Abstract

The goal of this 4th year project is to build a fully functioning assistive armband that takes input from electromyography sensors, processed by an Android device application, and driven by electroactive polymer (EAP) actuators. This paper outlines the process of testing the feasibility of different EAP actuators, the developed sensor input, and signal processing systems that have been accomplished at this early point in the design timeline.

The Team:

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Actuator Technology for the External Assistance of Muscles

1.0 Introduction

Assistive technology aims to give the beneficiary movement and strength in the most convenient and organic way by replicating smooth muscle-like motion. These devices, not to further burden the beneficiary, are intended to be silent, compact, and lightweight; this design requirement precludes using conventional combustion or electric motor systems. The emergent field of electroactive polymer (EAP) actuators, has demonstrated their capability to withstand the stress and strain needed, while being lightweight and energy efficient. [1][2]. Despite the promising research into the use of EAPs for assistive technologies, their adoption has yet to be seen.

Muscle movement and activity can be collected as analog data through Electromyography (EMG) sensors [3]. Using the muscle itself as an input provides one of the most intuitive control mechanisms possible for assistive technology. This analog data is then analyzed and processed by an Android application, and then decides if the muscle is doing work. If the application detects muscle activity, it sends a signal out on one of the device's speaker ports, which will control the actuation of the EAP actuators. The sensors and actuators then need to be integrated onto an exoskeleton structure that the user can comfortably operate. The Actuator Technology for the External Assistance of Muscles (ATEAM) project will aim to realize this novel assistive system.

2.0 Motivation

The concept for the ATEAM project originated around the growing amount of researching into: soft robotics [4]; dielectric actuators (DEAs) [1][5]; conductive polymer actuators (CPAs) [1][6]; and shape memory alloys (SMAs) [7]. All of these technologies are being investigated as novel actuators with distinct stress, strain, and deformation properties which designates EAPs as candidates for external artificial muscle tissue.

The literature review lead the project to not further investigate SMAs feasibility at this time. SMA's slow repetition of motion, and high energy loss due to thermal dissipation [7][8] contrasted from the practicality requirement for the ATEAM's design aspiration. SMA's were chosen not be further investigated at this point in time, thus the project team has proceeded with EAP actuator technology.

This project aims to create the integrating system to implement EMG technology with novel EAP actuators connected with a wearable exoskeleton structure, with the end goal of producing a working prototype of an armband by the end of the design timeline: March 2015. The armband will aid the user with the contraction of the bicep muscle for a lifting motion. The system will demonstrate the efficient, silent and lightweight advantages of EAP actuators over conventional systems.

The armband will have extensive computational and data acquisition abilities, which will help precisely and quickly detect contraction of the bicep muscle, and activate the actuators in real-time.

3.0 Background

3.1 Electromyography sensing

In the initial stages of development, the group tested different potential EMG electrodes including:

- textile electrodes;
- Silver/ silver chloride disposable electrodes;
- Silver chloride electrodes (after removing the adhesive, electrode paste).

As predicted by published characterizations [9], the signals produced by the textile electrodes were very dependent both on contact pressure and the individual wearing the electrodes. Similarly, the paste-stripped electrodes were subject to external factors, making them unreliable candidates for sensors. As a result, disposable silver/ silver chloride EMG electrodes were used for sensing muscle contraction.

Earlier research into conductive polymer actuators (CPAs) revealed several examples of conducting polymer (CP) materials used as highly sensitive EMG electrodes. Examples of investigations into these materials can be found in [10] and [11]. Currently, the group is not investigating the CP option since very accurate signals were produced from the disposable electrodes as previously noted. However, CPs may be investigated in the future as a reusable alternative to the disposable electrodes.
Figure 1 below shows the current experimental setup of disposable adhesive silver/silver chloride EMG electrodes attached to the subject's bicep.

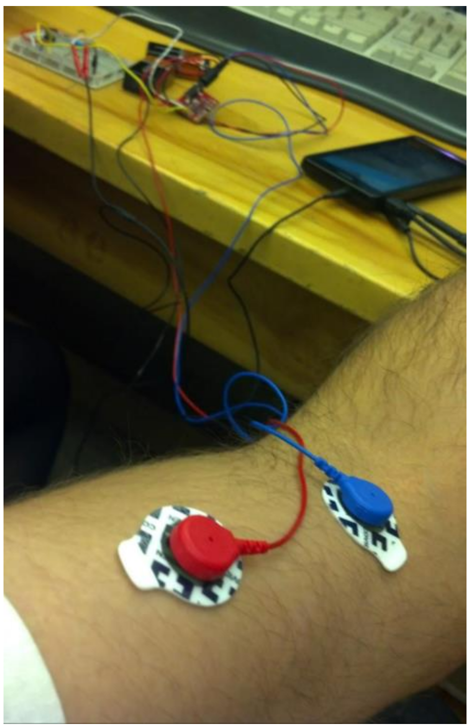


Figure 1: Disposable adhesive silver/silver chloride EMG electrodes attached to the subject's bicep. The red and blue electrodes form the differential pair, as (to be discussed in Section 3.2). The third reference electrode is hidden. The image also shows connections between the electrodes and Muscle Sensor, Muscle Sensor and input attenuation circuit, and input and phone (also discussed in Section 3.2).

3.2 Signal processing

This section will give an overview of the processing that was used in order to analyse EMG signals and create the power necessary for actuation. To convert the EMG signals obtained from the electrodes into a manipulatable analog signal, an external device called the Muscle Sensor v3 was purchased from Advancer Technologies. The Muscle Sensor takes input from three electrodes (a differential pair across the bicep and a third on bone as a reference) and converts the 0-5mV input EMG signal into a 0-9V smoothed and rectified analog signal [12].

In order to properly trigger motion in the actuators upon a muscle contraction, some basic signal processing needed to be performed on the output from the Muscle Sensor, which would require a processing unit. A smartphone device was chosen to be the processing unit for several reasons: smartphones are portable, easily available, have high enough computing power to handle real-time processing of the data, and contain an built-in analog-to-digital converter at the headphone jack. An Android device was chosen specifically over other platforms due to group members' comfort with them, and because there is a large amount of open-source code and community support for Android programming. For the initial testing, a Sony Xperia SP was used to characterize typical input and output power and voltage limits.

The microphone port for an Android device was utilized to receive input data from the Muscle Sensor and convert the data to a digital format for manipulation [13]. According to readings of the microphone input, it was determined the microphone port on the phone expects an input in the range of ±0.1V; outside this range, the signal is clipped to the maximum digital value. This was confirmed after observing the signal using the Android application OsciPrime Oscilloscope Legacy, seen in Figure 2. However, the Muscle Sensor outputs in the range of 0-9V, so the signal from the Muscle Sensor is therefore attenuated using a combination of a voltage divider, op-amps, and diodes to match the expected input on the phone's microphone port.

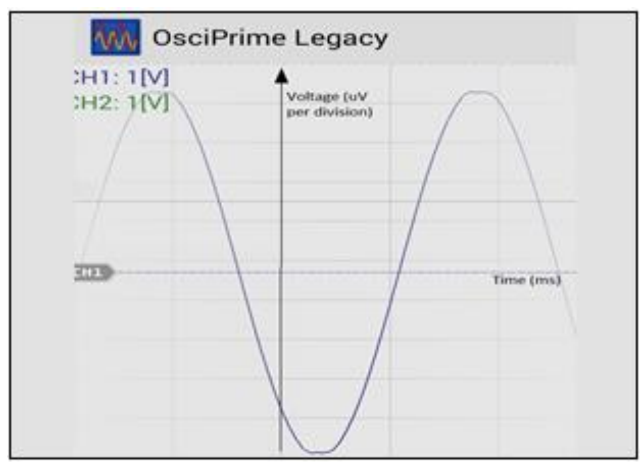


Figure 2: The OsciPrime visualization of slight voltage clipping of microphone input at 100mV.

The digitized values were in the range 0 to 1024, (or 10 bits), which was used to detect the state (contracting or relaxing) of the bicep muscle. After some collection and examination of data (Figure 3), a threshold to distinguish between the "on" and "off" states of the muscle was determined, and based on the state of the muscle a signal was outputted on one of the smartphone's headphone speakers.

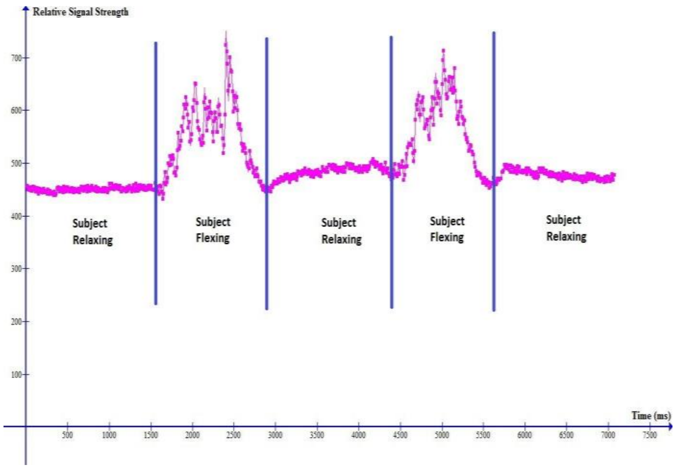


Figure 3: The x-axis of the graph represents time [mS], and on the y-axis is recorded the digital value received from the Muscle Sensor. It can be seen that the person to whom the sensor is attached is relaxing his/her arm in the first, third and fifth segment, and flexing in the second and fourth segment of the graph.

The output signal from the audio jack needs to go through several stages before it can be used to control actuation. The Android device can output 0.7V (peak to peak) from a single headphone speaker at about 50 uW (measured using a 1MOhm oscilloscope), but this output is not sufficient for significant actuation. In order to create actuation from the dielectric polymer actuators, the output must be converted into a minimum of 500 V to 1000 V DC signal (see Section 3.3). Additionally, most portable high voltage transformers operate at 40 kHz to 80 kHz (for example CCFL transformers) and the Android device headphone output is limited by frequency to the audible range (20 Hz to 20 kHz). In order to create useful output, a circuit that amplifies the initial signal and uses it to initialize a higher frequency signal which is then amplified and rectified is utilized; this process is shown in the block diagram in Figure 4.

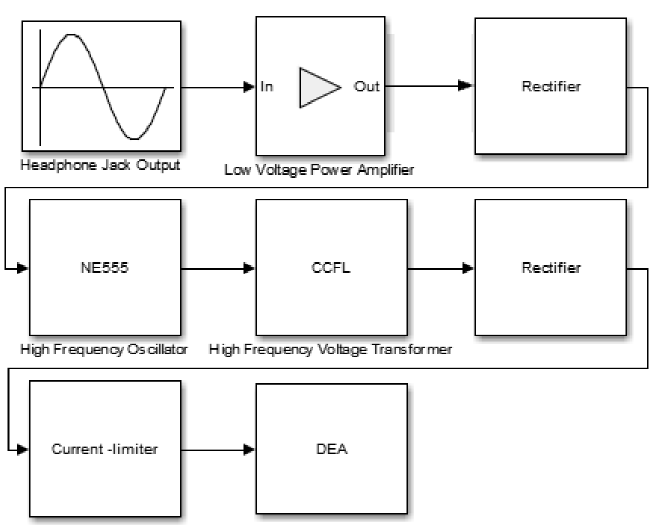


Figure 4: Block diagram of stages between smartphone headphone jack output and input to dielectric actuators (DEAs). The headphone jack output is amplified and rectified to be used as an enable (activation) signal for a high frequency oscillator created using an NE555 chip. This high frequency signal is then amplified using a CCFL voltage transformer, and then rectified to be used as the DC input to the DEAs.

Currently, the EMG input has been processed in the Android device and used to create output through the headphone jack at frequencies between 20 Hz and 20 kHz. Due to the potential damage to the phone and high voltage nature of the output for the DEAs, this portion has been tested in parts. Each block, with the exception of the current limiter, is functioning separately and simulations of the output circuit in LTSpice are showing positive results. The next steps are determining the amount of current limiting required, and creating a current limiting circuit. Once all simulations show expected, reproducible and satisfactory outputs, the final step will be implementing the stages of the output circuit.

3.3 Electroactive Polymers

The 2 major classes of EAPs explored in this project are: the dielectric actuator; and conductive polymer actuator.

The conductive polymer actuators function by having a conductive polymer submerged in an electrolyte. By applying a charge to the conductive polymer, (or to the electrolyte) the ions will intercalate into until a charge equilibrium can be achieved. The physical presences of the ions cause conductive polymers such as to swell [1]. Poly(3,4-ethylenedioxythiophene) (PEDOT) was the primary conductive polymer of choice for initial CPA feasibility testing. The initial testing showed electroactive chromatic changes as shown in Figure 5 below. These results were consistent but actuation results created reliability concerns. The project team continued forward with DEA development for the armband, this is further addressed in Section 4.0.

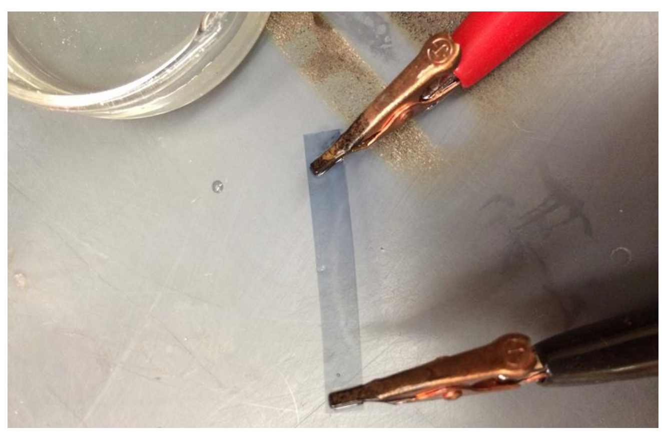


Figure 5: Conductive PEDOT strip dipped in a Lithium perchlorate solution with an applied +5V on the upper clip. A reversible visible change in color can be seen.

The dielectric actuator is effectively described as a parallel-plate capacitor with a compliant elastomer instead of a rigid dielectric between the parallel-plates. The dielectric elastomer allows the Coulomb force between the anode and cathode to reduce the height of the dielectric; thus creating a strong contracting actuation proportional to the coulomb potential:

$$F = \frac{(Area)\epsilon_0 V^2}{2r^2} \qquad (1)$$

DEAs can be modeled off this simple principle but as seen in the above equation the force is exponentially dependent on voltage (V^2), inversely exponentially proportional to the distance between the plates (r^2) and to the relative permittivity of the material. The Coulomb force has to be able to not only pull a load but to also deform the elastomeric material. The elastomer's deformation, stress, displacement and stretch are all vital material properties for the actuators ability to act as artificial muscle tissue [8][9]. Actuator design one may want to maximize the force balanced with the limitation of the material's ability to withstand extremely high voltages without breaking down. The effect of dielectric breakdown is destructively seen in of Figure 6.

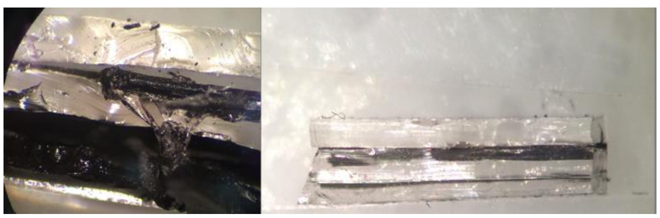
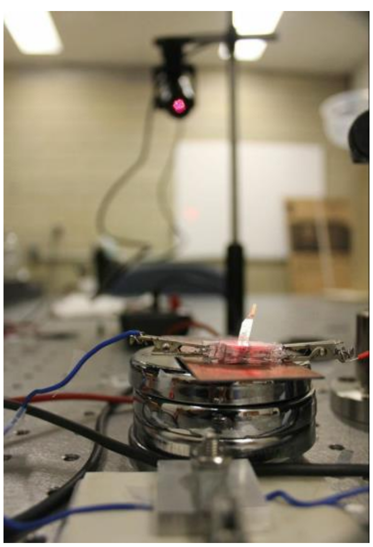


Figure 6: Dielectric breakdown of ~0.500mm Sylgard (on the left) and the DEA before testing.

The breakdown of the material is one of the most significant limiting design factors with DEAs due to the high operational voltages [14][15]. The actuators in this project are designed to use a lower voltage (~0.5kV) and to keep the system heavily current limited for safety reasons during user operation.

4.0 Actuator Lab results



Due to an aggressive project timeline, the exploration into the CPA's feasibility as an actuator for the ATEAM armband was deemed impractical at this time. Given that the timeline permits, the CPA will be explored in further detail.

4.1 Test Apparatus

The DEA's showed immediate actuation with a simple height change test apparatus as shown in the figure 7 below. The apparatus uses a Uniphase 1103p laser to reflect off of a reflective sheet on to a measurement screen. The reflective sheet is positioned such that it uses the thickness of the DEA to control the angle of reflection. This creates the dependence on the position of the reflected laser dot on the thickness of the DEA itself. As the thickness of the DEA actuates the reflected laser's movement on the measurement screen can be tracked.

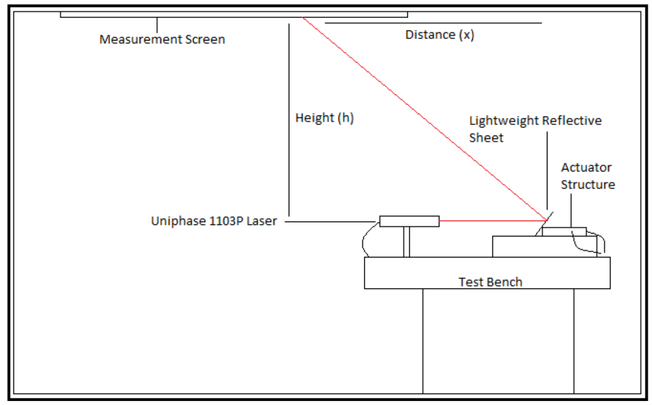


Figure 7: Diagram of the actuator test apparatus.

The configurations of different actuators with varying fabrication techniques were tested only in binary for actuator and not yet for percent deformation.

Where the percent deformation for characterizing different type of DEA can be found with geometry to be:

$$\Delta\theta = \frac{-arctan\left(\frac{h\Delta x}{x(x-\Delta x)+h^2}\right)}{2} \qquad (2)$$

$$\frac{\Delta z}{z_i} = \left(tan(\theta_i + \Delta\theta) - tan(\theta_i)\right)tan(\theta_i) \qquad (3)$$

Where θ_i is the initial angle of the actuator, $\Delta \boldsymbol{x}$ is the displacement of the laser dot observed, and Δz is the change in height of the actuator.

4.2 DEA Structures

The materials chosen to be the dielectric elastomer were Sylgard-184, and Sylgard 186. After a literature review of previously used materials [1][5][8][14]-[16]. These two brands of Sylgard possess ideal material properties while also being inexpensive, readily available, easily fabricated [16], and while avoiding urethane based materials due to high current dangers for armband application [17]. The strong dielectric and material properties of these elastomers are found in Table 1.

Table 1: Sylgard important dielectric and material properties [17][18].

Material	Dielectric strength (kV/mm)	Durometer Hardness, Shore A	Heat Cure Time (100^0 celcius)
Sylgard 184	19	43	35 minutes
Sylgard 186	13	25	1 hour

The structures tested all showed consistent actuation through a minimum of four independent trials. Shown below in Figures 8 through 10 are three different configurations of actuators investigated on the test bench being cycled through 0 to 500V 3-5 times cycles per trial.

All fabrication of Sylgard 184 sheets was done by drop casting into a petri dish in different increments by mass. Small sections of layers free from visible impurity were cut out in the varying sizes needed for single stack DEA tests. Figure 8 shows the one of the initial actuators to confirm actuation on the test apparatus.

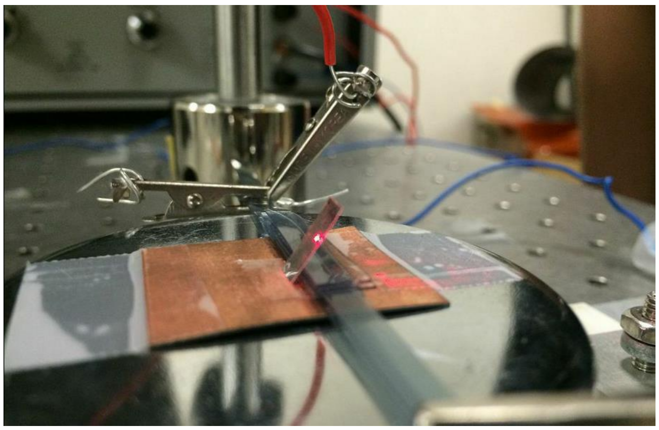


Figure 8: Sylgard 184 and Conductive foil bi-layered stack.

These sandwich style actuators were used only to demonstrate if actuation was achievable at our upper voltage limit (0.5kV) since they did not maintain structure when transferred.

Figures 9 and 10 show DEA where a solid state fabrication process was used involving a Sylgard 184 strip cured with the conductive layers in a method developed for this project.

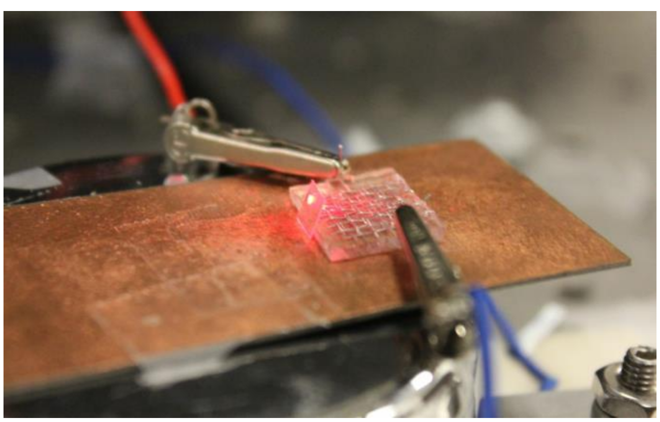


Figure 9: 8mm wide Conductive foil with ~0.5mm Sylgard 184.

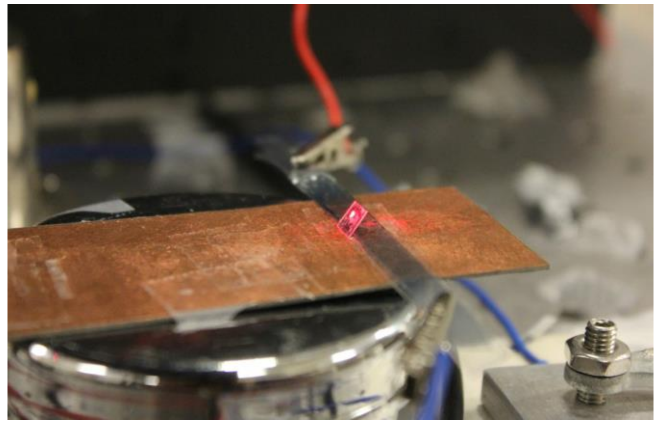


Figure 10: Steel Mesh bi-layer fabricated in 15g Sylgard 184 batch.

At the time of the writing of this article the Sylgard 186 testing had just begun as a look into using materials with lower shore hardness. 3D-printed molds for casting the high viscosity materials in more complex geometries are currently being designed. This geometries' will explore bending DEAs and linear actuators. The final step before finalizing actuator cell designs to be fixed into the exoskeleton structure is to quantitatively characterize Sylgard 184, Sylgard 186 and dielectric gel's for percent deformation, life span, stress, applied strain capabilities, and practicality of different geometries' implementation in the assistive armband.

5.0 Conclusion and Future Goals

The short term goals are creating and implementing dielectric actuators in series and creating a system from the EMG signals to the dielectric actuators such that the user flexing creates actuation. In proceeding steps, the group will construct the actual armband exoskeleton and attach the actuators along this structure. In order to build this exoskeleton, the actuator structures need to be built. In parallel with the exoskeleton construction, a feedback loop will be designed to inform the Android device of armband position and implemented. The final structure should be a full system which encompasses muscle contraction to actuation with feedback. We hope that this project's innovative look at assistive technology can help in ways conventional technology maybe be limited in.

In the long term, the armband's technology can be expanded to any part of the human body with more precise applications such as recording muscle movement patterns and playing back the movements for the user with valuable applications in teaching and programmable rehabilitation.

References

[1] T. Mirfakhrai, J. Madden and R. Baughman, 'Polymer artificial muscles', *Materials Today*, vol. 10, no. 4, pp. 30-38, 2007.

[2] P. Madden, J. Madden, P. Anquetil, N. Vandesteeg and I. Hunter, 'The Relation of Conducting Polymer Actuator Material Properties to Performance', *IEEE Journal of Oceanic Engineering*, vol. 29, no. 3, pp. 696-705, 2004.

[3] C. Drewes, 'Electromyography: Recording electrical signals from human muscle.' C. Drewes, Ed. 2000.

[4] S.Morin, Y.Shevchenko, J. Lessing, and et al. *Using "Click-e-Bricks" to Make 3D elastomeric Structures*, Advanced Materials, 2015 pp.5991.5999.

[5] K. Kim and S.Tadokoro*, Electroactive polymers for robotic applications.* London: Springer, 2007. Section 2.

[6] R.Baughman, 'Conducting polymer artificial muscles', *Synthetic Metals*, vol.78. no.3, pp.339-353, 1996.

[7]J. Madden, N. Vandesteeg, P. Anquetil, P. Madden, A. Takshi, R. Pytel, S. Lafontaine, P. Wieringa and I. Hunter, 'Artificial Muscle Technology: Physical Principles and Naval Prospects', *IEEE Journal of Oceanic Engineering*, vol. 29, no. 3, pp. 706-728, 2004.

[8] J. Shintake, S. Rosset, B. Schubert, D. Floreano and H. Shea, 'A Foldable Antagonistic Actuator', *IEEE/ASME Trans. Mechatron.*, pp. 1-12, 2014.

[9] I. Beckmann, C. Neuhaus, G. Medrano, N. Jungbecker, M. Walter, T. Gries and S. Leonhardt. "Characterization of textile electrodes and conductors using standardized measurement setups." Physiol. Meas. 31 (2010). p. 233-247.

[10] R. A. Green, S. Baek, L. A. Poole-Warren, and P. J. Martens., "Conducting polymer-hydrogels for medical electrode applications.", Sci. Technol. Adv. Mater. 11(2010). 014107.

[11] C. Pylatiuk, M. Muller-Riedere, A. Kargov, S. Schulz, O. Schill, M. Reischl and G. Bretthauer. "Comparison of Surface EMG Monitoring Electrodes for Long-term Use in Rehabilitation Device Control." IEEE 11th International Conference on Rehabilitation Robotics. Kyoto Internation Conference Center, Japan, June 23-26, 2009.

[12] Three-lead Differential Muscle/Electromyography Sensor for Microcontroller Applications, v3. Advancer Technologies, 2013.

[13] Y.Kou, S. Verma, T.Schmid, and P.Dutta, 'Hijacking power and bandwidth from the mobile phone's audio interface', ACM 978-1-4503-0473-3-10/12 , University of Muchigan, 2010.

[14] Y. Fujihara, T. Hanamoto and F. Dai, 'Fundamental research on polymer material as artificial muscle', *Artif Life Robotics*, vol. 12, no. 1-2, pp. 232-235, 2008.

[15] G. Kofod, 'Dielectric elastomer actuators', Ph.D thesis, The technical University of Denmark, September 2001.

[16]W. Lai, 'Characteristics of dielectric elastomers and fabrication of dielectric elastomer actuators for artificial muscle application', Graduate theses and Dissertations, Iowa State University, 2014.

[17] Dow Corning, 'Sylgard® 184 Silicone Elastomer', 2014. [Online]. Available: http://www.dowcorning.com/DataFiles/090276fe80190b08.pdf [Accessed: 10- Sep- 2014].

[18] Dow Corning, 'Sylgard® 186 Silicone Elastomer', 2014. [Online]. Available: http://www.dowcorning.com/DataFiles/0902770180dd95fc.pdf. [Accessed: 10- Sep- 2014].