Development and Control of Concise Semi-Active Ankle Prosthesis

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Abstract: - The Concise Semi-Active Ankle Prosthesis is developed and controlled in this paper. Modern ankle prostheses are devices that exchange the absent limbs, making it possible for amputees to run again. Utilizing both parallel and series spring systems, the compact semi-active ankle prosthetic is created. The leaf series springs, parallel springs, and a cam make up the ankle prosthesis's compact, semi-active model. To minimize torque and power consumption as much as possible, the parallel and series springs must work as a nonlinear system. The concise semi-active ankle prosthetic was modeled using the CAD-CAM software. ANSYS was employed to implement a finite element analysis of the model. Static structure simulation is carried out with a loading force of 1500 N representing the personal weight. The model is imported into MATLAB/Simulink to be controlled after being run via ADAMS for dynamic analysis. The performance of the model with control is extremely close to that of the non-amputee ankle, according to simulation data. To test this model, the concise semi-active ankle prosthetic is manufactured as is modeled using the CAD-CAM program. The control system consists of a DC motor, Arduino uno, H-bridge, encoder, and current sensor. The motion of the compact semi-active ankle prosthetic is extremely comparable to the motion of the non-amputee, according to the simulation and testing results.

Key-Words: - Concise Semi-Active Prosthetic Ankle, control system, PI controller, Ansys, CAD-CAM, experimental result

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1 Introduction

The human leg is a magnificent machine that allows humans to do a wide range of activities, [1]. Approximately 2 million persons in the United States have had limbs amputated, with an estimated 185 000 amputees per year. One noteworthy example is major lower limb amputation, which affects more than 600,000 individuals in the United States, disproportionately affects underprivileged groups, and is predicted to double in occurrence by the year 2050, [2], [3], [4], [5], [6], [7]. Lower limb amputations are traumatic events that can significantly affect an amputee's physical, psychological, and social well-being. The loss of movement is the most obvious side effect after lower limb amputation. Amputees are more prone to falls and crashes since they cannot walk without some sort of support. Following an amputation, routine tasks that were previously taken for granted become significantly more challenging or impossible to complete, [7]. However, while prosthetic limbs can replace a missing part of the human body, their effectiveness is dependent on the patient's level of activity and health. The amputee may restore part of their typical capability by using a prosthetic leg to replace their amputated limb, [8]. The goal of this field's research is to enhance the artificial leg's features to more closely resemble the functionality of a human limb. Amputees may now walk more comfortably and use fewer energy thanks to prosthetic limbs that more closely replicate the functions of a human limb.

Ankle foot prostheses must be the same weight and size as a normal ankle. Electric motors and batteries are utilized to power the prostheses. At least the ankle prostheses must work for one day after full charging. The researchers should try to minimize the electrical motor's power needed. The energy efficiency of prostheses is very important. Because the stiffness of the ankle is variable, the passive elements are used. This passive element is used to store and return the energy, [8]. The passive component can be connected to the motor in series or parallel.

In [7], the authors described an effective powered ankle-foot prosthetic (PAFP) employing a series change stiffness actuator. The Springs with Gear Five Bars (SGFB) actuator to create the Series Elasticity Actuator (SEA), the SGFB mechanism was joined with the conventional SEA. The new SGFB actuator combines the benefits between the SEA and the GFBS actuators in terms of mimicking human ankle biomechanics and lowering motor maximum power. A power ankle prosthetic (PAFP) was actuated by an Elasticity Parallel Actuator (EPA). The direct drivetrain and EPA, which had been founded on the biomechanics of a typical ankle of a person, were used to measure PAFP's energy demands initially. The electromechanical design of the changeable compliance operator used in dynamic ankle prosthetics, as well as the research that was conducted with it is given in [9]. Two small inertia measurement units on the shank and foot collected data are inserted, [10]. To predict ankle energy, machine learning techniques were used, providing a viable alternative to traditional methods. Torsional fat springs were presented for the ankle of a person’s series elastic muscle tendon operator mechanism to deliver great deformation and compliance values, [11]. Because a torsional fat spiral spring can only create torque in one direction, the suggested torsional fat spring uses two torsional fat spiral springs in opposite orientations. In [12], a mechanical transmission dependent on a chain pulley enabling human-like actuators in human-scale prosthetic ankle joints was proposed. A power ankle-foot prosthesis, [13], was created to provide a large variety of motivations and sufficient force for a stride in which you force. A mechanical transmission based on a chain pulley was invented for human-like actuators in human-scale artificial ankle joints. The increasing effects of a prosthetic that actuates both the ankle and the knee joints, [14], were examined to the effects of ankle actuated alone. A 2-degree-of-freedom ankle-foot prosthesis with electromyographic control was developed to help amputees with their rock climbing, [15]. The passive elastic components were designed to minimize the number of actuators needed in power ankle prosthetics. The difficulty is to produce most of the non-linear ankle response with the passive component while keeping the torque for the ankle from the actuator low, [16]. A novel PAFP, [17], was constructed using a small parallel springs system. Actuators ‘torque and power requirements are reduced using the parallel springs mechanism, which consists of two linear springs and a cam. In [18], the force of friction was considered in the cam spring mechanism. A power biarticular transtibial prosthesis that combines a commercial PAFP with a controlled robotic knee orthosis was described, [19].

In [20], the authors described a hydraulic damping design for use in a foot with a trans-femoral prosthetic. The hydraulic prosthetic foot joint uses two one-way flow control valves to regulate the dynamic response in dorsiflexion and plantar flexion, respectively. A carbon prosthetic ankle combined with springs and a damper could help to better simulate natural foot movement.

Every year a lot of accident happens and most of them result in fatal injuries. One of the injuries is the amputation. This research is an effort toward a better solution for foot amputation. The CSAAP helps to achieve close to natural gait motion for an amputated person. The focus of this research is to design a new reliable and sustainable prosthesis, study the material properties to determine the strength, optimize the cost and actuator assembly, control the motion of the CSAAP model, and manufacture this model. The Concise Semi-Active Ankle Prosthetic (CSAAP) is modeled in this paper using a DC motor, leaf series springs, and parallel springs with a cam profile mechanism. When the heel strike cannot meet bandwidth demands, the leaf series springs are employed to raise the level of the stiffness of this spring that preserves the pulleys and belt transmission from damage. During the control dorsiflexion phase, the parallel springs with the cam profile mechanism are employed to maximize the stored energy and return it during the subsequent push-off stage. As a result, the motor's demand for power is reduced. This allows us to use a lower-powered motor. The cam's design reduces the required actuator torque. The energy efficiency of CSAAP is a key component in lowering their weight. The electric motor is reduced, and the batteries are downsized after employing the cam. The energy system's efficiency has improved, [8]. CAD-CAM is used to model the prototype of a CSAAP. The ANSYS program is used to do the static structure analysis for the model. After using four groups of material, one group is chosen because this group is suitable for the system and available in Egypt. The results showed that this model's mobility is extremely similar to that of a non-amputee ankle. The PI controller is used to control the motion in this model, [21]. The Signal constraint is used to optimize the parameters of the PI controller algorithm, [22]. The results presented that the torque and the power exhaustion of the motor were reduced. The required power from DCEM (118.5 watts) is less than the power of a non-amputee (300 watts) because the leaf series springs and parallel springs with a cam mechanism are used in this model. So, the weight of the battery is reduced.

The remainder of this document is as follows: Section 2 discusses the mechanical design of the CSAAP, Section 3 discusses the development of a finite element model, Section 4 gives simulation and control of CSAAP, the CSAAP model is shown in Section 5, and the whole paper is concluded in section 6.

2 Mechanical Design of the Concise Semi-Active Ankle Prosthetic

The CSAAP model, [23], is made up of active and passive components that work together to generate total positive power at the ankle joint during the stance stage of walking. Figure 1 describes the schematic, which was created with the help of a CAD-CAM program. In this design, the prototype of the CSAAP contains a carbon fiber foot, actuator, leaf series springs, and parallel springs with a cam profile mechanism. During heel strike, the carbon fiber foot is used to absorb shock. An electric motor, pulleys, a belt, and a ball screw make up the majority of the actuator. The CSAAP is actuated by a DC motor taking into account the demands of the actuator regarding torque, speed, and peak power. The DC motor's output is transferred to the ball screw via a pair of specially-made pulleys and belts. The pulley-belt transmission is being used to minimize the velocity of the DC motor. To make sure that the artificial limbs keep moving effectively and efficiently, maximum accuracy and sensitivity are needed in ankle joints, and a ball screw nut is being used. Pulley drives used for the rigid transmissions always perform well in terms of operation stability, dependability, and transmission efficiency, [24]. Also, the ball screw nut is used to transfer the rotational motion into linear motion.

The leaf series springs are used to increase the level of series elasticity that successfully protects the gearbox from destruction during heel striking when the gearbox is unable to fulfill bandwidth requirements, [16], [25]. A cam, follower, and spring element make up the nonlinear parallel springs system. The cam profile was designed in [16], so the nonlinear parallel springs mechanism can mimic human ankle dorsiflexion stiffness. A particularly constructed ramp section is integrated into the cam form, allowing the springs to be unloaded by turning the ankle sufficiently in the plantarflexion direction. A conventional pyramid adapter is also included on the gadget's top, which is used to connect the device to a patient's prosthetic socket. The model of the CSAAP is developed. After using the series and parallel spring and cam, the model is behaving as the non-amputee ankle. The amount of power used is reduced. The peak power of the motor P_{peak} is 118.5 watts whereas the reference peak power P_{ref} is 300 watts, [26].

$$P_{dec} = \left(1 - \frac{P_{peak}}{P_{ref}}\right) * 100 = \left(1 - \frac{118.5}{300}\right) * 100 = 60.5\% \quad (1)$$

So, the decrease in the power of the motor P_{dec} is about 60.5% as in equation (1).

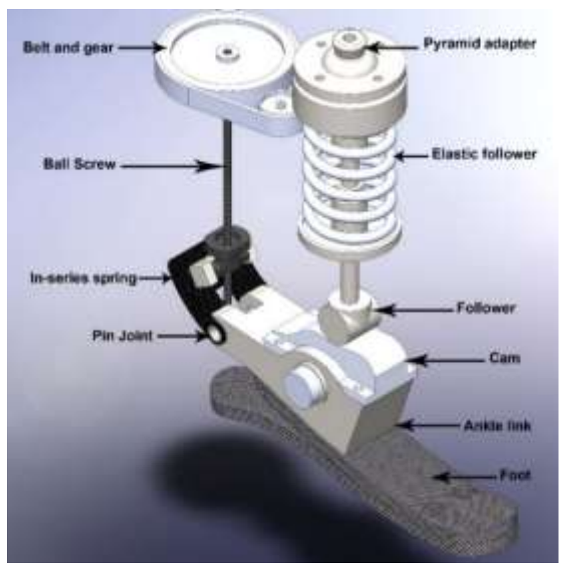

Fig. 1: the model of the PSAAP in CAD - CAM program

3 Analysis using the Finite Element Method

To evaluate the reliability, stress distribution, and deformation of the designed model. The FE Model is used to analyze constraints to assess the model's dependability, stress distribution, and deformation. The natural gait motion of the human foot during walking, as seen in Figure 2, is being studied to better comprehend the many stages of motion. These stages are determined over the entire stride of the foot, which includes complete dorsiflexion and plantarflexion to finish the step.

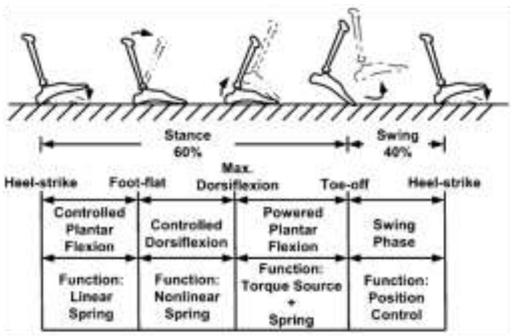

Fig. 2: Ordinary human ankle biomechanical when walking on level earth, [21]

After evaluating the stance and swing of the foot, there are three primary postures where foot stress points require attention. The three-foot postures that will be evaluated are heel strike, foot flat, and toe-off. A separate material is used to make each component of the prosthetic, [27]. Depending on the item's functionality and strength needs, these materials are assigned to it Table 1.

Table 1. The materials for each part

Part	Material
Foot base	PFR
The cam and follower mechanism	AL alloy
Ankle link	AL alloy
Series leaf springs	PFR
Screw and nut	Stainless Steel
cover	AL alloy

The default option for prosthetic meshing is used. The element order is programmed with a relevancy Centre, which works effectively and enables quick changes. The medium determines the mesh size. Figure 3 illustrates the tetrahedral mesh. Table 2 shows the total number of elements and nodes.

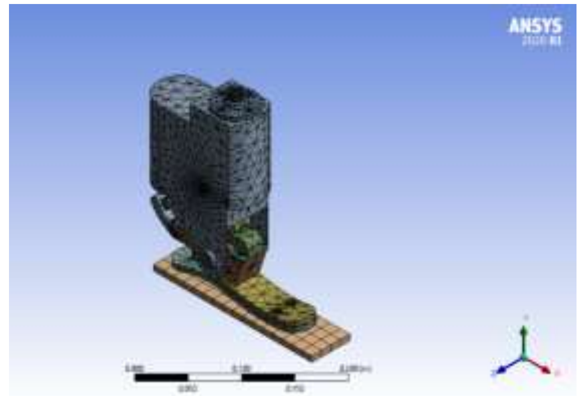

Fig. 3: The CSAAP model mesh in ANSYS

Table 2. No. of nodes and elements.

Nodes	53179
Elements	26616

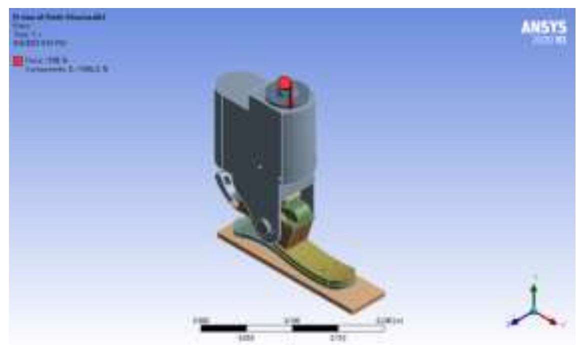

Fig. 4: The PSAAP model's vertical downward force (weight)

On the static structural tab, the boundary condition is set. The boundary condition is the entire load applied to the prosthesis. One of the burdens is the weight of the person who uses the prosthetic. There is persistence when the entire body's weight is on one of the feet when walking because a normal human being has two feet that split the body's weight. As a result, the simulation is built on this concept when a human's complete body weight is supported by one foot. A human being's average weight is 100 kg. With a 50% safety factor, the vertical downward weight works out to be 150kg. This load is modeled in ANSYS by providing a vertical downward force of 1500N on the pyramid adaptor, as exhibited in Figure 4.

The gravitational force on the housing is the second load on the prosthetic. The standard gravity of the earth is 9.8m/s2. As demonstrated in Figure 5, apply gravity to the housing in a negative Y direction.

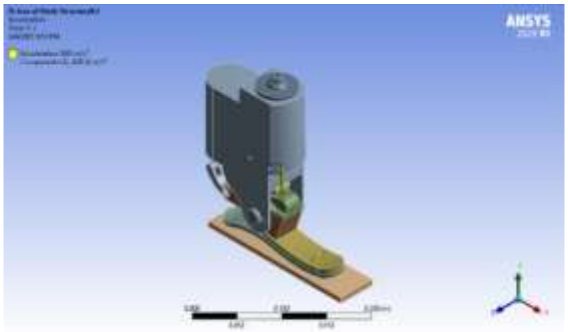

Fig. 5: Gravity force for the PSAAP model

3.1 Results of Finite Element Analysis

The results from the Ansys workbench for the three phases. The following are the outcomes of the three stages:

3.1.1 The Flat Foot

The first possibility to consider is the flat foot. A vertical force of 1500 N is delivered to the top of the prosthetic, as shown in Figure 4. The following solutions will be driven by the simulation: Figure 6 and Figure 7 show the equivalent stress and total deformation, respectively. In this stage, the highest equivalent stress is 0.23777MPa, which is less than the equivalent stress of the weakest material. $4.8247*10^{-5}$ m is the overall deformation, which is quite little.

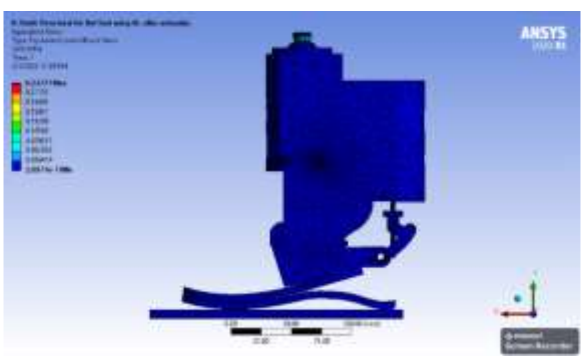

Fig. 6: Equivalent stress during the flat foot stage

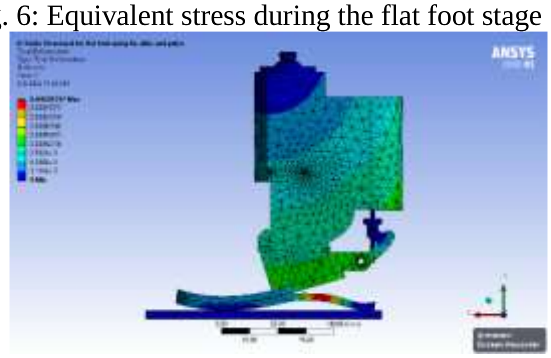

Fig. 7: Total deformation during the flat foot stage

3.1.2 Heel Strike

The second case study, as shown in Figure 8, is for the heel strike phase of the prosthetic's motion. The heel strikes the ground first in a normal gait motion, at this point, all of the body weight is on the foot region, and the heel part compresses to store elastic strain energy, which releases when the foot is lifted off the ground, thus, the heel design aids in the walk while converting potential energy into kinetic energy. Equivalent stress (as shown in Figure 9) and total deformation (as shown in Figure 10) are the following solutions. In this stage, the highest equivalent stress is 0.18152 MPa, which is less than the equivalent stress of the weakest material.

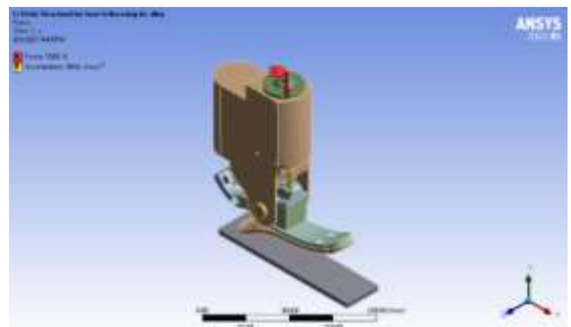

Fig. 8: The CSAAP model during the heel strike

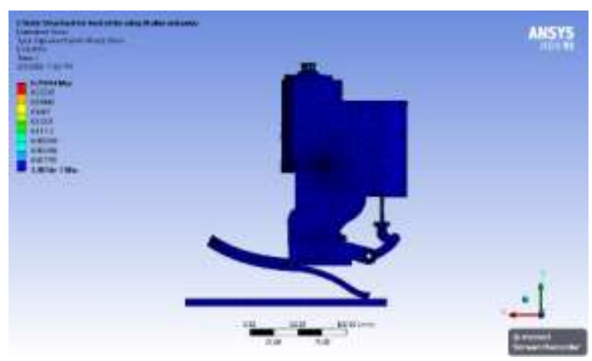

Fig. 9: Equivalent stress during the heel strike stage

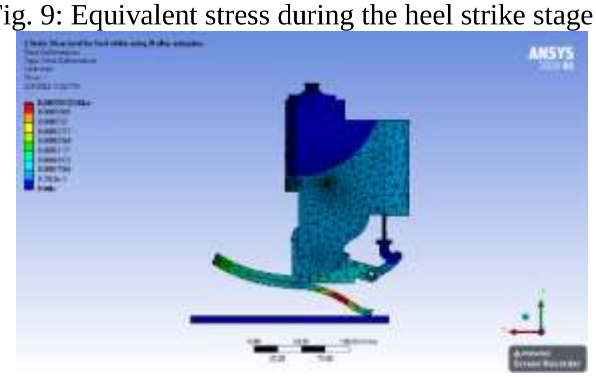

Fig. 10: Total deformation during the heel strike stage

3.1.3 Toe Off

The next case study focuses on the toe-off phase of the prosthetic motion (Figure 11). When the entire walking cycle is in its final phase, the entire body goes on a toe in a typical gait action. The stored potential energy is turned into kinetics at this stage, assisting the actuator's triggered push for lift-off. Equivalent stress (as shown in Figure 12) and total deformation (as shown in Figure 13) are the following solutions. In this level, the highest equivalent stress is 0.13598 MPa, which is less than the equivalent stress of the weakest material. The total deformation is relatively minimal at 3.5899*10-4 m.

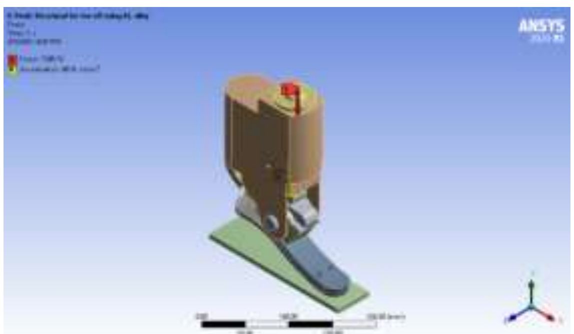

Fig. 11: The CSAAP model during the toe-off stage

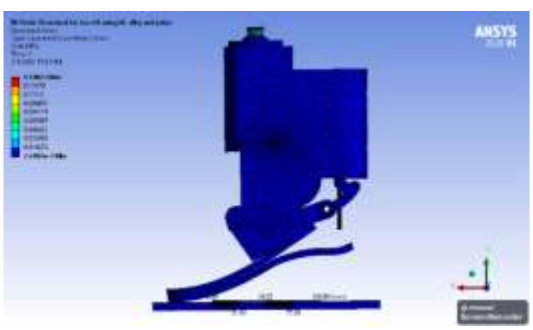

Fig.12: Equivalent stress during the toe-off stage

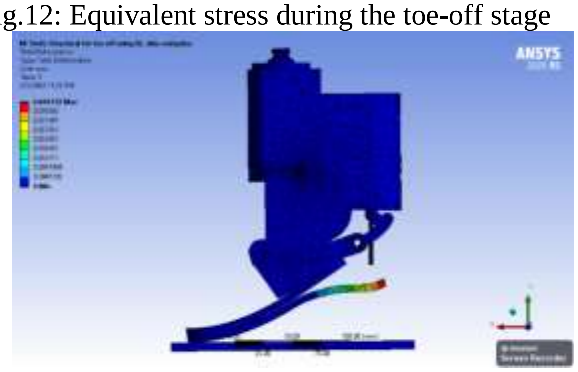

Fig. 13: Total deformation during the toe-off stage

4 Control of Concise Semi-Active Ankle Prosthetic

The control system of the CSAAP was designed and implementation by using MATLAB/SIMULINK and a DC motor. After the CSAAP was modeled by using CAD-CAM, the model needs to be controlled. Before control, the analysis of dynamics must be done. As seen in Figure 14, the CSAAP prototype is loaded into ADAMS for the analysis of the dynamics. The main imported prototype into ADAMS consists of the ground, pieces, joints, and so forth. As seen in Figure 15, the prototype of the CSAAP from ADAMS is imported into MATLAB/Simulink, and the control simulation system's block diagrams are created using the PI control algorithm. By integrating it with other control modules in MATLAB and Simulink, the control simulation system block diagram is created, [22]. The control system of the CSAAP begins with the reference speed. The reference speed is compared with the feedback speed of the DC motor. The error between the reference and feedback speed is entered into the first PI controller algorism. In the co-simulation between MATLAB/Simulink and ADAMS, the tuning parameters of PI controller algorism cannot be done by the PI block alone. So, the parameter of this PI controller is tuned by the signal constraint, [22]. The signal constraint is a block in MATLAB/Simulink that is used to tune the parameter of PI controller algorism. The reference current is produced by this PI controller. The DC motor's feedback current and the reference current are compared. To account for the discrepancy between the reference and feedback currents, a second PI controller method is modified. The voltage comes out of the second PI controller. This voltage is inputted into the DC motor model. The DC motor's specifications are listed, [28]. In MATLAB/Simulink, a block diagram of the control system of the CSAAP virtual prototype is created based on the PI control algorithm, as shown in Figure 16 and Figure 17.

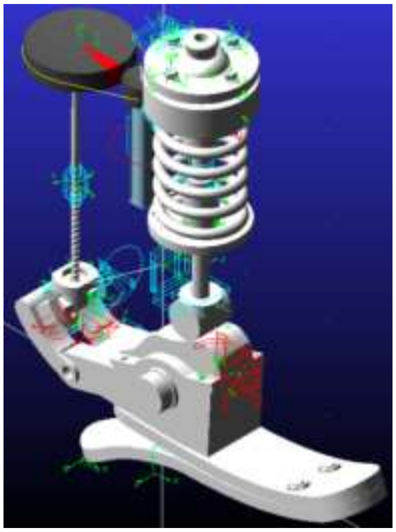

Fig. 14: the model of the CSAAP in the Adams program

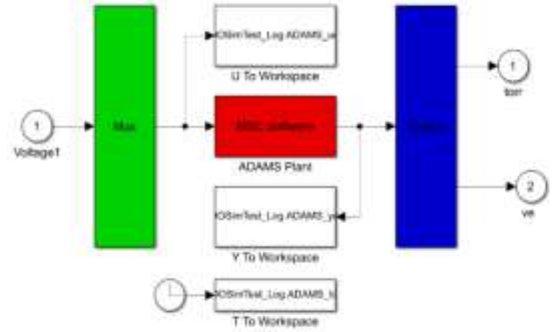

Fig. 15: Control plant generated by ADAMS

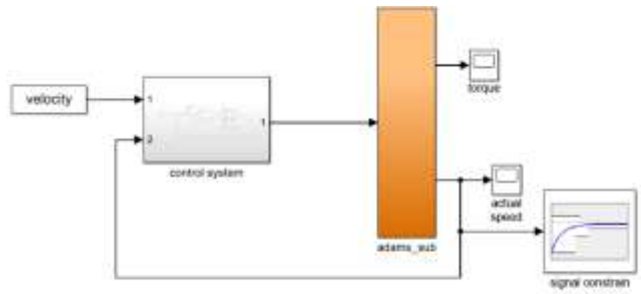

Fig.16: Simulink model of PI control algorithm

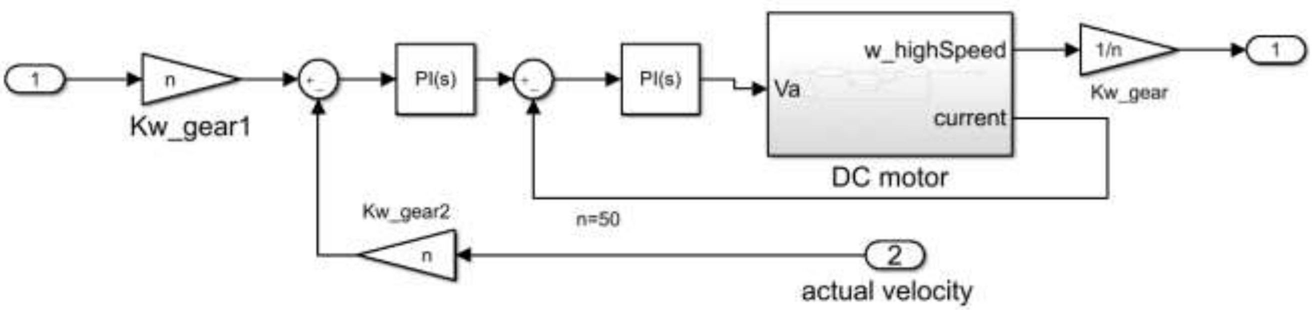

Fig.17: the control system of the PI control algorithm

4.1 Simulation Results

The simulation of the CSAAP is very important before manufacturing the CSAAP model. This model's simulation is done using CAD/CAM and MATLAB/Simulink programs. The results of the CAD-CAM (without control) and MATLAB/Simulink (with control) simulations for gait evaluation of the CSAAP during level-land walking are presented in this part. The source of the reference data is in [29]. From CAD-CAM and MATLAB/Simulink, Figure 18 depicts the motor's power. The motor is turned off during the controlled dorsiflexion CD stage because the Parallel spring element reproduces the desired torque. The advantages of this design are that it reduces the motor's operating time and the amount of energy used on the motor windings. It's worth noting that the CD stage's negative mechanical energy is stored in the parallel spring element. In the Powered Plantar Flexion PP stage, the energy stored in the Parallel spring is released to help the motor advance the user. In comparison to a reference peak power of 300 watts, the motor's max output is 118.5 watts, [29]. The motor's power is consequently decreased by roughly 60.5 percent. In the SP, the prosthesis is returned to its equilibrium point while utilizing very little electrical power. The reference torque, [29], the ankle torque from CAD-CAM, and the ankle torque from MATLAB are all quite close in Figure 19. The difference between CAD-CAM and MATLAB/Simulink may be due to the delay of feedback signals. The motor torque from CAD-CAM and MATLAB/Simulink is shown in Figure 20. Pulley-belt transmissions may provide a maximum motor torque of 58N.m. The torque produced by the Parallel spring, as shown in equation (2), is the difference between ankle torque and torque from pulley-belt transmissions.

$$T_{ac} = T_r - T_{kp}, \tag{2}$$

Where T_r $(N.m)$ is the reference torque, T_{ac} (N.m) is the actuator that provides the active torque, and T_{kp} $(N.m)$ is the produced torque by the parallel spring.

Figure 21 displays the motor speed as an input speed signal to the model, as well as motor velocity from CAD-CAM and MATLAB/Simulink. The MATLAB/Simulink velocity is extremely close to the input signal. The position for the non-amputee ankle and the position for SAAFP from CAD-CAM and MATLAB are shown in Figure 22. In the first 60% of the gait cycle, the MATLAB posture for SAAFP is quite similar to the reference position, [29]. Figure 23 displays the ankle torque-angle behavior for an 84.4 kg person using CAD-CAM and MATLAB/Simulink. Because CAD-CAM ankle torque-angle behavior is uncontrollable, there are variances between them. The ankle angle-torque behavior in MATLAB/Simulink is remarkably similar to the reference ankle angle-torque behavior in [30]. The ankle angle-torque behavior will be the same as the reference ankle angle-torque behavior when a high-performance control system is utilized. Points (A), (B), (C), and (D), respectively, depict the phases of the foot as shown in Figure 23 at heel strike, foot flat, maximal dorsiflexion, and toe-off. The gait phases (CP, CD, PP, and SP) are represented by the segments (A)-(B), (B)-(C), (C)-(D), and (D)-(A), which correspond to the ankle torque-angle behaviors during those phases. The human ankle joint has different spring inclinations during CP and CD, as can be seen in segments (A)-(B) and (B)-(C), respectively. Figure 24 depicts the motor's power under various loads. This demonstrates that the SAAFP can be utilized for people of various weights. The highest power of the motor is 118.5 watts when the weight of an amputee human is 83.3kg. When the amputee's weight reaches 100 kilograms, the motor's maximum output is increased to 160 watts. When the amputee's weight reaches 125 kg, the motor's maximum output rises to 242 watts. As a result, the SAAFP model can be used for a wide variety of human amputee weights. Figure 25 depicts the motor power at various speeds. It has been demonstrated that the SAAFP can be utilized for walking or running at various speeds. The motor power must be increased as the velocity increases.

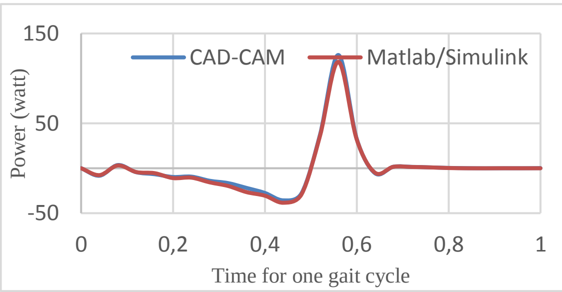

Fig. 18: The motor power (watt) from CAD-CAM and MATLAB/Simulink

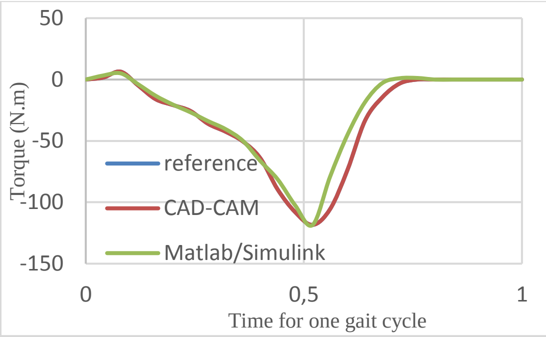

Fig.19: the reference torque for the ankle and the torque for the ankle from CAD-CAM and MATLAB/Simulink

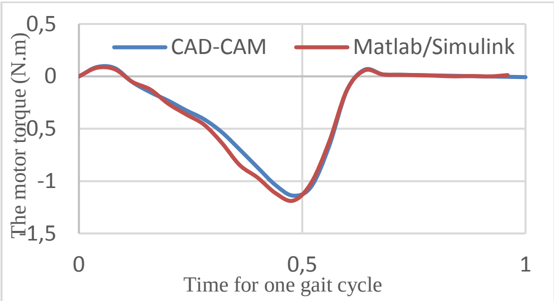

Fig. 20: The torque for the motor from CAD-CAM and MATLAB/Simulink

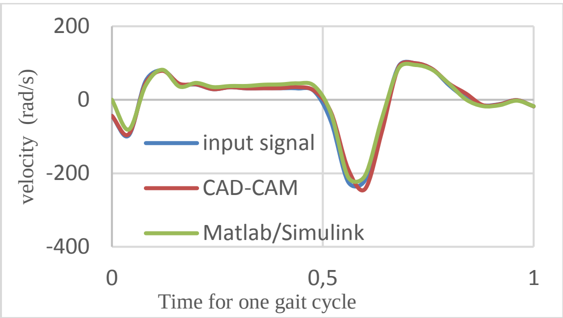

Fig. 21: The input signal and the velocity of the motor from CAD-CAM and MATLAB/Simulink

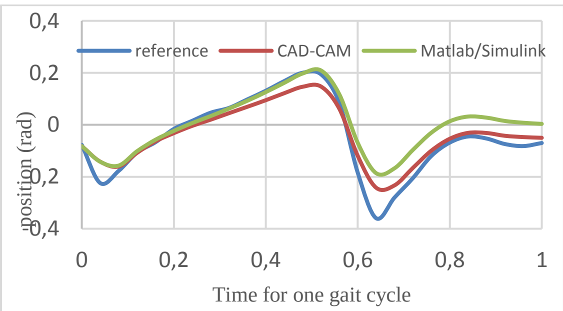

Fig. 22: the reference position and position from CAD-CAM and MATLAB/Simulink for the SAAFP

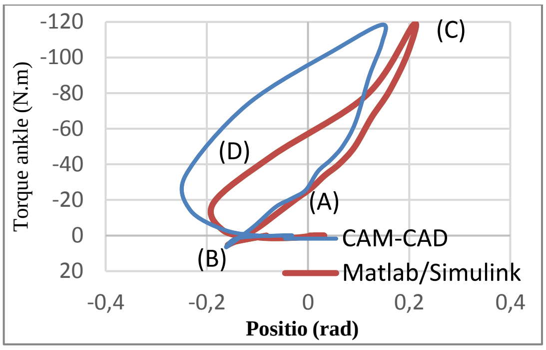

Fig. 23: the ankle torque–angle behavior of the prosthesis for one gait cycle from CAD – CAM and MATLAB/Simulink

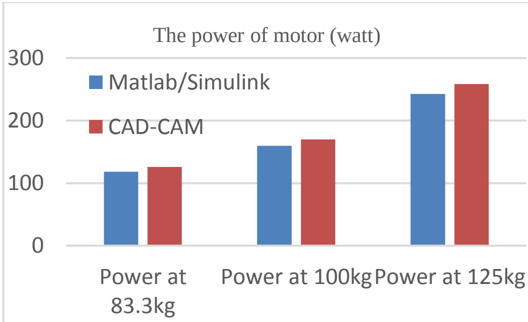

Fig. 24: the comparison of the power of the motor under different loads

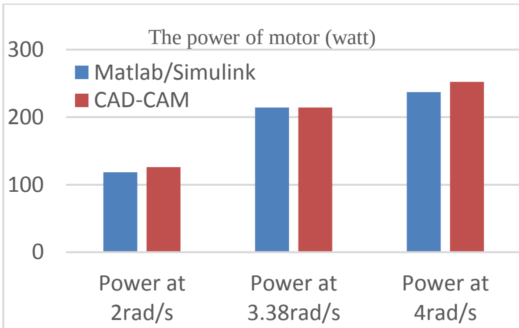

Fig. 25: the comparison of the power of the motor under different velocities

5 Concise Semi-Active Ankle Prosthetic Practical Model (CSAAP)

The practical model of the proposed CSAAP model is assembled and controlled in this section. The result is shown that the motion of the CSAAP is very similar to the motion of the non-amputee.

5.1 Experimental Setup

The practical CSAAP model is very important to prove that the simulation of the CSAAP model is useful for amputees. The material of each part is chosen according to what is available in Egypt and is good for the model. The mechanical parts are manufactured by using a CNC machine. Each part is manufactured from material according to Table 3 based on the EF and the available material in Egypt. The foot base and series leaf springs are manufactured from PFR material because carbon fiber is not found in EGYPT markets.

Table 3. the materials for each part

Part	Material
Foot base	PFR
The cam and follower mechanism	AL alloy
Ankle link	AL alloy
Series leaf springs	PFR
Screw and nut	Stainless Steel
Cover	PLA

The CNC machine to manufacture the parts needs each file of the parts for the model. Figure 26 displays each part after manufacturing. The CSAAP is assembled as shown in Figure 27. It is shown that the practical model is the CSAAP model in Figure 1.

A) The follower and the parallel spring

B) The foot parts

C) The cam and its base parts

D) The nut part

E) The bolt and one of the pulleys

F) The series spring part

G) The cover part
Fig. 26: Manufactured parts

Fig. 27: the semi-active ankle prosthesis

5.2 Control System

The PID control algorithm is used to control the CSAAP model as shown in Figure 16 and Figure 17. this control is executed by using ARDUINO UNO.

The CSAAP model needs to 200-watt DC motor. The Maxon 200-watt brushless DC motor is suitable because of its power and lightweight. This motor cannot be used because of costly. In the experiment, the 200-watt DC motor is not needed because the personal weight is not used. So, the small motor can be used with small personal weight. The encoder measures the motor speed. The current sensor is used to measure motor current and H-bridge is used for motor deriving.

The control system begins with the reference speed. The reference speed is compared with the actual speed of the DC motor. Two cascade PI-controller is designed for motor speed and current control. The parameter of this PI-speed controller is tuned by the signal constraint, [22]. The reference current is produced by this PI controller. The DC motor's actual current and the reference current are contrasted. The inner loop PI-current controller receives the error between the reference and the real current. The reference motor voltage is produced by the second PI controller. According to the positive or negative of the reference speed, the direction of the DC motor is specified if it is forward, backward, or stopped. The schematic of the control circuit is displayed in Figure 28. The program as the control is written in the Arduino program.

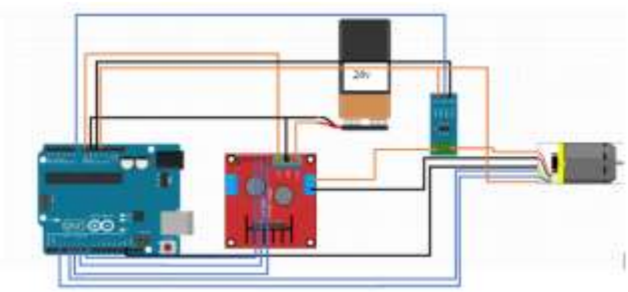

Fig. 28: The schematic of the control circuit

6 Practical Results

The model of CSAAP is connected to the control circuit. Then the CSAAP is operated on to know if it simulates the motion of the ankle for non-amputee. The CSAAP is taken as a video. Then this video is divided into many images. Figure 29 displays the stages of the gait cycle. These stages consist of the heel-strike, foot-flat, and Max. Dorsiflexion, toe-off, and heel strike again. The Figure displays that the gait cycle of this model is very similar to the gait cycle of a non-amputee.

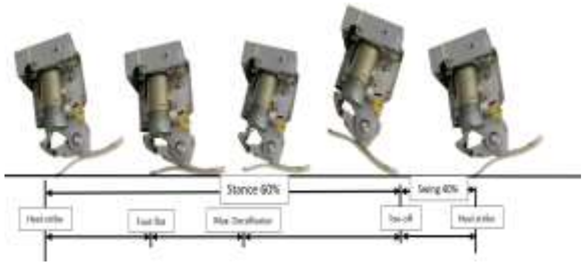

Fig. 29: The images of the CSAAP model from the video

7 Conclusion

The CSAAP with series springs and a non-linear Parallel spring mechanism was developed. The non-linear series and Parallel spring were realized by using the cam. The non-linear behavior of series and Parallel spring contributed to minimizing the power consumption and torque of the motor. The SAAFP with series springs and non-linear Parallel spring mechanism was modeled by using the CAD-CAM program. The model of the CSAAP is imported into ANSYS Workbench by using CAD Configuration Manager. An ANSYS Workbench was used to perform a finite element analysis of the CSAAP model structure. Static simulation is carried out with a loading force of 1500 N representing the amount of the personal weight. Finite element analysis showed that the CSAAP model can be successfully used for personal 100 Kg weight. The SAAFP was imported into ADAMS for dynamic analysis. In MATLAB/Simulink, the SAAFP virtual prototype model from ADAMS is imported. The CSAAP is manufactured. The Arduino, H-bridge, encoder, and current sensor are used to control the model practically. The results displayed that the power consumption and the torque of the motor were minimized.

The performance of this model can be summarized as:

1-the motion of the CSAAP is very similar to the motion of a non-amputee.

2- the required power from the DC motor (118.5 watts) is less than the power of a non-amputee (300 watts) because the series and parallel spring with the cam mechanism is used in this model.

3- The CSAAP model has the same size as the normal ankle.

4. This model weighs about 1.8 kg, which is less than the predicted weight of a severed limb of

roughly 2.3 kg. The DC motor employed in the current study has a large weight since the necessary DC motor, which weighs less, is highly expensive.

In future work, a more effective control system for the CSAAP will be done. The ANSYS program will be used in transient analysis to improve the dynamics of the CSAAP model. The design of the CSAAP will be developed.

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