System-Level Design of an RFID Sweat Electrolyte Sensor Patch

Daniel P. Rose-*IEEE Student Member*, M. Ratterman, Daniel K. Griffin,
Linlin Hou, Nancy Kelley-Loughnane, Rajesh K. Naik, Joshua A. Hagen,
I. Papautsky *IEEE Senior Member*, and Jason Heikenfeld *IEEE Senior Member*

***Abstract*— Wearable digital health devices are dominantly found in rigid form factors such as bracelets and pucks. An adhesive RFID sensor bandage (patch) is reported, which can be made completely intimate with human skin, a distinct advantage for chronological monitoring of biomarkers in sweat. In this demonstration, a commercial RFID chip is adapted with minimum components to allow potentiometric sensing of mM ionic solutes in sweat, and surface temperature, as read by an Android smart-phone app (*in-vitro* tests).**

I. Introduction

SWEAT is one of few examples of non-invasively accessed biofluids, with potential advantages in measurement of inflammatory biomarkers compared to saliva [1] and potentially superior time-resolved (chronological) readings of biomarker concentrations compared to saliva and urine. Sweat access can be locally stimulated using FDA-approved iontophoresis (Wescor Macroduct), and recent tattoo-like sweat-sensing demonstrations further include measurement of lactate, ammonium, and sodium [2-4]. A particularly lucrative application could be hydration and heat-stress monitoring through electrolyte balance (Na^+, K^+, etc.), of use for athletes, military personnel, first-responders, and others working in extreme-conditions. Electrolyte sensing is of further value, because the high salinity of sweat can confound other biomarker readings, hence electrolyte concentrations need to be base-lined. Realizing such wearable sensors could be achieved several ways, including wearable textiles[5], tattoos[3, 6, 7], and form-factors such as those seen commercially in digital bracelet products such as Nike+ and Fitbit. What has not been previously demonstrated is a complete wireless sensor with wearability comparable to a simple Band-Aid, very low cost, communication with standard cellphones, robust shelf-life, and a design which automatically lends itself to maximum time-resolved readings of sweat.

This paper was submitted for publication on 16 Jun 2014. This work was supported in part by the U.S. Air Force and the National Science Foundation EECS Division. (#1347725).

D. P. Rose, M. Ratterman, & D. K. Griffin are with the Department of Electrical Engineering & Computing Systems at the University of Cincinnati, Cincinnati, OH 45221 USA (email: rosedp@mail.uc.edu, ratterme@mail.uc.edu, griffid6@mail.uc.edu, respectively).

L. Hou, was with the University of Cincinnati, Cincinnati, OH 45221 USA. She is now with Intel Corporation, Hillsboro, OR 97124 USA (e-mail: linlin.hou@gmail.com).

J. Heikenfeld & I. Papautsky are faculty with the Department of Electrical Engineering & Computing Systems at the University of Cincinnati, Cincinnati, OH 45221 USA (email: heikenjc@ucmail.uc.edu, papauti@ucmail.uc.edu, respectively).

J. A. Hagen, N. Kelley-Loughnane, & R. Naik are with the Air Force Research Labs, Dayton OH 45233 USA (email: joshua.hagen.1@us.af.mil, nancy.kelley-loughnane@us.af.mil, rajesh.naik@us.af.mil, respectively).

An adhesive RFID sensor bandage (patch) is reported, which can be made completely intimate with human skin, and therefore minimize dead volumes of sweat for improved chronological monitoring. A commercial RFID chip is adapted with minimum components to allow potentiometric sensing of electrolytes in sweat, and also a reading of surface temperature. Together, electrolytes and temperature could be of use for hydration and heat-stress monitoring. The patch is battery-free, powered and read wirelessly using the ISO 15693 standard currently implemented by many Android phones. From *in-vitro* solution to smart-phone readout, a dynamic range of 235 mV to 255 mV is achieved with 20 mM to 70 mM range and 96% accuracy in the detection of Na^+ concentration. The sensing electrodes can be folded over to be in direct contact with skin, or paper microfluidics can be used to wick sweat from the skin and to the sensors. The wearability of the patch has been shown up to 7 days, and includes a protective textile which provides a feel and appearance similar to a standard Band-Aid or transdermal patch. This work provides a complete integration of the key materials, electronics, microfluidics, and ergonomics, required for a wearable sweat sensing patch, paving the way for future wearable sweat sensor development and *in-vivo* testing.

II. Top-Level Design Considerations

A. Patch Size

Two patch sizes were chosen and demonstrated. The smaller size of a typical Band-Aid was chosen (~25 x 60 mm) for high user acceptance. A larger 70 x 40 mm patch was also demonstrated, a size similar to bandage that might be placed over a knee. To allow a maximally thin form factor, potentially longer shelf-life, and low cost, the patch was designed for battery-free RFID (radio-frequency ID, inductively powered). RFID requires ~10-21 loops of a coiled antenna to power the electronics (depending on patch size). The resulting area interior to the coil is more than sufficient for placement of the necessary electronics and the sensor electrodes (as visible in Fig. 1). The larger of the two patch antenna sizes was found to be more resilient to variations in antenna fabrication (antenna resonance), and provides greater reliability in communication.

B. Communication

For communication, most modern smartphones have, or will have, the capability to establish wireless connections and transfer data via a near-field communication RFID protocol. The patch is designed to operate on the standard ISO-15693 as a vicinity device. This provides the desired versatility in communication through stand-alone RFID readers as well as customizable applications ('apps') for RFID-enabled Android smartphones. The ubiquity of smartphones provides an

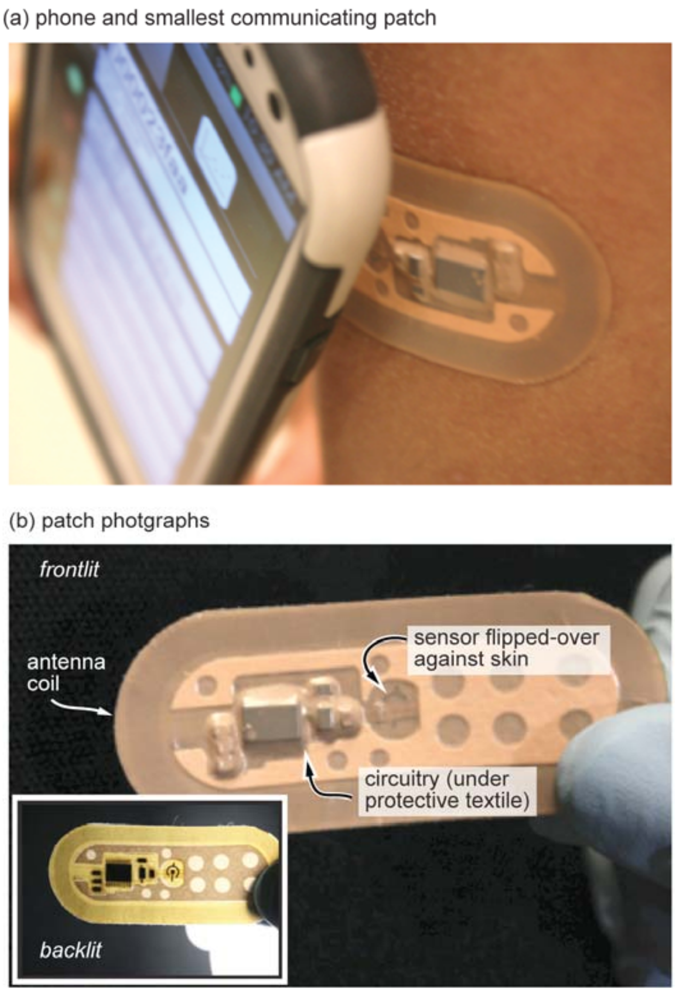

Fig. 1 Device photographs (a) shown on body and (b) to illustrate electronic and sensor features.

easily accessible computing platform with increased memory/storage capabilities, thereby eliminating the need for on-patch data logging and the additional circuitry, cost, size it would require.

A. Flexible & Wearable

Flexibility is of paramount concern in wearable electronics, along with strength and durability. The flexible printed circuit board (PCB) is built from Dupont Pyralux – a combination of flexible, conformal polyimide and a thin copper foil. The high heat tolerance of the polyimide allows for electronics to be attached by solder reflow. Furthermore, solder reflow allows surface-mount packaging which eliminates need for through-holes that would result in protrusions that would cause discomfort when the patch is worn. For packaging and skin adhesion, a survey of numerous medical-grade textiles from 3M was conducted to determine which materials would provide maximum adhesion to the wearer's skin and highest durability to protect the patch itself. Double sided medical adhesive tape was used below the patch, whereas above the patch, a medical textile covering was added to protect the patch and improve visual aesthetics (all shown in Fig. 1).

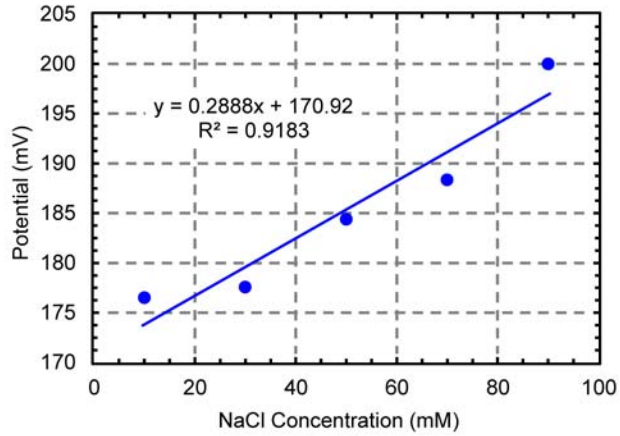

Fig. 2 Calibration curve showing sensitivity and linearity of patch to Na+.

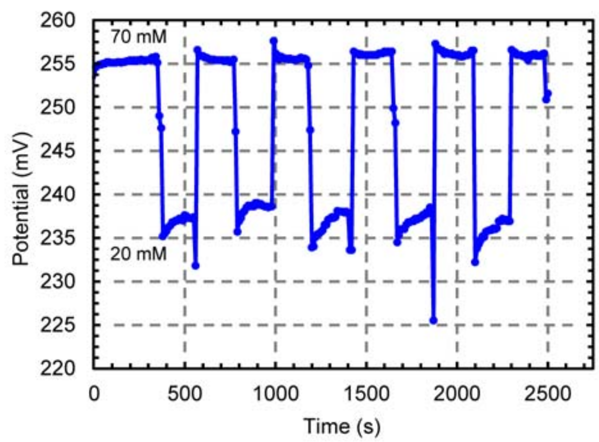

Fig. 3 Patch results for concentration variations from 20mM to 70mM NaCl.

B. Basic Electronics and Programming Functionality

The system level functionality of the patch begins and ends with communication between the reader device and the patch (Fig. 4). The reader device initiates the communication and requests identification from the patch. The patch responds by load modulating the inductive coupling between itself and the RFID reader. Custom commands for programming include reading, writing to memory registers, sensor configuration, power management, and other functionalities not required for this present work.

The primary chip in the patch is a small Melexis MLX90129 RFID transponder chip which has both basic sensor and energy harvesting capabilities. A significant amount of programming development was required to enable a smart phone to turn the RFID chip into an electrolyte sensor. The chip's sensor protocol manages tasks assigned to the MLX90129's sensor pins such as input connections, voltage output to external sensors, and enabling the internal temperature sensor. Four input pins provide differential measurement of potential from externally connected sensors (e.g. ion selective electrodes in this work). Two internal connections can enable the output of the onboard temperature sensor to be used as input into the multiplexer. Configuration of the multiplexer determines which inputs are passed to the

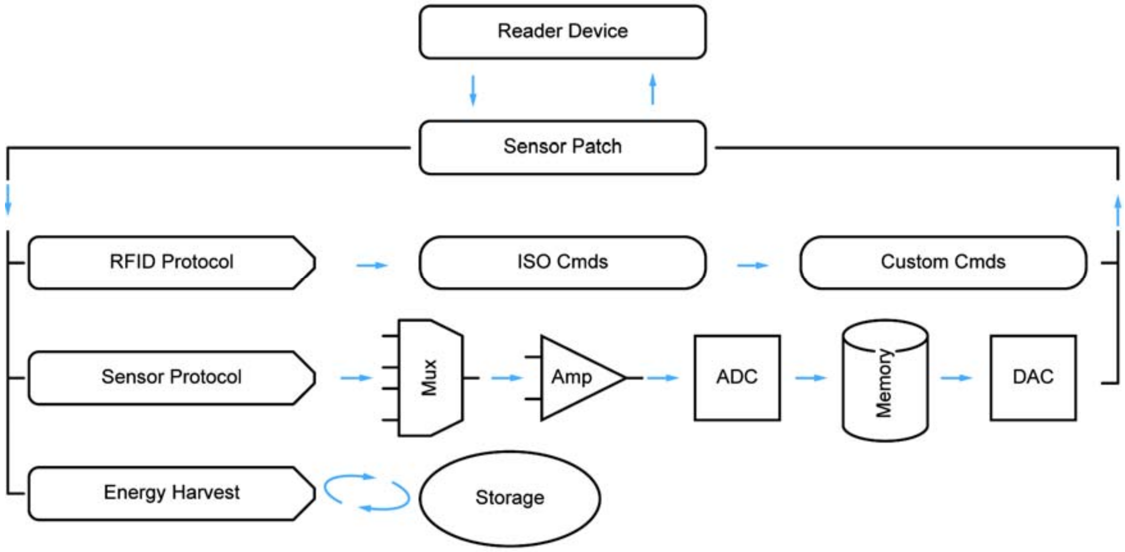

Fig. 4 System level block diagram

first programmable gain amplifier (PGA). If enabled, the digital to analog converter (DAC) can set the offset used in the optional second PGA.

III. Experimental Testing And Results

A. *Wearable Antenna Performance*

The assembled RFID circuit was tested to ensure that the resonate frequency was near enough to the target frequency of 13.56 MHz that the device would communicate with the reader. To confirm antenna tuning, a vector network analyzer (VNA) was connected to an ISO standard calibration loop-probe, per ISO10373-7, to enable contactless measurement of the patch's frequency response. The VNA displays the loop-probe's reflection coefficient by measuring the S-parameter S11 as a function of frequency. A reduction in the reflection measured by the VNA correlates to an increase in transmission of the electromagnetic waves at a specific frequency, via absorption of the radiated energy by a tuned and coupled device. Patch communication was shown in wearable format, including arm placement which induces curvature on the entire patch.

B. *Sensor and Electronics Performance*

The patch and sensors were in-vitro characterized by simple pipetting of various NaCl concentration solutions onto the patch sensor. This involved both calibration and comparisons to commercially available ion-selective electrode (ISE) sensors (Denver Instruments 300741.1). Firstly, a baseline of RFID ADC values were taken from the response of the previously mentioned commercial ion-selective electrode sensor and used as a reference for the integrated patch sensor. Using this baseline for the patch sensor a calibration curve was then generated which shows comparable linearity 0.91 to the commercial ISE of 0.95, allowing adequate distinction of different concentrations of Na+ in solution (Fig. 2). The range of 10mM to 90mM was chosen to ensure the detection range would be +/- 10 mM beyond physiologically relevant ranges for hydration monitoring [8, 9]. The patch sensors show a slope of 57 mV/decade which closely corresponds to other reports of solid-state ion-selective electrode sensors[10].

Sensitivity of the fabricated sensors in the integrated patch was 0.3 mV/mM which was slightly lower than that the commercial sensor sensitivity of 0.5 mV/mM (the sensitivity is the ratio between output signal and measured property). Therefore the patch is less sensitive to changes in NaCl concentration in the 20mM to 70mM range than the commercial electrodes. This sensitivity discrepancy can be expected due to the limitations of the ADC on the RFID patch, since the reference voltage for the ADC is generated by an outside field, and the sample rate for the converter is slow (< 3Hz), which impairs stabilization on a specific voltage.

Accuracy is an important characteristic for a sensor and measures how close sensor is able to determine the true value of a given concentration. To measure accuracy of our RFID Na+ sensor, we repeatedly measured 50mM of NaCl (n = 7), which was expected to yield 185mV based on the calibration curve (Fig. 2). Response of our patch was measured to be 177 +/- 5 mV. These results show that at 50mM NaCl, our patch sensor exhibits 96% accuracy. Precision is another important characteristic for a sensor and in essence it illustrates sensor variability. From the experimental data, the patch sensor exhibited 28% precision.

To further explore the potential for continuous hydration monitoring in sweat, concentration of NaCl was varied repeatedly every 4 minutes from 20mM to 70mM over a period of 45 min. Na+ concentration in sweat can vary from 20mM to 70mM depending on body hydration status [8, 9]. As shown in Fig. 3, the sensor exhibited good repeatability and stability during this measurement. The average high measurement was 255mV, which corresponds to 70 mM based on the concentration curve. The coefficient of variation across the 6 high concentration measurements was 0.1%, indicating good repeatability. Similarly, the 5 low concentration measurements yielded approximately 237 mV, with 0.8% variability. The response time at low concentration was slower, and is a function of the mass transport across the sensor ion-selective membrane. As with most electro-chemical sensors, drift of the response signal can be an issue and result in the inability to collect stable data. Our sensor exhibited slight drift (+/-3mV/5mM), however by increasing sampling frequency and averaging the values we were able to compensate for this difference. Future work will involve further stabilizing of the Ag/AgCl reference electrode to increase stability of the measured values. Collectively, these results suggest that the developed sensor is suitable for sweat electrolyte monitoring.

IV. Acknowledgements

The authors would like to thank Theresa Gerten, Senior Biologist at 3M, for guidance and sampling of skin adhesives, Air Force Research Labs (711th Human Performance Wing and Materials and Manufacturing Directorate) for collaborative and financial support, as well as the National Science Foundation.

References

[1] G. Cizza, A. H. Marques, F. Eskandari, I. C. Christie, S. Torvik, M. N. Silverman, T. M. Phillips, and E. M. Sternberg, “Elevated neuroimmune biomarkers in sweat patches and plasma of premenopausal women with major depressive disorder in remission: the POWER study,” *Biological psychiatry,* vol. 64, no. 10, pp. 907, 2008.

[2] M. J. Buono, C. S. White, and K. P. Connolly, “Cholinergic sensitivity of the eccrine sweat gland in trained and untrained men,” *Journal of dermatological science,* vol. 4, no. 1, pp. 33-37, 1992.

[3] T. Guinovart, A. J. Bandodkar, J. R. Windmiller, F. J. Andrade, and J. Wang, “A potentiometric tattoo sensor for monitoring ammonium in sweat,” *Analyst,* vol. 138, no. 22, pp. 7031-7038, 2013.

[4] A. J. Bandodkar, D. Molinnus, O. Mirza, T. Guinovart, J. R. Windmiller, G. Valdés-Ramírez, F. J. Andrade, M. J. Schöning, and J. Wang, “Epidermal tattoo potentiometric sodium sensors with wireless signal transduction for continuous non-invasive sweat monitoring,” *Biosensors and Bioelectronics,* vol. 54, pp. 603-609, 2014.

[5] S. Coyle, K. T. Lau, N. Moyna, D. O'Gorman, D. Diamond, F. Di Francesco, D. Costanzo, P. Salvo, M. G. Trivella, D. E. De Rossi, N. Taccini, R. Paradiso, J. A. Porchet, A. Ridolfi, J. Luprano, C. Chuzel, T. Lanier, F. Revol-Cavalier, S. Schoumacker, V. Mourier, I. Chartier, R. Convert, H. De-Moncuit, and C. Bini, “BIOTEX—Biosensing Textiles for Personalised Healthcare Management,” *Information Technology in Biomedicine, IEEE Transactions on,* vol. 14, no. 2, pp. 364-370, 2010.

[6] A. J. Bandodkar, V. W. S. Hung, W. Jia, G. Valdés-Ramírez, J. R. Windmiller, A. G. Martinez, J. Ramírez, G. Chan, K. Kerman, and J. Wang, “Tattoo-based potentiometric ion-selective sensors for epidermal pH monitoring,” *Analyst,* vol. 138, no. 1, pp. 123-128, 2013.

[7] W. Jia, A. J. Bandodkar, G. Valdes-Ramirez, J. R. Windmiller, Z. Yang, J. Ramirez, G. Chan, and J. Wang, “Electrochemical Tattoo Biosensors for Real-Time Noninvasive Lactate Monitoring in Human Perspiration,” *Analytical chemistry,* vol. 85, no. 14, pp. 6553-6560, 2013.

[8] M. J. Patterson, S. D. R. Galloway, and M. A. Nimmo, “Variations in regional sweat composition in normal human males,” *Experimental Physiology,* vol. 85, no. 6, pp. 869-875, 2000.

[9] T. Verde, R. J. Shephard, P. Corey, and R. Moore, “Sweat composition in exercise and in heat,” *Journal of Applied Physiology,* vol. 53, no. 6, pp. 1540-1545, 1982.

[10] S. Walsh, D. Diamond, J. McLaughlin, E. McAdams, D. Woolfson, D. Jones, and M. Bonner, “Solid-state sodium-selective sensors based on screen-printed Ag/AgCl reference electrodes,” *Electroanalysis,* vol. 9, no. 17, pp. 1318-1324, 1997.