Supporting Information

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A Sensitive and Biodegradable Pressure Sensor Array for Cardiovascular Monitoring

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Supporting Information

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Supplementary Figure S1. Pressure response curves from two consecutive linear loading-unloading cycles performed on one representative element of the sensor array of 4x5 pixels, where it is shown that hysteresis is negligible.
Supplementary Figure S2. Cycling tests and stability of pressure response over 8000 cycles on a flat surface. a) Measured capacitance and b) relative capacitance change as a function of number of cycles. c) Pressure response curves measured before (red circles) and after (blue crosses) >8000 cycles of pressure loading-unloading. A small signal drift is measured, illustrated by the fact that $C_0$ (capacitance measured when no pressure is applied) increases by 16% and $C_{max}$ (capacitance measured when 80 kPa is applied) increases by 4% after 8000 cycles. This change has however little impact on the pressure response curves measured as illustrated in (c), in particular in the present context of relative blood pressure measurement.
Supplementary Figure S3. Device response at applied pressures in the range 0 to 270 kPa. The robustness of the sensor is illustrated by the unaltered pressure response curves after several runs. 1st run: a pressure is applied from 0 to 10 kPa and back, twice. 2nd run: the same is performed for a pressure ranging from 0 to 270 kPa. A total of 6 runs are performed in total, with reproducible pressure response curves that illustrate the sensor robustness at high pressures. The maximum pressure that is experimentally applied on the sensors is 270 kPa (experimental setup limitation). However, the maximum pressure that can be theoretically resolved with this system, based on a signal to noise ratio SNR ≥ 2 and for a minimum pressure to resolve of Δp ≥ 2 kPa, is calculated to be ~890 kPa. This theoretical value is calculated based on the sensor output characteristic (above 50 kPa the sensitivity decreases with increasing pressure following a logarithmic trend), and it doesn’t take into account the fact that the sensor integrity might be altered when applying such high pressures (not experimentally tested).
Supplementary Figure S4. Device operation while bent. a) Pressure response curves for various bending radii. b) Pressure sensitivity as a function of bending radius, measured at different pressure ranges, and normalized by the sensitivity in the flat configuration. For radii above 27 mm and pressure in the range 50-60 kPa, no significant change in the sensitivity is observed. In the pressure range 5-10 kPa, a slight decrease in sensitivity with increasing bending radius is measured, the sensitivity at radius 27 mm being ~91% of the value measured in flat condition. Similarly, in the pressure range 0.1-2 kPa, the sensitivity at radius 27 mm is ~83% of the value measured in flat condition.
**Supplementary Figure S5.** Tensile and compression tests on PGS samples after 1, 2, 3, 4, 5, 6 and 7 weeks of *in vitro* degradation. a) Applied tensile strain, b) measured tensile stress, and resulting stress-strain curves c) after 1 week, and d) after 7 weeks of incubation in PBS solution at 37 °C. e) PGS tensile stress-strain curves for weeks 0 to 7 of incubation, the tensile modulus decreases with increasing incubation duration. f) Applied compressive strain, g) measured compressive stress, and resulting stress-strain curves h) after 1 week, and i) after 7 weeks of incubation. j) PGS compressive stress-strain curves for weeks 0 to 7 of incubation. PGS stripes (width 4.4 mm, length 5 mm, thickness 2.4 mm) and PGS cylinders (diameter 7 mm, height 2.4 mm) are cut with laser from a 2.4 mm-thick PGS film. After 1, 2, 3, 4, 5, 6 and 7 weeks, 7 PGS stripes and 7 PGS cylinders are removed from incubator, washed in deionized water, dried with towel and air jet, and dried in vacuum during 24 hours. For each stripe, tensile tests are performed at 6 different strain levels (ε = 0.05 to 0.3), and for each cylinder, compression tests are performed at 5 different strain levels (ε = 0.05 to 0.25). For all experiments, the strain rate is 1 mm/mm per minute. A very small hysteresis is observed both in tensile and compression stress-strain characteristics, even after 7 weeks of incubation.

**Supplementary Figure S6.** Tensile and compression moduli of PGS after 1, 2, 3, 4, 5, 6 and 7 weeks of *in vitro* degradation. (a) Tensile modulus and (b) compression modulus as a function of incubation duration.
Supplementary Figure S7. Response characteristics of the pressure sensor after being stored for one week (a) in an oven at a temperature of 45 °C and relative humidity of 40%, (b) in a freezer at a temperature of -6 °C and relative humidity of 56%. (c) Wrist pulse wave signals measured with sensors having been exposed to extreme storage conditions. Even though (a) and (b) show a decrease of the maximum capacitance for both storage conditions because of the aging of the magnesium electrodes, the sensitivities at low and high pressure regime are not affected by the harsh storing conditions, and the sensors are still able to successfully measure the wrist pulse wave after one week of storage in extreme conditions (c).
Supplementary Figure S8. Pressure sensor response when exposed to a sweaty patient. (a) Different sweating conditions are evaluated using artificial sweat (Artificial Eccrine Perspiration 1700-0020, Pickering, artificial sweat pH 4.5 solution containing amino acids, minerals and metabolites). The evaporation rate is 0.000364 kg per hour. The wrist pulse wave is successfully measured when the sensor is placed on top of (b) dry skin, (c) wet skin, (d) when the skin is entirely covered with artificial sweat. (e) Evaluation of PHB/PHV top and bottom substrates as a protection layer against the corrosion of the metallization layers. A droplet of artificial sweat is deposited on top of the sensor, covering the all measurement area. The sensor output (measured capacitance) is stable over time with less than 1% standard deviation.