Pressure Sensor Development based on Dielectric Electro Active Polymers

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***Abstract*—The Dielectric Electro Active Polymer's (DEAP's) sensing capabilities is one of the main trio-characteristics of the material applicable area's, the trio-formations as applicable use are actuator, transducer and last but not least sensor. It is noted here that one of the main value propositions whenever DEAP material is used is the dual characteristics as the sensing/actuating capability.**

In the following work, the DEAP membrane will be modeled and the relation between the key variables (pressure & capacitance) will be determined. Hence, such a relation depends on the geometrical shape of the used membrane. So on, conceptualization is carried out to propose alternative solution for the sensor design using the DEAP and the laminate material.

In general, the DEAP material has proven to be a very good sensor for pressure taking the advantage of flexibility, wide range of operation and last but not least the sensitivity. The theoretical model is benchmarked against the acquired data from the tests, good correlation has been recorded. The desired requirements for accuracy and measuring intervals are satisfied, showing a promising potentials for the DEAP material in pressure sensing in general, and pressure sensing application in specific.

Keywords*: *Conceptualization, Dielectric Electro Active Polymers, membrane, modeling, pressure vs. capacitance, sensor, smart material

I. The Smart Material Technology Potentials

Dielectric Electro-Activated Polymers (DEAP) are silicone material with compliant electrodes, thus can be presented as being capacitive elements whose values changes as the elements geometry is modified. The 'soft' properties of the elastomer material make it's deformation as a function applied forces or pressure feasible, hence, providing a potential ground for developing electro-mechanical force or pressure transducer.

The scope of the following work is to evaluate the DEAP performance as a transducer to sense and measure the change in pressure, more specifically the pressure range defining the human blood pressure sensor's partial range. It is a fact that Danfoss PolyPower A/S has done the basic work to use the DEAP material as a stimulus to actuate blood flow restriction. Measuring pressure based on the capacitive principle is well known as well using the piezo-devices [1], there should be good opportunity to leverage existing designs and components with respect to measuring circuits using DEAP's.

In order to reach the desired objectives, the following studies are performed:

- Characterizing the pressure sensing capability of DEAP
- Developing the capacitive measuring circuits
- Establish Control loops with adaptive feedback necessary to implement the measurement methodology
- Generate software to display the results (Not live)

As discussed already, the goal of this project is to research the feasibility of using the DEAP material for pressure sensing, thus exploring the potential of state-of-art application potential as the blood pressure measuring device range.

The motive behind exploring the Blood Pressure measuring device over other dependent pressure applications is the fact that the the primary BP measuring techniques suffers from the following problems:

- Auscultatory technique cannot be used in noisy environment
- The observations differ from observer to another
- A mechanical error might be introduced into the system e.g. mercury leakage, air leakage, obstruction in the cuff etc.
- The technique does not give accurate results for infants and hypotensive patients

As a area of research encompassing diverse aspects, a comprehensive list of specifications will be addressed in the next section, the requirements are derived from similar measurement systems which will not be addressed in this work as the main objective is to validate a feasibility of concept.

II. Dielectric Electro Active Polymer's

Electro Active Polymers (EAP) are polymers that are able to change size and/or shape when submitted to electrical stimulation. There are two types: Ionic and Electronic. The ionic types work by using ion exchange between two electrodes when a DC-voltage source is applied: The ions are transported from the positive electrode to the negative in the solution between them, causing a swelling in one side and a shrinking in the other [2].

Respectively, the pros and cons are large bending displacements with the application of low voltages, but the response is very slow and with a low actuation force. The electronic types of EAP utilize two forces:

1. The Maxwell forces between two electrodes where a HVDC (High Voltage Direct Current) source is applied, cre-

978-1-4577-2119-9/12/$26.00

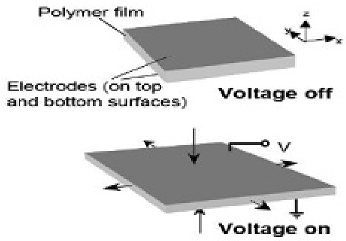

Fig. 1. Applying voltage to the electrodes makes the electronic EAP's change its shape

ating opposite electrical charges that attract each other.This attraction between the charged electrodes squeezes the soft but non-compressible polymer in the middle, causing it to expand in the directions orthogonal to the applied pressure as shown in Figure 1.

2. The electrostrictive forces are created when randomly-aligned electrical polarized molecules or dipoles within the dielectric material (Figure 2A) are subjected to an electric field: The opposite sides of the domains become differently charged and attract each other (Figure 2B), reducing material thickness in the direction of the applied field from within the material [3], like the Maxwell forces do from the surface [4].

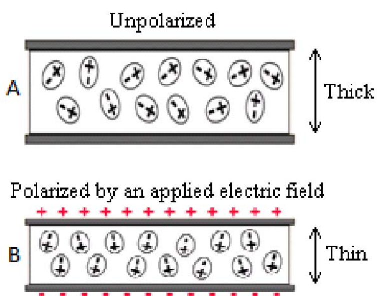

Fig. 2. With an electric field alligning the dipoles in the dielectricum,the electrostrictive forces pulls the material together

The electronic EAP element can be considered as a variable capacitor, which gives it a major advantage in being able to hold the actuator strain when the electrical charge stays on the electrodes. It can also function as a sensor where applied mechanical work changes its capacitance,but the disadvantage is the high voltage required to strain it [5] [6] [7].

Comparatively EAPs can be seen to have many advantages for actuation and a large range of dynamic response: Low power consumption along with the ability to hold strain when not activated, high energy density [8], large flexibility at low forces are extremely desirable properties for its potential use in many core electromechanical applications [9] [10], and a large potential also exists for using EAP for creating new technologies and new market areas [11], one of these areas is energy harvesting as feasibility study shown that the DEAP is a candidate with many value propositions over other smart materials [12].

III. The pressure sensor specifications

In order to design the sensor, some specification has been drawn as a work frame to be benchmarked and used for drawing conclusions:

Material: DEAP and laminate from Danfoss PolyPower A/S

Design: Array of sensors to cover the supporting bone of the artery. Each sensor size should be small relative to the artery

Operation: capacitance change as a function of applied pressure

Accuracy: Better than 2 mm Hg must be achieved with 75% of the measurements

Dimensions: 10x10x10 cubic mm for one sensor, and an array of 5 sensors, covering a cubic area of 50x50x10 of mm^2.

Capacitive measurement circuit: 12 digits resolution of measurements

Reading Interval: It is the pressure and time reading interval, the pressure reading interval of 30-180 mm Hg with more than 100 measurements per second on individual sensor.

Measurement Stability: same mean value in more than 75% of the measurements

Calibration: Digital calibration though software

Software Specifications: Software will be developed using LabView to display the results of the measurement including systolic and diastolic BP values

IV. The pressure sensor conceptualization

Through out the progress of the project, there have been designed three different types of pressure sensors as test pieces, where each new version tends to solve problems seen in the previous one. In this section, the three different models will be presented, thus explaining its structures and reasons for failure, and how did these reasons contribute in the new design/model of the new pressure sensor. Note that all pressure sensors used here are made of electroactive composite of two layers back-to-back laminated and the connection to the electrodes has been achieved through an electrically conductive tape supplied from 3M [13].

A. Concept I : DEAP membrane

As shown in Figure (3), the first model of the pressure sensor is a circular electroactive composite of 2 layers back-to-back laminated whereby a conductive tape is connected to both the bottom and the top electrode providing better area connection. The wires are connected to the conductive tape from one side and to the capacitometer on the other side.

While testing this sensor, the read value of capacitance on the capacitometer went totally unstable, indicating that there is a problem in the contact/connection to the electrodes. The reason behind loosing contact to the electrode is that the elongation in the stiff direction was outside the expected, leading us to a break in the coated silver electrodes. A new design of pressure sensor had to be developed to try to keep in contact to the whole active area of the electroactive composite/sensor.

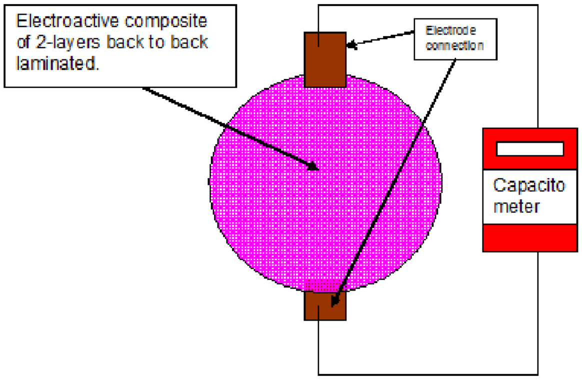

Fig. 3. Test piece structure of pressure sensor: Model 1/3

B. Concept II : DEAP membrane + strip

As shown in figure (4), the 2nd model of the pressure sensor is almost the same as the first model, with one extra part made of a composite layer. The reason for placing this part is to give stronger contact to the electrodes trying to solve the problem with the first model. When the electrode break across the stiff side it will break longitudinally (across the extra part) so this part will bring back the connection to the electrodes.

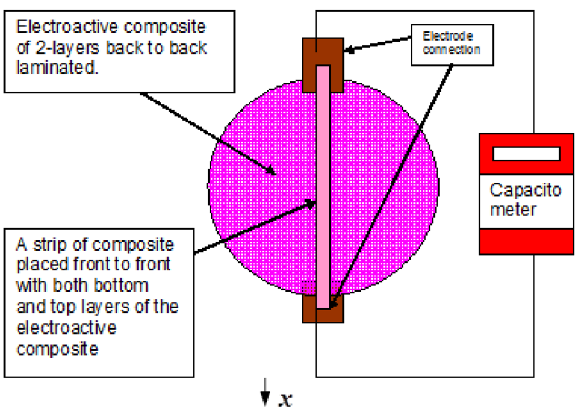

Fig. 4. Test piece structure of pressure sensor: Model 2/3

While testing this sensor as shown in figure (5), a stable value has been read on the capacitometer, but when increasing the pressure level, the capacitance has been decreasing. This indicated that the electrodes were still fracturing; resulting in the reduction is area between electrodes reducing the observed capacitance. As a result, the extra part added to the pressure sensor have solved a part of the problem stabilizing the reading on the capacitometer, but it did not solve the problem of the electrode being broken, and capacitance measurements does not express the capacitance of the whole area between the 2 electrodes of the sensor. Again a new design for a new test piece model of pressure sensor had to take place.

C. Concept III : Polymer membrane + DEAP strip

As shown in figure (6), the 3rd model of the pressure sensor is totally different from the first 2 models, since there

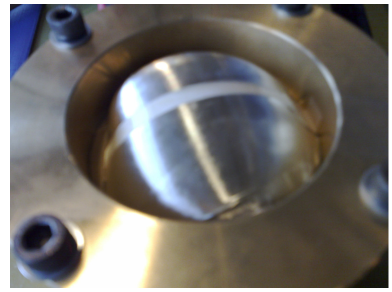

Fig. 5. Deformation of the test piece (Model 2/3) structure under pressure effect

have been introduced a new idea. A circular cured elastomer membrane of thickness 1 mm has been used a carrier to carry the pressure sensor which is made in a rectangular shape. The pressure sensor is placed on top of the membrane, and its role is now a follower which will follow any movement or any shape which the membrane could take due to pressure.

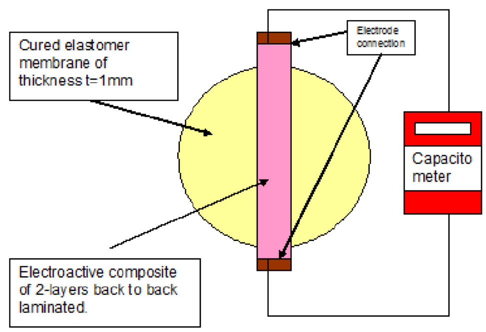

Fig. 6. Test piece structure of pressure sensor: Model 3/3

The pressure sensor is having the stiff direction on its width, which under deflection can only undergo negligible change in the width and only the length will be deformed ensuring that the electrode across the stiff direction of the composite will keep unbroken. The thickness of the pressure sensor is so much smaller then the thickness of the membrane and the width of the sensor is unconstrained, in thus a structure it is assumed that the pressure sensor will track the deflection of the membrane carrying it. While testing this mode, the values read on the capacitometer were expected in coordinate with the theoretical model which will be derived later.

V. Capacitance vs. Pressure theoretical model

The theoretically derived equations in this section are based on the following restrictive assumptions:

- The membrane is flat and of uniform thickness
- The composite is isotropic and homogeneous
- The maximum deflection due to applied pressure should be small-not more than 30% of the thickness of the plate
- All forces, loads, and reactions are applied normally to the plane of the plate
- The plate is not stressed beyond the elastic limit

- The plate should not be too thick. For a round diaphragm it should not be more than 20% of the diameter
- The plate deflection is due mostly to bending; therefore, the median plane of the plate endures no tensile forces
- The pressure sensor thickness is much smaller than the thickness of the plate, and the width of it is unconstrained, thus the pressure sensor is assumed to track the deflection of the plate

Based on the following assumptions, the deflection y of a diaphragm with fixed edges loaded by pressure P at any radial distance r can be written as:

$$y = \frac{3(1-\mu^2)P}{16Eh^3}(a^2 - r^2)^2 \tag{1}$$

Where the maximum deflection y_0 occurs at the center where r=0:

$$y_0 = \frac{3(1-\mu^2)Pa^4}{16Eh^3} \tag{2}$$

For a very thin diaphragm as the one used, especially one operating in the range of large deflection (y_0/h > 5) can be considered a membrane. A membrane in the flat position cannot support a load, its load-supporting capacity developing only with deflection. Theoretically, a membrane has no flexural rigidity and hence no bending stresses. As a matter of fact, the characteristic equations for a membrane may be obtained from the equations for a flat diaphragm with large deflections by assuming the flexural rigidity D is:

$$D = \frac{Eh^3}{12(1-\mu^2)} \tag{3}$$

The shape of the elastic surface of a membrane under pressure is almost spherical in shape.The characteristic equation of a membrane as developed by Andreeva (1946) is:

$$\frac{Pa^4}{Eh^4} = \frac{7-\mu}{3(1-\mu)}\frac{y_0^3}{h^3} \tag{4}$$

where μ = 0.5 assuming incompressible material

The general characteristic equation for a flat diaphragm taking into account both the bending and tensile loads may be obtained by the method of superposition. The equation thus obtained will give satisfactory results for diaphragms at any deflection. As already shown, the characteristic equation for a flat diaphragm for small displacements as expressed.

$$\frac{Pa^4}{Eh^4} = \frac{16}{3(1-\mu^2)}\frac{y_0}{h} \tag{5}$$

The strain vector for the electroactive composite can be defined as:

$$e = s\sigma \tag{6}$$

For a DEAP sheet consisting of elastomer film with compliant electrodes on both sides which will be used throughout this work for modeling the pressure vs. capacitance relation. The deformation of the model results in a new length, width and thickness l, w, h respectively,The ratio of deformed and actual dimensions are defined as and shown in Figure 7 :

$$\alpha_l = \frac{l_0}{l} \tag{7}$$

$$\alpha_t = \frac{w}{w_0} \tag{8}$$

$$\alpha_h = \frac{h}{h_0} \tag{9}$$

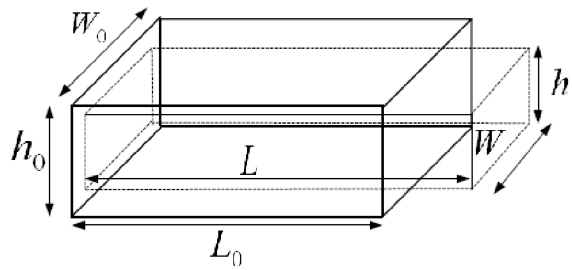

Fig. 7. Deformation of the EAP material

It is assumed incompressible with a Poisson's's ratio of 0.5 (Poisson's ratio defines by the ratio between the transverse strains divided by the axial strain), therefore the conservation of volume states:

$$V = V_0 \tag{10}$$

the following can be rewritten as:

$$l.w.h = l_0.w_0.h_0 \tag{11}$$

Assuming the width remains constant, since the changes in width (stiff direction) are much smaller compared to change in length (compliant direction):

$$\alpha_h = \alpha_l = \alpha = \frac{h}{h_0} = \frac{w_0.l_0}{w.l} = \frac{l_0}{l} \tag{12}$$

For the capacitance of the sensor the following in used:

$$C = \epsilon_0\epsilon_r\frac{A}{h} = \epsilon_0\epsilon_r\frac{lw}{h} = \frac{1}{\alpha^2}\epsilon_0\epsilon_r\frac{l_0w_0}{h_0} = \frac{1}{\alpha^2}C_0 \tag{13}$$

Assuming that the applied pressure will deflect the membrane in semi-elliptical shape (close to), So on, the circumference equation of the ellipse to calculate for the new length, knowing that:

$$Circumference \approx \pi[3(l_0 + y_0) - \sqrt{(\frac{3l_0}{2+y_0})(\frac{l_0}{2+3y_0})}] \tag{14}$$

Divide to circumference by 2 to get the actual length:

$$l = \frac{\pi}{2}[3(l_0 + y_0) - \sqrt{(\frac{3l_0}{2+y_0})(\frac{l_0}{2+3y_0})}] \tag{15}$$

substituting equation (2) into equation (15) to get:

$$l = \frac{\pi}{2}[3\left(l_0 + \frac{3(1-\mu^2)Pa^4}{16Eh^3}\right) - \sqrt{\left(3\frac{l_0}{2} + \frac{3(1-\mu^2)Pa^4}{16Eh^3}\right)\left(\frac{l_0}{2} + \frac{9(1-\mu^2)Pa^4}{16Eh^3}\right)}] \quad (16)$$

Substituting equation (12) into equation (13) to get the expression for the capacitance:

$$C = \frac{l^2}{l_0^2}C_0 \quad (17)$$

The initial capacitance C_0 can be calculated from the basic capacitance equation for a parallel plate capacitor using the following formula:

$$C_0 = \epsilon_0\epsilon_r\frac{l_0 w_0}{t_0} \quad (18)$$

Substituting equation (16) into equation (18) and solving for C to the final form for the capacitance vs. pressure model as shown in equation (19):

$$C = \frac{\pi^2}{4}\frac{1}{l_0^2}C_0[3\left(l_0 + \frac{3(1-\mu^2)Pa^4}{16Eh^3}\right) - \sqrt{\left(3\frac{l_0}{2} + \frac{3(1-\mu^2)Pa^4}{16Eh^3}\right)\left(\frac{l_0}{2} + \frac{9(1-\mu^2)Pa^4}{16Eh^3}\right)}]^2 \quad (19)$$

VI. Model validation

Figure (8 shows a comparison made between the theoretical calculations based on the developed theoretical model and the experimental results achieved through the test run. An error plot is added in red showing C_0 is off by 1.7%, the rest of the error is in capacitance measurement accuracy (0.5% + 10 pF, approximately 30 pF i.e. 7%). Some of the error is due to the pressure measurement accuracy. As a conclusion from the graph, the model predicts the behaviour well within the measurement system accuracy for the given pressure range (0-3000 Pa).

A foundation from this work shows that while designing a DEAP pressure sensor, is the stiffness on one direction of the composite must be taken into careful consideration. If the stiff direction is broken or damaged due to the pressure i.e. the electrode is broken it is impossible to get correct measurements of the capacitance. A solution for this problem would be to choose the compliant direction of the composite to deflect under pressure, and trying not to disturb the stiff direction of it.

VII. Conclusion

A novel investigation into the area of using Dielectric Electro Active Polymer material and laminate for pressure sensing has been conducted successfully. A state-of-art membrane has been conceptualized, modeled and developed, afterward the relation between the key variables (pressure & capacitance) has been determined.

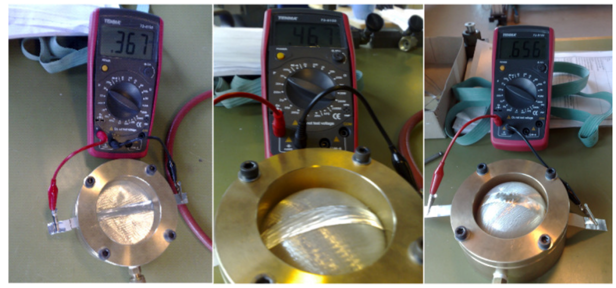

Fig. 9. Deformation of the EAP material

The potential of using DEAP material for applications whereby higher pressure requirements is valid by stacking extra layers of film, thus increasing the resistance of the material when deformed. The pressure range when testing one layer proves a high correlation within an approximate range of 0-23 mmHg (1.7% mean error) , this correlation can be generalized and expanded for a higher pressure range of 120-180 mmHg which is the requirement for Blood pressure sensing.

The output of the work expand the horizon of using the DEAP material for pressure sensing applications. The high reliable properties of the DEAP such as flexibility, formability, low weight and the ability of integrating it in a surface provides a ground breaking potential for applying it in many applications such as pressure, strain, force or shear stress sensing.

Acknowledgment

The authors would like to thank the engineering staff at Danfoss PolyPower A/S, Nordborg, Denmark for supplying us with the adequate material which enabled us to verify the precision of the static and dynamic models of the Dielectric Electro Active Polymer material. Moreover, special thanks to Professor Geir Hovland for supplying us with his extensive knowledge in the area of research and development.

References

[1] D. Best, L. Caffery, F. Gerbovics, D. Nogueiraeidrich, M. Johnson, K. Mitsubayashi, A. Monteiro, M. Snow, A. Sparenberg, A. Stranieri *et al.*, "Poster presentations-short papers/abstracts," *Telehealth for every nation, community and home.*, p. 81.

[2] M. Shahinpoor, Y. Bar-Cohen, J. Simpson, and J. Smith, "Ionic polymer-metal composites (ipmcs) as biomimetic sensors, actuators and artificial muscles-a review," *Smart Materials and Structures*, vol. 7, p. R15, 1998.

[3] L. Bay, T. Jacobsen, S. Skaarup, and K. West, "Mechanism of actuation in conducting polymers: osmotic expansion," *The Journal of Physical Chemistry B*, vol. 105, no. 36, pp. 8492–8497, 2001.

[4] R. Pelrine, R. Kornbluh, and J. Joseph, "Electrostriction of polymer dielectrics with compliant electrodes as a means of actuation," *Sensors and Actuators A: Physical*, vol. 64, no. 1, pp. 77–85, 1998.

[5] P. Sommer-Larsen, J. Hooker, G. Kofod, K. West, M. Benslimane, and P. Gravesen, "Response of dielectric elastomer actuators," in *Proceedings of SPIE*, vol. 4329, 2001, p. 157.

[6] Y. Bar-Cohen, "Electroactive polymers as artificial muscles-reality and challenges," *Structure*, 2001.

[7] ——, *Electroactive polymer (EAP) actuators as artificial muscles: reality, potential, and challenges*. Society of Photo Optical, 2004.

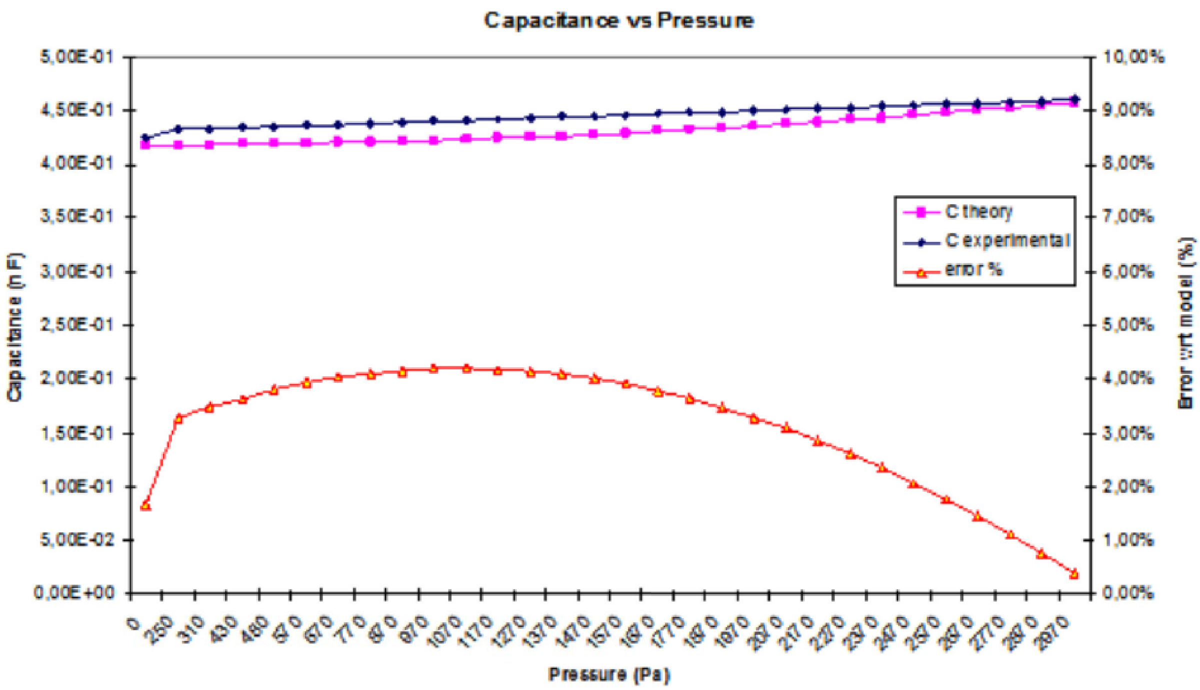

Fig. 8. A graph comparing the theoretical expectations to the experimental test results

[8] P. Brochu and Q. Pei, "Advances in dielectric elastomers for actuators and artificial muscles," *Macromolecular rapid communications*, vol. 31, no. 1, pp. 10–36, 2010.

[9] R. Pelrine, P. Sommer-Larsen, R. Kornbluh, R. Heydt, G. Kofod, Q. Pei, and P. Gravesen, "Applications of dielectric elastomer actuators," in *Proceedings of SPIE*, vol. 4329, 2001, p. 335.

[10] Y. Bar-Cohen, "Electroactive polymers as artificial muscles-capabilities, potentials and challenges," *Handbook on Biomimetics*, vol. 11, no. 8, 2000.

[11] J. Madden, N. Vandesteeg, P. Anquetil, P. Madden, A. Takshi, R. Pytel, S. Lafontaine, P. Wieringa, and I. Hunter, "Artificial muscle technology: physical principles and naval prospects," *Oceanic Engineering, IEEE Journal of*, vol. 29, no. 3, pp. 706–728, 2004.

[12] Y. Iskandarani, R. Jones, and E. Villumsen, "Modeling and experimental verification of a dielectric polymer energy scavenging cycle," in *Proceedings of SPIE*, vol. 7287, 2009, p. 72871Y.

[13] C. Calhoun, J. Berg, D. Koskenmaki, and R. Swinehart, "Electrically conductive adhesive tape," Feb. 11 1992, uS Patent 5,087,494.