A capacitance pressure sensor using a phase-locked loop

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Abstract—We are using a Hercules (model #F4-4R, 100 psi) pressure sensor to measure the pressure between the foot and shoe. An interface circuit converts the capacitance change into voltage. Over the pressure range from 0 to 1300 kPa, the capacitance changes from 275 to 580 pF. A 555 timer circuit converts the capacitance into a frequency range from 30 to 63 kHz. A phase-locked loop (PLL) converts this frequency to voltage from 0 to 5 V, which is then filtered using a first-order, low-pass filter, having a corner frequency of 20 Hz to reduce the ripple to 10 mV. The sensor’s hysteresis is about 8 percent at 40 degrees Celsius (C) and 12 percent at 20 degrees C. The sensor has a maximal nonlinearity of 8 percent and a worst-case nonrepeatability of 7 percent. Its temperature coefficient is −0.147 percent per degree C. Its spatial sensitivity decreases nonlinearly from 1 to 0.17 from the center towards the periphery. The sensitivity of the system is 2.77 mV/kPa and the temperature drift is +0.53 percent per degree C. We monitor the pressure at 7 locations under each foot (the rear and the front heel, great toe, and 4 of the 5 metatarsal heads). A portable data-acquisition system permits continuous monitoring for 7 minutes. Test results for pressure distribution for normal walk and run are presented. Results are useful when studying normal and abnormal gait, and for possibly providing feedback (sensory substitution) to diabetic patients with insensate feet in order to help them dynamically adjust pressure distribution under their feet.

Key words: foot pressure, gait, insensate feet, phase-locked loop, pressure sensors.

INTRODUCTION

Measurement of the pressure between the foot and the shoe requires a thin sensor. The Hercules pressure sensor (model #F4-4R, 100 psi [or 690 kPa], Allegheny Ballistic Laboratory, Rocket Center, WV) has a circular cross section with a diameter of 17 mm and thickness of 2.4 mm. The pressure applied to the sensor increases the capacitance from 275 pF at 0 kPa to 580 pF at 1300 kPa. Figure 1 shows the transverse section of the sensor. The sensor is constructed with 4 layers of mica as a dielectric medium, sandwiched between 5 layers of corrugated metal. It is coated with black rubber for protection. Applied pressure flattens the corrugations and moves the metallic plates closer to each other, thus increasing the capacitance. The sensor is not damaged by large overloads, withstanding pressures up to 3000 kPa, because flattening of the corrugations does not yield the metal. Figure 2 shows its capacitance change with pressure. The response is nearly linear up to 500 kPa. The sensor shows a maximal nonlinearity of 8 percent up to a pressure of 1300 kPa, which is the maximal plantar pressure. Figure 3 shows that the spatial sensitivity of the bare sensor decreases nonlinearly from 1 to 0.17 from the center towards the periphery.

The circuit

The capacitance change of the sensor varies the frequency of a 555 timer. The next stage converts the frequency to voltage, which is achieved either by
Figure 1.
Four layers of mica sandwiched between 5 corrugated metal layers, give 4 capacitors in parallel.

Figure 2.
Capacitance increases with pressure.

Figure 3.
Spatial sensitivity decreases with radial distance. The ordinate shows the increase in capacitance over no-load value for a 5-N force applied at a point. The increase when a 5-N force is applied over the entire sensor area is 8.6 percent.
using a frequency-to-voltage converter (3) or by using a phase-locked loop. The voltage is filtered by a low-pass filter and then offset and amplified to obtain the required output range. Figure 4 shows the entire circuit diagram excluding the amplification stage.

The frequency of the timer is:

\[ f = \frac{0.722}{RC} \]

where \( C \) is the capacitance of the sensor. The value of \( R \) is chosen to give the frequency output from 30 to 63 kHz. The 1-kΩ resistor connected to pin 3 ensures that the output voltage equals approximately \( V_{cc} \) (1). The 10-nF capacitor, connected to pin 5, prevents any noise from altering the pulse width (1).

Conventionally, a phase-locked loop (PLL) is used to track a signal’s frequency coherently and recover it from noise. Figure 5 shows the basic building blocks of the PLL (2). The multiplier multiplies the input voltage of the timer by the output voltage of the voltage-controlled oscillator (VCO). The VCO has a natural frequency that can be varied by changing the dc component of the input voltage. Let

\[ V_{in} = A_{in} \sin(2\pi f_{in}t) + \text{noise} \]

and

\[ V_r = A_r \sin(2\pi f_r t + \beta) \]

where \( V_{in} \) is the output voltage from the timer with amplitude \( A_{in} \) and frequency \( f_{in} \) and \( V_r \) is the output voltage from the VCO with amplitude \( A_r \) and frequency \( f_r \). The product is:

\[ V_{in} \cdot V_r = (A_{in} \cdot A_r / 2)[\cos(2\pi(f_{in} - f_r)t - \beta] - \cos[2\pi(f_{in} + f_r)t] + \text{noise} [A_r \sin(2\pi f_r t + \beta)] \]
The low-pass filter eliminates the terms containing higher frequency components. After low-pass filtering:

$$V_{in} \cdot V_r = (A_{in} \cdot A_r/2) \cos\left[2\pi(f_{in} - f_r)t - \beta\right]$$

A feedback loop varies the frequency, $f_r$, of the PLL. It gets locked when $f_r$ equals $f_{in}$ and yields a dc output voltage of magnitude $(A_{in} \cdot A_r/2)\cos(\beta)$. The dc voltage which generates a frequency $f_r (= f_{in})$ from the VCO corresponds to the output frequency of the 555 timer.

The PLL has a provision for selecting the locking range. The range from $f_{\min}$ to $f_{\max}$ gives the voltage variation from 0 to $V_{cc}$ (= 5 V) where

$$f_{\min} = 1/[R_2(C_1 + 32 \text{ pF})]$$

and

$$f_{\max} = f_{\min} + 1/[R_1(C_1 + 32 \text{ pF})]$$

The values of $R_1$, $R_2$ and $C_1$ are chosen to obtain the locking range from 22 to 63 kHz, which covers the entire frequency range of the timer output (30 to 63 kHz). The two frequency ranges should be the same to obtain an output variation from 0 to 5 V. We made our frequency range slightly larger to allow some variation in the characteristic of the sensor and the circuit components due to temperature change. An amplification stage is not required in the PLL for gain setting (unlike the frequency-to-voltage converter). Hence, it results in fewer active components.

In our design we did not use a low-pass filter after the multiplier stage. The problem in our application when using a low-pass filter in the PLL is that the locking would occur only when the frequency difference $(f_r - f_{in})$ is less than the corner frequency of the filter. If the corner frequency of the filter is raised in order to increase the locking range, then another filter at the output stage is required to filter out the ripple. Thus, the conventional use of a PLL for frequency-to-voltage conversion would result in more components. We elected to omit the low-pass filter inside the PLL, and instead filtered the resulting output containing high-frequency components at the output stage using a first-order, passive, low-pass filter with a corner frequency of 20 Hz. We chose a filter corner frequency of 20 Hz because lower corner frequencies caused a reduction of peak pressures measured under the foot. The output ripple is less than 10 mV peak-to-peak (for a full-scale of 4.5 V), which corresponds to 0.22 percent. Thus, the ripple error is within the resolution of the 8-bit A/D converter (0.4 percent) that is used to acquire the output voltage of the circuit. To achieve lower ripple, we could raise the frequencies of the timer and PLL stage, rather than reducing the corner frequency of the filter. At lower timer frequencies, lower ripple could be obtained by using higher-order filters.
RESULTS AND CONCLUSIONS

We studied the sensor characteristics of various sensors using the PLL circuit. The results for the sensor with worst-case hysteresis are presented. Figure 6 shows the calibration curve of the sensor for increasing and decreasing values of applied pressure at different temperatures and at different maximal pressures for a calibration time of 5 s. At 20 degrees C, the sensors show hysteresis up to 12 percent for a calibration time of 30 s. The hysteresis reduces to 8 percent at 40 degrees C. The hysteresis at 20 degrees C for a calibration time of 5 s is 14 percent. The temperature coefficient measured for the sensor was $-0.147$ percent per degree C. The worst-case nonrepeatability of the sensor is 7 percent. Each sensor cost $150.
Figure 7 shows the calibration curves for increasing and decreasing pressures at 20 degrees C for a calibration time of 5 s. The sensitivity of the system is 2.77 mV/kPa, and the temperature drift of the PLL circuit is +0.53 percent per degree C. The current consumption of the system is 40 mA (for 14 channels).

Figure 8a shows the pressure distribution at 7 peak pressure points under the left foot while walking. Figure 8b shows similar results on the same subject while running. The maximal peak pressure of 600 kPa was observed under the front heel while walking and a maximal peak pressure of 700 kPa was observed under the second metatarsal head while running.

Error analysis

The worst-case error in pressure measurement is obtained by adding up the individual errors. The error analysis is done for pressure measurement up to 1300 kPa. Following are the factors contributing to the total error:

1. **Nonrepeatability:** Although the worst-case nonrepeatability of the sensor is 7 percent, the maximal error due to nonrepeatability of the sensor can be reduced to 3.5 percent by calibrating the sensor just before and after the walking/running test. The average of the two calibration tables is used as the final calibration table.

2. **Output voltage ripple:** The final voltage output contains a 10-mV peak-to-peak ripple. The resolution of the 8-bit A/D converter is $5/2^8=20$ mV. The ripple can give a maximal error of 1-bit, which would correspond to 0.4 percent ($=20$ mV/5 V) error in the final pressure measurement.

3. **Nonlinearity:** In spite of using a calibration table for voltage to pressure conversion, nonlinearity of the sensor gives an error in measurement due to the bit resolution. The bit resolution of the system in the pressure range 0-500 kPa and 500-1300 kPa is 3.58 kPa and 25.6 kPa respectively. Hence, for pressure measurement at about 1300 kPa, the maximal error in pressure measurement is about 2 percent ($=25.6/1300$).

4. **Hysteresis:** The maximal hysteresis at 40 degrees C is 8 percent. Since the body temperature is about 37 degrees C, all the calibrations are done at 37 degrees C. The maximal error due to sensor hysteresis (and system time constant) is about 8 percent.

5. **Temperature effect:** The temperature coefficient of the sensor is $-0.147$ percent per degrees C. The sensor temperature during the test is about 37 degrees C, which remains constant throughout the test, as the sensors are always in contact with the body temperature. The temperature drift of the system is 0.53 percent per degrees C. A maximal variation of about 5 degrees C can be expected.
Figure 8a.
Pressure distribution in kPa at 7 peak pressure points under the right foot for a normal subject while walking. (A. rear heel, B. front heel, C. 1st metatarsal head, D. 2nd metatarsal head, E. 4th metatarsal head, F. 5th metatarsal head, G. great toe.) Timing marks at 1.05 second intervals.

Figure 8b.
Pressure distribution in kPa at 7 peak pressure points under the right foot for a normal subject while running. (A. rear heel, B. front heel, C. 1st metatarsal head, D. 2nd metatarsal head, E. 4th metatarsal head, F. 5th metatarsal head, G. great toe.) Timing marks at 1.05 second intervals.
during the test period which would give the maximal error of 2.7 percent in the measurement of the pressure.

The other minor sources of error in pressure measurement are the mismatching of the mechanical compliance of the sensor and that of the insoles, and locating the peak pressure points under the foot for insole construction. The maximal possible error in peak pressure measurement is about 17 percent.

DISCUSSION

Zhu et al. (6) reported maximal peak pressure of 700 kPa during the normal walking of normal subjects using conductive polymer pressure sensors. Thus, the results obtained using our system are in agreement with the results obtained by Zhu et al. (6). Cavanagh et al. (2) reported a maximal peak pressure of 1900 kPa for barefoot measurement at ulcer locations for diabetics. Conductive polymer pressure sensors have very poor sensitivity and bit resolution at higher pressures which result in a large error in measurement of high pressures. Our system has a distinct advantage when measuring pressures over 1000 kPa because the Hercules sensor has better linearity over the range of 0 to 2000 kPa than conductive polymer pressure sensors.

The temperature sensitivity of our system is comparable to that which uses conductive polymer sensors. However, our sensors cost more and the circuit is more complex.

Kothari et al. (4) described the method for calibration and insole construction with the Hercules sensors.

Another proposed circuit for the measurements from capacitive sensors uses counters to count each frequency for 3 ms/sensor (42 ms for 14 channels), which could give about a 24-Hz (=1/42 ms) sampling rate. The microprocessor controls the multiplexer, and selects one channel at a time. To obtain a higher sensitivity, the frequency should be high enough to include up to 256 counts and the microprocessor should reset the lower frequency (at 0 kPa) to zero before beginning the counts.

APPLICATIONS

We have evaluated commercial capacitive sensors for measuring the pressures between the foot and shoe. We monitor the pressure at 7 high peak pressure areas under each foot (the rear and the front heel, great toe, and 4 of the 5 metatarsal heads). Results should prove useful in the study of normal and abnormal gait. A portable data-acquisition system permits monitoring for 7 minutes.

A pressure sensor system could possibly provide sensory feedback substitution to diabetic patients who have insensate feet in order to help them improve the pressure distribution under their feet (5). Such patients do not receive adequate information about the pressure distribution under their feet. A repetitive high pressure exerted at one location under the foot may result in tissue breakdown and ulcer formation without their knowledge. Ulcer progression can result in the amputation of the foot. Thus, sensing pressure distribution under the foot and providing information back to the patient could be useful.

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REFERENCES