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Epidermal Passive RFID Strain Sensor for Assisted Technologies

Osman O. Rakibet, Christina V. Rumens, John C. Batchelor, Senior Member IEEE
and Simon J. Holder

***Abstract*—An epidermal passive wireless strain sensor using RFID tags is presented. The tag is intended to detect eyebrow or neck skin stretch where paraplegic patients have the capability to tweak facial muscles. The tag is designed on a Barium Titanate loaded PDMS substrate and is assessed to demonstrate the strain gauge sensitivity and repeatability as a function of skin stretch.**

***Index Terms*—RFID; wireless sensing; passive sensing; strain gauge; assisted living; paraplegia.**

I. Introduction

As well as logistics, mobile healthcare, homeland and personal security has application in sensor networks, [1-3]. Radio Frequency Identification (RFID) also has potential application for assisted living and rehabilitation systems aiding disabled and incapacitated people.

In applications such as powered wheelchair control, the navigation and control systems needs to be responsive in real-time and offer users reassuring robust support, especially in collision avoidance. It is important that assistive technologies incorporate a degree of intelligence, and must be sufficiently dynamic to recognize and accommodate for patients providing inputs of varying accuracy. Robotic assistance employed in the healthcare arena must therefore emphasize positive support rather than adopting an intrusive or over-supportive role [4], especially in rehabilitation scenarios where patients should be encouraged to gain increased independence as they learn to manage a condition as their needs change with time. This issue is the subject of SYSIASS, a European Commission funded project where autonomous powered wheelchair technology is supported by sensors to prevent collisions with door frames, static objects, and people.

A mouth mounted RFID tag acting as a tongue touch controlled switch has been proposed for joystick or mouse control on a powered wheelchair to offer a control interface for severely incapacitated wheelchair users with limited or no movement in their arms, fingers or hands [5]. Ultra High Frequency (UHF) RFID is proposed in this paper offering similar functionality for the same purpose, but in this case acting as the enabling technology for skin strain sensors that detect muscle twitches in the face and neck.

The strain sensor can be attached above the eyebrows or around the neck where many paraplegic patients have movement capability. The skin stretch associated with facial muscle tweak leads to tag geometry distortion and this is detected as a function of transmitted threshold power, which is the power required to turn on the RFID tag transponder for a known read distance. When combined with a proximal wheelchair mounted read antenna, there is an opportunity to monitor the extent and direction of the twitch and therefore control a wheelchair with respect to skin stretch.

II. Strain Gauge RFID Tag

Owing to the high permittivity and high loss tangent of skin and muscle tissues [5], the design of efficient RFID tags on, or close to, skin is highly challenging. In addition to this the dielectric properties of tissue vary according to location in the body and also between individuals. Very low profile skin mounted UHF RFID tags have poor radiation efficiency and it is important to obtain a good impedance match between the tag transponder chip and the antenna terminals if read ranges of more than a few cm are to be achieved [6].

The passive UHF RFID tattoo tags in [6, 7] were designed to withstand ordinary skin flexing for inkjet printed or sputtered conductors onto transfer paper. However, owing to micro-cracking of the conductors with applied strain, the resulting antennas suffer from poor efficiency and reduced gain. Additionally, the stretched structures may not cycle well and do not regain their original performance after being relaxed [8]. This challenge is addressed here where elastic conducting fabric is mounted on a substrate of polydimethylsiloxane (PDMS) which is loaded with barium titanate ($BaTiO_3$). This material was selected to obtain a low-profile elastic structure with a defined and invariant permittivity value significantly above free space and the $BaTiO_3$ loading was adjusted to control the ε_r value. Additionally, the low chemical reactivity and non-toxicity of the material make it well suited for epidermal application [9]. Loaded PDMS structures have been described in [10] to create flexible tags, while the aim here is to create a structure where the entire tag including the conductor can stretch.

III. Elastic PDMS Substrate Formulation

PDMS elastomers are formed using viscous linear PDMS, liquid cross-linker and a catalyst. Mechanical properties, such as elasticity are easily modulated by varying the cross-linker density i.e. the molecular weight of the linear PDMS and cross-linker concentration. $BaTiO_3$ is a ferroelectric ceramic powder, with high relative permittivity values of up to 4000 at room temperature [11]. As well as frequency, the permittivity of $BaTiO_3$ is dependent on its crystalline phase, temperature, dopants and importantly, the synthesis route which includes purity, density and grain size [12].

The substrates were fabricated by mixing $BaTiO_3$ and PDMS before cross-linking occurred using tetraethyl orthosilicate (TEOS) via a Sn (II) catalysed condensation reaction. The permittivity value of the substrates was controlled by varying the weight percentage of $BaTiO_3$ as shown in Fig. 1. PDMS substrates with 28.4 wt% $BaTiO_3$ loading produced a measured relative permittivity value of $\varepsilon_r = 3.4$ which was deemed suitable for the strain gauge design as higher loading percentages compromised the elasticity.

To synthesise the 28.4 wt% $BaTiO_3$ substrate, a silanol-terminated PDMS, viscosity 18000 cSt, (DMS-S42) (M.W. 77,000, Fluorochem Ltd.), tin (II) 2-ethylhexanoate (95%, Sigma Aldrich), toluene (analytical reagent grade, Fisher Chemicals) barium titanate (< 2μm particle size, 99.9% trace metal basis, Sigma Aldrich) and tetraethyl orthosilicate (99%, Sigma Aldrich) were used as received. Homogenous mixing of elastomer components was achieved using a labortechnik speed-mixer (RCT basic IKA).

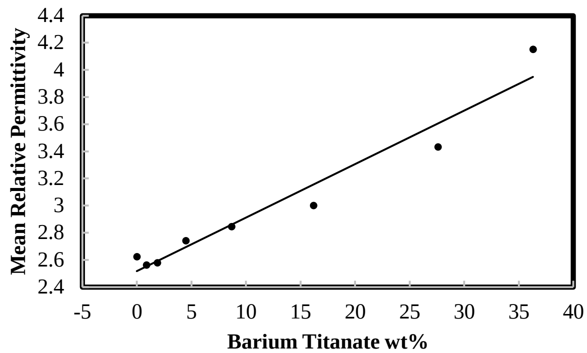


Fig. 1 Relative permittivity of $BaTiO_3$-PDMS composites (3GHz)

Silanol-terminated PDMS (12g, 0.156 mmol), $BaTiO_3$ (4.8g, 20.6 mmol), tetraethyl orthosilicate (0.070cm^3, 0.313mmol) and toluene (3.47cm^3, 35.6 mmol) were added to a glass beaker and speed-mixed for 30 minutes. Tin (II) 2-ethylhexanoate dissolved in toluene (0.074 cm^3, 0.148 mmol) was then added to the mixture and speed-mixed for 60 seconds before being poured into the circular mold. A flexible filling knife was drawn down over a PTFE circular mold (diameter = 80 mm, height = 1 mm) to ensure a uniform height elastomer which was allowed to cure at room temperature for 2 hours before being placed into an oven at 60ºC for 72 hours. The resulting substrates were stretchable, soft, flexible and water resistant.

IV. Tag Construction and Properties

To facilitate convenient epidermal mounting, a low-profile tag with overall dimensions reduced by 23% compared to that of [6] is shown in Fig. 2 with the principal dimensions listed in Table I. The tag conductor was simulated in CST Microwave Studio on a 0.046 mm thick Mylar sheet ($\varepsilon_r = 3$) attached to the loaded PDMS substrate with $\varepsilon_r = 3.43$. To represent human tissue, a 2 layer stratified rectangular phantom was included with 154×160 mm^2 upper surface area and comprising a 26mm top layer of combined skin and fat and an underlying 20mm thick muscle layer. The skin/fat layer was modelled with ε_r of 14.5 and conductivity σ of 0.25 S/m, and the second layer with ε_r of 55.1 and σ of 0.93 S/m as suggested in [13].

The permittivity of the loaded PDMS substrate was determined experimentally at 3GHz by the waveguide transmission method described in [14]. The substrate material was supported by expanded polystyrene foam and placed in an E-Band rectangular waveguide of cross section 58mm×29mm. Comparison of the scattering parameters, with and without the PDMS sample present, confirmed the dielectric constant of the loaded substrate to be 3.4. To verify this result, a physical prototype of the tag was fabricated by etching the design of Fig. 2 onto a copper clad thin Mylar sheet which was attached to the PDMS substrate. Measurements were taken using a Voyantic TagformanceLite UHF RFID characterisation system which accurately records transmitted threshold power from tags at known distance. Read range was calculated from measured reader power and found to be unchanged when the tag was placed either on the loaded PDMS substrate or on Perspex which was known to have a permittivity value of 3.4.

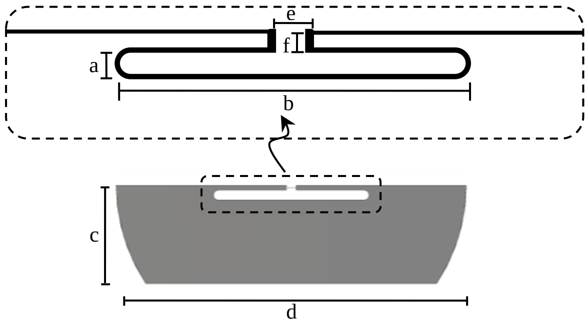


Fig. 2 Geometry of the RFID strain sensor

TABLE I STRAIN RFID SENSOR DIMENSIONS

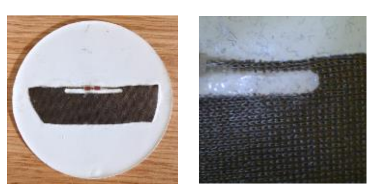
Parameters	Symbol	Length (mm)
Slot Width	a	2
Slot Length	b	32
Tag Width	c	20
Tag Length	d	72
Chip Length	e	2
Feed Line Thickness	f	1

Simulation of the tag using CST Microwave Studio with a 1 mm thick substrate (ε_r of 3.4) indicated a radiation efficiency of 82%, as opposed to 50% for a tag mounted directly on skin. In order to establish the accuracy of the body phantom and the PDMS material values, a full tag prototype was assembled on the loaded PDMS sample using an NXP UHF RFID chip with -15dBm sensitivity and input impedance 18-j125 Ω. The entire structure was placed on the skin of a volunteer's forearm using adhesive tape and the threshold power was measured with the Voyantic system. A read range of 1.6m at 868 MHz was measured, corresponding to the simulated S_{11} null frequency.

To develop a fully functioning prototype strain gauge, a stretchable conducting Lycra® fabric containing silver threads was used [15]. This material comprises highly elastic nylon based fibres with metallisation formed from embedded silver nanoparticles. The Lycra® conductor was laser cut with rounded corners in the slot ($a \times b$) to avoid slightly cutting into the feed lines at the slot ends, Fig. 3(b).

The silver Lycra® antenna was attached to the PDMS substrate during the curing process so no adhesive was required. Preliminary studies showed premature placing of the Lycra® caused a loss in conductivity as the semi-cured composite penetrated the fabric and coated the silver fibres. The coating insulated the parallel conducting strands in the fabric, increasing the resistivity, and the narrow feed lines of width f were particularly susceptible to conductivity degradation. To remedy this, the main body of the antenna was placed onto the composite 75 minutes into curing while the feed lines were placed at 95 minutes. The composite was then left to cure at room temperature for a further 25 minutes before being placed in the oven for 72 hours. The resulting tag showed excellent Lycra®-antenna adhesion with no visible PDMS seepage and the similarity of Young's Moduli between the two meant they could be strained with no visible wrinkling around the conductor edges caused by differential stretch. The circular PDMS substrate was 80mm in diameter and 1mm in thickness.

The Lycra® had thickness of 0.5 mm and its conductivity was measured to be 800 S/m with a four probe ohmmeter (Rhopoint milli ohmmeter, Micron Technology Ltd). The read range was found to be only 60% that of the copper prototype when tested on skin owing to the poor Lycra® conductivity. The prototype is shown in Fig. 3 and the peak read range has occurred at 868MHz.



(a) (b)

Fig. 3(a) PDMS with Lycra®, (b) Expanded view of rounded slot end.

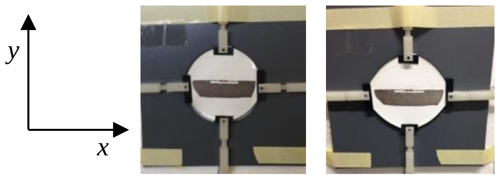
V. TAG STRAIN RESPONSE

In order to assess the tag performance under strain, the power necessary to obtain a tag read was measured for various percentage strains with the tag clamped into an adjustable PTFE jig. To examine the stretch effect in the *x*- and *y*-axis directions, the tag substrate was clamped at the right and left edges (x-axis), and the top and bottom edges (y-axis) respectively.

Stretch on the *x*-axis distorted the substrate into a left-right aligned elliptical structure making the slot longer and narrower (*a* decreases and *b* increases). Conversely, *y*-axis stretch causes the slot to become shorter and wider (*a* increases and *b* decreases). Fig. (a) and Fig. (b) illustrate the unstrained and 10% *x*-axis strain conditions respectively where strain is expressed as a percentage change in *a*, (positive for *x* stretch and negative for *y*).

VI. RESULTS AND DISCUSSION

In order to apply consistent and accurate strain to characterize the tag, a jig was constructed. When in the jig, the tag was not on skin, so the read range was reduced. However performance remained sufficient for measurements to be taken on the VoyanticLite system.



(a) (b)

Fig. 4 Tag on PDMS substrate of 1mm height and conducting Lycra® 0.5mm thick (a) unstrained, and (b) *x*-axis 10% stretched tag.

The principal of sensing is similar to that described in [16, 17] where a sensed parameter detunes the tag and the magnitude of the change is detected as an alteration in measured threshold power. In this case the threshold power changes due to a physical distortion of the tag and possible changes in Lycra® conductivity which alters the tag input transmission coefficient, τ, and the antenna gain.

Each strain measurement was repeated five times to remove fluctuations due to narrow band fading and averaged results for the unstretched and the stretched tag are presented in Fig. . The unstretched tag was worse matched and therefore exhibited a narrow matched bandwidth, though still sufficient to respond. Positive strains correspond to x-axis stretch where *a* increases and *b* decreases, while negative strains indicate y-axis strain where *a* and *b* have the opposite trends. It was found that about 26dBm transmit power was required at 868MHz to activate the unstretched tag while for 10% *x*-axis stretching, where the slot became longer and narrower, the impedance match improved and the required power reduced to 23.7dBm. When the tag was strained in the y-axis, the required power increased to 27.4dBm.

To assess the performance over time, the tag remained on the jig in a 10% stretched state for one week, after which the measurements were repeated. As shown in Fig. 5, at the operating frequency (868MHz), the difference between the new and original measurements was only 0.2dBm which indicated good repeatability.

Fig. 6 shows the strain gauge transfer response relating strain to required reader power. Although a polynomial would offer a close fit, in order to simplify the read/sensing process the response is approximated to a linear trend as would be expected for elastic materials. Although the threshold power is dependent on both the gain and the transmission coefficient of the tag, a linear slope is still observed as a reasonable fit. The trend indicates a sensitivity of 0.25dBm/percent strain with a maximum error of 2.4% from measurement. Using the original calibration trend after one week results in maximum errors of 1.8% and 0.8% for x- and y-axis strains respectively. The reason for the apparent improvement in calibration over time is because even though the slope of the later measurement trend line increases to 0.28dBm/percent, most of the 1 week data points lie quite close to the original trend and there is no significant outlier. Although this particular effect could not be expected in practice, nevertheless, it can be concluded that the strain gauge remains well calibrated after being stretched over significant periods of time.

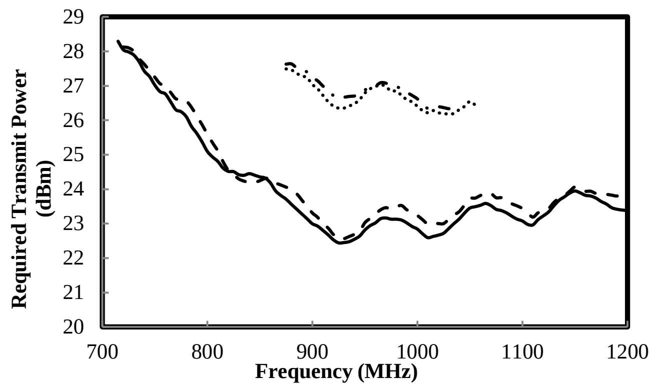


Fig. 5 Transmitted power vs. frequency graph with *x*-axis stretch

-5.3% Stretch 10.4% Stretch

-5.3% Stretch +1 week 10.4% Stretch +1 week

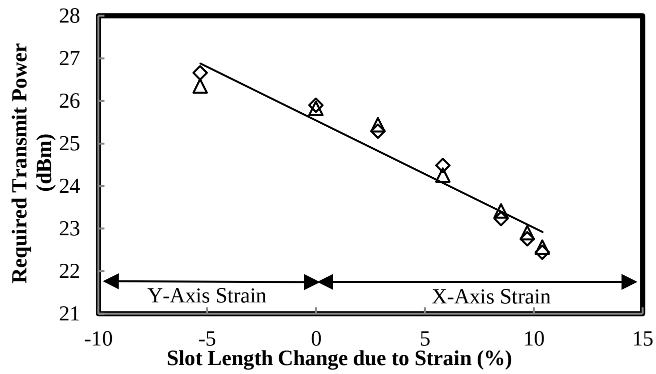


Fig. 6 Required tramsitted power as function of slot length

Δ First Measurement, trend line, ◇ measurement + 1 week.

VII. Conclusion

The principle of a passive epidermal strain sensor enabled by a UHF RFID tag on a flexible PDMS substrate has been demonstrated. The increase in permittivity associated with barium titanate loading in the substrate allowed the effect of tag detuning to be emphasized as a function of stretch, and additionally, the substrate material parameters led to an acceptable performance when mounted directly onto skin. A linear fit was found to offer reasonable agreement for calibration with strains of up to 10% causing read transmit power differences of about 4dB. This should be easily detectable after averaging over multiple reads to remove fast fade fluctuations. Measurements on a volunteer show 1 cm displacement occurs for a raised eyebrow which is twice the stretch observed in Figs 5 and 6.

To be a practical and conformal transfer mounted sensor, the design should be reduced in size, and the substrate will be investigated for performance as a thin film significantly less than 1mm in height. The response with repeated/cycled stretch must also be assessed to evaluate the tag performance in long term use. More work also needs to be done to investigate the recovery of the tag from stretching to see if any plastic deformation or hysteresis is found.

For use in a muscle tweak sensor for a joystick or mouse, the associated reader system would be integrated into a powered wheelchair and calibration would be required to account for variability in the distance between tag and read antenna. The close proximity between antennas would lower the required reader transmit power and reduce the effect of multipath and possible interference in hospital environments. Finally, the unique ID code embedded into each tag transponder facilitates multiplexing in 4 vector control systems such as joysticks.

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