Supporting Information

Miniaturized Piezo Force Sensor for Medical Catheter and Implantable Device

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Author contributions:

*: Developed the interface electronics for the medical catheter and implantable device and designed the force sensor, respectively.

‡: Developed the medical catheter.

•: Supervised the work.

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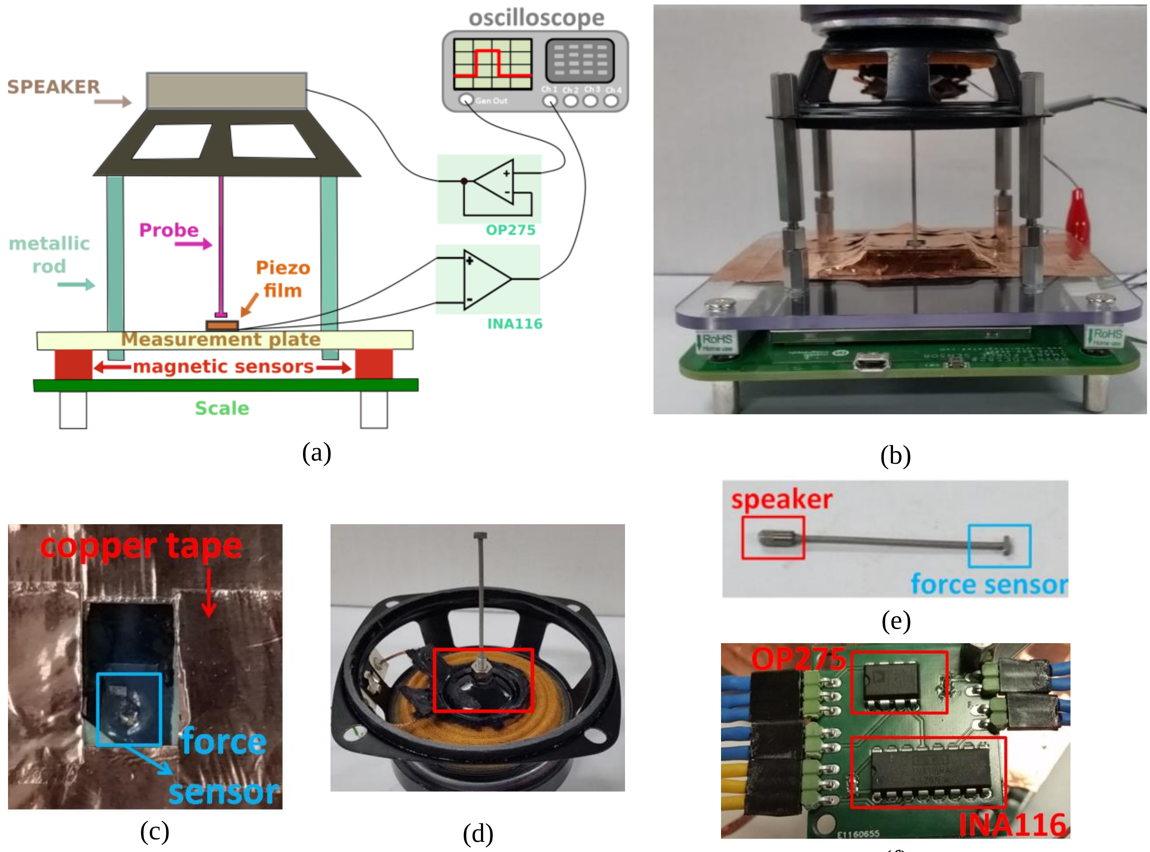
Experimental Setup for Cyclic Testing of the Force Sensor

A setup consisting of an audio speaker (FSR 8, Visaton, Haan, Germany) and ABS-polymer probe was assembled to apply axial loads directly over the force sensor located centrally within the structure displayed in Figure S1. The speaker produces movements of the probe in the vertical direction according to the programmed electrical signal. The non-conductive plastic probe touches the force sensor at one end (called "hammer-like" termination), with adjustable height through a millimetre-sized metallic screw-nut glued to the membrane of the speaker in the other end. Vibrations of the membrane induce displacements in the micro and lower millimetre ranges for the probe that reach the force sensor. To achieve repeated load application, the speaker is stimulated by a rectangular voltage signal with controllable amplitude, frequency, and duty-cycle. Since the input impedances of traditional speakers are too low ($\leq$ 8 Ω), they cannot be directly interfaced with the output signal produced by a waveform generator (MSO-X 3054A, Agilent Technologies, Santa Clara, CA, USA), without degrading drastically its amplitude. To solve this interfacial issue, an operational amplifier (OP275, Analog Devices, Norwood, MA, USA) in a buffer topology was intercalated in-between to deliver a theoretical zero Ohm signal to the speaker. In order to establish a dependency between the applied rectangular signal with the force produced by the probe, the entire structure was fixed on the top of a commercial force sensing platform (MAXREFDES82, Maxim Integrated, San Jose, CA, USA) that can detect force as low as 1 mN (or, equivalently, 0.1 g in weight) and points-of-contact all over the measurement plate. Then, the platform sends the measured values through an Universal Serial Bus (USB) interface to the recording computer, after initial calibration in order to zeroing any pre-loaded weight at the beginning of the experiments (e.g. entire metallic structure and speaker), therefore converting directly the applied voltage signal to an equivalent of force. An experiment was performed by applying repeatedly the same signal (20% duty-cycle, frequency of 2 Hz, no voltage offset) with



(a) (b) (c) (d) (e) (f)

Figure S1. (a) Contextualization image of the experimental setup idealized for electrical characterization and force application over the piezo film (force sensor). (b) Actual picture of the assembled setup, consisting of an audio speaker (top) supported by metallic rods attached to the force measurement plate of a commercial scale (bottom), with an acoustically activated probe acting on the force sensor (middle). (c) Force sensor (centre) surrounded by copper tape along its external perimeter to shield for electromagnetic interferences. (d) Image of the speaker employed in the setup (and facing upwards) to show the attachment mechanism between the vibrational membrane and the probe (metallic screw-nut). (e) ABS-polymer probe reaching the force sensor at the "hammer-like" termination. (f) Electronic circuit board developed to interface the speaker with the waveform generator of an oscilloscope (OP275) and between the force sensor and recording channel (INA116).

different amplitudes in the range $\in$ [1, 5] V to the platform while measuring the developed force, obtaining the linear relationship exhibited in Figure S2. Conversion from force to pressure equivalent involved only dividing the obtained values by the area of force application over the sensor (1 mm x 0.5 mm).

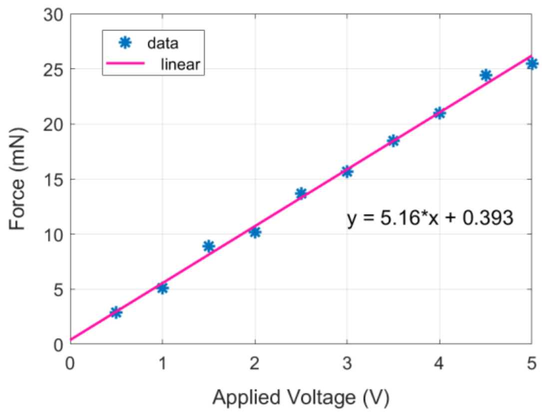


Figure S2. Experimental data and linear fit obtained between the voltage signal applied to the speaker (amplitude) and force detected by the commercial scale.

Electronic Design of the Interface for the Medical Catheter

The developed force sensors attached to the medical catheter were connected to a customized printed circuit board (PCB), whose electronic schematic is shown in Figure S3a. Two amplifiers (INA116, Analog Devices, Norwood, MA, USA) were employed to amplify the signals originated from the sensors with gain set to 100 V/V, followed by low-pass filtering (cut-off frequency of 125 Hz) and digitization by a 16-bit analogue-to-digital converter (ADC), with sample rate set to 250 samples *per* second (SPS) in two different channels of the embedded ADC (AD974, Analog Devices). The digital samples are afterwards sent to a microcontroller (MCU, PIC24FJ64GA002, Microchip, Chandler, AZ, USA) before being transferred to a computer through an USB chip transceiver (FT234XD, FTDI Chip, Glasgow, UK) at speeds of 115200 bits *per* second (bps) and galvanically-isolated transmission lines. The type of digital isolator employed on electronics (ADUM1201, Analog Devices) works as a dual-port bridge connecting the signals arising at the PCB side powered by batteries only (± 9V) to the electronic part supplied by the USB connection.

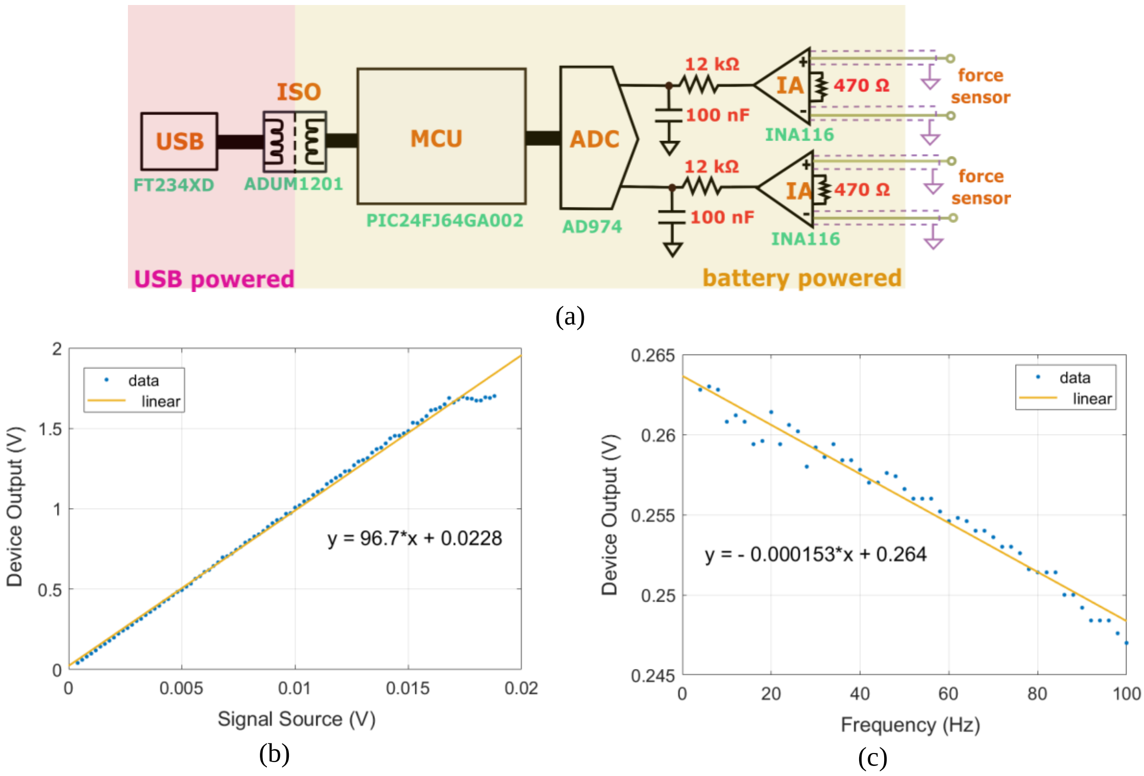


Figure S3. (a) Simplified schematic of the PCB developed to acquire the output voltage signals generated by the force sensors attached to the medical catheter, with galvanically-isolated transmission lines to an external computer. (b) Electronic characterization of the complete acquisition channel as a function of the amplitude of a sinewave (10 Hz frequency) injected at the terminal inputs of the IA. (c) Output voltage swing as a function of the frequency of the sinewave, with constant amplitude set to 3 mV.

Electronic characterization of the complete acquisition channel was additionally performed to evaluate the performance of the channel to a sinewave signal applied at the input terminals of a single IA, without attachment of the force sensor. Different amplitudes and frequencies of the sinewave were tested, obtaining the curves in Figures S3b and S3c, respectively, which were afterwards used to assess the linearity between input-output signals at the level of the IA, as well as output voltage swing in the lower frequencies (≤ 100 Hz). Discussion on the obtained levels is done in the main part of the manuscript.

Electronic Design of the Implantable Device and Pressure Measurement Principle

The electronic schematic for the developed implantable device is shown in Figure S4a. An NFC energy harvesting module (NT3H1101, NXP Semiconductors, Eindhoven, The

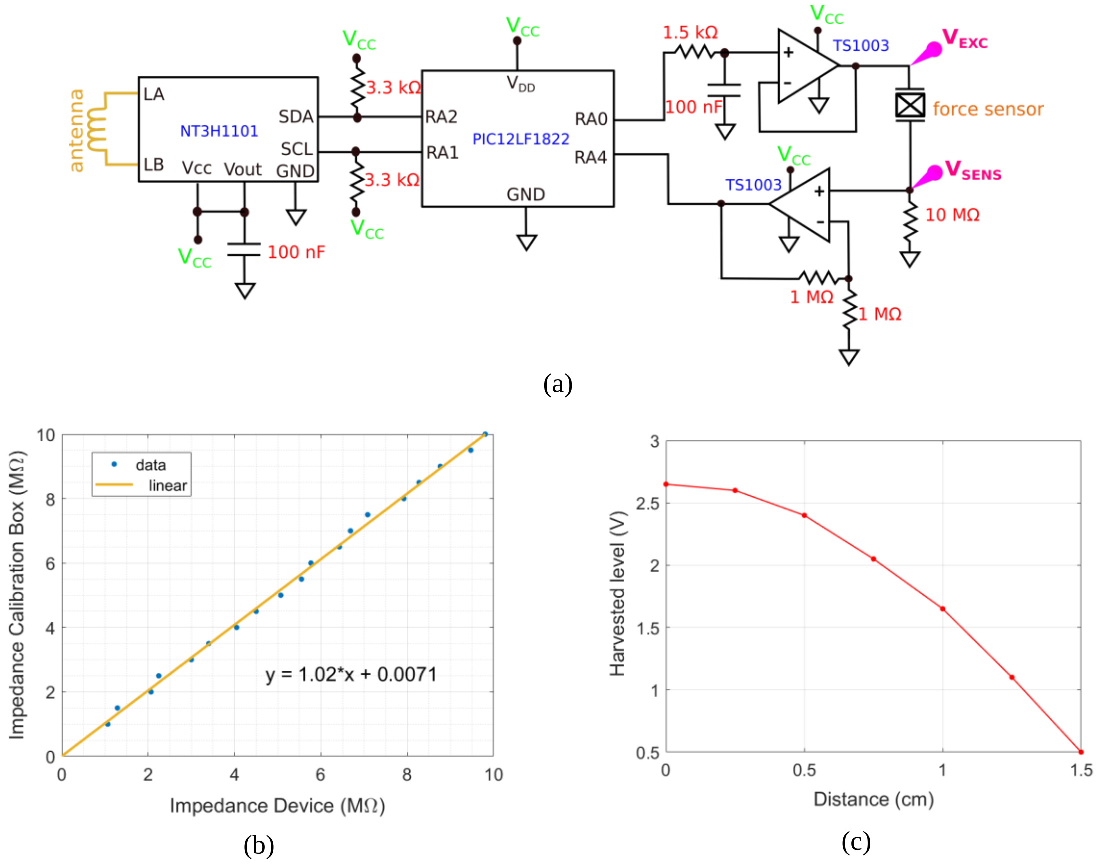


Figure S4. (a) Electronic schematic for the implantable device interfacing a single force sensor, with NFC power harvesting and data communication lines. (b) Impedance level measured by the implantable device for different resistors (calibration box) substituting the force sensor. (c) Harvested voltage profile as a function of the distance between the implantable device and mobile phone in open air, with aligned antennas along the transmission axis.

Netherlands) was used to generate electric power from the RF field produced by a mobile phone through a 30-turns loop antenna. Rectification of the AC signals is also performed inside the NFC module, delivering a filtered DC voltage level that can be used by the embedded microcontroller or MCU (PIC12LF1822, Microchip) and operational amplifiers (TS1003, Silicon Labs, Austin, TX, USA) interfacing directly the proposed force sensor. We employed a voltage-divider topology to measure the impedance (amplitude only) of the piezo element (force sensor) when stimulated by a signal with amplitude of 1.8 V (harvested level) and frequency of 1 kHz. This signal is directly imposed over one input terminal of the force sensor through a buffer amplifier configuration and generated by the MCU, firstly as a square signal followed by low-pass filtering (cut-off frequency around 1 kHz) to allow only the fundamental frequency to pass relatively undisturbed by the filter. The voltage signal, V_{SENS}, arising in-between the force sensor and the external resistor (10 MΩ) connected to the ground can then be calculated as,

$$V_{SENS} = \frac{10\ \mathrm{M\Omega}}{10\ \mathrm{M\Omega} + Z_{force}} V_{EXC} \tag{1}$$

where Z_{force} is the impedance amplitude of the force sensor and V_{EXC} the excitation signal. Amplification of V_{SENS} is performed by a non-inverting amplifier with gain set to 2 V/V before digitization by the microcontroller. Excitation at 1 kHz was selected in accordance to the experimental measurements performed with a single force sensor connected to a commercial impedance analyser (as described in the main manuscript), in which a lower impedance value (≤ 10 MΩ) is obtained in comparison to DC.

Digitization of the piezo voltage samples is performed inside the internal ADC of the MCU with 10-bit resolution, followed by estimation of the amplitude of impedance in the digital domain by the Discrete Fourier Transform (DFT), necessary to calculate the real and imaginary coefficients of the acquired signal component at 1 kHz. A programmed acquisition routine inside the MCU uses 40 samples that span 5 consecutive periods of the excitation signal (or, equivalently, 8 kSPS) to produce the previous coefficients as,

$$\mathrm{Real}\{X_{25}^{SENS}\} = \sum_{n=0}^{N-1} x_n \cos\left(-j2\pi \times 40\frac{n}{N}\right),\ \mathrm{Imag}\{X_{25}^{SENS}\} = \sum_{n=0}^{N-1} x_n \sin\left(-j2\pi \times 40\frac{n}{N}\right) \tag{2}$$

where x_n is the 40-sample data vector, n is the sample index, N the total number of samples and X_{25}^{SENS} the frequency component at 1 kHz. Conversion into impedance amplitude is performed already outside the implantable device by Eq. 3, which involves solving Eq. 1 in relation to Z_{force} and considering a constant value for V_{EXC}, as the excitation signal is not acquired by the implantable device to save computational cycles (inside the microcontroller) and associated conditioning electronics.

$$Z_{force} = 10\ \mathrm{M\Omega} \cdot \left[\frac{V_{EXC}^{r.m.s.}}{\sqrt{\mathrm{Real}\{X_{25}^{SENS}\}^2 + \mathrm{Imag}\{X_{25}^{SENS}\}^2}} - 1\right] \tag{3}$$

The computed coefficients (2 x 40-bit) are then transferred to the NFC module through digital communication lines. This module can hold up a maximum of 1 kB of data, which can be readily accessed by the RF field (contactless measurement) produced by the mobile phone. The update rate for the proposed estimation routine is around 50 ms, which implies sampling pressure signals with a maximum frequency of 20 Hz only.

Figure S4b shows the response of the implantable device with known resistor values intercalated at the terminals of the force sensor (replacement) and provided by a calibration box (type 008-B, Cropico, Peterlee, UK). The tested values were in the range from 1 MΩ to 10 MΩ in 500 kΩ steps, with the linear fit obtained between the resistance values and those estimated by the implantable device also depicted in the graph. This calibration curve was programmed afterwards inside the mobile phone's *app* to estimate more accurately the impedance variations produced by the force sensor when applied to real experimentation.

Finally, in terms of power consumption, the implantable device requires 100 μA of current for proper operation, which combined with the minimum voltage level for device operation (1.8 V) results in a consumption profile of 180 μW. The harvested voltage profile as a function of the gap between the implantable device (reception) and the mobile phone (transmission) is depicted in Figure S4c, when both the antennas are perfectly aligned along the transmission axis. Within these conditions, the operation of the device can be assured for distances up to one centimetre from the phone.