Open access • Journal Article • DOI:10.1016/0924-4247(93)80019-D

Capacitive sensors: When and how to use them☆ — Source link

Robert Puers

Institutions: Katholieke Universiteit Leuven

Published on: 01 Jun 1993 - Sensors and Actuators A-physical (Elsevier)

Topics: Capacitive displacement sensor, Capacitive sensing and Piezoresistive effect

Sensors and Actuators A, *37–38* (1993) 93–105

Capacitive sensors: when and how to use them*

Robert Puers
Katholieke Universiteit Leuven, Departement Elektrotechniek, E S A T – M I C A S , Kardinaal Mercierlaan 94, B-3001 Heverlee (Belgium)

Abstract

Capacitive sensors for the detection of mechanical quantities all rely on a displacement measurement The movement of a suspended electrode with respect to a fixed electrode establishes a changing capacitor value between the electrodes This effect can be measured and if the mechanical quantity controls the movable electrode, a sensor is realized Since the value of the capacitor is directly related to its size, and a small capacitor means high noise susceptibility, capacitive sensors should be as large as possible Capacitive pressure sensors have been developed with success for industrial applications, where large membrane sizes are not a critical issue However, most centres of expertise in silicon sensors show an interest in exploiting silicon technology to produce capacitive pressure sensors as well From the above, this miniaturization trend appears to be an unsound idea On the other hand, the principle of capacitive sensors allows the realization of measuring systems with so far unknown performance Indeed, the capacitive sensor reveals distinct advantages when compared to its piezoresistive counterpart high sensitivity, low power consumption, better temperature performance, less sensitive to drift, etc Nevertheless, only a minor fraction of the market for pressure sensors is taken up by capacitive-type sensors When observing the characteristics of capacitive sensors, it may seem surprising to encounter so few devices in real-world applications The reasons for the lack in breakthrough can be found in the design complexity and the requirements for a matched sensing circuit This paper will extensively discuss the justification of the choice for this research effort, and will elaborate on the techniques to fabricate the devices based on electronic manufacturing procedures Basically, silicon capacitive sensors differ from piezosensors in that they measure the displacement of the membrane, and not its stress! This has important implications on the final assembled device less package-induced problems can thus be expected However, a far more important issue is their extremely high sensitivity, together with a low power consumption These issues make them especially attractive in biomedical implant devices, or in other telemetry applications, where power is not randomly available So far, this is the only field of success for these sensors However, due to the interesting detection principle, new fields of application emerge, offering unique and superior performance when compared to available sensors Uniaxial accelerometers are a good example, where extremely high cross-sensitivity reduction can be obtained

Introduction

Capacitive-type sensors will generate an electrical signal as a result of the elastic deformation of a membrane, as is the case for other sensors, such as the piezoresistive types However, it is not the built-up stress in the membrane that causes the signal, but rather its displacement This phenomenon is the essential difference from other sensors and results in unique properties, which will be illustrated later on The basic structure of a capacitive sensor always consists in a set of plates of surface A separated by a distance d (Fig 1) One obtains for the capacitance value at rest

$$C_0 = \varepsilon\left(\frac{A}{d}\right) \tag{1}$$

where ε is the dielectric constant (permittivity) of the medium between the plates (usually air) One can see

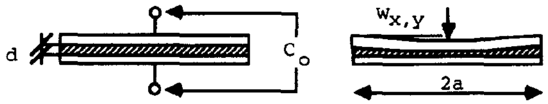

Fig 1 Schematic representation of a capacitor, consisting of two plates and a dielectric with thickness d To the right, the non-parallel movement of the plates is represented

that to change the capacitance value, one can either change the surface of or the distance between the plates Most capacitive sensors rely upon the change of the distance d rather than the change in surface A, although the latter phenomenon is intrinsically linear The reason has to be sought in a more elaborate structure and hence a more difficult assembly [1] If one is able to deform the dielectric layer over its entire surface with the distance Δd, such that the plates remain parallel, one has

$$C = \varepsilon\left(\frac{A}{d + \Delta d}\right) \tag{2}$$

*Invited paper

0924-4247/93/$6 00

From eqns (1) and (2), and for $\Delta d \ll d$, one obtains

$$\frac{C}{C_0} \approx 1 - \frac{\Delta d}{d} \tag{3}$$

which reveals a linear relationship

From eqn (1) one can also obtain the expression for the sensitivity of this sensor

$$\frac{\Delta C}{\Delta d} = -\varepsilon \frac{A}{d^2} \tag{4}$$

which illustrates that the sensitivity will increase with the square of the distance d between the electrodes For a small gap a high sensitivity can thus be expected

However, most sensors cannot be constructed such that expression (2) is valid, that is, the plates often do not move in parallel In most cases the capacitance change will be caused by a deformation of one of the electrodes itself (see Fig 1, right) Expressions (2) and (3) no longer hold Equation (2) should be written as

$$C = \int_{x=0}^{a} \int_{y=0}^{a} \frac{\varepsilon}{(d-w)} \, dx \, dy \tag{5}$$

where w, a function of x and y, represents the local deflection as a consequence of the pressure One can define v as the mean value of the displacement of the entire electrode

$$v = \frac{1}{A} \int_{x=0}^{a} \int_{y=0}^{a} w \, dx \, dy \tag{6}$$

so eqn (5) becomes

$$\Delta C = C_0\left(\frac{v}{d-v}\right) = C_0\left(\frac{v}{d}\right) \tag{7}$$

The sensitivity to pressure can be obtained from this expression There is a non-linear relationship If, however, the displacement is small compared with the membrane thickness, v equals about one quarter of the maximal deflection at the centre of the membrane [2], allowing an estimation of the performance of the sensor to be made

Justification of capacitive sensors

In order to understand the underlying reasons for the growing research in this field, one has to consider the limitations of the so-called well-established techniques, such as the piezoresistive devices One of the fields that has always been looking ahead for more and tighter specifications for sensor systems is without doubt medical engineering Many pressures can and must be monitored in the human body in order to allow proper diagnosis and therapy Since some of these measurements must be done inside the patient, they require extremely stable, temperature insensitive and accurate devices Solid-state strain-gauge devices have already been on the market for a long time [3] and have been widely used for arterial pressure measurement for more than 20 years [4] Whereas these systems are very useful and accurate for measuring dynamic pressure changes, they show deficiencies when it comes to long-term measurements of low-pressure signals This is because these sensors drift about 100 Pa per day If the pressure signal is in this range, they become unsuitable A possible solution is the use of fluid-filled catheters, which are coupled to precise manometer systems outside the patient However, this approach is generally accepted to be cumbersome, may cause inaccuracies in dynamic signals and restrains the patient It is unsuitable for application in telemetry devices Moreover, for such applications, the power consumption of the sensor and its signal-processing network become important, since telemetry system have to rely on a battery The limited amount of power makes the use of piezoresistive devices difficult, especially for long-term continuous monitoring This is especially the case for implantable systems, where the size of the battery must be kept to an absolute minimum, or where even no battery can be tolerated [5] These considerations gave rise to the development of capacitive-type pressure transducers to achieve an improved static accuracy [6, 7] and to provide low-power operation of the entire system

Construction methods

In contrast to piezoresistive or any piezosensitive transducers, for the capacitive type of sensor the stress in the membranes is of no direct relevance to the transduction phenomenon From a construction viewpoint, this offers distinct advantages, for example, a much larger tolerance can be accepted with respect to the deposited structures on the membrane In fact, the membrane itself acts as a transduction medium, and not the deposited structures (e g, resistors have to be placed very accurately to obtain maximal sensitivity, and are very susceptible to misalignment)

Complete discrete construction

Given its intrinsic simplicity, the capacitive sensor construction is not that obvious First there are the specific requirements with respect to the tolerances in the construction, especially concerning the electrode separation Moreover, the choice and homogeneity of the dielectric layer are not evident Discretely built sensors have been realized in commercial applications, for applications requiring a high sensitivity or for small-signal differential pressure measurements (e g, in flow-

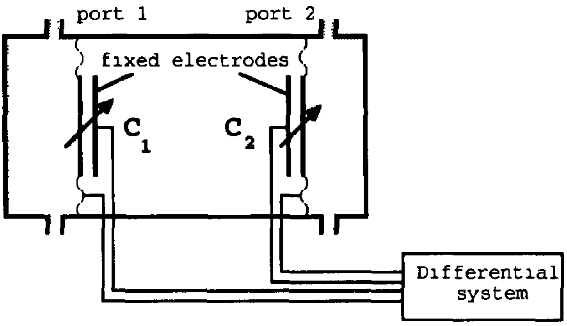

Fig 2 Example of a commercial differential capacitive pressure-measuring device The diameter of the membranes is about 10 cm, and the output is delivered under the 4–20 mA standard

measuring devices) Figure 2 illustrates a cross section of such a device [8] Since parasitic effects should be reduced, these sensors are generally large constructions, because the capacitive value (at rest) is directly proportional to the surface, and the parasitic capacitance remains approximately constant For industrial control systems the large size is usually acceptable

Figure 3 illustrates two capacitive sensors, built for medical implant devices The one on the left is a sensor built to cope with the drift problems of strain-gauge sensors It was intended to monitor intracranial pressure [9], and is a good example of what the state of the art was some 15 years ago The size of the sensor is 7 mm by 2 mm, with a spacing of 25–75 μm between the diaphragm and its base The conductive metal surface, which forms the fixed electrode, is fired on a glass layer The zero-pressure capacitance was 8 pF, and varied up to 16 pF for a 100 kPa signal On the right-hand side of the Figure is a more recent version of the same construction, introduced for use in cardiology Here, the prime concern was high sensitivity coupled to small size and low power consumption, since the device is intended for long-term implantation [10] A titanium barrel of 2 5 mm diameter is filled with a glass slug, fused to its inner wall A layer of gold is sputtered on the concave surface of the glass, and the movable counter electrode is welded consecutively One obtains a capacitance value of a mere 1 5 pF A signal-modulating circuit is added to boost the output of the sensor so that it can be transmitted over the long catheter wires The entire system is encapsulated into a hermetically sealed titanium package for biocompatibility purposes It is developed by a commercial pacemaker company (for automatic defibrillator triggering)

Though the fabrication sequence is extremely difficult, these kinds of sensor constructions still seem to be very attractive to designers, with simplicity as the key issue to justify their use As another example, the capacitive principle has also recently been proposed for the detection of angular position [11] Here, printed circuit board technology is used to form overlapping patterns on two different boards, which change their common surface with changing angular position Hence, the total capacitor surface A is changed, and thus C varies The conductive surface (electrodes of the capacitor) is shaped in order to improve the linear relationship with the angle

Thick-film technology

Since most mechanical sensors are in a rather high cost-price category, the relatively low fabrication cost of thick-film circuits seems an attractive alternative for the realization of sensors This has been extensively the case for piezoresistive sensors, where for large-market applications, such as the automotive sector, successful designs have been elaborated [12] Capacitive devices can be realized in thick-film technology as well, although virtually no research has been performed in this direction Standard thick-film technology allows capacitors to be realized using a multilayered structure Special pastes with a high dielectric constant are available to obtain a reasonable capacitor value with a small geometry However, the compliance of this dielectric, mostly consisting of a glass matrix, is virtually zero, rendering the structures impractical for sensor use Other materials should be adopted Silicone rubber is one of them This has been exploited in a device [13] by using flexible polyimide film carriers Figure 4 shows a thick-film variant of this sensor, a force transducer It will give a better linear performance when fabricated in thick-film

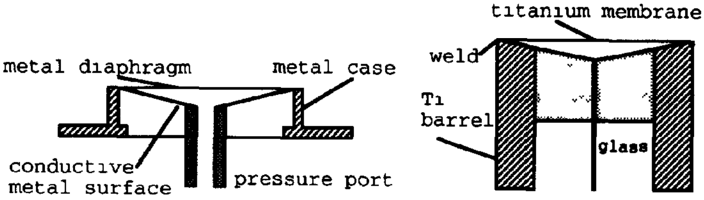

Fig 3 Examples of discretely built capacitive pressure sensors left, an early intracranial pressure sensor (7 mm diameter, membrane thickness = 12 μm), right, a more recent version (2 5 mm diameter) for applications in cardiology

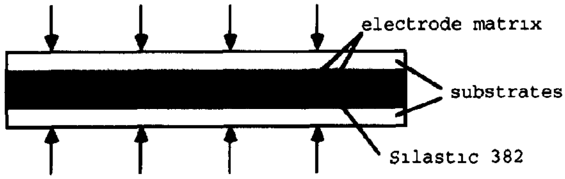

Fig 4 Capacitive thick-film force transducer based on a compliant dielectric

technology, since the ceramic substrates give more rigidity to the individual electrodes Another possibility is to make use of the mechanical properties of the ceramic substrate itself, and measure the deflection of a suspended membrane by the capacitive principle This comes close to the device of Fig 2, but there the ceramic electrode itself does not flex The only apparent drawback, i e , the low initial capacitive value, typically between 1 and 20 pF for this structure, can well be overcome by incorporating some impedance converting network on the same thick-film substrate The field is still open for some good (and inexpensive) research!

Silicon technology

The use of silicon as a construction material has long been demonstrated and has been exploited by many centres [14] Some 20 years ago, the first major efforts were undertaken to fabricate miniaturized sensors based on existing processing sequences used in electronic circuit fabrication, and expanding these to suit the specific requirements for the realization of novel devices These new techniques or processing sequences are often referred to as micromachining This involves all kinds of three-dimensional structuring of silicon to fabricate mechanical sensors The first efforts were primarily aimed at the feasibility of the realization of sensor devices as such, focusing primarily on the performance of the transducing mechanism of these sensors, and looking into possibilities to realize specific mechanical structures [15–19]

However, sensors always have to be considered as part of an overall system, where they act primarily at the input stage of a usually complex process Therefore, and also because the (electrical) signal out of the transducer is usually extremely weak, the sensors must be essentially coupled to some electronic networks to enhance and process their signals Since the sensor devices are fabricated in silicon, the idea of integrating electronic networks on the same die is quite natural However, from the fabrication technology viewpoint, this approach is far from natural Sensor fabrication requires dedicated processing sequences that conflict with circuit processing, impeding the merging of both devices in the first attempt However, by careful processing sequence design, one can achieve positive results Often, the circuit processing will have to be performed at the beginning, then the wafers are post-processed in the micromachining foundry

With respect to the more usual piezoresistive devices, the specific merits of capacitive devices have to be found in their simplicity, i e , they require less extensive processing (only an electrode is needed) on the membrane, and are less susceptible to misalignment errors [20] Once a process has been set up, a high degree of reproducibility can be expected This apparent 'simplicity', however, should be considered carefully The interelectrode distance is a very important parameter in the design and fabrication, and any lack of control will be reflected in a loss of reproducibility Figure 5 shows a common layout of such a capacitive sensor Misalignment in the x or y direction will have minor influence on the performance of the device However, tight control of the distance z is mandatory for a reproducible device z in fact reflects the distance between the two electrodes This can be affected both by the cavity depth and by all the intermediate layers between the glass and the silicon wafer Special care must be taken with the feedthrough of the electrode connections to the bonding pads (or the processing network) It is best to make use of a diffused layer in order to minimize step coverage failure On the other hand, an aluminium path 100 nm thick can still be bonded hermetically The Figure also reveals two possible interconnection schemes The one at the right is best with respect to a parasitic capacitance, since the two bonding pads are on the highly isolating glass layer The interconnection

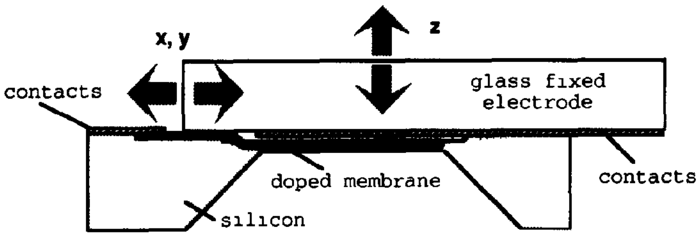

Fig 5 Usual construction for a bulk micromachined silicon capacitive pressure sensor Contacts can be made on the glass substrate or on the silicon The arrows define possible misalignments in the assembly

on the left is preferable when electronic processing circuitry is added

No matter what method is selected, care must be taken in dicing the bonded wafers one of the wafers, usually the glass one, must be cut precisely in order not to touch the underlying structure Also, this underlying structure should, evidently, not be bonded to the glass, which requires special precautions in the design stage

This means that, although intrinsically much simpler then the piezoresistive device, the biggest challenge of the capacitive device lies in its packaging This also forms the origin of most of its problems

One of the specifications directly influenced by the packaging technique is the temperature sensitivity of the sensor depending on the sealing technique one can realize absolute, relative or trapped-gas pressure sensors The first is sealed under vacuum, the second is sealed such that the reference chamber has a connection with the outside world or any other medium (to realize a differential measuring set-up) In the case of entrapped gas, extra non-linearity is added for two reasons first, the gas itself will be compressed for any movement of the membrane This effect can be diminished by providing extra volume in the sensor reference cavity, e g , at the edge of the sensing membrane The effect of thermal gas expansion, on the other hand, is much larger, and cannot be reduced by any means It can reach values of up to 1% of full scale per °C The only solution is to use vacuum-sealed pressure sensors

Overview of developed devices

An attempt is made here to give an overview of capacitive devices currently under development or of devices which have been demonstrated so far Table 1 gives the most important details of about 40 different or related capacitive-type sensors, all fabricated in silicon technology Although it would be intriguing to comment on all of these papers, time and space do not allow us to do so It is therefore left to the reader to pick out relevant devices to assess his own state of the art In this text, only a fraction of the Table will be discussed This will be done to justify the arguments given here and to illustrate general research policies

Pressure-sensing devices

First, it is apparent that the pressure sensor is the most successful device (60%), at least at the early stages of research in this field This has to match the field of application most of the devices are developed with low power consumption in mind Indeed, a capacitive device consumes just as much power as its measuring circuit The immediate application lies in the biomedical field, where (especially for telemetry purposes) reduction of power consumption is the major design criterion The first device was developed for use in cardiology [21], and aimed at low power consumption In a second version, an oscillator circuit was added in bipolar technology to yield an overall current drain of 20 μA at a supply from 2 5 to 10 V Figure 6 illustrates a cross section of this pioneering structure

It has been steadily improved to incorporate more advanced electronics (for temperature compensation) and finally to include a dummy reference capacitor [22, 23] Ko and his co-workers introduced the idea of incorporating a dummy capacitor in the same cavity to reduce the effects of parasitic capacitance They also compared square with circular shaped electrodes and illustrated that the highest sensitivity can be obtained for a circular electrode (of about 36% of the central area of the diaphragm [24, 25] Pressure sensors for biomedical applications were the very first technology drivers to simulate research in this direction [26–28] Numerous devices followed the first ones, with ever-decreasing sensor sizes Rest capacitive values of a mere 0 5 pF [29, 30] and, in a later development stage, even going down to 0 3 pF [31], have been reported by the Michigan group of Wise Evidently, such low intrinsic capacitive values cannot be handled without the matching circuitry close to them In the last two papers, the capacitor and the processing network were separate chips, avoiding compatibility problems when trying to merge the two devices on one die An interesting approach to further miniaturizing the size was proposed in ref 30, where first anodic bonding is performed, after which almost the full frame of the membrane die is etched away (see Fig 7) This is an excellent example of how standard techniques can still lead to novel and successful designs The circuit chip can be processed by any foundry The only drawback of this approach is the protection of the circuit chip when applied *in vivo*

The group of Esashi in Japan has proposed a series of interesting combinations of sensors with a matched measuring circuit in one package [32–34] All of these devices are based on the approach of first bonding the wafers and then doing a consecutive backside etch to yield a high degree of miniaturization Furthermore, a special technique is adopted to connect the device all electronics are encapsulated, and only the backside of the capacitive membrane is exposed to the outside world To contact the electronics, a glass feedthrough is realized by ultrasonic drilling, sputtering coating of the hole and finally connecting the lead wire by conductive epoxy (see Fig 8) Although the sealing is more advanced in this device, more caution is required for the realization of the CMOS/micromachined device chip

In the search to improve sensor specifications many more solutions have been proposed Special concern

TABLE 1 Details of capacitive-type sensors fabricated in silicon technology

Authors	Year	Type	Dimensions (mm^2)	C_0 (pF)	Etch	Seal	Circuit	Application	Discussion	Ref
Sander, Knutti and Meindl	1980	pressure	3 × 3	22	bulk	anodic	Schmitt oscillator, bipolar	cardio	early device	21
Ko, Bao and Hong	1982	pressure	2 × 4	?	bulk, hydrazine	anodic	integrated, bipolar	general	ring versus square membrane, drift aspects	24
Lee and Wise	1982	pressure	4 × 4	12	bulk, KOH	anodic	separate	general	effect sealing on TCO and TCS	26
Ko, Shao, Fung, Shen and Yeh	1983	pressure	2 × 4	6	bulk, hydrazine	anodic	integrated CMOS	biomedical	ref capacitor integrated with sensing capacitor	25
Smith, Prisbe, Shott and Meindl	1984	pressure	2 × 6	22	bulk	anodic	integrated oscillator, bipolar 10 μm	cardio	temperature compensation	22
Hanneborg and Ohlkers	1985 1990	pressure	4 × 4	?	bulk	sputtered glass	separate chip	general	low TCO and drift	36, 37
Smith, Bowman and Meindl	1986	pressure	2 × 6	22	bulk	anodic	integrated oscillator, bipolar 10 μm	cardio	comparison with piezoresistive devices	23
Chau and Wise	1987	pressure	1 × 5	0.5	EDP, bulk	anodic	separate	cardio	miniaturized	30
Miyoshi, Akiyama, Shintaku, Inami and Hijkigawa	1987	pressure	?	8	surface etch	n.a.	no	general	Ni diaphragm, sacrificial layer, cheap large batch prod	49
Shoji, Nisase, Esashi and Matsuo	1987	pressure	2 × 3	10	bulk, EDP	Si-Si fusion	CMOS	biomedical	direct fusion bonding	27
Furuta, Esashi, Shoji and Matsumoto	1989	pressure	3.5 × 0.7	3.5	bulk, KOH	anodic	separate, CMOS	cardio	small assembly, complete backside etch	33
Puers, Peeters and Sansen	1989	pressure	2 × 3.5	10	bulk, KOH	anodic	no	general	FEM analysis & linearization	39
Backlund, Rosengren, Hok and Svedbergh	1990	pressure	3 × 3	25	KOH bulk	fusion bonding	*LC* circuit only	eye-pressure	*LC* tuned by pressure-transponder system	69
Kandler, Eichholz, Manoli and Mokwa	1990	pressure	array 81 × 100 μm∅	2	sacrificial layer, polysilicon	n.a.	SC CMOS	general	preliminary results	47

Matsumoto, Shoji and Esashi	1990	pressure	2 × 1 7		bulk, KOH	anodic	integrated, CMOS	cardio	throughhole connection, backside etch	34
Puers, Peeters, Vanden Bossche and Sansen	1990	pressure	2 × 3 5	10	bulk, KOH	anodic	separate, CMOS	biomedical	parasitic capacitance rejection	40
Puers, Vanden Bossche, Peeters and Sansen	1990	pressure	1 8 × 2 2	10	bulk, KOH	anodic	separate	cardio	miniaturized biocompatible package	45
Artyomov, Kudryashov, Shelenshkevich and Shulga	1991	pressure	4 5 × 4 5	7	bulk	?	no	general	three electrode, linearization by reducing electrode	42
Ji, Cho, Zhang, Najafi and Wise	1991	pressure	1 4 × 0 4	0 3	bulk	anodic	separate	cardio	subminiature assembly	31
Kudoh, Shoji and Esashi	1991	pressure	2 3 × 3 7		bulk, KOH	anodic	integrated CMOS oscillator	general	advanced assembly, electrical feedthroughs	35
Kung and Lee	1991	pressure	0 4 × 0 5		surface etch, KOH through	n a	integrated NMOS	general	polydiaphragm, miniature size	46
Nagata, Terabe, Fukaya, Sakurai, Tabata, Sugiyama and Esashi	1991	pressure	5 × 5		bulk, TMAH	anodic	integrated CMOS oscillator	general	frequency output, linearized response	52, 53
Rosengren, Soderkvist and Smith	1992	pressure	2 × 2	3	bulk, KOH	fusion bond	no	test	linearization techniques by constructions	42
Schnatz, Schoneberg, Brockherde, Kopystynski, Mehlhorn, Obermeier and Benzel	1992	pressure	8 4 × 6 2	6	bulk	anodic	integrated, CMOS SC	general	includes band-gap reference for temp correction	47
Suminto, Yeh, Spear and Ko	1987	pressure, acceleration	2 × 3	8	bulk, EDP	anodic	separate, CMOS	general	membrane with central mass for linearity	28
Puers and Vergote	1992	movement	1 × 1	4	KOH, bulk	anodic	separate	animal monitoring	composite suspension membrane	63
Petersen, Shatel and Raley	1982	accelerometer	0 3 × 0 1	0 004	surface etch, EDP	n a	one CMOS stage	general	silicon oxide beam, with Au deposited mass	53
Rudolf	1983	accelerometer	1 5 × 2 6	1	KOH bulk	anodic	separate	general	plate suspended on torsion bars	54

TABLE 1 (continued)

Authors	Year	Type	Dimensions (mm^2)	C_0 (pF)	Etch	Seal	Circuit	Application	Discussion	Ref
Rudolf, Jornod, Bergqvist, Leuthold and Bencze	1987, 1990	accelerometer	8 × 6	20	KOH bulk	anodic	separate	spacecraft	force balancing μg resolution	55, 56
Olney	1988	accelerometer	2 8 × 3 6	10	bulk	anodic	separate	space, flight control	commercial device	64
Schlaak, Arndt, Steckenborn, Gevatter, Kiesewetter and Grethen	1990	accelerometer	7 × 7	16	bulk, EDP	anodic	separate	position sensing	μg resolution, symmetric suspension	58
Seidel, Riedel, Kolbeck, Muck, Kupke and Koniger	1990	accelerometer	35 × 3 5	10	bulk	anodic	separate	general	good linearity	60
Suzuki, Tuchitani, Kato, Ueno, Yokota, Sato and Esashi	1990	accelerometer	3 × 4 5	9	bulk	anodic	separate	general	servo action	57
Kloeck, Suzuki, Tuchitani, Miki, Matsumoto, Sato, Koide and Sugisawa	1991	accelerometer	3 × 4 5	9	bulk	anodic	separate	general	transparent electrodes on glass allow observation	59
Payne and Dinswood, and Goodenough	1991	accelerometer	±5 × 5		surface etch	n a	integrated CMOS	automotive	fully integrated device, no bulk micromachining	65 66
Peeters, Vergote, Puers and Sansen	1991	accelerometer	3 6 × 3 6	12	bulk, KOH	fusion bond +anodic	separate	general	highly symmetrical, tunable damping	61
Ura and Esashi	1991	accelerometer	5 × 6	15 ×2	bulk, hydrazine	anodic	separate	general	silicon-oxinitride suspension	62

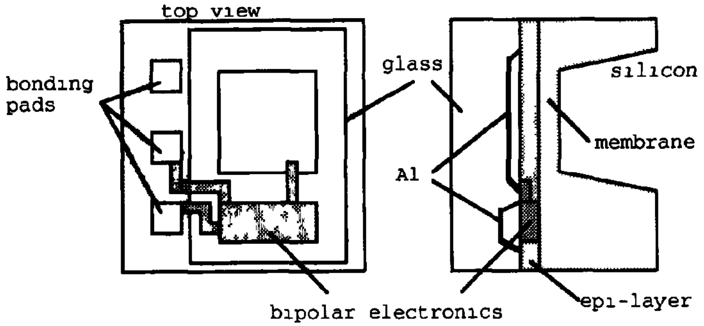

Fig 6 The Stanford capacitive pressure sensor, measuring 3 mm × 3 mm and incorporating a bipolar converting circuit, as proposed in 1980

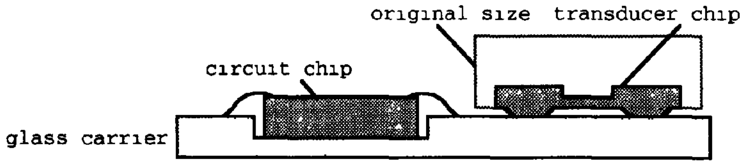

Fig 7 In order to obtain minimal size, the sensor chip is firstly anodically bonded to the glass carrier, and then etched down using EDP, which stops on the boron-doped diffused structure, forming the miniature frame and membrane (dotted line shows the original chip size) The processing chip itself is embedded in a cavity in the glass carrier

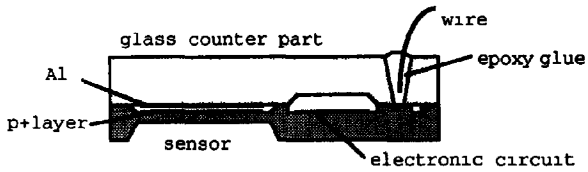

Fig 8 Fully sealed device incorporating an electronic circuit for detection of the capacitive changes A glass feedthrough provides the connection to the outside

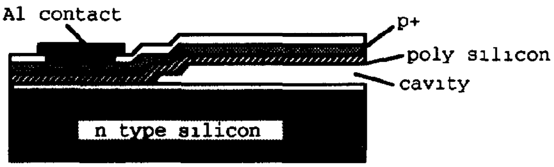

Fig 9 Principle of a surface micromachined pressure sensor by etching the oxide from underneath the polysilicon, a cavity can be created The doped p^+ region is used as the movable electrode for the sensor

has been given by the Norwegian group [35–37] to lower stress effects in the membrane (which cause temperature effects in the sensitivity curve) This can be achieved by retaining the idea of bonding the silicon chip to a glass carrier, but using thin sputtered glass layers on silicon substrates instead In conjunction, many efforts have been spent in better understanding capacitive transducing phenomena, and more precisely many proposals have been made in order to reduce parasitic effects, either by special circuitry or by specific electrode configurations, and-to look for a better linear response of the device [38–43]

Also, more recently, more and more efforts are being made to merge CMOS processes with capacitive pressure sensors [44–48] The main drive behind this approach is to minimize parasitic effects, and to look for acceptable process flows for the silicon foundries This has partially become possible by the use of sacrificial layer technology, first introduced by Guckel and Burns [49] The method consists in etching away the silicon dioxide from between the upper LPCVD polysilicon and the n-type silicon, as shown in Fig 9 A cavity is realized, which is equal to the original oxide thickness, e g , 1 μm The formation of stress-free polysilicon is a possible difficulty, but the advantage is a higher yield per wafer (since no space-consuming backside etching is required), and an automatic over-pressure stop for pressure sensors [50] Also, the double-sided alignment that is required in bulk micromachining can be avoided Moreover, this process is fully compatible with the standard CMOS process

Other approaches were to use an alternative to alkaline-based etchants, such as TMAH, to perform the bulk etch of the silicon, thus avoiding possible contamination of the CMOS process by KOH [51, 52]

Accelerometers

Accelerometers form the second type of devices for which the capacitive measuring principle has proved successful Although the first devices, developed only ten years ago, were rather primitive [53, 54], the first one showed a degree of miniaturization never obtained in any of the later devices It consisted of a cantilevered beam, made out of silicon oxide, the end of which was provided with a deposited dot of gold, which acts as a seismic mass The capacitance value was a mere 4 fF, such that a buffer stage had to be implemented on the same die, in immediate proximity of the cantilever The second device was nothing more than a suspended plate that served as the seismic mass The device was soon developed into a more sensitive accelerometer, using a mesa structure to detect the acceleration [55, 56] The capacitive principle also allowed it to be used reciprocally, i e, by applying an electrostatic voltage, the electrode can be repositioned with respect to the original driving force By doing so, the voltage required to restore the equilibrium position becomes a measure of the acceleration This principle is better known under the term 'force balancing' and became popular in accelerometer development because it allows the range of the device to be expanded by several orders of magnitude [57–59]

Other designs do not focus on force-balanced systems, but concentrate on the geometry of the structure as a means to improve the dynamic behaviour One possibility is to look for highly symmetrical structures The immediate merit of this approach is to obtain extremely high cross-sensitivity reduction [60, 61] Both the symmetry of the structure and the capacitive detection principle guarantee a specification of this parameter which is some orders of magnitude better than that of the best commercial devices in other technologies Figure 10 gives a three-dimensional impression of this accelerometer

In some designs, the spring elements of the accelerometer are no longer silicon and other layers can be used This is especially the case for designs where miniaturization becomes important the size of the seismic

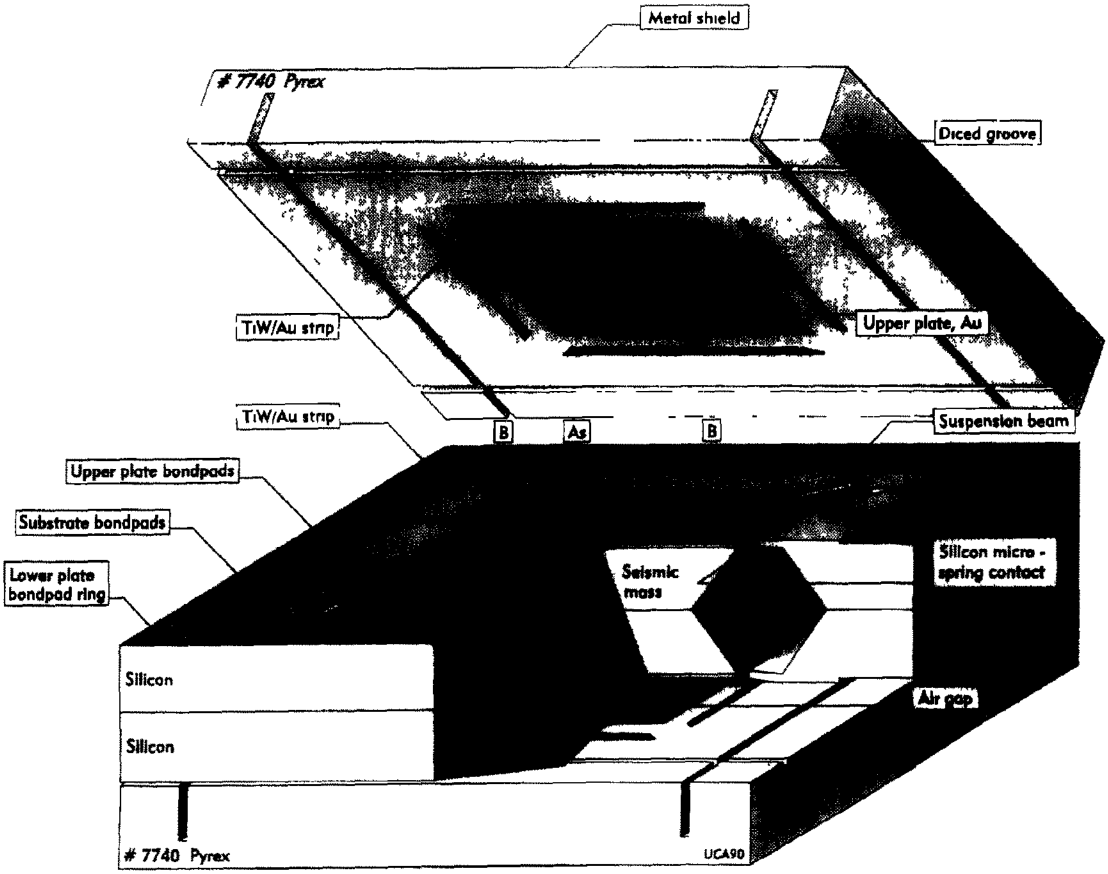

Fig 10 Three-dimensional view of a fully symmetric accelerometer, realized as a four-layer structure, including both silicon fusion and anodic bonding [61]

mass becomes too small to be susceptible to the desired acceleration Thinner suspension systems can be a solution In two cases, silicon nitride membrane is used [62, 63] The first uses a set of 64 silicon nitride beams to suspend the seismic mass, whereas the second device uses a composite membrane of nitride and oxinitride in order to cancel stress effects This device can therefore be made as small as 1 mm × 1 mm × 1 mm

It is intriguing to encounter fully commercial capacitive acceleration sensors, a fact which does not appear with pressure sensors The reason has to be sought in the fact that the capacitive principle allows for a much better performance than any other detection principle This is particularly the case in space and military applications, where there is a strong need for precise accelerometers The Endevco device is a good example [64] of such a commercial product It contains a well-calculated damping system, and it is shown that the device is much more sensitive, stable and has a higher resonance frequency than its piezoresistive counterpart

Another commercial product is the Analog Device acceleration sensor, developed for the automotive sector (airbag system) [65, 66] It is a very interesting device, built around an interdigitized structure, realized by reactive ion etching of a free-standing polysilicon layer (surface micromachined) The axis of sensitivity is thus in the plane of the sensor itself What is surprising is to see this company, with so far no known experience in micromachining, capable of putting a high state of the art device into the market (surface micromachined, chip surface = 10% sensor, 90% signal processing, includes force balancing, sensitive in the chip surface), obviously dwarfing all the research efforts of so many other centres

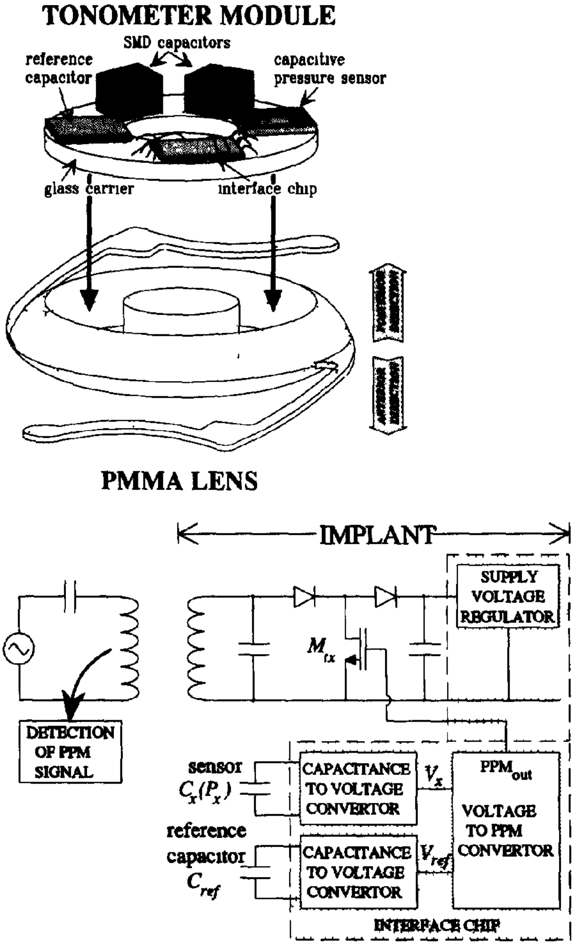

Fig 11 Schematic drawing of the posterior chamber lens, equipped with the telemetric tonometer

Example of biomedical device development

A good example for the justified application of capacitive sensing principles is the biomedical field, especially when information has to be transmitted from the inside of the (human) body to the outside world For such telemetry systems, low power consumption, high sensitivity, low temperature drift and a good stability of the sensors are a prerequisite Capacitive sensors can cope with these demands As an example, the project of eye monitoring is given In ophthalmology, the eye pressure is an important parameter However, it is a very slowly varying signal and is moreover very small in comparison with the atmospheric pressure which on average lies 2000 Pa above it, with a daily cyclic variation in the range of 500 Pa In order to be able to monitor this signal on a continuous basis, implantable eye-pressure monitoring devices have been proposed They all rely on capacitive sensors [5, 67, 68] Whereas the first two rely on a purely passive system (one *C*, one *L* which act as a resonant circuit), the last approach is a fully active device, which picks up its power from a tuned coil system It was developed in our laboratory by Van Schuylenbergh and Peeters, and is illustrated in Fig 11 [5] A commercially available PMMA lens is machined down to leave only the central optic part A cylindrical space of 3 5 mm ID and 8 mm OD by 0 5 mm is available Total power delivery by the tuned coil system is limited to 100 μW The Figure illustrates the circular carrier in glass, which contains the receiving coil on its bottom part, and the interconnection diagram for the electronic circuits and the (fixed) sensor electrode at the top side A dedicated switch capacitor circuit is used to make a differential measurement between the reference capacitor and the pressure sensor The interface chip, shown in Fig 12, is an adapted version of a former development [40], and runs at a mere 35 μA

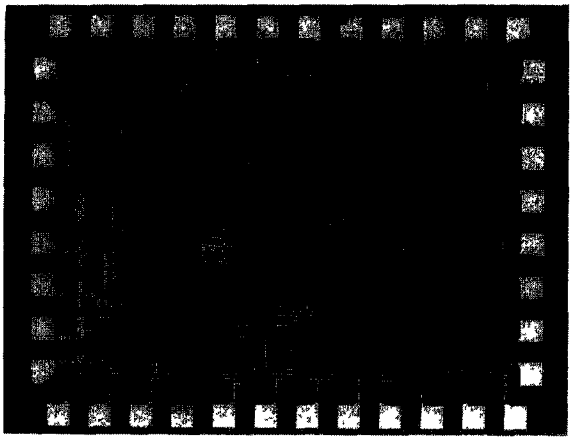

Fig 12 Microphotograph of the custom-designed CMOS switched capacitor circuit It measures 1 9 mm × 2 4 mm and accepts unregulated supply voltages between 4 5 and 15 V

References

1 U Zelbstein, Capteurs de pression de fluides, in G Asch (ed), *Les Capteurs en Instrumentation Industrielle*, Bordas, Paris, 1987, pp 595–599

2 S Timoshenko and S Woinowsky-Krieger, *Theory of Plates and Shells*, McGraw Hill, New York, 1959

3 E Konigsberg, Capabilities of implatable transducers, in L Harmison (ed), *Res Animals in Med*, DHEW (NIH) 72-333, 1973, pp 1145–1150

4 H Sandler, T Fryer and B Datnow, Single channel pressure telemetry unit, *J Appl Physiol*, *26* (1969) 235–238

5 K Van Schuylenbergh, E Peeters, B Puers, W Sansen and A Neetens, An implantable telemetric tonometer for direct intraocular pressure measurements, *Abstr 1st Eur Conf Biomed Eng*, *Nice, France, 1991*, pp 194–195

6 J Atkinson, D Shurtleff and E Foltz, Radio telemetry for the measurement of intracranial pressure, *J Neurosurg*, *27* (1967) 428–432

7 T Fryer, Telemetry of intracranial pressure, *Biotelemetry Patient Monitoring*, *5*(2) (1978) 88–122

8 Philips Industrial Automation, *Data sheet of PSC-Transmitter PD3 for Differential Pressure and Flow Measurements*, 1987

9 T Fryer, A pressure telemetry system utilizing a capacitance-type transducer, *Biotelemetry III*, Academic Press, New York, 1976, pp 279–282

10 A Moore and K Anderson, An implantable chronic pressure sensor, *Proc 4th Int Conf Solid-State Sensors and Actuators (Transducers '85), Philadelphia, PA, USA, June 7–11, 1985*, pp 189–192

11 R Wolffenbuttel and R Van Kampen, An integrable capacitive angular displacement sensor with improved linearity, *Sensors and Actuators A*, *25–27* (1991) 835–843

12 B Puers, S Pasczcynski and W Sansen, Assessment of thick film fabrication methods for force (pressure) sensors, *Sensors and Actuators*, *12* (1986) 57–76

13 M Neuman, A Berec and E O'Connor, Capacitive sensors for measuring finger and thumb tip forces, *IEEE Frontiers Eng Comp in Health Care, 1984*, pp 436–439

14 K Petersen, Silicon as a mechanical material, *Proc IEEE*, *70* (1982) 420–457

15 K Bean, Anisotropic etching of silicon, *IEEE Trans Electron Devices*, *ED-25* (1978) 1185–1193

16 A Bohg, Ethylene diamine-pyrocatechol-water mixture shows etching anomaly in boron doped silicon, *J Electrochem Soc*, *118* (1971) 401

17 T Jackson, M Tischler and K Wise, An electrochemical P–N junction etchstop for the formation of silicon microstructures, *IEEE Electron Devices Lett*, *EDL-2* (1981) 44–45

18 G Wallis and D Pomerantz, Field assisted glass-metal sealing, *J Appl Phys*, *40* (1969) 3946–3950

19 A Brooks and R Donovan, Low temperature electrostatic silicon to silicon seals using sputtered borosilicate glass, *J Electrochem Soc*, *119* (1972) 545–546

20 S Clark and K Wise, Pressure sensitivity in anisotropically etched thin-diaphragm pressure sensors, *IEEE Trans Electron Devices*, *ED-26* (1979) 1887–1896

21 C Sander, J Knutti and J Meindl, A monolithic capacitive pressure sensor with pulse period output, *IEEE Trans Electron Devices*, *ED-17* (1980) 927–930

22 M Smith, M Prisbe, J Shott and J Meindl, Integrated circuits for a capacitive pressure sensor, *Proc IEEE Frontiers Engin Comp in Health Care, 1984*, pp 440–443

23 M Smith, L Bowman and J Meindl, Analysis, design and performance of capacitive pressure sensor IC, *IEEE Trans Biomed Eng*, *BME-33* (1986) 163–174

24 W Ko, M Bao and Y Hong, A high sensitivity integrated circuit capacitive pressure transducer, *IEEE Trans Electron Devices*, *ED-29* (1982) 48–56

25 W Ko, B Shao, C Fung, W Shen and G Yeh, Capacitive pressure transducers with integrated circuits, *Sensors and Actuators*, *4* (1983) 403–411

26 Y Lee and K Wise, A batch-fabricated silicon capacitive pressure transducer with low temperature sensitivity, *IEEE Trans Electron Devices*, *ED-29* (1982) 42–48

27 S Shoji, T Nisase, M Esashi and T Matsuo, Fabrication of an implantable capacitive type pressure sensor, *Proc 5th Int Conf Solid-State Sensors and Actuators (Transducers '87), Tokyo, Japan, June 2–5, 1987*, pp 305–308

28 J Summto, G Yeh, T Spear and W Ko, Silicon diaphragm capacitive sensor for pressure, flow, acceleration and altitude measurements, *Proc 4th Int Conf Solid-State Sensors and Actuators (Transducers '87), Tokyo, Japan, June 2–5, 1987*, pp 336–339

29 H Chau and K Wise, Scaling limits in batch fabricated silicon pressure sensors, *Proc 3rd Int Conf Solid-State Sensors and Actuators (Transducers '85), Philadelphia, PA, USA, June 7–11, 1985*, pp 174–177

30 H Chau and K Wise, An ultra-miniature solid-state pressure sensor for a cardiovascular catheter, *Proc 4th Int Conf Solid-State Sensors and Actuators (Transducers '87), Tokyo, Japan, June 2–5, 1987*, pp 344–347

31 J Ji, S Cho, Y Zhang, K Najafi and K Wise, An ultraminiature CMOS pressure sensor for a multiplexed cardiovascular catheter, *Proc 6th Int Conf Solid-State Sensors and Actuators (Transducers '91), San Francisco, CA, USA, June 24–28, 1991*, pp 1018–1020

32 K Furuta, M Esashi, S Shoji and Y Matsumoto, Catheter-tip capacitive pressure sensor, *Tech Digest, 8th Sensor Symp*, *Japan*, 1989, pp 25–28

33 Y Matsumoto, S Shoji and M Esashi, A miniature integrated capacitive pressure sensor, *Tech Digest, 9th Sensor Symp*, *Japan*, 1990, pp 43–46

34 T Kudoh, S Shoji and M Esashi, An integrated miniature capacitive pressure sensor, *Sensors and Actuators A*, *29* (1991) 185–193

35 A Hanneborg, T Hansen, P Ohlkers, E Calson, B Dahl and O Holwech, A new integrated capacitive pressure sensor with frequency modulated output, *Proc 3rd Int Conf Solid-State Sensors and Actuators (Transducers '85), Philadelphia, PA, USA, June 7-11, 1985*, pp 186-188

36 A Hanneborg and P Ohlkers, A capacitive silicon pressure sensor with low TCO and high long-term stability, *Sensors and Actuators, A21-A23* (1990) 151-154

37 A Hanneborg, M Nese, H Jacobsen and R Holm, Silicon to thin film anodic bonding, *MME'92 Workshop Digest, Leuven, Belgium, 1992*, pp 9-19

38 B Puers, E Peeters and W Sansen, CAD tools in mechanical sensors design, *Sensors and Actuators, 17* (1989) 423-429

39 B Puers, E Peeters, A Vanden Bossche and W Sansen, A capacitive pressure sensor with low impedance output and active suppression of parasitic effects, *Sensors and Actuators, A21-A23* (1990) 108-114

40 P Pons, G Blasquez and N Ratier, Harmonic response of silicon capacitive pressure sensor, *Sensors and Actuators A, 25-27* (1991) 301-305

41 V Artyomov, E Kudryashov, V Shelenshkevich and A Shulga, Silicon capacitive pressure transducer with increased modulation depth, *Sensors and Actuators A, 28* (1991) 223-230

42 L Rosengren, J Soderkvist and L Smith, Micromachined sensor structures with linear capacitive response, *Sensors and Actuators A, 31* (1992) 200-205

43 P Pons and G Blasquez, Transient response of capacitive pressure sensors, *Sensors and Actuators A, 32* (1992) 616-621

44 B Puers, A Vanden Bossche, E Peeters and W Sansen, An implantable pressure sensor for use in cardiology, *Sensors and Actuators, A21-A23* (1990) 944-947

45 J Kung and H Lee, An integrated air-gap capacitor process for sensor applications, *Proc 6th Int Conf Solid-State Sensors and Actuators (Transducers '91), San Francisco, CA, USA, June 24-28, 1991*, pp 1010-1013

46 M Kandler, J Eichholz, Y Manoli and W Mokwa, CMOS compatible capacitive pressure sensor with read-out electronics, in H Reichl (ed), *Microsystems*, Springer, New York, 1990, pp 574-580

47 F Schnatz, U Schoneberg, W Brockherde, P Kopystynski, T Mehlhorn, E Obermeier and H Benzel, Smart CMOS capacitive pressure transducer with on-chip calibration capability, *Sensors and Actuators A, 34* (1992) 77-83

48 S Miyoshi, H Akiyama, H Shintaku, Y Inami and M Hijikigawa, A new fabrication process for capacitive pressures sensors, *Proc 4th Int Conf Solid-State Sensors and Actuators (Transducers '87), Tokyo, Japan, June 2-5, 1987*, pp 309-311

49 H Guckel and D Burns, Planar processed polysilicon sealed cavities for pressure transducer arrays, *IEDM Tech Digest, 1984*, pp 223-225

50 H Guckel, D Burns, H Busta and J Detry, Laser recrystallized piezo-resistive micro-diaphragm sensor, *Proc 3rd Int Conf Solid-State Sensors and Actuators (Transducers '85), Philadelphia, PA, USA, June 7-11, 1985*, pp 182-185

51 H Terabe, Y Fukaya, S Sakurai, O Tabata, S Sugiyama and M Esashi, Capacitive pressure sensor for low pressure measurements with high overpressure tolerance *Tech Digest, 10th Sensor Symp, Japan, 1991*, pp 133-136

52 T Nagata, H Terabe, S Sakurai, O Tabata, S Sugiyama and M Esashi, Digital compensated capacitive pressure sensor using CMOS technology for low pressure measurements, *Proc 6th Int Conf Solid-State Sensors and Actuators (Transducers '91), San Francisco, CA, USA, June 24-28, 1991*, pp 308-311

53 K Petersen, A Shartel and N Raley, Micromechanical accelerometer integrated with MOS detection circuitry, *IEEE Trans Electron Devices, ED-29* (1982) 23-27

54 F Rudolf, A micromechanical capacitive accelerometer with a two-point inertial mass suspension, *Sensors and Actuators, 4* (1983) 191-198

55 F Rudolf, A Jornod and P Bencze, Silicon microaccelerometer, *Proc 4th Int Conf Solid-State Sensors and Actuators (Transducers '87), Tokyo, Japan, June 2-5, 1987*, pp 395-398

56 F Rudolf, A Jornod, J Bergqvist and H Leuthold, Precision accelerometers with μg resolution, *Sensors and Actuators, A21-A23* (1990) 297-302

57 S Suzuki, S Tuchitani, K Sato, S Uneo, Y Yokota, M Sato and M Esashi, Semiconductor capacitance-type accelerometer with PWM electrostatic servo technique, *Sensors and Actuators A21-A23* (1990) 316-319

58 H Schlaak, F Arndt, A Steckenborn, H Gevatter, L Kiesewetter and H Grethen, Micromechanical capacitive acceleration sensor with force compensation, in H Reichl (ed), *Microsystems*, Springer, New York, 1990, pp 617-622

59 B Kloeck, S Suzuki, S Tuchitani, M Miki, M Matsumoto, K Sato, A Koide and Y Sugisawa, Motion investigation of electrostatic servo-accelerometers by means of transparent ITO fixed electrodes, *Proc 6th Int Conf Solid-State Sensors and Actuators (Transducers '91), San Francisco, CA, USA, June 24-28, 1991*, pp 108-111

60 H Seidel, H Riedel, R Kolbeck, G Muck, W Kupke and M Koniger, Capacitive silicon accelerometer with highly symmetrical design, *Sensors and Actuators, A21-A23* (1990) 312-315

61 E Peeters, S Vergote, B Puers and W Sansen, A highly symmetrical capacitive micro-accelerometer with single degree of freedom response, *Proc 6th Int Conf Solid-State Sensors and Actuators (Transducers '91), San Francisco CA, USA, June 24-28, 1991*, pp 97-100

62 N Ura and M Esashi, Differential capacitive accelerometer, *Tech Dig 10th Sensor Symp, Japan*, 1991, pp 41-44

63 B Puers and S Vergote, A subminiature capacitive movement detector using a composite membrane suspension, *Sensors and Actuators A, 31* (1992) 90-96

64 D Olney, Acceleration measurement using variable capacitance, *Proc Sensors Nuremberg '88, Germany, 1988*, pp 149-160

65 R Payne and K Dinswood, Surface micromachined accelerometer a technology update, *SAE Int Automotive Eng Congr, Detroit, MI, USA, 1991*, pp 127-135

66 F Goodenough, Airbags boom when IC accelerometer sees 50g, *Electronic Design*, (8 Aug) (1991) 45-56

67 C Collins, Miniature passive pressure transensor for implanting in the eye, *IEEE Trans Biomed Eng, BME-14* (1967) 74-83

68 Y Backlund, L Rosengren, B Hok and B Svedbergh, Passive silicon transensor intended for biomedical, remote pressure monitoring, *Sensors and Actuators, A21-A23* (1990) 58-61