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Concept and Simulation of a Portable Pneumatic Exoskeleton for Orthopedic Rehabilitation of the Elbow

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Abstract:

Exoskeletons are by definition an external cast that fits around a body part and can be modified to perform assistive and rehabilitative functions when natural motion of that part is hindered or compromised. These exoskeletons can even provide strength enhancement and aid with performing day to day activities of that part-in this case the upper arm. This paper analyzes the concept of a pneumatically driven exoskeleton which will perform the hinge like action of the elbow for rehabilitation in case of injury or paralysis. The objective of building such a prototype is to make it portable, light weight on the injured arm and aesthetic for daily wear purposes. Integrating an electronic driver unit, a pneumatic circuit and a mechanical power transmission an assembly is created to perform this hinge like movement of the elbow. It was found that the exoskeleton can provide assistive motion ranging from a gradual lift of the forearm to a slightly more accelerated rate of motion. The exoskeleton was found to lift the forearm in approximately 39 s at 12psi of pressure and can run for approximately 13 hours at a stretch which makes it a suitable device for elbow rehabilitation. This prototype can be scalable as per the user's arm dimensions and it is convenient to wear on the upper arm while carrying on with daily activities.

Keywords*: Elbow, Mechanical power transmission, Orthopedic Rehabilitation, Pneumatics, Portable Exoskeleton*

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ISSN: 0011-9342
Issue: 07 | Pages: 6876 - 6889

I. INTRODUCTION

The exoskeleton is an external cast which can be used to maneuver, protect or even provide a greater magnitude of strength to the upper limb allowing for the heightened efficiency and performance. The limb may be weakened due to diseases or reasons such as paralysis, stroke, muscular atrophy and different kinds of injuries. Elbow rehabilitation is required specifically in some conditions like “Lateral Epicondylitis” [1] ligamentous injuries, types of arthritis affecting the elbow [2] and even temporary paralysis of the arm. Focusing mainly on the physiotherapy exercise “Elbow Bend”, the exoskeleton assists the hinge like movement of the elbow for healing of the joints, tendon, ligaments and lateral muscles. [3]

Each prototype differs as per their energy input and actuation methods but the fundamental design and applications have stayed the same. Recent innovations have explored the idea of a wearable pneumatic sleeve that is easier on an injured or paralyzed limb. Such an assistive device can be used for: Rehabilitation of the entire arm as well as specific parts of the arm, as an assistive device in cases where the natural motion of the arm is hindered or even provide strength enhancement to the arm and as a prosthetic aid to a handicapped individual.

Morales and R. Badesa [4] conducted an in-depth research about the types of pneumatic rehabilitation devices made and about what kinds of robotic systems have been designed. Focusing mainly on the types of robotic exoskeleton systems and their characteristic features. In the work of Shen Yang, Ferguson et al. [5], their exoskeletal framework consists of a human upper extremity having seven degrees of freedom: elbow extension/flexion, forearm pronation/supination. An exoskeleton's degrees of freedom (DoFs) can be used to define structures.

On the other hand, the work of H. Xu et al. [6] describes a mechanism where the foundation is a wheeled frame that is used to transport items and support the motion system and the adjustment unit. The most critical aspect of this is the motion module. On exploring the plausibility of a soft and wearable pneumatic exoskeleton, Li, B, Greenspan, et al. [7], came up with a trademarked prototype called “Playskin Air” which depicts the modelled prototype exoskeleton which can assist a child of 11 years of age to be able to lift and drop down his arm as required via the usage of an air bladder which fits like a cuff around the user’s arm and can be inflated and deflated as required by a solenoid valve. On inflation of this cuff, the sleeve will tend to move the arm up and down when inflated and deflated respectively.

In further developments of such prototypes, the work of Rahman, M., Cristobal, et al. talks about the production, design and control of the ETS-MARSE. [8-9] The design entails the shoulder joint motion support portion- It has 2 motors and two connections (two links: A, B), and two potentiometers to help with horizontal and vertical flexion motion. The ETS-MARSE is made up of an arm chain, a sliding link (C), a fixed link (D), a motor, a tailored open-type bearing, a ring gear, a backlash preventive gear, and a potentiometer that helps with the rotation of the shoulder joint.

Zeroing in on exoskeletons built for the elbow exclusively, Mae Irshaidat et. al., [10] designed

ISSN: 0011-9342
Issue: 07 | Pages: 6876 - 6889

a soft wearable exoskeleton using muscle actuators and a control system which coordinates the delivery of the input between the actuators and the air tubes secured in a braided mesh. John Nassour et. al., [11] created a soft wearable pneumatic exoskeleton that reduces muscle activity and fatigue by combining a human-machine interface with pneumatic actuation to assist the elbow.

The underlying issue with the aforementioned prototypes is that most of these exoskeletons consist of a bulky framework dependent on an inconsistent pneumatic input. Although portable, these systems require the user to be seated at the exoskeleton and have them perform rehabilitative actions on the arms using a defined trajectory. The mechanisms which implement the use of robotic devices for actuation are heavy on the arm, not easily portable and are aesthetically compromised. Speaking of the soft and wearable kind of exoskeletons, the kinds that utilise air bladders and tubes, these components cannot be completely leak proof throughout the life cycle of the product and secondly, these apparatuses cannot be worn with ease and without discomfort as for the system's functioning, the bladder tightens around the arm and also presses down it. Thus, even though the circuit works, it cannot be worn for longer periods of time.

II. MATERIALS AND METHODOLOGY

The materials used in this exoskeleton are required to be mechanically strong, fatigue and corrosion resistant while not weighing down the arm. The designed components include the rack and pinion, the pulley which will assist this cable driven mechanism of the exoskeleton and the upper arm brace of the exoskeleton. Four materials were considered for the design of the spur rack and pinion- ABS Plastic, Aluminium 6061, Stainless Steel 304 and Cast Iron. Out of these, Stainless Steel 304 and Aluminium 6061 were selected due to their mechanical properties, corrosion resistance and compliance with the applied load cases. For structural compatibility and light weight, the pulley was designed to be made of Aluminium 6061. The arm brace was designed to be made of Stainless Steel 304 sheet metal to uphold the entire circuit assembly integrated with various components. Considering the interaction of this plate with the human arm, factors like sweat and moisture too could affect the sheet metal. To prevent unwanted corrosion and rusting of the base plate of the entire assembly, SS304 was deemed fit. The cable was selected to be an open cable [12] made of Nylon 6 due to its high yield strength, tensile properties and high elasticity. [13] The spring too is made of SS304 to bear high compressive stresses. [14]

2.1 Geometry and components of the exoskeleton

Design of the pulley, rack and pinion:

A pulley of diameter 60 mm and 5.5 mm groove with an 8 mm bore is selected considering the average human upper arm dimensions for aesthetic purposes and to rewind more of the cable that subtends the forearm on each revolution of the pinion gear. This pulley is made of Aluminium 6061 alloy.

The design of the rack and pinion: [15]

Spur gear (pinion):
Pitch diameter: 38 mm
Z= 36
Bore diameter: 10 mm
Module = 1
Material: Aluminum 6061
Toothed shaft (Rack):
Length: 400 mm
Type: 15 × 15
Material: Stainless steel 304

Pressure exerted on the bottom (piston disc) of the rack = 12 psi = 8.2737 N/ cm2

$$\text{Pressure} = \frac{Force}{Area} \tag{1}$$

Area of the piston disc = πr2 = π(10 mm)2 = 314.159 mm2 = 3.142 cm2

$$F = P \times A \tag{2}$$

Now that we have the tangential force that the Rack exerts on the pinion,

$$F_{2T} = F_N = \text{Tangential force} \tag{3}$$

$$F_N = mg.u + ma + Fe \tag{4}$$

Where: ma= 0; Fe = 0

ISSN: 0011-9342
Issue: 07 | Pages: 6876 - 6889

Thus, $u = \frac{F_N}{mg}$ (frictional factor) (5)

u= 0.264

Torque developed on pinion:

$T_N = F_N \times Rp$ (6)

where Rp is the radius of the pinion wheel

Design Torque (T_{NV}) = $T_N \times Sb$ (7)

Where Sb is the safety factor which is taken as 3 for vertical arrangements of rack and pinion.

Moment generated on the pulley due to rotation of the pinion at the same magnitude.

The pinion completes one revolution on meshing with the rack for 12 cm. An approximate rewinding distance is determined to be between 35-40 cm for the average human arm (through Pythagoras Theorem) to lift the forearm till a 150˚ angle. It was found that on 3 revolutions of the pinion, the pulley will rotate ~2 times to complete the rewinding process.

Design of the Spring:

Dimensions of the spring= 3.5 cm height and 3 cm width.

Maximum force that the spring will be subjected to-

$F = -kx$ (8)

x = 1.5 cm (allowed deformation)

k = was determined to be 15.1 N/cm.

Dimensions of the arm brace:

The arm brace was designed so as to encase the upper arm of the elbow only, to allow free movement of the elbow without any hindrances. The average human upper arm is around 30-35 cm long [16] and the brace was designed to be 26.5 cm long so as to fit comfortably and aesthetically without restricting the movement of the upper arm entirely. This brace consists of a Stainless Steel 304 sheet metal (265 × 40 × 2.5) on which all the components are mounted. This brace is secured at both ends with an elastic band to ensure no oscillatory movement occurs during the operation of the exoskeleton.

ISSN: 0011-9342
Issue: 07 | Pages: 6876 - 6889

The exoskeleton consists of various standard components assembled along with the designed ones to provide maximum efficiency and comfort to the user while rehabilitating their elbow-

- A Nickel metal Hydride rechargeable battery pack [17] which is required to ensure a minimum operation of 10 hours between 2 charging cycles to provide a power input to the circuit.
- A DC/DC relay [18] is needed because-

Battery voltage is un-stabilized power and fluctuates based on current drawn by the circuit. As batteries discharge their voltage drops. The relay here does the job of a transistorized constant voltage transformer ensuring steady state power is provided for safe functioning of the equipment down the chain. It steps down the 24V voltage from the nickel hydride battery to 12V for the input required by the compressor.

- A compressor is need to supply an input of 12 psi continuous supply and it generates adequate torque to cause the required degree of rotation. Model number: D250BL [19]. It is compact and light weight.
- A mini pressure regulator [20] is required because it Provides flexibility to regulate output pressure at various levels 5-12 psi. As the arm weight of humans is not a constant and condition of the damaged arm may demand variable speed of flexion, the regulator permits efficient pressure supply regulation.

Table 1. Components with their dimensions and weights

Components	Dimensions (mm)	Weight (g)
Spring	35 × 20	2
Toothed shaft (Rack)	200 × 15 × 15 (20 m base disc)	10
Gear (pinion)	38 × 15 (Z= 36)	15
Pulley	60 × 15	88
Mounting Bracket	265 × 40 × 2.5	49.8
Compressor	41 × 27 × 17	25-34
Pressure Regulator	72 × 38 × 50.8	113
DC/DC relay	42.82 × 22 × 17.42	16.78
Air tube	2.4 mm bore	Negligible
Air bladder	80 × 40	2.5-3

ISSN: 0011-9342
Issue: 07 | Pages: 6876 - 6889

2.2 Methodology:

An electrical input from the nickel metal hydride battery pack activates the pneumatic circuit via a battery-operated mini air compressor. This air compressor passes air through a pressure regulator which allows for a range of pressure from 5psi to 12psi and can be adjusted as per the user's comfort. This regulator passes air into an air bladder instead of a heavy pneumatic cylinder. This air bladder pushes a piston which subtends a toothed shaft or the rack. This toothed shaft meshes with a gear of similar tooth profile- the pinion, which is in turn fitted onto the face of a pulley. This pulley will then rewind a cable which is attached to a band that encases the forearm that needs to be lifted. Once the air is released, this shaft drops down onto a spring to reduce the jerk and this downward meshing subsequently rotates the gear in the opposite direction which then unwinds the pulley and lowers the cable- thus lowering the forearm gradually.

Actuation and Driver unit:

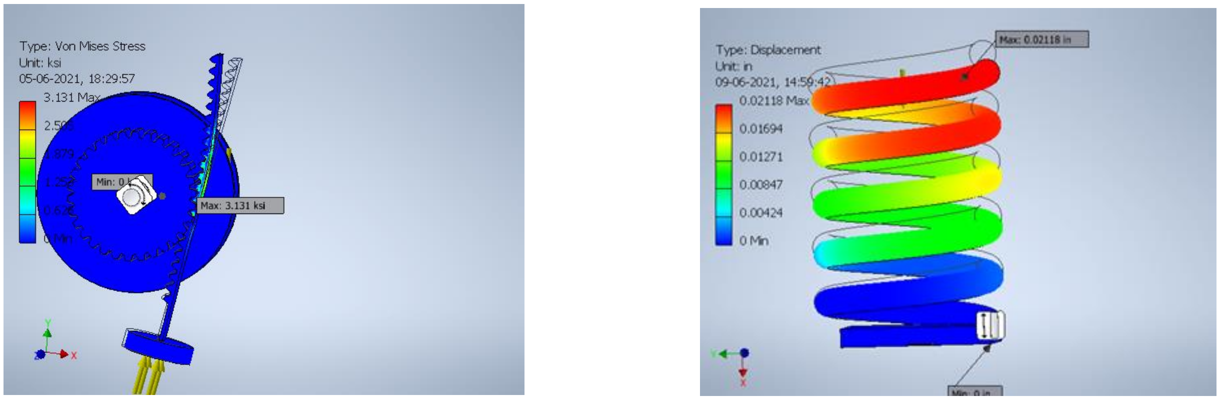
The actuation is provided by a rechargeable battery pack which is estimated to be able to provide power to the circuit of the exoskeleton for up to 13 hours due to its capacity of 1300 mAh running at 0.1 a per cycle. Through the datasheet of this battery pack, it is understood that even if the rotor of the compressor rotates at 1.5 times the original speed, the battery can power the circuit for a minimum of 7 hours. A pneumatic driver unit is chosen due to its light weight and omission of the need to include a storage tank for hydraulic oil or water which could compromise on the portability of the system and be a cause of discomfort to the user. The air bladder is installed to contain the air input with a uniform magnitude of pressure, provide a gap time to prevent jerky motion of the arm on actuation and to host a spring upon which the piston can fall back on, thus reducing the jerk on the arm when the circuit is switched off.

III. DATA ANALYSIS AND INTERPRETATION

3.1. Verification of designed components

To verify whether the materials selected for the designed components were correct, an FEA was conducted.

Design Engineering
ISSN: 0011-9342
Issue: 07 | Pages: 6876 - 6889



(a) (b)

Figure 1. (a) Simulation results depicting the Von mises stress on the components; (b) displacement observed in the spring

Table 2. Numerical range values of FEA parameters

Parameter	Minimum	Maximum
Von Mises Stress	0.00184413 N/mm2	21.58927 N/mm2
1st Principal Stress	-10.55842 N/mm2	21.000879 N/mm2
3rd Principal Stress	-34.460617 N/mm2	6.7576 N/mm2
Displacement	0.041868 mm	6.82929 mm
Safety Factor	0.926385 ul	15 ul
Contact Pressure	0 N/mm2	65.259015 N/mm2

The Von Mises Stress value determines whether at that point the material or component will fracture or showing yielding. [21] As displayed by the table and legends, the maximum value of Von Mises stress was found to be at the point of contact between the Rack and pinion. Yet, this value is well below the maximum specified limit and hence it can be concluded that the pulley, rack and pinion will not show any yield and hence the design is safe from fracturing when the respective load cases are applied to it. The 1st Principal Stress gives the value of the maximum tensile stress that the designed components are facing on conducting an analysis. As the upper limit is observed to be 3.046 ksi or 21.0014 N/mm2. The maximum tensile stress appears to be at the point of contact between the rack and pinion with a value that is well below the upper limit of this stress. This adheres to the force developed in the component as a reaction to the forces applied on it. In response to the tangential force and torque developed between the rack and pinion, this tensile stress is observed to be moderate through which we can expect a negligible elongation in the component but not detrimental to the design. As observed in the table and the figure of the 3rd Principal stress, the upper limit of this stress is 0.98 ksi or 6.756 N/mm2. The analysis results show that the compressive stresses are well within the permissible ranges and that no visible

ISSN: 0011-9342
Issue: 07 | Pages: 6876 - 6889

compression will be observed when the components are in operation. The rack shows a significant displacement of around 6 mm from its original position at its extreme end once the pinion meshes with the bottom half of the rack. Contact pressure is a ratio of the applied load to the area of contact.

Table 3. Numerical Range of FEA parameters for the spring

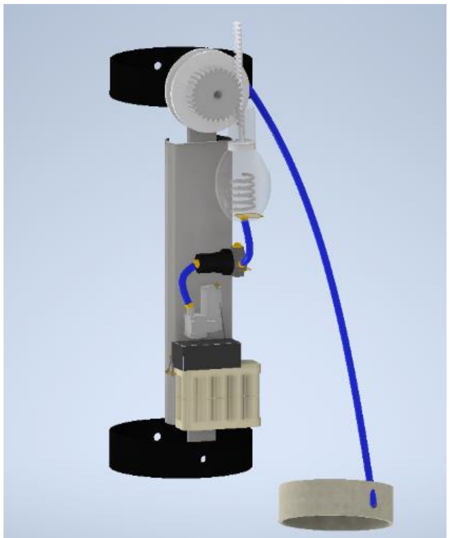
Parameter	Minimum	Maximum
Von Mises Stress	0.0025912 N/mm2	279.70789 N/mm2
1st Principal Stress	-71.38349 N/mm2	167.55432 N/mm2
3rd Principal Stress	-240.999 N/mm2	34.384223 N/mm2
Displacement	0 mm	0.5379 mm
Safety Factor	0.89379 ul	15 ul

The Von Mises Stress induced in the spring upon being subjected to a compressive load from the piston of the rack is almost 0 throughout the spring except for the bottommost coil which shows a slightly higher value. This means that this point- although safe from fracture or yielding, is definitely the weakest point of the spring and on increasing the magnitude of the load applied, it could show a possible fracture. It is observed that the topmost coil shows the least amount of tensile stress induced but as we descend the coiled in the spring, the tensile stresses start to increase in magnitude. Referring to the table and legends in the figure, it is clear that the value of maximum tensile stresses generated is well below the critical limit and hence, either no elongation of the component will be observed or there will be a negligible elongation in terms of micro or nano meters. The 3rd Principal stress or the compressive stresses induced in the coils of the spring show that although the value of compressive stress is high, it has not reached the upper limit that is 34.384 N/mm2. The compressive stress is observed to be higher in the first 2 coils as well as the mount of the spring. These results are complaint with the predicted compression that would occur in the spring on application of the loads

The displacement that occurs in the spring is as expected. The top two coils displace the most on being subjected to the descending force of the piston of the rack. The spring shows a deformation of 2 mm. The safety factor of the spring is low due to the actual deformation being low, yet the lowermost coil has the least factor of safety. This value is so low that with an increase in force it could show a point fracture in the coil. However, the overall structure of the spring is such that it is designed for the maximum load case which won't be the condition during operation. Thus, we can determine that the design of the spring is safe and compatible.

Design Engineering
ISSN: 0011-9342
Issue: 07 | Pages: 6876 - 6889

3.2 FEA of the pneumatic exoskeleton prototype:


(a)

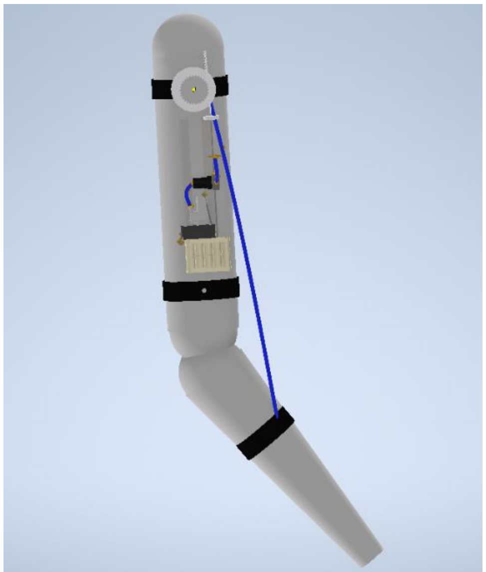

(b)

Figure 2. (a) Pneumatic exoskeleton assembly and; (b) Pneumatic exoskeleton assembly represented on a mannequin arm

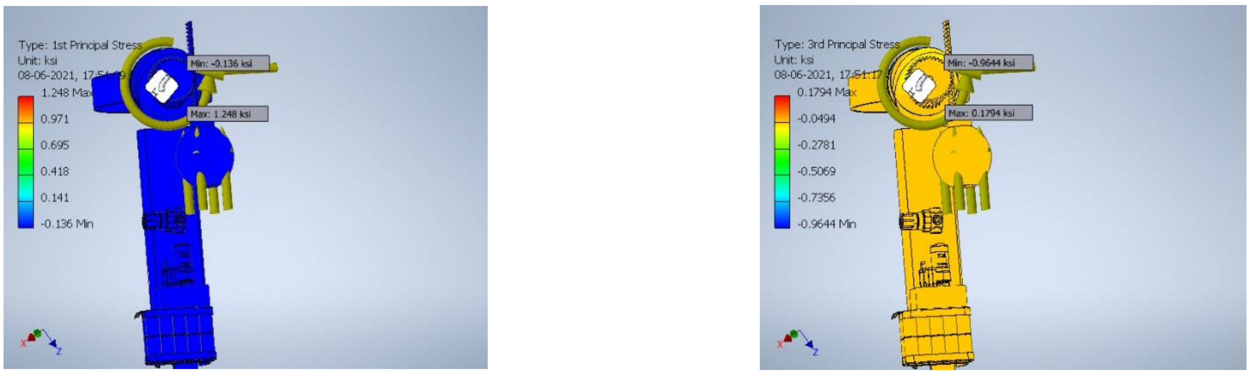
Table 4. Numerical values of FEA parameters in the analysis conducted on the prototype

Parameter	Minimum	Maximum
Von Mises Stress	0 N/mm2	7.40703 N/mm2
1st Principal Stress	-0.93647 N/mm2	8.60507 N/mm2
3rd Principal Stress	-6.44225 N/mm2	1.236829 N/mm2
Displacement	0 mm	0.00000974524 in
Safety Factor	15 ul	15 ul
1st Principal Strain	-0.0000000800404 ul	0.0000395382 ul
3rd Principal Strain	-0.0000305301 ul	0.0000000226141 ul
Contact Pressure	0 N/mm2	15.10634 N/mm2

ISSN: 0011-9342
Issue: 07 | Pages: 6876 - 6889

On closely analysing the effect of these stresses and strains on the prototype:

On conducting an FEA of the entire prototype, it was observed that the Von Mises Stresses developed in the entire prototype were almost negligible or rather 0 N/mm2. This means that the entire prototype including the arm brace will not fracture or yield at any point and is hence safe to use on the human arm.



(a) (b)

Figure 3. (a)1st principal stress on the exoskeleton; (b) 3rd principal stresses on the exoskeleton- they were observed to be high.

The tensile stresses developed in the entire prototype range from -0.9376 N/mm2 to 0.972 N/mm2 which are extremely low and almost negligible in terms of having an impact on the design. Thus, we can expect no visible or obvious elongation in any of the components or the prototype assembly as a whole. The 3rd principal stress denotes the compressive stresses induced in the entire prototype which in this case range from -1.9174 N/mm2 to -3.4060 N/mm2. Although the magnitude of the compressive stresses induced in on the higher side of the range mentioned in the table, it is not high enough to cause a deformation in the prototype and is within the permissible limit. Hence the design is safe from elongation and compression of the components and assembly.

The most significant displacement is observed to occur in the pulley as expected. Since the pneumatic circuit and rack and pinion are designed to rewind the pulley and subsequently lift the forearm, the maximum displacement is expected to occur in the pulley itself. The rest of the prototype stays stationary and does not move when fitted on the human arm. The safety factor of the entire exoskeleton is shown to be high (15) as the prototype needs to be safe and risk free to be usable on the human arm.

The analysis shows that the exoskeleton might exert a slightly high 3rd principal stress and strain on the arm but apart from this, there is no contact stress or extra deformation observed which could cause any backlash on the arm. The factor of safety is high due to the risk of injury or discomfort on the arm and overall, the design is safe and compatible for use.

ISSN: 0011-9342
Issue: 07 | Pages: 6876 - 6889

3.3 Multi Body Dynamic Analysis of the exoskeleton on a human arm:

After the analysis deemed the prototype safe, its operation on a human arm was simulated to determine the time period in which angular displacement of the forearm would take place, angular velocity of the forearm and the arc length covered by the forearm in motion.

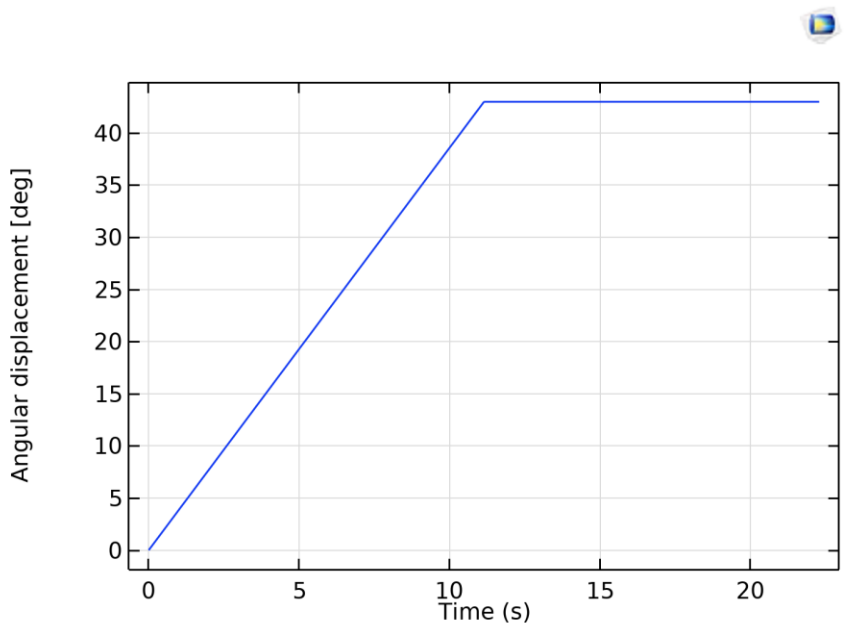


Figure 4. Plot displaying angular displacement of the graph Vs time

Through Simulative data, it was concurred that the forearm would be lifted by 45˚ from its initial position in about 11.42 seconds. The exoskeleton is designed to lift the forearm from 0˚ to 150˚ degrees in the entire period of operation. Since the motion is synchronized and continous, it is expected that the exoskeleton will be able to raise the forearm upto 150˚ in approximately 38.067 seconds.

Design Engineering

ISSN: 0011-9342
Issue: 07 | Pages: 6876 - 6889

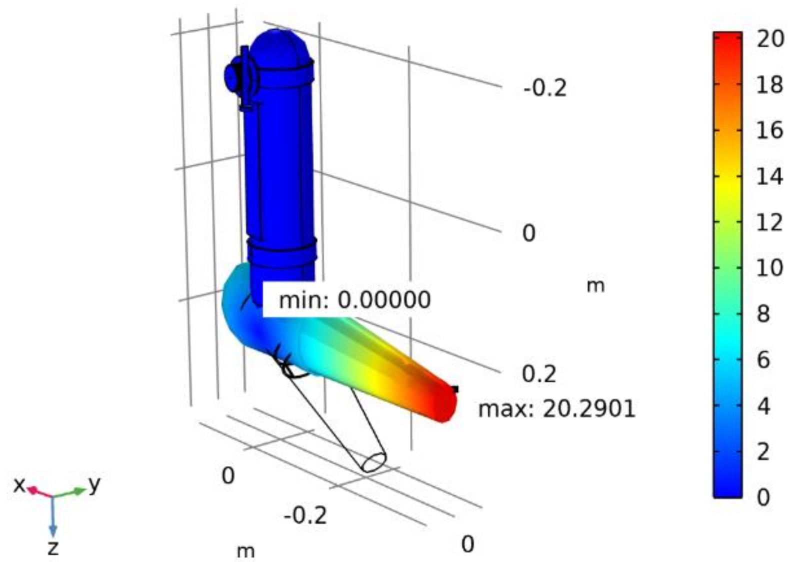


Figure 5. A simulation depicting the displacement of various points along the forearm when the exoskeleton in is operation.

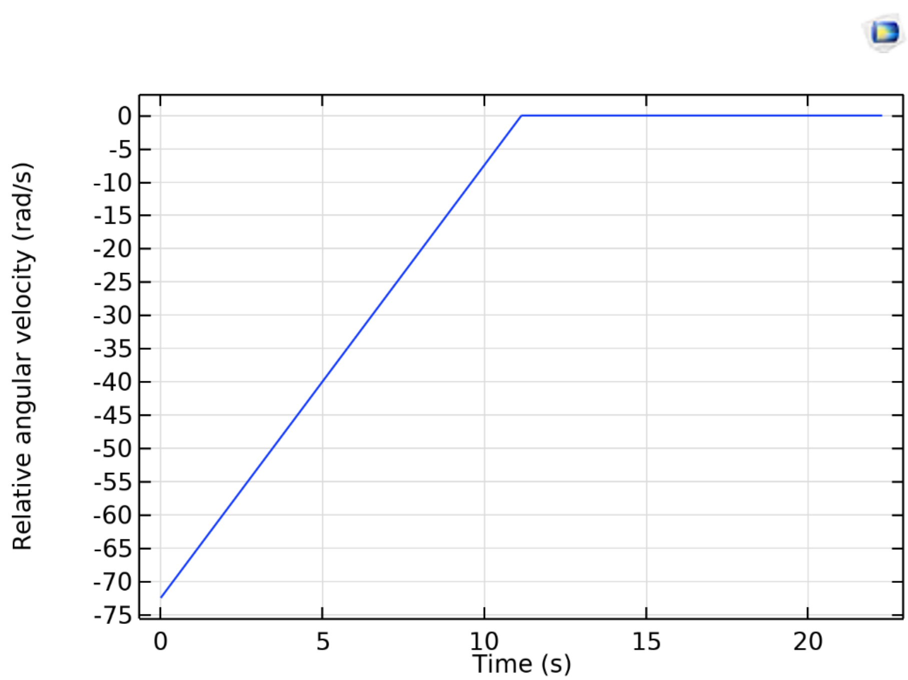


Figure 6. Plot displaying the angular velocity of the forearm Vs time taken

Since the motion of the arm is presented in a counterclockwise manner, the angular velocity depicted on the graph is portrayed with a negative magnitude. From the graph it is evident that angular velocity increases linearly but shows a constant nature after the forearm is raised by 45˚. Thus, it can be concluded that angular velocity rises continously till the point of the 45˚ and then raises the arm at a constant angular velocity for the rest of the operative period.

III. CONCLUSION

On comparing with pre-existing prototypes, the design is more easily portable, light weight, aesthetic and ergonomically viable. This prototype concentrates on the rehabilitation of the elbow exclusively, unlike the pre-existing prototypes. Since this prototype weighs 551g in total (as per analysis results), it is convenient to wear on the upper arm while carrying on with daily activities. The components are scalable as per the user's arm dimensions. The noise generated is negligible as per the component datasheets. In comparison with pre-existing prototypes, this model is convertible to a strength enhancement device as well as a rehabilitative device. It is portable, rechargeable and can be used daily.

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