Design of an Artificial Foot using Series Elastic Actuator

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Received: 15 March 2014, Revised: 12 June 2014, Accepted: 30August 2014

Abstract: Investigation of natural foot behaviour shows that it's kinetic and kinematics functions can be compared to mechanisms containing simple mechanical elements such as springs, dampers and actuators. Hence, appropriate design of this mechanism may give rise to an artificial foot that might help amputees properly. Studies show that whatever the biomechanical behavior of an artificial foot is similar to that of natural foot, where it might be utilized consuming less energy. In this study, required parameters for design of an artificial foot including spring and electromechanical actuator were examined by investigating biomechanical behavior of natural foot in walking level. Moreover, an algorithm was suggested for design of foot prosthesis and according to this algorithm, a prototype was designed and fabricated.

Keywords: Artificial Foot, Series Elastic Actuator, Walking Gait, Biomechanical

Reference: Kheirollahi, H., Goudarzi, A. M., Fazeli, M., and Imani, M., "Design of an Artificial Foot using Series Elastic Actuator", Int J of Advanced Design and Manufacturing Technology, Vol. 7/ No. 3, 2014, pp. 83-91.

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1 INTRODUCTION

The statistical studies indicate that diseases, accidents and wars are the principal causes to increase the number of amputees in the world. These studies also show that most of mutilation disabilities are related to musculoskeletal. Aside from the magnitude of severity of aforementioned causes in increasing disabilities, a well-developed planning is needed to support amputees. Therefore, the rehabilitation activities became more desirable than the past. The objective of this study is to propose a proper solution to improve the life quality of foot amputees.

Foot prostheses can be classified based on their power source as passive, active and semi-active prostheses. In passive foot prosthesis the power is supplied by amputee muscles and controlled by passive components of the prosthesis. In this case, the energy usage of prosthesis in a well-defined condition can be more than two other types. This type of prosthesis is the earliest one and is very common in market.

Active foot prosthesis is designed to supply necessary torque in each foot angle, similar to an intact foot. This type of prosthesis needs a considerable amount of energy. In addition, in order to be more applicable in human activities, this type of prosthesis must instantaneously react at different speeds whether low or high speed. In high speed activity, not only makes some problems in controlling motors or actuators, but also increases the required energy for motion of the apparatus. Using low weight and portable actuators is one of the important factors in design of active foot prostheses. Heaviness is the greatest disadvantages of prosthesis functioned by hydraulics, pneumatics and artificial muscular actuators.

By now, numerous studies were performed to develop an appropriate active prosthesis for foot amputees. Stein & Flowers developed a foot prosthesis moved by hydraulic actuators [1]. Popovic & Schwirtlich designed a foot prosthesis moved by a battery [2]. Another prosthesis developed by Klute et al., was moved by pneumatic actuator [3], [4]. Sup et al., introduced an improved foot prosthesis moved by pneumatic actuators [5]. Versluys et al., purposed a design of foot prosthesis using artificial muscle [6].

In semi-active foot prosthesis, the principal idea is to manage correctly the amputee's muscle forces by means of a controllable mechanism. Control of the mechanism requires an external power source just to activate the control devices. Using the amputee's muscle forces, caused by semi-active foot prosthesis, significantly reduced the indispensable external power in comparison to active foot prosthesis. This study finds its originality in this context of foot prosthesis. To this purpose, Bellman et al., designed an artificial foot functioned by an electrical motor linked to a linear spring as tendon to produce proper torque at foot joint [7]. By analyzing the mechanical behavior of limbs in a healthy individual during a walking gait cycle, the behavior of foot can be determined. The output data can be used for an artificial foot design. To replicate the foot behavior, a controllable mechanical design is needed. For this purpose, spring components are used for saving and consuming mechanical energy during a gait cycle and a controllable electromotor is used to regulate spring stiffness according to its requirement.

In this study, we have simulated the behavior of the proposed model similar to the real world case in order to determine the design parameters of prosthesis. For this purpose, the mechanical behavior of designed prosthesis was simulated to accomplish an optimal design. The optimal conditions are producing the required torque in the end of stance phase of each gait cycle and also reducing the overall work in the gait cycle [8]. With respect to the acquired results from simulation, an empirical prototype was fabricated and it's mechanical and control behaviors were tested in a gait cycle condition to validate the results of virtual simulation.

2 MOTION ANALYSIS OF FOOT IN WALKING GAIT

The definition of "foot angle" is shown in Fig. 1. The human foot angles, measured on a walking gait condition, were reported by several references that each of them specifies the behavior of a group of people in a given walking condition [9], [10], [11].

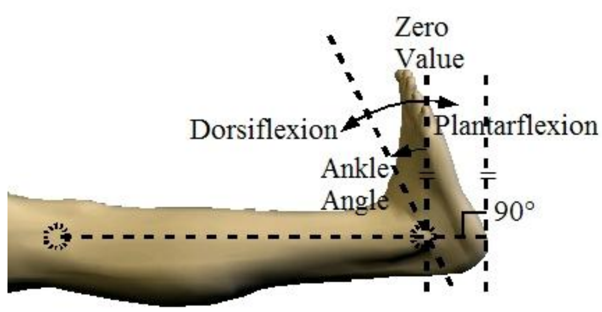

Fig. 1 Definition of foot angle

During a walking gait cycle, two principal phases can be distinguished which represent the mechanical behavior of a foot joint:

1) **Stance (St) phase**: During this phase, the foot is in contact with ground and is adopted by body forces as well as ground reactions. The applied constraints are altered in accordance with foot position. By considering the foot situation, the stance phase can be divided into three other sub steps classified as;

Controlled Plantar Flexion (CP), Controlled Dorsi Flexion (CD) and Powered Plantar Flexion (PP).

2) **Swing (SW) phase**: In this phase, the foot is separated from ground and in absence of the ground contact forces, the foot rotation can be directed by a negligible muscular power applied to foot joint. The swing phase is accomplished when the heel strikes to ground.

By considering the hip, knee and foot angle and body mass distribution of an individual and by neglecting the joint's friction effects, the human's foot torques in a given gait condition can be calculated [12], [13]. Based on data reported in [13], the biomechanical behavior of a foot can be simulated by simple mechanical components as follows:

a) **CP phase**: In this phase, foot absorbs energy and correlation between torque and foot angle is quasi linear (Fig. 2). According to Table 1, whereas the work done in this step is negative, an amount of energy must be wasted or saved. So the desired prosthesis can achieve to this step using an appropriate linear spring.

b) **CD phase**: In this phase, foot absorbs energy and correlation between torque and foot angle is nonlinear (Fig. 3). So, the desired prosthesis can accomplish this phase using a nonlinear spring or a combination of spring and damper element. Using a nonlinear spring to save energy is more appropriate, because in the next phase an amount of energy must be released and it can be supplied from the energy saved in this phase. A nonlinear spring with a variable stiffness can be realized by a combination of an actuator series by a spring which is known as "series elastic actuator".

c) **PP phase**: In this phase, released energy of foot and correlation between torque and foot angle is quasi linear (Fig. 4). Although the done work in this phase is positive, it is greater than the sum of energy that can be saved in the earlier phases. So, to achieve this step, desired prosthesis need an external source of energy in addition to what is saved in previous phases. By considering these conditions, the PP phase can be modeled by a combination of spring and actuator.

d) **SW phase**: Whereas in swing phase, the foot is not in contact to ground, the imposed force on foot is created by its inertia (Fig. 5). So, by an appropriate design and constraining paw by a simple revert spring, it can be expected that after swing, paw is placed on its appropriate position in starting of stance phase without imposing any external force.

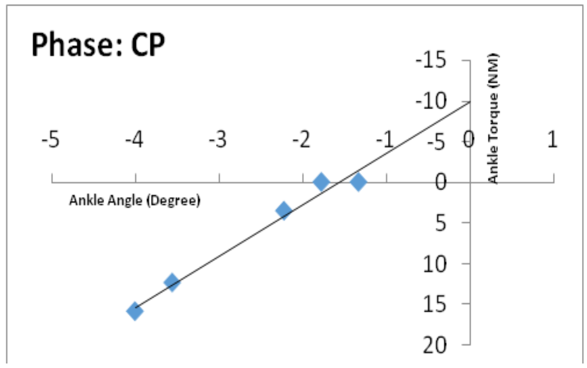

Fig. 2 Torque versus angle in CP phase

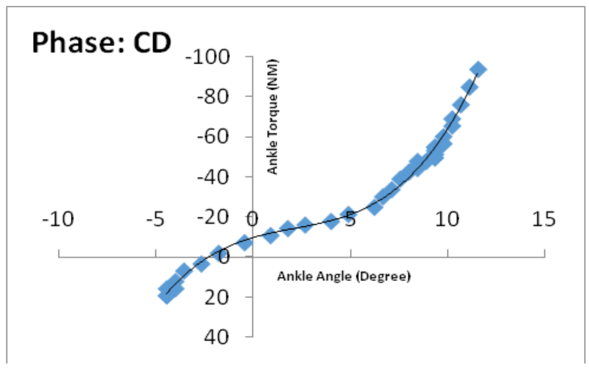

Fig. 3 Torque versus angle in CD phase

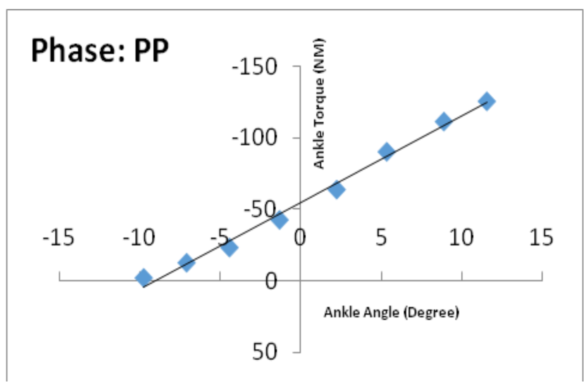

Fig. 4 Torque versus angle in PP phase

Table 1 Required pure work of healthy foot for fulfilling each step in walking gait (J)

Required work in CP step (W_{CP})	Required work in CD step (W_{CD})	Required work in PP step (W_{PP})	Required work in Sw step (W_{sw})	W_{PP} - ($\lvert W_{CD}\rvert$+ $\lvert W_{CP}\rvert$)
-0.27	-11.7	+15.2	~0	+3.25

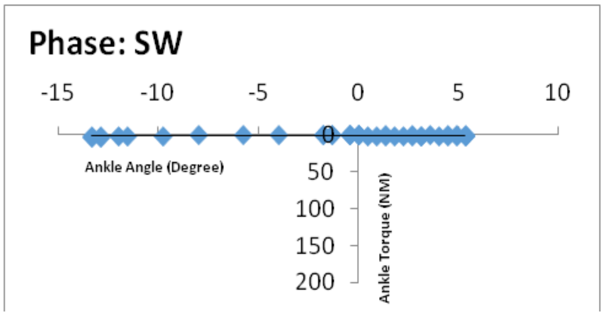

Fig. 5 Torque versus angle in SW phase

Figure 6 shows torque versus angle of a foot during a gait cycle and Table 1 shows the required muscular pure work of foot joint to achieve each step in a walking gait. By considering the cited data, it can be concluded that the foot joint is subjected to the significant torques during stance phase versus the swing phase. Therefore final design of prosthesis must satisfy the serious mechanical conditions due to the stance phase of walking gait.

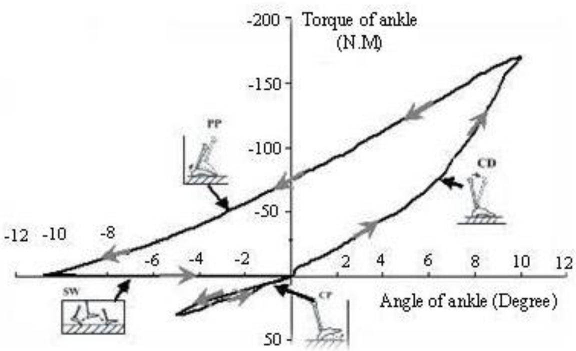

Fig. 6 Torque versus angle of foot during walking gait

Based on the provided technical solutions, a conceptual prosthesis equipped with several springs and a controllable actuator can well satisfy the low energy requirement of an amputee during a walking gait cycle. The proposed foot prosthesis mechanism is composed of two torsional springs that one of them is series to an actuator. One spring acts in CP phase and another one acts in CD phase. Assuming that mechanical behavior of a healthy foot is equivalent to its artificial replacement, the goal is choosing appropriate and optimum elements for prosthesis design.

3 DETERMINING OF OPTIMAL DESIGN PARAMETERS

To evaluate the design parameters of desired prosthesis during a walking gait condition, the mechanical effects of related limbs have to be considered. Figure 7 shows a two dimensional model representing mechanical behavior of a foot in sagittal plane. This model is composed of three actuators placed respectively on hip, knee and foot joints. In the healthy model, all joints are free of constraint, but in injured model, except the foot joint which is restricted by two torsional springs, the other joints do not have any constraint.

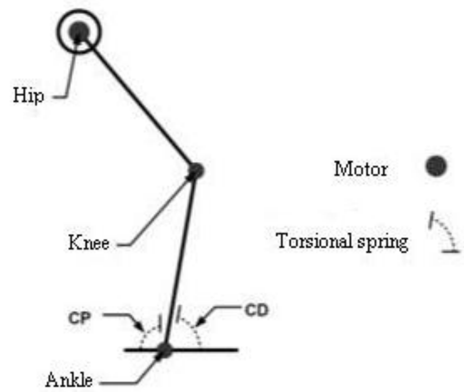

Fig. 7 Two dimensional model of a foot composed of mechanical elements

4 GUIDLINE OF OPTIMIZATION

The actuators of hip and knee must be governed so that these joints represent the same kinematic behavior as the healthy joints. Also the foot prosthesis has to be controlled so that it follows the kinematic behavior of healthy foot using minimum external energy. Regarding to the reported torque, angle, and power quantities for three foot's joints [13], a fitting function is composed in which the energy consumption of foot prosthesis is compared with the healthy foot. By minimizing this function, the optimal design parameters can be evaluated. This function named "fitness" can be defined as follows:

$$fitness=\sum_{\text{Joint}}\sum_{j=0}^{t_{Gc}-1}((P_{bi}(j+1)-P_{mi}(j+1)).(t_{j+1}-t_j))^2$$

$$t_j=\frac{j}{t_{Gc}}t_{Gaitcycle} \quad , \quad t_{Gc}=100 \tag{1}$$

Where:
"j" is the sequential number of time sector.
"$t_{Gaitcycle}$" is the total time of an overall cycle.
"t_{Gc}" is the number of time sector in a gait cycle.
"t_j" is the beginning moment of j^{th} time sector.
"P_{bi} $(j+1)$" is the power of joint (i) in healthy foot which is measured at $(J+1)^{th}$ time sector.
"P_{mi} $(j+1)$" is the power of joint (i) in injured foot which is measured at $(J+1)^{th}$ time sector.

To simplify the power calculation, it will be considered that angular velocity of a joint is constant across each time sector. Thereby Eq. (2) represents the power imposed to joint (i) at j^th time sector.

$$\begin{cases} P_{mi} = \tau_i.\dot{\theta}_i \\ \tau_i = k_i.\theta_i + b_i.\dot{\theta}_i + \tau_{mi} \end{cases} \tag{2}$$

Where, "$\boldsymbol{\tau}_i$" is the total torque applied to joint (i), "θ_i" and "$\dot{\theta}_i$" are respectively angular rotation and angular velocity of joint (i), "τ_{mi}" is the portion of torque imposed by motor of joint (i) and "k_i" and "b_i" are respectively stiffness and damping factors of joint (i).

Assuming that hip and knee joint do not have any constraint in mechanical model, their stiffness and damping factors are negligible and total required torque has to be provided by their own actuators. In this case, if the powers of actuators of hip and knee joints in mechanical model are considered equal to what are applied by musculoskeletal, on relevant joint of healthy body, their resultant fitness function are nil. As a result, the fitness function is just depended on behavior of foot joint. So, for the equal interval of time, Eq. (1) can be simplified as follows:

$$fitness = \min\left(\sum_{j=0}^{t_{Gc}-1}((P_b(j+1) - P_m(j+1)).\Delta t)^2\right) \qquad \Delta t = t_{j+1} - t_j \tag{3}$$

Where, "*P_b (j+1)*" and *"P_m (j+1)"* are respectively the power applied to healthy foot and mechanical foot, across *(J+1)th* time sector. Whereas the artificial foot joint is modeled by two springs and an actuator without damper, the joint power defined by Eq. (2) is accomplished by a combined effect caused by an actuator torque and the spring torque. In this case, there are two factors to realize the required power for foot prosthesis motion, one is the actuator power and another one is spring stiffness. So, by finding the optimal spring stiffness, the smallest value of actuator power can be obtained, while the corresponding foot joint fitness function is minimized. By considering the periodic motion of foot during each gait cycle, the stored energy in springs across one step of gait cycle will be released.

5 INVESTIGATION OF OPTIMUM ELEMENT RESULTS

Regarding the mentioned criteria to obtain the optimum design of foot prosthesis, two designs were considered. In the first design, the foot has no actuator and is composed simply of two springs acting in planar flexion and dorsi flexion conditions. In the second design, the mechanism was improved by adjoining a controllable actuator to the corresponding spring of dorsi flexion. By using some appropriate commercial software, the mechanical behaviors of designed systems versus the healthy foot during a walking gait were simulated (Fig. 8) and optimal design for foot components were characterized. By considering a suitable linear spring to achieve CP phase, three options are studied for other foot spring. They are respectively, linear springs, 2^nd, and 3^rd order nonlinear springs. In each option, the optimal spring designs which accomplish the "fitness" condition during a walking gait were calculated.

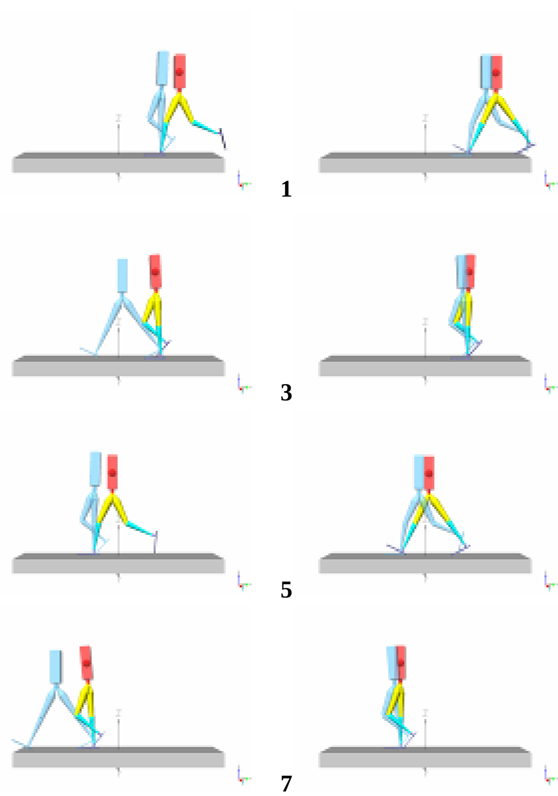

Fig. 8 Designed foot compared to healthy one

Figures 9 and 10 show foot torque and power during a walking gait cycle caused by each of these experiences. As shown in Fig. 9, before the torque reaches to its maximum amount, a foot prosthesis with a second order spring ($F=kx^2$) introduces roughly the same torque as healthy foot. Afterwards, the two foot torque charts are spread. During a walking gait, the maximum foot torque occurs when the foot power becomes positive in the end of CD phase.

To reduce the torque difference between second order spring prosthesis and normal foot, the prosthesis design is accomplished by joining a controllable actuator with CD spring to convert it to a virtually variable stiffness spring. As the actuator is able to produce appropriate offset torque to compensate eventual defects, regardless of the spring coefficient, the corresponding "fitness" function can be achieved by precise real time controlling of the corresponding actuator (see Figs. 9 and 10).

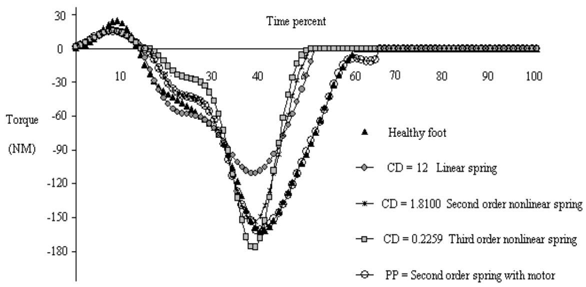

Fig. 9 Foot torque during a walking gait tested on different designs

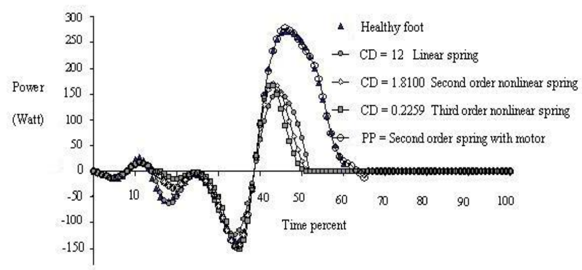

Fig. 10 Foot power during a walking gait tested on different designs

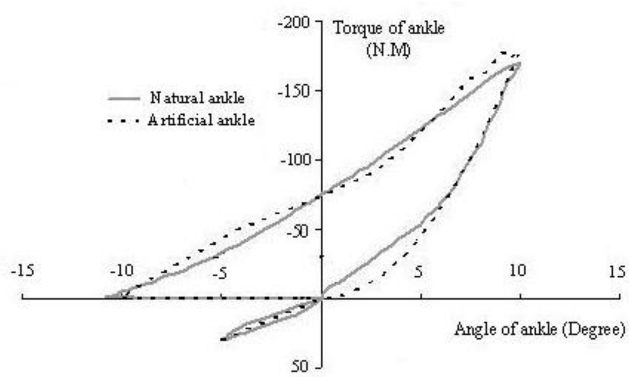

Fig. 11 Torque versus angle of foot prosthesis compared to that of healthy foot

Table 2 shows the mechanical specification of components of such defined series elastic prosthesis, to accomplish the best torque adaptation with a healthy foot during a walking gait cycle. Figure 11 compares foot torque versus angle caused by such series elastic actuator designed prosthesis with regard to the healthy foot behavior. As shown in the simulated model, the defined artificial foot has good agreement to the behaviors of a natural limb.

Table 2 Mechanical requirement for a series elastic actuator designed foot with second order nonlinear spring

Phase	Actuator	Spring
CP	----	K_{CP}=4.5 N.m/deg
CD	----	K_{CD}=1.8 N.m/deg^2
PP	$T_{offset}=-0.00447\theta^3-0.64\theta^2+4.4342\theta+67327$ ("θ" is foot angle)	K_{CD}
Swing	Return to initial position of foot	---

6 PROPOSED DESIGN

To simplify the control algorithm, a new design was proposed so that a nonlinear spring was replaced by two linear springs situated in parallel and one of them was linked to DC motor. In the proposed design, to accomplish each phase of foot movement, the following technical solutions were provided:

1. CP phase: This phase is achieved by a linear spring.
2. CD phase: This phase is modeled by two other linear springs parallel to CP's spring and a controllable DC motor which is series by one of springs. At the beginning of CD phase, the motor and related spring has no action on foot and the foot is constrained by two other springs. The torque versus angle domain value of such composed spring is restricted, and it cannot essentially satisfy the mechanical requirement during the CD phase. Otherwise, at the end of this phase, although the foot prosthesis angle reaches the same amount as in healthy foot, its torque cannot satisfy the condition that is in a healthy foot. Therefore, an additional torque is needed to offset the foot torque to realize the healthy foot condition. By using the actuator and its coupled spring, the external work introducing the cited offset torque is supplied. The exerted work is

accumulated as potential energy in springs and will be released in PP phase.

3. PP phase: In this phase, the actuator is adjusted to convert spring potential energy to foot work.

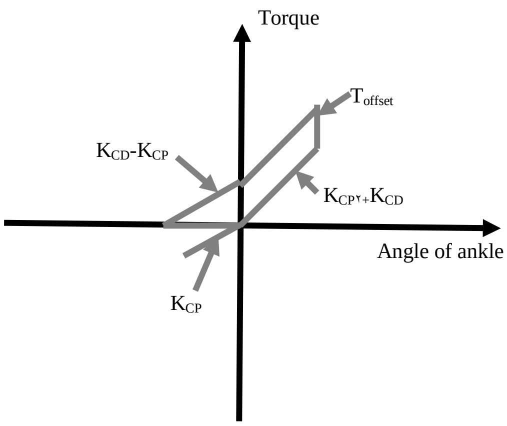

Fig. 12 Schematic of linear model and method of its performance

Figure 12 shows conceptual design of the proposed design. Figures 13 and 14 respectively show angle versus time and torque versus angle related to the designed foot prosthesis in virtual simulation. The results show that the designed foot can be appropriate replacement for natural foot because its biomechanical behavior has appropriate equality to that of natural foot.

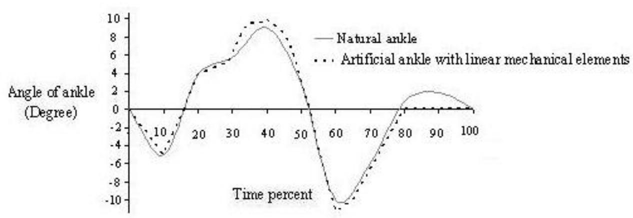

Fig. 13 Angle versus time of foot joint

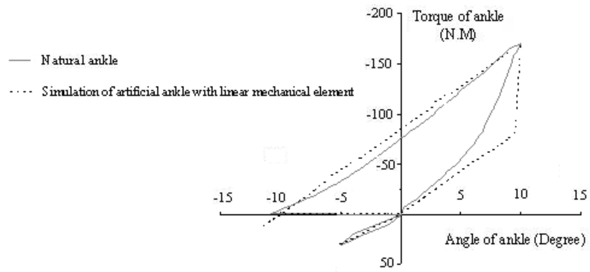

Fig. 14 Torque versus angle of foot joint

7 INITIAL PROTOTYPING

To validate the proposed design, an empirical prototype was fabricated (Fig. 15). This prosthesis includes a variable stiffness spring, composed of a spring series by a solar geared 12V DC motor via a ball screw mechanism, and two other passive springs (Fig. 16). The springs are limited by leg and foot. One of the passive springs which is installed at heal side of foot joint constraints foot at dorsi flexion and the other springs installed at paw side of foot joint constraints foot in plantar flexion. The torque exerted by motor is controlled via an appropriate real time algorithm attended by 4 sensors; two of them measure force exerted by motor and foot joint rotation, and the two others are used to indicate heal strike and paw strike.

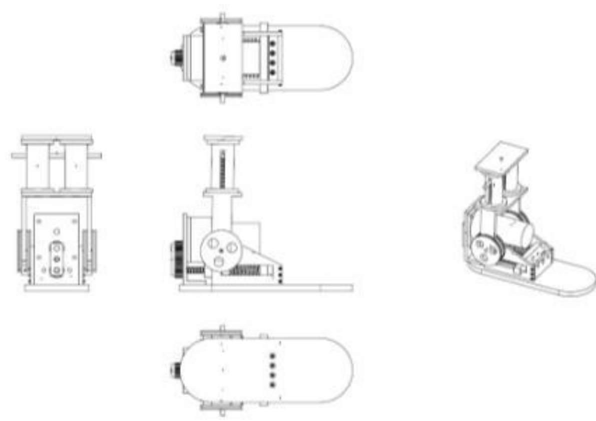

Fig. 15 Prototype design

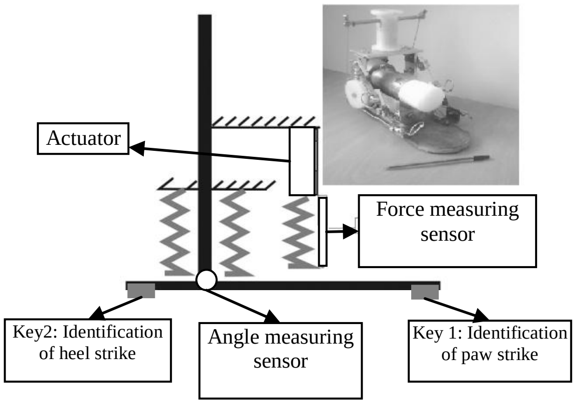

Fig. 16 Prototype layout

Figure 17 shows the layout of prosthesis control algorithm during a walking gait. According to this algorithm four critical foot conditions are detected indicating the start time of a motion phase during a walking gait cycle. To do each motion phase, the motor is controlled via an appropriate PID algorithm closed by a PWM driver (Fig. 18).

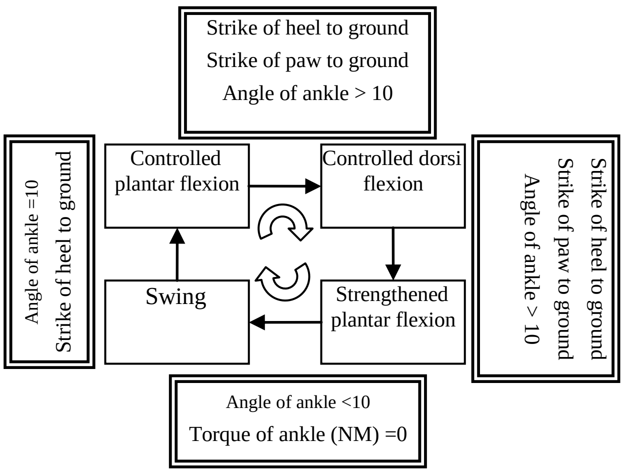

Fig. 17 Layout of prosthesis control algorithm

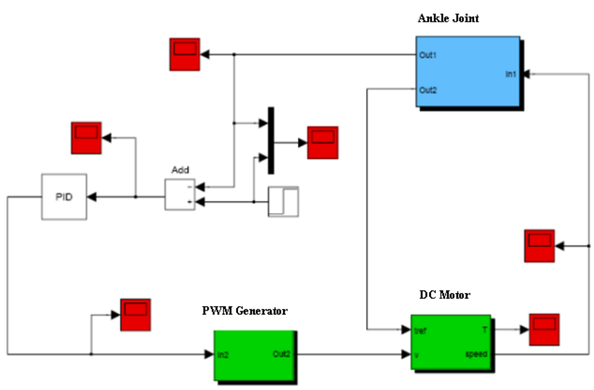

Fig. 18 Block diagram of prosthesis control algorithm

Commercial software was used to process the input data and apply the outcome commands to direct the actuator. The results of test of this mechanism during a walking gait were shown in Fig. 19. As it is definite, this mechanism can make torque in each angle so that it can have appropriate equality to the simulated model.

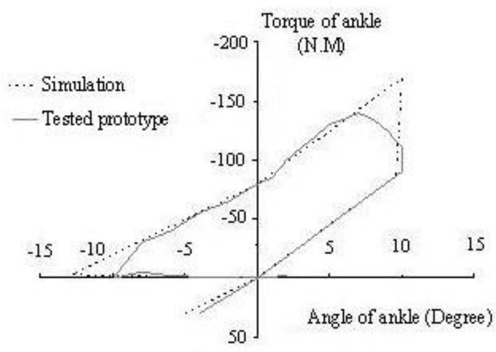

Fig. 19 Results of the initial prototype test

8 CONCLUSION

In this research the biomechanical behavior of foot during a walking gait was estimated using different artificial mechanism composed of linear and nonlinear springs and one of them was actuated via a controllable motor. It was proved that an artificial foot, composed of a linear spring and a second order spring, behaves roughly like a healthy foot during CD and CP phases and otherwise they branch out. To correct the divergence, the prosthesis was redesigned using a controllable motor linked to a nonlinear spring.

It was shown that, regardless of the spring specification, its linked motor can compensate the spring failure using an appropriate real time control. To simplify the control algorithm, a new design was proposed so that the nonlinear spring was replaced by two linear springs situated in parallel and one of them was linked to motor.

Mechanical ability of the proposed design was simulated and virtually was compared with a natural foot. Finally, according to the proposed design, a prototype was fabricated and its mechanical behavior was simulated using a PID control algorithm. Despite some weakness in prototyping, the torque versus angle has good agreement with that of healthy foot.

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