Haptic Interfaces Using Dielectric Electroactive Polymers

Muzaffer Y. Ozsecen, Mark Sivak and Constantinos Mavroidis[1]

Department of Mechanical and Industrial Engineering, Northeastern University
360 Huntington Avenue, Boston MA 02115

ABSTRACT

Quality, amplitude and frequency of the interaction forces between a human and an actuator are essential traits for haptic applications. A variety of Electro-Active Polymer (EAP) based actuators can provide these characteristics simultaneously with quiet operation, low weight, high power density and fast response. This paper demonstrates a rolled Dielectric Elastomer Actuator (DEA) being used as a telepresence device in a heart beat measurement application. In the this testing, heart signals were acquired from a remote location using a wireless heart rate sensor, sent through a network and DEA was used to haptically reproduce the heart beats at the medical expert's location. A series of preliminary human subject tests were conducted that demonstrated that a) DE based haptic feeling can be used in heart beat measurement tests and b) through subjective testing the stiffness and actuator properties of the EAP can be tuned for a variety of applications.

Keywords: dielectric elastomer, electroactive polymer, haptic interface, human subject test

1. INTRODUCTION

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 [1]. DEAs are made using EAP films, they operate silently, they are soft and their compliance is close to that of the human body. EAPs 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.

There are several examples of DEAs being used in biomedical applications. DEAs have been proposed as implantable diaphragm muscles [1]. In addition to body conformity, their efficiency enable the use of small size batteries hence allowing their use in remote operations [1]. Larger sizes of EAP films and actuators could be used for large areas of the human body. The EAP 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 [2]. 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 [3]. This orthosis works silently and applies continuous loading on the finger for faster recovery. 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. MRI compatibility of the actuator itself has already been documented [4] 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 / or 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.

Dielectric EAP's fall in the category of electric EAP's and are also referred as Dielectric Elastomers. Response of the electroactive film is the result of an electrostatic charge on the electrodes. When charged with opposite polarity, the electrodes attract each other creating a pressure (ρ) on the film. The compression on the film decreases the thickness and the area increases to keep the volume constant. The generated pressure results in a linear motion in a rolled type DEA

[1] Professor-Corresponding Author, Tel: 617-373-4121, Fax: 617-373-2921, Email: mavro@coe.neu.edu, Webpage: http://www.coe.neu.edu/~mavro

Sensors and Smart Structures Technologies for Civil, Mechanical, and Aerospace Systems 2010, edited by Masayoshi Tomizuka, Chung-Bang Yun, Victor Giurgiutiu, Jerome P. Lynch, Proc. of SPIE Vol. 7647, 764737 · © 2010 SPIE · CCC code: 0277-786X/10/$18 · doi: 10.1117/12.847244

since it constraints the film on two sides. When the voltage is switched off, the DEA contracts back to its original shape. The following electromechanical model [6,7] describes the pressure exerted on the silicon layer ρ as:

$$\rho = \epsilon_r \epsilon_o E^2 = \epsilon_r \epsilon_o \left(\frac{V}{t}\right)^2 \tag{1}$$

In addition to the pressure generation on the film, an EAP film changes capacitance with strain. Since the electrodes are separated from each other with a non-conducting polymer, DE is also considered as a variable capacitance device. The capacitance C is represented using the following model:

$$C = \epsilon_r \epsilon_o \frac{A}{t} \tag{2}$$

Where ϵ_r and ϵ_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. The film thickness is in the order of micrometers and the area depends on the application and it is in the order of meters. The pressure on the film creates the force along the direction of elongation and it is directly proportional to the square of the voltage. As can be seen from Equation (1), a nonlinear response of the force is expected with varying voltage values. However, the change in the capacitance is directly related to the change in the film thickness and the area since the dielectric constants are fixed.

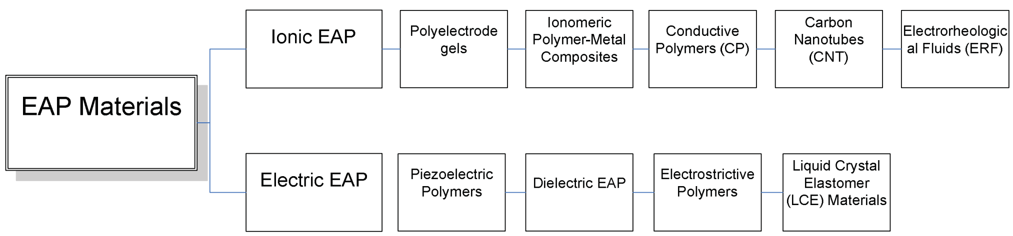

Figure 1: Classification of EAP Materials

The goal of this paper is to investigate the possible use of DEAs as a building block to provide haptic interface solutions to the medical field. 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).

2. EXPERIMENTAL SYSTEMS FOR ACTUATOR EVALUATION

Two experimental testing systems were constructed to evaluate the DEA performance and capabilities and compare it to a LVCM (see Figure 2). The Rolled DEA was manufactured by the company Danfoss-PolyPower (http://www.polypower.com) via rolling sheets of EAP film around an axis to form a cylindrical shape. By constraining the rolled films with passive polymer parts from both ends, planar elongation is transformed into linear motion in the direction of the roll axis (z axis). The DEA extends resulting in a linear motion when actuated with high voltage, and contracts back to its original shape when the voltage is returned to zero. Compared to electromagnetic motors and linear actuators, EAP's have higher power density. They are lightweight, have large displacements and better efficiency at lower speed operation [8-10]. Linear Voice Coil Motors (LVCM) are electric motors that consist of a 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 2: Linear Dielectric Elastomer Actuator (LEFT), Linear Voice Coil Motor (RIGHT)

Table 1 presents some important performance parameters of the DEA being used in these experiments and of the LVCM. The specific rolled DEA has lower strain levels compared to the LVCM. DEAs are slower compared to LVCM due to their inherent impedance effects. The maximum force that could be applied with respect to actuator weight 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 an MRI environment.

Table 1: Actuator Properties for Evaluation

Property	DEA	Voice Coil
Maximum stroke	1.2mm	57.2mm
Maximum force-to-weight	58.26 (N / kg)	61.26 (N / kg)
Maximum power-to-weight	7 (W / kg)	29.6 (W / kg)
MRI Compatibility	Good	Poor

A) Dielectric Elastomer Actuator

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 3-LEFT. The actuator motion and interaction forces were regulated through LabView by using an inverse model that maps the required voltage value to the maximum force generated by the actuator at the interaction point [11]. The system was operated under open loop control. A low voltage command signal was amplified by a high voltage amplifier (TREK 609D-6) to run the actuator. Data acquisition was done using a National Instruments USB 6216 card. A conventional laptop with USB port was used to run the LabView code and data acquisition.

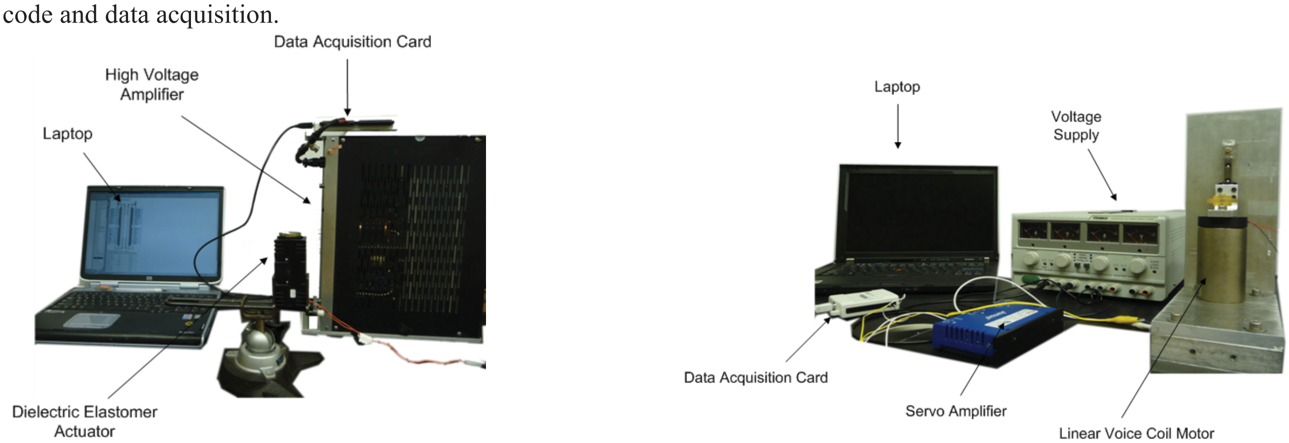

Figure 3: Dielectric Elastomer Actuator Setup (LEFT) and Linear Voice Coil Motor Setup (RIGHT)

B) Linear Voice Coil Motor

The second setup consisted of a LVCM (Moticont Linear Voice Coil Motor, LVCM-051-089-01) rigidly attached to an aluminum test base (see Figure 3-RIGHT). 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 actual current of 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. The desired current was sent to the controller through an analog input with the analog voltage supplied from a National Instruments (NI) USB 6009 terminal. The DC voltage to run the servo amplifier was supplied by a TENMA 72-2080 DC power supply.

3. NETWORK HEART RATE HAPTICS EMULATION

To demonstrate that it is possible in using a DEA as a component in a haptic system, we investigated the possibility in remotely monitoring and haptically measuring the heart rate of a subject as shown schematically in Figure 4. The heart rate of a subject was collected as he pedaled a stationary bicycle via a chest band (Polar) (as shown in Figure 5). It was then wirelessly sent to a signal receiver attached to a National Instruments DAQ card, and then it was 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 subject's heart beat and remained false in between beats.

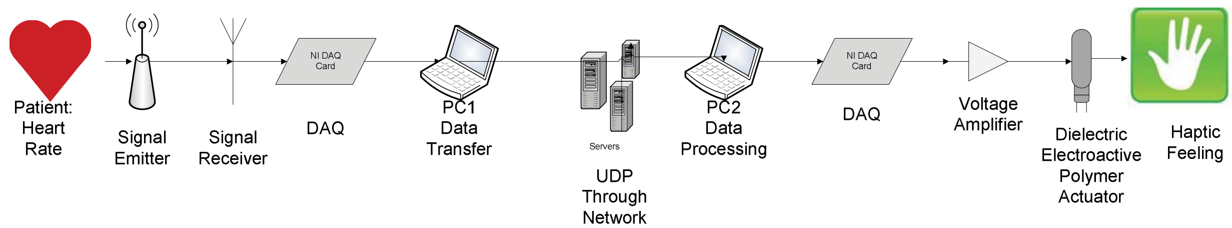

Figure 4: Haptic Heart Rate Emulation Diagram

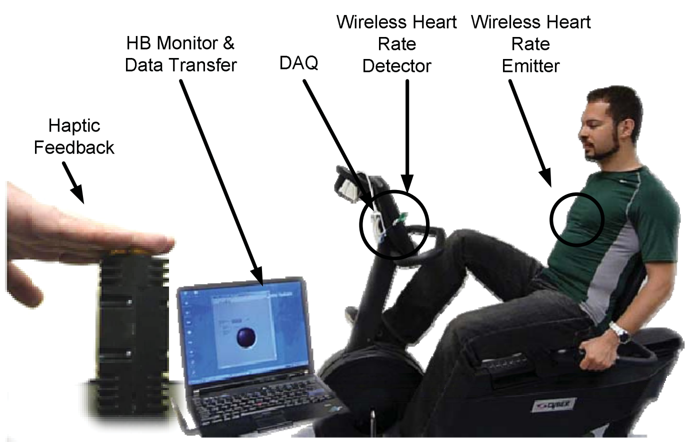

Figure 5: Human Subject Biking on One Side of the Network for Data Collection (RIGHT); Haptic Feeling on the Other Side of the Network Using a DEA

From PC1 the data traveled over the local network to PC2 via UDP. The data was sent as a string to preserve the digital nature 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 haptically mapped correctly by the DEA. A second subject in the area of the PC2 (i.e. this second subject could be a medical expert in a remote location) could then “feel” the heart beat of the first subject using the haptic feedback of the DEA. As well as transferring heart rate information, with data acquisition tools developed in the LabView environment, it was possible to graph the continuous heart rate information and detect the rate of change of the heart beat in time. Figure 6 shows the digital signal that was acquired by the heart rate sensor and then transferred as input to activate the DEA. 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.

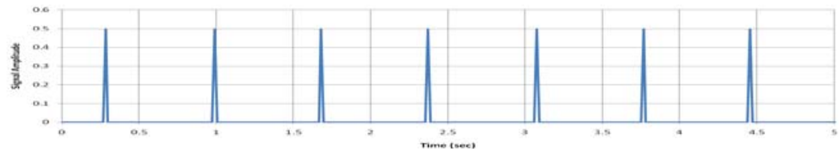

Figure 6: Heart Rate Data as Pulse Signal

4. HUMAN SUBJECT TESTS

A series of human subject tests was conducted to compare the human haptic perception when DEA and LVCM are used in the simple force-feedback setups described in Section 2. The set of experiments involved the usability of the actuators as haptic interfaces and considerations regarding how comfortable the actuator felt to the user. Tests consisted of two main parts: a) providing objective and b) subjective evaluation. In the objective evaluation part there are two tasks for the subjects to perform that would allow quantification of the user response on the comfort and quality of force 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 is 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.

A) Protocol

In the objective testing experiments, types of signals used included: 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 signals with duration of 15 seconds were generated using LabView with a predetermined number of impulses (ticks). Ticks refer to the peaks of the signal introduced during the task. The variable for comparing results is the error that they make while guessing the tick count. The error is calculated as the difference between the predetermined value and the guess of the subject. The predetermined value was selected as an odd number (15, 17 and 19) to eliminate the probability of the 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 [12].

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 is 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 a 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 (Figure 7). This test constitutes the subjective part of the experiments as the result that was captured by this data is based on human preference, and it is dependent on how subjects perceive the actuators.

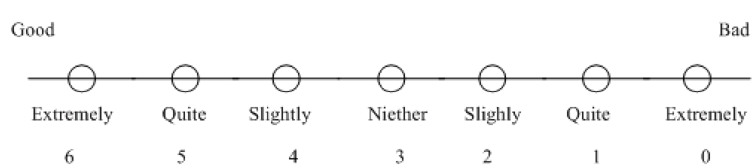

Figure 7: Scaling used for objective tests

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 asked the subjects to place their hand on top of the Dielectric Elastomer and the Linear Voice Coil actuators for the same period of time under the same conditions. The test subjects were selected from Northeastern University's student population. 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 didn't have any effect for this system under study, therefore left-handed and right-handed participants were subject to the same test procedure. Volunteers were asked to visualize the three signals mentioned above. Experiments were constructed not to exceed 15 minutes.

B) Results

Figure 8 shows 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 in counting how many times the actuator ticks.

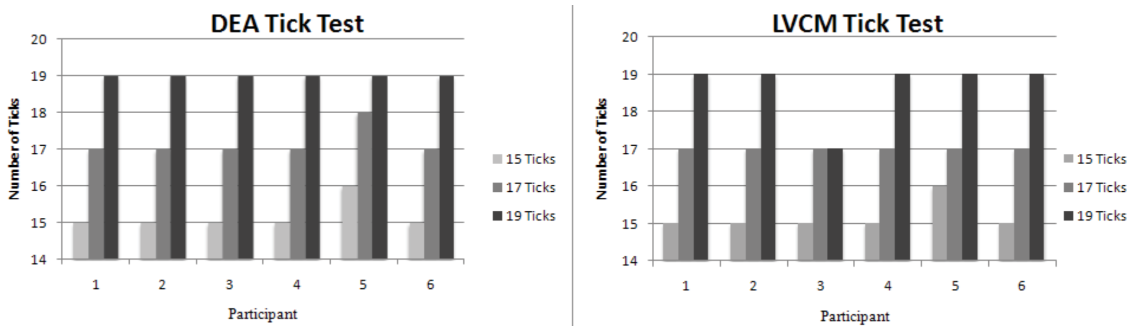

Figure 8: Tick Count Test results for DEA and LVCM

Table 2.Wave Estimation and Subjective Test Results

Participant	Dielectric Elastomer Actuator						Linear Voice Coil Motor					
	Wave Estimation			Questions			Wave Estimation			Questions		
	Test 1	Test 2	Test 3				Test 1	Test 2	Test 3			
	sine	square	saw tooth	comfort	smoothness	overall	sine	square	saw tooth	comfort	smoothness	overall
1	TRUE	TRUE	FALSE	4	4	3	FALSE	FALSE	FALSE	3	3	4
2	TRUE	TRUE	TRUE	5	4	4	TRUE	TRUE	TRUE	5	5	5
3	TRUE	TRUE	TRUE	3	5	5	TRUE	TRUE	TRUE	2	1	2
4	FALSE	FALSE	TRUE	4	2	5	TRUE	FALSE	FALSE	5	5	5
5	TRUE	TRUE	TRUE	5	5	5	TRUE	TRUE	TRUE	6	6	6
6	TRUE	TRUE	TRUE	3	2	3	TRUE	TRUE	TRUE	4	6	5

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. The left side of Table 2 presents the wave estimation test results. The total number of false estimations was 3 for DEA and 5 for LVCM. Only two of the participants (1 & 4) both made estimation mistakes. The subjective test results are depicted on the right side of the table for both actuators. 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 more in both the tick count and the wave estimation tests.

C) 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 suggest that participants felt more comfortable using the LVCM and forces felt smoother. This shows that, participants perceived the applied forces incorrectly while feeling comfortable. The participants' subjective response on the DEA has smaller standard deviation compared to LVCM, and this shows 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 are in the order of 1-2%, which implies the necessity to increase the sample size. One of the points that needs 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. CONCLUSIONS

Dielectric Elastomer based systems have certain characteristics that could be tailored towards specific use in a medical setting and haptic interfaces in particular. The possible use of DEAs as haptic heart rate emulators for patient monitoring was investigated in this paper. In order to evaluate how good a DEA can perform in haptic interactions a comparison with a LVCM was performed using preliminary human subject tests. These tests showed that the DEA performed at least as good as the LVCM in simple haptic feedback tasks. Subjective answers could be interpreted as participants had slightly different perception and preferences over the feeling of touch.

ACKNOWLEDGEMENTS

The authors would like to thank Danfoss PolyPower A/S of Denmark (http://www.polypower.com/) for providing the DE actuator samples and for their technical support regarding the Dielectric Electroactive Polymer Technology.

REFERENCES

[1] Bashkin, J.S., Kornbluh, R., Prahlad, H. and Wong-Foy, A., “Chapter 21: Biomedical Applications of Dielectric Elastomer Actuators”, [Biomedical Applications of Electroactive Polymer Actuators], Carpi, F. and Smela, E., Editors, Wiley-Blackwell, Wiltshire 395-410 (2009).

[2] Choi, R.H., Koo, I.M., Jung, K., Roh, S., Koo, J.C., Nam, J. and Lee, Y.K., “Chapter 23: A Braille Display System for the Visually Disabled Using a Polymer Based Soft Actuator”, [Biomedical Applications of Electroactive Polymer Actuators], [Biomedical Applications of Electroactive Polymer Actuators], Carpi, F. and Smela, E., Editors, Wiley-Blackwell, Wiltshire, 427-442 (2009).

[3] Carpi, F., Mannini, A. and De Rossi, D., “Chapter 24: Dynamic Splint-Like Hand Orthosis for Finger Rehabilitation”, [Biomedical Applications of Electroactive Polymer Actuators], Carpi, F. and Smela, E., Editors, Wiley-Blackwell, Wiltshire, 443-460 (2009).

[4] Carpi, F., Khanicheh, A., Mavroidis, C., De Rossi, D., "Silicone Made Contractile Dielectric Elastomer Actuators Inside 3-Tesla MRI Environment", IEEE / ASME Transactions on Mechatronics, 13(3), 137-142 (2008).

[5] Plante, J.S., Devita, L., Tadakuma, K. and Dubowsky, S., ”Chapter 22: MRI Compatible Device for Robotic Assisted Interventions to Prostate Cancer”, [Biomedical Applications of Electroactive Polymer Actuators], Carpi, F. and Smela, E., Editors, Wiley-Blackwell, Wiltshire, 411-425 (2009).

[6] Carpi, F. and De Rossi, D., “Dielectric Elastomer Cylindrical Actuators: Electromechanical Modelling and Experimental Evaluation”, Materials Science & Engineering C*,* 24(4), 555-562 (2004).

[7] Bar-Cohen, Y., [Electroactive Polymer (EAP) Actuators as Artificial Muscles: Reality, Potential, and Challenges], The Society of Photo-Optical Instrumentation Engineers, Bellingham, 529-577 (2004).

[8] Pelrine, R., Kornbluh, R. and Joseph, J., “Electrostriction of Polymer Dielectrics With Compliant Electrodes as a Means of Actuation”, Sensors & Actuators: A. Physical*,* 64(1), 77-85 (1998).

[9] Wissler, M., Mazza, E. and Kovacs, G., "Circular Pre-strained Dielectric Elastomer Actuator: Modeling, Simulation and Experimental Verification", Proc. SPIE 5759, 182 (2005).

[10] Sarban, R., Oubaek, J. and Jones, R., "Closed-Loop Control of a Core Free Rolled EAP Actuator", Proc. SPIE 7287, 72870G (2009).

[11] Ozsecen, M. and Mavroidis, C., "Nonlinear Force Control of Dielectric Electroactive Polymer Actuators" Proc. SPIE 7642, 81 (2010).

[12] A. Fisch, "Development and Testing of Haptic Interfaces Using Electro-Rheological Fluids", Rutgers University, Ph.D. thesis, (2007).