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Exoskeleton Arm With Pneumatic Muscle Actuation

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***Abstract*- For centuries now, humans have developed machines for tasks which are too labor intensive for species cannot do. So, creative imagination and subtle engineering has led to the development of the powered exoskeleton. It is a device which can be worn over the human body. A powered exoskeleton enables a human to perform tasks which are beyond the physical prowess by amplifying the muscular movements. We have outlined the process of developing an exoskeleton arm which increases the load lifting capacity of a human. The primary actuation of the exoskeleton relies on the longitudinal contraction of a group of Mckibben muscles or pneumatic air muscles (PAMs). The PAM has a thin-walled, rubber bladder placed inside an axially stiff but radially compliant braided sleeve. As the rubber bladder expands due to an increase in pressure, the diameter of the combined sleeve and bladder assembly easily changes in the radial direction and the PAM shortens in the axial direction. As the consequence of this interaction, a large contraction force produced can perform external work at rapid rate. However, non-linearity exists as the pressure changes in the bladder because its area expands proportionally to the square of the diameter. Also as the outer sheath material moves, its length is dependent on trigonometric relationships involving the outer sheath material, which are non-linear. Since, one end of the PM is attached above the elbow joint of the armature and the other end below the elbow joint (PAM forms the hypotenuse and biceps and forearm forms the other two sides of the triangle), the armature performs the lifting action due to muscle contraction.**

Keywords— Pneumatic Artificial Muscle (PAM), Exoskeleton, Exoarm, Robot.

I. INTRODUCTION

One of the proposed main uses for an exoskeleton would be enabling a soldier to carry heavy objects (80–300 kg) while running or climbing stairs. Not only could a soldier potentially carry more weight, he could presumably wield heavier armor and weapons. Most models use a hydraulic system controlled by an on-board computer. They could be powered by an internal combustion engine, batteries or fuel cells. Another area of application could be medical care, nursing in particular. Faced with the impending shortage of medical professionals and the increasing number of people in elderly care, several teams of Japanese engineers have developed exoskeletons designed to help nurses lift and carry patients.

Exoskeletons could also be applied in the area of rehabilitation of stroke or Spinal cord injury patients. Such exoskeletons are sometimes also called Step Rehabilitation Robots. An exoskeleton could reduce the number of therapists needed by allowing even the most impaired patient to be trained by one therapist, whereas several are currently needed. Also training could be more uniform, easier to analyze retrospectively and can be specifically customized for each patient. At this time there are several projects designing training aids for rehabilitation[1].

II. LITERATURE SURVEY

2.1 HISTORY

The first exoskeleton structure to assist walking, jumping and running was invented by Nicholas Yagn in 1890. It contained compressed gas bags to power it. The first totally functional and powerful exosuit was developed by General Electric in association with United States Military in 1960s which lifted 110kg with effort reduction by the factor of 10. However, pneumatic muscles were first developed in the 1950s for use of artificial muscle and then commercialized by Bridgestone Rubber (Japan) in 1980s. It was then when PAMs were used in an exoarm[2][3].

2.2 COMPARISION WITH EXISTING TECHNOLOGY

The main distinction between the design described in this paper and the existing ones is that a power assist type exoskeleton has been selected rather than a more expensive power amplification device. The cost required to design and manufacture such a device was reduced. But the main difference is apparent through the selection and control of Mckibben air muscles as the main power actuator. Some implementations, which use hydraulic actuators, need internal combustion engines to compress the non-compliant fluid. This implementation uses a limited supply of pressurized, compliant gas, which reduces the power consumption by a considerable amount. Another major difference is the use of Mckibben air muscles provides with a higher power-to-weight ratio than an exoskeleton using DC motors. Finally, the implementation of a combination of flex and EMG sensors, allows monitoring of the signals sent to the muscles, and potentially reduces the effort required to activate the sensors.

III. PROBLEM DEFINITION

3.1 REQUIREMENTS OF AN UPPER-LIMB EXOSKELETON

The requirements of an active upper-limb exoskeleton are different in accordance with the purpose of the device. The

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upper-limb exoskeleton exoarms also directly interact with the human user, safety becomes an important requirement. The exoskeleton for wrist motion assist has provided the axes deviation of wrist flexion/extension axis and wrist radial/ulnar axis. Movement of the center of rotation of shoulder joint according to the upper-arm motions must be considered to cancel out the ill effect caused by that in design. If upper-arm motions also have to be assisted by the exoarm as well as forearm motion, a mechanism that allows moving of the center of rotation of the shoulder joint must be considered in the upper-limb exoskeleton. This mechanism is considered in to cancel out the ill effects caused by the position difference between the center of rotation of the exoarm shoulder and that of the human shoulder. The mechanical singularity should not be occurred within the workspace of the exoarm. Some designs have specially considered this in their designs. Although above explained important requirements have been fulfilled, researchers should consider following aspects. The exoarm for wrist motion assist should have individual axis for wrist flexion, wrist extension, wrist radial deviation and wrist ulnar deviation motions. Mechanical designs of upper-limb exoskeletons can be further improved to reduce their inertia. The weight of the exoarm system also affects its portability [4].

3.2 PAMs FOR EFFICIENT MOTION ASSIST

The problem with exoskeletons as of now, lies primarily in its cost and the requirement of a high density power supply. To achieve the best power to weight ratio, most exoskeletons use hydraulics with non-compliant fluids. However, use of non-compliant fluids leads to higher power consumption and heavier compression gear. Hence, for the scope of this project, we have opted for a lighter system which uses a compliant fluid (air) in the form of Pneumatic muscles.

3.3 SENSING MOTION OF THE HUMAN ARM AND CONTROLLING THE MUSCLE

Currently, the methods we propose on using to proportionally control the air muscle involve a combination of carefully placed EMG electrodes and flex sensors. These will provide the feedback based on the users motion. The plan of control involves a feedback transducer positioned on the joints of the mechanical harness which gives angular feedback. This limits the linear motion of the PAMs and the input to the system is given based on the EMG and flex signals generated by the arm [5].

IV. METHODOLOGY

4.1 ARM FORCE CALCULATIONS

One of the main data requirements for the commencement of the design phase was the distribution of forces on the human arm. This was done to have a better understanding of the muscular reduction already available to an average human. A layout of the positions at which the forces would act was made in SolidWorks, which allowed estimation of the forces effectively. Force exerted by the bicep muscle as a function of distances at which the muscle is attached from the rotary joint is given as:

$$F_b = \frac{[r_2 \times M_a + r_3 \times M_l]}{r_1} \qquad (4.1.1)$$

The table 4.1.1. shows the calculations of the forces exerted on the arm.

r_1	r_2	r_3	M_a	M_l	F_b
0.04	0.16	0.38	2.5	30	295
0.05	0.16	0.38	2.5	30	236
0.06	0.16	0.38	2.5	30	196.66667
0.07	0.16	0.38	2.5	30	168.57143
0.08	0.16	0.38	2.5	30	147.5
0.09	0.16	0.38	2.5	30	131.11111
0.1	0.16	0.38	2.5	30	118
0.11	0.16	0.38	2.5	30	107.27273
0.12	0.16	0.38	2.5	30	98.333333
0.13	0.16	0.38	2.5	30	90.769231
0.14	0.16	0.38	2.5	30	84.285714
0.15	0.16	0.38	2.5	30	78.666667
0.16	0.16	0.38	2.5	30	73.75

Table: 4.1.1: Force on harness versus lifted load

F_b	α	θ	β	β(rad)	sine(β)	F_{yc}
295	20	25	85	1.482778	0.9961289	293.858016
295	20.5	25	85.5	1.4915	0.9968577	294.0730195
295	21	25	86	1.500222	0.9975107	294.265651
295	21.5	25	86.5	1.508944	0.9980878	294.4358957
295	22	25	87	1.517667	0.998589	294.5837407
295	22.5	25	87.5	1.526389	0.9990142	294.7091748
295	23	25	88	1.535111	0.9993634	294.8121883
295	29	25	94	1.639778	0.9976217	294.2984083
295	29.5	25	94.5	1.6485	0.9969826	294.1098635
295	30	25	95	1.657222	0.9962676	293.8989439
295	30.5	25	95.5	1.665944	0.9954768	293.6656654
295	31	25	96	1.674667	0.9946103	293.4100458
295	31.5	25	96.5	1.683389	0.9936682	293.1321045
295	32	25	97	1.692111	0.9926504	292.8318627
236	20	40	70	1.221111	0.9394806	221.7174228
236	20.5	40	70.5	1.229833	0.9424331	222.4142075
236	21	40	71	1.238556	0.9453139	223.0940716
236	21.5	40	71.5	1.247278	0.9481227	223.7569635
236	22	40	72	1.256	0.9508595	224.4028327
236	22.5	40	72.5	1.264722	0.9535239	225.03163
236	23	40	73	1.273444	0.9561157	225.6433078
236	23.5	40	73.5	1.282167	0.9586348	226.2378193
236	24	40	74	1.290889	0.961081	226.8151194

Table: 4.1.2: Force on harness versus lifted load

where,

r_1: Distance of bicep attachment from elbow joint in meters

r_2: Distance of CG from elbow joint in meters

r_3: Distance of load from elbow joint in meters

F_b: Upward force exerted by the bicep in Newtons

M_a: Mass of Forearm in Kg

M_l: Mass of load in Kg

α : Angle of flexion in degrees

β : Angle of muscle w.r.t. forearm in degrees

θ : Derived angle in degrees

4.2 ACTUATOR DESIGN

Once the required force was calculated, possible dimensions for the pneumatic muscle assembly can be estimated. Since the force characteristics of a PAM are non-linear in nature, only maximum stall force was calculated. A bulky assembly was not desirable, since the apparatus is to be worn by a human. It also has to be form fitting, so as to allow the maximum possible range of motion available to the human body. Hence, the design was made, to strike a balance between power and size. The diameter for the actuator was determined based on the pneumatic muscle equation [6]. Force generated by the muscle can be given as:

$$F = P\frac{dV}{dX} \qquad (4.2.1)$$

F = Force generated

P = Fluid pressure

V = Volume of actuator

X = Contraction of actuator

Calculated values are shown in the table below-

Dia	Force(N)	force(kg)	4*force	reduction force
0.01	21.67699	2.167699	8.670796	2.435616776
0.011	26.22916	2.622916	10.49166	2.947096299
0.012	31.21486	3.121486	12.48595	3.507288158
0.013	36.63411	3.663411	14.65364	4.116192352
0.014	42.4869	4.24869	16.99476	4.773808882
0.015	48.77323	4.877323	19.50929	5.480137747
0.016	55.49309	5.549309	22.19724	6.235178948
0.017	62.6465	6.26465	25.0586	7.038932484
0.018	70.23345	7.023345	28.09338	7.891398356
0.019	78.25393	7.825393	31.30157	8.792576563
0.02	86.70796	8.670796	34.68318	9.742467106
0.021	95.59552	9.559552	38.23821	10.74106998
0.022	104.9166	10.49166	41.96665	11.7883852
0.023	114.6713	11.46713	45.86851	12.88441275
0.024	124.8595	12.48595	49.94378	14.02915263
0.025	135.4812	13.54812	54.19247	15.22260485
0.026	146.5364	14.65364	58.61458	16.46476941
0.027	158.0253	15.80253	63.2101	17.7556463
0.028	169.9476	16.99476	67.97904	19.09523553
0.029	182.3035	18.23035	72.92139	20.48353709
0.03	195.0929	19.50929	78.03716	21.92055099
0.031	208.3159	20.83159	83.32635	23.40627722

Table: 4.2.1: Actuator length, Dia(diameter) and force lifting capacity.

Diameter in meters, Force in Newtons and all other forces are converted to Kg.

4.3 COMPONENT SELECTION AND ACTUATOR PLACEMENT

After the actuator's dimensions were fixed, the effect of the actuator placement on the output force could be analyzed. This was done using SolidWorks Layout designer. A sample layout using arm lengths, measured practically, was created and a stable geometry with length manipulation, for the actuator lengths, both extended and contracted was determined. Figure 4.3.1 shows the actuator placement and the arm geometry.

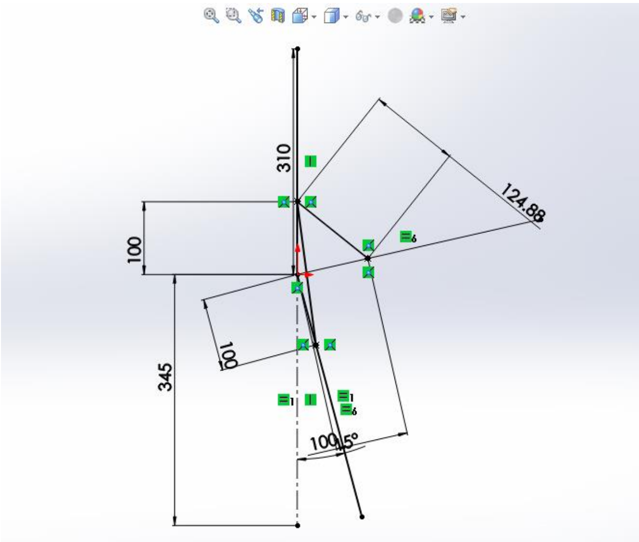

Fig: 4.3.1: SolidWorks model showing PAM actuator placement (All dimensions in mm).

The next part of the preparation involved selecting the silicon tube for the elastic chamber inside the mesh of the pneumatic muscle. Estimations for the pressure and temperature standards for the tube, were made based on the force requirements. The pressure required would be a direct function of the force applied. The temperature was calculated based on the pressure-volume relation from Boyle's gas law. Based on the information from a market survey, a list of possible configurations was generated. The tube was selected having inner diameter 18mm, outer diameter 24mm, i.e., 2mm wall thickness. Minimum wall thickness was calculated using Barlow's formula.

$$P = \frac{2St}{D} \qquad (4.3.1)$$

where,

P = Pressure in tube [Max pressure taken at 6 bar]

S = Ultimate tensile strength

t = Tube wall thickness

D = Tube diameter

4.4 ADAPTER AND FIXTURE DESIGN

SolidWorks has been used to design the actual physical components of the PAM. Material properties for the tube and adapter fitting is specified and the total weight of the system was estimated to be 150gm. The pneumatic fittings and the required adapters were designed to reduce leaks and withstand the expansion stress of the tube. Figure 4.4.1 shows the mechanical design for the PAM tube.

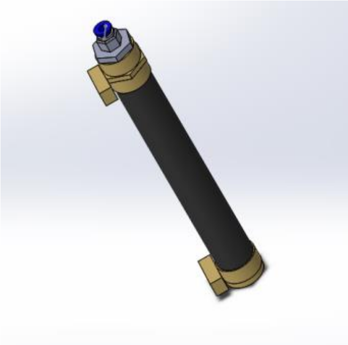

Fig: 4.4.1: SolidWorks 3D CAD of the PAM with adapter

4.5 PNEUMATIC VALVE SELECTION

Required valve is a three stage valve, In-Hold-Exhaust. For this purpose a proportional valve is available in the market. But two 3/2 pneumatic valves have been used to reduce the cost. It works exactly as a proportional valve and can be controlled at a wide range of open-close frequency. The arrangement is shown in figure 4.5.1.

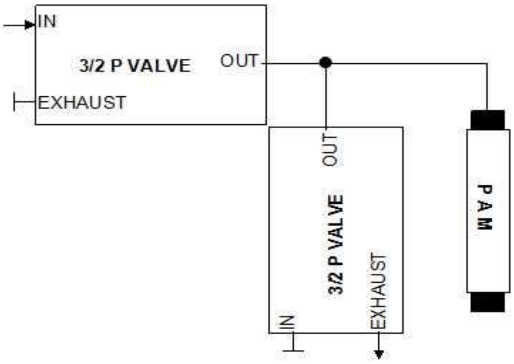

Fig: 4.5.1: Valve configuration for air flow control

4.6 SENSOR SELECTION

EMG Sensor Control: Electromyography (EMG) is an electrodiagnostic medicine technique for evaluating and recording the electrical activity produced by skeletal muscles. EMG signals are essentially made up of superimposed motor unit action potentials (MUAPs) from several motor units. For a thorough analysis, the measured EMG signals can bedecomposed into their constituent MUAPs. MUAPs from different motor units tend to have different characteristic shapes, while MUAPs recorded by the same electrode from the same motor unit are typically similar. Notably MUAP size and shape depend on where the electrode is located with respect to the fibers and so can appear to be different if the electrode moves position. Rectification is the translation of the raw EMG signal to a single polarity frequency (usually positive). The purpose of rectifying a signal is to ensure the raw signal does not average zero, due to the raw EMG signal having positive and negative components. It facilitates the signals and process and calculates the mean, integration and the fast fourier transform (FFT). The rectified and processed signals from the MUAPs are then mapped into singular instances which determine the motion of the muscle being sensed. This will be used to control the Pneumatic muscles on the harness[7][8].

Figure 4.6.1: proposed placement of EMG sensors

4.7 SIGNAL APMLIFIER AND CONDITIONER CIRCUIT

Amplifer circuit for EMG signal amplification consists of four stages. 1st stage is the differential amplifier stage using INA106. 2nd stage is for voltage gain. 3rd is the rectifier stage and 4th is the inverter stage. The total gain of the circuit is 100. The circuit introduces the noise of 50mV making it difficult to extract the original signal. For which we have used Kalman filtering which will be discussed later.

4.8 VALVE CONTROL CIRCUITRY

Required valve is a three stage valve, In-Hold-Exhaust. For this purpose a proportional valve is available in the market. But two 3/2 pneumatic valves have been used to reduce the cost. It works exactly as a proportional valve and can be controlled at a wide range of open-close frequency. The arrangement is shown in figure 4.2.3.

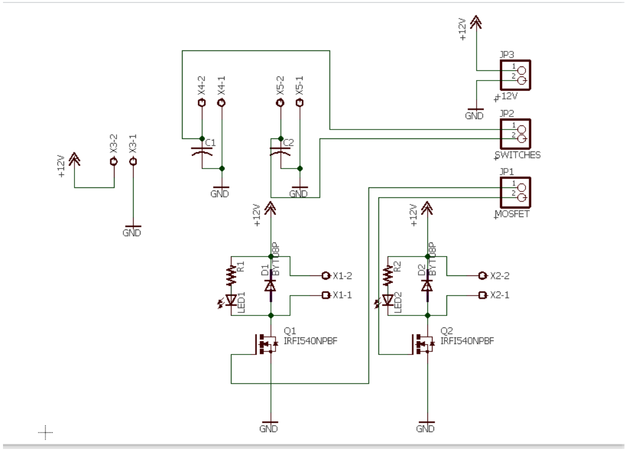

Fig: 4.8.1: Valve configuration for air flow control

The solenoid valves will be controlled via control signal generated by a microcontroller. The microcontroller will generate PWM signal of a specific frequency for the proportional on-off action. This frequency will depend on the feed of the processed EMG signal to microcontroller.

4.9 PROGRAM

MATLAB Kalman filtering: Matlab program running the readings from the EMG electrode through a Kalman filter. The conditioning circuit sends the signal to the microcontroller's ADC. The quantized signal is then sent to Matlab through the serial port.

Muscle control program: Once the Kalman filter coefficient is obtained, it is used to create an embedded version of the program. The platform used is an Arduino Mega 2560. The program detects peaks in the EMG signal and generates a PWM signal to control the pneumatic valves. The change in number of peaks increases or decreases the duty cycle of the signal going to the valves. If no peaks are detected the valves are set to hold mode. The following figures show the raw signal and the signal at the different stages of processing :

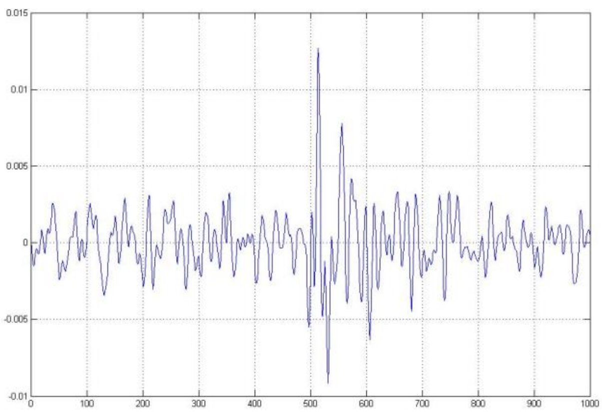

Fig 4.9.1: Raw EMG Signal.

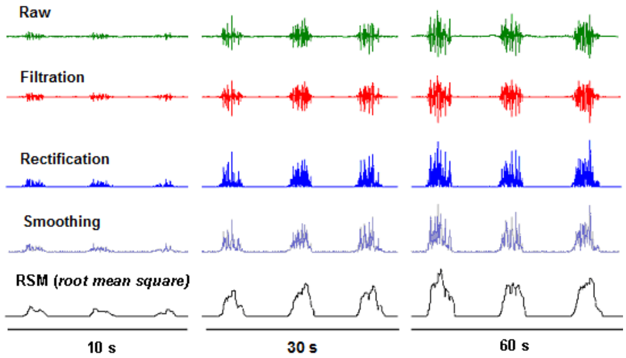

Fig 4.9.2 : Signal at different stages of processing[9].

V. IMPLEMENTATION

5.1 PROTOTYPING

A prototype of the rib-cage assembly was made using PVC circular pipes with 1" diameter. The prototype served two functions: It allowed us to confirm the dimensions we had designed with. It also allowed us to optimize the positions of the joints. A similar harness was made for the shoulder and arms. Thus we were able to successfully optimize and verify the dimensions and orientations of the joints.

5.2 PNEUMATIC MUSCLE ASSEMBLY

As per the required force lifting capabilities for this project, each component of the pneumatic muscle was chosen carefully. As per the force calculations mentioned earlier following specifications were incorporated for the assembly.

1. Silicon tube with the inner diameter 15mm and outer diameter 18mm, wall thickness 3mm.
2. Braided sleeve of the diameter 16mm.
3. Hose with outer diameter 13mm.
4. PU connector ¼ 6mm.

Figures below shows the separate components and the assembled version of the pneumatic muscle.

Fig 5.2.1: Components of the muscle

Fig 5.2.2: Assembled muscle

5.3 FABRICATION OF ALLUMINIUM CHASSIS

A fixture for welding according to the measurements in the CAD model using SolidWorks was made. Before selecting the material for the harness, we did the Finite Element Analysis (FEA) on SOLIDWORKS which is a numerical technique for finding approximate solutions to boundary value problems for partial differential equations. It subdivides a large problem into smaller, simpler parts, called finite elements. SOLIDWORKS Simulation uses the displacement formulation of the finite element method to calculate component displacements, strains, and stresses under internal and external loads. The welding will be done in the coming week. The welding fixture and the chassis are shown in the figure 5.3.1 and 5.3.2.

Fig 5.3.1: Welding fixture

Fig 5.3.2: Fully fabricated/welded chassis

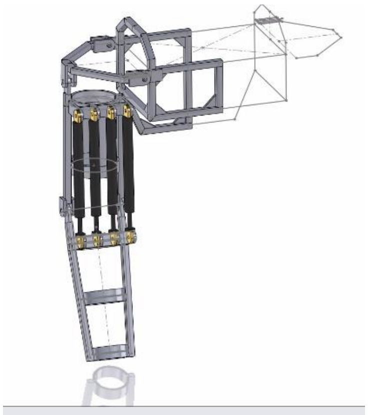

Fig 5.3.3: Screen shot of the SolidWorks CAD model of complete mechanical assembly

VI. LOAD AND CONTRACTION TESTING

Tests for the load and contraction data were performed by fixing the PAM vertically and attaching a load to the free end. The tests used a 10 Kg load and tested the contraction at various pressures. Table 6.1 shows of the results.

All measurements fro a PAM with silicon tube dimensions 18x15mm with 20 mm diameter mesh			
Load (in Kg)	Restin length (in mm)	Contracted length (in mm)	Pressure (in bar)
No	290	260	All pressures from 2 to 6
10	290	282	2
10	290	273	2.5
10	290	270	3
10	290	265	3.5
10	290	263	4
10	290	260	4.5
10	290	260	5
10	290	260	5.5
10	290	260	6

Table: 6.1: Load test results

VII. PROPOSED CONTROL STRATEGY

After the required force and contraction is achieved the proposed control strategy for the PAM involves attaching it to a harness and measuring the displacement in response to various input signals. The input and output data will then be logged and the System Identification toolbox in MATLAB will be used to derive the state space model for the system. Once the state space model has been derived, a PID controller will need to be realized for effective control of the muscle. By applying constant mass M to end of the system, the dynamics of the PAM can be described in equation (7.1).

$$M\ddot{y} + B(p)\dot{y} + K(p)y = F_c(p) - M_g \quad (7.1)$$

where,

$$F_c = 2.43 * p - 1.29 * 10^{-3} * p^2 \quad (7.2)$$

$$K = 5.71 + 0.0307p \quad (7.3)$$

$$B = \begin{cases} 1.01 + 0.00691p\,; Inflation \\ 0.6 - 0.000803p\,; Deflation \end{cases} \quad (7.4)$$

Where y is the contraction length of the PAM, p is the fluid pressure inside the muscle [10].

Simultaneously EMG signals from the upper arm muscles will be acquired using a precision amplifier circuit and sent to a 10-bit ADC. These signals will then be passed through a low pass filter in MATLAB and will be used to derive another state space model based on Hill's muscle model. This would provide us with a clear indication of the required motion the PAM must make to move in sync with the human body.

VIII. CONCLUSION

The main objective for this paper has been the enhancement and the assist of natural upper body motion of the human skeleton. Here, calculations of the required forces and estimation of the dimensions required were finalized. After the prototype test and the fabrication of a mechanical harness for the actuators, the potential applications would be in diverse fields such as defense, physiotherapy and manufacturing. The proposed method to use EMG electrodes will enable people with muscular defects to still be able to perform daily tasks like a fully functional human.

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