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Design, control, and prototyping of a series elastic actuator for an active knee orthosis

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A B S T R A C T

The use of elastic elements in actuators can be effective in improving their performance. The present paper examines the different layouts of the elastic element in actuators, choosing the appropriate design, and then designing a series elastic actuator for use in an active knee orthosis. For this purpose, the design elements of actuators were extracted using the knee data for walking. In the next step, considering the effect of the spring stiffness on the energy consumption and actuator's peak power, its optimum amount for gait was calculated. Afterwards, the effect of adding the spring has been shown at an optimal value. Since the human knee in a gait cycle involves position and force control, designing an efficient controller to track the force and position of the knee during the gait cycle is the next section of this study. Due to the specific application of this actuator, its volume and size are important for the user. For this reason, it is necessary to design and construct an actuator with a suitable size and weight, with good output. For this purpose, a lightweight and low volume actuator was fabricated.

1. Introduction

War, accidents, illness and genetics are among the factors that cause disability. According to statistics released in 2015, in the United States, 7% of the population, that is about 22.5 million, are suffering from Ambulatory Disability (Erickson et al., 2016). One of the most common types of lower limb disability is the inability to walk correctly due to a knee joint failure. The failure in the performance of this joint, in addition to causing severe pain while walking, leads to a reduction in the speed and function of the disabled person and his imbalance. The use of an orthosis is one of the common ways to help people with impaired walking. These devices make it easier for the person to use that limb, by helping the person in controlling the particular limb. Although an inactive orthosis helps the person with disabilities to use their body, they also have limitations. On the other hand, injured people who attend rehabilitation sessions after the incident, due to excessive pressure and fatigue, have meetings shorter than one hour, also, between the two sessions of practice, the patient walks with the pattern that he/she is accustomed to it. This increases the duration of treatment and reduces the effectiveness of training sessions.

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The use of an active orthosis with a controllable actuator in addition to speeding the process of improving the walking pattern and rehabilitation process, can be effective in restoring the daily lives of affected individuals. The purpose of this research is designing and building an active orthosis with an acceptable weight and size for the knee joint, which, in addition to helping the injured in training sessions, also has the ability to be used outside the clinical environment for daily routine affairs. The focus of this study is on the use of a series elastic actuator to optimize energy consumption and peak power in addition to better interaction with the body due to its low mechanical impedance and power to weight ratio. Pratt et al first introduced an exoskeleton for the knee that takes advantage of a series elastic actuator (Pratt et al. 2004).

In many cases an active orthosis for the knee has the task of teaching and accompanying the patient in rehabilitation exercises (Weinberg et al., 2007). In this case the orthosis was created by means of a brake fluid that provided the torque required to prevent knee bending in the stance state. Researchers in the Laboratory of Biomedical Robotics and Biomicrosystems have introduced a torsional series elastic actuator for a knee orthosis (Sergi et al., 2012). The torsional spring of this actuator has been designed and built for this purpose. Yu and colleagues have introduced a knee-ankle foot orthosis for stroke patients and for outside the clinical environment (Yu et al., 2013). Hassani and his colleagues also presented a knee active orthosis designed to help in training sessions (Hassani et al., 2014). Shan and colleagues have introduced a knee active orthosis to help walking that uses electromyography signals, they showed that the active orthosis helps reduce muscle activity (Shan et al., 2016). In the remainder of this article and in the second section, we will examine different actuator layouts and select the appropriate actuator. Then we try to design the selected actuator. The means by which the orthosis is controlled is also presented in the third part. In the fourth section, the results are presented and in the final section the conclusions of the work and suggestions are presented.

2. Methods

Elastic tissues, such as tendons, ligaments, and muscle, play an essential role in optimal energy use and body balance during walking (Ker et al., 1987; Blickhan, 1989; Cavagna et al., 1977; Farley et al., 1993; Hogan, 2002; McMahon, & Cheng 1990). The human body uses these elements to compensate for some of the energy lost when it collides with the ground by absorbing and storing energy and releasing it at the time of the removal of the leg. This is an inspiration to many designers of orthotics and prosthetics that use elastic elements to reduce the walking metabolism (Elliott & Herr 2012).

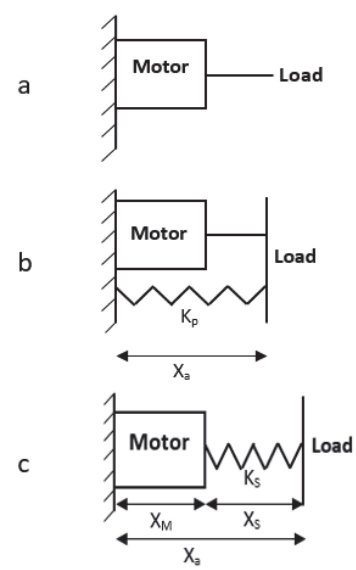

Fig. 1. Schematic view of different actuator layouts; (a) Direct Drive; (b) Parallel Elastic actuator; (c) Series Elastic actuator.

2.1. Spring optimization

In this section, we first examine three layouts including: Direct Derive (DD), Parallel Elastic Actuator (PEA) and Series Elastic Actuator (SEA). Fig. 1 shows the schematic of these three layouts.

The criterion for selecting the optimal actuator is the peak power and energy consumption. The peak power for direct drive is obtained from

$$P_M = F_K \cdot \dot{x}_a \, , \tag{1}$$

where P_M is the motor power, F_K is the knee force and $\dot{x}_a$ is the rate of actuator length change.

Also the power for the series and parallel actuator is calculated from Eq. (2) and Eq. (3) respectively; K_P is the parallel spring stiffness and Δx_a is the spring length change.

$$P_M = \left(F_K - (K_P \cdot \Delta x_a)\right) \cdot \dot{x}_a \tag{2}$$

$$P_M = F_K \left(\dot{x}_a + \frac{\dot{F}_K}{K_S}\right) \tag{3}$$

The next criterion is the orthosis energy consumption. Power is used to calculate the energy (E).

$$E = \int |P_M| \, dt \tag{4}$$

The reason for using the absolute value of the power in Eq. (4) is that the DC Motor needs power for both pushing and resistance to movement. In Eq. (1), parameters x_a and F_K are obtained from the walking geometry and clinical data, and as a result, the energy required is dependent on K_S. In the next step, in order to find the optimal value of the spring constant for reducing energy consumption and maximum power in each actuator state the spring constant was changed from 1 kN/m to 1000 kN/m with steps of 1 kN/m, and the amount of power and energy consumption of the orthosis for the various stiffness coefficients of the spring were calculated in each mode. The results show that a series elastic actuator during walking has a better performance than the other two layouts.

2.2. Design

A commercial sample of a Knee Brace is used to connect the actuator to the knee. A sliding potentiometer was used to calculate the displacement of the spring and consequently the actuator output force. This potentiometer shows the displacement with an accuracy of 0.1 mm. The Power of this actuator is taken from a 100-watt BLDC from Maxon's EC-4pole series. The 7075-T651 aluminum alloy has been used to reduce the weight of the actuator, whose ultimate strength is close to steel and has a density of about one third of the steel. Using the knee data, the maximum force of the actuator was calculated to be 1050 N for compression and 800 N for tension. The actuator's plates were modelled and analyzed using the ABAQUS software. The Maximum Distortion Energy Theory of Failure was chosen as the criteria for part performance evaluation. Comparing the results with 7075-T651 Aluminum limits, shows that the plates could withstand the working conditions. Fig. shows the stress distribution of the motor plate.

The plate which is connected to the ball screw support plate, was assumed to be fixed at the contact surface and a load of 1050 N was applied to the motor plate pin (which is connected to orthosis frame).

The analysis of the end mount of the actuator is shown in Fig. 3. Two rods are fixed at the holes on the opposite sides of the plate; in the simulation, the plate is fixed in those holes. The load of the actuator is applied to the hole in the middle, which is where the actuator connection is screwed on.

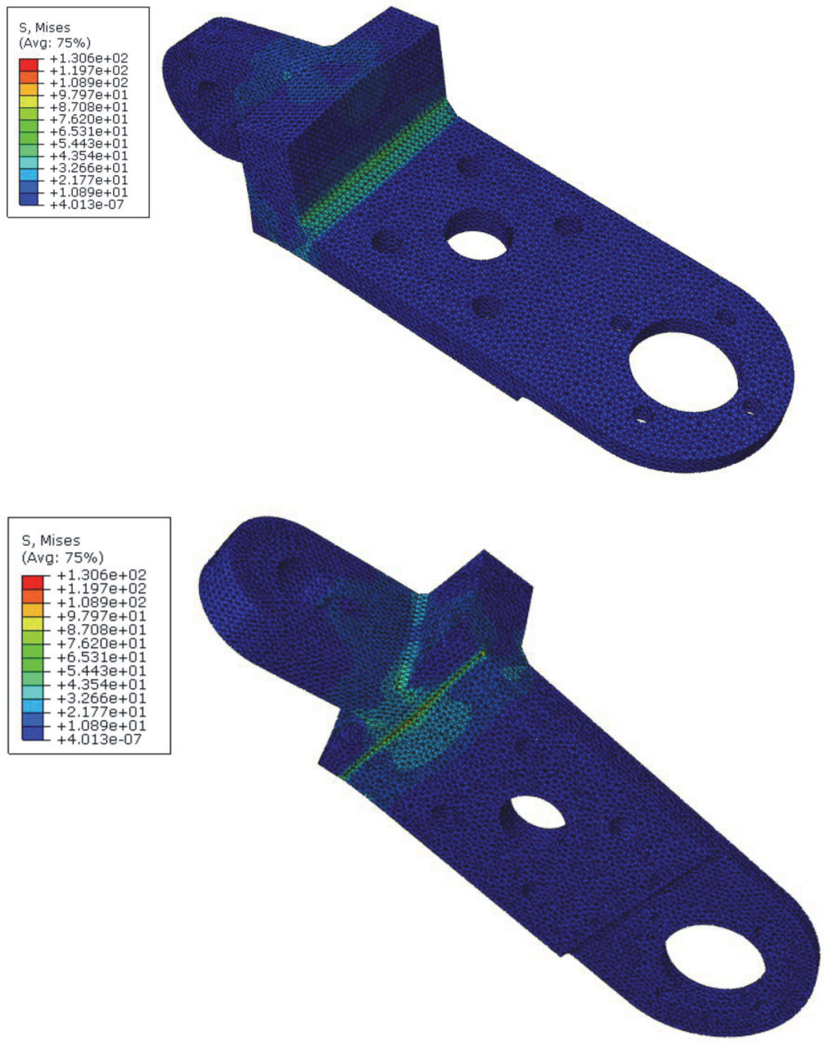

Fig. 2. Motor plate von Mises stress distribution

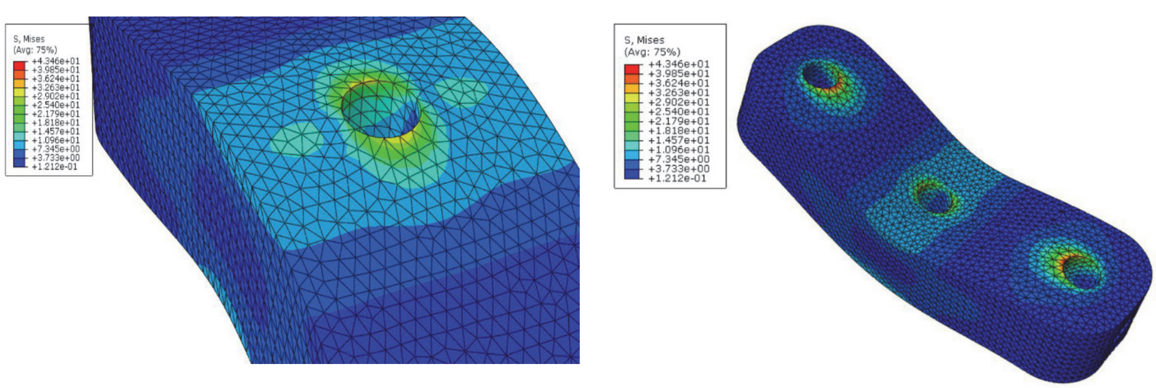

Fig. 3. End mount von Mises stress contour

The nut plate is shown in Fig. 4. In the stress analysis, this plate was fixed in the nut contact surface and tension/ compression load was applied in the springs' contact. Comparing the results of the finite element stress analysis with the yield and fatigue strength for 7075-T651 aluminum (ASM International Handbook 1990) shows that the actuator parts are safe under the loading conditions. The actuator model is shown in Fig. 5.

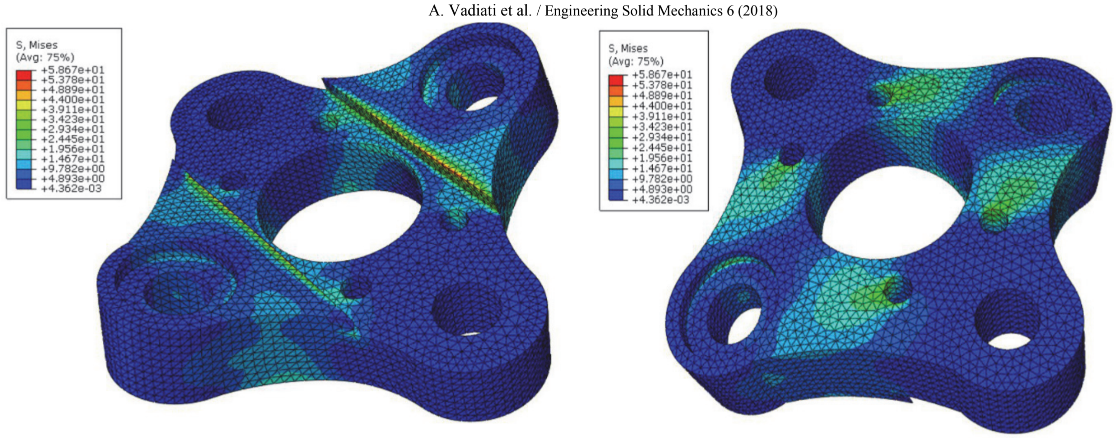

Fig. 4. Nut plate von Mises stress contour.

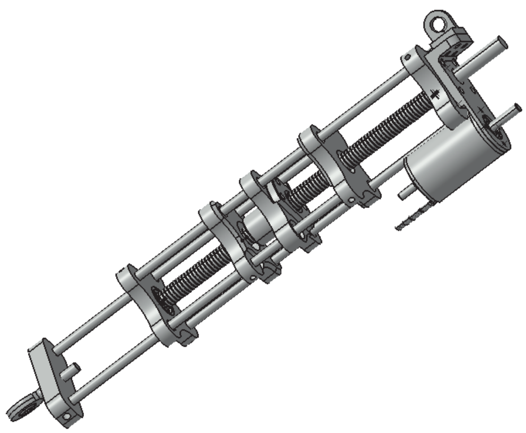

Fig. 5. CAD model of SEA actuator

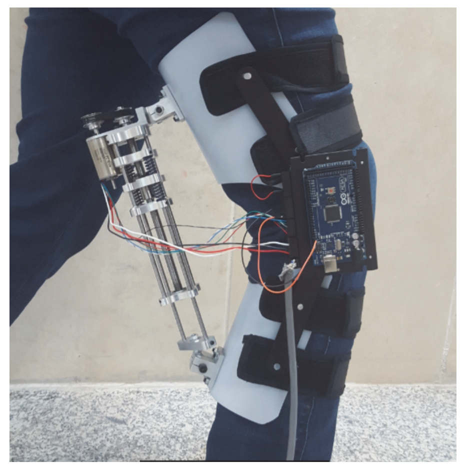

Fig. 6. Active knee orthosis prototype mounted on healthy subject.

The most important task of the active orthosis is to create an auxiliary torque in the knee joint. This action can be done rotationally or linearly. In this research, knee torque is produced by linear force and with the help of a ball screw. In this plan we have avoided the direct connection of the electric motor to reduce the length of the actuator, reduce the possibility of damage to the Motor and open the designer's hand in changing the speed and torque input to the ball screw; the motor motion is carried by belt and pulley. The Motor used in this research is EC-4pole 30. Finally, the series elastic actuator has the features listed in Table 1. The active knee orthosis is shown in Fig. 6.

Table 1. Series elastic actuator specifications.

Parameters	Values
Weight	710 g
Initial length	30 cm
Final length	45 cm
Diameter	6 cm
Max. linear speed	0.41 m/s
Continues force	510 N
Max. force	1200 N
Minimum resolvable force	<10 N
Dynamic range	120
Operating voltage	18-48 Volts
Max. current	6A

2.3. Control

The first step in simulation is the generation of the desired path for the actuator by means of the gait data. For this purpose, using the spline function, fitting the polynomial functions on the data, the force, position, velocity and acceleration functions of the actuator in terms of time for a walking cycle were obtained. In the next step, by placing the obtained functions in the mathematical relations, the desired position of motor is obtained and is shown in Fig.7.

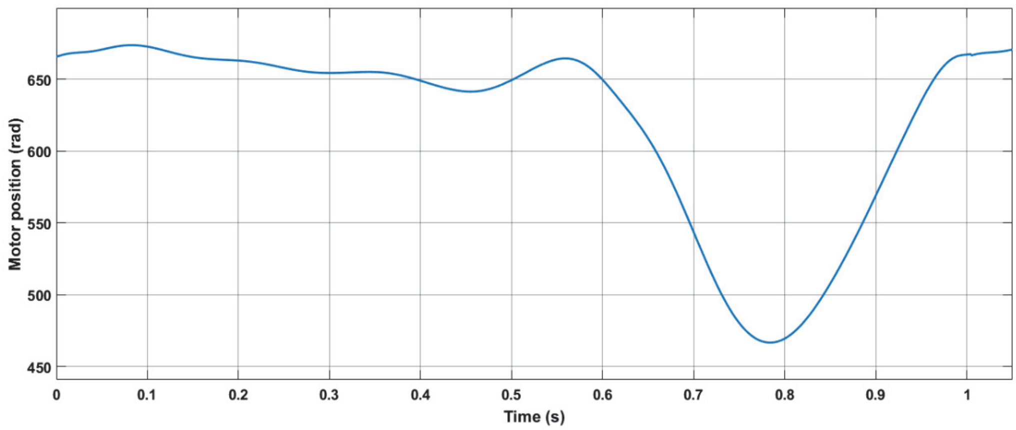

Fig. 7. Desired path of actuator DC motor.

The Computed Torque Method (CTM) is the control method used in this research. By rewriting the robot dynamics equations, we have the following general form:

$$M(q) \cdot \ddot{q} + V(q) \cdot \dot{q} + G(q) = B(q) \cdot \tau \tag{5}$$

In the above relation, the coefficients matrices will appear in the form of numbers, because the knee orthosis has one degree of freedom. By rewriting the motion equation, we have:

$$J_t \cdot \ddot{q} + \frac{np}{2\pi}(F_K + M \cdot \ddot{x}_a) = \tau \tag{6}$$

in which J_t is the total SEA inertia (motor, ball screw, pulleys and equivalent ball screw nut mass inertia), q is the motor position, n is the pulley ratio, p is the ball screw lead, tau is the motor torque and M is the mass of the actuator elements between the spring and calf.

In this method, we consider the linear feedback $q_d(t) - q(t)$ as an error. By calculating the error to calculate the torque output of the controller, one can write:

$$\tau_c = M \cdot (\ddot{q}_d + u) + N_d \tag{7}$$

where u is obtained from Eq. (8):

$$u = -K_P - K_V \cdot \dot{e} - \int K_t e(t)\, dt \tag{8}$$

The block diagram of the CTM controller is shown. In Fig. 8. A trial and error method has been used to select controller coefficients. For this purpose, the proper PID coefficients were selected by applying a step input to the system. The values of these coefficients are given in Table 2.

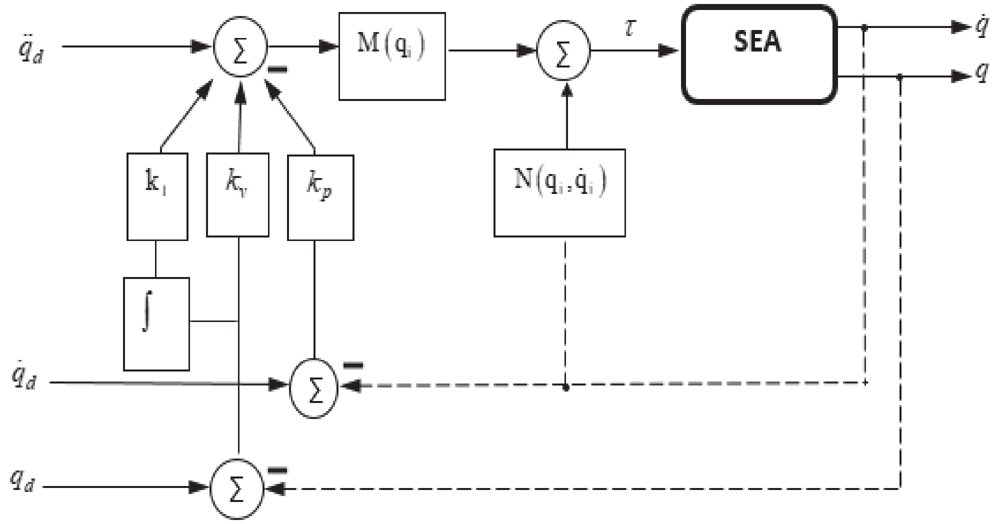

Fig. 8. Block diagram representing CTM controller.

Table 2. Coefficients of PID controller.

P	1000
I	250
D	300

3. Results

For ground level walking, the peak power for three modes of actuator have been shown in Fig. 9.

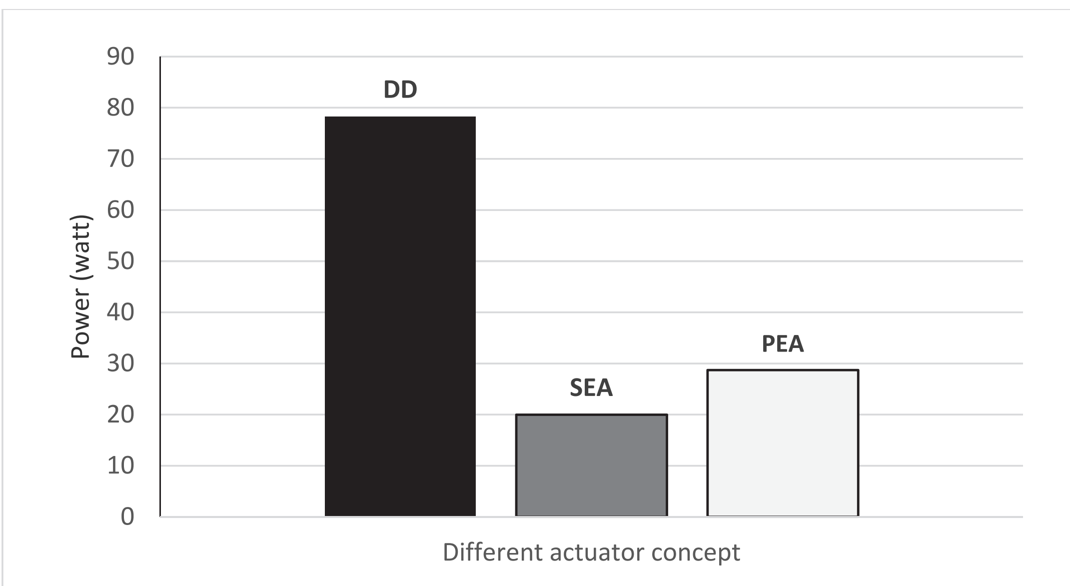

Fig. 9. Peak power for three different actuator concepts.

The energy consumption of the three models is shown in Fig. 10.

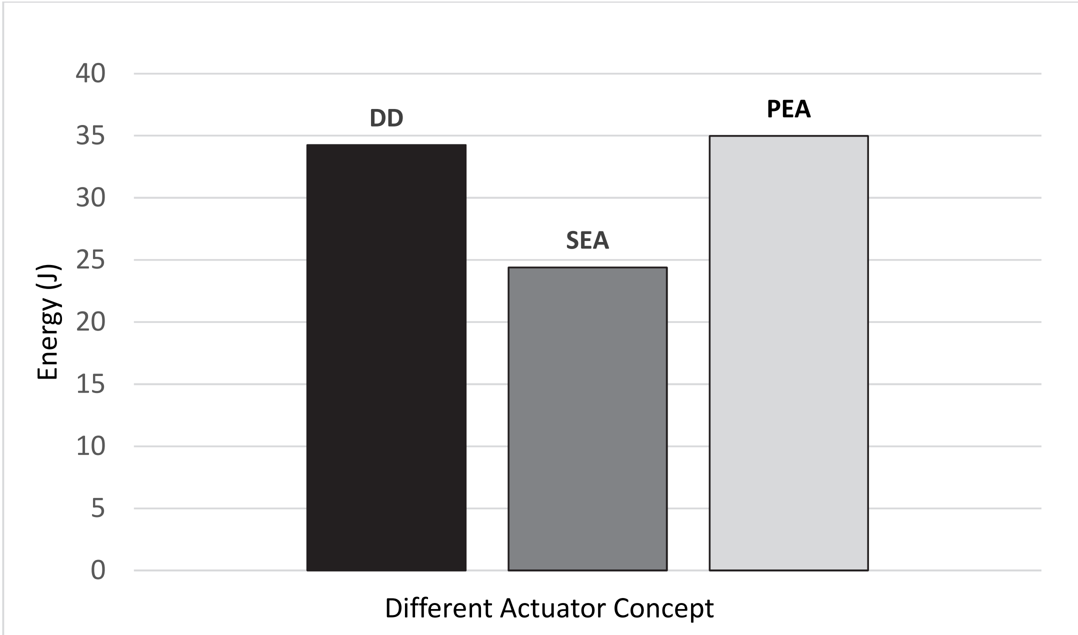

Fig. 10. Energy consumption for three different actuator concepts.

The results show that the series elastic actuator has the most suitable layout for walking. Fig. 11 shows the power changes in terms of the spring constant.

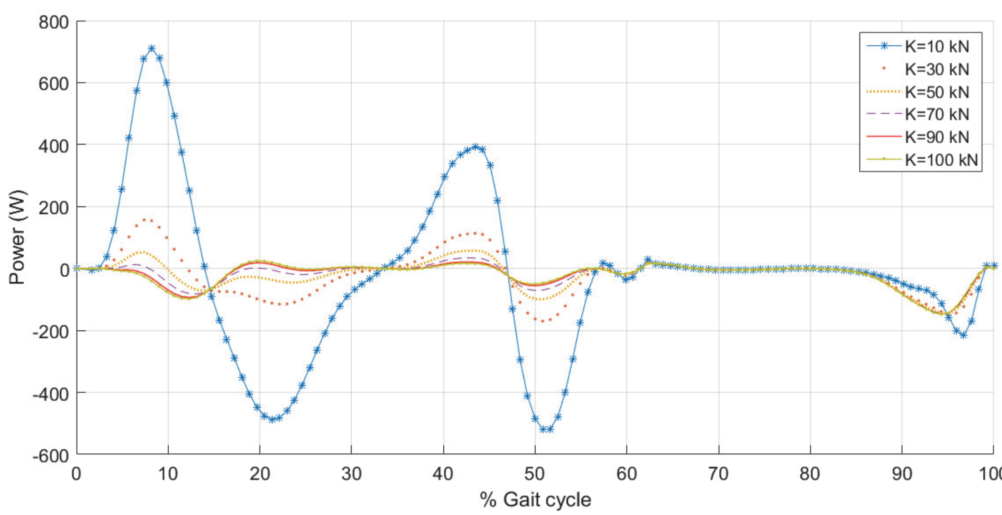

Fig. 11. Required power of SEA in gait cycle for different spring constants.

This diagram shows that the peak power decreases by increasing the spring constant. Fig. 12 shows the relationship between energy consumption and the spring constant.

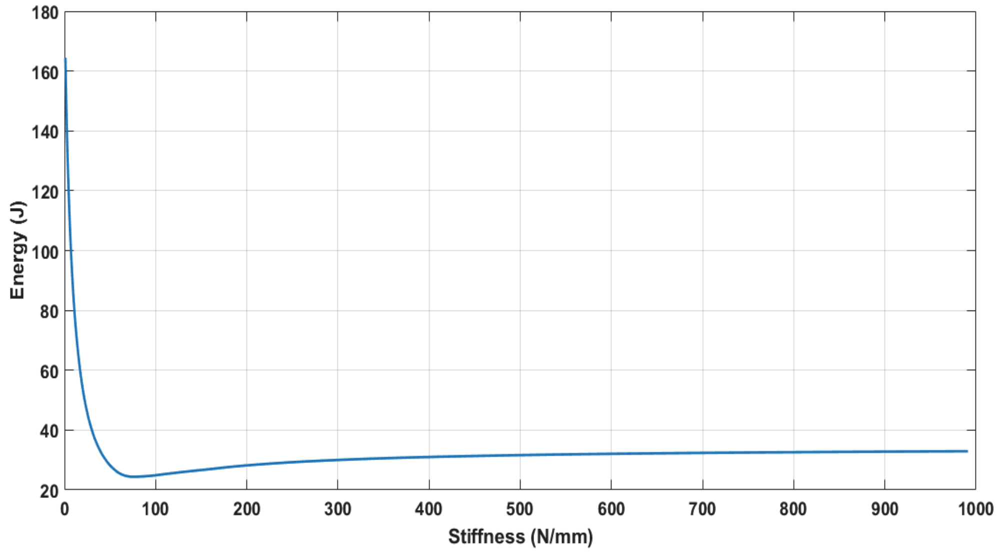

Fig. 12. Energy consumption of SEA for different spring constants

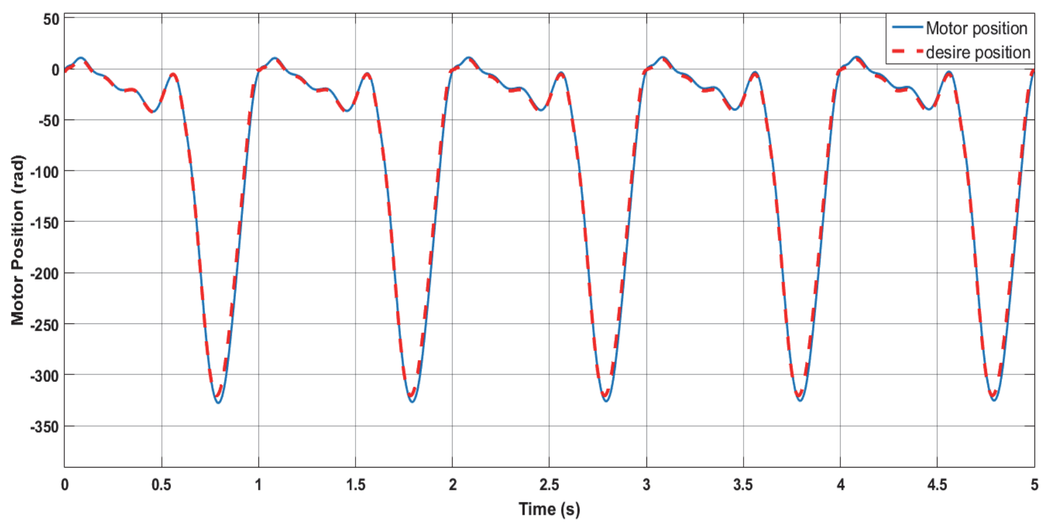

Fig. 13. Actual and desired SEA motor position

The Mid-level control of orthosis consists of two phases of stance and swing. The knee torque reaches its maximum in the stance phase. In this phase, it is necessary to control the actuator force. In the swing phase, only the knee position is important. Consequently, in this phase, it is necessary to control the knee position. Fig. 13 shows the output of the controller for position against the actual position of the knee and Fig. 14 shows the position error.

Based on motor specifications, the torque limits of 0.15 Nm were applied to simulation.

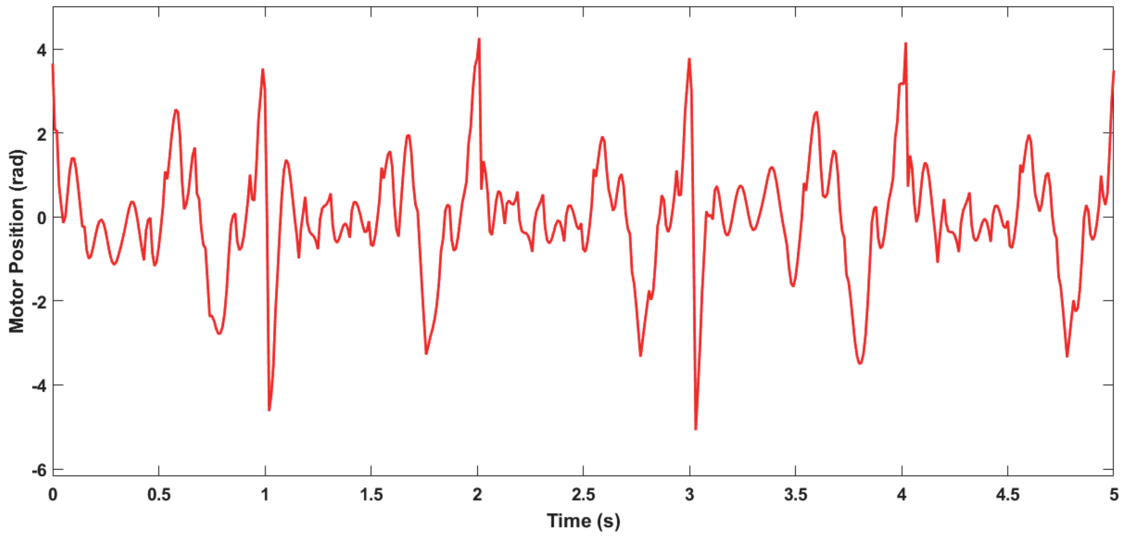

Fig. 14. SEA actuator motor position error

As is shown in Fig. 14, the error is less than 4 radians. Fig. 15 shows the output of the controlled force against the actual force of the knee.

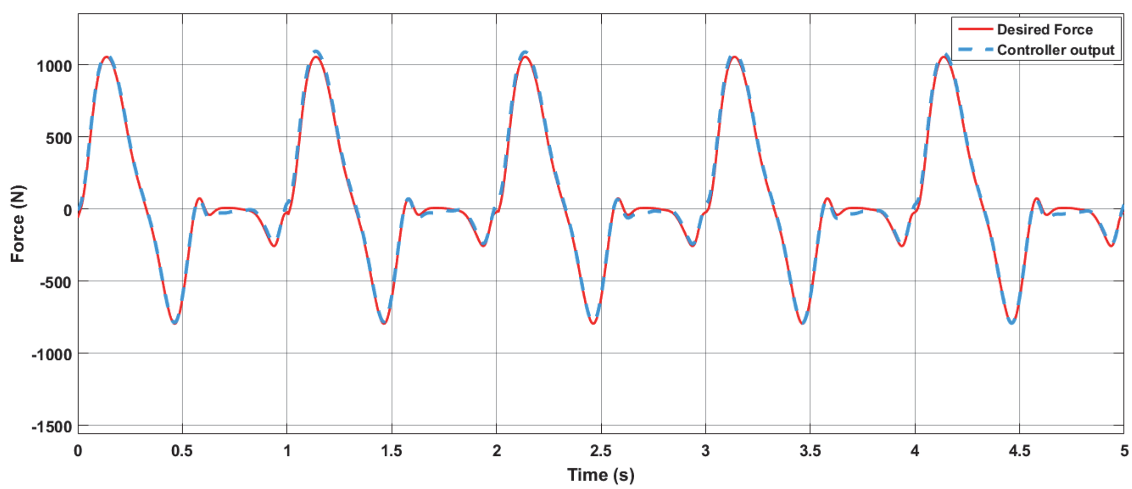

Fig. 15. Actual and desired actuator force output.

The controller error in this case is less than 10% which is acceptable. Fig. 16 Show this error.

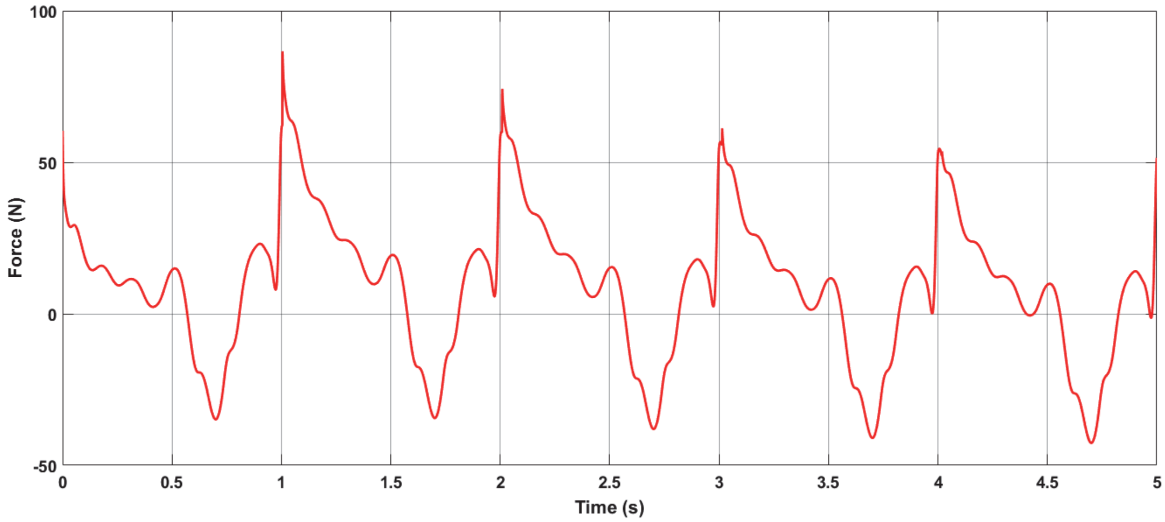

Fig. 16. SEA actuator output force error.

4. Discussion

As shown in Fig. 9 and Fig. 10, the series elastic actuator is far better both in terms of energy consumption and peak power in comparison to the other layouts. This actuator reduces energy consumption by 28.5% compared to the direct connection of the motor to the orthosis. For the peak power this actuator can cause a reduction of 74%. Although a PEA will decrease peak power by 60%, the use of the parallel elasticity will always increase energy consumption; since the lowest stiffness considered is 1kN/m, the minimum energy consumption when using a parallel spring shows a slight increase.

As is evident in Fig. 12, energy consumption in the SEA will drop initially, as stiffness increases, but after reaching the optimal amount of K=75 kN/m, changes in energy, with increasing stiffness are small. The Peak power of the SEA follows the same trend and the minimum value occurs at K=83 kN/m (Fig. 11); it can be seen that the power requirements for stiffness values higher than the optimal amount stay confined to the 50kN/m stiffness power. Although the optimal stiffness for peak power and energy consumption do not coincide, due to the small changes in stiffness values higher than the optimal amount, when choosing a stiffness equal to 83kN/m, which yields minimum peak power, an acceptable energy consumption is also achieved.

As can be seen in Fig. 13 and Fig. 14, the position error of the motor is sufficiently low. The reason for the peak of the error at the end of each gait period is the saturation of the motor which was considered based on the motor data. During the saturation time the motor falls behind from the desired path and leads to position and consequently, force errors (Figs. (15-16). As is evident, the maximum force error coincides with the maximum position error.

5. Conclusion

Using series springs in the actuator and calculating the optimal spring coefficient, it can be said that the construction of an active orthosis that is used daily is one step closer to reality. In this study, a series elastic actuator was designed for use in an active knee orthosis, and eventually an actuator with a mass less than 1kg was developed that provides the ability to produce the required knee joint force and speed. It was then shown that the control of the force and position of the actuator by the selected method is suitable.

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