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Flexible Membrane Micro Flow-rate Threshold Flow Sensor

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Abstract: A piezoresistive flow sensor that functions as a threshold flow switch, independent of input supply voltage, is designed, fabricated, and tested. The test flow rate is in the 10s ml/hr range. The sensor is fabricated using doped hydrogenated amorphous silicon thin film with MEMs on flexible polyimide substrate. The flow sensor is highly sensitive and able to distinguish between distilled water and 0.1 % NaCl saline solution in terms of threshold switch-on flow rate. The switch-on threshold of the flow sensor is observed to be independent of the applied input voltage. These turn-on threshold levels are adaptable to user's application. *Copyright © 2007 IFSA.*

Keywords: Flexible substrate, Micro flow rate sensor

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1. Introduction

Excessive brain fluid or Cerebrospinal fluid (CSF) accumulation in the brain and its surrounding because of hydrocephalus is detrimental [1]. Shunt pumps have been designed [2] and put into clinical practice to alleviate this life threatening anomaly [3]. Slow abnormal accumulation of fluids in the ventricles or subarachnoid space of the brain [4,5] needs to be removed via the CSF shunt system at an equilibrium rate to maintain an osmotic liquid pressure balance in the cranial. However, the conventional shunt systems are not able to monitor, control and effectively prevent over and under siphoning [2]. Nevertheless, flexible thinned silicon chip pressure sensors to remotely monitor blood flow to alert patients of anginapectoris and myrocardial infarction, and other fluid pressure has been constructed by Shin and coworker [6]. A flexible 3 mm x 3mm silicon substrate with a p+ diaphragm on 50nm/300nm Cr/Au electrode on Pyrex glass substrate acting as the variable capacitor for the inductance-capacitance (LC) circuit has been developed for biomedical applications, such as intracranial pressure monitoring, that are useful for hydrocephalus patients [7].

This article presents a new piezoresistive [8] based flow sensor design for slow continuous flow rate monitoring with a predetermined turn-on flow rate threshold. The flow sensor discussed is constructed with strain sensitive n+ a:Si-H piezoresistive elements deposited using a PECVD system as previously reported [8]. Since SiNx encapsulation and polyimide base are biocompatible that will resist adhesion of protein constituents and non-degradable, it can be included and implanted inside the body [9]. A precise switching threshold flow rate of a designed flow sensor is important to ensure that the shunt is activated to draw away the excessive CSF and prevent any CSF accumulation. The precise threshold switching flow sensor is preferred over a continuous flow meter to reduce battery power consumption (battery's life), shunt's pump wear and tear (pump's life), and the surgical costs of the shunt's pump replacement.

2. Device Design and Realization

The flow sensor's piezoresistors layout is as shown in Fig. 1. The two inner sensing elements (R2 and R4) are designed to measure the maximum amount of flexing by the flow of the incompressible media, while the outer sensing elements (R1 and R3) are mostly unaffected by the lateral compression and acts as the Wheatstone bridge's background temperature adjusters/compensators. The 2D and 3D outer and inner piezoresistors deformation details are modeled as illustrated in Fig. 2 and Fig. 3 respectively for completeness. The flow pressure of the fluid is applied from under the substrate.

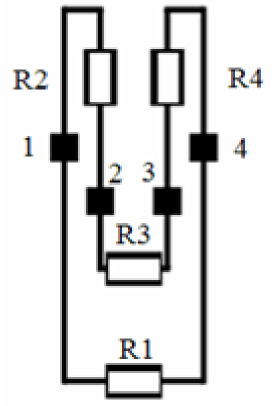

Fig. 1. Flow sensor design layout where label R1-R4 are the piezoresistors and label 1-4 are the connection pads.

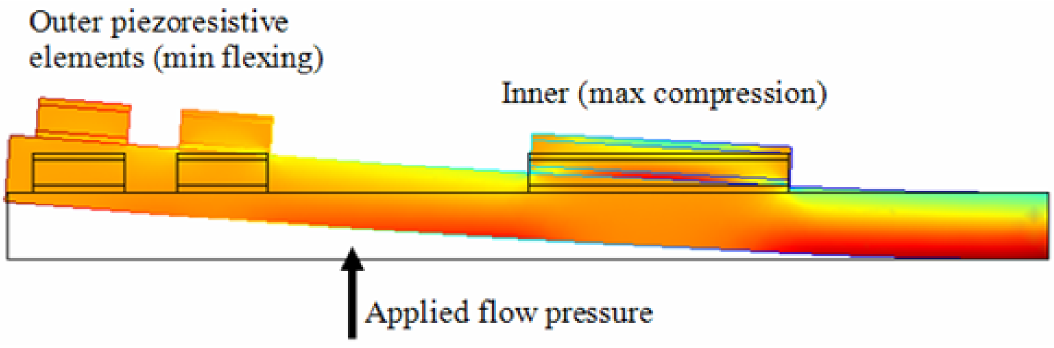

Fig. 2. 2D FEM modeled outer and inner piezoresistors deformation of the flow sensor. Piezoresistors R1 and R3 experience minimum bending and are located on outer cantilever edge while R2 and R4 situated in the inner location see a maximum compressive deformation.

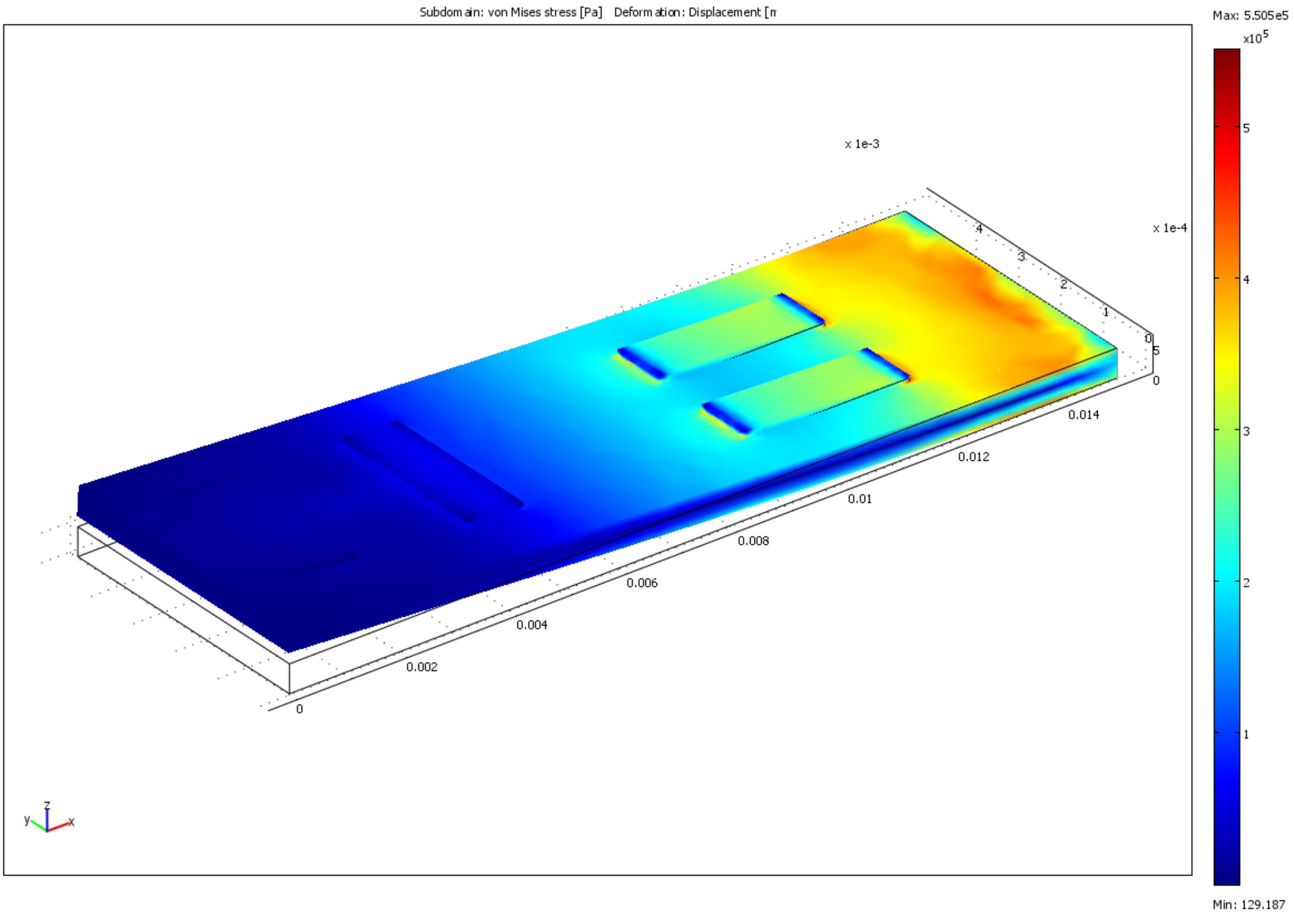

Fig. 3. 3D flow sensor simulations result showing the outer (R1 and R3) and inner (R2 and R4) piezoresistive sensing elements deformation with applied flow pressure from the bottom. The color indicator bar on the right illustrates the amount strain experienced.

The sensor processing flow starts with a 350nm SiNx adhesion deposition, followed by the 270nm n+ doping of a:Si-H via a Plasma Enhanced Chemical Vapor Deposition (PECVD) system on 51 micron clean polyimide substrate, Kapton E. The conduction traces are 10nm Cr/ 240nm Au sputtered using E-beam. The flow sensor is next patterned by standard photolithography method, RIE and chemical wet etched, 200 nm SiNx passivated, and annealed for 1 hr under forming gasses [8]. The flow sensor structural layout is illustrated in Fig. 4.

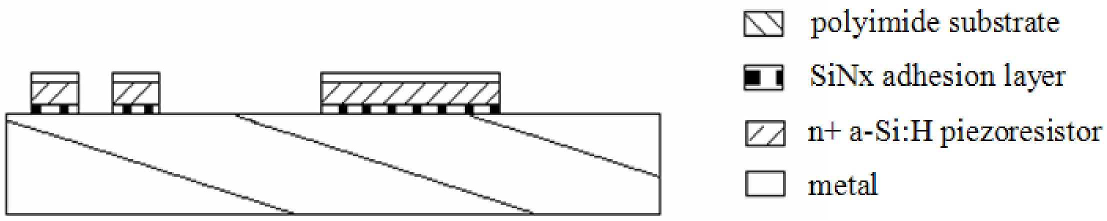

Fig. 4. Flow sensor schematic.

Each piezoresistive element is measured 0.6 mm W x 1.0 mm L, with a cross sectional area of 0.6 mm^2 and resistivity equal to 0.0033 Ωm. Gold wires (0.0007” diameter) are used to connect the sensor into the full Wheatstone bridge according to the electronic schematic given in Fig. 5. The sensor output is next connected to a operational amplifier circuit to amplifier the sensor's output before being measured with a Fluke 8520A digital multi-meter.

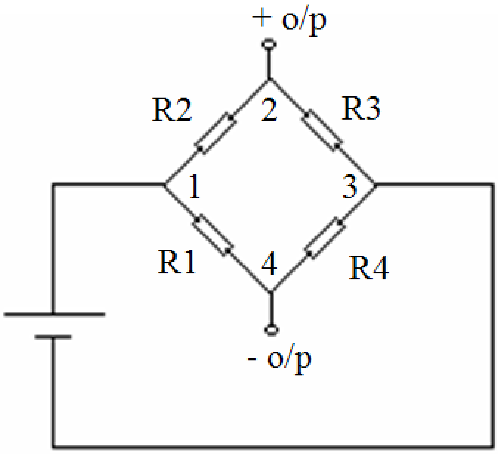

Fig. 5. Schematic diagram of sensor connected in a full Wheatstone bridge configuration. Points 1-4 are as defined in Fig. 1.

Since the flow sensor's inner piezoresistors are compressively altered by the bending as a result of the applied flow pressure from beneath, the actual resistance (R_s) changes is therefore negative or decreasing for sensing element length compression. This can be inferred from Equation (1),

$$R_s = \rho L/A, \tag{1}$$

where, R_s is the sensor piezoresistance, ρ is the resistivity, L is the length of the sensing element, and A is the total cross sectional area of the sensing element. As the sensor bends, the strained length L is reduced corresponding to a decrease in sensor resistance.

The flow sensor's output voltage for a Wheatstone bridge, is

$$\frac{V_{sensor}}{V_{in}} = \frac{A}{2}\left[\frac{R_s - R}{R_s + R}\right], \tag{2}$$

where, V_{sensor}/V_{in} is the ratio between sensor output (measured) voltages and the Wheatstone bridge input power. R_s is the sensing arm resistance (inner piezoresistors) and R is the reference arm resistance (outer piezoresistor) and A is defined as the gain of the operational amplifier. The sensor's output is consequently measured to be directly proportional to these decreasing resistance changes.

The experimental setup schematic is shown in Fig. 6. The sensor's Wheatstone bridge is powered by a HP E3630A constant current power supply. The flow rate is regulated with a mini-pump variable flow meter by Control Company. Fig. 7 is the flow sensor mounted on a stainless steel support adapter that is next attached to the circular drip pan illustrated by Fig. 8.

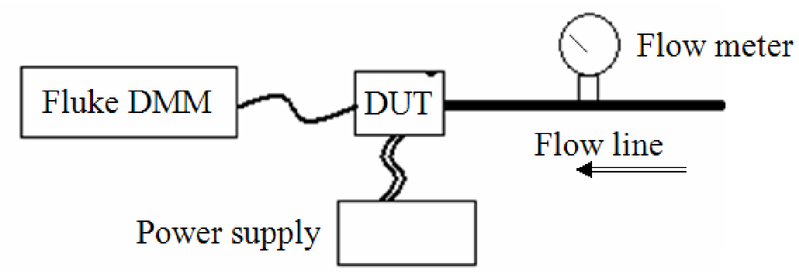

Fig. 6. Experimental setup schematic.

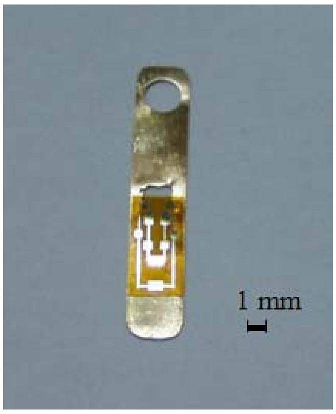

Fig. 7. Flow sensor mounted on a support adapter.

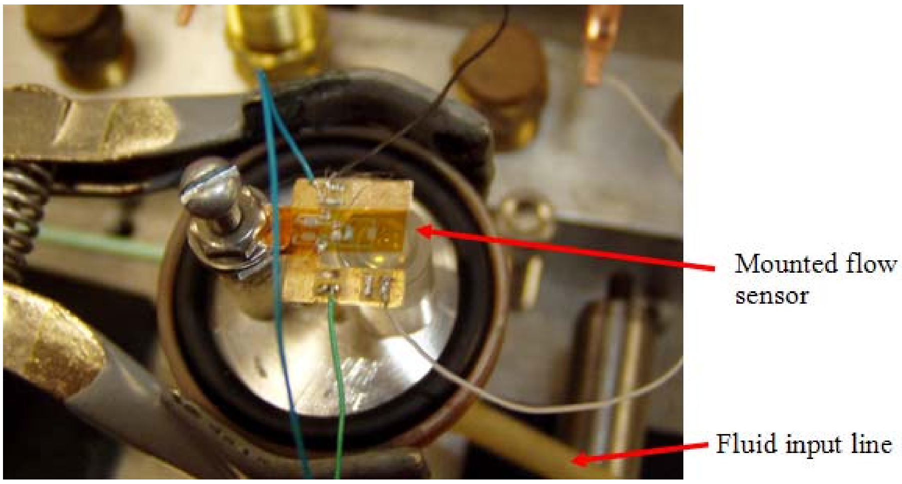

Fig. 8. Flow sensor experimental setup with sensor connected in a Wheatstone bridge configuration.

3. Results

The graph plotted in Fig. 9 is the typical measured flow sensor reading using a slow-moving incompressible fluid and a Wheatstone bridge input voltage of 6.0 Vdc. The sample fluid tested is twice distilled water introduced over different flow rates. The total length of the protruding cantilever measured from the mounting screw is approximately 17.69 mm. The flow sensor tests show a flat dc response below a threshold flow rate of approximately 4 ml/hr, followed by a roughly linear gradient with respect to the increasing flow rates. The empirical flow sensitivity observed is 0.004.1 V/ml/hr at standard laboratory conditions (1 atm, 25°C).

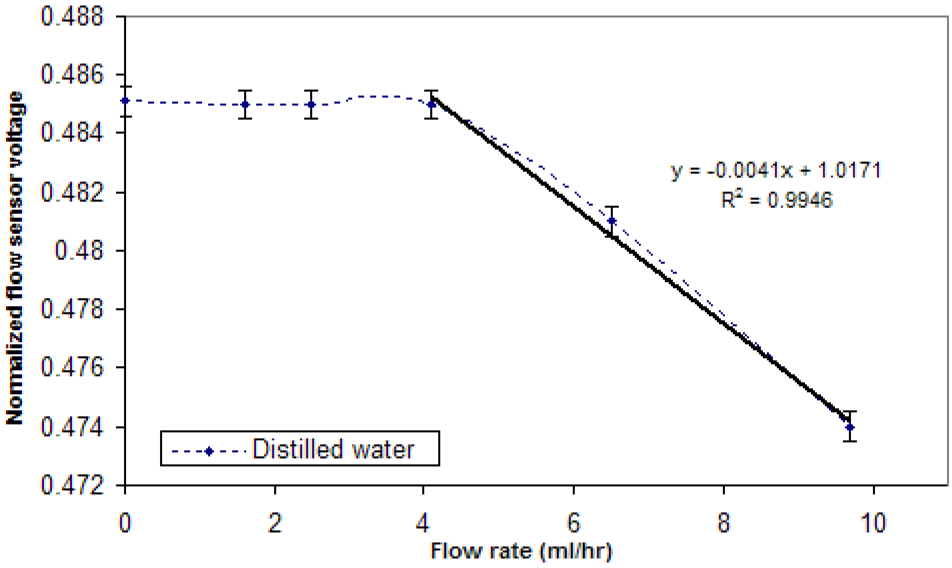

Fig. 9. Flow sensor response to continuous slow distilled water flow at 6.0 Vdc.

The input voltage dependence of the flow sensor is also investigated by applying different input voltages to the device. It is determined that the switch-on flow threshold is independent of the supplied voltage. The fluid used is low concentration saline water (1000 – 3000 ppm) that mimics the body's CFS. The saline solution is prepared with a NaCl tablet (BioChemika) at pH 7. The protruding cantilever length is 15.82 mm. From the measured data, in Fig. 10, it is found that the initial flat dc responses are proportional to the applied voltages. The switch flow threshold is found to be constant, at approximately 16 ml/hr, for all the different applied voltages. This constant switching threshold is crucial for any implanted shunt pump system because the designed flow sensor would still activate the device at the correct preset flow rate level even as the battery operated system is weaken and aged. The nominal CSF secretion measured for a young child to a healthy adult ranges from 16 - 20 mL/hr [5] by which the designed flow sensor switch-on flow rate threshold could be successfully employed in hydrocephalus application.

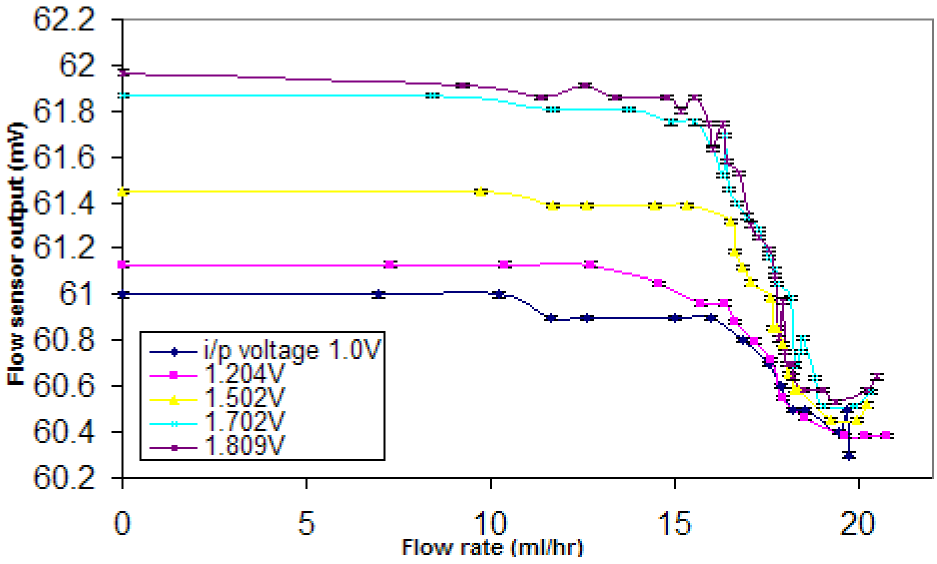

Fig. 10. Flow sensor responses as a function of input voltage for slow-moving continuous flow.

The reason for the different turn-on flow rate threshold between the twice distilled water and 0.1% saline solution experiment is due to the physical length of the protruding cantilever. Hence, a custom switch-on threshold flow rate can be adjusted physically with ease by flow sensor's mounting. This tunable cantilever flow sensor switching flow rate threshold design is applicable to any precision switching applications that have weight restrictions.

4. Conclusion

A simple piezoresistive flow sensor design is realized using Si processing technology and is capable of measuring continuous slow-moving flow rates. The observed sensor flow threshold/sensitivity is independent of input voltage and surrounding temperature. A linear relationship between the sensor output response voltages and slow moving flow rate of 0.0041 V/ml/hr is measured for distilled water as deduced theoretically. A 16 ml/hr switch-on threshold for slightly saline (brackish) fluid is independent of applied voltage. The designed flow sensor with its custom adjustable turn-on threshold range is needed for hydrocephalus devices. Different saline solutions and viscosity dependent in-vivo experiments are needed to effectively quantify the sensor for possible in-situ application.

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Guide for Contributors

Aims and Scope

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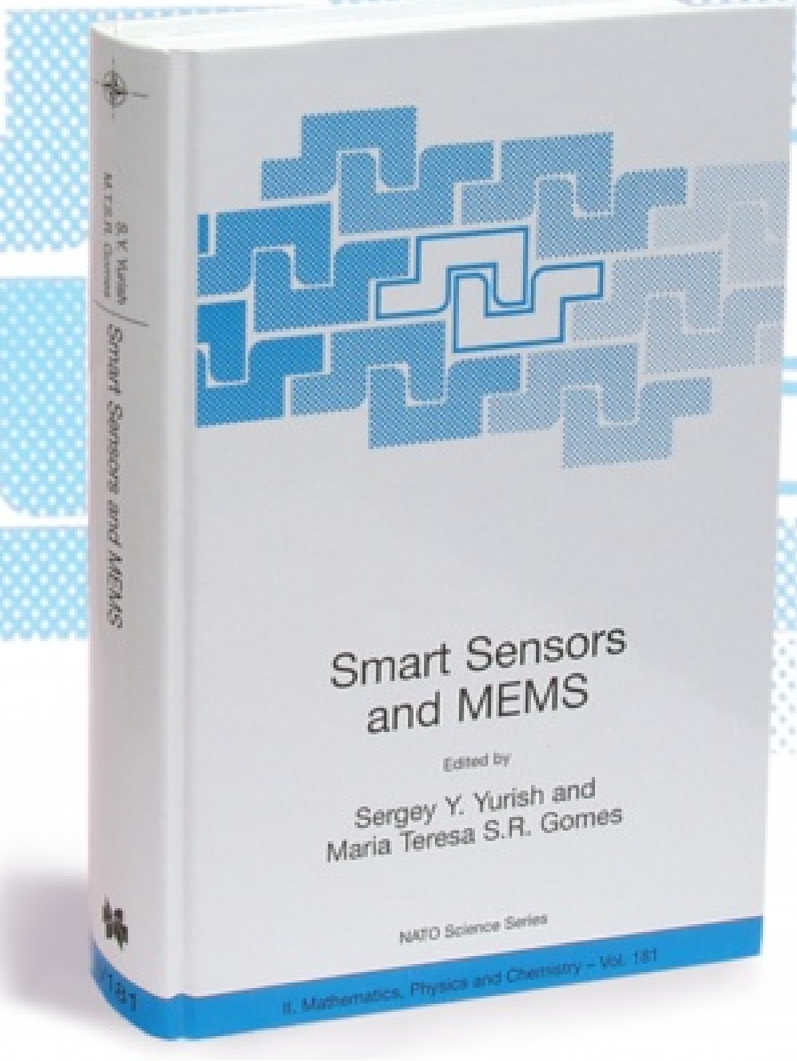

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