Cardiac Sensor Preliminary Experimentation

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1 Abstract

A simple Alginate based Hydrogel and Hall effect sensor were tested for appropriateness as a high sensitivity strain sensor. The Alginate hydrogel fractured at 66% strain and showed significant plastic deformation during repeat loading at 10% strain. The guage factor of Alginate was measured at 2.79. The Hall effect sensor was calibrated using a Rational curve fit with 5 coefficients (achieving a R-squared error of 0.98 and RMSE of 0.30). In testing phase the sensor achieved errors as high as 23%, standard deviation of 5.6 and a mean error of 6.4 %. With the equipment available, a resolution of 5 μm was achieved. The optimal resolution of the Hall effect sensor however, could be much higher. Based on the highest resolution of strain measurements, the smallest measurable stress was approximated at 300 and 20 Pa for Elastic and Cardiac tissue respectively.

2 Objectives

Develop a high sensitivity strain sensor for small displacements ($< 5mm$) and small stress values ($0.005Pa > \sigma > 0.03Pa$). The purpose of this sensor is for measuring forces imposed on cardiac tissue during a) in lab training, and b) implanted in living organisms.

Therefore, the aim of this project is two-fold;

Objective 1. Develop a bio-compatible sensor solution for in vivo measurements

Objective 2. Develop a non bio-compatible sensor solution for in vitro measurements

Objective 1. is the main project goal, if successful would achieve both objectives. However given the difficult nature of the task, Objective 2 was run in parallel. Both objectives are presented in this report.

3 Introduction

There are many families of strain sensor, however, ionic conducting hydrogels show great potential for electrical properties as well as bio-compatibility [1].

Alginate is a naturally occurring, bio-compatible hydrogel [2] that is used for medical applications such as wound healing, drug delivery, in vitro cell culture, and tissue engineering.

Alginate based Hydrogels have been used for the application of strain sensors, achieving high gauge factor and stability [3] [4]. In these cases, double network hydrogels were manufactured for improved mechanical properties such as improved toughness and self-healing.

Alginate hydrogel networks conduct electricity through the mobility of charged ions [5]. This is affected by the use of different crosslinkers. As the hydrogel is stretched, the charge pathways are limited and the resistance increases [3].

No research found suggested the use of a single network Alginate based Hydrogel as a strain sensor. The limitations of which are that they do not exhibit high stretchability; for example, an alginate hydrogel ruptures when stretched to about 1.2 times its original length [6]. However, for low strain applications such as this, high stretchability is not a requirement. Therefore the focus of this project has been on exploring the potential of a simple Alginate Hydrogel as a strain sensor.

For Obj. 2, a reliable strain sensor with high sensitivity is required. Although a non-biocompatible sensor would be used pre-dominantly in vitro, there is merit for a sensor that could operate within the body within a contained biocompatible membrane.

There are multiple options for measuring displacement using currently available sensors. Ultrasonic sensors measure distance based on the principle of propagation time of ultrasonic waves. Ultrasonic transducers have been shown to achieve less than 1 % error [7]. However, calculations of distance are sensitive to changes to the speed of sound in different mediums, which can vary. In addition, the sensors are generally bulky. For these reasons, it would be an inappropriate choice for potential in vivo measurements.

Hall effect sensors measure the strength of a magnetic field. The strength of a magnetic field has an inverse square relationship to distance, hence it can be used to measure distance after calibration. Hall effect sensors have been shown to measure linear displacements up to 0.3 μm for the range of displacement of 1 mm [8]. Hall effect sensors are also very compact and are not affected by any surrounding non magnetic medium. Therefore, potentially a good solution for allowing future in vivo measurements, given a progression of bio-compatible membranes.

4 Materials and Experimental Set-Up

4.1 Objective 1: Alginate Hydrogel

4.1.1 Materials

The experimental procedure from [5] was used to produce the Alginate samples using Sodium Alginate and Calcium Chloride. Sodium Alginate and Calcium Chloride were obtained from Amazon.

4.1.2 Preparation Procedure

Sodium Alginate powder was blended with distilled water to produce a 2M solution. $CaCl_2$ crosslinker was dissolved in distilled water to produce a 1M solution. Filter paper was cut to size and soaked in crosslinker solution. This was placed between two parts of the mould and screwed in place. Sodium Alginate solution was then placed in the mould cavity. A scraping tool was then used to remove excess solution. The second filter paper section was placed on top of the mould to allow even diffusion of crosslinker solution. Fig. 1 shows the mould design. The sample was removed after 10 minutes for complete gelling and used immediately to prevent deterioration.

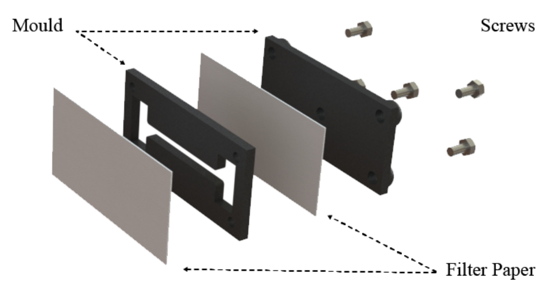


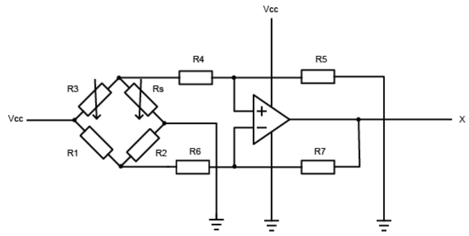
Figure 1: Exploded view of Alginate Mould

4.1.3 Circuit

A Wheatstone bridge circuit was used to detect changes in resistance of the hydrogel sample R_s. This was coupled with differential operational amplifier (LM324AN) with variable gain (depending on values of R5 and R7). The circuit diagram is shown in fig. 2. The circuit design is based off [9].

4.2 Test Rig

A test rig was built that achieves linear movement using a stepper motor and threaded rod. This was used to apply strain to the samples. Fig. 3 shows the



Component	Value (kΩ)
R1	0.3
R2	0.3
R3	100
R4	1
R5	1, 51
R6	1
R7	1, 51

Figure 2: a) Circuit Diagram b) Resistor Values

set-up. This set-up gave a maximum resolution of 5 μm per step. Parts were 3D printed in PLA using a Prusa i3 3D Printer. Data was transferred from the Arduino, via serial communication to MATLAB for processing.

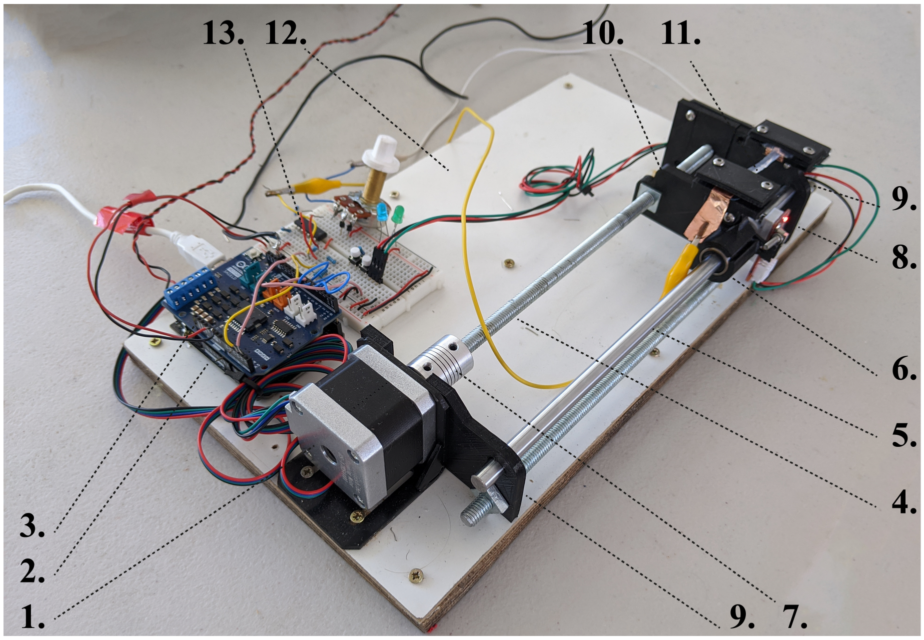


Figure 3: Test Rig

Number	Description	Cost (£)
1	NEMA 17 Stepper Motor	8.99
2	Arduino Uno	8.99
3	Arduino A000079 Motor Shield Rev3	12.90
4	Threaded shaft (BH03016 M8 x 300 mm, 1mm pitch)	1.96 (x2)
5	Linear shaft (BQLZR - 8 mm diameter)	7.70
6	LM8UU Linear Bearings	0.66
7	Aluminum Shaft Coupler (5mm to 8mm)	1.59
8	Mechanical End Stop Switch	1.39
9	3D Printed Mounts	-
10	3D Printed Carriage	-
11	Alginate Sample being stretched	-
12	MDF Mounting board	-
13	Electronics on Breadboard	approx 10 (See Fig. 2a)
14	Hall Effect Sensor (49E SS49E)	0.79
15	Neodymium permanent magnet (N42 - 2mm diameter x 1 mm thickness)	0.09
TOTAL		46.07

Table of component values

4.3 Objective 2: Hall Effect Sensor

Hall effect sensor (49E SS49E) and Neodymium permanent magnet (N42 - 2 mm diameter x 1 mm thickness) procured from Amazon. The hall effect sensor is one potential solution for reliably and repeatably measuring displacement. A change in magnetic field causes a change in resistance of the sensor. Once calibrated, the sensor could provide repeatable measurements for displacement, for measuring strain of samples in the lab. Fig. 4 shows the sensor and magnet mounted on the test rig.

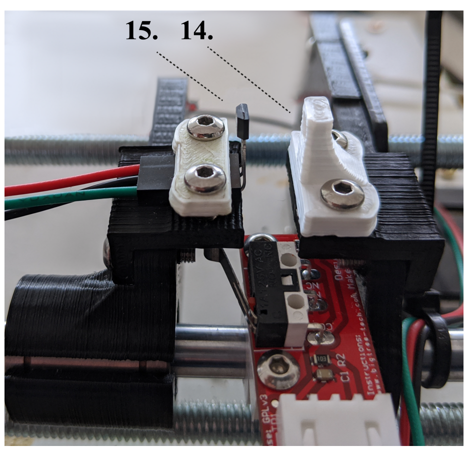


Figure 4: Hall Effect Sensor and Magnet Mounting

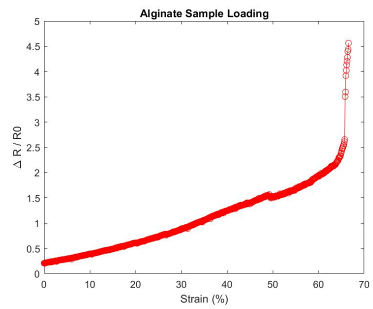
5 Experiment

5.1 Biocompatible Alginate Sensor

A simple alginate sample (see Fig. 1)was tested for change in resistance when under applied strain.

Initial test of material was performed without signal gain (R5, R7 = 1 k Ω). Material was stretched by 20 mm (ϵ = 66 %) in steps of 0.025 mm. The loading curve is shown in Fig. 5a). The sample fractured at around $\epsilon = 65\%$. The Gauge Factor was measured to be 2.79.

Repeat loading cycles were performed with 51x signal gain (R5, R7 = 51 k Ω). 5 cycles of 3 mm ($\epsilon = 10\%$) showed significant creep in resistance. This can be seen in Fig. 5 b).



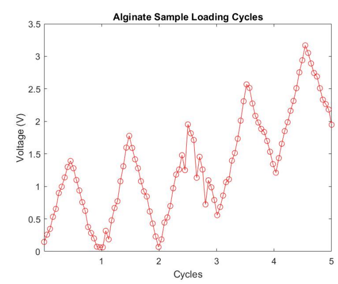


Figure 5: a) Loading of Alginate sample b) 5 Alginate Sample loading cycles at 10% strain

5.1.1 Discussion

What is evident in Fig. 5 b) is the upwards creep of resistance readings on repeat cycles. This could be a caused by plastic deformations in the material, due to limited elasticity. Alternatively this creep could be caused by changes in contact resistance between the sample and the electrodes as the material is stretched and slips.

5.2 Non-Biocompatible Hall Effect Sensor

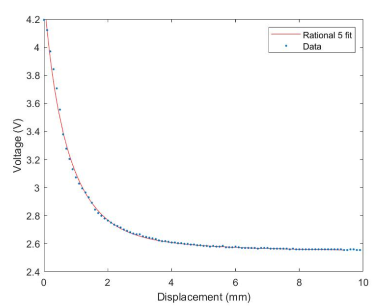
The Hall effect sensor and permanent magnet were mounted on the test rig as shown in Fig. 4. The Hall effect sensor was supplied with 5V, and separated up to a displacement of 10 mm. It must be noted that the orientation of the magnet may affect reading values.

5.2.1 Calibration

The sensor showed high sensitivity with low displacements $< 5mm$. The curve was modelled using a variety of different fits which are shown in Fig. 6. The fitting models were evaluated for number of coefficients, R-squared and root mean squared error (RMSE). The best scoring model was the Rational 11. However, Rational 5 was selected because it had comparable r-squared and RMSE values with less than half the number of coefficients. The Rational 5 fit is shown in Fig. 7.

Fit	Coefficients	R-square error	RMSE
Rational	11	0.9902	0.3022
Rational	5	0.9885	0.3170
Rational	3	0.9833	0.3234
Rational	2	0.9788	0.4248
Polynomial	7	0.9574	0.6213
Power	2	0.9232	0.8080
Polynomial	4	0.8060	1.2976

Figure 6: Table of models and fitting parameters



Coefficient	Value
p1	-0.8683
p2	5.926
p3	-9.32
q1	-4.391
q2	4.708

Figure 7: a) Calibration Curve of Hall Effect sensor b) Table of coefficients of calibration curve

The rational fit is described in Equation 1.

$$x = \frac{p1V^2 + p2V + p3}{V^2 + q1V + q2} \tag{1}$$

5.2.2 Validation

The sensor was tested in order to evaluate the accuracy of displacement approximations. Fig. 8 shows the disparity between values from the hall effect sensor and readings directly from the stepper motor (as this is the best means of measuring displacement available). A maximum error value of 23 %, standard deviation of 5.6 and a mean error of 6.4 % was found.

Repeat cycles of 2 mm displacement were performed. As seen in Fig. 9. These show consistency up to 100 cycles.

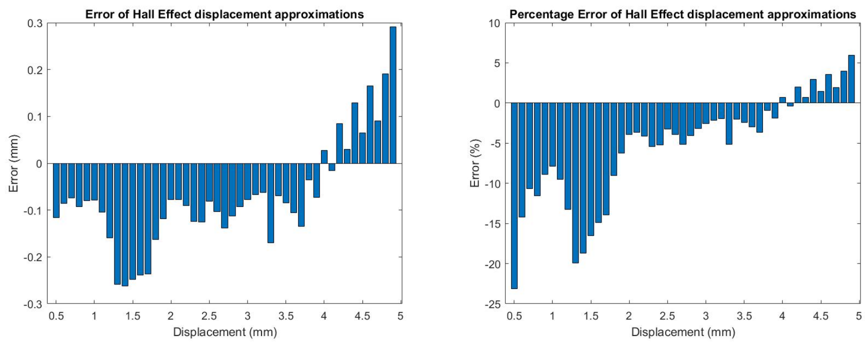


Figure 8: Error in sensor measurements a) mm b) %

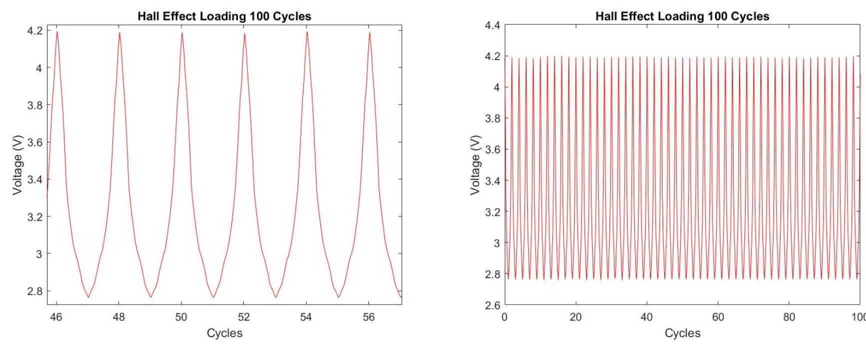


Figure 9: Loading cycles of Hall effect sensor a) 5 b) 100

5.3 Elastic as cardiac tissue

5.3.1 Approach

Elastic band was used to emulate cardiac tissue for validation purposes as unable to access lab. The alginate sample was secured to the band using Gorrilla Glue. Once the Young's Modulus of the elastic had been established, approximations of stress can be determined from measurements of strain. This is shown in the Equation 2 below.

$$\sigma = E/\epsilon \tag{2}$$

5.3.2 Results

Empirical data was used to find Young's modulus of elastic.

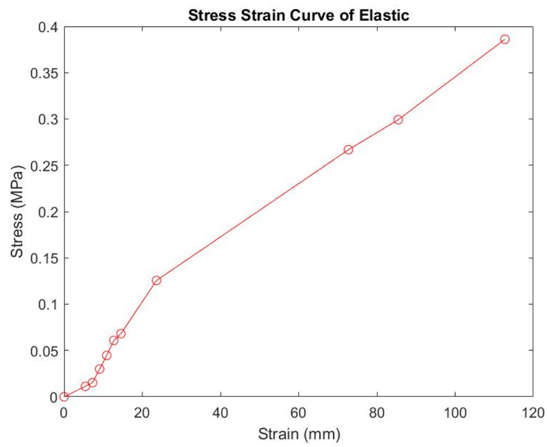


Figure 10: Stress Strain Curve of Elastic

found to be approximately 0.3 MPa which is within the range of values for cardiac muscle (0.02 to 0.50 MPa) [10].

A non-linear region in the stress strain graph (Fig. 10) was identified and hence the elastic band was pre-tensioned to ensure measurements are taken within the linear region.

Given the limitations of the test rig resolution of 5 μm, and the established value of E, the smallest theoretical stress measurable is 300 Pa. This is derived from Equation 2, with an assumed initial sample length of 5 mm. The same calculation was performed with the lowest Young's Modulus value of cardiac tissue (0.02 MPa), to achieve a theoretical stress measurement of 20 Pa.

Key Metric	Elastic	Cardiac Tissue
E of Elastic (MPa)	0.3	0.02
Smallest Measurable Stress (Pa)	300	20

The Hall effect sensor was tested at 100 repeat cycles while mounted to elastic band. Fig. 11 shows the results. There is gentle but consistent upward trend in Voltage readings. This can be compared to Fig. 4 b) where the Hall effect sensor was mounted directly to the rigid mount. The difference is possibly a result of slip. The elastic may slide ever so slightly from inside the clamped mount. Thus resulting in the distance between the sensor and magnet getting incrementally smaller, producing higher Voltage readings.

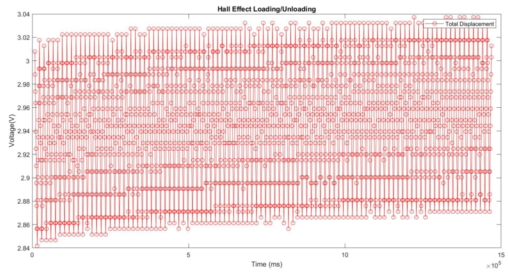


Figure 11: Hall effect sensor mounted to elastic 100 loading cycles

6 Discussion

6.1 Alginate

The sample fractured at around$\epsilon = 65\%$. The Gauge Factor was measured to be 2.79. This is an impressive achievement, performing better than other hybrid network Alginate Hydrogel (GF = 0.478, 0.348) [3],[4]. However, this significantly lower than the state of the art high sensitivity strain sensors achieving values up to 160 [11].

Upon repeat loading, there is a significant creep in Voltage readings. It is difficult here to deduce with absolute certainty the cause of the rise. However, upon observation, the samples do exhibit some plastic deformation, even under low strain. There is also the possibility of contact area change and slip. Observing a similar (but less significant) phenomena while loading Elastic (see Fig. 11) suggests that the mechanism for mounting samples is not sufficient.

6.2 Hall effect

The results from the Hall effect sensor are promising for meaurements in the sub 5 mm range. The rational model is a good fit and cycles show high repeatability. The error testing however is still high (maximum 23%). The cause of which is likely to be a zeroing error. The measurements of displacement are taken directly from the stepper motor encoder and depend on the position of the mechanical end-stop. Therefore, slight variations in end-stop position will produce an offset error. Therefore, a more reliable form of measurement should be used to validate the accuracy of calibration.

In addition, the Hall effect is greatly affected by the orientation of the permanent magnet and sensor. Slight variation in the relative orientation will provide errors. Therefore, a robust design should be developed.

6.3 Elastic Mounting

Mounting both sensor types to the elastic was achieved successfully with glue. However, it remains untested as to how it would work on cardiac tissue. The approach of approximating stress relies on a valid and consistent value of Young's Modulus. In terms of achieving the target stress value of $(0.005Pa > \sigma > 0.03Pa)$, the experimental set-up was not successful, achieving approximate values of 300 Pa. In addition, the use of Young's modulus of the elastic based on empirical data is tenuous. Fig. 2 shows the variance in stress/strain gradients and thus a flaw in approximations.

7 Further Work

Based on the outcomes of this research, significant further work should be undertaken. The use of a more robust, elastic and bio-compatible sensor should be considered. Solutions such as hybrid network hydrogels, with enhanced Gauge Factor may provide an appropriate solution.

The use of Hall effect sensor for in vitro measurements has the most immediate potential, achieving sensitivity in the correct range. However, more thorough calibration with and testing with higher precision equipment is recommended. A resolution in the sub nm range would be required to achieve measurements in the target stress region.

A sensitivity analysis of sensor orientation and a robust design for the sensor should be carried out to ensure that unwanted changes in orientation does not affect future test results.

In addition further validation of the stress approximation approach should be carried out.

Further assurances should be taken to ensure secure fastening of tissue samples. This would be required to guarantee the effect of slip does not interfere with test results.

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