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A novel sensor for measuring the inner pressure of catheters for clinical use

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Abstract—In numerous clinical procedures requiring the infusion of a fluid (gas or liquid) it is necessary to monitor the infusion pressure. To this purpose, several devices have been developed so far. Such devices are generally disposable, with a pressure transducer in direct contact with the infusion fluid. In this paper we describe the design and development of a dry infusion pressure measurement system, sensing the pressure-induced radial displacement of the external surface of polymeric catheters. Such displacement is first amplified using a near-to-singularity elastic beam and then transduced using a Hall effect displacement sensor. We start describing the working principle of the whole system, then we detail the design process. Finally, we present the static calibration on two widely used biomedical catheters.

Index Terms—Pressure sensor, Sensing infusion pressure, Hall effect pressure sensor.

I. INTRODUCTION

Many clinical procedures involve the infusion of a pressurized fluid into an anatomical compartment. In such procedures, infusion pressure must be monitored for safety reasons. Important examples include the evaluation of intra-abdominal pressure [1], [2], [3] and pressure monitoring during epidural needle insertions [4], [5], [6].

Most of the existing commercial devices comprise a pressure sensor inserted along the fluidic line using a 3-way valve or directly embedded in the syringe body or in the plunger.

In both configurations, the pressure sensor is in direct contact with the infusion fluid. To solve sterility issues, the sensors are disposable. In the case of reusable sensors it is necessary to avoid any contact between the device and the infusion fluid to mitigate contamination risks.

Tesei et al. [5] has recently addressed the problem of monitoring the pressure exerted by the clinician on a syringe plunger during an epidural needle insertion. In their work, Tesei et al. mounted a piezoresistive force sensor (FSR) on the syringe plunger [5]. Evidently, because of the friction between the plunger and the inner surface of the body of the syringe, such measure only provides an indirect estimation of the infusion pressure, unless friction is properly characterized before infusion.

Another approach for mitigating contamination risks without friction issues has been recently proposed by Kartmann et al. [7], who presented a capacitive pressure sensor capable

to sense the catheter expansion caused by fluid pressure. In their configuration the fluid acts as the dielectric material between two electrodes. During catheter expansion both the distribution and the quantity of the dielectric (i.e. the fluid) vary, thus affecting the capacitance. The proposed sensor, calibrated using DI water, exhibits good sensitivity (0.135 V/kPa) and resolution (3.75 mmHg for pressures up to 22.5 mmHg or 12.0 mmHg for pressures up to 262.5 mmHg), but its metrological performances depend on the dielectric properties of the fluid itself.

In this paper we describe a novel dry pressure sensor capable of evaluating fluid pressure from the measurement of the radial displacement of the external surface of a polymeric catheter. An *ad hoc* designed mechanical system, mainly consisting in a close-to-singularity beam, amplifies the small radial displacements, which are then measured using a Hall effect sensor. The relation between radial displacements and internal pressures have been found using static calibration procedures.

The proposed sensor does not include sliding surfaces and is therefore not affected by friction phenomena. Moreover, as it will be shown later, its response is essentially linear.

Although the transduction principle can be theoretically adopted to develop pressure sensors for a variety of applications, we derived the design requirements considering a specific application scenario, i.e. the treatment of intervertebral disc degeneration (IVD).

IVD is indeed among the main causes of chronic low back pain [8] and its medical treatments include conservative, invasive or regenerative strategies. The last one is receiving an increasing attention, since it represents a promising approach. Regenerative strategies consist in delivering Advanced Therapy Medicinal Products (ATMP), such as growth factors, stem cells or hydrogels, into the intervertebral space, until the physiological pressure of the intervertebral disc is restored. For a healthy subject lying prone the value of the intervertebral disc pressure is about $p_{max} = 835 \text{ mmHg}$ [9]. In this paper we consider p_{max} as the reference maximum pressure to be measured.

The paper is structured as follows. Section II is dedicated to the description of the mechanism adopted to measure the radial deformation of the catheter. A first subsection provides the tools, derived from the Theory of Elasticity, to evaluate the radial deformation induced by pressure; a second subsection is devoted to the design of the mechanism to amplify displacements. Numerical results are collected in dedicated paragraphs. Section III describes the actual prototype assembly, while

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Section IV is dedicated to the experimental calibration. Finally, discussion and conclusions are presented in the last section (Section V).

II. DESIGN

In this section we will present the theoretical background behind the design of the pressure sensor. In particular, we will first review some classical results from Theory of Elasticity relating inner pressure and outer deformation of a catheter, which is modeled as an elastic tube in regime of small deformations. Then, we will describe the mechanical means adopted to amplify the radial displacements in order to improve the sensitivity of the sensor.

Such amplification means hinges a round a slightly flexed beam, i.e. close to a singular configuration, which is compressed when the catheter expands. The displacement of the mid-point of the beam, read by a Hall effect sensor, is then related to the inner pressure using a calibration procedure. As detailed below, the design of the displacement amplification device is mainly based on FEM simulations.

A. Evaluation of catheter radial deformation

The catheter is modeled as an elastic tube (Young modulus: E ; Poisson's ratio: ν) with internal and external radii respectively denoted with r_i and r_e . The (relative) internal pressure is denoted by p_i . Such pressure is the only load acting on the tube.

Deformations (ϵ) are linear functions of principal stresses (σ). Subscripts r , θ and l will respectively stand for radial, tangential and longitudinal. The constitutive equations are:

$$\begin{cases} \epsilon_r = \frac{1}{E}(\sigma_r - \nu\sigma_\theta) \\ \epsilon_\theta = \frac{1}{E}(\sigma_\theta - \nu\sigma_r) \\ \epsilon_l = \frac{\nu}{E}(\sigma_r + \sigma_\theta) \end{cases} \quad (1)$$

The principal stresses σ_r and σ_θ are:

$$\begin{cases} \sigma_r = -\frac{1-\rho^2}{\rho^2} \frac{\gamma^2}{1-\gamma^2} p_i \\ \sigma_\theta = \frac{1+\rho^2}{\rho^2} \frac{\gamma^2}{1-\gamma^2} p_i \end{cases} \quad (2)$$

where $\gamma = r_i/r_e$ and $\rho(r) = r/r_e$ ($\gamma < \rho(r) < 1$).

The radial displacement $u(r_e)$ of the outer surface can be calculated as $u(r_e) = r_e \epsilon_\theta$. Equations 1 and 2 yield to:

$$u(r_e) = r_e \frac{2}{E} \frac{\gamma^2}{1-\gamma^2} p_i \quad (3)$$

The maximum radial displacement occurs when $p_i = p_{max}$.

1) *Numerical values of radial displacement:* Two commercial catheters in PVC ($E = 6.0$ MPa) have been chosen for the calculation of the radial displacements. One catheter, in the following referred to as catheter 1 (Likset, MULTIMEDICAL srl), has $r_i = 0.75$ mm and $r_e = 1.25$ mm; the second catheter, in the following catheter 2 (Deltaconnector, DELTA MED srl), has $r_i = 1.5$ mm and $r_e = 2$ mm.

Using equation (3) the respective maximum radial displacements (when $p_i = p_{max}$) can be calculated:

$$u_{mod1}(r_{ecat1}) = 2.37 \mu m \quad (4)$$

$$u_{mod2}(r_{ecat2}) = 8.68 \mu m \quad (5)$$

Such maximum displacements are too small to be measured with a good resolution. Therefore, a displacement amplification means is deemed necessary.

B. Amplification mechanism

Axially compressed straight beams buckle if the compression force exceeds Euler's critical load. During buckling, a small displacement of one end of the beam corresponds to a large displacement of the mid-point of the beam in the direction orthogonal to the beam axis. The beam can be considered as an elastic transmission, with an input port at the end where the compressive force is applied and an output port corresponding to the mid-point, where the displacement e_0 is produced (Fig. 1). In regime of small displacements, the output deflection, albeit small, would be much larger than the input displacement. This principle is at the basis of the displacement amplification mechanism adopted in the sensor. In our case, the input force is provided by the radial expansion of the catheter, while a displacement sensor (Hall effect sensor) is connected to the output port. It should be noted that a straight beam would exhibit a transverse displacement at its mid-point only when the compressive force exceeds the critical Euler's load. It means that the transmission mechanism would not amplify any input displacement corresponding to an input force below the critical load. This would result in the insensitivity of the sensor to small radial displacements of the catheter, i.e. to small internal pressures.

To circumvent this limitation, a pre-curved beam is adopted in place of a straight one. If the beam is a circular arc, its curvature can be expressed as the angle β shown in Fig. 1. Another advantage of using a pre-curved beam in place of a straight beam is that the direction of flexion caused by compression is geometrically determined.

We should also recall here that the compressive force is produced by an inherently compliant structure, i.e. the pressurized elastic catheter. An ideal transmission mechanism for displacement amplification should be perfectly compliant, i.e. it should have no stiffness counteracting the expansion of the catheter. This requirement is evidently impossible in practice, since the beam has its own stiffness.

Therefore, the angle β is a design parameter to be defined as a trade-off between two opposite requirements: small angles (i.e. almost straight beams) would generate larger displacement amplifications, but would correspond to a stiffer transmission opposing catheters expansion. Conversely, larger β would generate smaller amplification, but would result in more compliant beams, allowing larger catheters expansion.

To identify an adequate value for β , a simple lumped element model was adopted, where both the pressurized catheter and the curved beam are represented as linear springs mounted in series (Fig. 2). Here, k_c represents the stiffness of the catheter, while $k_f(\beta)$ represents the stiffness of the curved beam. Evidently $k_f(\beta)$ is a decreasing function of β .

The length of the spring k_c varies from l_0 to l'_0 as the internal pressure increases from 0 to p_{max} (Fig. 2). When not

Fig. 1: Parameters describing the geometry of the near-to-singularity beam: e_0 is the vertical displacement due to the application of the compressive force exerted by the pressurized catheter (5); β is the curvature of the pre-curved beam. One end of the beam is fixed (1), the other one is constrained so that can only translate horizontally (2). (Left) Rest configuration: the catheter is unpressurized (3). (Right) The pressurized catheter (5) exerts a compressive force on the beam, consequently the translating end (2) moves horizontally and the beam is bent. A magnet (4) placed on the beam allows to measure the vertical displacement of the mid-point of the beam by means of a Hall effect sensor. It is possible to relate the mid-point displacement to the catheter inner pressure.

Fig. 2: An internal pressure (red arrows) acting inside the catheter causes a radial expansion. Both the catheter and the beam can be modeled as a series of two springs: k_c and k_f respectively are the stiffness of the catheter and the beam.

in contact with the beam, as an effect of the internal pressure spanning its whole range, the free end of the spring k_c would span a distance $\Delta l = l'_0 - l_0 = 2u(r_e)$ according to (3).

When the second spring $k_f(\beta)$ is mounted in series to k_c (Fig. 2), the contact point A moves to a new equilibrium position, spanning a distance s ($s < \Delta l$).

With reference to fig. 2, at equilibrium, the force on the catheter (F_c) equals the force on the curved beam (F_f):

$$F_c = k_c(\Delta l - s(\beta)) = k_f(\beta)s(\beta) = F_f(\beta) \quad (6)$$

Therefore:

$$s(\beta) = \Delta l \frac{k_c}{k_f(\beta) + k_c} \quad (7)$$

and

$$F_f(\beta) = \Delta l \frac{k_f(\beta)k_c}{k_f(\beta) + k_c} \quad (8)$$

When the catheter is in contact with the beam, its expansion causes the mid-point of the curved beam to move vertically spanning a distance $e(\beta)$. In regime of small displacement $e(\beta)$ is directly proportional to the compressive force ($F_f(\beta)$). Therefore, the following function:

$$g(\beta) = \frac{e(\beta)}{F_f(\beta)} \quad (9)$$

only depends from the geometrical and mechanical parameters of the beam.

The function $e(\beta)$ can be derived from (8) and (9) as follows:

$$e(\beta) = \Delta l k_c \xi(\beta) \quad (10)$$

where $\xi(\beta)$ is:

$$\xi(\beta) = \frac{k_f(\beta) g(\beta)}{k_f(\beta) + k_c} \quad (11)$$

The optimal value of β is the one which maximizes the objective function $\xi(\beta)$. To maximize $\xi(\beta)$ the following quantities must be evaluated: k_c , $k_f(\beta)$ and $g(\beta)$. The evaluation of this quantities requires the adoption of numerical means.

1) *FEM simulations*: FEM simulations were performed on catheter 1, being the one with the highest radial stiffness (k_c). The stiffness k_c was evaluated by simulating the compression of the catheter. With reference to Fig. 3, a rectangular probe A applies a transverse force F (ranging from 0 N to 6.4 N) on the external surface of the catheter.

The simulations, performed using Comsol Multiphysics (COMSOL Inc., Burlington, MA, USA), return the radial displacements (δ , in μm) of the nodes in contact with surface A. The stiffness k_c is evaluated as $k_c = F/\delta$. Assuming a linear model for the catheter deformation (i.e. the constitutive equations), k_c is independent from the imposed internal

Fig. 3: Virtual test bench for the evaluation of the catheter stiffness. After imposing a vertical force F , the radial displacement δ of the catheter wall is calculated through FEA simulations. The internal relative pressure p was set to 75 mmHg.

Fig. 4: Force/displacement plot of the catheter for several applied load.

catheter pressure. It is worth noting that imposing an higher internal catheter pressure cause the reduction of possible numerical noise effects. Therefore, the catheter is pressurized with an internal pressure $p = p_{max}$. To minimize boundary effects, the width of A was chosen to be ten times (i.e. 25 mm) the external radius of catheter 1. The force (F) versus displacement (δ) plot is shown in Fig. 4. The value of k_c is $k_c = 14000 \text{ N/m}$.

To determine $k_f(\beta)$ and $g(\beta)$, simulations were performed on curved beams, keeping constant the width and height of the cross-section ($w \times h = 12 \times 0.3 \text{ mm}^2$) (Fig. 1) and the length of the beam axis (50.5 mm). The angle β was varied from 3° to 21° by increments of 2° and a displacement ($\Delta l = 4.75 \mu\text{m}$) was imposed to the free end of the beam, the other one being clamped. FEA simulations provided the force $F_f(\beta)$ and the displacement e_0 . $k_f(\beta)$ was evaluated as $F_f(\beta)/\Delta l$, while $g(\beta)$ was calculated as the ratio $e_0/F_f(\beta)$. Finally, $e(\beta)$ was analitically evaluated using eq. (10).

As expected, $\xi(\beta)$ exhibits a maximum, which in our case occurs at $\beta = 9^\circ$ (fig. 5). For this value, we have $k_f(\beta = 9^\circ) = 15760 \text{ N/m}$, $g(\beta = 9^\circ) = 7.153 \cdot 10^{-4} \text{ m/N}$ and $e(\beta = 9^\circ) = 26.51 \mu\text{m}$. Such value of β was adopted in the final design of the beam.

Fig. 5: Objective function ξ as a function of β . The displacement is maximum when $\beta = 9^\circ$.

Fig. 6: Schematic view of the pre-curved beam interacting with the catheter. Left: rest configuration of the system; Right: deformed configuration of the system.

When the catheter C expands, the moving probe A is pushed purely horizontally due to the constraint constituted by the double pendulums E. The probe A bends the curved beam B. When the curved beam bends, the magnet D moves vertically along its axis of a quantity $e(\beta)$. The displacement $e(\beta)$ is measured using an Hall effect sensor.

III. PROTOTYPE DEVELOPMENT

Figure 6 provides a schematic overview of the sensor. The moving probe A is in contact with the catheter C and supported by two pairs of double pendulums E, assuring purely horizontal movements. The moving probe, pushed by the expanding catheter, bends the curved beam B. A magnet D is positioned in the middle of the beam, where the maximum deflection occurs ($e(\beta)$). When the beam bends, the magnet (ams, AS5000) moves vertically along its axis. A digital Hall effect displacement sensor (ams, AS5510) with an output ranging from 0 to 1024 counts, measures such displacement.

The body of the sensor is comprised of three parts (F, G, H in Fig. 7) 3D printed in Polyamide (PA 2200). Part G works as a rigid frame for the curved beam B and the double pendulums E. Moreover it holds the PCB I with the Hall effect displacement sensor L. A clamp M positioned at the mid-point of the beam, holds the magnet. Catheters C with different diameters can be inserted thanks to an adjustable mechanism supported by part H, constituted by a radial bearing mounted eccentrically with respect to its axis. Turning the lever N, the outer rings of the bearing pushes the catheter against the moving probe.

Fig. 7: Assembly sequence of the sensor: Part G, the housing of the beam, is inserted in part F. Part H embeds the mechanism for the insertion of catheters with several diameters. It fixes the catheter C to parts G and F. Turning the lever N, the catheter C is pushed against the double pendulums E (dimensions are in millimeters).

IV. EXPERIMENTAL CHARACTERIZATION AND CALIBRATION

A first set of experimental measures was performed by imposing known displacements in the range $0 - 8 \mu\text{m}$ with steps of $2 \mu\text{m}$ to the free end of the beam using a linear slider (Newport, M460P-Series, Linear Stage). Through these measurements it was possible to relate the displacement of the moving probe (A, Fig. 6) with the output of the Hall effect sensor, directly related with the deflection of the beam. For each known displacement, the measurement was repeated five times ($N=5$) in order to consider the effects of extraneous variables on the measured data. Possible noise sources are: i) calibrations set-up vibrations; ii) random thermal electronic noise; iii) electromagnetic interferences; iv) slight differences in the positioning of the catheter over the measurements.

For each sample the precision interval was calculated considering a normal distribution of the measurements about the sample mean value. The precision interval i was obtained by multiplying the sample variance σ by a coverage factor t . For a Student's distribution with 4 degrees of freedom the coverage factor is $t = 2.770$. The interval $i = \pm \frac{t\sigma}{\sqrt{N}}$ represents the precision interval, given at a probability 95%, within which

Fig. 8: Calibration curve obtained giving in input to the measurement system a displacement. Confidence interval = 95% [10]

one should expect any measurement value to fall. In fig. 8, the result of the calibration procedure has been reported. The measured sensitivity of the device is $S = 33.62 \text{ count}/\mu\text{m}$.

Keeping constant the input to the measurement system at

Fig. 9: Calibration set-up: 1. PC for data acquisition and analysis; 2. three-way valve; 3. catheter; 4. syringe for pressure generation; 5. manometer; 6. sensor.

$0 \mu m$, the resolution (Res) of the device was evaluated as the square of the standard deviation of the output divided by the sensitivity: $Res = \frac{\sigma^2}{S} = \frac{0.98^2}{33.62} = 0.028 \mu m$.

Finally, the whole system was calibrated pressurizing the two catheters with known pressures ranging from 0 to 100 $mmHg$, in order to relate radial displacements of the tubes and deflections of the beam. For each known pressure, the measurement was repeated five times.

A schematic of the calibration set-up is shown in Fig. 9. A three-way valve connects the sensor to a syringe (pressure source) and a manometer (Mercurial Sphygmomanometer, ICO Medical), used as reference standard for calibration. The catheter displacement Δl is amplified by the developed mechanism. Then, the amplified displacement ($e(\beta)$) is read by the Hall effect sensor. A microcontroller ARDUINO UNO (not shown) communicates with a PC trough RS232 protocol.

In Fig. 10, calibration curves are shown for the two catheters. For catheter 1 the sensitivity and resolution respectively are: $S(r_{ecat1}) = 0.047 \text{ count}/mmHg$, $Res(r_{ecat1}) = \frac{\sigma^2}{S} = \frac{0.98^2}{0.047} = 20.77 mmHg$.

For the second catheter (catheter 2) sensitivity and resolution respectively are: $S(r_{ecat2}) = 0.14 \text{ count}/mmHg$ and $Res(r_{ecat2}) = \frac{\sigma^2}{S} = \frac{0.98^2}{0.14} = 7 mmHg$.

V. DISCUSSION

In many diagnostic or therapeutic procedures the pressure of infusion fluids has to be monitored for safety reasons. The most common measurement approach consists in the insertion of a pressure transducer along the fluidic line by means of a 3-way valve. In this configuration the transducer gets in contact with the infusion fluid. Because of the resulting contamination of the transducer, the device should be sterilizable or disposable. The aim of this work is the development of a dry pressure sensor, i.e. not in contact with the infusion fluid. The working

Fig. 10: Calibration curves for the catheter 1 (up) and catheter 2 (down). Confidence interval = 95% [10].

principle of this sensor is based on the measurement of the radial expansion of the external surface of a catheter, as in [7].

This radial displacement is amplified by an *ad hoc* designed mechanical system consisting in a close-to-singularity beam. The sensor has been conceived having in mind as target application scenario the treatment of the intervertebral disc degeneration by means of regenerative strategies, involving the infusion of ATMPs.

The infusion has to be continuously monitored not to exceed physiological intradiscal pressure values. As a design constraint, we have chosen a measurement range compatible with the intradiscal pressure of a healthy individual lying prone (75 $mmHg$ relative pressure [9]).

Two catheters, namely catheter 1 and catheter 2, commonly used in such surgical procedure, have been selected. Due to their mechanical and geometrical characteristics the maximum radial displacements corresponding to the application of the maximum pressure ($p = p_{max}$) respectively are $2.37 \mu m$ and $8.68 \mu m$.

Encouraging results have been obtained in terms of measurement range and resolution. Our sensor measures pressures in the range of 0 – 75 $mmHg$ with a resolution of 20 $mmHg$ for catheter 1 and 7 $mmHg$ for catheter 2.

The high radial stiffness ($k_c = 14000 N/m$) of catheter

Fig. 11: Sensitivity (S) and resolution (Res) of the sensor vary with the catheter external diameter (r_e), under the hypothesis of keeping the Young's modulus ($E = 6 MPa$) and the catheter wall thickness ($r_e - r_i = 0.3 mm$) constant.

1 results in a larger value of resolution than catheter 2. However, the confidence interval around the target pressure $p_{max} = 75 mmHg$ is equal to $i = \pm 16 mmHg$. For this reason pressure values in the range $[59-91] mmHg$ cannot be distinguished. In the worst case it would happen that the sensor measures a pressure equal to p_{max} , while actually the intervertebral disc pressure has risen up to 91 mmHg. However, the pressure range that the intervertebral disc can tolerate without damages is well over 91 mmHg [9]. Therefore, catheter 1 can be also considered suitable for the proposed scenario. Although developed for a particular application, the sensor can be easily adapted for other clinical contexts. In fact, resolution and sensitivity both depend from the geometric and mechanical characteristics of the catheter, as quantified by Eq. 3.

As shown in Fig. 11, keeping constant the material of the catheter (E , ν) and its thickness ($r_e - r_i$), sensibility (S) and resolution (Res) strongly depend from the external radius (r_e).

By way of example, another possible application field could be pain treatment through epidural blockade during labour. In this context, Lechner et al. [6] developed a device to identify the epidural space by continuously monitoring the pressure of saline solution flushed by the epidural needle during its insertion. Pressure changes are indicative of the anatomic compartments being passed through to reach the epidural space. After proper identification of the epidural space, a dose of local anesthetic is administered. Typical pressure values measured during the insertion range from 60 mmHg to 350 mmHg [6].

Considering only the range in which our sensor has been calibrated for known displacements, if a PVC catheter with $r_e = 0.7 mm$ and $r_i = 0.4 mm$ is chosen, the sensor would

be suitable for such application (Fig. 11, Application 1). In this case, a sensitivity $S = 0.04 count/mmHg$ and a resolution $Res = 26 mmHg$ can be obtained in the range 0 – 442 mmHg.

Another possible application is the measurement of intra-abdominal pressure. This parameter is used in clinical practice as a guide to intra-peritoneal pathologies diagnosis and as a predictor of renal function. It is commonly measured using gastric balloons [2] or urinary catheters [3] through the use of three-way valve systems with pressure sensors in contact with the measuring fluid.

In this case, the measured pressures are between 0 – 30 mmHg. A PVC catheter with $r_e = 1.8 mm$ and $r_i = 1.5 mm$ can be used in conjunction with the sensor proposed in this work, having a resolution $Res = 0.77 mmHg$ and a sensitivity $S = 3.67 count/mmHg$ in the range 0–36 mmHg (Fig. 11, Application 2).

Depending on the chosen clinical scenario, values in different pressure ranges have to be measured and a catheter with the proper dimensions and material has to be selected. Therefore any change of the application scenario implies a re-calibration procedure. Once the catheter has been selected and the calibration procedure has been performed, it is not required to repeat the calibration sequence for every use.

VI. CONCLUSION

The proposed system represents a proof-of-concept of a novel transduction means for the development of non-invasive and non-disposable sensors, for the cost-effective continuous monitoring of infusion pressure. A design optimization is required for a future mass production of the sensor. Currently the Hall effect sensor range (i.e. 1024 counts) is partially utilized. Therefore, the device can be improved selecting a proper magnet dimension and magnet-sensor distance. This implies to change the fabrication technology in order to guarantee the assigned geometrical tolerances of the realized parts. However, it worth noting that the variability on the magnetic properties and geometrical specifications of the magnets can affect the repeatability of the sensor performances. In order to overcome this issue, a calibration process is required both for each assembled device and, as mentioned in Sec.V, for each application scenario. Finally, future work will be also devoted to the integration of a microcontroller allowing the final user to select among several already calibrated catheters.

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