Analysis of Tongue Pressure Sensor for Biomedical Applications

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Abstract— **The tongue is an important muscle and the contact with the hard palate during the articulation of a syllable or during swallowing is fundamental. Patients who have had cerebrovascular or other neurological disorders may have impaired speech and swallowing problems due to decreased ability to control the tongue. In this work, a device with the aim of providing a non-invasive aid for the rehabilitation is described. The proposed device has been designed with the purpose of measuring the tongue pressure on the palate directly in the oral cavity and transmitting the data wirelessly. The device is minimally invasive, because no cable is used to connect the pressure sensor placed in the oral cavity with the reading unit placed outside. A first prototype was developed and preliminary testing data for the analysis of the sensor behavior have been performed. A specific experimental setup has been designed and realized and a testing protocol has been defined and adopted. The obtained preliminary experimental results show a wide measurement range, up to 100 kPa. Application fields of this device are the treatment of people with swallowing or phonetic disorders.**

Keywords— Intraoral sensor, implanted sensor, wireless sensor, physiological monitoring, pressure sensor, thick-film sensor.

I. INTRODUCTION

The number of patients who had cerebrovascular or other neurological disorders (e.g. Parkinson's disease) is continuously increasing. Usually these patients have trouble swallowing and articulating a sound and they require rehabilitative care [1-2]. The possibility of measuring the tongue pressure against the hard palate during syllable articulation or swallowing is important and, in fact, the contact between tongue and palate is of vital importance for both functions.

The techniques usually used are videofluorography (VF) or videoendoscopy (VE). In [3-5], the authors analyzed qualitatively the tongue coordination and jaw movements in chewing and swallowing via VF and VE. However, the application VF widely and repeatedly is not recommended because of the dangers of radiation exposure for patients. Therefore, in the literature, studies on the measurement of tongue contact against the hard palate in a non-invasive manner have been studied recently. In [6-11], different types of pressure sensors are used to measure the tongue pressure. These techniques are based on a sensor in contact with the palate and connected via cables to a conditioning electronics positioned outside the oral cavity. For example in [9], the authors report a pressure sensor for measuring the tongue pressure on the palate using a palatal plate with seven pressure sensors. However, the presence of wired connections, between the oral cavity and the reading unit placed outside, may impair sound articulation or swallowing. Furthermore, the presence of cables can be perceived by the patient as invasive. This problem is widely reported in the literature and different approaches are proposed [12-15]. To overcome this problem, several authors propose devices transmitting information wirelessly between the inside and the outside of the oral cavity. In [13], the authors describe the architecture and the operating principle of a new wireless system for balance control for the falling prevention. The device measures if and where the tongue is in contact with the palate, but does not incorporate a pressure sensor. In [14], the aim was to develop a wireless telemetry method to complement the shortcomings of existing methods that measure the intra-oral pH. In [15], it is reported a prototype system for wireless control of different interfaces. The wireless intraoral device is activated by the tongue, the signals are sent to tens of meters to a coordinator wireless transmitting commands from the intra-oral device to the computer. The sensor operates as a switch by sending the contact information.

In this research, we propose a wireless intra-oral device for measuring the tongue pressure on the palate. The presented system is designed to measure the tongue pressure on the palate and wirelessly transmit data to a reading unit placed outside the oral cavity. The device consists of seven sensors connected to an electronic circuit for conditioning and data transmission. The sensors are fabricated using screen printing technique on a plastic substrate at a low temperature. Screen printing technique consists in the deposition of a film of a few microns in thickness in a single step using a mask. This technique allows the manufacture of low cost, light, flexible and biocompatible sensors. The target applications may be different: first of all, these sensors can be used in wearable systems [16] or in implantable systems within the human body, in passive devices or connected to power harvesters [17].

In a previous publication [18], a first device was presented and preliminary experimental results reported up to 80 kPa. We now present an improved version of the device. A new conditioning circuit is introduced and the building blocks are briefly described. Furthermore, the sensor is tested with a new experimental apparatus and according to proper testing protocols, designed so to highlight the correlation between the exerted stress and electrical output. Accurate analyses for a

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wide field of measurement up to 100 kPa have been performed. The preliminary experimental results are reported and discussed.

II. WIRELESS INTRAORAL DEVICE

The device consists of a thick film sensor fabricated on a plastic sheet (Fig. 1). The sensor is connected to a circuit for signal conditioning and data transmission outside the mouth wirelessly. The sheet's thinness is considered effective to reduce discomfort in the oral cavity.

The system can be fully or partially enclosed in a plastic casing of biocompatible material. The casing can be thermoformed to fit the patient's oral cavity geometry.

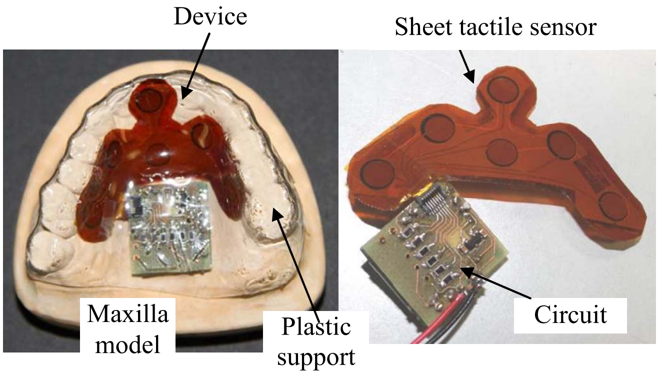

Fig. 1. Images of the tongue pressure sensor enclosed in a biocompatible plastic support and placed on a maxilla model.

A preformed maxillary model of an adult subject was used as a reference for the design of the sensor geometrical characteristics. The sensor implements six measurement points (P1 - P6), Fig. 2 .The measurement point locations are decided on the basis of the dental arch and anatomy of a standard adult [9]. The sheet size and the distance between the measurement points are shown in Fig. 2. Each measurement point has a diameter of 3.2 mm. Two measurement points (P1 - P4) are placed along the midline, two (P5 - P6) are in the rear and two (P2 - P3) laterally.

In addition, the sheet on which the sensor is manufactured has a geometry that can be adapted to the oral cavity curvature, and the circuit has small size (about 18 x 20 x 5 mm) and is positioned in an area not affected by the tongue contact on the palate.

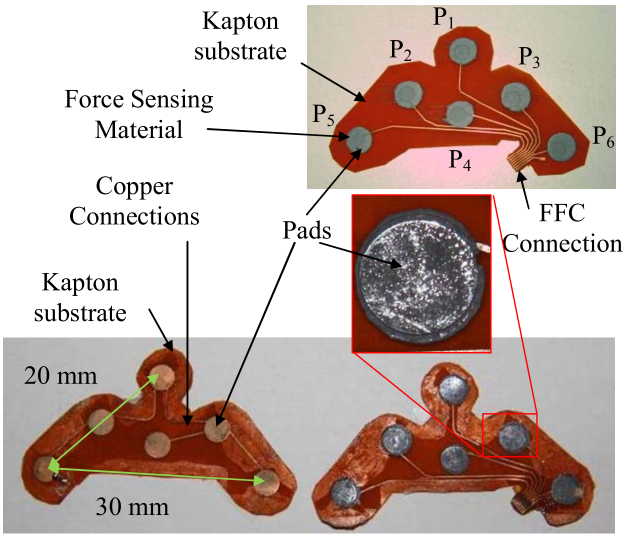

Fig. 2. Images of the two sensor layout parts and a zoom of one pad with the force sensing material deposited.

The sensor is made from two sheets of Kapton film (25 microns thick) laminated with copper on it (35 microns thick). Electrodes and connections are obtained by using the photolithographic technology, and then a pressure-sensitive material, (CreativeMaterials DS118-44) is deposited on the copper electrodes (Fig.2). This material is deposited by screen printing techniques and cured at a low temperature of 120 °C for about half an hour. In the end, an adhesive is applied to connect together the two sheets, so the sensor thickness is about 150 microns.

Copper connections and thick film depositions are then hermetically contained within the two sheets. The outer layer is then Kapton which is a biocompatible material. The operating principle of each measurement point is as follows: the electrical resistance in the absence of load is almost infinite, and decreases when a force is applied perpendicular to the surface.

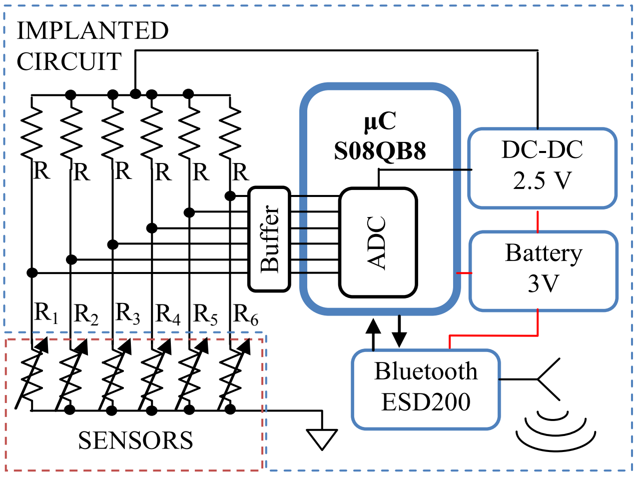

Fig. 3. Block diagram of the implanted system.

The block diagram of the system for the measurement of the tongue pressure is shown in Fig. 3. The conditioning and the transmission circuit are operated by a low-power microcontroller (FreescaleS08QB8) powered by a 3 V button cell battery. The battery voltage is stabilized by a DC-DC regulator with output 2.5 V, this voltage is used as a fixed reference for the analog-digital converter and as a reference voltage for the resistive dividers.

The measuring principle is based on a measurement of the voltage divider as shown in Fig 3, the resistance R is about 30 MΩ. The sampling frequency is about 70 Hz, but it can be changed by software. The analog-digital converter (ADC) resolution is 10 bits, which gives a resolution of about 2.5 mV, equal to theoretical minimal resistance variation of about 100 kΩ for a sensor resistance range of 30 MΩ - ∞.

The voltage divider is converted from the ADC and sent via serial communication to the Bluetooth module (ParaniESD210). Thus, the data are transmitted in real time to a personal computer.

III. EXPERIMENTAL SETUP

For the sensor behavior analysis, an experimental setup has been designed ad hoc (Fig. 4). An electromechanical dynamometer (Instron, mod. 3366), employed under load control and in compression conditions, used for the sensor characterization aiming to vary the applied pressure over each measurement point.

The sensor was fixed to a flat compression plates and the load was applied by means of an ad hoc prepared compression rod, whose diameter was approximately that of the pad; this allowed to easily calculate the pressure on the sensor, by dividing the measured force by the rod cross section. The machine is equipped with a load cell able to detect the force exerted on the material with a resolution of 0.01 N. The conformation of the machine ensures that the force is exerted perpendicular to the measurement point. The machine is connected to a computer allowing to control the scrolling speed of the beam and the exerted force.

Each individual measurement point was connected to a circuit specially designed and shown schematically in Fig 4. The sensor is connected to a resistive divider, and then to an amplifier (OPA2320) in voltage follower configuration. We have chosen, at this stage, powering the circuit between +5 V and - 5 V in order to focus on the sensor response rather than on the optimization of the conditioning circuit.

The output voltage was measured using a Hewlett Packard 34401A multimeter. The tests were carried out at room temperature (about 24 °C). Furthermore, a temperature characterization was performed using the climatic chamber Angelantoni (UY110).

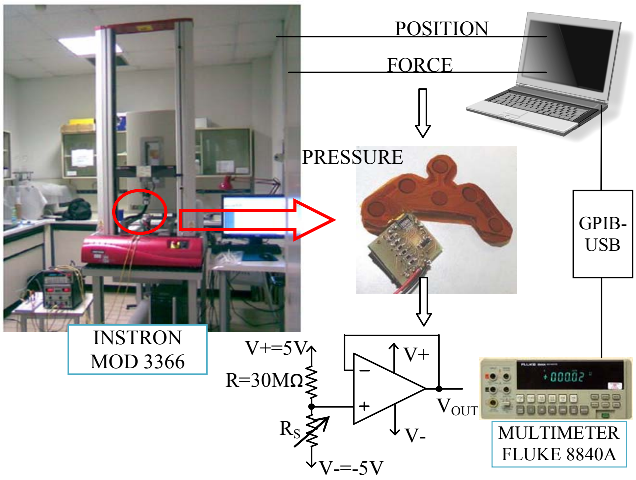

Fig. 4. Experimental setup adopted for the characterization of the sensors.

IV. PRELIMINARY EXPERIMENTAL RESULTS

Preliminary experimental tests were carried out to evaluate the sensor outputs. The specimens were subjected to loading-unloading ramp at fixed loading rates, while simultaneously measuring displacement, load and voltage variations. In this paper, a preliminary analysis on one measurement point is shown. A detailed analysis of all measurement points is in progress and will be reported in the extended version.

In Fig. 5, the applied pressure trend and the corresponding deformation measured during 35 seconds is shown. The deformation is the displacement of the force application point with respect to the point of early contact with the sensor, when pressure is slightly above 0.1 kPa.

The pressure was applied from 0.01 MPa to 0.1 MPa in a cyclic manner, with the loading phase of about one second, and the unloading phase of about 5 seconds. This protocol allows testing the sensor in overloaded conditions. In fact, in the literature, the average maximum pressure elevation of the tongue on the palate is 57.5 ± 15.1 kPa [3], which is about half of the maximum pressure applied during the test, whereas the minimum value of 0.01 MPa was chosen to always ensure a constant pressure on the sensor during all cycles in order to evaluate the possible signal drifts.

In Fig. 5, a progressive sensor crushing with a consequent thickness reduction is visible as testified by the drift of the displacement signal to slightly higher values. This reduction is about 9 um after the application of six cycles. This crushing even though small is perhaps due to the high pressure at which the measuring point is subjected (0.1 MPa). Lower pressure application, as well as expected for the reference application [3], will allow a considerably reduction of the phenomenon.

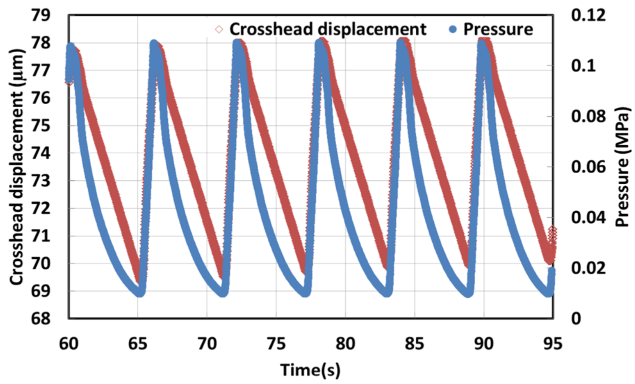

Fig. 5. Crosshead displacement and pressure values for one measurement point during six cycles.

In Fig 6, the measured voltage values (V_{OUT}) with respect to the applied pressure during six cycles are shown. An increasing pressure generates a decrease in the measurement point's resistance value and consequently a decrease in the voltage value. For a variation of about 0.1 MPa the voltage has a variation of about 4 V.

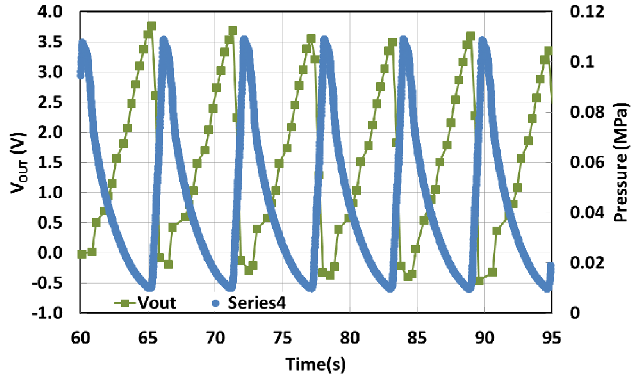

Fig. 6. Output voltages and pressure values for one measurement point during six cycles.

Figure 7 shows the mean values and standard deviations of the measured voltage values vs. pressure during the two tests repeated for six cycles each. As can be observed, at high pressures the deviation is greater, this phenomenon is due to the drift analyzed previously and as said, it can be reduced for pressure values up to 60 kPa.

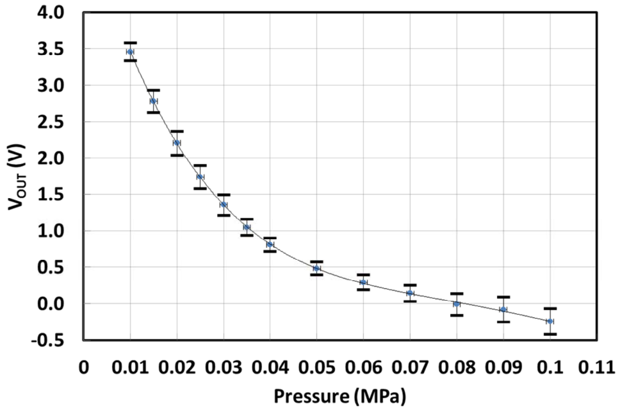

Fig. 7. Output voltages vs. pressure values up to 100 kPa for a single measurement point (mean±1SD).

In human applications, the temperature influence on implanted sensors must be controlled. The temperature influence on the sensor was analyzed and the preliminary experimental results are shown in Fig 8. The temperature has been made to vary between 15 °C and 38 °C and a maximum hysteresis of about 0.1 V for the sensor output voltage (V_{OUT}) has been calculated (Fig. 8). Also in this case the sensor has been subjected to an wide-range test, since the temperature variation of 23 °C is excessive for the reference application. The oral temperature variation is almost the body temperature and for healthy subject it is stable at around 36-37 °C.

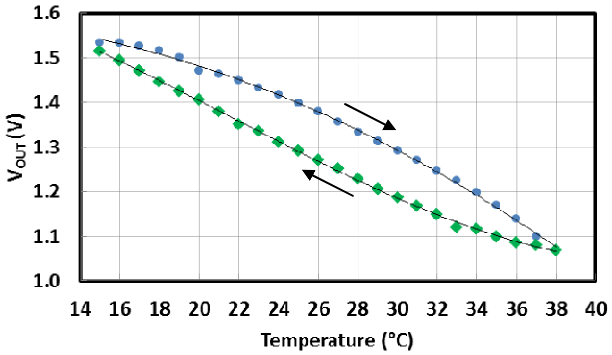

Fig. 8. Output voltage variation for a temperature cycle from 15 °C to 38 °C.

V. CONCLUSIONS

In this paper, a wireless sensor for measuring the tongue pressure on the palate is presented. The proposed device consists of six sensors and a conditioning and transmission circuit. The sensor is fabricated with screen printing technique on a plastic substrate at low temperatures. The conditioning and transmission circuit is introduced and the building blocks are briefly described. The sensor thickness and the compact circuit size can reduce the discomfort in the oral cavity. Preliminary experimental results are reported and discussed.

The characterization shows an operation for a wide range of pressure up to 100 kPa. A validation of the system is in progress with the medical staff with the aim to carry out different tests on patients. In fact, application fields for this device are the treatment of people with language or swallowing disorders.

REFERENCES

[1] O. Chenu, N. Vuillerme, J. Demongeot and Y. Payan, "A Wireless Lingual Feedback Device to Reduce Overpressures in Seated Posture: A Feasibility Study", PLOS one 4(10), 2009, pp. 1-8.

[2] C.M. Steele, G.L. Bailey, S.M. Molfenter, E.M. Yeates and K. G. Martin, "Pressure profile similarities between tongue resistance training tasks and liquid swallows", J. of Rehabilitation Research & Development 47(7), 2010, pp. 651–660.

[3] H. M. Clark and N. P. Solomon, "Age and sex differences in orofacial strength", Dysphagia 27(1), 2012, pp. 2-9.

[4] T. Ono, K. Hori, KI. Tamine and Y. Maeda, "Evaluation of tongue motor biomechanics during swallowing—From oral feeding models to quantitative sensing methods", Japanese Dental Science Review 45(2), 2009, pp. 65–74.

[5] Y.J, Kumakura, K. Hori, KI Tamine and T. Ono, "Differences in biomechanical features of tongue pressure production between articulation and swallow", J. of Oral Rehabilitation 39(2), 2012, pp.118-25.

[6] F. Yoshioka, S. Ozawa, Y.I. Sumita, H. Mukohyama and H. Taniguchi, "The pattern of tongue pressure against the palate during articulationg glossal sounds in normal subjects and glossectomy patients", J. Medical Dental Science 51, 2004, pp. 19-25.

[7] M.J. Mcauliffe, E.C. Ward and B.E. Murdoch, "Intra-participant variability in Parkinson's disease: an electropalatographic examination of articulation", Advances in Speech–Language Pathology 9(1), 2007, pp. 13–19.

[8] J. Yano, I . Kumakura, K. Hori, K.I. Tamine and T. Ono, "Differences in biomechanical features of tongue pressure production between articulation and swallow", J. of Oral Rehabilitation 39, 2012, pp. 118–125.

[9] K. Hori, T. Ono, K. Tamine, J. Kondo, S. Hamanaka, Y. Maeda, J. Dong and M. Hatsuda, "Newly developed sensor sheet for measuring tongue pressure during swallowing", J. of Prosthodontic Research 53(1), 2009, pp. 28-32.

[10] K. Yanagisawa, I. Takagi and K. Sakurai, "Influence of tongue pressure and width on tongue indentation formation", J. of Oral Rehabilitation 34; 2007, pp. 827–834.

[11] C. Jeannin, P. Perrier, Y. Payan, A. Dittmar and B. Grosgogeat, "Tongue pressure recordings during speech using complete denture", Materials Science and Engineering C 28, 2008, pp. 835-841.

[12] A. Cadei, A. Dionisi, E. Sardini, M. Serpelloni, "Kinetic and thermal energy harvesters for implantable medical devices and biomedical autonomous sensors", Measurement Science and Technology 25 (1), 2014, art. no. 012003.

[13] N. Vuillerme, N. Pinsault, O. Chenu, A. Fleury, Y. Payan and J. Demongeot, "A wireless embedded tongue tactile biofeedback system for balance control", Pervasive and Mobile Computing 5, 2009, pp. 268–275.

[14] J.H. Ro, S.Y. Ye, J.H Jung, A.Y. Jeon, Y.-J Kim, I.C. Kim, C.H. Kim and G.R. Jeon, "Development of Indwelling Wireless pH Telemetry of Intraoral Acidity", World Academy of Science, Engineering and Technology 30, 2007, pp. 361-365.

[15] Q. Peng and T.F. Budinger, "ZigBee-based Wireless Intra-oral Control System for Quadriplegic Patients", Proceedings of the 29th Annual International Conference of the IEEE EMBS, Cité Internationale, Lyon, France, August 23-26, 2007, pp. 1647-1650.

[16] M. Borghetti, E. Sardini, M. Serpelloni, "Sensorized glove for measuring hand finger flexion for rehabilitation purposes", IEEE Transactions on Instrumentation and Measurement 62, 2013, pp. 3308-3314.

[17] A. Cadei, A. Dionisi, Sardini, M. Serpelloni, "Kinetic and thermal energy harvesters for implantable medical devices and biomedical autonomous sensors", Measurement Science and Technology 25, 2014, 012003.

[18] E. Sardini, M. Serpelloni, R. Fiorentini, "Wireless intraoral sensor for the physiological monitoring of tongue pressure", 2013 Transducers and Eurosensors XXVII: The 17th International Conference on Solid-State Sensors, Actuators and Microsystems, TRANSDUCERS and EUROSENSORS 2013, art. no. 6627010, 2013, pp. 1282-1285.