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Contact force sensor for artificial hands with a digital interface for a controller

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Abstract

A force sensor, for use with an artificial hand, needs to be small, robust, low power, cheap and easily interfaced to a controller using digital techniques. The prototype featured in this paper uses capacitance effects to measure the strain on an elastic polymer foam. Low power consumption results in a device that can be supplied from a miniature battery thereby requiring only signal wires to the controller. A non-linear model accurately describes the characteristic of the sensor, requiring the estimation of only three parameters. The device has been tested up to 20 N but is capable of measuring greater forces.

Keywords: force, capacitance, polymer foam, non-linear model, artificial hands, robot feedback control

(Some figures in this article are in colour only in the electronic version)

1. Introduction

In the design of an artificial hand for the disabled, there is overriding requirement to make the prosthesis as lightweight as possible and to constrain the mass to less than 500 g [1]. Yet, in order to improve the functionality and usefulness of a hand, the addition of feedback control and hence the measurement of finger position, object-slip and the focus for this paper, force (or touch), is required. Each sensor must fit the appropriate place in the structure of a hand but also is required to be as small as possible. For an anthropomorphic hand the palmar side contains force sensors especially at the fingertips. Typical commercial hands do not have closed loop controls with force feedback and only a few designs have sensors embedded in their structure.

For this application, the standard features for any transducer are required (low temperature drift, good accuracy, good repeatability and insusceptibility to EM interference) and the specifications shown in table 1.

No single method is appropriate for tactile sensing [2] and the aim is to select a suitable technology with the least shortcomings. The majority of tactile sensing research has been conducted in the area of robotics. Unfortunately this has limited application within prosthetics due to the size, mass, power or signal sensing factors.

Table 1. Specifications for a prosthetic force sensor.

- Forces up to 100 N
- High sensitivity to small forces
- Integral power supply
- Lightweight
- Little hysteresis
- Low power consumption
- Not easily damaged by large impact forces
- Robust
- Service period of six months
- Simplicity in construction and mounting
- Simple digital interface to a controller
- Small size with an area less than 100 mm^2
- Thin in depth for mounting on fingers and palm

Force sensitive resistors (FSR) [3] have found application in medical engineering for the measurement of the pressure distribution under the foot of a person with diabetes. They are suitably thin and small with a simple interface but have high drift and low accuracy. A strain gauge, while providing good accuracy and stability with proven technology, requires a bridge amplifier with the consequent large size and high

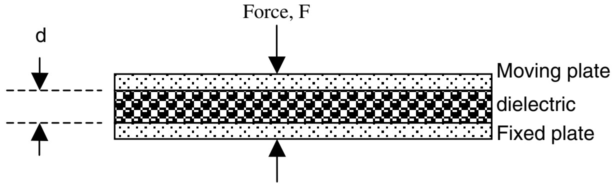

Figure 1. Parallel plate capacitor with centre elastic material.

power consumption. Optical transmission, using an LED and photodiode at either ends of an elastic tube, has provided a reliable sensor with integral acoustic slip sensing [4]. However this device is complex to manufacture. Hall effect sensors are a low cost solution but need to be turned off by the controller when not in use otherwise they dissipate too much battery power with the subsequent increase in system complexity [5]. Piezoelectric sensors require signal-conditioning circuits for static measurement and there is some difficulty in separating the piezoelectric and pyroelectric effects. However this technology may provide a good solution with small size and low power. Zhu and Spronck [6] have reported that capacitive tactile sensors have little hysteresis, creep and temperature dependency making them an attractive technology for object manipulation.

The market for artificial hands is relatively small and prohibits the use of high cost sensors using a sophisticated technology [6–8]. However a special sensor which is easy to manufacture in low volumes would provide an incentive to the introduction of closed loop control into commercial prosthetic design with the subsequent improvement in functionality. The novelty of the work presented in this paper is primarily in the prosthetics context. However it also has the potential to be applied to other areas where an inexpensive, small and low power device is required.

2. Sensor

The sensor consists simply of two parallel plates separated by an elastic material that also acts as a dielectric (figure 1). Applying a force across the plates increases the capacitance.

2.1. Elastic material

The material between the two plates should yield sufficiently under low loads, in an ideal elastic (no hysteresis) and repeatable manner. The obvious material is spring steel since this has the potential to produce a sensor with little hysteresis. Standard helical compression springs could be used but would be custom made with the subsequent increase in cost. Disc springs or Belleville washers are ideal devices for high loads in small spaces. With a conical shape they are loaded axially and can be subjected to static and dynamic forces. Under high loads they have a small deflection making them ideal for this application. A test sensor was constructed using two disc springs, which had a useful and linear characteristic but a large and unacceptable amount of hysteresis. This effect is due to the friction between the washers and the plate surfaces. It can be reduced with suitable lubrication but this restriction was considered to be a poor feature of any design.

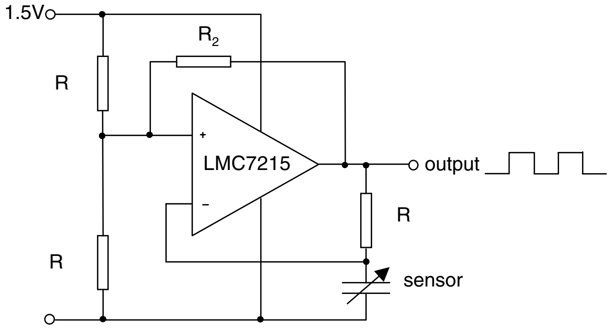

Figure 2. Square wave oscillator.

A material that fills the whole space between the two plates would make construction of the device both easier and therefore less expensive to manufacture. Metals do not deform sufficiently under low loads while polymers exhibit hysteresis under strain. However polymeric foams, made by dispersing a gas in a solid have the advantage of a choice of density and cell geometry. The gas-phase composition modifies the physical properties of the solid-phase plastic substrate. These foams can be manufactured with a low compression set (ASTM standard) giving them good stability and low hysteresis.

2.2. Capacitance

The sensor is based on the principle of estimating the capacitance between two conducting plates separated by an insulator. Now

$$C = \frac{\varepsilon_0 \varepsilon_r A}{d} \tag{1}$$

where C is the capacitance, ε_0 is the permittivity of free space, ε_r the dielectric constant of the material in the gap, A is the area of the plates and d their separation. The capacitance is a function of the geometric configuration and dielectric constant. Linking the change in force applied across the plates to a change in either A, d or ε_r through a suitable elastomer creates a sensor.

For ease of construction, the separation of the plates is used in which case the capacitance is an inverse function of the force. This follows from

$$F = k(d_0 - d) \tag{2}$$

where F is the force across the plates, k is a constant and d_0 is the separation with zero force.

Combining equations (1) and (2) gives

$$C = \frac{\varepsilon_0 \varepsilon_r A k}{k d_0 - F}. \tag{3}$$

Thus capacitance varies as the inverse function of force.

2.3. Square wave oscillator

A convenient method of converting a change in capacitance into a signal is to use a relaxation oscillator (figure 2).

Charging and discharging currents for the sensor flow through the resistor, R. The resistors, R_1 form a potential divider to provide the centre voltage of the supply and allow for a square wave output. Hysteresis for the comparator (LMC7215) is provided by the resistor R_2.

The circuit oscillates with a frequency given by

$$f = \frac{1}{2CR\ln[(R_1 + R_2)/R_2]}. \tag{4}$$

Combining equations (3) and (4) (eliminating C) shows that as the force increases the frequency decreases linearly as follows:

$$f = \frac{kd_0 - F}{2R\varepsilon_0\varepsilon_r Ak\ln[(R_1 + R_2)/R_2]}. \tag{5}$$

3. Experimental results

3.1. Polymer foams

Several polymer foams were studied to identify one with suitable mechanical and electrical properties. Samples were bonded between two brass shims (300 mm^2) with cyanoacrylate. The whole assembly was bonded to a very high-density polycarbonate strip to provide a stable platform. Each sample was loaded to 8 N and unloaded with three replicates. Samples with a high hysteresis were rejected. The change in capacitance with applied load was measured using a standard 555 timer circuit and a supply voltage of 5 V.

Experimentally, the circuit (figure 2) was found to have a stable oscillation with a capacitance greater than 0.8 pF. To meet this requirement, the area, A, in equation (1) has to be more than 49 mm^2 (minimum diameter of 7.9 mm) for a relative permittivity of 2.6 and a thickness of 1.4 mm.

3.2. Sensor

The circuit and sensor components are mounted on a printed circuit board (PCB), see figure 3. An 8 mm diameter piece of foam was bonded to the copper side of the upper plate made from Pyralux. On top of this, a shim formed the upper force plate made from PEEK (0.7 mm thick) which was bonded to the polyamide surface of the Pyralux. The foam was bonded to the lower plate. A ‘tang’ from the upper plate was rolled to form a loop and soldered to the PCB. The assembled device is shown in figure 4 together with an adult human small finger to illustrate the size of the sensor. The lead-lengths between the sensor and the inputs to the capacitor were made as short as possible to reduce the effects of adding stray capacitance to the sensor.

3.3. Load tests

Static forces were applied to the sensor using steel weights and the output frequency recorded using a frequency counter. Loads were applied incrementally to nearly 20 N and then unloaded. Three replicate experiments were carried out. Figure 5 shows the experimentally observed frequency against applied force.

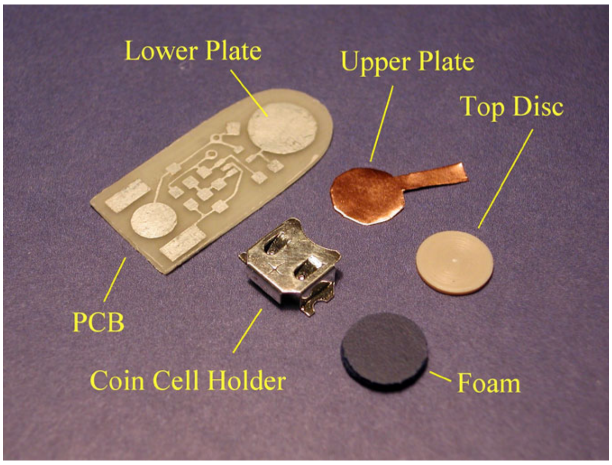

Figure 3. Components of the sensor.

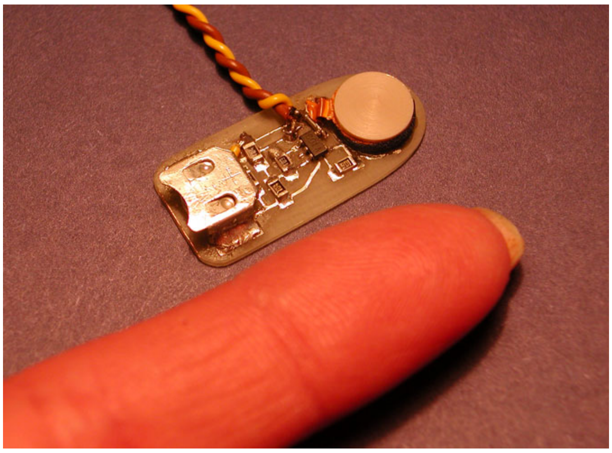

Figure 4. Sensor and human finger.

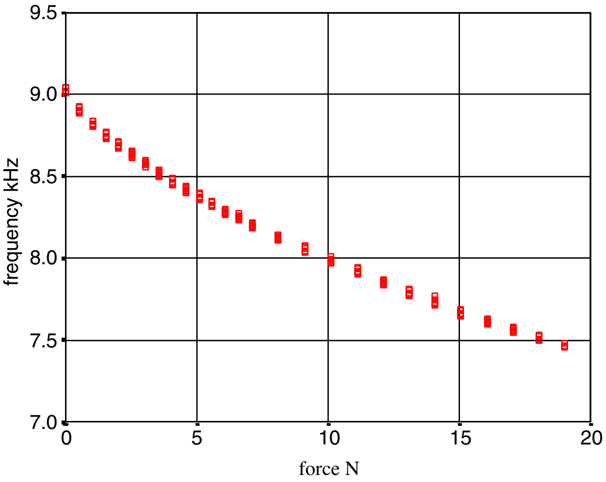

Figure 5. Observed frequency against force.

3.4. Non-linear model

Inspection of the characteristic (figure 5) demonstrates that while the frequency decreases with increase in force (equation (5)) the relationship is non-linear. A monotonic model was fitted to the data of the form given below.

$$f = B_1 - B_2F^{B_3}. \tag{6}$$

Non-linear regression was used to determine estimates of the three parameters, B_1, B_2 and B_3. The output from the

Table 2. Summary of parameter values and statistics.

Parameter	Estimate	Asymptotic standard error	Asymptotic 95% confidence interval Lower	Asymptotic 95% confidence interval Upper
B_1	9.046 9231	0.005 1306	9.036 79	9.057 06
B_2	0.229 2875	0.003 7901	0.221 80	0.236 77
B_3	0.660 4190	0.005 0015	0.650 54	0.670 30

computer package SPSS gave an R-squared value of 0.998 77 demonstrating a very good fit to the model. Table 2 summarizes the parameter values.

Figure 6 shows the residuals (Hz) against observed frequency.

3.5. Power and weight

The sensor supply current is 3 μA at a voltage of 1.5 V. The device, complete with a coin-cell battery weighs 0.96 g.

4. Discussion

The sensor should be maintenance-free for a period of at least 6 months, at which point the battery could be replaced. For 6 months (4368 h), the circuit requires 13 mA h which can be supplied from a standard energizer 321 battery (14 mA h, 6.8 mm diameter and 1.65 mm thick). This is a significant advantage as the connections from the sensor to the controller are reduced from three to two wires (signal and ground). Future prosthetic hands will have large numbers of sensors so any savings with interconnections decreases the mass of a hand and improves its reliability. This design philosophy also anticipates that future sensors may be self powered, require minimal power consumption and be intelligent [9–11].

Three sensors could be mounted on a finger: one at the tip and one on each of the two other phalanges, making a total of 12 sensors for the four fingers. The thumb could have two while six could be sited on the palm. This arrangement has a mass of 20 g assuming a mass of 1 g per sensor including its interface circuits and interconnecting wires. To provide a sufficient number of sensors across the palmar surfaces represents 5% of the total mass (400 g) of a recent lightweight prosthesis [1]. Improvements in the sensor will reduce this value further, opening up the possibility of putting more sensors in a hand to improve functionality without a significant mass penalty.

A square wave output allows for a simple interface to a digital signal processor or microcontroller using a counter/timer to measure the output frequency. This technique has been successfully applied to accelerometers (Analogue Devices).

The absolute value of the slope of the characteristic decreases as the force increases (figure 5), forming a higher sensitivity for light grip forces, which is a desirable feature for this application. From equation (6) the slope is given by

$$\left|\frac{\mathrm{d}f}{\mathrm{d}F}\right| = B_2 B_3 F^{(B_3-1)}. \qquad (7)$$

The slope decreases with increasing force, as the parameter B_3 is less than one.

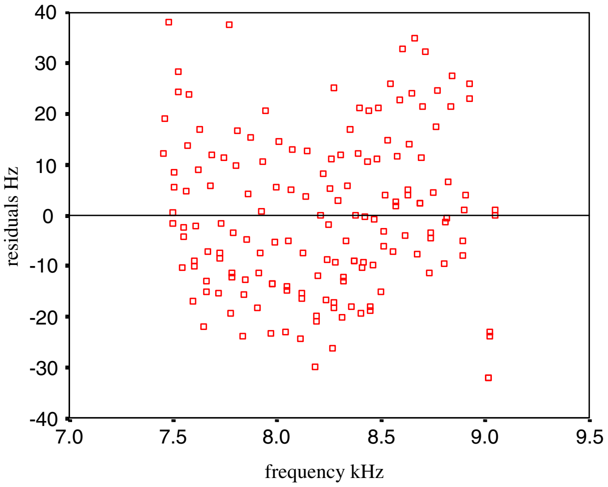

Figure 6. Residuals against observed frequency.

Observed values of frequency were substituted into equation (6) to estimate the applied forces and these were subtracted from the actual applied forces. These absolute errors have a mean value of 0.82% and a standard deviation of 0.62%. The model for the sensor (equation (6)) demonstrates a potential accuracy of less than 1% which is a significant improvement compared to sensors like FSR. The force can be calculated by a microcontroller from the output frequency and the three parameters stored in memory.

It is possible that a local minimum can be found for the non-linear parameter estimation. The residuals show a fairly uniform distribution across the range of data points (figure 6), which is evidence that a global minimum has been found. Also the estimation was repeated a few times with different starting values and found to converge to the same parameter values.

A test was made of how consistent the parameter fit is for different data sets. The data was divided into three sets and parameter estimates obtained (table 3).

The mean parameter values are only very slightly different from those estimated using all the data together (tables 2 and 3). The variances are small which demonstrates that the estimation is robust to noise, otherwise, if the parameter values were very different within the three data sets, the estimation would have been deemed to be ill conditioned. Parameter B_2 has the highest coefficient of variation (standard deviation as a percentage of the mean), followed by B_1 and then B_3.

Various effects contribute to the drift of frequency in a relaxation oscillator. Temperature fluctuation and power supply regulation are common to most circuits. An artificial hand operates over a temperature range from 0 to 40 °C so the addition of a small voltage regulator is required for the stable operation of the sensor. Stray and parasitic capacitance effects need careful consideration. A polymer is likely to have an insulation resistance in the range 10^4–10^8 MΩ μF, which is higher than say the low leakage, found in tantalum capacitors (1–100 MΩ μF). Polymer capacitors have very low dielectric absorption (typically <0.01%). This effect is a hysteresis-like internal charge distribution that causes a capacitor that is quickly discharged and then open-circuited to appear to recover some of its charge. Despite all of theses potential hazards the prototype showed very little drift. However it is anticipated that improvements in this respect will be needed.

Table 3. Mean, variance and coefficient of variation for the parameter values.

Parameter	Data set 1	Data set 2	Data set 3	Mean	Variance	Coefficient of variation
B_1	9.065 1253	9.033 5620	9.042 0837	9.046 9237	2.67×10^{-4}	0.180
B_2	0.230 0479	0.229 1326	0.228 6837	0.229 2881	4.83×10^{-7}	0.303
B_3	0.659 7644	0.660 2445	0.661 2484	0.660 4191	5.73×10^{-7}	0.115

5. Conclusions

A very small, lightweight and low power sensor has been developed for use with artificial hands. It has a maximum sensitivity at low forces for griping lightweight objects. The sensor has been tested to a force of 20 N but has the potential to measure higher forces. A non-linear model accurately describes the relationship between the output frequency and applied force. It could be equally applicable to other sensors where polymer foam separates the plates of a capacitor. An integral small battery powers the sensor for up to six months. The output is a variable frequency square wave, which allows for a simple and serial interface to a digital signal processor or microcontroller.

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