including creating and absorbing mechanical work and carrying out other tasks such as generating force or activating the muscles. Therefore, all of the metabolic energy is directed toward the system. When energy is lost in muscles, it is expressed as heat. The prosthesis in the mechanical system swaps energy to both the muscles and the body. Energy exchange also takes place between the different segments of the body (including the prosthetic leg) through kinetic and gravitational potential energy, and this process is highly efficient in terms of mechanics. The dissipation of mechanical energy in body segments occurs primarily through damping during collisions, which is a minor factor. Additionally, there may be some energy loss due to friction caused by the prosthetic feet slipping on the ground, deformation of the ground, or air resistance, but these factors are insignificant in normal situations. Fig. 1 (a) shows the energy system for transfemoral amputees walking.

2.2 Active Energy-saving Prostheses

Another source of energy (active energy) can be introduced in the form of electricity, for instance. Although the human may bear a smaller proportion of the total energy input (and eventual dissipation) of the system, the overall energy economy may suffer if the total energy input increases. This has been observed with powered devices that aim to decrease the metabolic energy expenditure of individuals with transfemoral amputations. Although a powered device has the potential to improve the overall energy economy, this has not always been the case in practice. Two potential ways in which a powered device could improve the overall energy economy have been identified in theory. The first method involves the powered device functioning similarly to an unpowered device, with minimal electricity used only to regulate the timing of mechanical components like clutches. The second method involves the motor doing the work with higher efficiency, which could replace positive mechanical work from muscles. Fig. 1 (b) shows the energy system for transfemoral amputees walking with a powered actuator (active energy).

2.3 Passive Energy-saving Prostheses

In contrast to the powered case, there is no extra energy supply available. Therefore, the only approach to reducing metabolic energy consumption is by reducing the overall energy dissipation of the system, or alternatively, by enhancing the overall energy efficiency of the system. It is important to note that the only change in energy flow for individuals with transfemoral amputations is the inclusion of components such as springs that store and transfer mechanical energy (passive energy) within the system. Therefore, decreasing the metabolic rate using passive parts is similar to altering the person's morphology in a way that makes locomotion more energy-efficient. Fig. 1 (c) shows the energy system for transfemoral amputees walking with an unpowered actuator (passive energy). The idea of our research depends on this energy system which is suitable for completing all research aims.

3. Conceptual Design

This mechanism has been created to incubate the prosthetic leg in a manner that allows energy to be stored and released during walking. The study concentrated on famous prostheses to enhance the fundamental movements of amputees. Below, the fundamental principles of the design and the diverse functional capabilities of different components of the prosthetic leg are illustrated.

3.1 Polycentric knee

The knee joint is the most crucial component of the prosthetic leg when an amputation occurs above the knee (Soriano et al., 2020). As a result, the prosthetic knee became essential for conducting a study and evaluating its function and performance primarily (Mohanty et al. 2020; Phanphet et al. 2017). The functioning mechanism of the prosthetic knee is analyzed and assessed to establish a connection with the suggested prosthetic leg and combine it together with the new energy-saving mechanism.

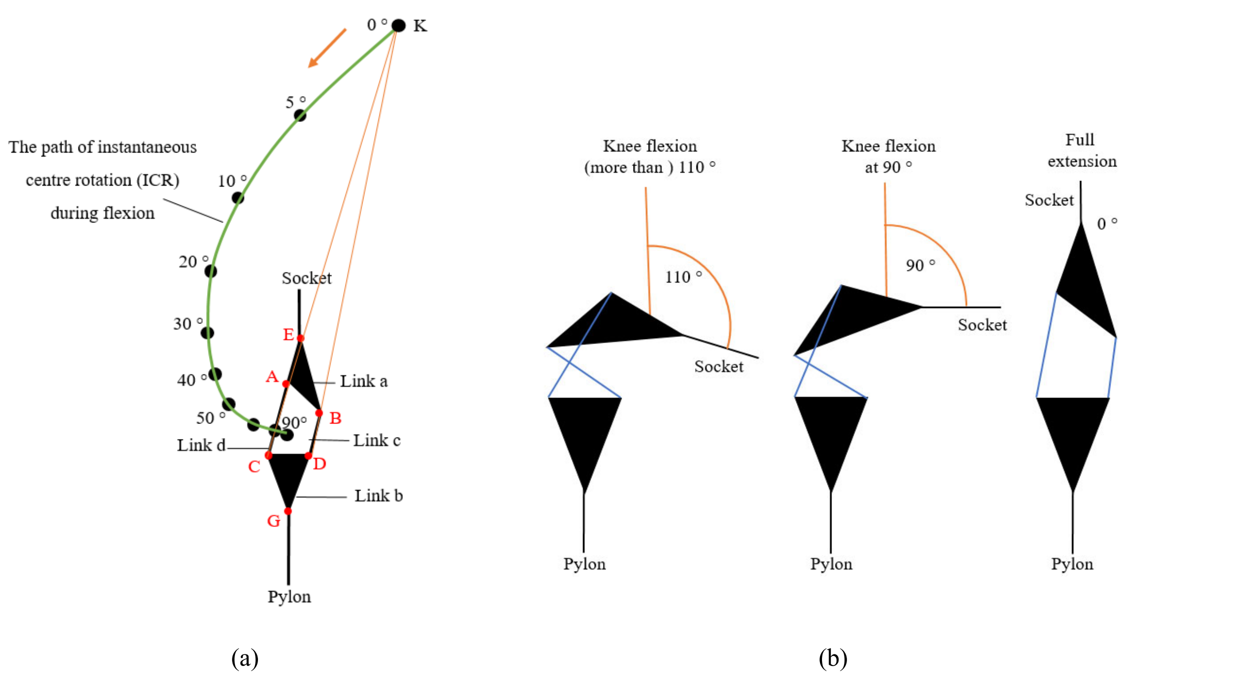


Fig. 2. (a) The bending synthesis and instantaneous knee center of the knee, (b) The flexion angles of the polycentric knee.

There are three types of four-bar linkage mechanisms that are utilized in lower limb prosthesis, as explained in reference (El-Sayed, Hamzaid, & Abu Osman, 2014): (1) the four-bar linkage with elevated ICR, (2) the four-bar linkage with hyper-stabilization, and (3) the four-bar linkage with voluntary control. For individuals who have limited ability to control stability or for geriatric amputees, it is recommended to use the four-bar linkage with elevated ICR, as depicted in Fig. 2 (a) and 2 (b).

This mechanism has a long anterior link and a short posterior link, which have been designed to provide maximum stability during heel contact. The ICR is situated behind the load line, which allows for full extension during heel contact without requiring the amputee to exert a hip extension moment. During push-off, the amputee generates a hip flexion moment and can move the load line behind the ICR. The initiation of knee flexion requires minimal effort.

3.2 Five-bar foot with Springs

Prosthetic feet are designed to restore the daily activities and functional capabilities of lower-limb amputees to their pre-amputation levels (Ventura et al., 2011). The mobility of the lower-limb amputee is a crucial indicator of the quality of the prosthetic foot (Shepherd et al., 2018; Wurdeman et al., 2018).

The prosthetic foot, which stores energy and returns it while walking, behaves like a Spring-loaded Five-bar foot, as shown in Fig. 3. Based on conceptual analysis, it consists of five links (a, b, c, d, and e) and two compressed springs (1 and 2). In the heel strike, spring 1 is in a state of compression (100%), while spring 2 is uncompressed, as shown in Fig. 3 (a). Half of the compressed energy is transmitted from Spring 1 to Spring 2, and thus both springs are compressed (50%) in the mid-stance, as described in Fig. 3 (b). In the case of heel-off, the rest of the compressed energy is transferred to Spring 2 (compressed 100%) and the energy starts to release, as explained in Fig. 3 (c). Both springs 1 and 2 are uncompressed in the swing phase, as described in Fig. 3 (d).

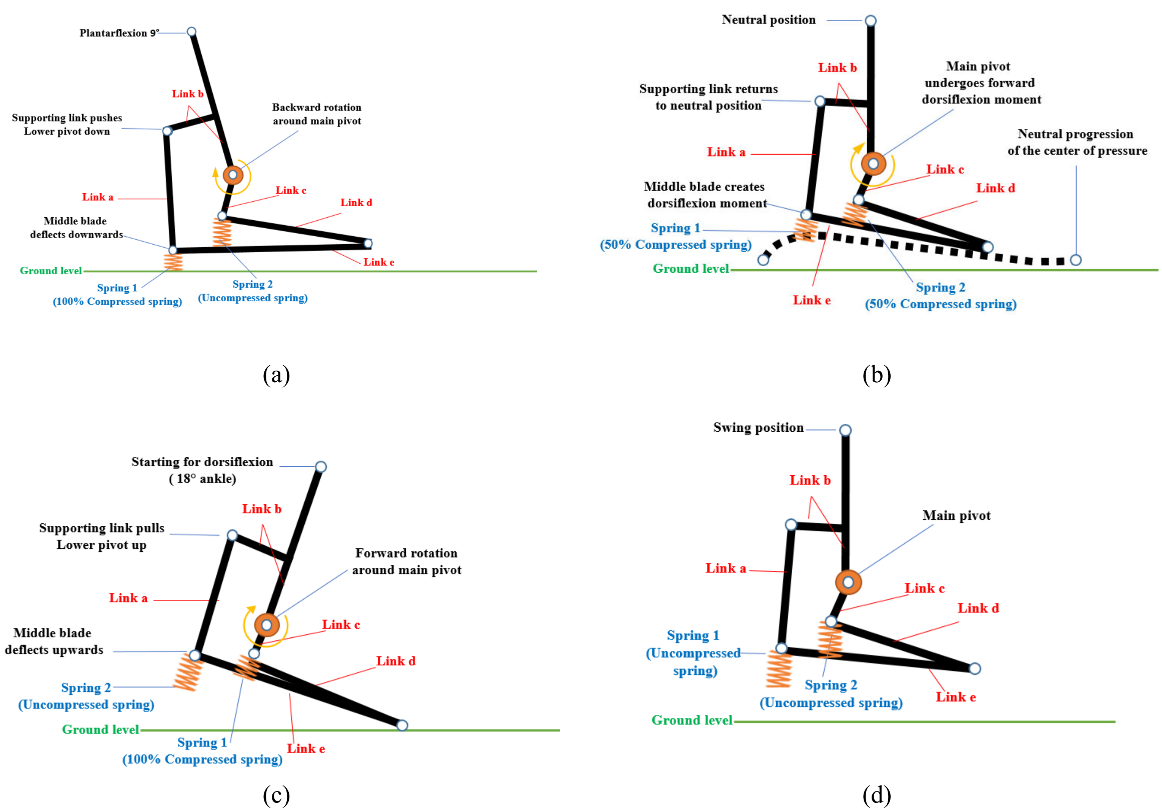


Fig. 3. The schematic of the Spring-loaded Five-bar foot during some phases of the gait cycle. (a) Heel strike, (b) Mid stance, (c) Heel off, (d) Swing position.

3.3 Energy-saving mechanism (pylon spring & ratchet clutch mechanism)

The energy-saving mechanism is composed of a schematic structure where the polycentric knee is connected to the spring-loaded five-bar foot through the pylon. Additionally, the clutch is connected to the pylon, and a rope (Kevlar strands) extends from the clutch to the spring, which is then connected to the foot. Fig. 4 displays the schematic of the energy-saving mechanism for the transfemoral prosthetic leg.

The purpose of the schematic is to understand the way energy is stored and released during the standing phases and also the movement of the parts successively within the gait walking cycle. This helps us to conclude and visualize the mechanism in a clear and uncomplicated way and to realize any overlap between the parts of the prosthetic leg. It is a recognized fact that the onset of movement occurs when energy is transferred from the stump to the various parts of the prosthetic leg. As a result, the body moves forward within the scope of one step.

It is well-established that individuals who have undergone an above-knee amputation depend on a healthy back and the energy provided by a healthy leg. Therefore, it requires a huge amount of energy to maintain their mobility. The energy supply principle involves selecting a foot that stores energy in one stage and releases it during another stage. Furthermore, a concept has been introduced in which a tension spring and clutch are utilized during the standing stage of the walking gait cycle.

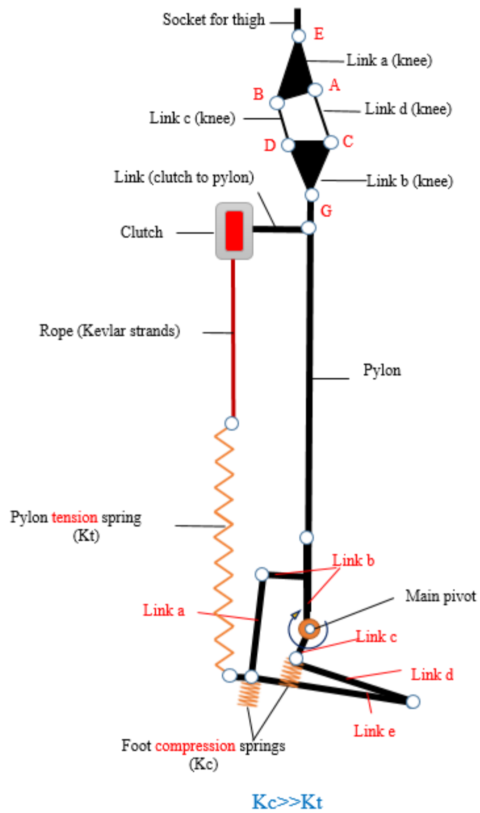


Fig. 4. The schematic of the energy-saving mechanism for the transfemoral prosthetic leg

The design of a ratchet-clutch mechanism with the spring is to the natural leg (including the natural foot) to reduce metabolic energy. As well known, the spring-loaded five-bar foot behaves like the natural foot. Energy storage inside the compressed spring (in the spring-loaded five-bar foot) starts from the heel strike until the heel off. On the other hand, the energy saving inside the tension spring begins after flat foot until before toe-off, as shown in Fig. 5. We note that the value of Kc is greater than the value of Kt. Therefore, every spring is working alone without any interface or obstacle between them. Finally, the tension spring adds extra energy to the prosthetic leg which reduces the energy of muscle during walking.

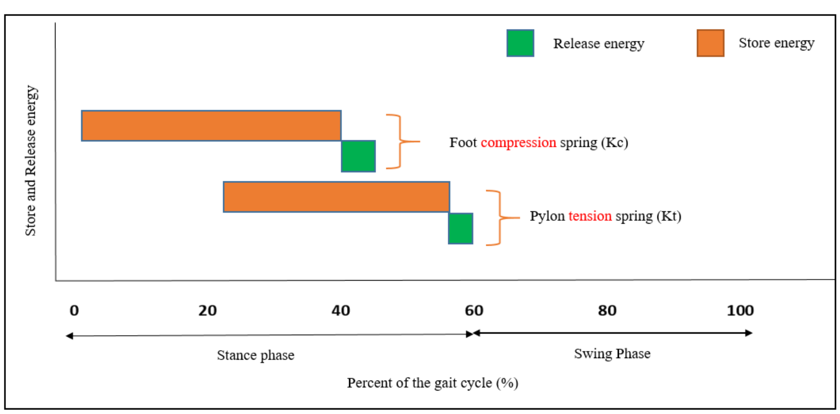


Fig. 5. Store and release energy for the pylon tension spring and foot compression spring during the gait cycle.

4. Detailed Design

It is created with the intention of offering various functional enhancements. The enhancements aimed to decrease the energy required for walking and replicating the behavior of a natural leg. The design for each part of the prosthetic leg is detailed below and the mechanism for attaching them together.

4.1 Knee assembly

For the current research, the commonly used prosthetic knee joint is referred to as 3R36. As well known, the prosthetic knee (3R36) is a famous knee, a passive knee, and a polycentric knee; these reasons made it very convenient for choosing it in the mechanism design of the prosthetic leg. Also, some transfemoral amputees can replace knee 3R36 with what suits them from passive prosthetic knees.

The assembled 3R36 comprises two sets of components: lower and upper, as shown in Fig. 6 (a) and 6 (b). These parts are interconnected using linkage bars (anterior and posterior). The posterior linkage bars are of lesser length than the anterior linkage bars. The linkage extremities of the bars are mainly connected by specifically designed internal screws and pins. These connections can provide greater mobility while in motion. A distinct component of the upper part of the 3R36 knee joint connects the entire joint with the socket, whereas another component of the lower part of the 3R36 knee joint connects the entire assembly with the pylon.

As per the identifications provided by the Ottobock Company for the 3R36 knee joint, the greatest flexion angle of the knee is approximately 110 degrees, which is selected and demonstrated in Fig. 6 (c). The spring present within the 3R36 knee joint manages and regulates the swing phase, while the kinematic analysis of the Instantaneous Center of Rotation (ICR) plays a fundamental role in maintaining the stability of the stance phase (Al-Maliky & Chiad, 2021).

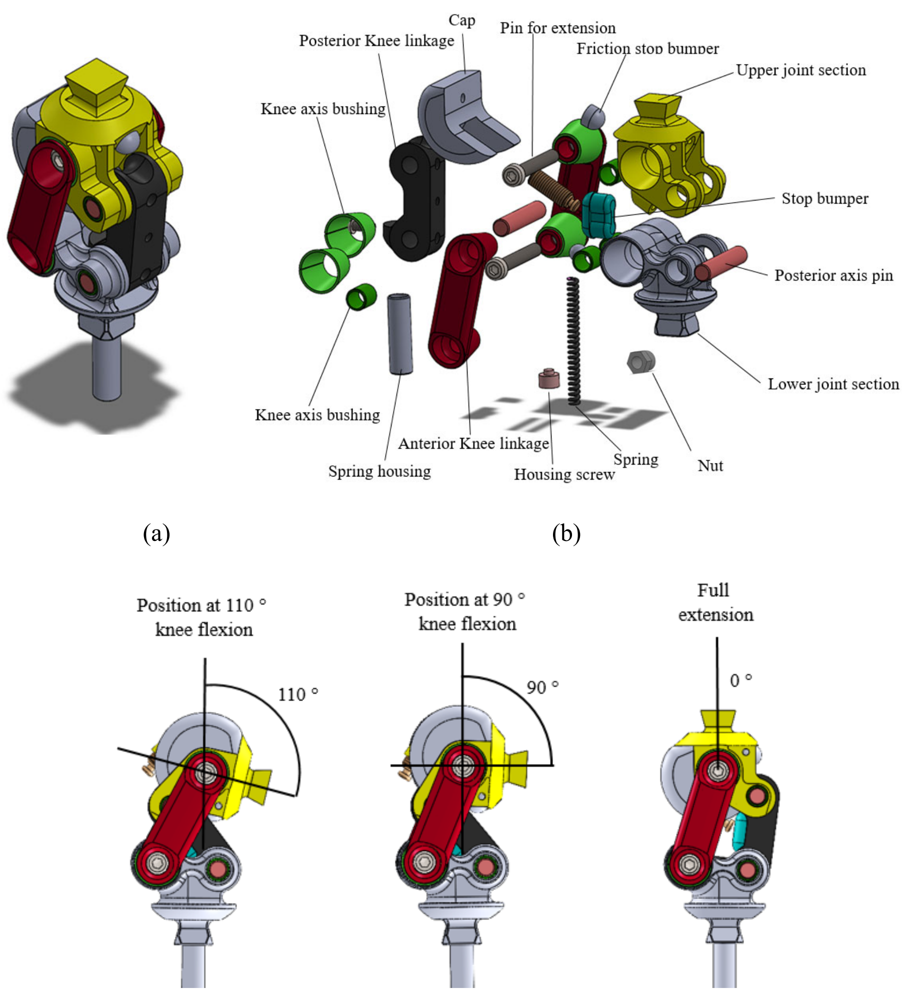


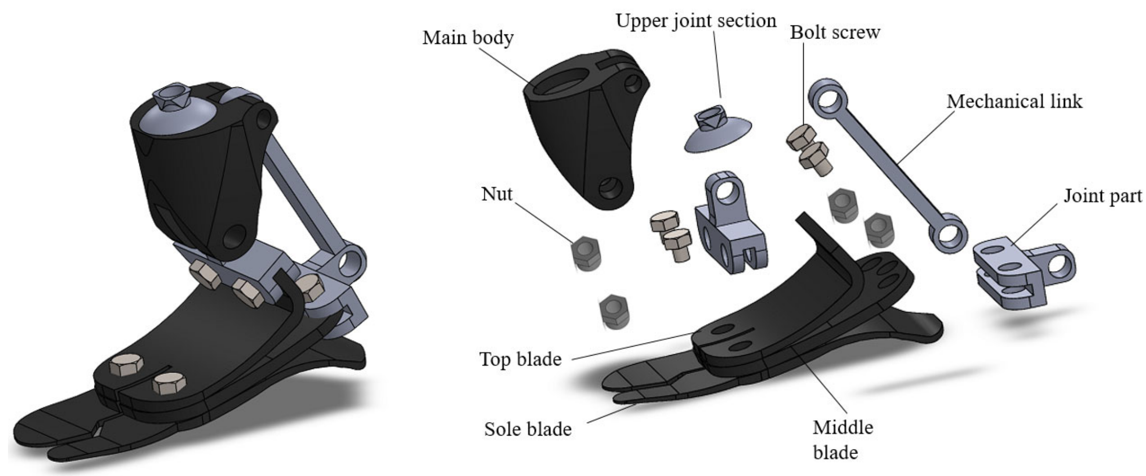
Fig. 6. The polycentric prosthetic knee joint assembly (3R36). (a) Whole view, (b) Explosion view, (c) The bending angles of the 3R36 knee.

4.2 Ankle and Foot Assembly

The mechanical function of prosthetic feet significantly affects important aspects of mobility, such as walking energy expenditure and stability (Major et al., 2014, 2016). The ESAR (Energy Storing and Return) prosthetic ankle-foot is characterized as the contemporary and renowned foot design. The ESAR prosthetic foot has been chosen for the prosthetic leg mechanism design, for the current study, due to its shock-absorbing capabilities, high plantarflexion and dorsiflexion range, ability to store and return energy during gait, widespread usage, and passive nature. The ESAR prosthetic foot recommended for lower-limb amputees emulates the spring-like behavior of the natural ankle-foot unit during walking (the gait cycle) (Shepherd et al., 2018).

Based on the specifications of the Pro-Flex® prosthetic foot from Össur, the ESAR foot functions as a specific spring system that stores energy during the mid-stance phase of the gait cycle and then releases it to propel the ESAR foot during a late stance (Hansen & Starker, 2018). During the stance phase, each blade of the ESAR foot undergoes compression and then is released. Specifically, at the start of the stance phase, the sole and middle blades of the foot store energy, which is then transferred to the middle and top blades at the delayed stance or loaded toe. Finally, all the stored energy is utilized to propel the foot into the second stage of the gait cycle, known as the swing phase (Tryggvason et al., 2020).

The ESAR foot assembly consists of two sets of components: flexible and rigid. The flexible components are made up of three leaf springs, which include the top, middle, and sole blades, as shown in Fig. 7. The remaining components of the ESAR prosthetic foot are entirely inflexible. Some parts are connected by bolt screws. The foot is connected from above to the pylon, which in turn is linked to the rest parts of the prosthetic leg. Mostly, the ESAR foot covers by silicone foot cover.



(a) (b)

Fig. 7. The foot and ankle joint assembly (ESAR). (a) Whole view, (b) Explosion view.

4.3 Pylon compliant and Ratchet clutch Assembly

Our design's key feature is a clutch mechanism that uses the concept of regulated energy storage and release. This mechanical control system is capable of storing and releasing the spring's energy at precise intervals during the stride to ensure optimal performance. We have designed a smart clutch that differs in the internal parts from the traditional clamp of Dr. Collins (Collins et al., 2015) but is similar in principle to suit the requirements of the prosthetic leg. The design involved using springs, pins, and motion restrictions to regulate the locking and unlocking of a ratchet and pawl mechanism. This enabled us to engage or disengage the parallel springs at different points during the act of walking. The intelligent clutch has benefits because it relies on the spring linkage's linear motion, which is controlled by changes in the angle of the prosthetic ankle joint, instead of using electro-mechanical switches to determine when the pawl should be locked or released.

The passive mechanism of all parts involved placing the clutch at the rear of the pylon for the prosthetic leg. The Kevlar strands of the clutch are firmly linked to the linear tension spring, which is connected to the heel section of the ESAR foot.

(a)

(b)

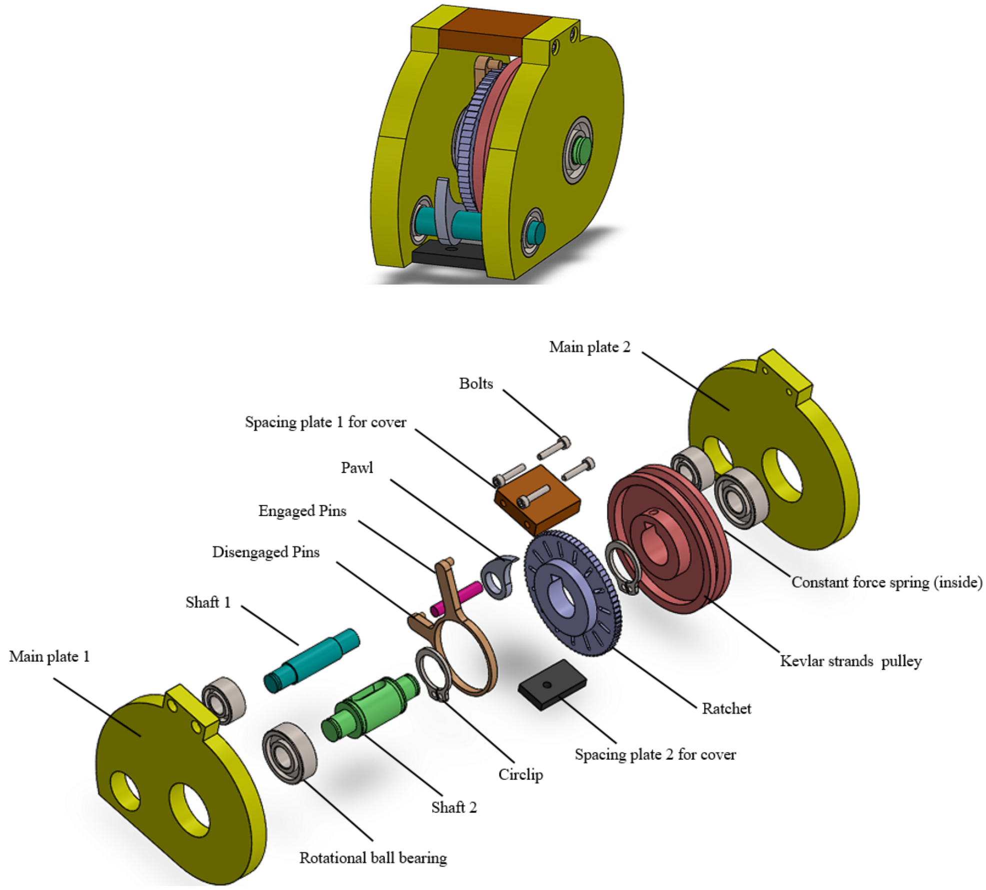
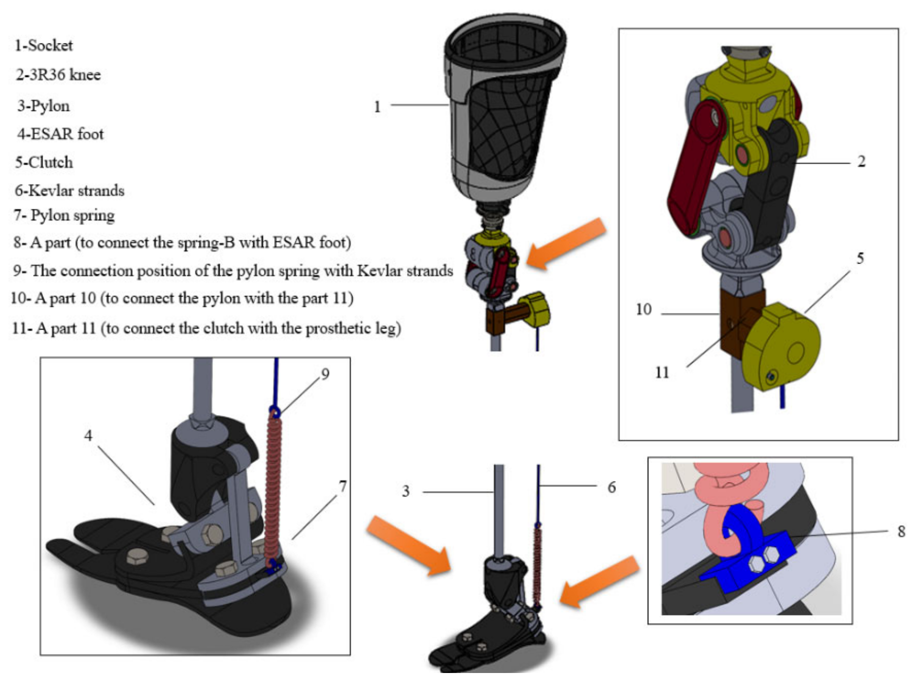


Fig. 8. The smart-clutch assembly. (a) Whole view, (b) Explosion view.



(a)

(b)

Fig. 9. The 3D schematic of the suggested mechanism for the prosthetic leg. (a) The prosthetic leg parts, (b) Whole prosthetic leg view.

In Fig. 8, which shows an exploded view of the 'Smart-Clutch' components, it is important to observe specific features such as the ratchet, timing pins (engaged and disengaged pins), and pawl. The movement of the ankle joint is transferred to the rotation of the ratchet via the Kevlar and linear spring linkage during the stride. The timing pins are positioned to start and stop the engagement of the pawl at critical points throughout the process.

The mechanism comprises primary components that join with the upper prosthetic leg to the ESAR foot, both in the front and back. Fig. 9 (a) shows the principal parts of the above-knee prosthetic leg: Socket (1), 3R36 knee (2), pylon (3), and ESAR foot (4); also, the added parts for linking the mechanics: Clutch (5), Kevlar strands (6), and pylon spring (7). Fig. 9 (b) displays the 3D schematic of the suggested mechanism for the whole prosthetic leg.

The connection method of the mechanism is to store the energy during the stance phase, then release it during the swing phase. We connected the ESAR prosthetic ankle-foot joint with the smart-clutch and linear tension spring (pylon spring (7)). The connection method begins by attaching the clutch (5) to the upper back of the pylon (3) using the screws. Then, the rope (Kevlar strands) (6) comes out of the clutch and is attached to the pylon spring (7). Finally, the pylon spring connects to the back side of the ESAR foot by a ring. The spring is positioned such that it generates torque comparable to that produced during human walking.

The specifications of the socket type and the pylon type are not addressed, because these parts are considered bonding parts for the purpose of completing the prosthetic leg.

5. Results & Discussions

The proposed prosthetic leg mechanism is designed to provide multiple functional improvements, including reducing the energy required for walking, preventing the prosthetic foot from scraping while walking, and mimicking the behavior of a natural leg. Also, the design of the mechanism has a way that is suitable for providing stability during the stance phase by assisting with the control of hip muscles. Additionally, it is well-suited for facilitating smooth bending during the push-off phase.

The present design utilizes mechanical input to activate or deactivate a parallel spring with elastic properties. The graph Fig. 10 (a) shows the results of the stance phase in a single stride, from the moment the heel strikes the ground (0%) to the push-off phase (60%).The rotation of the center of pressure around the ankle joint causes the spring to store elastic energy between (22%) into (58%) into the stride cycle. The maximum value of potential energy is (50%) from the stride cycle. During the push-off phase (58%) into the stride cycle, the stored elastic energy in the spring is released rapidly and it contributes to all of the positive work that propels the ankle.

The clutch is initiated by permitting downward motion of the linkage (which includes Kevlar strands and a linear spring) until the moment of heel strike, as shown in Fig. 10 (b 1). The pin that keeps track of timing activates a mechanism made up of a ratchet and a pawl, which prevents any further downward movement of the linkage.

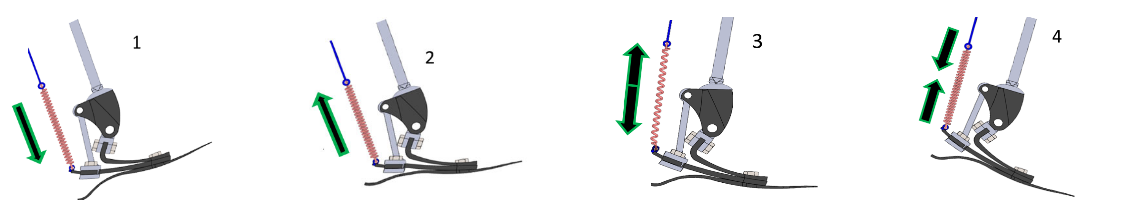
After the ratchet and pawl mechanism restricts further downward movement of the linkage at maximum plantar flexion; a constant tension spring in the clutch generates a reaction force to take up slack in the system while the foot plantar flexes until reaching the foot-flat position, as shown in Fig. 10 (b 2). After foot-flat position, when the ankle starts dorsiflexing towards mid stance, the clutch becomes locked and the linkage transfers force to stretch the linear spring, accumulating energy from the body's center-of-mass Fig. 10 (b 3). This energy is then released during push-off, producing positive mechanical work and moving the body forward by providing the necessary force to achieve propulsion Fig. 10 (b 4). After the positive mechanical work is performed, a second timing pin is activated at maximum plantar flexion, which then disengages the ratchet and pawl mechanism. This allows the ankle to dorsiflex during the swing phase, thus resetting the cycle.

This design refers to the energy that can be stored in the spring due to its deformation, which is determined by the amount of stretching and the spring constant (represented by 'k'). In order to store energy in the proposed modular transfemoral prosthesis, the amount of stretching in the spring is measured at different points in the stride cycle, while the spring constant ('k') remains constant, as shown in Fig. 10 (c). Subsequently, the spring exerts a force that is equal in magnitude and opposite in direction, which propels the foot toward the next step. During the swing phase, the leg moves quickly, which enables amputees to reduce the energy needed for walking and lessen the amount of muscle effort required.

Fig. 11 (a) explains the typical motion of the suggested prosthetic leg for individuals with transfemoral amputations throughout a single walking gait cycle. The typical motion variations, during thigh rotation at different stages of the walking cycle, show the prosthetic leg positions (from the hip down to the ESAR foot) and the points of contact between the prosthetic foot and the ground. The analysis of the mechanism's behavior illustrates its ability to enhance stability during the stance phase by assisting in the control of hip muscles.

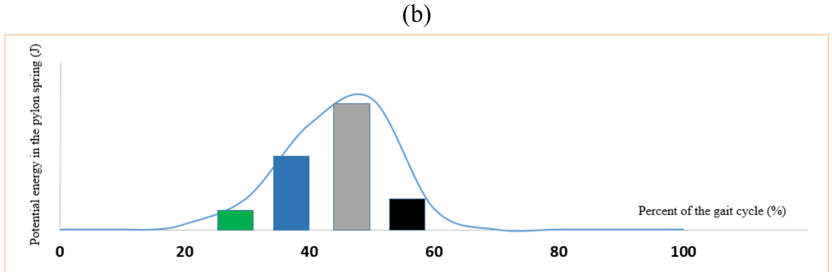
Fig. **11** (b), 11 (c), and 11 (d) displays the typical changes in the angles (degree) of the knee, ankle, and hip of the prosthetic leg throughout the process of walking. The proposed prosthetic leg's typical movement for individuals with transfemoral amputations during a single walking gait cycle closely mimics the regular patterns of movement exhibited by healthy individuals.

(a)



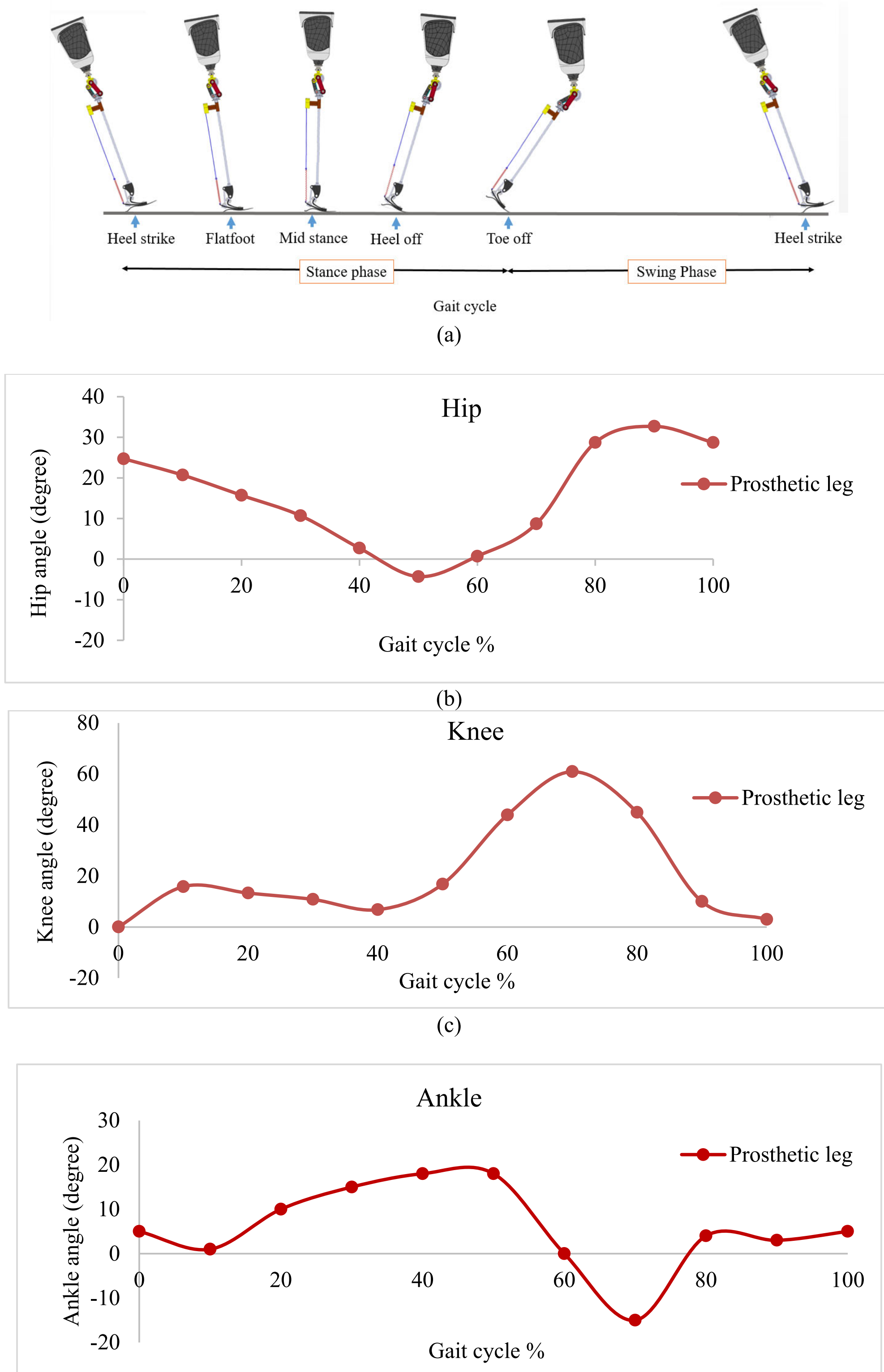
(Clutch engaged at heel strike) (clutch locked at flat foot) (Potential energy stored in spring) (Potential energy released from spring at push-off)

(b)



(c)

Fig. 10. The function of the passive mechanism. (a) The general behavior of the proposed transfemoral prosthetic leg to store energy, (b) The main proceedings of the smart-clutch, (c) The stored potential energy during the walking gait cycle and released it.



(d)

Fig. 11. (a) The typical motion of the suggested prosthetic leg for individuals with transfemoral amputations throughout a single walking gait cycle, (b) The knee angles during the walking gait cycle, (c) The hip angles during the walking gait cycle, (d) The ankle angles during the walking gait cycle.

6. Conclusions

This article suggests a new method for aiding above-knee amputees with their mobility by introducing a passive mechanism. The study focused on established prostheses and utilized the 3R36 polycentric knee and ESAR foot. Accordingly, at the stance phase, the center of pressure rotates around the ankle joint and proceeds to forward naturally. This mechanism is also designed to connect the prosthetic leg from the pylon to the ankle joint in a way that can store and release energy, resulting in improved functionality of basic movements and increased energy efficiency. The inclusion of passive components eliminates the need for an external power source, making the product lightweight and easily transportable. The results of the study indicate that the springs within the mechanism mainly store elastic energy during the early stage of the stance phase and release it during the end stage of the stance phase, contributing to positive work that propels the ankle normally. However, further analysis of the entire prosthetic leg with an exoskeletal model of the human body is necessary to better understand movement patterns and muscle energy required during walking with this mechanism. Overall, this new method has the potential to significantly enhance the quality of life for amputees and should be further explored in future research.

Acknowledgement

The authors declare there is no conflict of interest regarding this research. They also declare there is no financial aid supporting this research work. All authors contributed equally to the manuscript.

References

Al-Maliky, F. T., & Chiad, J. S. (2021). Study and evaluation of four bar polycentric knee used in the prosthetic limb for transfemoral amputee during the gait cycle. In *Materials Today: Proceedings*. https://doi.org/10.1016/j.matpr.2020.12.709

Blaya, J. A., & Herr, H. (2004). Adaptive Control of a Variable-Impedance Ankle-Foot Orthosis to Assist Drop-Foot Gait. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, *12*(1), 24–31. https://doi.org/10.1109/TNSRE.2003.823266

Collins, S. H., Bruce Wiggin, M., & Sawicki, G. S. (2015). Reducing the energy cost of human walking using an unpowered exoskeleton. *Nature*, *522*(7555), 212–215. https://doi.org/10.1038/nature14288

Diller, S. B., Collins, S. H., & Majidi, C. (2018). The effects of electroadhesive clutch design parameters on performance characteristics. *Journal of Intelligent Material Systems and Structures*, *29*(19), 3804–3828. https://doi.org/10.1177/1045389X18799474

Diller, S., Majidi, C., & Collins, S. H. (2016). A lightweight, low-power electroadhesive clutch and spring for exoskeleton actuation. In *Proceedings - IEEE International Conference on Robotics and Automation* (Vol. 2016-June, pp. 682–689). https://doi.org/10.1109/ICRA.2016.7487194

Dollar, A. M., & Herr, H. (2008). Lower extremity exoskeletons and active orthoses: Challenges and state-of-the-art. *IEEE Transactions on Robotics*. https://doi.org/10.1109/TRO.2008.915453

El-Sayed, A. M., Hamzaid, N. A., & Abu Osman, N. A. (2014). Technology efficacy in active prosthetic knees for transfemoral amputees: A quantitative evaluation. *Scientific World Journal*, *2014*(July). https://doi.org/10.1155/2014/297431

Farris, D. J., & Sawicki, G. S. (2012a). Linking the mechanics and energetics of hopping with elastic ankle exoskeletons. *Journal of Applied Physiology*, *113*(12), 1862–1872. https://doi.org/10.1152/japplphysiol.00802.2012

Farris, D. J., & Sawicki, G. S. (2012b). The mechanics and energetics of human walking and running: A joint level perspective. *Journal of the Royal Society Interface*, *9*(66), 110–118. https://doi.org/10.1098/rsif.2011.0182

Ferris, D. P., Czerniecki, J. M., & Hannaford, B. (2005). An ankle-foot orthosis powered by artificial pneumatic muscles. *Journal of Applied Biomechanics*, *21*(2), 189–197. https://doi.org/10.1123/jab.21.2.189

Ferris, D. P., Gordon, K. E., Sawicki, G. S., & Peethambaran, A. (2006). An improved powered ankle-foot orthosis using proportional myoelectric control. *Gait and Posture*. https://doi.org/10.1016/j.gaitpost.2005.05.004

Gailey, R., Allen, K., Castles, J., Kucharik, J., & Roeder, M. (2008). Review of secondary physical conditions associated with lower-limb amputation and long-term prosthesis use. *Journal of Rehabilitation Research and Development*. https://doi.org/10.1682/JRRD.2006.11.0147

Galle, S., Derave, W., Bossuyt, F., Calders, P., Malcolm, P., & De Clercq, D. (2017). Exoskeleton plantarflexion assistance for elderly. *Gait and Posture*. https://doi.org/10.1016/j.gaitpost.2016.11.040

Galle, Samuel, Malcolm, P., Derave, W., & De Clercq, D. (2014). Enhancing performance during inclined loaded walking with a powered ankle–foot exoskeleton. *European Journal of Applied Physiology*, *114*(11), 2341–2351. https://doi.org/10.1007/s00421-014-2955-1

Goldfarb, M. (2013). Consideration of Powered Prosthetic Components as They Relate to Microprocessor Knee Systems. *JPO Journal of Prosthetics and Orthotics*. https://doi.org/10.1097/jpo.0b013e3182a8953e

Grabowski, A. M., & Herr, H. M. (2009). Leg exoskeleton reduces the metabolic cost of human hopping. *Journal of Applied Physiology*, *107*(3), 670–678. https://doi.org/10.1152/japplphysiol.91609.2008

Hansen, A., & Starker, F. (2018). Prosthetic foot principles and their influence on gait. In *Handbook of Human Motion*. https://doi.org/10.1007/978-3-319-14418-4_74

Kao, P. C., Lewis, C. L., & Ferris, D. P. (2010). Invariant ankle moment patterns when walking with and without a robotic ankle exoskeleton. *Journal of Biomechanics*, *43*(2), 203–209. https://doi.org/10.1016/j.jbiomech.2009.09.030

Lamoth, C. J. C., Ainsworth, E., Polomski, W., & Houdijk, H. (2010). Variability and stability analysis of walking of transfemoral amputees. *Medical Engineering and Physics*. https://doi.org/10.1016/j.medengphy.2010.07.001

Major, M. J., & Fey, N. P. (2018). Considering passive mechanical properties and patient user motor performance in lower limb prosthesis design optimization to enhance rehabilitation outcomes. *Physical Therapy Reviews*, *22*(3–4). https://doi.org/10.1080/10833196.2017.1346033

Major, M. J., Twiste, M., Kenney, L. P. J., & Howard, D. (2014). The effects of prosthetic ankle stiffness on ankle and knee kinematics, prosthetic limb loading, and net metabolic cost of trans-tibial amputee gait. *Clinical Biomechanics*. https://doi.org/10.1016/j.clinbiomech.2013.10.012

Major, M. J., Twiste, M., Kenney, L. P. J., & Howard, D. (2016). The effects of prosthetic ankle stiffness on stability of gait in people with transtibial amputation. *Journal of Rehabilitation Research and Development*. https://doi.org/10.1682/JRRD.2015.08.0148

Malcolm, P., Derave, W., Galle, S., & De Clercq, D. (2013). A Simple Exoskeleton That Assists Plantarflexion Can Reduce the Metabolic Cost of Human Walking. *PLoS ONE*, *8*(2). https://doi.org/10.1371/journal.pone.0056137

Mohanty, R. K., Mohanty, R. C., & Sabut, S. K. (2020). A systematic review on design technology and application of polycentric prosthetic knee in amputee rehabilitation. *Physical and Engineering Sciences in Medicine*, *43*(3), 781–798. https://doi.org/10.1007/s13246-020-00882-3

Mooney, L. M., & Herr, H. M. (2016). Biomechanical walking mechanisms underlying the metabolic reduction caused by an autonomous exoskeleton. *Journal of NeuroEngineering and Rehabilitation*, *13*(1). https://doi.org/10.1186/s12984-016-0111-3

Mooney, L. M., Rouse, E. J., & Herr, H. M. (2014). Autonomous exoskeleton reduces metabolic cost of human walking. *Journal of NeuroEngineering and Rehabilitation*, *11*(1), 1–5. https://doi.org/10.1186/1743-0003-11-151

Moxey, P. W., Gogalniceanu, P., Hinchliffe, R. J., Loftus, I. M., Jones, K. J., Thompson, M. M., & Holt, P. J. (2011). Lower extremity amputations - a review of global variability in incidence. *Diabetic Medicine*. https://doi.org/10.1111/j.1464-5491.2011.03279.x

Phanphet, S., Dechjarern, S., & Jomjanyong, S. (2017). Above-knee prosthesis design based on fatigue life using finite element method and design of experiment. *Medical Engineering and Physics*, *43*, 86–91. https://doi.org/10.1016/j.medengphy.2017.01.001

Sarvestani, A. S., & Azam, A. T. (2013). Amputation: A ten-year survey. *Trauma Monthly*, *18*(3), 126–129. https://doi.org/10.5812/traumamon.11693

Sawicki, G. S., & Ferris, D. P. (2008). Mechanics and energetics of level walking with powered ankle exoskeletons. *Journal of Experimental Biology*, *211*(9), 1402–1413. https://doi.org/10.1242/jeb.009241

Sawicki, G. S., Gordon, K. E., & Ferris, D. P. (2005). Powered lower limb orthoses: Applications in motor adaptation and rehabilitation. In *Proceedings of the 2005 IEEE 9th International Conference on Rehabilitation Robotics* (Vol. 2005, pp. 206–211). https://doi.org/10.1109/ICORR.2005.1501086

Schmalz, T., Blumentritt, S., & Jarasch, R. (2002). Energy expenditure and biomechanical characteristics of lower limb amputee gait: The influence of prosthetic alignment and different prosthetic components. *Gait and Posture*, *16*(3), 255–263. https://doi.org/10.1016/S0966-6362(02)00008-5

Shepherd, M. K., Azocar, A. F., Major, M. J., & Rouse, E. J. (2018). Amputee perception of prosthetic ankle stiffness during locomotion. *Journal of NeuroEngineering and Rehabilitation*. https://doi.org/10.1186/s12984-018-0432-5

Soriano, J. F., Rodríguez, J. E., & Valencia, L. A. (2020). Performance comparison and design of an optimal polycentric knee mechanism. *Journal of the Brazilian Society of Mechanical Sciences and Engineering*, *42*(5), 1–13. https://doi.org/10.1007/s40430-020-02313-6

Tryggvason, H., Starker, F., Lecomte, C., & Jonsdottir, F. (2020). Use of dynamic FEA for design modification and energy analysis of a variable stiffness prosthetic foot. *Applied Sciences (Switzerland)*, *10*(2). https://doi.org/10.3390/app10020650

Ventura, J. D., Klute, G. K., & Neptune, R. R. (2011). The effects of prosthetic ankle dorsiflexion and energy return on below-knee amputee leg loading. *Clinical Biomechanics*. https://doi.org/10.1016/j.clinbiomech.2010.10.003

Westerterp, K. R. (2013). Physical activity and physical activity induced energy expenditure in humans: Measurement, determinants, and effects. *Frontiers in Physiology*. https://doi.org/10.3389/fphys.2013.00090

Wiggin, M. B., Sawicki, G. S., & Collins, S. H. (2011). An exoskeleton using controlled energy storage and release to aid ankle propulsion. In *IEEE International Conference on Rehabilitation Robotics*. https://doi.org/10.1109/ICORR.2011.5975342

Wurdeman, S. R., Stevens, P. M., & Campbell, J. H. (2018). Mobility Analysis of AmpuTees (MAAT I): Quality of life and satisfaction are strongly related to mobility for patients with a lower limb prosthesis. *Prosthetics and Orthotics International*. https://doi.org/10.1177/0309364617736089

Yandell, M. B., Tacca, J. R., & Zelik, K. E. (2019). Design of a Low Profile, Unpowered Ankle Exoskeleton That Fits Under Clothes: Overcoming Practical Barriers to Widespread Societal Adoption. *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, *27*(4), 712–723. https://doi.org/10.1109/TNSRE.2019.2904924