Journal of The Electrochemical Society, 2021 **168** 017515

PressION: An All-Fabric Piezoionic Pressure Sensor for Extracting Physiological Metrics in Both Static and Dynamic Contexts

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The strategy of detecting physiological signals and body movements using fabric-based pressure sensors offers the opportunity to unobtrusively collect multimodal health metrics using loose-fitting, familiar garments in natural environments. (A. Kiaghadi, S. Z. Homayounfar, J. Gummeson, T. Andrew, and D. Ganesan, *Proc. ACM Interact. Mob. Wearable Ubiquitous Technol.*, **3**, 1–29 (2019)). However, many sensing scenarios, such as sleep and posture monitoring, involve an added static pressure from exerted body weight, which overpowers weaker pressure signals originating from heartbeats, respiration and pulse and phonation. Here, we introduce an all-fabric piezoionic pressure sensor (PressION) that, on account of its ionic conductivity, functions over a wide range of static and dynamic applied pressures (from subtle ballistic heartbeats and pulse waveforms, to larger-scale body movements). This piezoionic sensor also maintains its pressure responsivity in the presence of an added background pressure and upon integration into loose-fitting garments. The broad ability of PressION to record a wide variety of physiological signals in realistic environments was confirmed by acquiring heartbeat, pulse, joint motion, phonation and step data from different body locations. PressION's sensitivity, along with its low-cost fabrication process, qualifies it as a uniquely useful sensing element in wearable health monitoring systems.
 [DOI: 10.1149/1945-7111/abdc65]

Manuscript submitted September 19, 2020; revised manuscript received January 13, 2021. Published January 29, 2021. *This paper is part of the JES Focus Issue on IMCS 2020.*

The development of flexible and textile-based wearable pressure sensors[1–7] has provided the opportunity for continuous and real-time measurement of human physiological and biomechanical signals during daily activities. Pressure sensors are transducers that convert an exerted compression stress into a detectable electrical signal. Different transduction mechanisms have been introduced so far including triboelectricity,[8,9] transistivity,[10] capacitance,[11,12] piezoelectricity,[13] and piezoresistivity.[14] Piezoresistive pressure sensors are the most widely used type due to the simplicity of their structure, and the wide range of materials that can be selected along with low-cost fabrication methods, and easy read-out system required for signal extraction.[15]

A vast majority of piezoresistive sensors developed so far are on-skin sensors developed to detect subtle pressures (1 Pa–10 kPa) for touchpads and electronic skin applications.[16,17] However, to sense physiological signals such as pulse, respiration, and phonation the sensor range of detection must fall within a medium range of 10 kPa to 100 kPa. As expected, for larger-scale human motion detection such as sleep posture and footwear evaluation, the sensor must be able to sense compression stresses larger than 100 kPa.[14,17] This wide range of detection needed to capture important signals from the human body is one of the important challenges in designing on-body pressure sensors.

Most piezoresistive sensors are comprised of conductive nano- or microstructures dispersed in an elastomeric matrix and function via connection/disconnection mechanism.[17–20] The functionality and sensitivity of these sensors are highly limited by the poor bulk mechanical properties of the elastomer, in addition to the lack of breathability/water transport exhibited by these composites and other irritations arising from the skin-sensor interface.[20,21] Conceptually, textile-based sensors can overcome many of the problems posed by elastomeric skin-contact sensors. Textile-based sensors have been created by coating fibers with conductive inks[22] or intrinsically conductive polymers (ICPs).[23–25] However, these known iterations also suffer from major drawbacks. The high conductivity of the conductive coatings restricts the range of detection of these sensors to subtle pressures. Moreover, known textile-based sensors can only respond to either static or dynamic pressures—for example, once a static pressure is applied, these sensors lose their responsivity to added subtle pressure exertions due to their connection/disconnection mode of performance. Lastly, known textile-based sensors need to be used in tight-fitting clothing to be able to accurately capture signals, which makes them uncomfortable to wear and hinders their widespread adoption by a large population.

Here, we introduce an all-fabric piezoionic pressure sensor, called "PressION," that can be used to extract important physiological signals when loosely placed over various locations on the body. By taking advantage of ionic conduction, as opposed to electronic charge conduction, our sensor is capable of functioning over a wide range of applied pressures (several kPa to larger than 100 kPa), which can be further fine-tuned by modifying the conductivity of the ionic active layer. Importantly, the PressION is also capable of simultaneously responding to both static and dynamic pressures, meaning that subtle ballistic heartbeat, pulse and phonation signals can be recorded from various locations on the body, even in the presence of body weight exertions or other external pressures. To overcome challenge of tight-fitting clothing, we took advantage of the fact that, even with loose-fitting garments, there are still some parts of the garment that experience applied pressures by interacting with the body or against an external surface, such as a bed or chair, or a blanket placed over the body [Ref 29] (Fig. 1a). All these characteristics makes PressION an ideal candidate to be embedded in daily garments for future biomedical and psychological studies.

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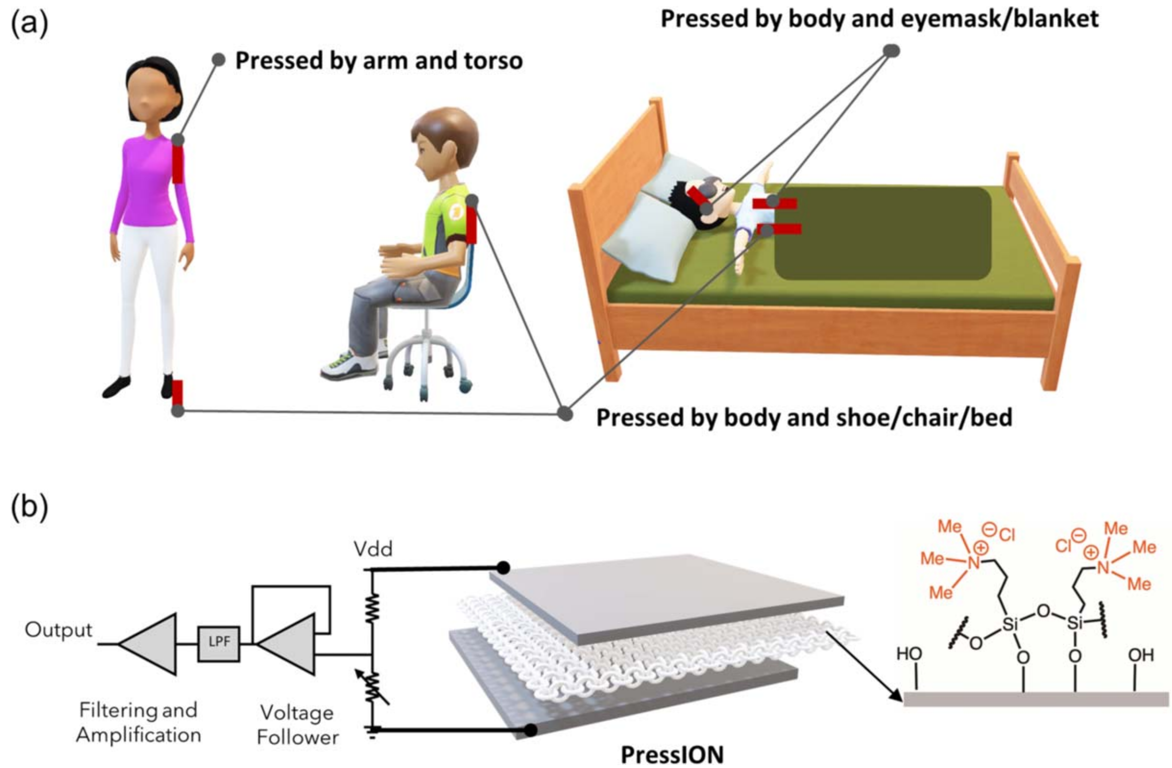

Figure 1. (a) Illustration of scenarios where an on-body pressure sensor may experience multiple applied pressures, which complicates extraction of desired physical and physiological signals. (b) Schematic illustration of PressION, a piezoionic pressure sensor for recording physiological signals, and the control circuit used for signal readout.

Experimental

Materials.—*N*-(triethoxypropylsilyl)-*N*,*N*,*N*,-trimethylammonium chloride, as a 50% solution in methanol, was purchased from Gelest. Silverized nylon was purchased from LessEMF.

Textile coating.—In a typical process, a 10 cm × 10 cm piece of cotton gauze was immersed into a 300 mL isopropanol solution containing between 5–25 volume% of a 50% solution of *N*-(triethoxypropylsilyl)-*N*,*N*,*N*,-trimethylammonium chloride in methanol and stirred for 30 min at 24 °C. Then the fabric was annealed at 120 °C for 2 h and rinsed with isopropanol.

Results and Discussions

The basic architecture of PressION follows all other piezoresistive sensors, i.e., an active layer sandwiched between two electrodes. However, the distinguishing features of this device are first, the all-fabric substrates, and second, the ionic conductivity of the active middle layer. Ions are classical particles that follow Newtonian mechanics, as opposed to quantum mechanics, and experience mass transport following Fickian dynamics. These characteristics mean that ionic conductivities are typically three-four orders of magnitude lower than those observed in electronic conductors and are, also, linearly dependent on the local ionic concentration. We posited that using ions as charge carriers in a pressure responsive device—a piezoionic device—would give us access to a wide detection range.

We identified a poly(siloxane) polymer, poly(*N*-propylsilyl-*N*,*N*,*N*,-trimethylammonium chloride), as an ideal ion-conductive coating for our desired application. Poly(siloxane)s are longstanding textile finishing treatments used to change the luster and handfeel of natural fabrics, such as a cotton; these coatings are also wash stable and resistant to mechanical washing.[26] The bulky quaternary ammonium side chain of the polymer used in this work remains surface bound to the underlying cotton substrate, but the smaller and biocompatible chloride anion remains mobile. Upon a solution-phase functionalization reaction (Fig. 2a) followed by a heat treatment and multiple rinsing processes, the siloxane moieties of this polymer bond covalently to the free hydroxyl groups of cotton. Scanning election microscopy (SEM) images (Fig. 2b) confirmed the formation of a conformal coating on the surface of cotton fabric. Energy-dispersive X-ray spectroscopy (EDX) confirmed the presence of silicon and chlorine atoms on the surface, which can only arise from the presence of the desired poly(siloxane) coating depicted in Fig. 2c. The uniform distribution of the atoms on the surface in the EDX map has already been reported in our previous study.[27]

To explore the effect of the concentration of the siloxane precursor used in the solution-phase functionalization process on the surface resistivity of the functionalized cotton gauze, we made five pressure sensors with five different concentrations of the siloxane precursor (5, 10, 15, 20, and 25 v/v%) and performed four-point probe measurements on the resulting coated fabrics. As shown in Fig. 2d, the data reveals that the surface conductivity increased with increasing siloxane precursor concentration. After 20 v/v%, the conductivity saturated, likely because the free hydroxyl groups present on the surface of the cotton are all bonded to siloxane moieties and the excess amount of siloxane molecules that are not surface bound are washed away by the rinsing process. The temperature dependence of the observed conductivity for a fabric coated in a 15 v/v% solution of the siloxane precursor confirmed that the primary charge carrier in this sample was ionic, not electronic (Fig. 2e).

The final PressION is created by sandwiching this ion-conductive fabric between two silverized nylon electrodes (Fig. 1b). The working mechanism of this pressure sensor is based on three major factors. Under the application of an inward pressure on the sensor, three simultaneous phenomena take place. First, as the air gap between the layers reduces, the number of available conductive percolation routes between the two conductive electrodes increases. Second, the overall thickness of the fabric decreases which leads to

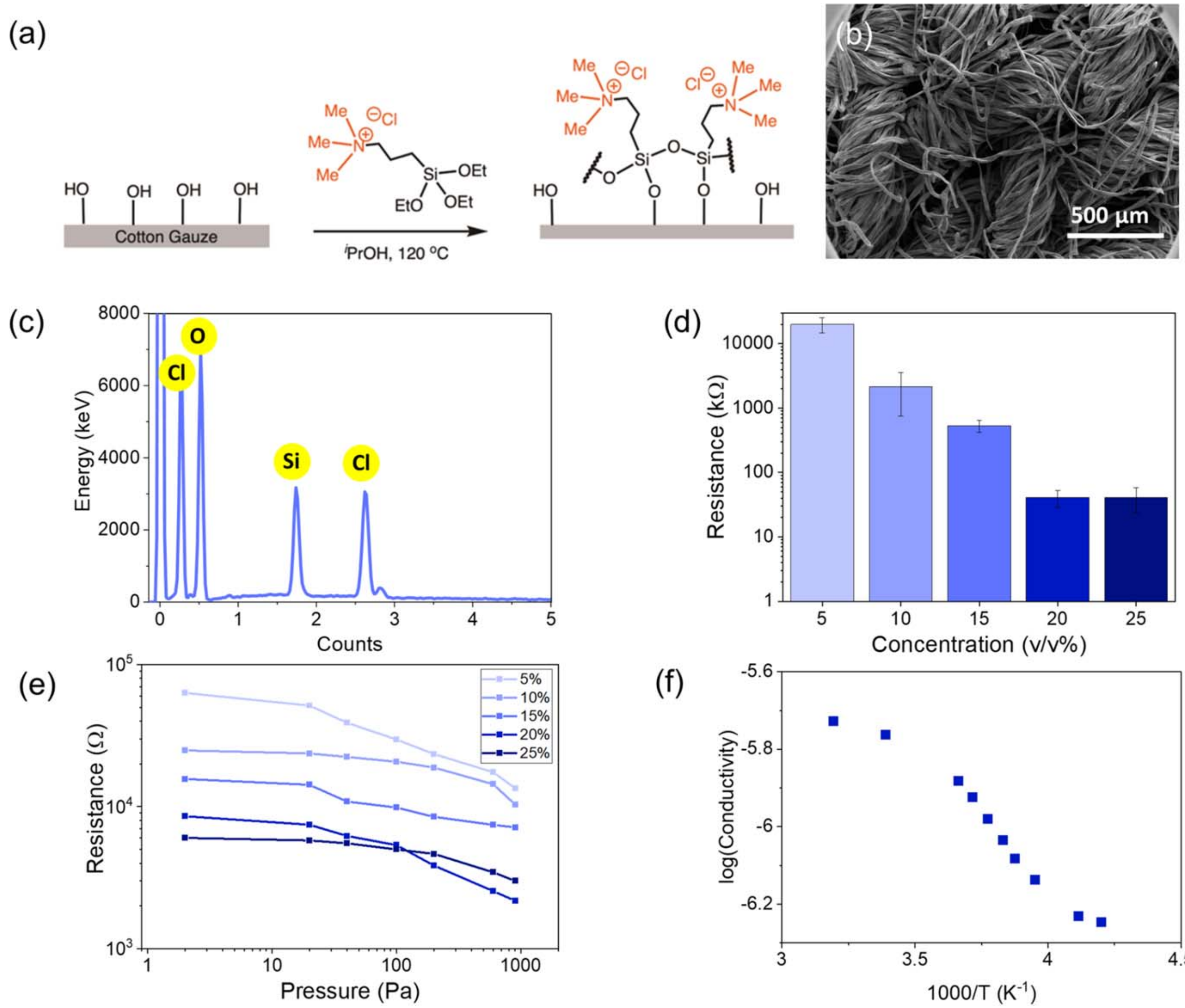

Figure 2. (a) The solution-phase functionalization process for creating ion conductive cotton. (b) Scanning electron microscope image of functionalized cotton gauze, indicating uniform coating. (c) Energy dispersive X-ray spectrum of functionalized cotton gauze showing the presence of silicon and chlorine atoms on the surface. (d) Observed surface resistance of functionalized cotton gauze with varying concentrations of siloxane precursor. (e) Pressure response of devices containing functionalized cotton gauze with varying surface resistance. (f) Temperature dependence of the observed conductivity of a device containing functionalized cotton gauze.

the change in the capacitance of the sensor. At the same time, the applied pressure leads to the physical movement and relocation of the mobile chloride ions in the ion-conductive middle layer, which gives rise to a piezoionic effect.[28] All these factors contribute to the reduction in impedance of the pressure sensor upon the application of pressure.

The medium-weave cotton gauze fabric used in this study lead to the best pressure sensitivity, over other fabrics of different weave densities and mesh sizes functionalized with the same poly (siloxane), in addition to showing lower saturation effects and higher signal stability over time. We also explored the effect of the concentration of the siloxane precursor used in the solution-phase functionalization process on the performance of PressION. As revealed in Fig. 2f, the response of the sensors stayed almost the same for all the precursor concentrations, but the absolute value of the voltage output varies due to the differences in surface resistivity. Since the sensor impedance with 15 v/v% concentration fell within the ideal range of resistivity that we look for our applications, we carried on our experiments with this optimal concentration.

Figure 1b shows the equivalent electrical circuit model for this pressure sensor. In order to capture coarse- and fine-grained changes in sensor resistance, we designed the microcontroller by adding a constant resistor (R_c) in series with the sensor, resulting in a voltage divider. This stage converts the resistance of the sensor to voltage following the formula:

$$V_o = \frac{R_s}{R_s + R_c} \times V_{dd}$$

Where V_o is the output voltage of the voltage divider, and R_s is the sensor resistance. Next, we placed a voltage follower stage to isolate the sensor from the amplifiers and to provide a low output impedance for next stages. The output of the voltage follower was not amplified and presents the static state of the sensor. Finally, we filtered the signals from unwanted sources in the analog domain and used a second order non-inverting amplifier to increase the signal-to-noise ratio and capture small voltage changes resulting from subtle vibrations, such as those from heartbeats and vocal cords.

As mentioned before, one of the major limitations of known pressure sensors is their failure to sense subtle static pressure signals in the presence of a second static base pressure. To simulate this scenario, we designed and performed an experiment (Fig. 3a) in which we, first, placed a 1 or 2 kg copper plate on top of the PressION (to approximate a static base pressure) and then applied an extra inward compression stress on the PressION using an extra weight. As seen in Fig. 3b, the sensor affords a temporary dynamic response upon being compressed by the extra weight, but then relaxes to a stable final readout whose absolute voltage value is lower than that afforded by the static weight alone. The difference between the initial baseline (with one weight) and the final readout with the added weight is the sensor response to the extra pressure. As seen in Fig. 3c. the sensor is still sensitive to extra applied pressures even under a constant base pressure of 2 kPa, which is the average body weight pressure exerted while sleeping on a bed.

In order to evaluate the sensitivity of the PressION to dynamic pressures, we performed an experiment in which we, first, placed a 1 or 2 kg copper plate on top of the PressION (to approximate a static base pressure) and then dropped a bouncing ball from the same height onto the sensor surface (Fig. 3d). The response time of the sensor was calculated to be 5 ms (Fig. 3e). As seen in Fig. 3f no significant change was observed in the dynamic signal in the presence of a static base pressure. Since the dynamic pressure is proportional to the kinetic energy of the dropped ball, and therefore to the height of the point from which the bouncing ball was released, we were able to change the impact applied on the sensor by changing the height of the release point. Although the signal power drops in the presence of a base pressure, PressION nonetheless responded linearly to an applied dynamic pressure, thus making it useful for capturing heartbeats and respiration while an individual is lying on the sensor (meaning that their exerted body weight acts as a static base pressure on the sensor).

Considering the fact that there are some key locations to detect physiological signals and body motions, we gently placed PressION on various parts of a participant's body to extract a plethora of different physical signals. We asked a participant to lie face-down on a bed, placed the PressION between their chest and the bed and were able to record heartbeats from the sensor (Fig. 4d). We captured respiration by placing the sensor on a participant's chest while the participant was lying face-down over it (Fig. 4h). Respiration could also be recorded when the participant was sitting or standing and the PressION was placed on the side, between the arm and torso (Fig. 4g), or gently held against the participant's back using a loose-fitting shirt (Fig. 4f). Artery pulses could be captured from the supraorbital artery on the face while wearing an eyemask (Fig. 4a). Comparatively weaker phonation signals were also detected while gently holding the PressION against a user's throat using a loosely-wrapped shawl (Fig. 4b) (in this case, an extra amplifier was employed). A reliable signal output was also obtained when a participant walked while wearing a shoe containing the PressION as an insole (Fig. 4e)—this data suggests a rugged platform to perform footwear evaluation studies. Lastly, placing the PressION on the inside of the elbow or back of the knee allowed for motion tracking at these joints (Fig. 4c). Each of these diverse signals can enable a host of deeper kinesiological and physiological studies and applications.

Conclusions

We introduce PressION, a first-of-its-kind all-fabric piezoionic pressure sensor that can be used to extract important physiological signals when loosely placed over various locations on the body. We identified an ion-conductive polysiloxane coating for cotton fabrics that allows for creation of pressure sensors capable of responding to a wide range of pressures, from the subtle pressures exerted by ballistic physiological signals to large-scale joint motions. We verified that the polysiloxane functionalized cotton fabrics were ion conductive using temperature-dependent conductivity measurements. We designed experiments to simulate sensing scenarios where subtle dynamic pressures are applied on top of a larger static base pressure and confirmed that the PressION responds linearly to

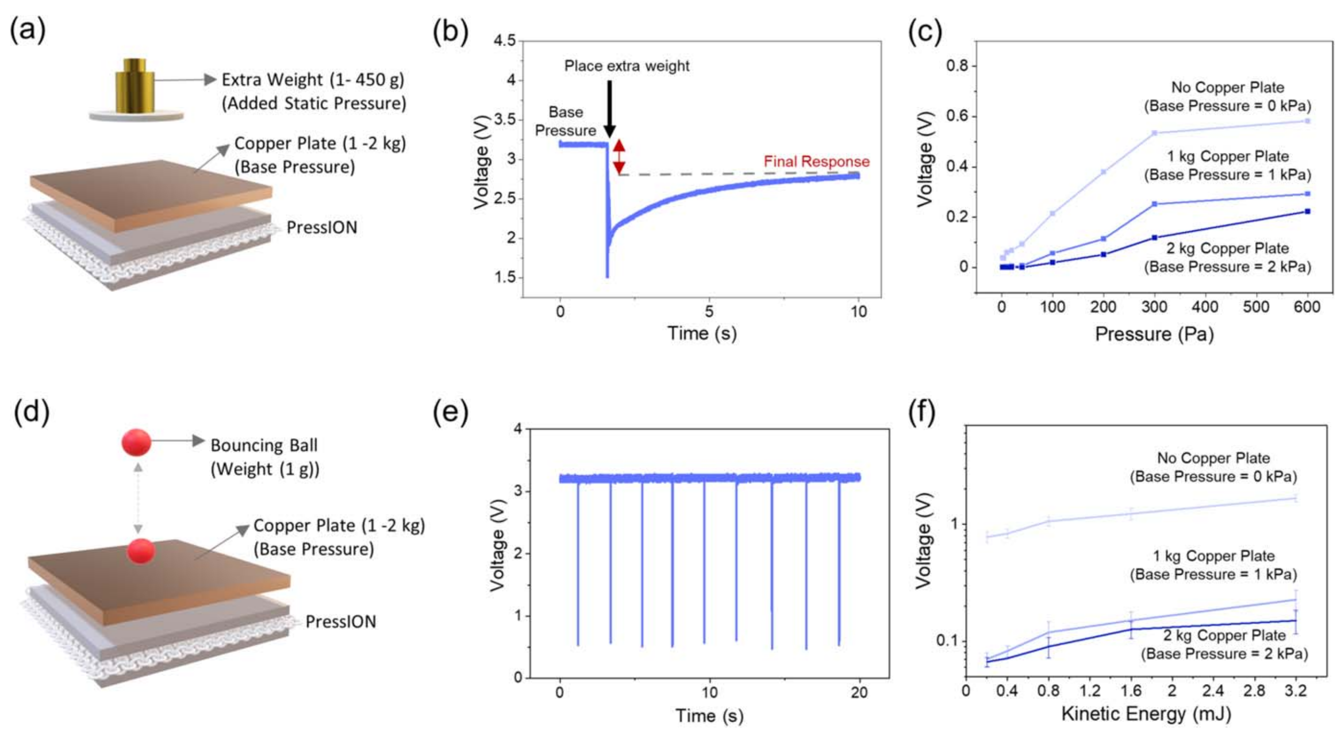

Figure 3. (a) Experimental setup to observe the effect of a base pressure on the static response of PressION. (b) Representative real-time signal output from the PressION for the experimental setup depicted in part (a). (c) Comparison of the response of PressION to static applied pressures in the presence of various base pressures. (d) Experimental setup to observe the effect of a base pressure on the dynamic response of PressION. (e) Representative real-time signal output from the PressION for the experimental setup depicted in part (d). (f) Comparison of the response of PressION to dynamic applied pressures in the presence of various base pressures.

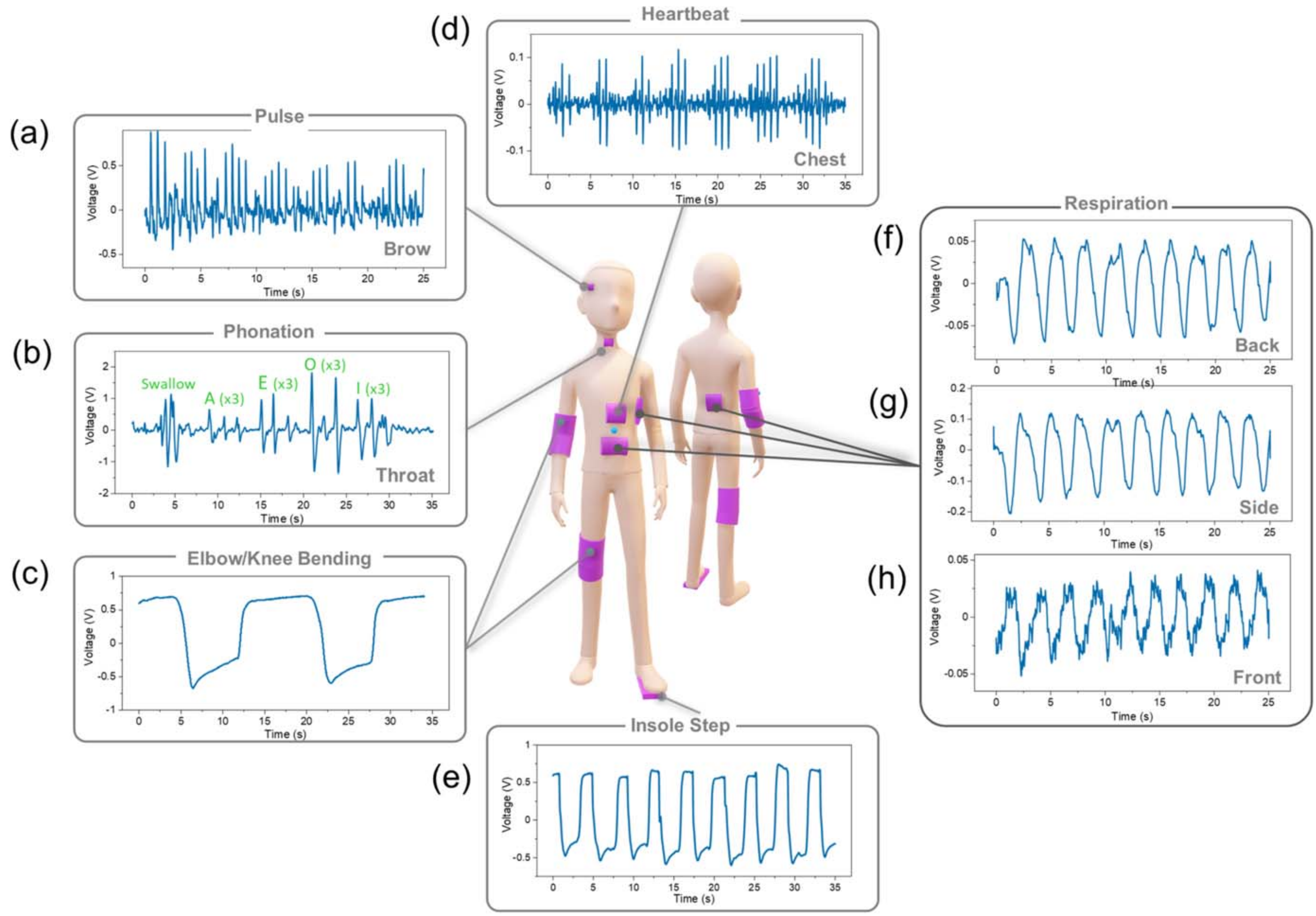

Figure 4. A variety of physiological signals can be extracted by gently placing PressION on different locations of the body: (a) artery pulse from the face; (b) phonation from throat; (c) joint motion at the elbow or knee; (d) heartbeat from the chest while lying face down on a bed; (e) steps from the sensor used as an insole in footwear; and respiration from the sensor placed on the (f) back, (g) side or (h) front of the body.

both static and dynamic pressures in the presence of base pressures as high as 2 kPa. We leveraged the pressure exerted between a body and external surfaces as an opportunity to embed this sensor into loose-fitting clothing and successfully demonstrated signal acquisition with the sensor placed in different body locations. We captured heartbeats from the chest while lying face-down on the sensor, respiration from the PressION placed on the side, back or front side of the body, phonation from the PressION gently placed on the throat, steps from the sensor used as an insole in footwear, artery pulses from the face, and joint motion signals with the PressION placed at the elbow or knee. The exceptionally wide range of physiological, joint motion and posture signals that can be revealed by PressION makes it a uniquely useful sensing element for wearable health and motion monitoring systems.

Acknowledgments

This work was funded by the National Science Foundation under agreement CSR Medium 1763524. S. Z. H. and T. L. A. also thank the David and Lucile Packard Foundation for providing partial support.

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