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Design and Evaluation a New Type of Semi-active Lower Limb with Knee Joint Booster

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ABSTRACT

This article presents a new lower limb orthosis for helping weak knees during human locomotion. The orthosis structure has 10 degrees of freedom. It utilizes a series elastic actuator, equipped with an elastic rope that transfers torque generated by the motor to the orthosis link. The performance of the proposed lower limb orthosis is virtually simulated by using ADAMS-MATLAB Co-simulation software. The orthosis is designed based on the anthropometric data of a normal human body with a mass of 76 kg and a height of 180 cm. The simulation scenario involves walking with an average speed of 1 m/s on a straight path, and the knee orthosis can bear 40% of the torque exerted by the knee joint during the gait cycle. The simulation aims to evaluate the effectiveness and efficiency of the orthosis in assisting the weak knee joint. The simulation results indicate that The orthosis reduced the knee joint torque by more than 13 Nm in a healthy person, which indicates lower forces on the weak knee. Moreover, the orthosis decreases the maximum energy needed per gait cycle, which implies a higher efficiency and reduces the metabolic cost of the body in the gait cycle.

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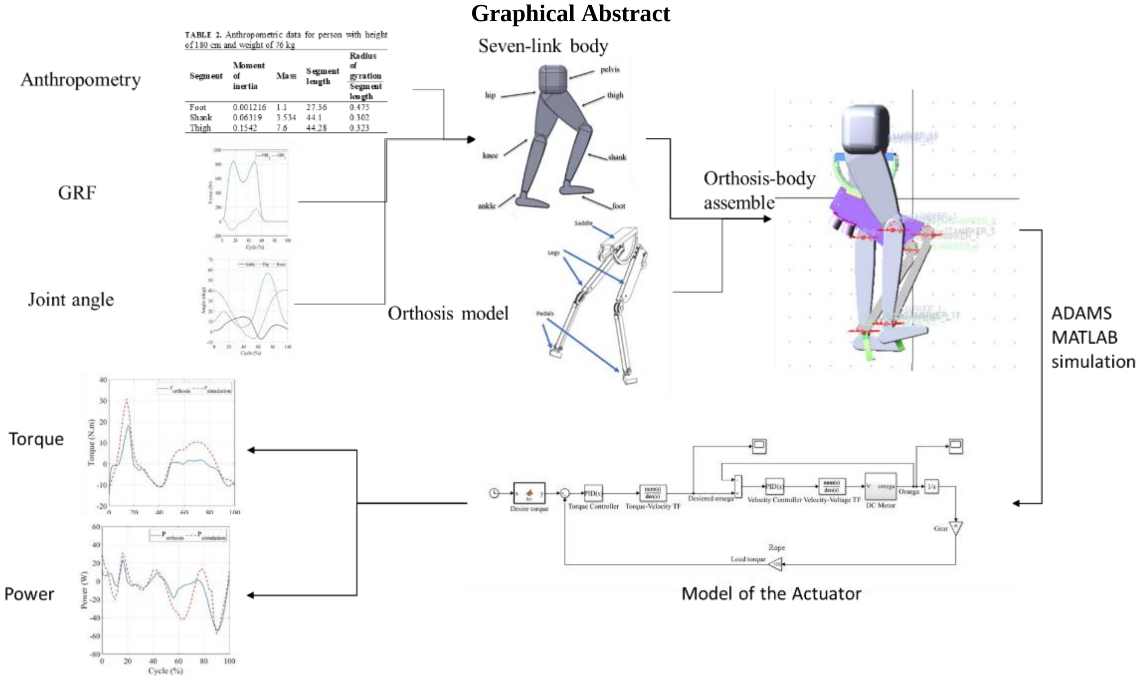
Graphical Abstract



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1. INTRODUCTION

The knee joint performs a vital role in human mobility, as it carries the weight of the body, enables lower limb swing, and absorbs impact from the ground. It is the largest joint in the body and experiences varying forces and torques while walking. The maximum amount of these torques are applied to the knee joint in the stance phase (1). The human movement relies on the knee joint's function and will be impaired if it is not working properly. Prior studies of wearable robots have noted the importance of rehabilitating people who have weak knee joints (2, 3).

Movement disorders affecting various parts of the body are prevalent worldwide, and they can stem from a variety of factors. The aging of the global population is one of the most important medical and social problems worldwide (4). Based on reports from the World Health Organization, it is anticipated that by 2030, approximately one in six individuals globally will be 60 years or above. This demographic shift is expected to lead to an increased demand for rehabilitation services, due to factors such as accidents, a growing number of stroke patients, weight gain, and weakness of some muscles like gluteal muscles, quadriceps, adductors, hamstrings, gastrocnemius, and soleus (5, 6).

Additionally, the lack of physiotherapists is one of the reasons for the shift to robot-assisted rehabilitation (7). The development of rehabilitation devices is of utmost importance in enabling patients to acquire the necessary ability to lead normal lives. These devices serve as a training tool, assisting patients in their physical and cognitive therapy, which in turn, enables them to regain their independence. The provision of such devices, therefore, plays a crucial role in the restoration of patients' physical and mental health.

In this manner, wearable robots are designed to reduce the torque on the joints and body parts (8), alleviate the exhaustion resulting from prolonged walking (9), and even enhance the speed of walking (10). Orthoses are a common types of wearable robot that provide external support for the neuromuscular and skeletal systems. They are worn parallel to the body's limbs and are used to assist patients in correcting or restoring movement functions. Orthoses can be classified into three types: active (11, 12), semi-active (13), and passive (14). Active and semi-active orthoses rely on an external power source for transmission and are mainly used by patients with limbs and joints diseases (15).

Although there have been notable improvements in the design of them, these devices continue to encounter some obstacles. Despite the notable technological advancements that have revolutionized the orthosis industry, the cost of these medical devices remains a pressing concern for a considerable number of individuals. Unfortunately, a significant portion of the population is still unable to afford the high prices of orthotic devices. It is an ongoing challenge to ensure safe interaction between orthosis devices and the human body. Comfort, ergonomic suitability, and potential skin irritation are important factors that must be fully addressed to ensure user satisfaction and long-term usability since these devices are intended to be worn or attached to the body.

In this manner, this article aims to improve the design of orthoses by addressing some of its weaknesses. By positioning the orthosis between the legs, the hands can move freely and the center of mass is brought closer to the body's own center of mass. This reduces the moments of inertia of the body. Moreover, the orthosis design eliminates interaction with the lower limbs, making it even safer to use with series elastic actuators (16). It also considered ergonomics when designing the knee structures to minimize misalignment, an aspect that has been overlooked in previous studies.

On the other hand in contrast of reported data (17, 18) by replacing the traditional spring and ball screw with elastic rope, the cost of the orthosis can be significantly reduced. The elastic rope connects the motor to the knee joint of the orthosis and takes up minimal space by being positioned between the orthosis link. Notably, unlike in previous studies, the user will not experience any additional weight from the orthosis (19, 20).

To summarize the content of this article, a newly designed lower limb orthosis was explored by modeling it in a walking position using Adams software. The orthosis includes two links for each leg, two pedals, and a saddle positioned between the legs. For this study, a series elastic actuator was chosen as it offers an electric motor and an elastic rope to regulate the pivot's stiffness. This stiffness is managed by a feedback loop that measures the torque and the pivot's angular displacement. A closed-loop torque control system with a reaction torque sensor is utilized that can precisely achieve assistive or resistive torques for different physical therapies. As a result, this controllable stiffness pivot mechanism offers a flexible and adaptable approach to support a weakened knee joint.

The remainder of this paper is structured as follows: In the second section, the physical model of the orthosis is described, along with a detailed explanation of the actuator model and its components. The third section covers the participant's properties, as well as the collection of his kinematic and anthropometric data. In section four, the knee torque of a healthy individual is simulated using Adams software, which is later evaluated using experimental data. Finally, the orthosis is implemented on it. The results of this implementation are presented and discussed in section five. The paper concludes with the presentation of the conclusions in the final section.

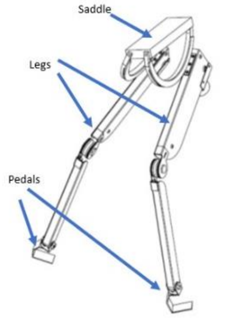
2. MODEL DESCRIPTION

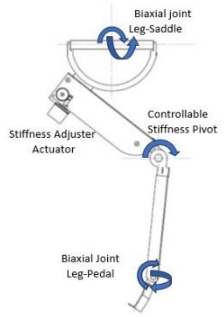
2. 1. ORTHOSIS The orthotic device comprises two interconnected links, namely the shank and thigh, and incorporates a saddle positioned between the user's legs. The design of the orthosis is intended to offer stable support to the lower extremities, enhancing the user's mobility and overall functionality Figure 1(a). The device is equipped with two degrees of freedom, allowing the user to move in both the sagittal and frontal planes. Additionally, it features a pedal that connects to the soles of an individual's shoes through a two-degree-of-freedom joint, allowing movement in both sagittal and transverse planes as shown in Figure 1(b). When simulating movement in the sagittal plane, there are six degrees of freedom, with one in each joint. The ankle and hip joints are aligned with those of a typical human model, but the knee joints are parallel to the wearer's knee, not in alignment with it as shown in Figure 2(a). The orthosis that has been suggested, when combined with the thigh and leg of the user's lower limbs, creates a four-bar mechanism as shown in Figure 2(b). Only the knee joints of orthosis are active and can perform mechanical work. To avoid disturbing the user and minimize the weight burden on the knee joint, it is recommended to place the motor at the back of the thigh link, near the center of mass instead of directly on the knee joint.

In order to enhance stability and mitigate torque during walking, the knee joint of the orthosis is connected to a motor via an elastic rope. The elastic rope is responsible for applying a 15 Nm preload in the direction of knee extension, thereby preventing the leg from bending while the person is in the stance position. During the swing phase, which accounts for 55-80 percent of the walking cycle, the motor produces a torque in the direction of flexion to assist the patient in maintaining their foot position. In order to maintain the safety of the patient, we gradually reduce the amount of applied torque.

It is important to be mindful that the use of orthosis aims to decrease the forces exerted on the knee joint. Therefore, it is crucial to ensure that the weight of the orthosis does not impede walking and that the auxiliary device's size does not hinder mobility. In general, reducing the weight of the orthosis and increasing its strength will increase the cost of the orthosis, and it may not be economically utilizable for all people in society.
The design must strike a balance between the cost of the design and the structural characteristics of the orthosis. The orthosis links are typically made of aluminum alloy, which is not only lightweight but also has high-strengt (21).

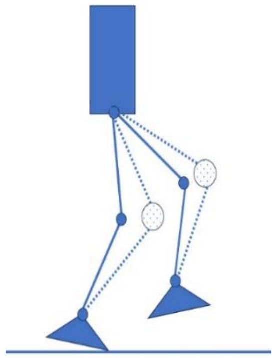
The present orthosis offers a noteworthy economic advantage over the existing series elastic actuator-based orthoses, particularly in terms of its power transfer system. This is accomplished through the elimination of





(a) (b)

Figure 1. Charactristics of designed lower limb orthosis (a) main mechanical component (b) device degrees of freedom



(a) (b)

Figure 2. Schematic of how orthosis act on knee joint (a) the simplicity of orthosis components and assumption of joints place (b) Forming a four-bar mechanism by orthosis links and individual's legs

components such as ball screws and replacement of springs with elastic ropes. The resulting cost reduction and improved efficiency make it a promising choice for business and academic settings alike. Moreover, the orthosis's design situated between the legs allows for unhindered movement of the hands. The orthosis can also support a portion of the user's body weight while they are sitting on a saddle that is in direct contact with the ground. This layout improves communication between individuals and robots.

2. 2. ACTUATOR The actuator which is selected to assist human gait is the series elastic actuator. Actuator components include four main groups motor, elastic agent, control system, and sensors.

1) Motor

The motor is one of the significant components of the series elastic actuator, which provides the necessary power and torque to assist in walking. power capacity and maximum torque are the two primary performance factors in properly selecting a motor for a specific application or use. A motor's capacity is determined by the maximum permissible power, which can be measured by multiplying the torque it generates by the angular velocity. The speed and torque of the motor are inversely related. The speed must be decreased proportionally if we want to increase the output torque. In order to increase output torque and transmit power, gearboxes are commonly utilized in robotic applications. For this particular research, the Maxon RE35 24V DC graphite brush motor was employed, which has a power rating of 90 watts and weights 340 grams. To further enhance performance, the study also utilized the GP42C planetary gearhead, which weights 465 grams and has a transmission ratio of 156:1. This combination of motor and gearbox is capable of consistently handling a maximum torque input of 15 Nm. The charactristics of proposed motor are listed in Table 1.

2) elastic element

This particular design employs an elastic rope as the elastic component. This component is installed within the orthosis thigh link to serve as a connection between the motor and the knee joint. The elasticity rope is connected to two flexible ropes that are carefully wrapped around pulleys, as illustrated in Figure 3.

Another advantage of this type of elastic design is that it is placed between the motor and the load and softens the power or torque transmitted from the motor to the orthosis link. It also acts as an energy buffer, storing energy when the constraint joint does negative work and helping it when it does positive work. By increasing the stiffness of the rope, the controllability of the system improves at higher frequencies. But it should be noted a hard elastic element is less sensitive to small torques, and this causes the accuracy of torque control to decrease. In addition, the stiffer the rope, its nonlinear terms increase and it no longer obeys Hooke's law. Spring constant values for walking aids are usually in the range of 100 to 300 Nm/rad (22), and the maximum torque they can apply is from 10 to 100 Nm (23). According to the mentioned points, the constant value of the spring was determined to be 170 Nm/rad.

3) Control system

Due to the interaction between the orthosis and the body, this device must be mechanically compatible with the anatomy of the wearer's body so as not to interfere with his walking (24). In the arms of industrial robots, a high impedance actuator is used in a large frequency bandwidth to effectively reject disturbances. But because both legs are in contact with the ground separately as a rigid body during gait, a low impedance actuator is used to increase safety and comfort and prevent resistance to human movement (25). In general, low impedance excitation means that the actuators control the force or torque applied to the load rather than the commanding position or angle. The series elastic actuator has

TABLE 1. Parameters of motor used to drive series elastic element

Nominal Voltage	24 V
Terminal Resistance	0.582 Ω
Terminal Inductance	0.191 mH
Torque Constant	0.0292 Nm/A
Back EMF Constant	0.0292 V s/rad
Rortor and Gearhead Inertia	88.1 g cm^2
Damping Constant	0.0013 Nms

Figure 3. Form of the elastic element inside hip link of orthosis

inherently low impedance due to the use of a spring between the gearbox and the load. In this design, the speed control loop provided by Robinson (26) is used for the motor. Using the speed controller in the motor overcomes some undesirable effects of the motor and gearbox such as non-linearity and stickiness, and it is much simpler than the flow controller from the point of view of implementation. By creating a torque controller on the speed controller in a cascade manner, the desired result can be achieved in the output of the actuator. This result applies as torque to the orthosis link. One of the most common control algorithms used in the industry is the Proportional Integrative Derivative (PID) control, This algorithm is widely applied in the industry because of its simplicity and easy implementation in control system (27).

4) Sensors

By getting feedback from key components, this structure can achieve extremely high control accuracy (28). The sensors that used in this study are: an encoder is placed at the end of the motor and before the gearbox to measure the motor's output speed, so it is possible to determine the type of gearbox and speed reduction ratio based on that information. A potentiometer measures the angle of the orthosis joint. By subtracting it from the gearbox output angle, it is multiplied by the spring stiffness and determines the final torque that enters the orthosis link. A potentiometer is also needed in the person's knee joint to detect the beginning of the gait cycle. It is also needed to turn on the motor to help the gait cycle. In addition, by using two switches in the toe and heel parts of a person's sole, it is possible to recognize the beginning and end of the stance phase.

3. BIOMECHANICAL CONSIDERATION

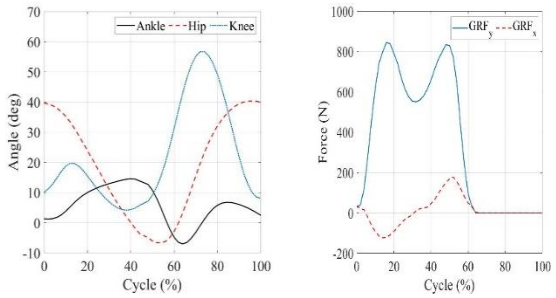
3. 1. Gait Analysis Walking can reveal consequential information about people's health status. By having the information obtained from how to walk and their detailed analysis, it is possible to treat many diseases and disabilities related to the lower limbs. This information can even prevent many problems.

The analysis of human walking is one of the most challenging issues in biomechanics due to the many nonlinearities in the walking equations, GRF, and the force exerted by the muscles.

Among the approaches to obtain the necessary information for gait analysis, the method of using clinical gait analysis, known as motion capture, is straightforward, more accurate, and more widely used than other methods. Motion capture laboratories are usually equipped with cameras and force plates. These devices are used to determine kinematic information and external forces acting on the body. In this regard, Kirtley (29) conducted various experiments on the way the organs are placed in the walking cycle and the forces transferred when the foot contacts the ground, the result of which is the basis of the current research (Figure 4). Using inverse kinematics has some advantages such as avoiding position error, that may occur in the use of the forward kinematic approach (30).

It is essential to understand how these angles are defined to examine the graph of joint angles during the gait cycle. In this article, joint motion in this plane is simply referred to as flexion and dorsiflexion (positive direction), extension and plantarflexion (negative direction) (31). It should be noted that this data is for a person who moves at 1 m/s on a straight path. Increasing walking speed decreases the percentage of the stance phase, which includes double support and single support, and increases the swing phase. Also, as the person's speed Increases, the maximum angle of the joints in the walking cycle increases.

In a general review of the position of the limbs in a walking cycle, it can be found that if movement is on a straight path and the body's speed remains constant, the position of the limbs relative to each other is repeated periodically. In other words, the information obtained from one walking cycle can be generalized to the entire



(a) (b)

Figure 4. The properties of a person with 76 kg weight with the speed of 1 m/s. (a) the measure of joints angle during the gait cycle (b) the ground reaction force (GRF) applied to the foot

TABLE 2. Anthropometric data for person with height of 180 cm and weight of 76 kg

Segment	Moment of inertia	Mass	Segment length	Radius of gyration Segment length
Foot	0.001216	1.1	27.36	0.475
Shank	0.06319	3.534	44.1	0.302
Thigh	0.1542	7.6	44.28	0.323

walking process in other cycles with an acceptable approximation. The vertical force of Earth's reaction is more than in other directions. At some stages of the walking cycle, this force reaches even higher and lower than body weight. One of the reasons for this is the decreasing and increasing acceleration of the lower limbs during the walking cycle. For example, when the foot hits the ground, the speed is negative, and its amount decreases, as a result, the acceleration will be positive, and according to Newton's second law, the sum of this positive value along with the force caused by the body's weight will enter the foot.

3. 2. Inverse Dynamics In this article inverse dynamics' method is used to obtain forces and moments in the joints whereby kinematic information and external forces are used. The inertia, size, and mass of each body part must be known to correctly estimate joint torques. Considering that it is difficult for each person to measure anthropometric characteristics, the size of these parameters is usually estimated from the determined values of corpses collected by Winter (32). This information for a subject with a weight of 76 kg and a height of 180 cm can be seen in the Table 2. Since many muscles cause movement in the joint, it is impossible to accurately measure which muscles contribute to applying force. But on the other hand, the inverse dynamics method is used of obtaining kinematic information, as well as forces and moments in joints and limbs.

where the moment of inertia is around the center of mass of the organs and is obtained according to the following formula:

$$I_i = m(L_i \rho_i)^2 \quad (1)$$

where L_i is the length of each member, ρ_i is the radius of gyration to the length of each member at the center of gravity.

4. SIMULATION

4. 1. GAIT In order to showcase the mechanics of walking, certain simplifications are commonly employed. Specifically, it is assumed that the upper body and hands remain stationary during human locomotion. Additionally, the lower limbs are presumed to be symmetrical, meaning that information gleaned from one leg can be applied to the other foot as well.

After simplifying the model, Adams software is used to design the seven-link model with rigid components. Due to the assumption that walking movements mostly occur in the sagittal plane, movements in other planes can be disregarded. As a consequence, the joints are modeled as a hinge joints. Use a simply hinged 1 degree of freedom system is designed to facilitate system control through the simplification of dynamic models; it also serves the purpose of improved durability. In reality, however, human joints do not simply rotate around one axis. The mass and center of mass of these links are determined according to Table 2. As a movement agent, a motion is placed on each of the joints, which is only able to move in the sagittal plane. The input function of each of these motions is the angular information obtained from the fitting of Kirtley's experiments are shown in Figure 4.

Interpolation involves identifying data points come before and after a given set of information. In Adams, the joint angle curve is imported as spline data, and a third-order spline curve is utilized to interpolate these points. Due to the limitations of the contact model in Adams and to correctly apply the GRF to the individualas foot, this force was introduced to the center of pressure (COP) the foot in each part of the cycle. center of pressure (COP) is the term given to the point of application of the ground reaction force vector. The ground reaction force vector represents the sum of all forces acting between a physical object and its supporting surface. The way to enter this force in the software is to consider during the stance phase, the center of pressure varies from the heel to the toe. Therefore, each percent of the cycle is attributed to a specific point of the foot.

With the help of the reference coordinate device, it is possible to calculate the position, speed, and acceleration of each component using kinematic equations based on the reference coordinates device. If it is assumed that the person moves at a constant speed, considering that the trunk is fixed during walking, then the horizontal acceleration for the trunk part is equal to zero. For example, for thigh link:

$$\begin{aligned} &a_t = a_{tr} + a_{t/tr} \\ &a_{t/tr} = \alpha_t \times r_t + \omega_t \times (\omega_t \times r_t) \\ &a_{t/tr} = \alpha_t \vec{k} \times (r sin\theta_h \vec{\imath} + r cos\theta_h \vec{\jmath}) + \\ &\omega_t \vec{k} \times [\omega_t \vec{k} \times (r sin\theta_h \vec{\imath} + r cos\theta_h \vec{\jmath})] \end{aligned} \quad (2)$$

where a_t *is* the acceleration of the thigh link, a_{tr} is the acceleration of the trunk and $a_{t/tr}$ is the acceleration of the thigh relative to the trunk.

To use the inverse dynamic method, we consider all the forces that come from the ground and the linear and angular acceleration of the links. This helps us calculate the forces, moments, and power for each joint.

The force of the foot determines from the kinematic equations according to the newton's second law. The torque of the ankle joint can obtain as follow:

$$\begin{aligned} &\sum M_a = I\alpha_a \\ &M_a + f_x r sin(\beta) + G_x r sin(\beta) + f_y r cos(\beta) - \\ &G_y r sin(\beta) = I\alpha_a \end{aligned} \quad (3)$$

where $\beta = \theta_h - \theta_k - \theta_a$ which respectively are the angle of hip, knee and ankle joints.

In the same way, the shank and thigh forces and torques are also calculated. In general, stance and swing dynamic equations are different. Because GRF component is disapeared in swing phase.

The results of gait simulation are shown in Figure 5. The comparison of torque in experimental data and simulation has an average error percentage of 9.51%. Comparison of power in laboratory mode and simulation has an average error percentage of 5.74%.

4. 2. ORTHOSIS Orthosis components, the links, and the physical model are transferred from the Solidworks to Adams. However, the actuator model is simulated in Simulink.

First, the motor model is created according to the specifications given in Table 1. This model includes two mechanical and electrical parts that follow the following two general equations:

$$V = L\frac{di}{dt} + Ri + e \quad (4)$$

$$T = J\frac{d\omega}{dt} + B\omega + T_L \quad (5)$$

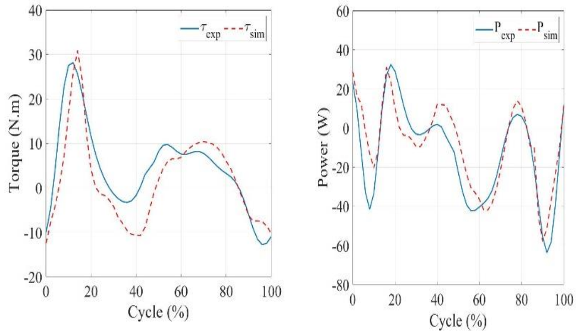
According to the explanations, the actuator model is simulated in the MATLAB/Simulink environment. This model includes an internal velocity PID controller, and a torque PID controller cascaded on the velocity controller. A gear system is installed between the motor output and the spring input. The gear ratio causes the motor output speed to be reduced while the torque transmitted to the orthosis links is increased.

Gear head modeling consists of several gears, which are alternated by a gain block in Simulink. The elastic element is also placed between the gearbox and the orthosis link, modeled using the block. The schematic model of the actuator can be seen in Figure 6.

The model of actuator applied on leg with orthosis is shown in Figure 7.

5. RESULTS AND DISCUSSION

The orthosis and body model were exported and The physical model of the orthosis was implemented on the body in Adams software (Figure 8). The placement of the orthosis was thoughtfully considered, as the wearer's center of gravity is closer to the inner part of the body, this ensures that it exerts minimal force on their lower limbs and does not impede their ability to walk. In addition, the hand movements are executed with ease and do not hinder movement while walking. In order to assess how orthoses affect an individual's legs, it is crucial to establish a linkage between the actuator model in Simulink and the physical models in Adams. This work is done with ADAMS-MATLAB Co-simulation. To define the orthosis model in Adams software, it is necessary to define an inputs and outputs. As previously stated, the input to the system is the torque that the actuator must apply to the orthosis. The system outputs



(a) (b)

Figure 5. Validation of the knee (a) torque and (b) power obtained in Adams simulation with Kirtly's experiment for healthy individual

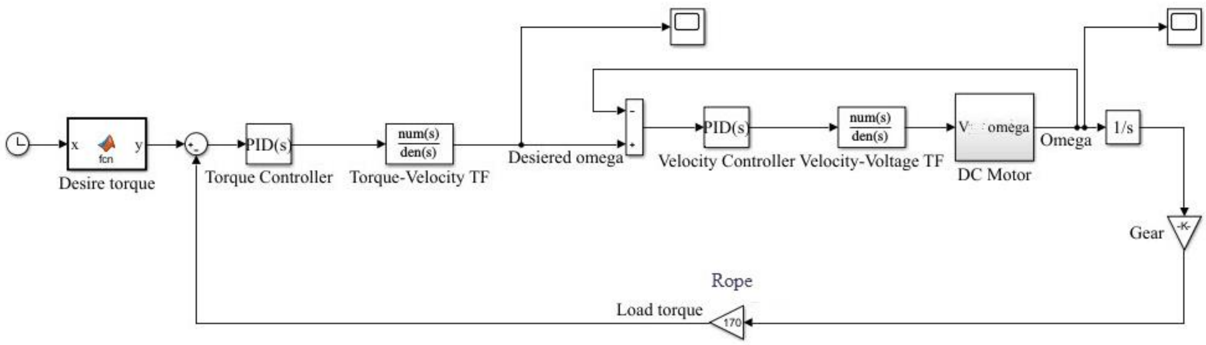


Figure 6. The schematic model of the actuator that simulated in MATLAB/Simulink environment. Both torque and velocity controllers are PID types. Transfer functions convert the output of controllers to the velocity input of the motor.

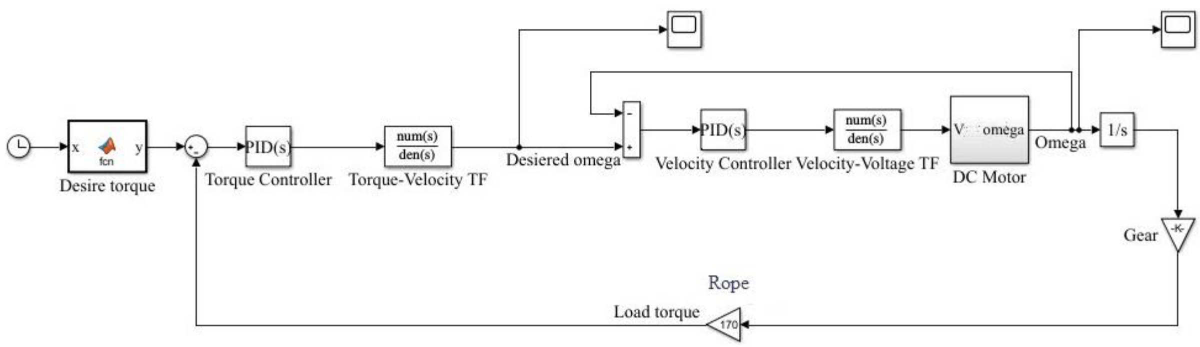


Figure 7. The model of actuator applied on leg with orthosis. The sensor measure angle of orthosis joint and reduced from the engine output angle. This angle multiplies by spring stiffness and applies torque to the orthosis link.

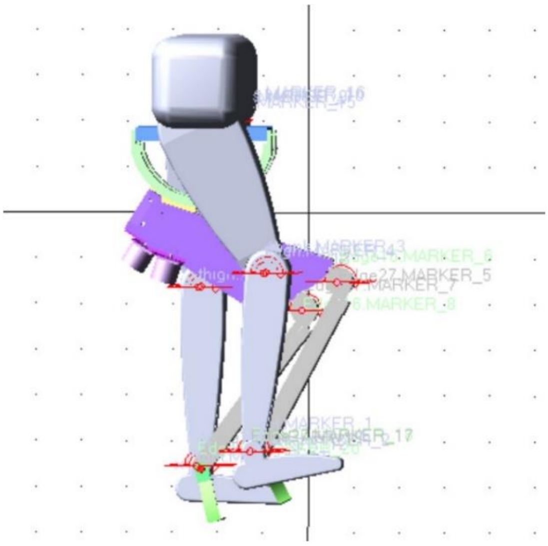


Figure 8. Physical model of orthosis applied on the body in MSC Adams

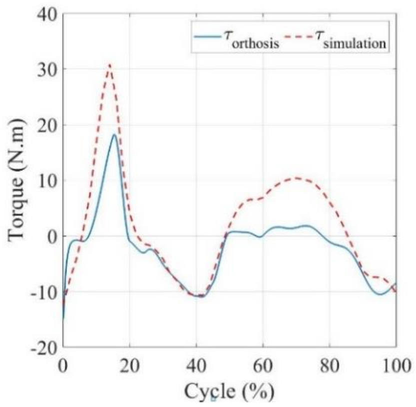


Figure 9. Reduce of knee torque with orthosis against without orthosis

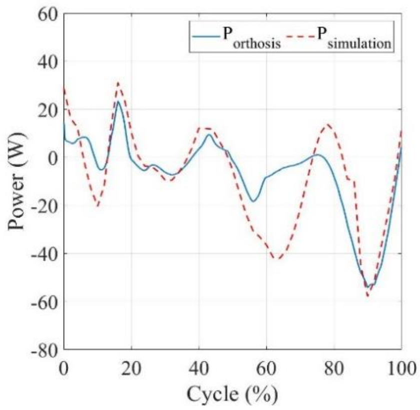


Figure 10. Comparison between knee power with orthosis and without orthosis

include a sensor located at the knee joint of the orthosis that provides angle information to the system, as well as the torque and angular velocity in the knee joint of the lower body model. These outputs are the ultimate objectives of this study. Finally, The model is exported to Simulink, the actuator's output is connected to the torque input defined in the orthosis joint, and an angle sensor measures information about the orthosis knee angle (Figure 7).

After running the simulation, the results are obtained as shown in Figures 9 and 10.

The data reveals that when the orthosis is present, the knee joint is subjected to a maximum torque of approximately 18 Nm. This value exhibits a significant reduction of over 40 percent when compared to the torque experienced in the normal state. It is worth noting that the reduction in torque can be attributed to the maximum output power of the motor. If a motor with a higher power is used, it would enable the knee joint to experience greater torques at the same velocity.

This device is particularly useful in the swing phase of a person's gait. The elastic rope present in the orthosis has pre-tension, which causes the orthosis link to extend at the beginning of the swing phase. However, the motor applies a torque to the orthosis link, preventing the extension and causing the foot to flex with the lowest possible torque step to zero. This action enables the person to pass through the swing phase with ease and efficiency, improving their overall gait pattern.

When it comes to evaluating the efficacy of an exoskeleton, a crucial metric to consider is the metabolic cost required to walk or run. This metric measures the amount of energy expended during locomotion. In technical terms, energy is the integral of power over time, which means that if an exoskeleton is designed to reduce the power required in the joints, it can also decrease the amount of energy consumed. In fact, studies have shown that exoskeletons can reduce energy expenditure by up to 21% per gait cycle. This energy reduction can have significant benefits for individuals who use exoskeletons, as it can reduce fatigue and increase endurance, making it easier for them to perform daily activities and engage in physical exercise.

6. CONCLUSIONS

The study presents a new lower limb orthosis that has potential benefits for people with knee disorders and those who want to reduce the torque on their knees while walking. The research emphasizes the importance of design factors such as weight, strength, cost, and anatomical compatibility to create orthoses that enhance mobility while prioritizing user comfort and safety. To accomplish this, the orthosis was virtually simulated using MATLAB and Adams and assembled with a seven-link human lower limb model to evaluate its efficacy during individual walking. The results of the study indicate that the application of the orthosis to a healthy leg reduces the maximum torque exerted on the knee joint during walking by approximately 13 Nm. This device can also be beneficial to individuals with a 40% weakness in their knee joint. Furthermore, the study reveals that users of this orthosis consume around 21% less energy when compared to normal walking. These findings suggest that the use of this orthosis can have a significant impact on the joint health and energy expenditure of individuals with knee joint weaknesses.

For future work, A preliminary version of this orthotic device can be fabricated to assess its effectiveness in reducing body torque and energy expenditure by placing it on individuals' lower limbs. Additionally, other joints in the orthosis can be equipped with either passive or active actuators to address other movement disorders or muscle weaknesses. This provides a more comprehensive approach to treating musculoskeletal conditions, as the orthosis can be customized to meet individual needs. By incorporating these specialized actuators, patients can experience increased mobility and functionality, leading to improved quality of life.

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Persian Abstract

چکیده

در این مقاله یک ارتز جدید اندام تحتانی معرفی می شود که برای کمک به زانوهای ضعیف در حین راه رفتن طراحی شده است. ساختار ارتز دارای ۱۰ درجه آزادی است. این دستگاه از یک محرک الاستیک سری، مجهز به یک طناب الاستیک استفاده می کند که گشتاور تولید شده توسط موتور را به لینک ارتز منتقل می کند. عملکرد ارتز اندام تحتانی پیشنهادی به صورت مجازی با استفاده از لینک نرم افزارهای شبیه سازی آدامز و متلب شبیه سازی شده است. ارتز بر اساس داده های آنتروپومتریک یک انسان عادی با وزن بدن ۷٦ کیلوگرم و قد ۱۸۰ سانتی متر طراحی شده است. سناریوی شبیه سازی شامل راه رفتن با سرعت متوسط ۱ متر بر ثانیه در یک مسیر مستقیم، با کمک ٤٠ درصدی ارتز زانو است. هدف از شبیه سازی ارزیابی اثربخشی و کارایی ارتز در کمک به مفصل ضعیف زانو است. نتایج شبیه سازی نشان می دهد که ارتز باعث کاهش بیش از ۱۳ نیوتن متر گشتاور مفصل زانو در یک فرد سالم می شود که نشان دهنده نیروی کمتر بر روی زانوی ضعیف است. علاوه بر این، ارتز حداکثر انرژی مورد نیاز در هر چرخه راه رفتن را کاهش می‌دهد که به معنی کارایی و استقامت بالاتر در چرخه راه رفتن است.