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Viable Design of an Arm Prosthesis System

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Abstract

A viable design of an arm prosthesis system is presented. The design has addressed some existing challenges by adopting a fresh approach. The design is demonstrated by a simulation model. Dielectric Elastomers (DEs) are implemented into the model. That leads to higher reliability and user-friendliness, because the system contains fewer components. In addition, DEs have the desirable feature of being able to act similar to human muscles, and that makes their integration into human body easier than other technologies. An existing mathematical model of DE material is used to represent the bicep muscle. A set of operational parameters is selected in order to specify the design details. These details include geometric dimensions, forces, power, and control. The proposed design is promising with a displacement of 5 cm and a resisting force of 34 N in response to an applied voltage of only 23 kV.

Keywords: *Arm prosthesis, Bicep muscle, Simulation, Dielectric elastomers*

1. Introduction

Upper limb amputations cause severe suffering to many people, which leaves their families and supporting institutions struggling. In prosthesis industry, the main challenge is to provide the amputee with a substitute that functions with a high sensation of reality. The performance of the prosthesis is highly related to the actuation method used. Several actuation methods were developed, starting with simple mechanisms passing through electrical motors and pneumatic actuators, and ending with artificial muscles. DEs have already been investigated as artificial muscles. Their properties have been determined and many researchers have stated mathematical models for these materials. However, so far, these materials have not been successfully integrated in artificial limbs. That is due to their high actuation voltage requirement, along with low output forces which limit their use. In this work, a new design is proposed which deals with these difficulties and promises a tangible improvement to the prosthesis design.

Field-activated polymers have several properties that make them attractive as muscle substitutes. They respond quickly and have high electromechanical coupling that can allow efficiencies as high as 80%. Moreover, they can exceed the peak power of natural muscle, allowing devices of size and mass comparable to natural muscles [11]. However, phenomena such as stress relaxation, hysteresis, creep, and viscoelasticity remain as challenges to the widespread implementation of DEs [17]. In addition, a high actuating voltage is needed to have DEs deform. A thoughtful idea to have the actuator supply itself with power is to integrate a DE block that is operated in the generating mode, converting the deformation of the elastomer to electrical energy [20].

Many researchers have dealt with DEs modeling such as references [1] to [7]. They used multiple combinations of geometry and conditions. However, only few of them used the material in real applications. Those applications generally involved relatively small forces

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and deformations. These results were presented in references [8, 9, and 10]. The shape of the DE material affects the properties as well as strain and voltage characteristics. One actuator configuration is an elastomer inflatable structure; Its operation principle is based on tensile stress, where the stress generated inside the structure is transformed into a linear force able to contract the artificial muscle [18]. Other possible configurations involve contraction of a stack actuator and expansion of a tube and roll actuator [19].

This work aims at simulating the arm motion with DE material as the actuator. As a result, a functional model of the arm prosthesis system is designed. The proposed design employs large dimensions. These dimensions enable the DE to generate relatively high forces. A solution for the high-voltage problem is presented as well. Using the proposed design, the DE actuator acquires better control precision. Also, an innovative configuration for mounting the DE block is proposed. This configuration can achieve the required path of motion while minimizing loads on the DE block.

2. Mathematical Model

The mathematical model is based on the combination of Elasticity and Dielectricity formulations. The elastic formulation is based on the two-term Ogden model [8]:

$$F = X'_2\, X'_3[\mu_1\, (\alpha_1^{(k1-1)} - \alpha_1^{(-1-k1)}) + \mu_2\, (\alpha_1^{(k2-1)} - \alpha_1^{(-1-k2)})] \tag{1}$$

Where,

F: Force (N).

X'2: Initial width of DE material (m).

X'3: Initial thickness of DE material (m).

μ_1, μ_2: Ogden model parameters (Pa).

α: Extension ratio.

k1, k2: Ogden spring constants

While the DE material can be represented as a capacitor with the following capacitance [9]:

$$C = \varepsilon_0\, \varepsilon_r\, A / X_3 \tag{2}$$

Where,

C: Dielectric Elastomer capacitance (H)

ε_0: Vacuum permittivity (8.854*10-12 F/m)

ε_r: Relative permittivity

A: Area of DE after deformation (m^2)

X_3: Deformed thickness of DE material (m).

In addition, two resistors are required; One in parallel, representing the small conducting current, and another in series, representing the electrode wiring and connections resistance. The resistances are R_L = 14.28 GOhm, and R_S = 1708 Ohm [9]. The force between both electrodes is determined using the Maxwell stress theory:

$$F_{maxwell} = (X'_2/X'_3)\, \alpha\, \varepsilon_0\, \varepsilon_r\, A\, V^2 \tag{3}$$

Where,

$F_{maxwell}$: Maxwell Stress Force (N)

X'2: Initial width of DE material (m).

X'3: Initial thickness of DE material (m).

α: Extension ratio.

ε_0: Vacuum permittivity (8.854*10-12 F/m)

ε_r: Relative permittivity

A: Area of DE after deformation (m^2)

V : Voltage (V)

The combined analytical model now is [7]:

$$F_{Ogden} = X'_2\, X'_3[\ \mu_1\ (\ \alpha_1(k_1\text{-}1) - \alpha_1(\text{-}1\text{-}k_1)\ \alpha_p - k_1) + \mu_2\ (\ \alpha_1(k_2\text{-}1) - \alpha_1(\text{-}1\text{-}k_2)\ \alpha_p - k_1) \tag{4}$$

$$F_{Maxwell} = (X'_2/X'_3)\ \alpha\ \alpha_p\ \varepsilon_0\ \varepsilon_r\ A\ V_2 \tag{5}$$

Where pre-strain is in width direction and $\alpha_p = X_2/X'_2$

3. Design Configuration

3.1. Compliant electrodes

By applying a certain voltage V across a definite thickness of a DE actuator, the resulting electrostatic pressure generates a strain depending on the square of the applied electric field. For an ideal device with electrodes made of a perfect conductor, this field would be V/z0. An electrode with small resistance (~kΩ) will not have a significant effect on the behavior [5]. Still second-generation DE materials can be used as electrodes themselves since they are electrically conducting. However, the strains generated are smaller than those for first generation materials, whilst the stress produced is comparatively large. Consequently, mechanical configurations have been utilized that can amplify these small strains and make use of their potentially high work capacity [16].

3.2. Material model

A commercially available 3M VHB acrylic elastomer (VHB 4910 and VHB 4905) appeared to be the most promising in terms of strain performance, with strains in excess of 380% reported for highly pre-strained films. The theoretical energy density of this elastomer is an impressive 3.4 MJ/m^3 and coupling coefficients as well as efficiencies as high as 90% are possible [4]. The Ogden parameters and Dielectric constants for modeling a dielectric Acrylic (3M VHB 4910) DE actuator are obtained from Kofod's G. [7]. The numerical model's parameters (3-term Ogden model) are as follows:

$$\mu_1 = 8.85\text{ kPa},\ \mu_2 = 84.3\text{ kPa},\ \mu_3 = -23.3\text{ kPa}$$
$$k_1 = 1.293,\ k_2 = 2.3252,\ k_3 = 2.561 \tag{6}$$

This material model will be simulated to represent the muscle actuator. As for the selection of the dielectric constant, experimental values vary within the range (3.2-4.7) [7]. The value of 4.7 is selected to enhance the accuracy of the model. Mathematical Models of the selected 3-term Ogden model are used for material simulation.

3.3. Muscle design

An electromechanical model is used to simulate the DE actuator. It is controlled by a high voltage DC converter. The converter is manipulated by varying a programming input, which is selected according to the desired angle of arm motion. Feedback control is achieved using an angular position sensor. In addition, the proposed prime mover is a conditioned EMG signal that is interpreted to resemble the desired angle.

3.4. Mechanical model

DE material properties as well as mathematical model have been presented in previous sections. In this section, DE material is put into action. The main purpose is to achieve the bicep muscle function adequately. This involves lifting the arm upwards, by means of DE material block installation. As a mechanical structure, the block dimensions must comply with the upper limb size. A free body diagram of the proposed design is shown in Figure 1, while the following equations describe the structure.

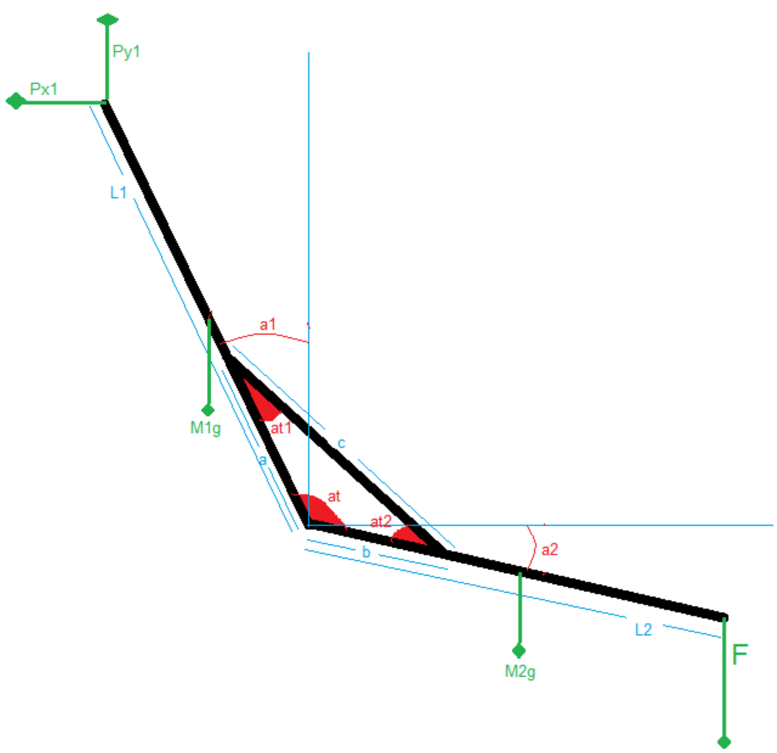

Figure 1. FBD of both arm levers with a connecting string

$$Px1 = 0 \tag{7}$$

$$Py1 = (M1 + M2)g + F \tag{8}$$

$$T = \frac{L2cos(\theta 2) * (M2g/2 + F)}{(bsin\theta T2)} \tag{9}$$

$$Px2 = Tcos(\theta 2 + \theta T2) \tag{10}$$

$$Py2 = M2g + F - Tsin(\theta 2 + \theta T2) \quad (11)$$

Trigonometric identities are used to simplify the solution as shown below:

$$\theta T1 = cos^{-1}((a^2 + c^2 - b^2)/2ac) \quad (12)$$

$$\theta T2 = cos^{-1}((b^2 + c^2 - a^2)/2bc) \quad (13)$$

$$\theta T = cos^{-1}((a^2 + b^2 - c^2)/2ab) \quad (14)$$

$$\theta 2 = \theta T - 90° - \theta 1 \quad (15)$$

Where $Px1, Py1, Px2, Py2$ are the reaction forces at the upper-arm-shoulder joint, and the upper-arm-forearm joint, respectively. The length of the upper arm L_1 and of the forearm L_2 is 40 cm and 30 cm respectively. Both lengths are estimated relative to average arm measurements [17]. DE actuator design depends on two main factors; the force *T* that the DE block must exert to achieve the desired motion, and the elongation of the DE block. In order to move bar L_2 upwards, one of the sides of triangle abc (shown in Figure 1) should change its length. Consequently, the angle will decrease. The DE block is placed at bar "a". When it extends, it slides the pivot joint between "a" and "c" along a keyway engraved in bar L_1, which pulls bar L_2 upwards.

3.5. Design parameters

Based on the aforementioned mechanical analysis, the required force relative to the translational strain of bar "a" (Figure 1) can be calculated. As the length of bar "a" increases, the angle of T decreases as shown in Figure 2. The multiple curves represent different iterations of the initial length of "a". Using Figure 2, a suitable length can be selected to achieve the range of rotation. Figure 3 shows that as the angle of T increases, the required force increases. The force and strain are inversely proportional to the length of "a" as shown in Figure 4 and 5. The operating point is marked in red as shown in Figures 2 to 5.

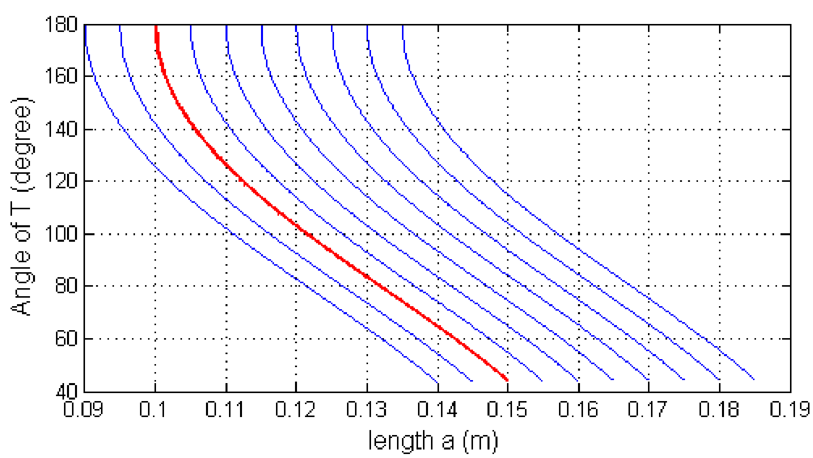

Figure 2. Angle of T vs. length of "a"

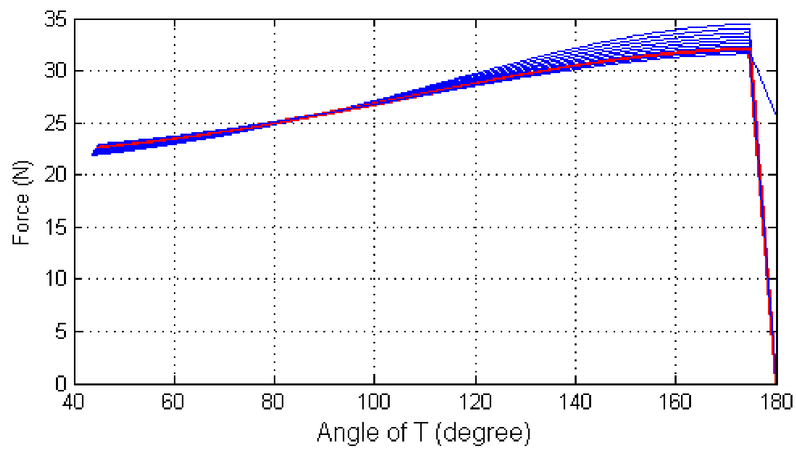

Figure 3. Force vs. angle of T

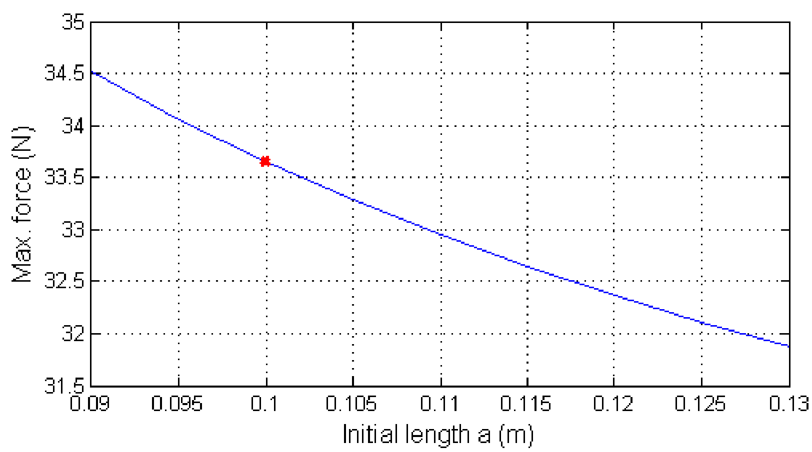

Figure 4. Maximum force vs. initial length of bar "a"

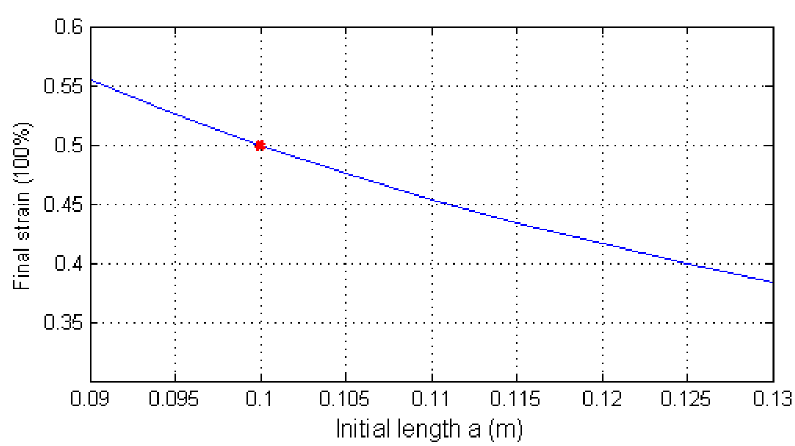

Figure 5. Final strain vs. initial length of bar "a"

The length of "b" (Figure 1) is inversely proportional to the force, but directly proportional to the strain, as shown in Figure 6 and 7 respectively. However, the design configuration limits the length of "b" to 3 cm. Consequently, the length of "c" will equal the sum of the lengths of "a" and "b" which adds up to 180^{o}. Based on the DE's model, the width has no effect on the required driving voltage as shown in Figure 8. On the other hand, the thickness has a major effect as shown in Figure 9.

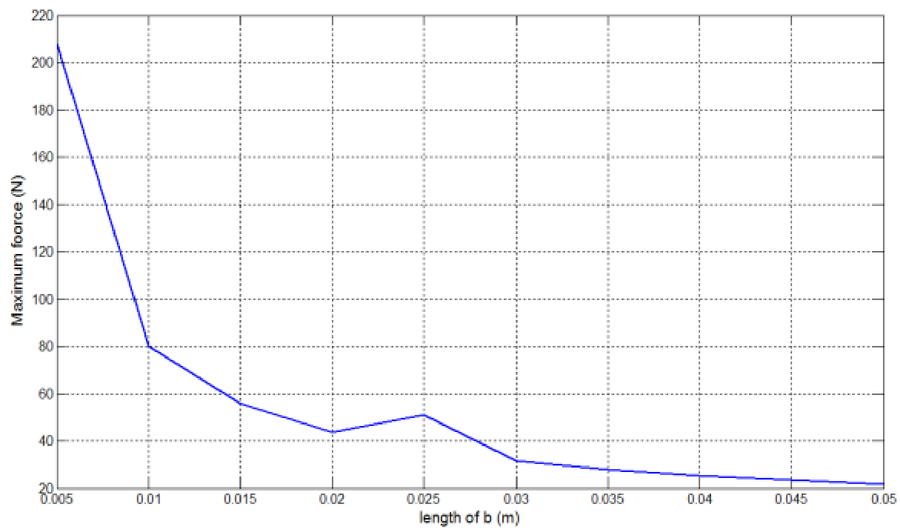

Figure 6: Maximum force vs. length of "b"

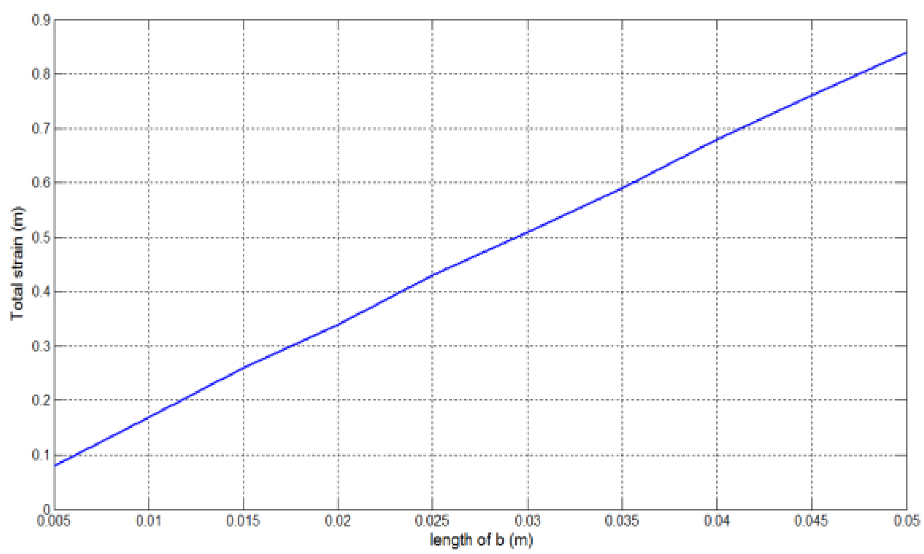

Figure 7. Total strain vs. length of "b"

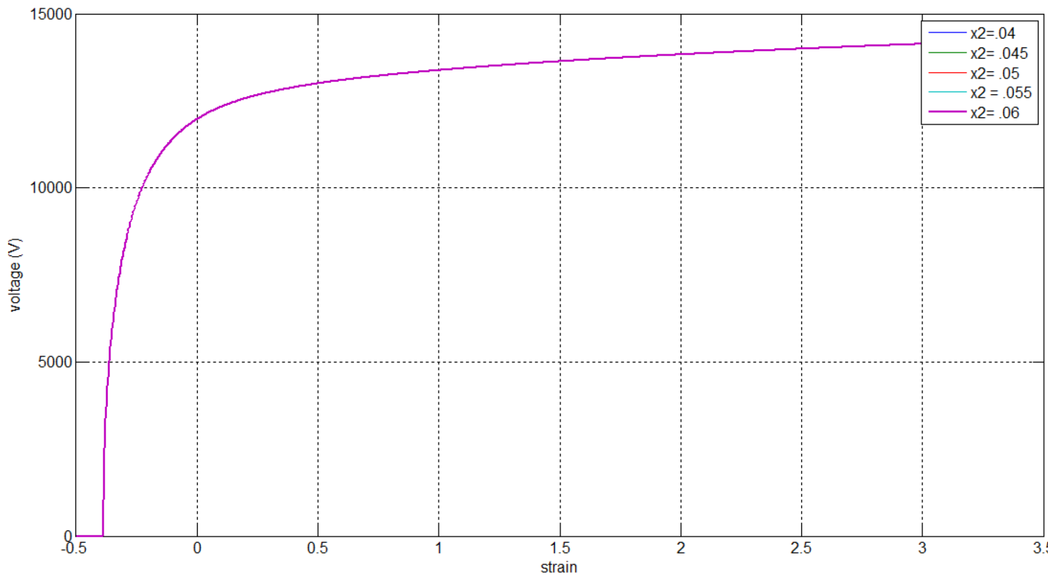

Figure 8. Voltage vs. s1train for multiple widths

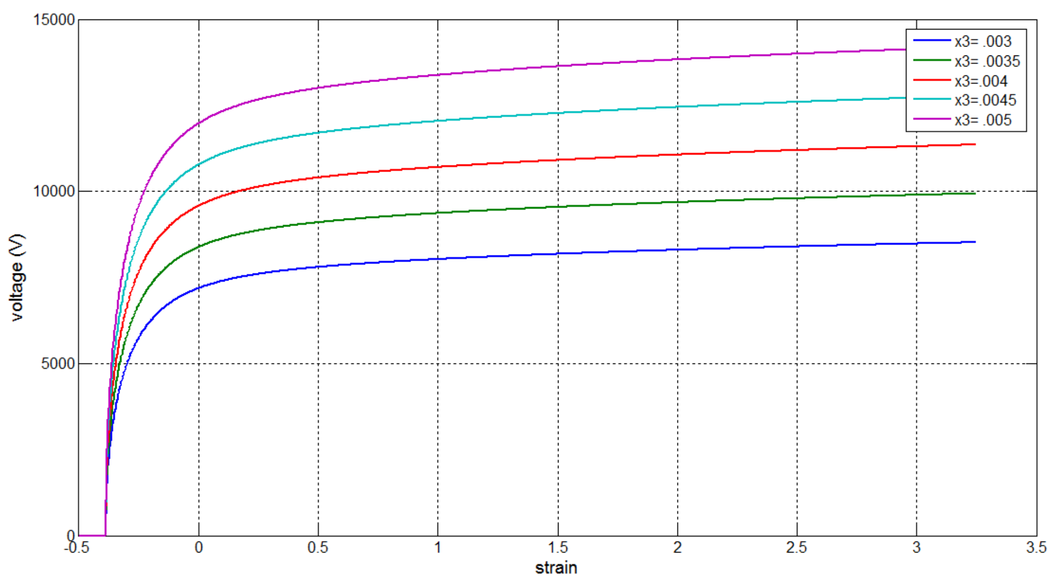

Figure 9. Voltage vs. strain for multiple thicknesses

The analysis performed so far was limited to the range of motion of the DE Block of 40° to 170° [17, 18]. That is due to hyperextension that the human muscle exhibits between the angular positions 170° and 180°. In that case, the DE block would exhibit a form of compression which is beyond the operating mode used. Table 1 lists all varying parameters and the final selected values.

Table 1. Selected parameters for the DE model

Design parameter	Selected values
DE block Length ($x_1 = a$)	10 cm
DE block Width (x_2)	5 cm
DE block Thickness (x_3)	0.5 cm
Length of bar "c"	13 cm
Length of bar "b"	3 cm
Angle $\theta 1$	0^o degrees
External force F	0 N/m
Prestrain %	400%
Angle θT	40^o – 170^o

The parameters shown in Table 1 are applied to the DE model. It is found that the voltage required for driving the actuator from 10 cm to 15 cm is 21.5 KV. As for the force, it ranges from 22.5 to 34 N/m. For example, to have a length "a" equals to 12 cm, and have the forearm move upward in an equivalent angle, the voltage on the DE terminals should be 19.5 KV. However, the DE needs to be energized at the start to pass the high voltage point which occurs at 10.03 cm. After that a decrease in the input voltage will allow the DE to reach the desired point and maintains its position. The value of 23 KV is found suitable to energize the DE and to compensate for losses through the drive circuit. The force required to achieve the desired strain are shown in Figure 10.

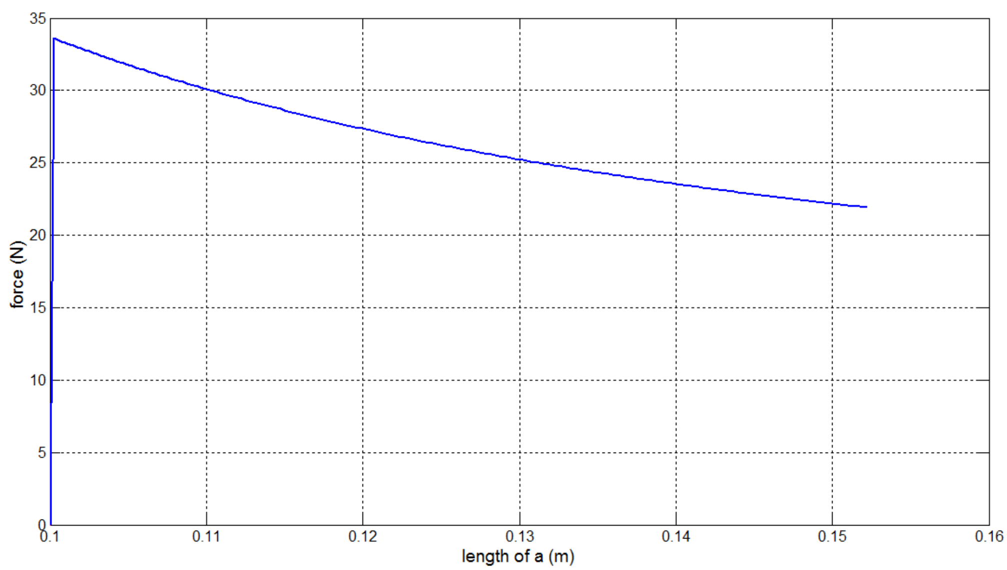

Figure 10. Force vs. length of "a"

4. Power Analysis

As shown by the analysis performed so far, the required voltage for the proposed design is in the order of Kilo Volts. This large amount of power is needed to drive and hold the DE block in the required position. This is one of the major challenges facing the success of such materials. However, thanks to great developments in the field of power converters, this challenge can now be tackled. It can be considered a matter of good installation, since size and output voltage is no longer a problem. A small DC converter device, suitable for the design at hand, is selected. The device converts a 24 Volts input into a maximum of 33 kV. This can be achieved as a response to a programming input that varies from 0-10 volts. The size of this device is remarkably small, therefore, it can be easily mounted on an amputee's arm with little consideration to its weight.

As for the power source, a rechargeable battery can be the best choice for powering the whole equipment. A pack of 20 1.2 DCV NiMH batteries are used to produce the needed voltage. The main disadvantages of such batteries are the relatively short discharge time and fast wear. An emerging solution would be to use an energy-storing super-capacitor. The main difference between super-capacitors and batteries is that the latter use chemical processes to store energy, while super-capacitors store energy through charge separation. This means the need for chemicals is reduced, which enables a longer life of super-capacitors, estimated at 1,000,000 charging cycle [19]. The power flow in the equipment is shown in Figure 11.

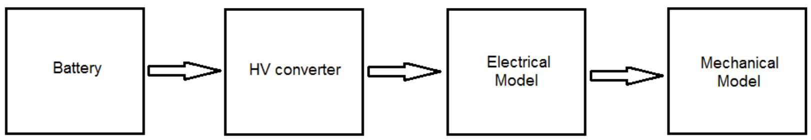

Figure 11. Power flow in the prosthesis model

In order to define the required input power and operating time for one full charge of the battery, power calculations must be carried out; The power in the transient region represents almost all of the power dissipated in the DE block. Since in the steady-state region, the arm maintains its position leading to a theoretically zero work output. In this case, the applied voltage is high, but the current is very low. Therefore, the dissipated electrical power is also close to zero. The analyses carried out thereafter represent power calculations in the transient region. Starting at the electromechanical conversion in the DE material, as voltage and current are applied by the source, the total energy stored in the material consists of electric field energy (W_{el}) and mechanical energy (W_m). The derivative of the two energies represents the power dissipated in the DE block:

$$output\ power\ P_{out} = \frac{d(W_{el} + W_m)}{dt} \tag{16}$$

$$W_m = F_{maxwell} d_{x_3} \tag{17}$$

$$W_{el} = \frac{1}{2} C\, V^2 \tag{18}$$

As for the input power:

$$input\ power\ P_{in} = \frac{dW_{in}}{dt} = IV \tag{19}$$

Figure 12 shows the time history of the input electrical power in the transient region. As shown earlier, DE materials are basically energy converters that convert electrical energy into mechanical energy, while operating in the actuating mode. The standard value of the efficiency of DE materials (Table 2) is 60-80%. Yet the efficiency calculated below is the total efficiency of a full cycle of motion, moving from 170° to 45°. The output work depends on the position and the calculated power values. Therefore:

$$\eta\ \% = \frac{W_{out}}{W_{in}} * 100\% = 49.6\% \tag{20}$$

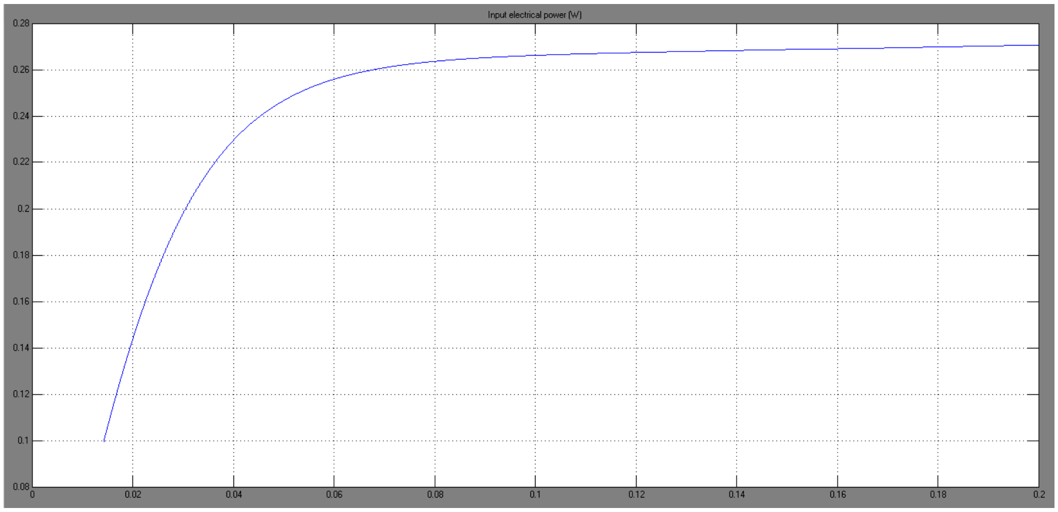

Figure 12. Input electrical power (W) vs. time (s)

The values of P_{in} , I, V are found using the DE block. They are needed to define the necessary output values of the high voltage converter. The driving voltage of the DE block is calculated as 23KV. This amount leads us to use a converter with a 10 watt maximum output power. This limitation leads to the problem of high current output of the converter. Therefore, a 14 MΩ resistor is used as input impedance for the DE block. This setup causes the drawbacks of increased response time and power losses. Referring to Table 2, the efficiency of the material is in the range 60-80%. The total efficiency is defined as:

$$\eta_{total} = \eta_{design} * \eta_{material} \tag{21}$$

Power losses are calculated relative to the efficiency of the design,

$$\eta_{design} = \frac{\eta_{total}}{\eta_{material}} = \frac{0.495}{0.6} = 0.827 \tag{22}$$

The estimated power requirements are summarized in Table 2. The data in Table 2 suggest that the frequency of motion is 1 cycle per second. According to this value, a battery of the selected type would serve 40 minutes before it requires recharging. Therefore, the use of ultra-capacitor is recommended as a storage supply.

Table 2. Estimated power requirements

	Battery	HV Converter	Model
Power		P_{out} = 10 Watt	P_{IN} = 10 Watt
Voltage	+24	V_{in} = [0,23KV]	V_{in} = 23000 V
Current	$I_{rated} = 6500\ mAh$ $I_{out} = \frac{6500}{3600}$ = 1.8 mA	I_{in} = load%*I_{in} rated = (23KV/30KV)*1 A = 0.83 mA	$I_{in} = \frac{Pin}{V_{in}}$ $= \frac{10}{23000} = 0.43$ mA
Time period	1 sec/cycle	0.11 sec/cycle	0.11 ec/cycle

5. Model Control

5.1. Human muscle control

Myoelectric control derives its name from the Latin word for muscle (myo), and the resulting by-product of electricity that muscle contraction creates. It is commonly called electromyographic (EMG). Although strictly speaking electromyography refers to the recording of myoelectric signals, rather than to the actual signals themselves. When a muscle contracts, an electric potential is produced as a by-product of that contraction. If surface electrodes are placed on the skin near a muscle, they can detect this signal [18].

The actual size of the signal is highly variable and depends on the thickness of the connective tissue, in addition to the quality of the contact between the electrode and the skin and the size of the electrodes. These factors vary from experiment to experiment so the size of an EMG is a largely qualitative measure, although it does increase in size with the activation level of a muscle. It would be convenient if there was a straightforward relationship between the EMG signal and the tension generated in a muscle. Sadly, earlier studies showed that this is not true [20]. Still, EMG signal can provide a good indication about the force and it's used widely in prostheses.

The signal must be amplified to amplitude in the range of 0 to 10 V before it can be used. The bandwidth of the surface EMG signals is 10 to 300 Hz, with most of the signals' energy in and around 100 Hz. The signal should pass through different stages before it can be used; First stage is a differential amplifier with 10,000 upward gains. It is used to amplify the EMG, because the small EMG signal is often superimposed on large common-mode signals. At these gain levels, the common-mode signals would saturate the amplifier in a single-mode configuration. Therefore, the differential amplifier can remove the large common-mode signals, leaving only the potential difference (EMG signal) between the electrodes to be amplified. Then the signal is rectified by taking the absolute value. The resultant signal is passed through a low-pass filter which acts to smooth the high frequency peaks, and produce a slowly varying trend [18]. This conditioned output voltage is interpreted in the controller as a predefined angle. This angle will pass a signal to the high voltage converter. The signal is considered a programming input voltage which will move the arm to the desired angle.

5.2. Prosthesis model control

The position of the arm joint is directly related to the EMG signal. Therefore, a position control is employed using position as a feedback. The input voltage signal and the output strain in an unloaded DE actuator are directly proportional. However, this is not the case in our model. The reason is that forces that affect the DE actuator as the arm rotates upwards decrease instead of increasing. This can be seen obviously by comparing Figures 13 and 14. The initial voltage required to move the arm is the highest voltage. Therefore, we need to provide the DE actuator with 23 KV until the maximum force is exceeded. Then the arm starts moving upwards. The voltage is then decreased to slow the arm down as it reaches the desired angle. Once this is accomplished, the angle is held at that position.

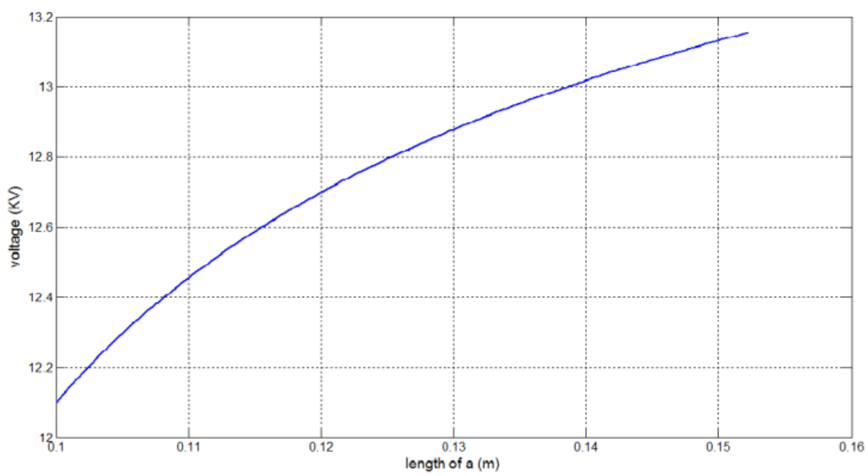

Figure 13. Voltage required for unloaded DE actuator

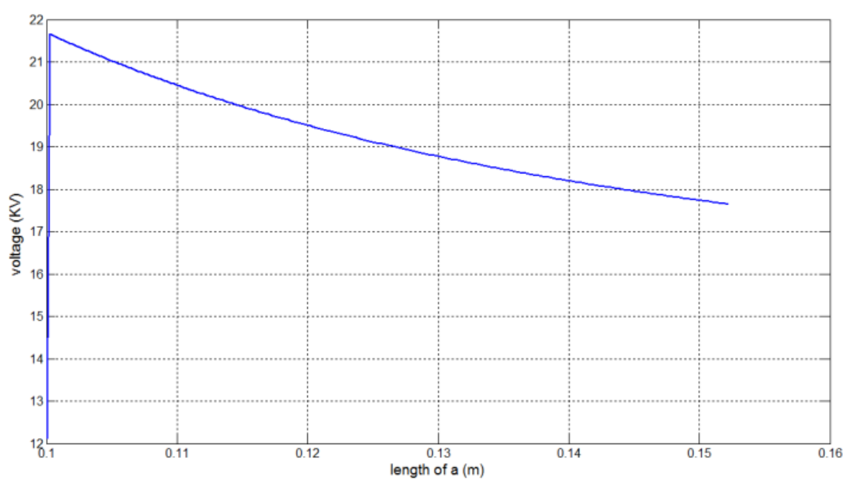

Figure 14. Voltage required for loaded DE actuator

The control flow chart is shown in Figure 15. The actual angle is fed back into the system, where it is compared to the converted EMG signal. The EMG signal is converted into the desired angle with a gain of 1000. This was deducted to increase the response time with less power dissipation. The main block is the DE material block shown in Figure 16. This block design is based on electro-mechanical coupling. In the coupled block, continuous calculation of DE capacitance takes place, which is a function of the material's deformation. The mechanical model joins Maxwell forces and load forces which are also a function of the material's deformation, then calculates the mechanical strain using Ogden's model.

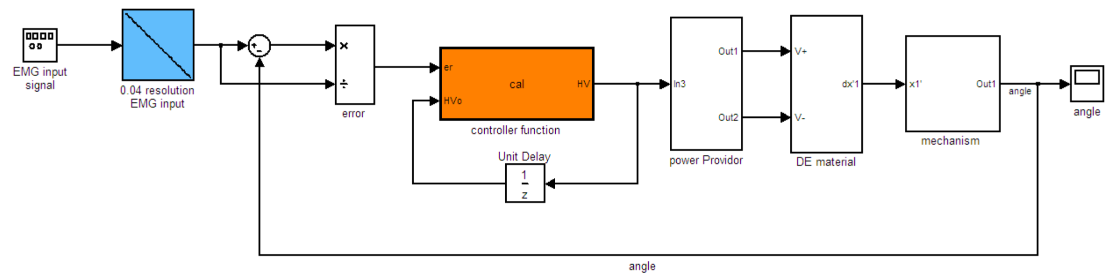

Figure 15. Control flow chart

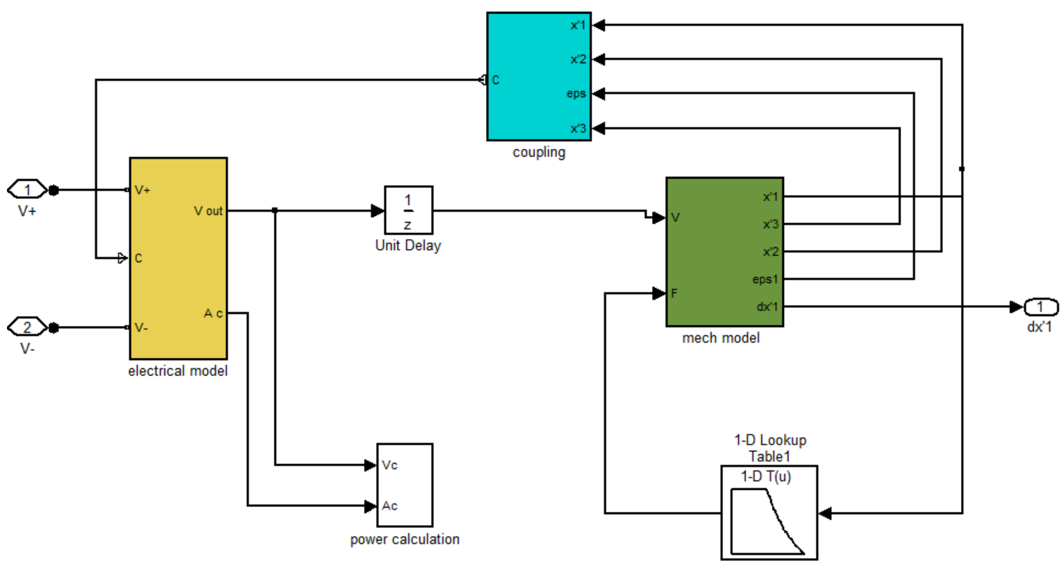

Figure 16. DE simulink block

6. System-level Simulation

After integrating all the components of the system, the next step is to simulate the whole package. The simulation is done in order to ensure that the system can perform the required tasks properly. The simulation shows that the system reaches steady-state position with maximum oscillations of ±0.25°. This results in a smooth upward movement. Figures 17-19 show the response of the system for multiple desired angles. These inputs are 660°, 920°, and 1440°, respectively. For an EMG train of signals, the input signals and output angles of the controller are shown in Figure 20 and 21 respectively.

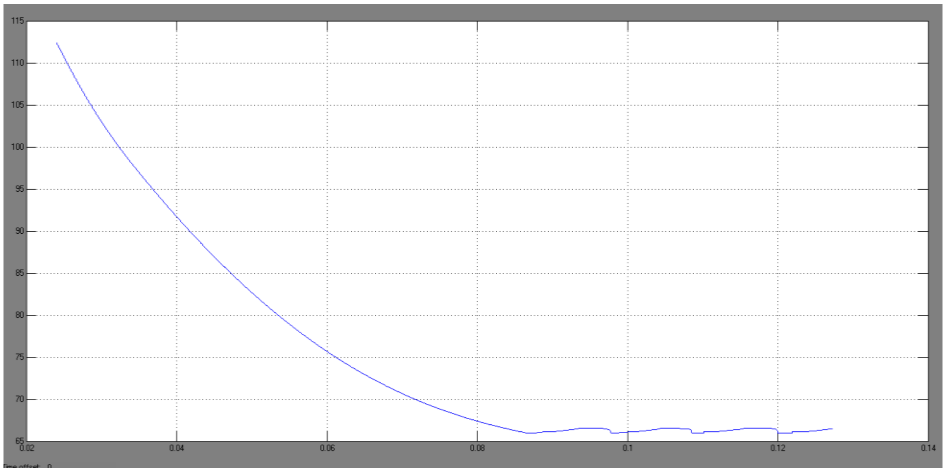

Figure 17. System response for a desired angle of 660°

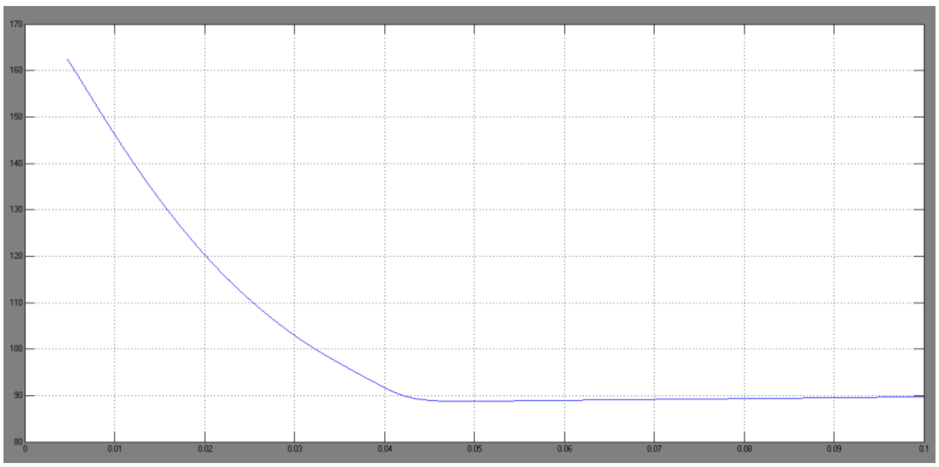

Figure 18. System response for a desired angle of 920°

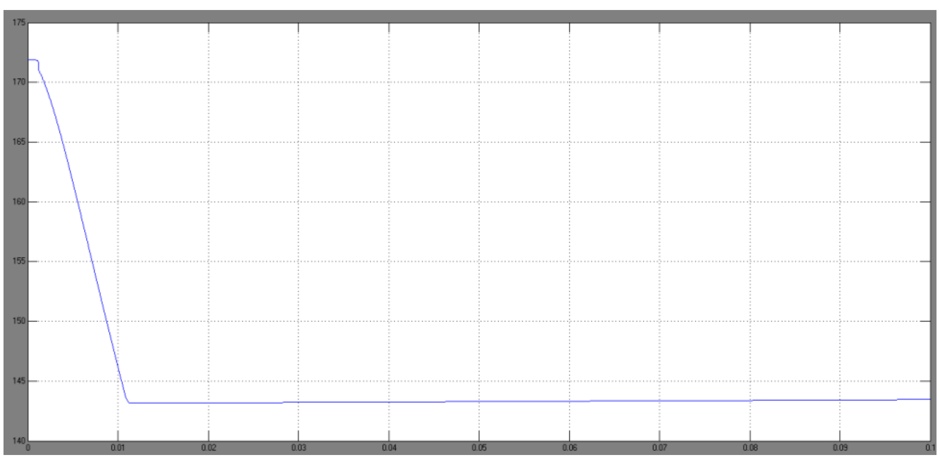

Figure 19. System response for a desired angle of 1440°

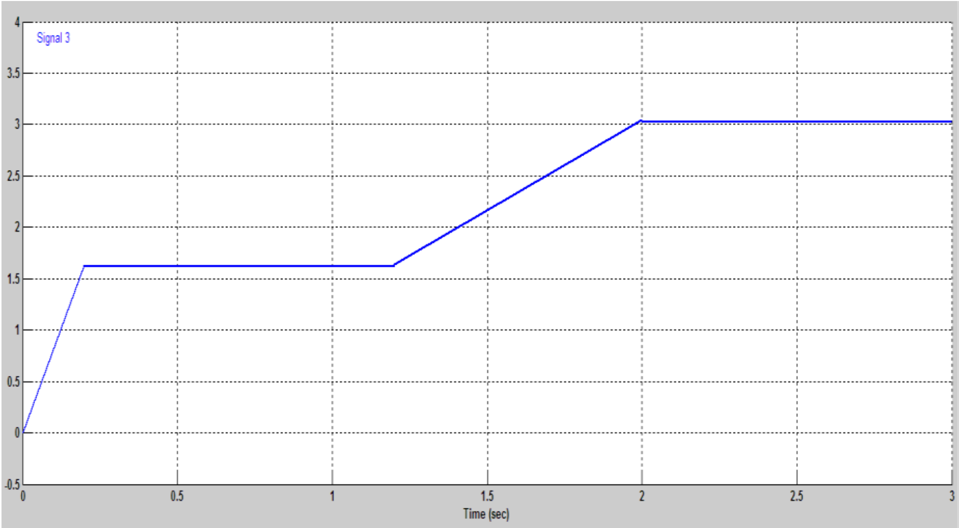

Figure 20. EMG input signal for the controller

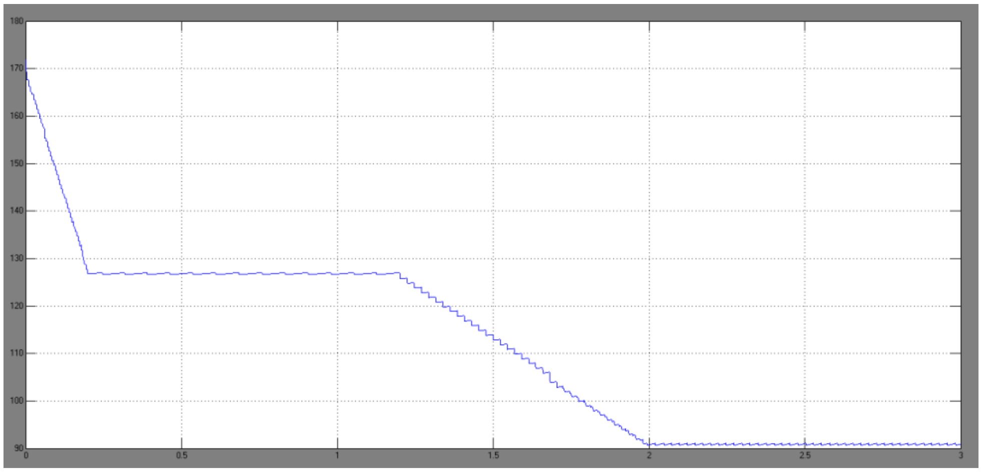

Figure 21. Output angle for the EMG input signal

7. Conclusion

An innovative arm prosthesis design is proposed. The design involves the implementation of DE material while ensuring feasible performance. One important reason for choosing this material is that it acts similar to human muscle. A multi-physical model of the proposed design is built and simulated. The model simulates the motion of the arm system. The necessary force and power computations are performed. A reasonable displacement is achieved in response to a practical voltage input. This result was possible due to careful synthesis of mechanical and electrical components. The methodology followed in this work can be utilized to design prosthesis for other parts of the body as well.

The initial simulation was based on a previous mathematical model of the DE material. It covers both elasticity and dielectric formulations. The prosthesis design is configured so as to provide the needed force and power. The proposed design has addressed some of the challenges associated with the implementation of DE materials, such as the problem of high-voltage requirement. A carefully-designed circuitry has reduced the required voltage to an acceptable level. The system parameters were tuned so as to reach the optimum operating point. Finally, the control mechanism of the system is designed to resemble the actual human arm as close as possible. This was verified by simulating the system using actual parameters, which showed that the model can accomplish the required displacements and forces.

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