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A flexible encapsulated MEMS pressure sensor system for biomechanical applications

N. K. S. Lee, R. S. Goonetilleke, Y. S. Cheung, G. M. Y. So

Abstract The use of pressure sensors made of conductive polymers is common in biomechanical applications. Unfortunately, hysteresis, nonlinearity, non-repeatability and creep have a significant effect on the pressure readings when such conductive polymers are used. The objective of this paper is to explore the potential of a new flexible encapsulated micro electromechanical system (MEMS) pressure sensor system as an alternative for human interface pressure measurement. A prototype has been designed, fabricated, and characterized. Testing has shown that the proposed packaging approach shows very little degradation in the performance characteristics of the original MEMS pressure sensor. The much-needed characteristics of repeatability, linearity, low hysteresis, temperature independency are preserved. Thus the flexible encapsulated MEMS pressure sensor system is very promising and shows superiority over the commercially available conductive polymer film sensors for pressure measurement in biomechanical applications.

1 Introduction

Micro electromechanical system (MEMS) technology is very promising and has been extensively researched and used in many applications in automotive, aerospace, industrial process control, and telecommunication systems (Trimmer, 1997). For example, micro pressure sensors are used for measuring levels in fuel tanks, pressure in engine cylinders and tires and air-bag-release applications (Mehregany and Huff, 1993), measuring blood pressure (Koester et al., 1996), de-icing in helicopters (Varadan et al., 1996) and so forth. Nowadays, the applications of micro sensors are growing exponentially due to their low cost, small size, and ability to sense the necessary physical parameters with minimum interference.

There are currently two views regarding the pace of MEMS commercialization: very slow, by many of MEMS promoters; and very fast, by many companies whose products are replaced by MEMS substitutes (Walsh et al., 1996). Even though MEMS pressure sensors (Motorola, 1994) are commercially available, the potential of these sensors has not been fully exploited. The commercialization process may be rapidly enhanced and expanded with suitable packaging technology. One promising application is in human interface pressure measurement in biomechanics.

Human-equipment interface pressure is strongly related to comfort, injury, and limb debilitation (Gross et al., 1994; Guirini et al., 1993; Mueller, 1992; Young, 1993; Wilson, 1992; Webster, 1991). Automobile manufacturers such as BMW, Fiat, General Motors, Honda, Nissan, Renault, Toyota and so forth have been using commercial pressure measurement systems to improve seating comfort. Seat pressure systems are also used by rehabilitation centers such as Kibi Kohgen Rehabilitation Center in Japan and Helen Hayes Rehabilitation Hospital in New York to reduce the formation of pressure sores in paraplegic patients. Thus, it is critical that the interface pressure be measured accurately and reliably over a long period of time in dynamic conditions.

Ideally, the pressure sensors that are used for human interface measurement should satisfy the following criteria (Ferguson-Pell et al., 1976):

- Linearity, low hysteresis, low creep (drift). Linearity is preferred, even though it can be overcome using sophisticated calibration methods. In the presence of many sensors, individual calibration (for example, as with conductive polymer pressure sensors (Ferguson-Pell et al., 1976) can be rather cumbersome. Measurement repeatability will be significantly altered if the reading depends on whether pressure is increasing or decreasing (hysteresis).
- Low temperature sensitivity in the range of 20 °C to body temperature of 37 °C (Ferguson-Pell et al., 1976). If the sensors are temperature sensitive, then temperature readings ought to be taken at the corresponding locations, and the calibration protocol would need to account for the temperature variations.

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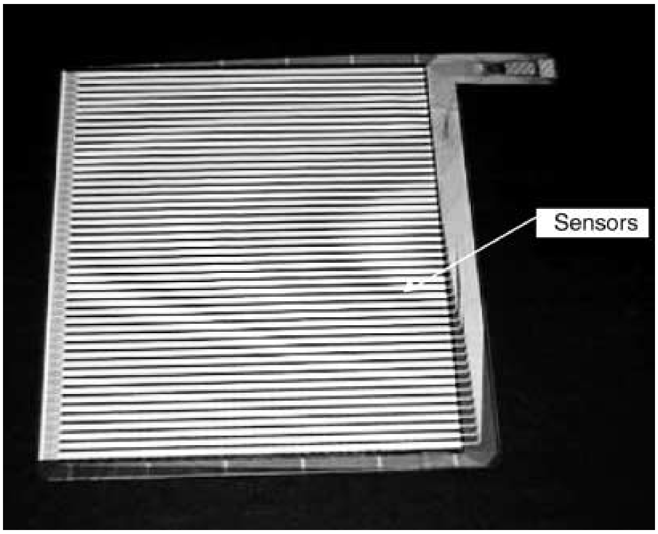

Fig. 1. Sensor layout of the conventional system

- Flexibility. A rigid sensor can affect the surface or interface characteristics between human tissue and the product surface. Hence, it is important that the sensor be flexible and conform to the tested contour such as in seats, shoes, etc.
- Thin. A sufficiently thin sensor is necessary if the geometric characteristics of the surface are to stay unaltered.
- Large range. In most cases, the loading on the sensor(s) can be equivalent to body weight. Hence it is important that the sensors have the ability to measure a large range of pressures (Typically 0–1500 mmHg or 200 kPa). For seat pressure measurements, a sensor that could measure a maximum pressure of 500 mmHg would be sufficient (Ferguson-Pell et al., 1976).
- Repeatability (durability) – Repeated measurements over a long period of time is necessary to quantify the dynamic changes of pressure.
- Unobtrusive to the measured subject, be easy to use and are cost-effective (Gross et al., 1994).

Due to the flexibility and thickness limitations, the sensors used for human interface pressure measurement have been primarily limited to resistive inks (conductive polymer film sensors such as those from FSA, TEKSCAN, Interlink, etc.). A typical layout of one type of commercial system is shown in Fig. 1. The sensors are arranged in a parallel line pattern and a flex-circuit acts as a base. The output of the flex-circuit is usually connected to a computer.

With sensors or sensor mats similar to those shown in Fig. 1, the common pressure sensitive materials have been elastomers. Most conductive polymer sensors yield a decrease in resistance with increasing pressure. This change is the result of an increased contact area between two sheets of polymer (or elastomer) such as mylar or ultem, coated or impregnated with conductive material. There are many performance issues associated with these types of sensors. The repeatability is very poor, and the repeatability from sensor to sensor is about 20% (Olson, 1991). The device is highly nonlinear, and the elastomers exhibit a large degree of hysteresis and creep. Furthermore, bending of the sensor film during measurement causes an additional problem as bending changes the contact resistance between the two sides of the film. Due to these drawbacks of conductive polymer pressure sensors, it is desirable to investigate alternative pressure sensors for absolute pressure measurement.

MEMS technology is a very promising alternative since it is rather well developed, commercially available and of high quality with potential for continuous improvement. But the commercially available MEMS pressure sensors are not quite suitable for biomechanical applications due to their packaging structure and design. If they are to be used for human-interface pressure measurements, they ought to be flexible and thin. Thus, the aims of this paper are to explore the following:

1. Design and develop a flexible encapsulated packaging system to enable MEMS sensors to be used in biomechanical applications.
2. Characterize the flexible encapsulated system that has been designed and developed.
3. Check if the proposed flexible encapsulated package system would degrade the performance of the original MEMS pressure sensor.
4. Provide a simple comparison of performance characteristics between the proposed flexible encapsulated package system and the commercially available conductive polymer sensors.

2 Development of the flexible capsule packaging system

2.1 Design concept

The design involved encasing the micro-sensor inside a flexible capsule filled with a pressure transmissible fluid. The functions of this flexible capsule are twofold:

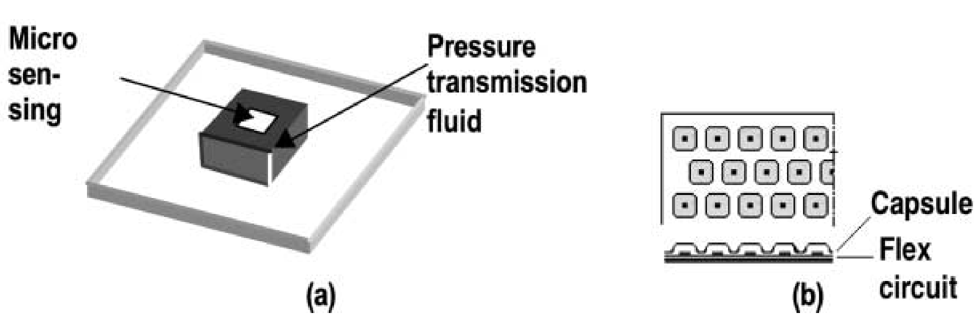

Fig. 2. a Schematic of individual MEMS pressure sensor capsule. **b** Top view schematic of flexible encapsulated MEMS pressure sensor mat design

1. It provides a flexible interface between the relatively rigid pressure sensor chip and human tissue (Fig. 2a).
2. It is a cost-effective solution to develop a complete pressure mat (Fig. 2b) as a relatively small MEMS chip can be used to serve a relatively large sensing area.

The repackaged micro sensor (Fig. 2a and b) or the encapsulated sensor consists of a flat plastisol sheet (Plastisol, 2000), a formed plastisol sheet, pressure transmission fluid, micro sensing chip and a flex circuit. The prototype bubble was $9 \times 9 \times 1$ mm. When pressure is exerted on the top of the capsule, the encapsulated fluid would transmit the pressure to the sensing chip. The sensing chip is hard-wired through a flex circuit underneath the plastisol sheet. Using an appropriate data acquisition system and software, the pressure exerted can be sensed and recorded for further analysis.

2.2
Design considerations

Even though air is a possible pressure transmission medium for the capsule, it has many disadvantages with reference to pressure measurement (such as large thermal expansion coefficient of air, and high compressibility). Thus, temperature dependence and non-linearity would be unavoidable. A non-compressible fluid such as liquid or gel (Esashi and Minami, 1994) is more appropriate for better performance. For ease of containment and handling, gel is more advantageous than fluid. However the higher viscosity of gel can introduce hysteresis and other nonlinear effects into the system. Hair styling gel has been used in this design because it is readily available and preliminary experiments showed that hair styling gel does not introduce any hysteresis or other unwanted effects such as non-linearity and non-repeatability. The steady rate viscosity sweep was performed on the hair styling gel using the Rheometrics (NJ, USA) RMS 800 system with parallel plates at 25 °C. The viscosity curve is shown in Fig. 3.

One of the key elements of the repackaged pressure sensor is the plastic encasing. It must not only hold the pressure transmission fluid and sensing chip, but also be soft and flexible. Plastisol, a mixture of polyvinyl chloride and plasticizer having a Youngs modulus 4.453 MPa (determined using an Instron 5560 Tensile Testing Machine) was used for the plastic encasing due to its high tensile strength (approximately 1200 psi or 8274 kPa), large elongation (approximately 300%), good heat resistance (excellent up to 225 °F), and satisfactory chemical resistance (to most acids and alkalis) (Plastisol, 2000). Plastisol molding is a fairly mature process and is therefore suitable for mass production (Aydin, 2000).

2.3
Prototype development

To assess how well the flexible encapsulated system concept would work, a prototype of a single micro-sensor package was built. The prototype (Fig. 4) was made to simulate the real system. For the sake of convenience, simplicity of implementation, and the ease of comparison of test results, a commercially available MEMS pressure

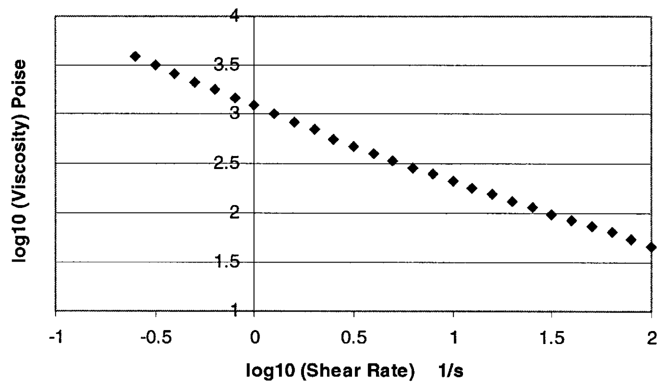

Fig. 3. The viscosity curve for hair styling gel

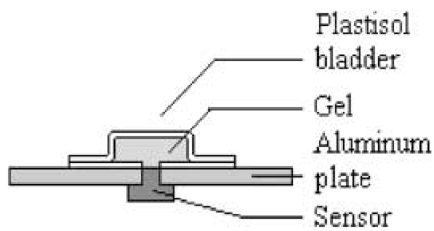

Fig. 4. Prototype of the single repackaged encapsulated sensor

sensor (Motorola MPX2300D[1]) was used instead of a "raw" sensor. This Motorola pressure sensor has been designed using a monolithic silicon piezoresistor, which generates a changing output voltage when the pressure applied is varied. The resistive element constitutes of a strain gauge and is ion implanted on a thin silicon diaphragm. Applying pressure to the diaphragm causes a change in resistance in the strain gauge, which in turn causes a change in voltage in direct proportion to the applied pressure.

The prototype consisted of a plastisol encasing, the Motorola micro-sensor and an aluminum plate. The aluminum plate will not be part of the finished pressure measuring system but it is important in the prototype for fixing and supporting the plastisol encasing and the sensor during performance characterization.

To fabricate the prototype, first plastisol sheets were made from liquid polymer. This was accomplished by filling an aluminum mold with the polymer and heating it to 250 °C for about 10 min. To form the cup shape of the encasing, the top sheet was subjected to an additional forming process using a cup shaped heated aluminum mold. Next, the top and bottom plastisol sheets were bonded cohesively and then cured for one day. Bonding was attempted using ultra-sonic welding, epoxy adhesives, and contact adhesives. A heat curable adhesive, MEK supplied by Respironics, Inc. was found to be the most suitable based on lamination strength. UV curable adhesives [for example, those from Dymax Limited (Dymax, 2000)] would work equally well.

[1]MPX2300D stands for Motorola Pressure X-Ducer (MPX), '2' stands for temperature compensated and calibrated, '300' is the pressure range in mmHg and 'D' stands for differential, basic element.

The plastisol encasing was laminated onto the aluminum plate and gel was injected through a perforation on the bottom flat sheet. The sensing head was then press fitted onto the opening in the aluminum plate (Fig. 5). The subsequent burst test showed that the bubble could withstand a pressure of over 1000 psi (6895 kPa).

3
Performance characteristics of the repackaged sensor

The prototype was subjected to a few different tests in order to evaluate its performance relative to the original MEMS sensor (Maudie and Wertz, 1997). The supply voltage for the MPX2300D sensor was set at the rated value of 6 V dc and applied to pin 1 of the sensor. Pin 4 of the sensor was the ground, while pin 2 was +OUT and pin 3 was −OUT. Other than the temperature dependency tests (Sect. 3.2), all other tests were performed at room temperature (around 20 °C).

3.1
Test setup

A simple test fixture was designed and built (Fig. 6a). During the test, standard test weights were placed on the aluminum platform (Fig. 6a) so that the pressure was transmitted onto the sensor while the aluminum support fixes and aligns the platform. In the hysteresis test to simulate the loading and unloading operations, three standard weights were placed on the platform one by one and the corresponding sensor output was recorded. Then, the weights were removed one by one (unloading) and the readings noted. The applied pressure was calculated using the following formula:

$$\text{Applied pressure}\,(P) = \text{weight (gm)} \times 9.81\,(\text{m/s}^2)/\text{area}\,(\text{m}^2) \qquad (1)$$

where area $= (9 \times 9 \times 10^{-6})$ m^2.

The experimental setup of checking temperature dependency comprised a temperature controller, temperature reader, two Type K thermocouples (accuracy ±3 °C), hot gun, power supply, voltmeter and the repackaged sensor prototype (Fig. 6b). During the experiment, hot air was blown onto the repackaged sensor. When the temperature of the encasing reached the preset level, the thermocouple sensed and fed the temperature information to the temperature controller, which then governed the power input to the hot gun. Meanwhile, the temperature reader sensed the temperature of the bubble throughout the experiment.

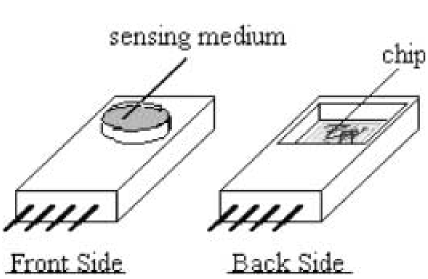

Fig. 5. Motorola pressure sensor MPX2300D

3.2
Temperature dependency

The temperature dependency of the repackaged sensor is shown in Fig. 7. Since the strain gauge of the MPX2300D sensor is an integral part of the silicon diaphragm, the manufacturer claims that there are no temperature effects due to differences in the thermal expansion of the strain gauge and the diaphragm. The properties of the strain gauge however are specified to be temperature dependent and temperature compensation and offset calibration have been built-in to the MPX2300D sensor using resistive elements. The differences seen among the differing temperatures at various applied pressures could be as a result of this compensation effect.

To better quantify the temperature effect with respect to the output at 20 °C, the following measure, E, was also used:

$$E + \frac{P_{\mathrm{T}} - P_{20}}{(T - 20)} \qquad (2)$$

where E = temperature effect in psi/°C at a given temperature (T) and pressure (P_{T}), P_{T} = actual sensor reading of the pressure at temperature (T), P_{20} = sensor pressure reading at 20 °C, T = actual temperature (°C).

Table 1 shows that the E values are rather small, indicating a relatively low temperature effect with respect to the ambient temperature of 20 °C.

3.3
Hysteresis effects

Most non-MEMS pressure sensors suffer from significant hysteresis (Ferguson-Pell et al., 1976). The repackaged sensor was tested even beyond its design maximum

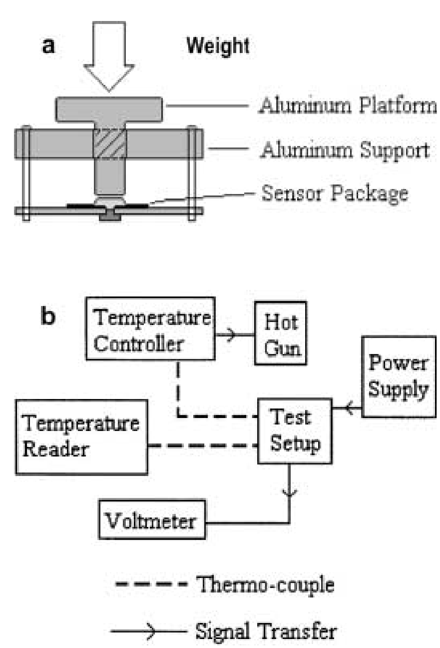

Fig. 6. **a** Experimental setup for testing static performance characteristics (repeatability, hysteresis, creep, stability and linearity). **b** Experiments for checking temperature dependency of the repackaged sensor

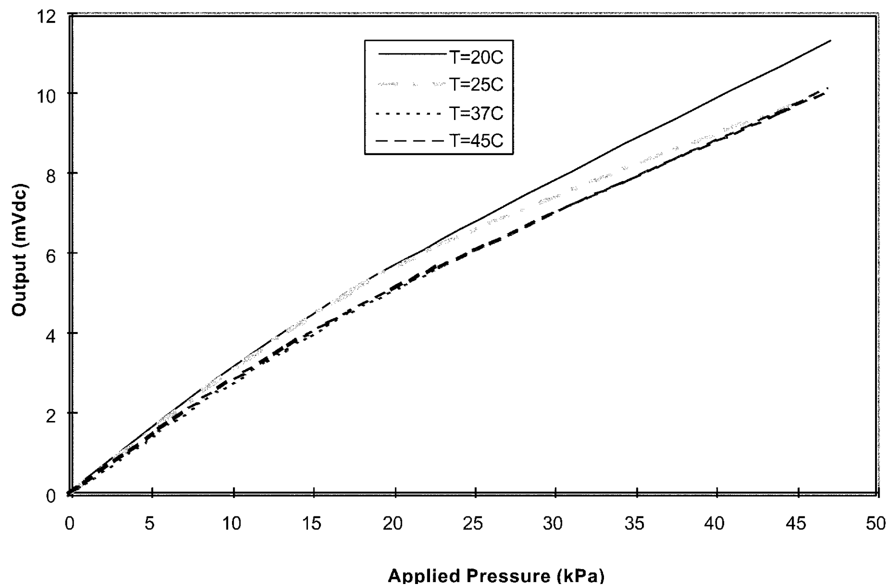

Fig. 7. Temperature effect of repackaged sensor

pressure of 40 kPa (test range was 0–6.828 psi or 47 kPa). However, in general, it would be best that the measured pressure be within the designed pressure for the sensor. Figure 8 shows a typical loading and unloading for the repackaged sensor and it shows very little hysteresis.

3.4 Linearity

The raw sensor is rated to have linearity and hysteresis amounting to −2 and +2 of % full scale span (FSS). A linear regression (independent linearity, Pallas-Areny and Webster, 1991) on the loading portion of Fig. 8 produced a good fit (mV = 0.235 × kPa + 0.5108; $R^2 = 99.3\%$, R^2 (adjusted) = 99.0%). R^2, the coefficient of determination, is the proportion of the total variation that is explained by the linear model (Blaisdell, 1993). This high value for R^2 suggests that the packaged sensor has very good linear performance even though the packaging may have altered this characteristic.

3.5 Repeatability

Repeatability is an extremely important feature of a pressure sensor. The performance of the repackaged sensor is satisfactory in terms of repeatability with a value of at most around 3.2% (Table 2) of the mean value.

Table 1. Temperature effect E (psi/°C)

Temp T (°C)	Pressure			
	0	1.569 psi (10.818 kPa)	3.322 psi (22.904 kPa)	6.828 psi (47.076 kPa)
25	−0.0068	−0.0023	−0.0114	−0.1322
37	−0.0027	−0.0121	−0.0215	−0.0362
45	0.0014	−0.0064	−0.0137	−0.0264

3.6 Stability

A constant load corresponding to 4.198 psi (28.94 kPa) was applied to the sensor during the stability test. The sensor output was recorded every 5 s for 144 s (Fig. 9). The results show that the stability of the repackaged sensor is between −2 and 3% FSS, while a majority of the readings are ±0.1%.

3.7 Creep

Creep is a percentage change increase in output over time while the applied pressure remains constant. By applying the same experiment pressure as in the stability test (i.e., pressure of 28.94 kPa), creep was evaluated. The sensor reading at $t = 0$ was 6.5 mV. The sensor output was recorded every hour for 12 h. The creep values after 3, 6 and 12 h were as follows:

$$\mathrm{Creep}_{3h} = (6.5 - 6.5) \times 100/6.5 = 0\% \quad (3)$$

$$\mathrm{Creep}_{6h} = (6.5 - 6.3) \times 100/6.5 = 3.1\% \quad (4)$$

$$\mathrm{Creep}_{12h} = (6.5 - 6.2) \times 100/6.5 = 4.6\% \quad (5)$$

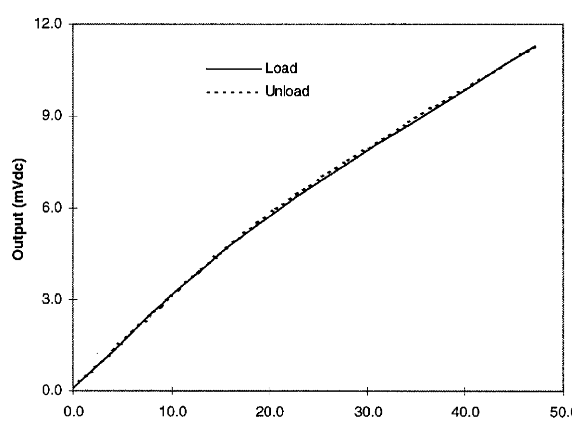

Fig. 8. Typical hysteresis test curve of the repackaged sensor

4
Comparison of performance characteristics with MEMS sensor

A comparison was made to evaluate the performance characteristics of the proposed flexible encapsulated package against the original MEMS pressure sensor. This enabled us to judge whether the packaging degrades the performance characteristics of the original MEMS pressure sensor. In addition, the proposed flexible MEMS package was compared against a commercially available conductive polymer sensor to determine its superiority.

4.1
Comparison of performance with original MEMS pressure sensor

The performance characteristics of the original MEMS were extracted from the Motorola pressure sensor device catalog (Motorola, 1994). As shown in Table 3, the repackaged sensor shows no sign of degradation of performance characteristics in terms of temperature dependency, hysteresis and zero pressure offset, and it shows only a slight percentage increase of full scale signal (FSS) in stability and possibly linearity. This result is rather remarkable considering the viscosity of the pressure transmission fluid that was used in the proposed approach, even though it was not fully optimized.

Table 2. Repeatability test of repackaged sensor

Applied pressure (kPa)	Mean of 10 samples (mV)	Standard deviation	Coefficient of variation (%)
4.792	1.53	0.05	3.157
10.859	3.37	0.05	1.433
16.927	4.82	0.04	0.875
29.062	7.49	0.03	0.422

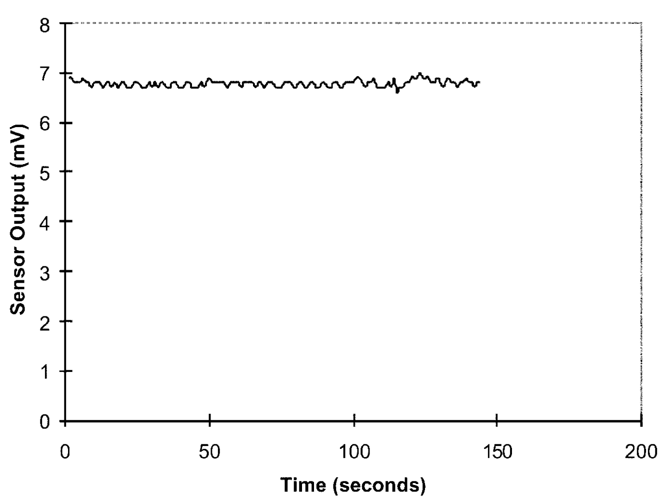

Fig. 9. Stability test result

4.2
Comparison of performance characteristics with commercially available conductive polymer pressure sensors

Olson (1991) has provided some typical performance characteristic curves of Interlink sensors that are used in biomechanical applications. As shown in Table 4, the proposed repackaged sensor has superior performance characteristics over the conductive polymer sensor in every category. In particular, the performance differences in hysteresis and linearity are very significant. There is practically no hysteresis (less than 2%) for our repackaged MEMS pressure sensor as compared to very strong hysteresis (25%) in the conductive polymer sensor systems.

The pressure output signal of our repackaged MEMS pressure sensor is basically linear as against the parabolic dependency of conductive polymer sensor systems.

Olson (1991) did not report any creep characteristics. However, Ferguson-Pell and Cardi (1992) have given creep values for short-term tests as shown in Table 5.

We measured the creep values over a longer period of time for the TEKSCAN sensors and they were as follows:

$$\mathrm{Creep}_{3h} = (7.17 - 5.06) \times 100/5.06 = 41.7\% \quad (6)$$

$$\mathrm{Creep}_{6h} = (10.10 - 5.06) \times 100/5.06 = 99.6\% \quad (7)$$

$$\mathrm{Creep}_{12h} = (11 - 5.06) \times 100/5.06 = 117.4\% \quad (8)$$

The creep data shown in Eqs. (6)–(8) for the commercial conductive polymer sensor are much larger than that of the corresponding creep data of the repackaged MEMS pressure sensor system. Over a 12-h period, the creep of commercial conductive polymer pressure sensor is over 100% versus less than 5% for the repackaged MEMS pressure sensor system.

Table 3. A comparison of the operating characteristics of the MEMS sensor (Motorola, 1994) and repacked sensor

	Raw sensor (Motorola, 1994)	Repackaged sensor
Temperature effect on offset (15–40 °C), psi/°C	±0.0057	Typical: 0.002 Max: 0.02 Min: −0.06
Hysteresis, % FSS	Min: −2.0 Max: 2.0 (Linearity + Hysteresis)	Max: 2.0
Zero pressure offset, mV	Min: −1 Max: +1	Max: 0.1
Stability, % FSS	±0.5	−1.8–1.5

Table 4. A comparison of the operating characteristics of the repackaged MEMS sensor and conductive polymer pressure sensor (Ferguson-Pell et al., 1976)

	Pressure sensor	
	Repackaged MEMS	Conductive polymer
Linearity	Good linearity with $R^2 = 99.3\%$	Highly nonlinear and it is basically a parabolic curve (Olson, 1991)
Temperature effect on offset (25–40 °C)	About 0.2–0.4%/°C (based on estimation from Fig. 6)	About 1%/°C (Olson, 1991)
Hysteresis, % FSS	Max: 2	Max: 25 (estimated from Fig. 9.2 of Olson, 1991)
Typical repeatability from sensor to sensor	6% (based on estimation from MPX7200 series (Motorola, 1994)	20% (Interlink, 1987)
Creep at 12 h	4.6%	117.4%

Table 5. % creep as reported in Ferguson-Pell and Cardi (1992) for two types of commercial sensors

Sensor tested	Applied pressure and time			
	50 mmHg		100 mmHg	
	2 min	10 min	2 min	10 min
TEKSCAN	12.5	19.4	10.6	16.6
FSA	3.3	4.4	2.1	7.1

5
Conclusion

The feasibility of a new flexible encapsulated packaging system for MEMS pressure for the biomechanical application has been conducted. Test data from a prototype has shown that the repackaged MEMS pressure sensor has minimal unwanted effect when compared against the performance characteristics of the original MEMS pressure sensor. Furthermore, the proposed repackaged sensor has much superior performance characteristics over the conductive polymer sensor in every category. In particular, the performance differences in hysteresis, linearity and creep are very significant. Therefore it appears that the current proposed flexible packaging approach for the MEMS pressure sensor holds good promise in the area of biomechanical application. Given that we could expect improvements in MEMS sensors, the repackaging of these sensors to cater to biomechanical applications shows much promise.

References

Aydin (2000) Product Technical Information: Dip-Molding. Vinyl Plastisol products manufacturing company of the Aydin Corporation. "http://www.aydin.com/moldeddevices/dip-molding_is.html"

Blaisdell EA (1993) Statistics in Practice. Orlando, Saunders College Publishing

Dymax (2000) Adhesives from Dymax Limited. http://www.dymax.com/homepage.htm

Esashi M; Minami K (1994) Packaged micro sensor, ETFA '94 1994 IEEE Symposium on Emerging Technologies and Factory Automation, pp. 30–37

Ferguson-Pell MW; Bell F; Evans JH (1976) Interface pressure sensors: existing devices, their suitability and limitations, in: Bedsore Biomechanics (Eds Kenedi R; Cowden JM; Scales JT), Baltimore, University Park Press

Ferguson-Pell M; Cardi M (1992) Pressure mapping systems. TeamRehab Report, pp. 26–32

Gross CM; Goonetilleke RS; Menon KK; Banaag J; Nair CM (1994) The biomechanical assessment and prediction of seat comfort, in: Hard Facts about Soft Machines (Eds Lueder R; Noro K), Taylor and Francis, pp. 231–253

Guirini J; Barry D; Chrzan J; Habershaw G; Lawrence M; Rosenblum B (1993) A preliminary analysis of the felted-foam dressing for the management of diabetic foot ulcerations using F-scan technology, Division of Podiatry. Deaconess Hospital/ Joslin Diabetes Centre, Boston, MA

Interlink (1987) Applications note IL-02: the Interlink force sensing resistor kit. Santa Barbara, CA

Koester DA; Markus KW; Walters MD (1996) MEMS: Small machines for the microelectronics age, MCNC MEMS Technology Application Center

Maudie T; Wertz J (1997) Pressure Sensor Performance and Reliability, IEEE Industry Application Magazine, pp. 37–43

Mehregany M; Huff MA (1993) Microelectromechanical Systems, IEEE Circuits and Devices, pp. 14–21

Motorola (1994) Pressure Sensor Device Data. DL200/D Rev 1. (MPX2300). Arizona, Motorola Semi-conductor Products Sector

Mueller MJ (1992) Etiology, Evaluation and Treatment of the Neuropathic Foot, Critical Reviews in Physical and Rehabilitation Medicine 3(4): 289–309

Olson J (1991) Conventional Pressure Sensors, in: Prevention of Pressure Sores (Ed Webster JG), Adam Hilger, pp. 121–141

Pallas-Areny R; Webster JG (1991) Sensors and Signal Conditioning, Wiley

Plastisol coatings (2000). http://www.pilotrock.com/cpt_3.html

Trimmer WS (1997) Micromechanics and MEMS, New York, IEEE

Varadan VK; Varadan VV; Bao XQ (1996) MEMS for measuring deflection, acceleration, and de-icing of a helicopter, in: Microelectronic Structures and MEMS for Optical Processing II (Eds Motamedi ME; Bailey W), Proc. SPIE 2881, pp. 208–218

Walsh S; Carr B; Mados H; Narang DS (1996) Commercializing MEMS – Too fast or too slow, in: Microelectronic Structures and MEMS for Optical Processing II (Eds Motamedi ME; Bailey W), Proc. SPIE 2881, pp. 12–24

Webster JG (1991) (Ed), Prevention of Pressure Sores, Adam Hilger

Wilson M (1992) Measuring in-shoe pressures to help predict underfoot comfort. SATRA Bulletin, pp. 6–8

Young CR (1993) The F-Scan system of foot pressure analysis. Clin Pediat Med Sur 10(3): 455–461

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