Calibration of sensors for reliable radio telemetry in a prototype flexible wound monitoring device

Nasir Mehmood a,⇑, Alex Hariz a, Sue Templeton b, Nicolas H. Voelcker c

a School of Engineering, University of South Australia, Adelaide, SA 5001, Australia
b Royal District Nursing Service, Adelaide, SA 5035, Australia
c Mawson Institute, University of South Australia, Adelaide, SA 5001, Australia

A R T I C L E   I N F O

Keywords:
Chronic wounds
Wireless sensing system
Sensor calibration
Sensor characterization
Chronic wound management

A B S T R A C T

With the growing financial burden of chronic wound management in clinics and hospitals, it is high time to devise wireless wound monitoring devices to be placed within wound dressing for continuously sensing the wound milieu. Sensors are a vital part of such monitoring system, enabling it to sense and measure variations in wound parameters. This paper describes the calibration, characterization and real-time testing of selected temperature, moisture, and pressure sensors deemed suitable for wound monitoring applications. The sensors were chosen on the basis of small size, non-invasiveness, reliable performance, low power consumption and their suitability for placement within wound dressings. All selected sensors were first individually calibrated and characterized using commercially-available software and measurement tools, they are then collectively interfaced to a custom-designed flexible radio-frequency transmitter device for real-time performance measurement within a clinical-grade wound bandage. Experimental results on a mannequin leg have validated the low-power operation, and reliable sensing and data transmission capabilities. The nominal measurement resolutions obtained for temperature, moisture, and pressure were 0.15 °C, 0.85–5 %RH, and 0.05–0.56 mmHg, respectively.

1. Introduction

Modern living is overwhelmed by the ever increasing costs of treatment and care for non-healing chronic wounds such as venous leg ulcers, diabetic foot ulcers etc. [1]. Australia alone spends almost $2.5 billion annually on wound management with about 4000 limb amputations per year [2]. Chronic wound management also creates a burden in terms of nursing care and hospitalization time. During 2010, almost 6.5% of total diseases and 31% of nursing activities were attributed to wound management in Australia [3]. The USA has spent almost $3 billion during 2006–07 [4], while the UK has spent £26 million during 2008–09 [1] for the treatment of non-healing chronic wounds. Throughout the world, this cost is on the rise with an estimate of over $20 billion in the near future [5].

Understanding the healing trajectory of wounds is important to mitigate the effects of non-healing wounds on society and on the economy. Wound healing is a complex biochemical process divided into four distinct phases (i) haemostasis, (ii) inflammation, (iii) proliferation, and (iv) remodelling [4]. A wound is properly healed up if all of these phases occur in their natural sequence. However, if one or more stages are prolonged for any reason, the healing process is impaired, resulting in delayed or non-healing wounds [6], [7]. Many factors contribute to impaired healing, including oxygenation, age, gender, infection, diet, hormones, stress, diabetes, medications etc. [4].

Covering a chronic wound with suitable dressing is the most economical and effective way of treatment [8]. For some wounds, appropriate pressure bandages are also applied to absorb the wound exudate and expedite healing [9]. Healing rate may be increased by providing external moisture to the wound site through moisture-retentive dressings [10]. When healing progresses, wound exudate gradually diminishes. At this stage, a special type of dressing is required (e.g. foams, hydrogels or hydrocolloids) to maintain the required moisture level at the wound site. Otherwise, the wound may not heal due to hypoxia [11]. Furthermore, wound-site temperature and pH level may also be used with other parameters, such as moisture level, nitric oxide concentration etc., to predict the status of wound healing [12], [13], and hence the proper time to change the dressings may be determined. However, in current clinical practice, most of the dressings are changed routinely without any scientific reason or
clinical evidence which complicates the problem of impaired wound healing.

The literature shows scarce efforts made for the development of diagnostic tools and devices with potential for effective monitoring of chronic wounds. Matzeu et al. [14] demonstrated an RFID-based skin temperature monitoring system obtaining a 0.2 °C temperature measurement accuracy. A flexible platinum-based miniaturized temperature sensor is proposed by Moser and Martin [15] to operate within 0-400 °C range. However, the sensor has not been used in wound dressings. McColl et al. [16] have developed an impedance sensor-based moisture monitoring system for wound dressings. Based on this principle, Ohmedics [17] has developed a commercial moisture monitoring device called WoundSense®. However, the sensing system is not designed to stay within the dressing for continuous moisture measurement. To date, no device exists for continuous monitoring of sub-bandage pressure, though some sub-bandage pressure sensors, such as Kikuhime®, are in practice, but they have very limited clinical application. Miniaturized pressure sensors have been designed and used for other applications such as intracranial pressure (Codman® [18] and Mejzlik [19]), intraocular pressure (Ning et al. [20]), spinal plates pressure (Sauscer et al. [21]) and for general in-vivo applications (Clausen et al. [22], Willlyan et al. [23] and Hill et al. [24]).

For a wound monitoring system, calibration and characterization of sensors is important in order to sense accurate information beneath the dressing. Incorrect sensor measurements will negatively impact on clinical decisions about the wound under observation. In our recent review article [25], we have highlighted the gap in potential use of modern sensors and wireless technology for wound monitoring applications. In this paper, we present methods for calibration and characterization of temperature, moisture, and pressure sensors deemed suitable to be interfaced with a wireless telemetric sensing system. The sensors were carefully selected considering their small size, low power operation, flexibility and minimal invasiveness to the wounds. The calibrated sensors were interfaced with the developed prototype sensing system, and the performance of the system was tested by placing the system and sensors under a compression bandage on a mannequin leg.

2. Methods and materials

2.1. Temperature sensor calibration

Monitoring temperature at the wound site is important in the sense that temperature rise gives an indication of a possible infection [26]. Although a number of miniature temperature sensors are commercially available, we have chosen LM94021B (Texas Instruments) temperature sensor, due to its compact size and configurable operation. This sensor can operate from a 1.5–5.5 V supply voltage with a typical 9 µA current consumption. Its gain can be controlled through a combination of gain select pins (G50 and G51). The size of this sensor is 2.15 mm × 2.40 mm × 1.1 mm (L × W × H), and the sensor has a nominal accuracy of ±1.5 °C over the temperature range 20–40 °C [27]. For testing and calibration purposes, a small 5 mm × 5 mm standard printed circuit board (PCB) was prepared (Fig. 1(a)).

The datasheet of LM94021B sensor contains the transfer table for various gain settings at 5 V supply voltage. We have tied [G51:G50] pins to logic 1 to obtain the highest gain of 13.6 mV/°C. Entering this transfer table into the MATLAB curve-fitting tool (cftool), the following mathematical relationship was obtained between temperature and output voltage.

\[ T(°C) = -0.07269 \left( \frac{°C}{mV} \right) \times V_{out}(mV) + 191.2(°C) \]  \hspace{1cm} (1)

The transfer function in Eq. (1) is linear with a 5 V supply voltage. (Fig. 1(b)). The temperature sensor was then calibrated over the 20–50 °C range using the Digitech QM 1535 multimeter.

2.2. Moisture sensor calibration

2.2.1. Active moisture sensor

An active sensor requires an excitation signal to produce an output. This signal may be a constant voltage or a constant current. In our system, we have used Honeywell’s HIH4031 piezoelectric moisture sensor [28]. The sensor (Fig. 2(a)) has the dimensions of 8.59 mm × 4.2 mm × 3.5 mm (L × W × H) and operates at 5 V supply voltage. Its measurement range is 0–100 %RH with normal accuracy of ±3.5 %RH. Typical current consumption of this sensor is 200 µA. The datasheet of HIH4031 sensor shows a dependency of output voltage on temperature.

For calibration and characterization, we used two moisture sensors with one sensor acting as a reference. These sensors were placed on opposite sides of a mannequin leg (Fig. 2(b)) and then covered with a commercial compression bandage Coban™ 2 (Fig. 2(c)), which was wrapped around the leg. Both sensors were powered up using a 5 V supply and their output data was sampled at 100 mHz using National Instruments data acquisition card NI DAQ 6009. After about 30 s, distilled water was sprayed near the dressing portion above the second moisture sensor. The experiment was conducted at 22 °C for about 6 h and the acquired data was plotted using Microsoft Excel (Fig. 3). The graph in Fig. 3 confirms that the Honeywell moisture sensor HIH4030 is suitable for wound monitoring applications. The following mathematical equations (Eq. (2) supplied by the manufacturer) [28] were used to calculate the moisture level at any temperature.

\[ \%RH = \frac{161.2903 \times V_{out} - 0.16}{1.0546 - 0.000216 \times \text{Temperature}(°C)} \]  \hspace{1cm} (2)

Moisture Level = |%RH(sensor2) − %RH(sensor1)| \hspace{1cm} (3)

2.2.2. Passive resistive moisture sensor

A passive resistive moisture sensor operates without any excitation signal and produces a change in resistance or impedance between its terminals in response to a change in moisture. We have used Multicomp’s HCZ-D5 device, which measures 10 mm × 5 mm × 0.5 mm (L × W × H) in size. It consists of two inter-digitized terminals on a substrate made of alumina (Fig. 4(a)). The rated humidity accuracy of this sensor is ±5 %RH, slightly worse than the accuracy of the active moisture sensor. However, its small size and delicate architecture makes it quite suitable for wound monitoring applications.

For our application, we have used the HCZ-D5 sensor as a variable moisture-sensitive resistor in a differential operational amplifier circuit (Fig. 4(b)). This circuit uses LM358 operational amplifier IC that is capable of producing rail-to-rail full swing output voltage. The sensor’s datasheet provides the value of its resistance as a function moisture level (20–90%RH) for the temperature range 5–60 °C. We have used these resistance values with the following equation (Eq. (4)) obtained through circuit analysis to determine the output voltage expression for the interface circuit in Fig. 4(b).

\[ V_{out} = \frac{R_{out} - R_{sens}}{R_{out} + R_{sens}} \times V_s \]  \hspace{1cm} (4)

where, \( R_{At} \) is the resistance between positive input terminal of LM358 and ground, \( R_{sens} \) is the variable resistance of the HCZ-D5 sensor and \( V_s \) is the supply voltage. The values of \( R_{sens} \) and \( V_{out} \)
are plotted against the moisture level for a selected temperature range of 25–40 °C (Fig. 4(c)).

Fig. 4(c) also shows the variation of sensor’s characteristic impedance and hence the output voltage with temperature. However, as the sensor will be placed within wound dressing where the nominal human skin temperature is expected to be around 35 °C, so the curve associated with this temperature is of particular interest for our system. Using the MATLAB curve-fitting tool, the following polynomial expression for the moisture level at 35 °C was obtained.

\[ M = p_1 v^3 + p_2 v^2 + p_3 v + p_4 \]  

(5)

where, \( p_1 = 1.228 \times 10^{-8} \), \( p_2 = -4.178 \times 10^{-5} \), \( p_3 = 0.05692 \) and \( p_4 = 20.2 \), while \( M \) is the moisture level in %RH and \( v \) represents the output voltage in mV. The HCZ-D5 sensor was tested for real-time moisture measurements within the wound dressing by placing it on a mannequin leg and powering it through the interface circuit (Fig. 5(a)). The output voltage was measured by taking some 8000 samples using National Instruments’ data acquisition card NI DAQ 6009 with a sampling rate of 1 Hz. It is evident from the graph (Fig. 5(b)) that this moisture sensor is capable of reliable moisture measurements in a wound environment. However, the initial response of passive sensor was slower than that of active sensor and it required additional interface electronics for its proper operation. Furthermore, due to the insensitivity of the device for moisture levels lower than 20 %RH and above 90 %RH (as the manufacturer did not supply impedance data for this range), the sensor can only measure moisture within the range 20–90 %RH.

2.3. Pressure sensor calibration

Pressure therapy via compression bandaging is an established clinical technique used to improve the healing rate of chronic wounds such as venous leg ulcers [29,30]. The pressure range varies from 14 to 40 mmHg depending on the type, volume and location of the wound [31]. While effective for treatment of chronic wounds, pressure therapy may be counter-productive if incorrect pressure is applied. Quite often, clinicians and nurses rely on their experiences to judge the adequacy of the sub-bandage pressure, as there is no objective mechanism to monitor the sub-bandage pressure during and after application of the bandage. Hence, a real-time sub-bandage pressure monitoring system may improve the efficacy of the pressure therapy and translate into improved chronic wound healing [30].

A number of sensors are available for measuring pressure in a gas or liquid [25,13], though, the majority of them is unsuitable for monitoring of sub-bandage pressure in compression bandages. For this application, the sensor needs to be flexible, non-invasive and of a minimal size (e.g. a few millimetres in diameter). After a thorough search, we have found that Interlink Electronics’ FSR series flexible force sensors [32] could be used as pressure sensors within a compression bandage. For a prototype system, we chose the FSR402 sensor with a size of 13 mm diameter and a 56 mm long stem (Fig. 6(a)).

Since the sensor is piezoresistive, it requires additional interface circuitry to operate it as a variable resistor in a differential amplifier configuration (Fig. 4(b)). A force applied at the sensor tends to reduce the impedance of the sensor which causes more current to flow through the circuit, resulting in a detectable change in output voltage. In order to obtain the transfer function of the FSR402 sensor, we performed an experiment (Fig. 6(b) and (c)), where the sensor was placed on a mannequin leg along with a commercial pneumatic pressure meter HPM-KH-01 for validation of pressure measurements. The pressure bandage (Fig. 2(c)) was gradually wrapped around the mannequin leg and pressure readings were taken from the pneumatic meter, while at the same time the output voltage was measured from the FSR402 sensor for pressure values ranging
from 1 to 40 mmHg in 1 mmHg interval. The data acquired in this experiment was plotted (Fig. 7) and the following empirical relation was obtained using the MATLAB curve-fitting tool.

$$P(\text{mmHg}) = p_2 V^2 + p_1 V + p_0$$

where, \( p_2 = 6.814 \times 10^{-6} \), \( p_1 = 1.254 \times 10^{-3} \) and \( p_0 = 1.783 \) and the voltage is expressed in mV.

3. Experiments using a compression bandage

For real-time validation of all the calibrated sensors working together, we designed a prototype flexible wireless transmitter (9.7 cm x 4.7 cm) and a receiver (4 cm x 4 cm), both operating at the radio frequency (RF) of 2.4 GHz using the IEEE 802.15.4 ZigBee protocol. The sensors were interfaced with the transmitter circuit using additional interface circuits designed for impedance and voltage conditioning. Both the transmitter and the receiver use Atmel’s Atmega128RFA1 integrated circuit as transceiver, with all other required components for their proper operation. The temperature, moisture, and pressure sensors were mounted on the mannequin leg at proper positions, and then an elastic compression bandage (Coban™2) was wrapped over them (Fig. 8(a) and (b)). Distilled water was sprayed externally over the surface of the bandage. The data acquired by the sensors in real-time was transmitted to the nearby receiver at an interval of 5 s. The receiver was
attached to the computer’s serial port. On reception of the full data packet, the receiver first saved and then sent the captured data to the PC serial port (Fig. 8(c)). A graphical user interface (GUI) was designed in Visual Basic to display the real-time information on temperature, moisture, and pressure, along with capture-time values (Fig. 9).

The GUI is split into five distinct regions of information i.e. control panel, file options, current values, graphical plots, and channel information. The serial port can be controlled using ‘control panel’ by selecting the available serial port (COM port), and by selecting the baud rate. Every time that the ‘Acquire’ button is pressed by the user, an ASCII character ‘U’ (equivalent to 55 H) is transmitted first as a synchronization protocol for a correct serial communication. After receiving the validated protocol character, the receiver sends the received sensor data over the serial port to the PC. The recent data is displayed in the ‘current values’ section as well as displayed on the ‘data plots’ region with the capture time shown along horizontal axis. All the captured values of temperature, moisture, and pressure are displayed with colour coding to distinguish them from each other. All the acquired data in a session is also printed in a separate text file for off-line analysis. The GUI also displays the transmitting device identification, frequency channel, and the received signal energy.

4. Analysis and discussion

The real-time performance of the selected sensors was tested extensively and monitored with the designed prototype RF transceiver system. The transmitted and received data was validated using commercial temperature, moisture, and pressure meters. The average errors in temperature, moisture and pressure measurements in the system were ±1.5 °C, ±3 %RH and ±2 mmHg, respectively. The measurement resolutions obtained for temperature, moisture, and sub-bandage pressure were 0.15 °C, 0.85–5 %RH, and 0.05–0.56 mmHg respectively using a 10-bit analog to digital converter (ADC) within the RF transceiver. The temperature resolution is more than sufficient to detect temperature variations in human skin, which range from 32 °C to 37 °C under normal conditions [14]. The moisture and pressure measurement resolutions are not uniform owing to nonlinear characteristics of respective sensors. In actual scenario, the temperature sensor will be placed on periwound skin and not directly on top of wound. The moisture sensor could be placed inside a foam dressing (e.g. Mepilex™, Allevyn™) over the wound, while the pressure sensor could be placed near ankle to monitor sub-bandage pressure. The measurements with the temperature and the moisture sensors were location independent, while the sub-bandage pressure measurements showed dependency on the sensor location on the mannequin limb, possibly due to changes in applied pressure on to uneven surface morphology around the limb. However, the optimum position of the pressure sensor can arguably be determined during clinical experiments.

Both the active and passive moisture sensors (HIH4031 and HCZ-D5) showed reliable and repeatable performance during experiments with and without the applied bandage. The active sensor showed a faster response as compared to that of the passive sensor. Moreover, it does not require any additional interface electronics for its operation except for an excitation voltage signal. However, the thickness (3.5 mm) of this sensor may be of concern.
Fig. 7. Transfer function of FSR402 sensor with $R_M = 1$ k$\Omega$ used in the interface circuit. The graph shows a nonlinear relationship between the applied pressure and the output voltage.

Fig. 8. (a) The sensors attached to the flexible prototype wireless sensing system mounted on mannequin leg before the application of bandage. (b) The wireless sensing system after the application of compression bandage. (c) Photo of the matched RF receiver and serial interface control circuit.

Fig. 9. A screen shot of the GUI developed to display the measured parameters acquired by the sensors. The GUI displays the current and previous values of temperature, moisture and sub-bandage pressure in graphical, textual and file format.
during its use in a compression bandage on a wound. Although this problem can be mitigated by placing the sensor distal to the wound site, it may still cause residual pain if not used with care.

The passive moisture sensor has the advantage of smaller size (10 mm × 5 mm × 0.5 mm) and therefore can be placed anywhere under the bandage. However, it gave a slower response to moisture within the bandage. It also required an additional interface circuit for its normal operation, which increased the size of the sensing system, though with minimal effect on power consumption.

The prototype sensing system has also been studied for power consumption and data transmission range. With one temperature sensor, two active or passive moisture sensors, and one pressure sensor interfaced, the average current consumption of the transmitter was 20 mA (measured using the Digitech Q.M1535 multimeter) at maximum transmitter output power (i.e., 3.5 dBm). The current consumption rose to a maximum of 24 mA only when the pressure sensor was pressed to its full range, while the other sensors had a minimum effect on current consumption during their normal and peak operation. The transmission range depended on the effective transmitter power, antenna parameters, and RF matching network. With the maximum transmitter power selected and using the recommended matching RF network, a transmission range of 4–5 m was achieved in an unobstructed environment. With all other factors remaining constant, the transmission range decreased when the transmitter and receiver were not facing each other, or when there was an obstacle between them. However, considering the clinical environment, the range of 4–5 m was deemed reasonable for reliable observations.

5. Conclusions

Evidence-based monitoring of chronic wounds, such as venous leg ulcers and diabetic foot ulcers, has drawn substantial attention because of soaring costs of their treatment. In current clinical practice, clinicians and nurses rely on their subjective judgement when applying or changing dressings on chronic wounds. Quite often, these dressings are changed unnecessarily without any scientific evidence about the state of wound. On the other hand, there is no objective mechanism to apply correct sub-bandage pressure on the wound site. These activities not only consume nursing time, but also create an unnecessary burden on healthcare budget. There is an opportunity to utilize modern sensors, biosensors, and wireless transmission technology to report to the patients and clinicians on the status of wound parameters from within bandages or dressings. Sensors play a vital role in extracting accurate information of wound parameters. In this paper, we have proposed and implemented calibration and characterization methods for carefully-selected temperature, moisture, and pressure sensors for reliable and accurate information acquisition of wound parameters. The calibration curves and mathematical expressions were obtained for each sensor using commercially-available software and measurement instrumentation. The performance of the sensors was validated by placing them on a mannequin limb with a compression bandage wrapped around them. We also designed a prototype wireless flexible sensing system, and a matched receiver to demonstrate the real-time performance of the calibrated sensors. The received data can be visualized on a computer screen. The average current consumption of the sensing system was measured as 20 mA, with all the sensors interfaced simultaneously. The system reliably works within a range of 4–5 m in an open environment.

Future work will include the design of the entire wireless wound sensing system (i.e., sensors, interface circuits and the RF transmitter) on a more compact flexible printed circuit material to be placed inside the dressing for a longer period of time.

Conflict of interest

The authors declare that there is no conflict of interest.

Acknowledgment

We are thankful to the Wound Management Innovation Cooperative Research Centre (CRC) Australia for sponsoring this research work (Project No. 2.10).

References


