DIELECTRIC ELECTROACTIVE POLYMER BASED BIOMEDICAL DEVICES: CONTROL, SENSING AND INTERFACING

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ABSTRACT

Electroactive Polymers (EAPs) have great potential to provide smart solutions to engineering problems in the fields such as robotics, medical devices, power generation, actuators and sensors. This is due to the fact that EAPs yield several important characteristics that are advantageous over conventional types of materials, like: lower weight, faster response, higher power density and quieter operation. In this thesis, the use of Dielectric Electroactive Polymer (DEAP) based actuators and sensors were studied in the following order: 1) Development of a nonlinear force control for PolyPower InLastor PUSH actuator, 2) Evaluation of the PUSH actuator for haptic use with respect to a Linear Voice Coil actuator, 3) Prototyping a novel pressure sensor based on DEAP to be used in a Blood Pressure Measurement Device (SyBeD).

Topic 1: Controlling the amount of force exerted during an interaction between an actuator and an object is crucial for certain applications, such as those involving a human and an actuator. To date there is little research into the force control of EAPs or their possible applications that utilize force control. This study presents a realtime nonlinear force controller for a Rolled type Dielectric Electroactive Polymer Actuator (DEA). To increase the response of the actuator, a control algorithm and an inverse model were derived using the actuator’s Nonlinear behavior. The force controller presented can enhance the safety and performance of this unique family of actuators, allowing for more advanced and efficient applications.

Topic 2: Quality, amplitude and frequency of the interaction forces between a human and an actuator are essential traits for haptic applications. This study demonstrates a rolled Dielectric Elastomer Actuator (DEA) used as a telepresence device in a heartbeat
measurement application. In this testing, the heart signals were acquired from a remote location using a wireless heart rate sensor, sent through a network and the DEA was used to haptically reproduce the heartbeats at the medical expert's location. A series of preliminary human subject tests were conducted to demonstrate that a) DEAP based haptic feeling can be used in heartbeat measurement tests and b) through subjective testing the stiffness and actuator properties of the EAP can be tuned for a variety of applications.

Topic 3: Providing cheap and accurate physiological measurement devices for primary healthcare increases the chances of early detection of diseases. This study focuses on the problems with the available blood pressure measurement devices and proposes how to improve current practices by employing an embedded DEAP sensor based pressure cuff and using wireless communication to form next generation diagnosis tools.
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1 INTRODUCTION

1.1 MOTIVATION

Biomedical devices help post stroke patients to re-gain their motor function abilities, decrease the time required for the healing of a scar, provide inexpensive and accurate physiological measurement tools for practitioners. There are many biomedical devices that improve the healthcare needs of both patients and healthcare professionals using conventional materials. As the technology continues to evolve, new materials and devices enable us to solve some of the problems that occur in health sciences, it is my responsibility to understand the developing technologies and to elaborate on how to use these for a higher purpose of helping people through engineering and medicine interaction.

Throughout my research in Biomedical Mechatronics Laboratory at Northeastern University, I worked to develop the know-how to understand more about the capabilities of Dielectric Electroactive Polymers (DEAP). This understanding expanded my knowledge regarding the areas of implementation of DEAPs to improve services for both the healthcare professionals and to patients as biomedical devices practices advance. In the light of these reasons, DEAP based actuators and sensors were studied to develop control algorithms for haptic interactions with people and cost-efficient sensory equipment for physiological measurements. The persistent research issue is “how we can build better and more efficient devices to help patients with new technologies, such as DEAPs?”
1.2 CONTRIBUTIONS

Throughout the research process, our efforts on establishing the know-how and experience on the use of Dielectric Electroactive Polymer based actuators and sensors resulted in two conference papers, a provisional patent application, an innovation prize in primary healthcare and new industry collaborations to further improve applied DEAP products.

*Nonlinear Force Control of Dielectric Electroactive Polymer Actuators* [1], is the first paper to present an experimentally established nonlinear force controller to control the interactions with a static environment using DEAP actuators. On the other hand, *Haptic Interfaces Using Dielectric Electroactive Polymers* [2] demonstrated the first application of DEAP actuator as a haptic heart rate emulator. Both papers were published by the International Society of Optics and Photonics (SPIE) and the novelties described were presented in Smart Structures / NDE conference held in San Diego, CA on March 2010. Both papers received well feedback from the Electroactive Polymer community.

The proposal: *Synchronized Blood Pressure Measurement Device* [3]: SyBeD, was selected as a top-ten finalist in a nationwide competition held by the Center for Integration of Medicine and Innovative Technology (CIMIT). The proposal received research funding to develop a prototype for showing proof of concept. The pressure sensor is the first example of a DEAP sensor to be used in a blood pressure measurement device. Following the initial tests, Northeastern University has filed a provisional patent application for the DEAP based pressure sensor prototype.
Our ongoing collaboration with Danfoss, PolyPower A/S has resulted to be very fruitful for both parties to understand the capabilities of DEAP powered products and applications with exchange of ideas for future developments to be followed on the research and development area of applications of Electroactive Polymer based devices. It also initiated the collaborations with Care2Wear AS on the wireless communication and DEAP based blood pressure monitor systems.
2 BACKGROUND

2.1 ELECTROACTIVE POLYMERS

Study of Electroactive Polymer (EAP) materials has emerged extensively since 1990s due to the efforts on polymer development and advancement in new material properties. Emergence of EAP materials with induced strains over 100% marks the substantial evolution followed after. However, the beginning of EAP studies dates back to 1880s when Roentgen conducted a study to charge and discharge a polymer fixed at one end and attached to a mass at the free end related to the observed volume change to thermal effects due to applied electric field. In 1899, Sacerdote related strain response to electrical stimulation with a mathematical model. The achievements were then succeeded with the discovery of piezoelectric polymer by Eguchi in 1925 [4]. EAP technology first observed over hundred years ago is now emerging and gaining both scientific and industrial interest from many fields to enhance current capabilities.

Materials and structures that respond to external stimuli by means of visible and measureable energy in a controlled way in real-time or near real-time are referred as smart materials. Piezoelectric materials and Shape Memory Alloys (SMAs) are commonly known examples of smart materials, and the research practices over these materials led to commercial use of these new technologies. EAPs are plastic materials that exhibit shape and size change due to applied voltage or current. These EAPs are also referred to as a part of smart materials. There are two main classes of EAPs, as classified in Figure 1 as: Ionic EAPs and Electronic EAPs. Ionic EAPs are Polyelectrode gels, Ionomeric Polymer-Metal Composites (IPMCs), Conductive Polymers, Carbon Nanotubes and Electrorheological Fluids and they work through mobility or diffusion of
ions. Electronic EAPs are Piezoelectric Polymers, Dielectric EAPs (DEAPs), Electrostrictive Polymers and Liquid Crystal Elastomer materials that work by applied electric fields or Coulomb forces.

![EAP Materials Diagram]

**Figure 1: Classification of EAP Materials**

As indicated above, the EAP family consists of a broad polymer network, coupled with either ionic or electrically driven means. Some of the classes have been thoroughly studied and everyday applications are available, on the other hand many of these topics remain as research areas that draw attention from research companies to develop future transducers. The EAP technology is advantageous to develop novel products with low energy requirement and manufacturing costs. Liquid Crystal Displays (LCDs) are a good example of such emergence. These displays rely on the moving and aligning liquid crystal (LC) elastomer materials, to display shapes and color. Essentially LCD displays use polarization filters, a reflector and electrodes to generate the shapes and colors we see in our laptop screen, TV sets, microwave programmers, etc. The advantages of using LCDs were greater compared to the previous technology of CRTs, therefore almost all monitors have been replaced by LCDs. This shows a very current example of integrating EAP technology to everyday life.
Another common application is the use of piezoelectric elements as transducers. Transducers are used as precision actuators in MEMS and nanotechnology devices, lens focusing in cameras, buzzers at homes, etc. A piezoelectric material is able to generate electricity when strained by mechanical means, the actuation principle relies on using electricity as an input to output strain, hence force. The strain levels for piezo elements are smaller compared to other EAPs, for example lead zirconate titanate crystal has a maximum strain rate of 0.1%. Although the strain level is quite low; these materials are very suitable for sensory applications where there is precision needed. Physik Instrumente (PI) is one of the companies that develop many of the piezoelectric transducers currently in use today.

Figure 2: Smart Materials Today: Liquid Crystal Display, Nano-precision Positioned, Carbon Nanotube Fibers [5-6]

One of the future technologies is projected to be nanotechnology based devices that is using nano-sized particles, assembling them to miniaturize the current products. The advantages of using nanotechnology are numerous; one of the building blocks of this technology is carbon nanotubes (CNTs). They exhibit extraordinary strength and very unique electrical properties that enable their use in electronics, optics, fabrics and
architecture. Some single walled CNTs have up to 5TPa of Young’s modulus, giving them the characteristic of being the strongest and stiffest materials that has been discovered. Production and characterization of CNTs have been studied for the past decade, and future obstacle that needs to be tackled is the assembly of CNTs for usable applications [7]. Many devices that are currently in use would be affected deeply by the availability of CNT replacements. Future applications with CNTs include: electronic parts, energy storage systems, development of new materials with CNT enhanced structures, gas, electrochemical, optical sensors, etc [8]. Liquid crystals, piezoelectric materials and carbon nanotubes have been introduced to stress the uniqueness and potential of EAPs on how to improve our daily life. Many EAP classes exhibit unique characteristics and enable new design opportunities, mostly due to their natural resemblance to human tissue and muscles.

Artificial Muscle is another synonym commonly used in the EAP community to exhibit the resemblance of EAP actuation mechanisms to human muscles. The mechanical energy generation from the electrochemical or electrical stimuli is similar for mammalian skeletal muscle and EAP actuators. Pei, Bar-Cohen and Madden reviewed a comparison for the properties of EAP actuators with mammalian skeletal muscle and the comparison is a useful benchmark to select actuators for specific applications. When compared to Ferroelectric Polymers, Conducting Polymers and NiTi Shape Memory Alloys, Dielectric Electroactive Polymers exhibit characteristics that are closer to the human skeletal muscle. DEAPs cover the limits of mammalian muscle strain (%), have greater work density (kJ/m³) than mammalian muscle. However one of the biggest drawbacks still is the cycle life, the lifetime of DEAP (10⁵) is still considerable shorter than a mammalian
One of the future goals of the EAP community is to generate EAP actuators that can replace injured human muscle tissue and act as active prosthetics for people with missing limbs.

### 2.2 DIELECTRIC ELECTROACTIVE POLYMERS

Dielectric Electroactive Polymers (DEAPs) are electromechanical transducers that convert one form of energy to another. They can convert electrical energy to mechanical energy, which makes them actuators. On the other hand this process is reversible; they can adapt mechanical energy to electrical energy as well and could be used as sensors and generators. Many different types of actuators and sensors are currently in use in industry applications, a broad range of fields that we sought for research and development was: consumer electronics, haptic devices and maybe most importantly, biomedical devices. DEAPs could at least be employed in these fields as: new generation input devices and rehabilitation machinery and physiological sensors.

Dielectric EAP falls in the category of Electric EAP’s and also referred as Dielectric Elastomers. Response of the electroactive film is the result of electrostatic charge on the electrodes. When charged with opposite polarity, electrodes attract each other creating pressure ($\rho$) on the film and causing to compress the polymer film. Hence the film thickness decreases and area increases to keep the volume constant. Since a rolled type DEAP actuator constraints the film on two sides, the pressure on the film results in linear motion. When the voltage is switched off, the DEAP actuator contracts back to its
original shape. Pelrine’s electromechanical model [10] describes the pressure exerted on
the silicon layer $\rho$ as:

$$\rho = \varepsilon_r\varepsilon_o E^2 = \varepsilon_r\varepsilon_o \left(\frac{V}{t}\right)^2$$  \hspace{1cm} (2.1)

In addition to pressure generation on the film, DEAP film changes capacitance with
strain. Since the electrodes are separated from each other with non-conducting polymer,
DEAP is also considered as a variable capacitance device. Capacitance, $C$, is depicted
below with the electrical model as [4, 11]:

$$C = \varepsilon_r\varepsilon_o \frac{A}{t}$$  \hspace{1cm} (2.2)

Where $\varepsilon_r$ and $\varepsilon_o$ are the permittivity of free space and the relative permittivity of the
polymer; $E$ is the applied electric field; $V$ is the voltage, $A$ is the area (length times width)
of the film and $t$ is the film thickness.

Figure 3: Stretched Electroactive Polymer Film
DEAP film thickness is on the order of micrometers and the area depends on the application and it is on the order of meters. Pressure on the film creates the force along the direction of elongation and it is directly proportional to the square of voltage. As can be seen from Pelrine’s model, a nonlinear response of force is expected with varying voltage values. On the other hand, the change in the capacitance is directly related to the change in the film thickness and the area since dielectric constants are fixed.

Once the physics was understood, the researchers are interested in how to make better polymers that can take advantage of this phenomenon and extend the limits of DEAPs. The improvement involves testing of new polymer material and compliant electrodes. Skov and Sommer-Larsen described physical and chemical properties of dielectric elastomers and building processes; whereas Kornbluh and Pelrine tested acrylic and silicone elastomer performances to find the advantages of silicone based DEAPs over acrylic ones and Kofod and Sommer-Larsen discussed the compliant electrode structure for higher performance [9]. The DEAP material used in the studies conducted and presented in this thesis is based on silicone and currently being manufactured by Danfoss, PolyPower A/S in Denmark. The main goal of DEAP community is to enhance the current practices of manufacturing and establish a broader description of material properties to build DEAP structures. In addition to the ongoing modeling and material characterization studies, novel applications to steer the DEAP manufacturing and design is necessary. The applications could mainly be grouped as DEAP based actuators, generators and sensors. All of these fields are highly promising and gaining more scientific interest every year. In the following sections, DEAP based actuators and
sensors are described to serve as a background to the studies conducted in the following sections.

2.2.1 DIELECTRIC ELECTROACTIVE POLYMER ACTUATORS

Dielectric Electroactive Polymers (DEAPs) combine actuation and sensing into a single material. When DEAPs are used as an actuator, they operate with high power density and operations at low speed do not introduce problems like friction; unlike electric motors, pneumatic and hydraulic systems that are frequently in use for industrial and robotic applications. There is an increasing scientific interest in the EAP actuator field that is resulting in development of new types of actuators and sensors. DEAP actuators have been introduced as rolls, tubes, stacks, diaphragms, extenders, folded, bimorphs and unimorphs [11]. DEAPs are very unique and functional since they combine large active deformations, high energy densities and fast response compared to other Electric EAPs.

The basic configurations of DEAP based actuators do not necessarily have to have external support structures to guide the polymer actuators, instead their design enables the motion itself. The DEAP thin film can be used as stacked, extender, bimorph/unimorph bending beam, diaphragm and tube configurations. The DEAP enables design flexibility to researchers and engineers; however there are practical challenges that need to be taken into account. The actuator design should be able to couple forces generated by electrostatics and external loading and prevent buckling. Designs can take advantage of the polymer material such as being formed into large areas as sheets, stretched for various contour shapes due to its flexibility, employed in
applications where transparency is needed and serve as structural component in addition to being the actuation mechanism of a device [9].

Figure 4: Configurations of Dielectric Electroactive Polymer Actuators[11]

Detailed configurations were suggested and presented in the DEAP literature. Most of the configurations are similar to the ones developed for piezoelectric materials, however DEAPs exhibit much higher rates of strain therefore the implementation is not the same as the piezoelectric examples. Bowtie and spider configuration enables effective coupling of the external load from the internally generated electrostatic force and better for large stroke applications. Roll and framed structures can exploit flexibility while being able to exert unidirectional forces and these configurations are mostly favored for high force output. Stacked configuration demonstrated by Carpi uses the actuator to pull objects on
the contrary to most of the configurations presented so far and it is also favored for high force applications. In fact, a finger rehabilitation device based on this configuration was proposed by Carpi and his group in University of Pisa, it is will later be discussed in the section 2.2.3. Finally diaphragm type configuration is favored when distributed loading is required to employ actuators on 2D surfaces or to use them as pumps.

Almost all configurations listed above are 1-degree-of-freedom (DOF), extension in planar or unidirectional field. However, by using a more complex configuration Rosenthal and Pei demonstrated the “spring roll” actuator that is capable of multi-degree-of-freedom actuation and self recovery through rolling four isolated DEAP film sheets around a spring [12]. 2-DOF and 3-DOF version of spring roll was demonstrated. 2-DOF roll is capable of 60-90 degrees maximum bending angle with 0.7-1.68N lateral force for its size being 6.8-9cm in length and 1.4-2.3 diameter. The 3-DOF roll with similar size has maximum bending angle of 20-35 degrees with lateral forces between 0.2-1N.
An emphasis for the explanation of the rolled DEAP is necessary for the following sections of this thesis, since the actuator that was used for force control in section 3 and haptic studies in section 4 is a rolled DEAP. Production of such an actuator consists of two phases; laminated DEAP film production is the first step and the actuator structure fabrication is the second. In the first phase the elastic material is developed with a corrugated surface so that compliant electrode structure can be established. The electrodes are deposited using physical vapor deposition (PVD), and lamination. In the second phase, two laminated films with the compliant electrode structure are placed back to back to each other and an elliptical winding tool rolls the multilayer structure. The rolling process requires precision and repeatability for better quality and homogenous actuator production. Figure 6 represents the laminated film before the rolling process.

![Rolled DEAP Film in Production Line](image)

Figure 6: Rolled DEAP Film in Production Line [13]

The unique characteristics of DEAPs enable designers and engineers to form new principles of actuation types and processes according to society and industry needs. In order to design and provide solutions to society’s needs, the advantages of using this
unique family of polymers should be well understood. Improving material properties and learning how to use these smart materials could be considered as parallel paths to be proceeded to utilize DEAP systems in the future. A better understanding of how DEAP operates can help in forming new actuators with greater capabilities over conventional actuators. Compared to electromagnetic motors and linear actuators, some DEAP actuators have greater power density and efficiency for operation at low speed because gearing is eliminated. The material and production costs are estimated to be lower than electromagnetic actuators and they come in greater size and shape variations [4]. DEAP actuators have promising opportunity for many applications in the fields of: consumer electronics, medical instruments, MEMS, biomimetics, robotics and haptics, since they can be manufactured for custom applications.

2.2.2 DIELECTRIC ELECTROACTIVE POLYMER SENSORS
The sensory use of DEAPs is described in this section. When used as a sensor, DEAPs work as a variable capacitor where capacitance change across the electrodes is observed with mechanical strain. Strain is both related to force and displacement; therefore it could be used as a position, a pressure or a force sensor depending on the way sensor is set up. Most of the work conducted on DEAPs focused on the actuation property so far; however they are highly advantageous for sensory applications as well. By measuring the capacitance across the electrodes on the elastomer, one can convert mechanical stimuli to electrical signals. Sommer-Larsen and Benslimane relate the capacitance change to the square of the strain ratio to [9]:
\[ C = C_0 \lambda^2 \]  

(2.3)

where \( C \) is the instantaneous capacitance, \( C_0 \) is the initial capacitance and \( \lambda \) is the material strain. The sensitivity of any sensor type that takes advantage of the strain of DEAP depends on the accuracy of capacitance measurement system. A capacitance meter, a capacitance to frequency converter or an RC timer could be employed to calculate the capacitance for the sensory system. DEAP based sensors provide design flexibility since the strain rates are on the order of 20-100\%. Applications with high strain ratios can take advantage of the lightweight, plastic sensors that could be developed by DEAPs. When it comes to industrial use, bandwidth of such sensors is an important aspect. The bandwidth and response time depend on the design, the elastomer material and the electronics that is used. Due to the elastomeric behavior of DEAP materials, fastest response time is on the order of a few milliseconds. Generally, DEAPs operate around a bandwidth of 100Hz. This makes DEAP based sensors suitable for medical operations that involve physiological measurements and low frequency position, force and pressure measurement devices.

The studies on sensory applications of DEAP materials were found to be limited due to the current emphasis on the actuation phenomena. Few of the examples include Carpi’s work on integrating sensors to buckling actuators for biomedical applications. He has included a strip of DEAP film on the diaphragm type actuator assembled as stacks to measure the displacement due to buckling. In addition to the buckling actuators, De Rossi stressed the advantages and usability of DEAP materials as wearable sensors as biomonitoring devices. In this aspect, shirts, gloves, socks, shoes could be embedded with DEAP materials to continuously gather data to use for medical, interaction media or
artistic purposes [14]. Embedding DEAPs involve an interdisciplinary approach of fabric design and DEAP know-how. Smartex has been incorporating smart material technology into their textile designs. The interactive fabrics could be incorporated for health monitoring, rehabilitation and sports medicine. “WEALTHY” (see Figure 7) is an example integrated system to that is equipped with 6 fabric electrocardiogram (ECG) electrodes, 4 fabric impedance electrodes and 9 insulated fabric connectors.

Figure 7: DEAP as a Wearable Sensor: Body Movement Study (left) [14], Smartex “wealthy” (right) [15]

DEAPs have high potential to be used in medical devices particularly suitable for developing wearable, light, silent and flexible measurement systems. The joint effort between the medical professionals and engineers will define the device properties. In order to develop medical devices, novel technologies provide the design flexibility on the other hand require know-how. Smart materials have been researched for decades now and
records show smart material based systems are starting to be used for risky operations [10]. Smart materials, namely DEAPs are highly modifiable. The film thickness, elastomer material in use and electrodes of a dielectric elastomer could be altered depending on the stiffness values required for operation at hand. Being able to grasp the use of a sensing structure that has tunable properties is a very important asset to improve the compliance range of physiological measurement devices. Many of researchers point out the practicality of compliancy in medical device development and patient monitoring systems. In addition to the research community, medical practitioners are in favor of using smart materials because it provides design flexibility for the product developer and improved comfort to the patient.

2.2.3 BIOMEDICAL DEVICES WITH DIELECTRIC ELECTROACTIVE POLYMERS

Studies have shown that biomedical applications of EAPs are one of the most promising fields that take advantage of the inherent characteristics of this polymer [16]. DEAPs are made using thin films, operate silently, have a soft texture and their compliance is close to that of the human body. DEAPs are highly modifiable because the film thickness, the elastomer material and the electrodes of a dielectric elastomer could all be altered depending on the stiffness values required for a specific application. Being able to use a sensing structure that has tunable stiffness properties is an important asset to improve the compliance range of human machine interfaces.

There are several examples of DEAPs being used in biomedical applications. DEAPs have been proposed as implantable diaphragm muscles. In addition to body conformity,
their efficiency enables the use of small size batteries hence allowing their use in remote operations. Larger sizes of DEAP films and actuators could be used for large areas of the human body. The DEAP material properties could also enable their use in small patches. A haptic device powered by a dielectric elastomer was developed to convey information for people with visual impairment using the Braille system as seen in Figure 8. The grid like structure of that device also enables fingertip haptic feedback for virtual reality operations and it could potentially be used in surgical settings.

Another example of a medical application of EAP actuators is the dynamic hand orthosis for finger rehabilitation operated by dielectric elastomer contractile actuators as seen in Figure 9. This orthosis works silently and applies continuous loading on the finger for
faster recovery. In this application, stacked DEAP actuators that are able to pull are installed and connected to a pulley to elevate the finger for rehabilitation exercises.

![DEAP Powered Finger Rehabilitation Orthosis Device](image)

**Figure 9: DEAP Powered Finger Rehabilitation Orthosis Device [18]**

Furthermore, a unique advantage of polymer based actuator and sensor systems is that they are Magnetic Resonance Imaging (MRI) compatible and hence they can operate inside MRI scanners [19]. MRI compatibility of the actuator itself has already been documented and results show that the electromechanical performance of the actuator does not change in the MRI environment, and the actuator does not affect the image quality. In this way EAP based sensorized and actuated medical devices could be made for use inside MRI scanners. For example, a robotic manipulator for prostate cancer interventions in the MRI environment 5 was built using EAP as the alignment element as seen in Figure 10.
Tactile display for Braille, Finger rehabilitation device and MRI compatible assistive device for prostate cancer interventions are unique examples of DEAP actuators being used in biomedical applications. DEAPs maintain certain characteristics in the design and implementation that helps use the compliance of these actuators. As can be seen from these examples, DEAPs could be employed in operations that require: pull, push, alignment, tactile feedback and sensory input. Biomedical use of DEAPs are widely applicable for many operations, they enable design and problem solving flexibility by eliminating additional mechanical parts. DEAPs provide a potential to enhance the capabilities of today’s biomedical devices due to their nature.

2.3 SUMMARY

Background information starting from what EAPs are to DEAPs in specific is given in this chapter to clarify the research area of this thesis. EAP phenomenon is over a century
old, however with only recent advancements of the EAP technology and availability of new materials made the current state of EAPs and DEAPs in particular attractive to scientific and research community. The reasons on the advancement for DEAP based actuators and sensors were described and capabilities have been presented with examples. Most of the research on DEAP field focuses on the development of new materials, structures and modeling. Similar efforts should be followed to develop new applications using currently available materials. The unique properties of DEAP materials suggest it is advantageous to use them in haptics, medical and robotic applications. In order to improve the performance of a DEAP actuator for its use in haptics and human interfaces, force control for DEAP actuator is necessary. In order to evaluate the usability of DEAP actuators as biomedical devices in contact with humans, certain tests need to be performed. In addition to the use of DEAP as an actuator, its capabilities as a sensors need to be studied to enhance the current sensor equipments.
3 FORCE CONTROL OF DEAP ACTUATOR

3.1 INTRODUCTION

Experimental and mathematical characterization of electroactive polymers is still being developed. Fox and Goulbourne recently demonstrated electromechanical dynamic response of a DE membrane subject to pressure and reported the sensitivity of electrical loading on the membrane [21]. Huynh, Alici and Spinks worked on system identification and validation of their conductive polymer actuator model [22]. Material selection for Dielectric Elastomer Actuator changes the mechanical properties of the film; therefore the control structure has to be altered accordingly. It was observed that although different materials had been used as the building block, nonlinear behavior of DEAs of the actuators was consistent. Precise modeling and control was adapted to tackle difficulties caused by nonlinearity in many studies. In control studies conducted with Ionic Polymer Metal Composites (IPMCs) [23-25] Richardson et.al. investigated the polymer performance with impedance control. Another feedback control method by Mallavarapu, Newbury and Leo used Linear Quadratic Regulator (LQR). Bhat and Kim further looked into hybrid force and position control strategy by implementing empirically obtained plant transfer function for precision control. On the other hand, conjugated and conducting polymer research follows a similar approach to improve performance by implementing closed loop control. Qi, Lu and Mattes demonstrated the importance of closed loop control for conducting polymer actuators in 2002. Later Fang, Tan and Alici demonstrated that robust adaptive control scheme requires less effort than of PID control [26-27]. Similar approach has taken place on the control of Dielectric Elastomer Actuators. Carpi and De Rossi experimentally validating their electromechanical model
of a cylindrical actuator made of dielectric elastomer with [11]. Toth and Goldenberg addressed the possible use of Dielectric Elastomer Transducer as a sensory subsystem and an actuator, hence by measuring strain and capacitance in real-time control development [28]. On the other hand, Chuc et.al. addressed self-sensing capability of force at DE actuators [29]. Sarban, OuBaek and Jones introduced a closed-loop control of a core free rolled EAP actuator, same with the one used in this paper, using a gain scheduling algorithm with PI controller [30]. Gisby, Calius, Xie and Anderson shared the results obtained from a control algorithm using PWM signals to control current [31].

As aforementioned, most of the work addresses the importance of position control. Possible applications of EAPs include: robotics, medical and haptic devices where humans have direct contact with these interfaces, which requires robust force controllers to ensure safety and proper handling for human-machine interaction. Force control is now an integral part of robotic manipulators for certain tasks (such as: machining and assembly operations) not only requiring precision but also requiring predetermined forces and safety of the tool at hand. New innovations in the actuator technology still employ control techniques that were developed for safety and handling. As a further approach to control DEAs, the force controller presented in this section, and to the best knowledge of the authors, the literature regarding force control of DEAs is very limited, and most of the effort to date has been directed towards position control and/or control applications where the sensory information is position only. An experimental method to develop force control over DEAs was proposed by first defining an inverse model for the DEA; later P, PI, and PID controllers with and without the inverse model as the feed-forward term was compared.
3.2 MECHANICAL SYSTEM DESCRIPTION

Experimental comparison of DEA enhanced performing controllers required characterizing the actuator and then implementing the control algorithm. A test bed was built assembling a DEA, a force sensor, and a position sensor, as depicted in Figure 11. Structure of the test bed parts were fabricated with Rapid Prototyping (RP) from Accura 40 resin using a Viper SLA 3D system in the Biomedical Mechatronics Laboratory at Northeastern University. This SLA material was selected because it provides greater design capabilities, better precision for alignment and it is lighter compared to machined metal parts.

![Figure 11: Test Bed Close-up View](image-url)
The design of the test bed was done using SolidWorks with an assembly of three separate parts. From bottom to top, these are the yellow colored parts in sequence: test bed base, median linkage and potentiometer mount. After the design is completed as a SolidWorks part file (.SLDPRT) it is saved as a stereolithography CAD file (.STL). The STL file was then exported to be processed at 3D Lightyear software for production. After defining support structures and assembling the parts on the tray in the software, a binary format file (.BFF) was created to be sent to the Viper SLA device. A close-up view of the RP parts are shown in the Figure 2 and the overall experimental system is presented in Figure 12.

Passive damping elements were attached as four legs for support and alignment to the test bed. Load cell (Honeywell 31 Mid series) is rigidly attached to both test bed base and to DEA (InLastor-Push Actuator manufactured by Danfoss PolyPower A/S) via a clamped RP structure for force feedback. Mechanical stops were used for controller development phase to keep the DEA fixed between load cell and median linkage. Median linkage was used to carry weights and transfer the displacement of the actuator to the linear potentiometer (Active Sensor CLS1311). The linear potentiometer was used for some other tests related to vibration measurements. A potentiometer mount was built to align and support a linear potentiometer and the median linkage.

The load cell reading was connected to the input port of National Instruments terminal (NI SCC-68). By capturing the force reading through a real-time data acquisition card (NI PCI-6259 M Series), error was calculated in the LabVIEW Real-Time software in comparison to desired force value and necessary control command was sent through
output port of NI terminal. Output signal was amplified by high voltage amplifier (TREK 609D-6) to run the actuator.

Figure 12: Experimental Test Bed

3.3 SOFTWARE SYSTEM DESCRIPTION

The DEA’s open loop performance was measured and a nonlinear closed force control was implemented using a real-time platform: LabVIEW real-time. The LabVIEW real-time software of the DEA control system allowed accurate timing characteristics.

The main reason to employ a Real Time Operating System (RTOS) versus a general purpose Operating System (OS) was the accurate timing characteristic of RTOS. In general, OS are optimized to run a variety of applications simultaneously to ensure all applications receive processing time. These operating systems are susceptible to interruptions from other input devices such as the mouse or keyboard; even the graphical
displays can alter the processing speed hence the timing accuracy. In addition to the accuracy issue, there is limited control from the user over how these tasks are handled by the processor. As a result, high-priority tasks can be taken over by lower priority tasks, making it impossible to guarantee a fast response time for critical applications. Regular DAQ hardware running on a general-purpose OS such as Windows cannot guarantee real-time performance. In contrast, real-time hardware running on real-time operating systems (RTOS) allows the programmer to prioritize tasks and enable accurate timing.

A real-time system consists of a target system that runs the RTOS and a host computer (usually a laptop or standard desktop PC). The code is developed on the host, and then deployed to the RT target via high speed Ethernet communication. The RT target streams critical parameters back to the host for monitoring and data recording.

The graphical user interface that was used to collect the force information and check the actuator performance is depicted in Figure 13. The parameters that held core importance was received from the RTOS to Windows OS for monitoring purposes. Following a clockwise direction in Figure 13, 1) real time (dt RT) and windows (dt WIN) timing gauges display the loop speed during operation to ensure accurate timing; 2) overshoot, settling time and rise time was displayed to monitor control performance; 3) desired input was commanded using a toolbar that enabled step, square and sine signals channels as well as a user defined custom input; 4) proportional, integral and derivative control gains were altered with sliding bars and the nonlinear feed-forward term (to be explained in the next section) was enabled/disabled from the button on top of the bars; 5) the effect of controller tuning parameters and the nonlinear feed-forward term were displayed as the voltage commanded by the controller by the gauges below the sliding bars as well as the
total voltage supplied; 6) a force versus time chart was adopted to display the desired force input and the resulting actuator output.

![Graphical User Interface](image)

**Figure 13: Graphical User Interface**

By using the tools generated in LabVIEW environment for the controller graphical user interface, it was straight forward to fine tune the feedback controller. The effects of gain parameters were instantly displayed both on the graph and at the overshoot, rise time and settling time display bars. Altering the reference input signal to check controller performance and fine tuning the controller by taking advantage of information displayed on the GUI helped to form different set of controllers. It was also possible to record all of the data available on the GUI to a separate text file (.TXT).
3.4 INVERSE MODEL IDENTIFICATION

Electroactive polymer actuator characteristics are described using nonlinear equations; hence to improve the performance of the actuator these nonlinear characteristics has to be taken into account for controller development. Feed-forward control helps improve response time and steady state error by eliminating the work that needs to be done by feedback loop. In order to address desired force to required voltage, an inverse model was developed. This subsection explains the development procedure and the inverse model.

Blocked force measurement was carried out for verification of the specifications provided by Danfoss PolyPower. Measurements were conducted via restraining the actuator’s movement along the direction of motion and reading the force output using a compression load cell. No preload was applied. Ramp input voltage was supplied with increments of 1V per 5ms, reaching up to 2250V in 11.25 seconds. Result of this experiment was depicted in Figure 14 as the response for blocked force measurement.

![Figure 14: Blocked Force Measurement for DEA with No Preload](image-url)
To enhance control capabilities, blocked force measurement was used to develop a model to provide feed-forward gains. The inverse model that was developed upon blocked force measurements assumed no loading condition, therefore adversely affected the transient response of the controller. To compensate for transient errors, series of data sets were collected with various step responses in a single graph. Step response graphs of the actuator provided more reliable data for the improvement of transient response for the control. Figure 15 represents the strategy followed in order to construct the inverse model.

Figure 15: Inverse Model Construction Schematic
Steady noise signals, maximum actuator force (from varying preloading conditions) and voltage readings were averaged to obtain the step response open loop characteristic of the actuator. Average values of force capabilities were then graphed with respect to voltage. Before starting the force measurements, noise in the setup was measured with respect to time and applied voltage signal. While force reading was continuously recorded at 5 kHz, blocked system at rest was excited by a voltage value and then left to rest for a few seconds. Voltage was applied as step input of increments of 250V from 0 to 2.25kV. At every voltage level, actuator’s maximum force capability was recorded. The same voltage values were applied five times and recorded values of data were averaged to find the best approximate for every given preload condition. Averaged results of incremental voltage values were then put together to form the average step response curve for one preload condition. The same procedure was held for more preloading conditions starting from 0 to 18N with 2N increments. During measurement the weights of the actuator and its housing were offset as well as the test bed structure. By averaging the best approximate values for every preloading condition (from 0N to 18 N); a final curve was generated that accommodated each preloading condition as an average of load and condition size. Inverting this curve resulted in a graph that depicted the estimated voltage for desired force output. Average results are presented in Figure 16 as stars, and a third degree polynomial is fitted to express the relation as a mathematical model and is presented as a polynomial equation given below. This look-up table type model provides the inverse model for the controller to estimate the voltage needed to meet desired force input values.

\[ v = 0.0078f^3 - 0.115f^2 + 0.7365f + 0.0327 \]  \hspace{1cm} (3.1)
As can be seen from Figure 14 and Figure 16, voltage and force has a nonlinear relationship with each other. Equation 3.1 depicts the nonlinearity of the force-voltage relationship; hence Figure 14 combines multiple loading conditions and takes averaging of nonlinear relations therefore such a graph is expected.

![Graph showing voltage vs force](image)

Figure 16: DEA Inverse Model with Multiple Preloading Conditions

### 3.5 OPEN LOOP RESPONSE

Open loop control structure employed the inverse model polynomial, presented in the previous subsection, as the open loop gain. The command signal was later amplified in the high voltage amplifier to activate the DEA. The open loop schematic of the system is depicted below in Figure 17. The disturbances in the system due to external loading were described as the haptic interaction between the DEA and human contact.
The open controller gain was calculated using equation 3.1 and the command signal was altered by the inverse model via changing desired force values. However, the system did not have the feedback coming from the force sensor; therefore both the transient and steady state errors were not compensated. Response of the system to a step input using only inverse model as the feed-forward term for the force control resulted in steady state error of 25% as presented in Figure 18. Uncertainties of the system and its predictability depend on many parameters that cannot be modeled all at the same time, therefore to compensate for the steady state error a closed loop control was be implemented.
Result of open loop studies provided a mathematical model that assisted the controller to predict how much voltage is required to apply desired force. Data collection was held such that it takes into account the transient response of the actuator and overshoot to assist the feedback loop. This way, error that needs to be compensated is desired to be decreased with implementation of closed loop control.

### 3.6 CLOSED LOOP CONTROLLER

Performance of open loop response of the system was limited to the accuracy of the inverse model. The averaging technique that was used to merge actuator responses with various preloading conditions forms the limiting factor in the inverse model. A closed loop controller was developed to compensate for both the steady state and transient errors after feed-forward term.

Closed loop controller evaluates the difference between force output reading and the force input reading and calculates the error to be compensated by the signal sent to the actuator to acquire desired output. Closed loop controller is a necessity for applications that require known forces to be applied in known periods of time and it would enable more precise control of the DEA.

A schematic of the closed loop system and the control structure is represented in Figure 19. Desired force input was evaluated in the controller; error from the previous loop was calculated. Specific gains multiply the error, its integral and its derivative; if the inverse model was enabled the polynomial model fed extra voltage command forward. The signal
generated was then filtered using a moving average digital filter to smooth out short-term fluctuations that would overload the voltage amplifier. After applying a saturation function to apply voltage values only in between 0-2.3kV voltage signal was applied to DEAP actuator and force output of the device was fed back to find the error.

![Figure 19: Closed Loop Control Structure of the DEAP System](image)

Evaluation of the controller started with using only inverse model, but the steady state error suggested the need for feedback control. The steady state error was present because of averaging used to form the inverse model with different loading conditions; the inverse model was fixed and did not get adapted for changing input types. Instead of focusing on actuator characterization and having a well defined transfer function, DEA actuator was considered to be a black box and its response was recorded to find control parameters.

Zeigler-Nichols PID tuning method was adopted to find and implement the required controller. This controller was able to compensate for error compensation of a system of an undefined transfer function. Zeigler-Nichols method was a starting point before the fine tuning of the controller. The Ultimate Sensitivity method was one of the methods that Zeigler and Nichols described in 1940s. In the ultimate sensitivity method, gain
parameters (Figure 20) were adjusted by evaluating the oscillations of frequency and the amplitude of the system at stability limit. To apply the method, the proportional gain ($K_p$) was increased until the system became marginally stable; the gain value obtained is the ultimate (critical) gain ($K_c$) and the oscillation period is called the ultimate (critical) period ($T_c$). [32] Below is a table that demonstrates ultimate sensitivity method’s parameter relations.

<table>
<thead>
<tr>
<th>Control Type</th>
<th>$K_p$</th>
<th>$K_i$</th>
<th>$K_d$</th>
</tr>
</thead>
<tbody>
<tr>
<td>$P$</td>
<td>0.5 $K_c$</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>$PI$</td>
<td>0.45 $K_c$</td>
<td>$\frac{1.2 K_p}{T_c}$</td>
<td>-</td>
</tr>
<tr>
<td>$PID$</td>
<td>0.6 $K_c$</td>
<td>$\frac{2 K_p}{T_c}$</td>
<td>$\frac{K_p T_c}{8}$</td>
</tr>
</tbody>
</table>

![Figure 20: Zeigler Nichols Tuning Based on Ultimate Sensitivity Method](image)

The steps described above were followed to find preliminary values for controller gains when applying the Zeigler-Nichols Method for controller development. Gain parameters were fine tuned later by using the LabVIEW real-time code (presented in Software System Description subsection) that runs at a frequency of 5 kHz to calculate the rise time, overshoot, steady state error and settling time, and to display these parameters. The goal of the controller was set to be fast transient response with less than 5% overshoot and no steady state error. Haptic applications require a fast and accurate force controller. According to given parameters fine tuning of the controller was implemented.
The previously described setup was designed to work on a real-time platform which allowed accurate timing characteristics. Real-time system consisted of a host computer and a target computer that runs Real-time operating system. Codes were developed in a host computer that ran on Windows OS, and then the codes were deployed to the target computer. Communication protocol used between the host and target computer was a high speed Ethernet. Streamed data was displayed and saved at the host computer, and was displayed in the graphical user interface.

A graphical user interface for controller tuning consisted of input/output force graph, numeric rise time, settling time and overshoot displays of the streamed data. The user was able to select the type of desired input signal from a list consisting of: step, ramp, sinusoidal and square functions; and entered desired signal amplitudes and frequencies. Gauge type of displays were used to show the voltage requirements of each gain in the controller, therefore user can see the voltage draw for the controller gain immediately after a change. Figure 13 presents the GUI part of the LabVIEW real-time code generated for controller development.

### 3.7 RESULTS

Tuning protocol for the controllers was described above, comparison according to their steady and transient responses by contrasting closed loop P, PI, PID controllers with and without feed-forward term as well as considering only feed-forward term in open loop is as follows. Open loop control with nonlinear feed-forward term had a steady state error of 19%, therefore proportional gain was implemented on the nonlinear controller. After
tuning the proportional gain $K_p$ (Nonlinear P controller) had 4.35% steady state error with 0.19 seconds settling time and 19% overshoot. To compensate for the steady state error first integral gain was introduced, (nonlinear PI controller) steady state performance improved up to 0.5% steady state error with 0.13 seconds settling time but an overshoot of 22.3%. To compensate for the overshoot this time derivate gain was implemented and resulting (Nonlinear PID) controller had 0.1% steady state error with 0.05 seconds settling time and only 1.18% overshoot. Also, the feed-forward term was eliminated and only proportional gain was re-tuned (Linear P controller). This controller had a 30% steady state error, 25.65% higher than the Nonlinear P with 3.5% overshoot; a 15.5% lower than Nonlinear P. Linear PI controller was then introduced; it performed 23.6% overshoot with 0.13 seconds of settling time and a steady state error of 0.01%. Steady state performance was improved compared to Nonlinear PI controller, transient responses of Linear and Nonlinear PIs were close to each other with 1% better performance on overshoot of Nonlinear PI both having same settling times.

![Comparison of Controllers](image)

Figure 21: Comparison of Controllers
Lastly Linear PID controller was tuned, it had no overshoot, settling time was 0.05 seconds and the steady state error was less than 0.01%. Comparison of controllers with and without the feed-forward term for P, PI and PID controllers suggested the use of integral term to eliminate the steady state error and derivative gain helped improve the transient response for the actuator as can be seen in Figure 21. Not only step response but also, sinusoidal and triangular response tests have been constructed and Figure 22 and Figure 23 present the comparisons for these tests.

Figure 22: Linear/Nonlinear PID Controller Closed Loop Response to Sinusoidal Input.

Amplitude: 4N, Frequencies: 1,2,5,10 Hz Accordingly.
PID gains were held the same while testing for sinusoidal and triangular waves. Frequencies of 1, 2, 5 and 10 Hz were selected for the operation range to cover frequencies that are relatively close to human haptic perception levels. For the 1 Hz frequency of sine wave, Linear PID lead the desired input where as the Nonlinear PID followed it. In 2 Hz both controllers performed similar. Looking at 5 and 10 Hz signals Linear PID starts leading after 2.0 seconds where as Nonlinear PID lags. Similar results were observed from the triangular wave tests. At all frequencies Linear PID started to lead the desired signal whereas Nonlinear PID had a better performance over the Linear
PID on triangular signals. Hence, the Linear PID controller was not able to hit 4N peak for triangular signals.

3.8 SUMMARY

Interaction between an actuator and an object is crucial for certain operations, such as those between a human and an actuator. Many applications in medicine, robotics and haptics include direct contact of human with mechatronic devices. To date there has been little research on the force control of DEAP actuators hence the possible applications that utilize force control. Development of a nonlinear PID controller was presented in this paper to address the need for force controllers in devices performing with DEAs. Gain tuning was done by using Zeigler-Nichols and fine tuning was performed for every controller with a custom GUI written in Lab-VIEW Real-Time. Improved performance in P and PI controllers with nonlinear feed-forward terms was observed for transient response. Continuous sinusoidal and triangle waves have been tested, to simulate possible haptic application signal types, and performance of Nonlinear PID has been observed to follow desired trajectory with less error than Linear PID controller. Future work to improve the force control capabilities will be robustness tests varying amplitude of the step signals without re-tuning. Later on, an adaptive method to form the feed-forward term and re-tune the gain parameters will be researched.
4 EVALUATING HAPTIC USE OF DEAP ACTUATORS

4.1 INTRODUCTION

Studies have shown that biomedical applications of EAPs are one of the most promising fields that take advantage of inherent characteristics of polymer material based systems [33]. DEAs are made using DEAP films, they operate silently, they are soft and their compliance is close to that of the human body. DEAPs are highly modifiable because the film thickness, the elastomer material and the electrodes of a dielectric elastomer could all be altered depending on the stiffness values required for a specific application. Being able to use a sensing structure that has tunable stiffness properties is a very important asset to improve the compliance range of human machine interfaces. DEAs as an implantable diaphragm muscle have already been tested. In addition to body conformity, DEAP’s efficiency enable use of small size batteries hence improving performance on remote operation [33]. Larger sizes of DEAP films and actuators could be used for large areas of the human body. On the other hand, material properties enable the use of small patches and generation of small scale actuators as well.

The possible use of DEAs as a building block to provide haptic interface solutions to the medical field is investigated in this section. The haptic interface example here uses for the first time a DEA as a haptic heart rate monitoring system to transfer the heart beat of a person to any location. In order to quantitatively evaluate the performance of the DEA for use in this haptic application it was compared in human subject testing with a Linear Voice Coil Motor (LVCM).
4.2 HAPTIC TESTS WITH ACTUATORS

Compared to electromagnetic motors and linear actuators, DEAs have higher power density. They are lightweight, have large displacements and better efficiency at lower speed operation [10, 34-36]. Electroactive Polymer based actuators have been under investigation for over two decades, however electromagnetic motors have been around for over a century. Linear Voice Coil Motors (LVCM) is one of the fundamental electric motors founded on electromagnetic theory. These motors consist of permanent magnet housing and a moving coil. When a current across the terminals of the motor coil is applied, it produces a magnetic field and the coil translates with respect to the permanent magnet. Motion control is provided by altering the magnitude and polarity of the current and the generated force is proportional to the current that flows through the coil. Figure 24 demonstrates both actuators that are described above.

Some of the important parameters of actuator performance for haptic applications are given below in Table 1 to demonstrate the actuators characteristics. Both manufacturers’ specification sheets and laboratory measurements were used to construct the table. The rolled DEA has lower strain levels compared to the LVCM however, other types of DEA actuators was shown to have higher stroke levels [37]. DEAs are slower compared to LVCM due to their inherent impedance effects. The maximum force that could be applied with respect to actuator weights is comparably close to each other. DEA was observed to have lower power rating compared to LVCM due to its smaller stroke characteristic. DEA’s can operate under very high magnetic forces which enable designing robotic systems that are used in MRI operating environments.
Figure 24: Linear Dielectric Elastomer Actuator, Linear Voice Coil Actuator

Table 1: Actuator Properties for Evaluation

<table>
<thead>
<tr>
<th>Criteria</th>
<th>DEA</th>
<th>Voice Coil</th>
</tr>
</thead>
<tbody>
<tr>
<td>Maximum stroke</td>
<td>1.2mm</td>
<td>57.2mm</td>
</tr>
<tr>
<td>Maximum force-to-weight</td>
<td>58.26 (N/kg)</td>
<td>61.26 (N/kg)</td>
</tr>
<tr>
<td>Maximum power-to-weight</td>
<td>7 (W/kg)</td>
<td>29.6 (W/kg)</td>
</tr>
<tr>
<td>MRI Compatibility</td>
<td>Good</td>
<td>Poor</td>
</tr>
</tbody>
</table>

4.2.1 DIELECTRIC ELECTROACTIVE POLYMER ACTUATOR SYSTEM

The first test setup consisted of a DEA (InLastor-Push Actuator manufactured by Danfoss PolyPower A/S) rigidly attached to the bench via a clamp, presented in Figure 25. Actuator motion and interaction forces were regulated through LabVIEW by using an inverse model that maps required voltage value to the maximum force generated by the
actuator at the interaction point [1]. The system was operated under open loop control;
low voltage command signal was amplified by a high voltage amplifier (TREK 609D-6)
to run the actuator. Data acquisition was done by a National Instruments USB 6216 card.
A conventional laptop with USB port was used to run the LabVIEW code and data
acquisition.

Figure 25: Dielectric Elastomer Actuator Setup

4.2.2 LINEAR VOICE COIL MOTOR SYSTEM

The second setup consisted of a LVCM (Moticont Linear Voice Coil Motor, LVCM-051-
089-01) rigidly attached to an aluminum test base that was machined is presented in
Figure 26. The LVCM was supported by the aluminum structure at the base and linear
motion was aligned by a linear motion guide (THK, RSR 9KM) rigidly attached to the aluminum body. A rapidly prototyped fixture was fastened on top of the LVCM to ease hand/finger grip. A closed-loop controller that regulates the current going to the coils of the voice coil actuator was used to control the force applied by the actuator. For this purpose, a commercially available servo amplifier (Junus) from Copley Motion was used. The amplifier can be programmed through its software CME2, which was used and the connection was established via an RS-232 protocol. The gains of the current loop were adjusted through this software. The controller measured the apparent current on the coils, compared that with the current commanded to calculate the error. A Proportional and Integral (PI) controller regulated the voltage based on this error, thus making sure that the desired current was actually being supplied to the voice coil actuator. The desired current was sent to the controller through an analog input with the analog voltage supplied from a National Instruments (NI) USB 609 terminal. The constant for analog command was set as 5 Volts = 1.0 Amps. The force constant of the voice coil actuator is 10.1N/A. Therefore, when a 1 Volt analog output command is given to the servo amplifier, the desired current is set as 0.2 Amps which amounts to 2.02 N. The DC voltage to run the servo amplifier was supplied by a TENMA 72-2080 DC power supply.
4.3 SOFTWARE DESCRIPTION

Both the DEA and LVCM setups employed similar LabVIEW codes to control the actuators and to acquire sensory data. Both codes were run on Windows OS at 40Hz, enabling a working frequency that was greater than the actual bandwidth of human force control perception limits that is up to 7 Hz [38]. The code generated for the human subject testing also included the actuator control algorithms embedded in it. The control scheme followed for the DEA was an open loop architecture as the one described in the section 3.6. The controller employed the inverse model to calculate the required voltages to run the actuator and sends the command signal to the high voltage amplifier. On the other hand, the LVCM system employed the closed loop controller provided by the commercial controller described above. The controller input was supplied according to the human subject test procedure described in the protocol section below. A similar
approach to the input signals generated with LabVIEW real-time system for force control was followed in Windows OS. The input requirements for the human subject tests were square, sinusoidal and sawtooth waves. In order to generate the difference between signal types and amplitude, variable signal sample size for these signals were generated in the LabVIEW environment and later assigned channel that was accessible by the test personnel. A snapshot from the front panel of the code used is displayed below in Figure 27.

![Figure 27: Front Panel of the Test Code](image)

The variable signal information was kept the same for both system setups, the only difference between DEA and LVCM software was the control algorithm used to send control signals. The input generator was able to provide the square sinusoidal and sawtooth signal types.
4.4 HUMAN SUBJECT TESTS

A series of human subject tests was conducted to compare the human haptic perception when DEA and LVCM were used in the simple force-feedback setups described in the previous subsections. The set of experiments involved the usability of the actuators as haptic interfaces and considerations regarding how comfortable the actuator felt to the user. The necessary approval has been given by Northeastern University’s institutional review board for Human Subject Research Protection rules. Tests consisted of two main parts: a) providing objective and b) subjective evaluation. In the objective evaluation part there were two tasks for the subjects to perform that would allow quantification of the user response on the comfort and quality of forces exerted by the actuator. Tasks that subjects were asked to complete in the objective testing included tick count and wave estimation. In the subjective measurement method, an assessment scale from zero to six was selected and subjects were asked to score the comfort, smoothness of both actuators and present an overall score based on the experience with both actuators after performing the objective evaluation tasks. The following subsections describe the test protocol, results and discussions on the human subject tests for the haptic evaluation of the aforementioned actuators. The human subject testing procedure is represented in Figure 28.

4.4.1 PROTOCOL

In the objective testing experiments, types of signals used include: impulse, sinusoidal, square and sawtooth. In the tick count tests subjects were asked to count the number of ticks they feel as the actuator follows several signal properties. To generate the ticks,
impulse signal with duration of 15 seconds were generated using LabVIEW with a predetermined number of impulses (ticks). Ticks referred to the peaks of the signal introduced during the task. The variable for comparing results was the error that they make while guessing the tick count. The error was calculated as the difference between the predetermined value and the guess of the subject. The predetermined value was selected as odd numbers (15, 17 and 19) to eliminate the probability of subject making up a value when they lose the count. It was noted from previous studies that round numbers are more expected by the subjects and any attempt to guess the correct number would be to round the number if they are not sure [39].

In the wave estimation tests, subjects were asked to tell what wave type they felt during the experiment. For this task, sinusoidal, square and sawtooth signals were applied at 1Hz
with peak-to-peak force amplitude of 3N with 1.5N offset introducing only push action. Guesses were saved as true or false and later compared among the two actuators to determine if there was any pattern that could relate to user perception. By looking at the subject responses it was possible to detect how the transitions between forces were acquired by the subjects. These two tasks were selected as an experimental way to determine the capabilities of the actuator from haptics point of view.

In the subjective testing procedure, volunteers were asked to score the comfort levels, smoothness of forces applied on their hand and the overall quality of the interaction forces of both actuators. For subjective score assessment, scaling from 0 to 6 was used, 0 being extremely bad and 6 being extremely good as depicted in Figure 29. This test constituted the subjective part of the experiments, the result that was captured by this data was based on human preference, and it is dependent on how subjects perceived the actuators.

![Figure 29: Scaling Used for Objective Tests](image)

In this experiment, it was feasible to ask the same subject to participate in multiple tasks and as a result within-subject testing was used to identify the attributes of the system. We had the subjects to place their hand on top of the Dielectric Elastomer and the Linear Voice Coil actuators for the same periods of time under same conditions. The preliminary
experiments were performed at the Biomedical Mechatronics Laboratory at Northeastern University in Boston. The test subjects were selected from the student population mostly MS and PhD volunteers from Northeastern University. Tests were constructed in accordance with the Human Subject Research Protection rules and regulations. There were 6 male subjects, all between 20 to 30 years old. They did not have any previous diseases or disabilities related to sensation, motion and coordination of the hand. Whether the subjects’ dominant hand was left or right, did not effect this system under the study, therefore left-handed and right-handed participants were subject to the same test procedure. Volunteers were asked to visualize three signals mentioned above, if not experimenter explained volunteers how they appeared. Experiments were constructed to not to exceed 15 minutes. In the first couple of minutes the test procedure was explained and subjects were asked if they have any health related problems that could raise biases on the test. Participants’ names were kept anonymous. A researcher was responsible to construct the tests, record subject’s answers, and monitor the software for data acquisition and actuator control. The testing parameters have been randomized among subjects, each of them different order and number of tests regarding the same parameters. Test time was kept short; this way there was no lag between the test and the response of the subject. The tables representing both wave estimation and subjective evaluation results are presented in the next section.

4.4.2 TEST RESULTS

Results of the human subject tests to determine the haptic interaction capabilities of DEA actuator in contrast to LVCM are presented in this section. Figure 30 and Figure 31 show
the tick count test results in which the lightest color is 15 ticks and the darkest is 19 and
the median color is 17 ticks. While testing the DEA, participant 5 made two counting
errors. In the LVCM the same participant made one error but at this time participant 3
also made a mistake while counting the ticks. By using DEA fewer errors were made to
count how many time the actuator ticks according to these tests. In addition to tick count
wave estimation test was held for objective measurement.

In the wave estimation tests, subjects were asked to determine what type of wave signal
was introduced by the actuator and the results were collected as true or false. Left hand
side of the below table for each actuator presents the wave estimation test results. Total
number of false estimations was 3 for DEA and 5 for LVCM. Only two of the
participants (1 & 4) both made estimation mistakes.

![DEA Tick Test](image.png)

Figure 30: Tick Count Test Results for Dielectric Electroactive Polymer Actuator
Figure 31: Tick Count Test Results for Linear Voice Coil Motor

Table 2: Wave Estimation and Subjective Test Results for DEA

<table>
<thead>
<tr>
<th>Participant</th>
<th>Wave Estimation</th>
<th>Questions</th>
</tr>
</thead>
<tbody>
<tr>
<td>Test 1</td>
<td>Test 2</td>
<td>Test 3</td>
</tr>
<tr>
<td>sine</td>
<td>square</td>
<td>saw tooth</td>
</tr>
<tr>
<td>1</td>
<td>TRUE</td>
<td>TRUE</td>
</tr>
<tr>
<td>2</td>
<td>TRUE</td>
<td>TRUE</td>
</tr>
<tr>
<td>3</td>
<td>TRUE</td>
<td>TRUE</td>
</tr>
<tr>
<td>4</td>
<td>FALSE</td>
<td>FALSE</td>
</tr>
<tr>
<td>5</td>
<td>TRUE</td>
<td>TRUE</td>
</tr>
<tr>
<td>6</td>
<td>TRUE</td>
<td>TRUE</td>
</tr>
</tbody>
</table>
Table 3: Wave Estimation and Subjective Test Results for LVCM

<table>
<thead>
<tr>
<th>Wave Estimation</th>
<th>Questions</th>
</tr>
</thead>
<tbody>
<tr>
<td>sine</td>
<td>comfort</td>
</tr>
<tr>
<td>FALSE</td>
<td>3</td>
</tr>
<tr>
<td>TRUE</td>
<td>5</td>
</tr>
<tr>
<td>TRUE</td>
<td>2</td>
</tr>
<tr>
<td>FALSE</td>
<td>5</td>
</tr>
<tr>
<td>TRUE</td>
<td>6</td>
</tr>
<tr>
<td>TRUE</td>
<td>4</td>
</tr>
</tbody>
</table>

Subjective test results are depicted on the right hand side of the tables (Table 2 and Table 3) for both actuators, listed as “questions”. Numerical scaling from 0 to 7 was used to grade the comfort, smoothness and overall level of satisfaction. While some of the participants were more comfortable and preferred the use of the DEA, others liked the LVCM better for regarding both the tick count and the wave estimation tests. Overall point given by the participants using the DEA were close to each other, however the LVCM presented a larger variance.

4.5 DISCUSSION

According to the preliminary results gathered with the objective tests, the participants were inclined to make fewer errors while counting the ticks and guessing wave types using the DEA. The subjective comparison scores suggested that participants felt more comfortable using the LVCM and forces felt smoother. This showed that, participants
perceived the applied forces incorrectly while feeling comfortable. The participants’ subjective response on the DEA had smaller standard deviation compared to LVCM, and this showed that their behavior towards using a DEA actuator was very similar. The LVCM had a higher standard deviation in subjective analysis which could be explained that not everyone favored the use of magnetic coils similarly. The differences were on the order of 1-2%, which implied the necessity to increase the sample size. One of the points that need to be addressed is that the DEA setup used the inverse model developed for a nonlinear force controller however; the LVCM was running under closed-loop force control. Although the DEA was working on open loop control its performance was at least as good as that of the LVCM setup.
5 NETWORK HEART RATE EMULATOR

To show the viability of using a DEA as a component in a haptic system a demonstration was staged in a laboratory setting where the heart rate of a subject was transmitted over a network and then mapped onto the DEA so that it could be “felt” by another subject. In this way the DEA was used as the force feedback component of a teleoperated system and it was the first time when DEA was used for heart rate monitoring. The force feedback via DEA was observed to be very close to an actual heart beat. The global schematic of the system can be seen below in Figure 32. The heart rate of the subject was collected as they pedaled a stationary bicycle as a digital signal via a chest band (Polar). It was then wirelessly sent to signal receiver that is attached to a National Instruments DAQ card to be transferred to the first computer (PC1) which also displayed a blinking light to present the instant heart beat information. The digital signal changed to true when the subjects heart beat and remained false in between beats.

Figure 32: Haptic Heart Rate Emulation Diagram
From the PC1 the data traveled over the local network to PC2 via UDP. The data was sent as a string so to preserve the digital data of the original signal. Once it arrived at PC2 the digital data was used to switch the inverse model on and off so that the heart rate of the subject was mapped correctly by the DEA. The second subject could then “feel” the heart beat of the first subject by the haptic feedback of the DEA. As well as transferring heart rate information, with data acquisition tools developed at LabVIEW environment, it was possible to graph the continuous heart rate information and detect the rate of change of heart beat in time.

5.1 MECHANICAL AND SOFTWARE SYSTEM SETUP

The network heart rate emulator system setup consists of an exercise bike. A Polar wireless heart rate chest strap to emitter the heart beat information signal and a wireless antenna to detect the Boolean signal. A NI 6009 USB data acquisition card was used to transfer the signal from the antenna to a remote laptop. By forming a network system in the Biomedical Mechatronics Laboratory, heart rate information was transferred through UDP connection. There was a second computer that was stationed by the actuator station. A second NI 6216 DAQ picked up the heart rate information, enabled signal monitoring and initialized actuator control at the other end of the network. A high voltage amplifier, TREK 609D-6 was used to supply the command voltage to the Danfoss PolyPower, InLastor PUSH actuator (DEA) to provide the haptic feedback to the user. A closer view of the biker’s end in the haptic feedback transfer setup is represented in the Figure 33 with a picture of the subject and setup for the heart rate input system.
5.2 RESULTS

The digital signal that was acquired by the heart rate sensor to switch the DEA is plotted for visualization in Figure 34. The same data was used to calculate the heart rate by measuring the time elapsed between pulses. This demonstration was performed in a laboratory environment but it can easily serve as a proof of concept for use between any two locations with internet access. This is important because both the equipment for the PC1 setup and the PC2 setup are portable. The demonstration worked very well and a user at the end of the network was clearly able to distinguish the heart beat information.
5.3 SUMMARY

The probable use of DEAP based actuators (DEAs) in medical and haptic applications were brought to the reader’s attention in this chapter. DEAP based systems have certain characteristics that could be tailored towards specific use in particular medical settings and haptic interfaces. Possible use of DEAs as haptic heart rate emulators for patient monitoring was investigated. In order to evaluate how good a DEA can perform in haptic interactions a comparison with a LVCM was documented with preliminary results of human subject tests. Although it was not conclusive, experiments showed that the DEA performed at least as good as the LVCM for haptic feedback tasks. Subjective answers could be explained by participants that had slightly different perception and preferences over the feeling of touch. A better identification in force quality and human perception of haptic devices could be accomplished with a larger sample of participants. Human factors in the design process for new actuator and sensors for haptic applications should be kept as an integral part of conceptualization designing process.

Figure 34: Heart Rate Data as Pulse Signal
6 DIELECTRIC ELECTROACTIVE POLYMER BASED BLOOD PRESSURE MONITORING SYSTEM

6.1 INTRODUCTION

This chapter explains the synchronized blood pressure measurement device that was proposed for the 2010 Primary healthcare innovation grant to Center for Integration of Medicine and Innovative Technology (CIMIT) for the development of DEAP based pressure sensors embedded in a pressure cuff. The advantages of the proposed new device are: a) the size of the device for BP measurement is considerably reduced; b) the accuracy for measuring BP is increased in a robust and adaptive way; c) it offers possibilities for wireless communication with a networked computer that could allow home based, continuous BP monitoring for telehealth applications; and d) it contains low-cost and disposable pressure sensing components. The technology is based on the use of Dielectric Electroactive Polymer (DEAPs). DEAPs are plastic, thin-film like materials that change their electromechanical properties upon application of a voltage and vice versa. In this project DEAPs was used to design very compact, low-cost and more accurate measurement devices for home-based monitoring of blood pressure. Hence, the mission of this project is to complete the development and demonstrate initial clinical use of the new BP measurement device, called SyBeD (Synchronized Blood Pressure Measurement Device).
6.1.1 SPECIFIC GOALS

The proposed BP measurement device consists of a DEAP based cuff design; an embedded microcontroller, user-friendly data storage and graphical user interface (see Figure 35). After the system is initiated by a start button, an DEAP based pressure sensor is used to provide feedback to the controller. The controller is the unit that controls the amount of pressure that has to be applied on the BP measurement location. This device combines precise BP measurement with modern wireless communication protocols, extending the remote monitoring system from patients to athletes and military personnel as well; hence making SyBeD one of the next generation remote monitoring technologies. In addition SyBeD features a disposable, low-cost pressure cuff with an embedded capacitive pressure transducer and a compact, user-friendly pressure measurement and display box.

Figure 35: Proposed Concept for SyBeD (Syncronized Blood Pressure Measurement Device).
The project’s specific aims are:

(1) To develop a novel DEAP based cuff system for BP measurement through bench and pilot clinical use testing into a finalized go-to-market stage “product”-device-integrating precise measurement with ease of use and data transfer.

(2) To demonstrate clinical utility and gather preliminary benchmark clinical testing data in side-by-side comparisons of new to existing measurement techniques and devices’ capabilities.

6.1.2 METHODOLOGY

6.1.2.1 SYBED'S LOW COST AND HIGH ACCURACY FEATURE

*SyBeD* is a non-invasive, mercury-free, automated blood pressure measurement device that uses the oscillometry technique to measure blood pressure from the brachial artery. The main difference of *SyBeD* compared to the devices in the market is the disposable and low cost nature of its cuff that is "synchronized" (note: the term "synchronized" used in this project to indicate shape and control parameter adaptation in real time) for different arm sizes and includes a novel DEAP film based pressure transducer in it. One of the biggest advantages of using a disposable cuff with a sensory element embedded in it is that, it removes the problem of (re-)calibration and hence it increases BP measurement accuracy.

It is well known that every sensory element needs to be (re-)calibrated after a certain time of use. Therefore, BP measurement devices generally require to be serviced regularly as well. If they are not regularly serviced, the pressure measurements would be different
than the actual value. An inspection showed that only 5% of the devices were regularly under maintenance in a major teaching hospital [40]. The inspection also revealed that 50% of the manometers in use were defected having thus a direct impact on the measurement quality and accuracy of the diagnosis. It is clear from these findings that, neither hospitals and health practitioners nor patients have the time to conduct regular maintenance to their BP measurement devices.

*SyBeD*'s cuff is designed to be disposable and have an embedded EAP based pressure sensor. These two design properties solve the calibration problem of BP measurement devices in an intuitive and low cost manner. Instead of following up maintenance schedules, users can replace their cuff after using it a predefined number of times. Furthermore, personal / disposable cuffs help eliminate the transmission of infectious diseases that can be transferred through skin contact when the same BP measurement device with the same cuff is used in different patients. Finally, selection of the correct occluding cuff size is an important parameter for accurate blood pressure measurement. *SyBeD* eliminates the accuracy problems that may arise by using the wrong cuff size especially in over-weight and elderly patients. The "synchronized" concept relies on helping patients determine the correct cuff size that matches their arm periphery through proper feedback and learning. This would be done by measuring patients arm size with a measuring tape to determine which range they fit into according to the guidelines for appropriate blood pressure cuff selection. The studies on effective blood pressure measurement depict that: the bladder length should be 80% and the width to be at least 40% of the arm circumference. Therefore recommended cuff sizes are presented in the Table 4 below [2].
In addition, *SyBeD* can adapt the controller gain parameters to different arm sizes in order to increase the accuracy of BP measurement by linking the cuff size information to the clip-on control box. The amount of air supplied is related to the cuff and bladder size. In order to control the speed of supplied air pressure, air flow velocity control gain will be adapted according to the cuff size. A series of pressurizing tests for different bladder sizes with an air pump will be conducted to determine the gain parameters. This would enhance to reaction time to achieve pressurizing rates specified with the oscillometry technique.

Table 4: Recommended Cuff Size for Blood Pressure

<table>
<thead>
<tr>
<th>Arm Circumference</th>
<th>Cuff Size</th>
</tr>
</thead>
<tbody>
<tr>
<td>22 to 26 cm</td>
<td>Small adult: 12 x 22 cm</td>
</tr>
<tr>
<td>27 to 34 cm</td>
<td>Adult: 16 x 30 cm</td>
</tr>
<tr>
<td>35 to 44 cm</td>
<td>Large adult: 16 x 36 cm</td>
</tr>
<tr>
<td>45 to 52 cm</td>
<td>Adult thigh: 16 x 42 cm</td>
</tr>
</tbody>
</table>

### 6.1.2.2 SYBED'S TELEHEALTH CHARACTERISTIC

Inclusion of a wireless communication element enables *SyBed* to be used for patient monitoring to detect variations in blood pressure and remotely notifying healthcare personnel (see Figure 36). Tele-operated systems are highly recommended for hospital, nursery and home use as they would increase the capabilities of the healthcare
professionals in nursing organizations where human resources are scarce. For many people like the elderly and those recovering from a major ailment and that they spend most of their time at home, it is very beneficial to be able to frequently send vital signals directly to their primary care physician for monitoring. In addition, the potential savings regarding time and resources of personnel and healthcare costs are worth considering. Some of the western European countries have already started investing in telehealth services to increase efficiency in providing healthcare to the aging population. For example, two pilot studies have been initiated in Denmark in June 2009 and a national roll-out of remote patient monitoring systems are expected to be completed by 2012 [41]. Furthermore, studies conducted on the effects of telemonitoring on hypertension and blood pressure control stated that patients who have been involved in telemonitoring of blood pressure had higher success rate to regulate their blood pressure levels compared to patients who have been involved in standard blood pressure monitoring processes [42-43]. In order to take advantage of these important findings telemonitoring devices should be easily accessible and they should contain wireless communication packages that can be linked to personal computers and cellular phones.
As well as providing the patient or the primary care personnel immediate and precise BP measurement, patients and medical practitioners will be connected to each other through the wireless network. Thus, practitioners will have the ability to outsource expert clinical services and export services to foreign satellite care sites even internationally. Availability of continuous wireless data transfer to health care personnel will enable detecting emergencies prior to them happening as well as storing effective data for further and detailed analysis of daily activities.

SyBeD supports telemonitoring applications with its Bluetooth communication protocol. The wireless communication enables direct transfer of the instantaneous blood pressure data to a central location through local access ports. The local ports consist of laptops and cellular phones that are compatible with Bluetooth communication. By simply installing a software package to their laptops or mobile phones, users will be able to transfer blood pressure information (and other signals and information) to their doctors (see Figure 36).

Figure 36: SyBeD's Telehealth Capabilities
6.1.2.3 COMPARATIVE ADVANTAGES AND DISADVANTAGES

There are about 74.5 million people of age 20 and above with high blood pressure (hypertension) in the US. Studies show that, of the people with high BP, only 77.6 percent were aware of their condition. Being aware of the condition is a life saver since hypertension is a curable disease if detected. Some companies have already introduced automatic BP measurement devices to the market in the $50-100 range. However, these devices present several problems including inappropriate BP cuff sizes that result in very large errors in BP measurement in certain patient populations (such as obese people) and inability to provide continuous (even 24h if needed) BP monitoring at the home setting. Medical practitioners and policy makers strongly agree that home based, automated, precise and in some cases continuous, non-invasive, BP monitoring could considerably reduce healthcare costs while it can improve prevention of fatal events due to high BP incidents [44].

Compared to the generic types of BP measurement equipment, SyBeD could be classified as an advanced automated spot-check device. It is mercury-free, portable, lightweight, compact, easy to use and contains no observer bias. In addition to these characteristics the nature of the cuff makes it a smart monitoring device as well. It is capable of determining the cuff size due to the connection port between the cuff and clip-on control box. Some of the automated spot-check devices have small memory options to save the previous blood pressure data. SyBeD takes it to the next level by simply integrating wireless communication into the portable clip-on control box, hence enabling instantaneous data monitoring and continuous data saving without memory problem.
The *SyBeD* concept focuses more on the home use and remote monitoring, therefore ease of operation and patient comfort are top priorities. Synchronized cuffs that enable patient specific data analysis are estimated to increase the accuracy of the blood pressure measurement. With the low cost manufacturing of embedded sensor pressure cuffs and *SyBeD*'s disposable features, recalibration related problems are eliminated and accurate, non-biased measurements can be provided.

### 6.1.2.4 ADVANTAGES OVER TYPICAL CURRENT PRIMARY CARE PRACTICES

The resulting synchronized cuff is designed to be fitted to every patient that needs BP monitoring and the blood pressure data can be transmitted to health care personnel over wireless communication protocol. Complications caused by inappropriate cuff sizes are eliminated by employing DEAPs as pressure sensing elements embedded in the pressurizing cuff. *SyBeD* alerts the user about his/her cuff size and reminds them to make sure they are using the correct one because selection of the correct occluding cuff size is an important parameter for accurate blood pressure measurement. The conformity of the patient with the blood pressure measurement device is another *SyBeD*'s point of interest. Our current research on DEAPs shows that material compliance is close to the human skin compliance, hence making DEAPs strong candidates to be used in wearable sensor technologies [2]. *SyBeD* is capable of adapting pressure measurement for various patients because it takes into account the differences between each patient.
6.2 MECHANICAL DESIGN

*SyBed* consists of two main modules as shown schematically in Figure 37. The first module is the Pressure Cuff equipped with an embedded pressure sensor. The second module is the Clip-on Control Box containing the data acquisition, wireless data transfer and pressure control electronics. The pressure cuff consists of a bladder to exert pressure to the artery for occlusion and an embedded DEAP based pressure sensor to regulate the pressure at the cuff-arm interaction and the CAD design to represent these features is depicted in Figure 38.

![Diagram of SyBeD Components](image)

**Figure 37: Schematic of the SyBeD Components**

The clip-on control box consists of a capacitance-voltage transducer, a mini scale pump, a printed circuit board to control and acquire data, and an LCD screen for display. A wireless (Bluetooth) connection provides communication with a central network in which blood pressure and pulse data could be monitored remotely. Pressure regulation and accurate pressure sensing play a vital role in measuring the blood pressure using
occlusive cuffs as the one used in *SyBeD*. Details on how these issues were addressed in *SyBeD*'s design are described in the sections below.

*SyBeD*'s DEAP based pressure sensors are embedded inside the cuff to enhance blood pressure monitoring capabilities of the device. A series of sensors could be assembled inside the cuff as represented in Figure 39. By measuring the pressure across all of these sensors it is feasible to have an accurate measurement of the blood pressure at the artery. The DEAP films are very thin, and the required sensory element is relatively small in quantity. All the embedded sensor elements are connected through embedded wires to a connection port. The connection port establishes both electrical and pressure connections via metal pins. The structure of the connecting ports is the same although the size of the cuff varies. Metal male and female pins are connected to each other when placed on the marked areas (represented in yellow in Figure 39). *SybeD*'s design makes it straightforward to connect the pressure cuff to the clip-on control box.

Figure 38: Virtual Representation of the SyBeD Design
6.2.1 PRESSURE CUFF

The pressure cuff consists of the cuff itself and the embedded sensory element as shown in Figure 40. The cuff consists of a bladder, a textile shield and Velcro strip. The bladder is used to be filled with air to pressurize and occlude the artery. It is covered with textile to act as a shield and as a supporting structure to distribute the pressure. The pressure required for the occlusion is provided by an air pump situated in the clip-on control box and the connection is done via elastic tubes. The textile cover also acts as the intermediary surface between the arm and the bladder. After the cuff is tightened manually, Velcro strips are used to secure the cuff around the arm. For ideal measurement, the cuff should be worn proximal to the brachial artery and the elbow joint; this will provide the occlusion required for the oscillometry method.
Wearable sensors are gaining more scientific interest and their use as textile like sensors and their implementation in wireless health services is highlighted in recent studies [45-48]. To provide the required pressure information SyBeD employs flexible, thin, wearable, DEAP based pressure sensors. The use of DEAP flexible films for wearable sensor development is very limited. In this project we used DEAP thin films that are manufactured in Denmark by our industrial collaborator PolyPower A/S (http://www.polypower.com).

6.2.2 CLIP ON CONTROL BOX

The clip-on control box, shown in Figure 41, was designed to be ergonomic, compact, portable, and user-friendly. The box consists of a printed circuit board (PCB) for electronics, a Liquid Crystal Display (LCD), a Bluetooth device for wireless
communication and an air pump as shown in Figure 42. The PCB is designed to acquire control pressure data and manipulate them for wireless transfer. The user commands are transferred using push buttons that control a) the power ON/OFF; b) the measurement initiation and c) the data saving/sending. The LCD is a two color screen, and will display the diastolic and systolic pressure levels as well as providing the pulse information.

![Figure 41: Clip-on Control Box for Measurement Electronics and Wireless Communication.](image-url)

There are three types of interfaces between the control box and the cuff: a) mechanical connection to attach the control box on the cuff; b) electrical connection to transfer the voltage change from the pressure transducer; and c) air connection to provide the air supply needed to apply the needed pressure on the arm.

The embedded sensor information is transferred to the clip-on control box via a circular-fit connector. The connector provides both pressure flow and data transfer. The hollow
structure in the middle allows the air flow and the bindings on the sides of the circular connections are mated to transfer the instantaneous capacitance values.

As shown in Figure 42 and Figure 43, the PCB contains: a) the circuit to measure the capacitance change across the EAP pressure sensor; b) filters and amplifiers that will clean and tune the signal output from the EAP pressure sensor and A/D converters to digitize the data for processing; and c) a microprocessor to regulate the pressure and calculate the blood pressure.
A capacitance to voltage converter microchip is utilized to convert the capacitance value to voltage as the pressure sensor’s output. The measured voltage value relates to a pressure reading at the embedded sensor in the pressure cuff. The reading observed from the sensory element (variable capacitor) will be filtered with low-pass and band-pass filters to eliminate noise. The low-pass filter will be used to alter pressure regulation accurately and the band-pass filter will be used to determine the fluctuations in the pressure reading. The output of the sensor will be differential; therefore differential amplifiers and filters are required. Digitization of the sensory information is followed after filtering. A microprocessor correlates the voltage value to pressure instantaneously. The processor and A/D converter should run simultaneously with same rate. The highest frequency signal will be acquired for the heart cardiac synchronous fluctuations, and it is around 5Hz. According to Nyquist sampling theory [49] minimum rate to acquire this information should be 10Hz. To characterize these fluctuations efficiently 5 to 10 samples should be recorded per fluctuation. Continuous acquisition of the pressure information is transferred to the air pressure control section of the box. By altering the
command signal to the air valve and pump, air flow that is used to pressurize the cuff can be controlled with a microprocessor. Microprocessor is the unit that runs the program to acquire the sensory signals, to control cuff pressure for the automated measurement procedure. Using the pressurization parameters and intervals that are well described in the oscillometry technique, the blood pressure measurement is taking place. In order to find the magnitude of the fluctuations, fluctuation signals will be examined continuously. Maximum value of the magnitude determines the Mean Arterial Pressure (MAP). Diastolic and systolic pressure values are determined with by using multiplication factors of 0.85 and 0.55 respectively that estimate blood pressure. It is also possible to calculate the heart rate, in addition to the blood pressure (MAP, DBP and SBP). Frequency of the fluctuations is dependent on cardiac activities; hence the heart rate is related to the frequency of fluctuations [50].

![Bluetooth Antenna Connected to the PCB](image)

Figure 44: Bluetooth Antenna Connected to the PCB
The graphical user interface is designed for ease of operation and clear presentation of data. An off-the-shelf LCD is used for displaying the systolic and diastolic pressure levels. The LCD screen also displays errors like: connection problems, too much noise in the environment, etc. In addition to the pressure information and error signals, the LCD displays the status of the data transfer, wireless communication and remaining power.

The layout of the GUI with the black circular push buttons is shown in Figure 44 and Figure 45. One ON/OFF button is used to control the power of the device. The "select" button starts the measurement while the "undo" one, resets the measurement for starting over. The "select" and "undo" buttons are also used to enable/disable the wireless communication with a local port.

![Figure 45: Layout of the GUI with Buttons](image_url)
Wireless connection with a central port will enable logging monitored data, and if desired send out alerts and notifications to the healthcare personnel and/or patients’ helpers via a software on the central computer. By using a Bluetooth antenna, pressure and pulse information of the patient could be sent to a remote location, or a central position where healthcare practitioners are notified regarding the current situation of the patient.

The use of the air pump is to provide pressure to fill the bladder inside the pressure

This bladder occludes the brachial artery. Once the artery is occluded the pressure fluctuations could be monitored and the blood pressure could be calculated. Most of the commercially available blood pressure monitors use diaphragm-driven air pumps. These pumps are well designed and very suitable for blood pressure monitoring devices that are stationary, i.e. stationed on table. On the other hand, it is not very preferable to use air pumps that would require more than 2 AA batteries for portable blood pressure monitoring devices due to weight related constraints. There are other diaphragm-driven small scale air pumps that could potentially be used for portable devices and mini-pumps that are more suitable for the clip-on control box. The preferred power consumption for the air pump is lower than 1.5W. Air pumps that would work within the 0-0.5A range would make it possible to use only 2 AA or rechargeable batteries. The specifications for such a blood pressure monitoring device are provided in

Table 5.
Table 5: Specifications for SyBeD and Its Components.

<table>
<thead>
<tr>
<th>PRIORITY</th>
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6.3 WORKING PRINCIPLE

SyBeD employs a novel miniature pressure sensor based on DEAPs. DEAP materials exhibit capacitance change when stretched and thus when pressure is exerted on their
surface it can be measured by measuring the capacitance change in the DEAP element. The SyBeD's pressure sensory element is designed as a diaphragm type capacitance pressure transducer as shown in Figure 46. The sensor is designed to be able to withstand pressure values of 0-300mm Hg, to be sufficient for probable highest systolic pressure value. The DEAP film is held in between two polyurethane (PU) layers. These layers are molded using a custom rapid prototype mold that holds the DEAP film and support structure in place for curing. The cured polyurethane is a rubber like plastic that is found in different hardness levels, thus by altering the PU layer thickness the working range for the sensory element could be modified. The specific DEAP material that is used in this work is the Dielectric Electroactive Polymer manufactured by Danfoss PolyPower A/S.

![Capacitive Transducer Element](image)

Figure 46: Capacitive Transducer Element

The core of the pressure sensor element takes advantage of the DEAP properties that allow the fabrication of flexible sensor products with compliant packaging. DEAP is
composed of a silicon film sandwiched between two electrodes that are isolated from each other. The DEAP film can be mechanically deformed and stretched in area. As the polymer in the middle is deformed, the electrodes get closer to each other and the electrical capacitance between the electrodes starts to fluctuate. In Figure 47, red and blue layers represent the electrodes and the white layer is the silicon material. Electrodes hold their compliant characteristic under tension; enabling the transfer of the change in capacitance due to the strain on the polymer material.

![No force applied vs Force applied](image)

*Figure 47: Capacitance on EAP Film Changes with Strain*

The capacitance of a DEAP film element can be calculated using equation 2.3, hence using thin plate theory [51] and assuming that the DEAP film thickness is constant, the change of capacitance in circular DEAP film elements enclosed within another polymer material that is clamped with respect to pressure when a distributed pressure loading is applied on the EAP surface is calculated from the following equation:

$$
\delta C = \varepsilon_0 \varepsilon_r \Pi r \left[ r - \frac{3}{8} \frac{p_0 r^2 (1-\nu)^2}{E \kappa} \right] \quad (6.1)
$$
where $\delta C$ is the change in capacitance due to pressure exerted on the film; $P_0$ is the pressure exerted on the sensor; $r$ is the radius of the active film; $v$ is the Poisson’s ratio for the surrounding polymer; $E$ is the elastic modulus of the polymer package; and $h_e$ is the thickness of the top cover of the sensor. Equation 6.1 shows that when pressure is applied on the DEAP element then a variation of its capacitance occurs. Hence, the DEAP element could be used as a variable capacitor to detect pressure changes when it is part of an analog circuit that serves as a capacitance to frequency converter (RC timer) [9] as shown in Figure 48. Using the circuit of Figure 48 and considering the EAP as a variable capacitor, then the EAP voltage output can be calculated from the equation below [52]:

$$V_c(t) = V(1 - e^{\frac{t}{RC}})$$

(6.2)

Where $t$ is time, $V$ is the input supply voltage, $R$ is the fixed resistor value and $C$ is the variable capacitance value of the DEAP element. For example, the capacitance of a DEAP film of 2cm x 2cm square surface with 100 µm thickness will be in the order of 100-200 pF. From Equation (3), by measuring the DEAP voltage output $V_c$, the capacitance change can be calculated and then from Equation (2) the pressure change.
6.4 FINITE ELEMENT ANALYSIS

Finite Element Analysis (FEA) has been performed using ANSYS Workbench. In this analysis, Polyurethane (PU; i.e. the polymer that is used to coat DEAP film as described in the previous subsection) has been analyzed. In this analysis the DEAP film has not been included as, due to its low elastomeric properties, its deformation will follow PU's displacement. The circular sensor depicted in the model has a diameter of 25mm and it is 2mm thick.

The displacement results are presented in Figure 49, Figure 50 and Figure 51. From the maximum deformation calculated, the deformed DEAP film area can be calculated and then using Equation 6.1 the change in the capacitance of the DEAP film is estimated. An isometric view of the sensory element analysis under pressure is presented in Figure 49.
The FEA analysis results depicted that, when a distributed pressure of 320mm Hg is exerted on the sensor, maximum displacement will occur on the red colored areas and it will not be more than 100µm. In the analysis figures, blue colored zones are the areas with the least displacement and green zones represent displacement of 40µm. The ribs restrain the motion of the sensor element to only one direction (compliant direction). This keeps the DEAP film from bending more than its elastic limit in the stiff direction. The electrode structure on the DEAP film breaks down when the film is stretched from its stiff direction. Under atmospheric pressure, ribs are aligned parallel to stiff direction. When the sensor is pressurized, the ribs bend along the compliant direction (as noted in Figure 51). By bending the ribs, the DEAP film is stretched on its compliant direction resulting in capacitance change.

The analysis result in Figure 50 shows the bending of the PU in its thickness direction. The resulting displacement is 0.04mm that is equal to 2% strain in the thickness mode of the sensor. This strain ratio for thickness mode is within the safe working range of DEAP.
film. DEAP film is allowed to strain 1% in the stiff direction; however the compliant direction is allowed to strain up to 30% without any damage to the electrode structure.

Based on these FEA results, it was concluded that it is effective to have the DEAP film positioned along the center of the sensor circular diaphragm with the compliant direction parallel to horizontal axis on Figure 51. By positioning the DEAP film along the center of the circular PU mold, DEAP film will be stretched on its compliant direction while being constrained from its stiff direction.
6.5 PROTOTYPE BUILDING

Preliminary prototypes were developed and pilot data was obtained to establish the usability of DEAPs as an embedded pressure sensor. The prototype sensory element, shown in Figure 52, was designed as a circular diaphragm type, capacitance pressure transducer. A rectangular DEAP film was suspended in between two circular polyurethane (PU) layers as described in section 6.3. These layers were molded using a custom, rapidly prototyped mold that keeps the DEAP film encapsulated inside the package.
The cured PU was available with different hardness levels, thus by altering the layer thickness the working range for the sensory element could be modified. The encapsulating PU layers were attached to the DEAP film to seal and strengthen the sensory element. Air sealing will prevent any possible air leaks that could affect both measurement and control of air pressure. In addition, the DEAP film itself is not able to withstand the forces generated by the pressurized air. The PU material shields the DEAP film and prevents it from yielding and keeps the film deflection under control due to its higher modulus of elasticity. Sensors with different pressure measurement capabilities can be produced simply by altering the encapsulate design and material elasticity.

6.5.1 MOLD DESIGN

A mold needed to be designed in order to build rapid prototype duplicates for cheap sensor production. The methodology to build the sensory element relies on the mold
design. After finalizing the design for the sensory element, the same CAD file was transferred into a new assembly where the mold tools of SolidWorks were employed. A cavity in the shape of the designed sensory element in a block material was created using the cavity tool in SolidWorks. The cavity tool enables to insert cavities in the shape of a desired part and it is a widely used technique in mold design. The pink part in Figure 53 represents the PU encapsulated DEAP pressure sensor, and the two yellow parts are the pieces that constitute the mold.

![Figure 53: CAD Design of the Mold to Embed the DEAP Film Inside PU Cover](image)

The mold has a reservoir with an inlet and outlet to let the liquid PU enter the chamber with sensor features, an opening for the DEAP film to be attached and an additional side bar to align the top and bottom pieces of the mold and through holes were cut at the edges to screw in both pieces. The sensory element was created by molding as shown in Figure
55. The pressure transducer itself was designed using CAD and then a negative of it was created as the basis for the mold design. The mold was then built using an SLA rapid prototype machine. The resulting mold pieces were later cured and assembled for use, as seen in Figure 54.

![Mold to Embed the DEAP Film Inside PU Cover](image)

**Figure 54: Mold to Embed the DEAP Film Inside PU Cover**

### 6.5.2 SENSOR BUILDING

The mold was then used to construct the sensory element with the embedded DEAP. The DEAP film was aligned to the center of the mold and its position was secured using screws around the mold. The mold was then harnessed so it could be centrifuged to remove all air from it. After the mold had been assembled, PU solution was prepared to be immediately injected inside the mold. Due to the nature of the PU material, there were air bubbles present inside the mold. Air bubbles were removed through a predefined outlet designed into the mold by centrifuging it. If necessary, additional PU solution was
injected after the centrifuge and centrifuge was redone to ensure that the PU material fills the mold. When there were no more air bubbles and the PU was well distributed around the mold, it was set to cure. Curing time for the PU was approximately 16 hours. The other operations necessary to create the prototype took approximately 2 hours so in a total of 18 hours, the prototype was finished. Figure 55 summarizes the processes followed through the prototype development stage. Once the PU was cured, the sensory element was separated and the mold was removed. The sensory element now has the DEAP film embedded inside the PU cover. The sensory element has vertical thin features that act as ribs to constrain the bending of the thin film in only one direction. These constraining ribs can be seen in Figure 55, pictures 1 and 6.

Figure 55: Manufacturing Process of DEAP Based Pressure Sensor Prototype
6.6 MEASUREMENT SETUP

The new sensor was tested for proof of concept and viability of pressure measurement. The test focuses on exerting a known pressure value on to the elastic membrane, deflecting the DEAP film inside and getting capacitance reading as the output from the sensor. To conduct the test, an experimental setup was assembled as shown in Figure 56.

![Experimental Setup to Measure the Capacitance Change](image)

Figure 56: Experimental Setup to Measure the Capacitance Change

The components of the experimental setup consist of a laptop that runs LabVIEW code for signal monitoring and data collection. A National Instruments data acquisition card was used to transfer voltage outputs of the pressure transducer instrument and of a capacitance to voltage converter to the LabVIEW environment. An analog low pass filter with a bandwidth of 19Hz and a third order inverse Chebyshev digital filter were used to filter noise. A Sensotec S2000 series instrument was used to calibrate and monitor an Omega PX26 pressure sensor attached at the test bed. A Boonton BD72 capacitance to voltage converter was used to convert the capacitance value to voltage and amplify the
signal to be acquired by the data acquisition card. In addition to the electronic equipment, a test bed was built by rapid prototyping to test the sensor as shown in Figure 57.

![Test bed to Constrain the DEAP Based Pressure Sensor and to Collect Pressure Information](image)

The pressure inlet tube was rigidly connected to the test bed for the pressurized air entry. The sensor was clamped to the test bed for sealing and for achieving a rigid connection. Inside the test bed, there was a cylindrical chamber that transfers the pressure to the sensory element and to a commercially available pressure transducer that was used to measure the chamber pressure, for future calibration needs of the DEAP pressure sensor. The electrodes were connected to either side of the thin DEAP film and the capacitance readout was transferred to the capacitance to voltage converter. The voltage converter represents the capacitance that is around 2pF as 2V, with mV resolution. The importance of digitization and sampling rate was discussed in section 6.2.2. We have picked 8 samples for fluctuation; hence the sampling rate was 40Hz. The values from both pressure transducers were collected in a graph, displayed in section 6.7.
6.7 RESULTS

The red line represents the actual pressure value inside the chamber in the test bed as measured by the off-the-shelf pressure transducer. The blue line represents the capacitance change on the DEAP based pressure transducer. From Figure 59, we can observe that under the applied pressure, the capacitance of the DEAP sensor changes that suggests that if a proper calibration curve is calculated then it will be possible to calculate the pressure applied on the DEAPs sensor by calculating its capacitance change.
Figure 59: Experimentally Measured Pressure and Capacitance vs Time for DEAP Based Pressure Sensor

Experimental results of the sensory element working inside the pressure cuff were collected in the graph presented in Figure 60. As can be seen from the figure, the change in capacitance follows the change in pressure very closely. The red line represents the actual pressure value inside the chamber in the test bed as measured by the off-the-shelf pressure transducer. The blue line represents the capacitance change on the DEAP based pressure transducer. The direct relationship between the pressure exerted at the artery, and the relative capacitance information was observed from Figure 60. The mathematical model representing the change in capacitance (Equation 2) and the experimental results with the sensor in between cuff and the brachial artery accord each other. These results are very exciting and important demonstrating the validation of our new sensor concept.
6.8 SUMMARY

In this project new, non-invasive, mercury less, automated blood pressure measurement device with wireless communication access, namely SyBeD was proposed. SyBeD is designed to a) reduce the size of current blood pressure measurement devices that conduct measurements from the arm on the brachial artery, b) increase the accuracy of blood pressure measurement through use of embedded pressure sensors at the pressure cuff in an adaptive and robust way, c) to be manufactured with embedded pressure sensors based on DEAPs that allow producing low-cost disposable cuffs. The details of the design, interfacing with wireless communication, working principle of the novel sensor, specification and material costs were presented in the section 6.2.
Preliminary analysis and experimental data results showed that, the novel sensory element works and was able to map pressure values to capacitance and it can provide sensory information when attached to a pressure cuff. The design considerations, process, molding and data acquisition procedures are explained in the sections 6.5 and 6.6.

To summarize, the mission of this project was to complete the development and demonstrate initial clinical use of the new BP measurement device, called SyBeD (Synchronized Blood ressure Measurement Device). Through SyBeD, we prefer to focus on the awareness of blood pressure related diseases and highlight the importance of early diagnosis using home-based physiological measurement systems. With this proposal, we would choose to demonstrate our interest in enhancing primary healthcare capabilities by producing SyBeD and making it cheap and easily accessible for patients in need; while creating new business opportunities and industrial partnerships. For this purpose we have teamed up with notable industrial and clinical experts in device manufacturing and clinical testing. For the support letters from our collaborators please see additional pages.

Northeastern University has filed a provisional patent application on Friday, May 28th, 2010. It is our goal to commercialize this technology using the existing relationship with our industrial partners.
7 CONCLUSIONS

7.1 RESULTS FROM USING DEAP IN VARIOUS APPLICATIONS: CONTROL, HAPTICS AND SENSORS

The use of DEAP based actuator and sensor was investigated with various applications from a biomedical point of view. The very nature of the DEAP material was observed to be very close to human compliance. Hence, allowing it to exert “natural” like forces and making it compatible for applications that involve human interactions. Studies have been conducted to characterize the DEAP based InLastor PUSH actuator to determine its capabilities and usability as a biomedical device. Open loop response has shown that closed loop controller was required for force control. In that respect a nonlinear force controller was empirically built to control the forces between the actuator and the objects it’s interacting with. The PID force controller with nonlinear feed-forward term showed better performance for dynamic test results; hence it was favored over the PID closed loop controller without the feed-forward term and other tested controllers.

After studying the usability of the actuator as a biomedical device, its use in haptic applications were tested through comparison with an off the shelf, electromagnetic motor of similar size and human subject tests were conducted to determine the haptic perception capability of both actuators by users. By implementing the inverse model developed through the force control studies, haptic capability of DEAP actuator was tested in open loop control with human participants. Results have shown that DEAP actuators are as good as electromagnetic motors to convey the information through sense of touch. Since the haptic communication for commercial electronics and personal devices would be taking human at its center, and the way people would feel comfortable is variable.
DEAPs results as having lower standard deviation for subjective tests compared to LVCM points out the general usability and acceptance of DEAP actuator. In the light of these findings a haptic application of DEAP actuator was demonstrated as a haptic heart rate feedback emulator. It was shown that the forces, bandwidth and inherent damping of the DEAP actuator are very suitable for emulating physiological signals as force feedback. The actuator was controlled from a remote location by the observed signals of a biker’s heartbeat and it was shown that DEAP actuators can be employed as telemonitoring devices.

The excellent capabilities of DEAP to emulate physiological signals led researching, if it will also be able to capture human physiological signals as a wearable sensor. The use of DEAP as a sensor was already described in the literature; however there weren’t many practical examples of DEAP based sensors. DEAP based pressure sensor is a novel pressure sensor that could be used as a wearable sensory device to measure the physiological signals that could be related to sensor strain. It is sealed and encapsulated with polyurethane which makes it a completely polymer based sensor. The preliminary analysis and test results showed that the change in capacitance on the thin DEAP film can be related to strain and various sensors could be developed with this technology. The main sensory equipments would be: displacement, force and pressure sensors based on DEAP film. It was noted that due to its lightweight and flexibility and its ease of design, DEAP materials are very suitable for wearable physiological measurement devices for humans.

One of the significant elements of robotic devices in contact with humans, force control, was investigated and a nonlinear force control was found to be most suitable for DEAP
actuator’s use for biomedical devices. The open loop controller found from force control studies was applied in haptic comparison tests, enabling the haptic evaluation of the device human subject tests. The evaluation tests have shown that DEAP actuator was able to convey physiological signal information haptically and it is also suitable for teleoperation applications. Finally the sensory use of DEAP material was presented as a pressure sensor prototype. The design and manufacturing of cheap, flexible pressure sensors were described and it was shown that pressure change could be related to the capacitance change due to strain of DEAP film encapsulated inside a polymer housing. The studies showed that DEAP based actuators and sensors are very suitable as force feedback or wearable sensory devices in contact with human body. In particular, physiological sensors and feedback devices can take advantage of DEAP material.

Throughout this thesis work, the first experimental force controller for DEAP actuator was represented, the very unique example of encapsulated DEAP pressure sensor was developed, and a provisional patent application was filed for the sensor. The DEAP based actuators and sensors are very promising for enabling the design flexibility of these materials. Customized actuators and sensors combining rapid prototyping with DEAP’s advantages will be sought as a future path.
BIBLIOGRAPHY


