Reprint from proceedings MEMS2000, Miyazaki, Japan

ADVANCED HYBRID INTEGRATED LOW-POWER TELEMETRIC PRESSURE MONITORING SYSTEM FOR BIOMEDICAL APPLICATIONS

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ABSTRACT

A new, highly miniaturized hybrid integrated solution for telemetric pressure measurement in biomedical applications is presented. The telemetrically powered ITES (Implantable telemetric endo-system) consists of a surface micro-machined capacitive type absolute pressure sensor fabricated in an eight-mask MOS-like process and two low-power ASIC for capacitance change read-out and telemetric data and energy transmission. An advanced flip-chip mounting and assembly technology is applied to overcome most of the drawbacks of hybrid integration and to fulfill space requirements of biomedical implants without monolithic sensor integration. This paper emphasizes system design considerations of the hybrid system like its partitioning and the mounting and assembly technology. The pressure sensor design and its associated read-out is discussed in detail.

INTRODUCTION

In the last years there have been a lot of high invest investigations in monolithically integrated smart sensor systems [1-5], but only few devices have been a commercial success until now. Reasons for this may be: If the sensor is fully integrated into the IC process flow, the sensor performance is limited by the physical properties respectively by the order of the used layers. If the sensor is completely processed after the electronics finished, the costs increase because of additional masks and thereby reduced yield. This holds also for additive technologies like on-chip electroplating. The high invest for complete monolithical integration in the electronics process is only justifiable in high volume markets. The highest invests are focused in the automotive area at the moment (pressure, acceleration and yaw-rate sensors).

Another solution is the use of separate sensor chips packaged together in specific housings with an ASIC using standard processes available through manufacturers worldwide. Advantages for this are: The two processes can be optimized separately, respectively remain unchanged for the electronics part. The sensor can be fabricated on older and much more simple fabrication lines because there is no need for very low dimensions in the sub-µm range for most physical sensor principles. A modern low-power low-voltage CMOS/ BiCMOS process with mixed-signal libraries for cell based, highly automated design can be used for the electronics part. The drawbacks are the higher costs because of raised effort for packaging, testing and calibration. Additional area on both chips is needed for the chip-to-chip interconnect. Reliability is somewhat lowered compared to the monolithical solution. A matching of technological parameters between the separately processed chips is not given. To overcome these difficulties a careful system design is important.

System Concept

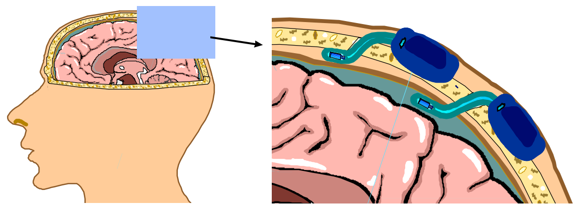

Fig. 1 Telemetric intracranial pressure measurement (ICP) application of ITES

The ITES system design shown in Fig. 1 and Fig. 2 [6,7] differs from all other known integration concepts because a separate surface micro-machined capacitive pressure sensor instead of a bulk micro-machined capacitive [8-11] or piezoresistive one is used.

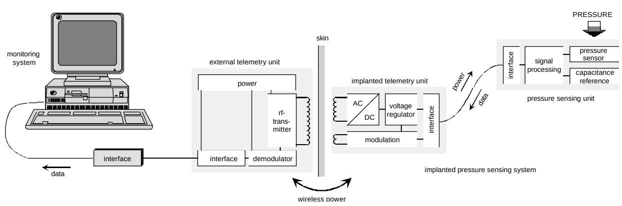

Fig. 2 ITES overview: implanted / external components

The capacitive pressure sensor comprises a pressure dependent capacitance $C_x = C_0 + \Delta C$ and a pressure independent capacitance C_0 for canceling out the offset capacitance. To build a sensor for measurement of absolute pressure the reference pressure inside the

sealed micro-cavity is kept well below 100 Pa = 1 mbar. This low pressure enclosed in the sealed cavity ensures the needed low temperature dependency of < 200 ppm/°C. Therefore no temperature sensor is needed for compensation in the small medical temperature range (approx. 35 °C – 42 °C). The read-out circuit is a micro-power (≤ 100 µA@3,5 V) differential SC-relaxation-oscillator with 10 bit resolution in a $\Delta C \approx 600$ fF measurement range giving a quasi-digital PWM output. A capacitance C_r defining the capacitive measurement range of the oscillator and an integrator feedback capacitance C_F are placed on the signal preprocessing ASIC. The PWM output of the signal preprocessing is transmitted via a flexible foil carrier based cable to the telemetry unit placed under the human skin. The whole system is powered by the transcutanous inductive telemetry link. The telemetry ASIC and the necessary micro-coil are hybrid integrated at this stage of development. The main disadvantage of further integration by using coil-on-chip technology is the large silicon area needed for the micro-coil. The data is transmitted by passive absorption modulation recently suggested by other researchers [12, 13].
The sensor and the telemetry ASIC are contacted to a flexible foil carrier by flip-chip technology. The carrier allows the necessary spatial splitting of the locations of the measurement point within the body and the transmission coil directly under the skin.

Surface micro-machined capacitive pressure sensor

As the systems key component a surface micro-machined capacitive absolute pressure sensor [0,8 mm x 2 mm x 0,5 mm] has been designed and fabricated. A slightly tensile stress optimized 1 µm thick polysilicon layer [14,15] is used as pressure sensitive membrane. The sensor is fabricated in a eight mask MOS-like process. The simple process and small area of the sensor ensure high yield. The sensor process flow is as follows:

1. Starting p^+-substrates (ρ = 10-30mΩcm)
2. Implantation of boron: front side substrate contacts
3. Implantation of phosphor: definition of ground electrode and feed through lines
4. Deposition and patterning of 225 nm low-stress LPCVD-Si_3N_4 layer as LOCOS mask and poly-Si membrane dielectric isolation
5. Growth of 850 nm thermal silicon dioxide (SiO_2) as sacrificial layer above the ground electrode
6. Deposition and patterning of 250 nm LTO (Low Temperature Oxide) as etch channels on top of thermal oxide
7. Deposition, doping with $POCL_3$, annealing and patterning of 1 µm poly-Si membrane
8. Sacrificial layer etching in HF 50%, rinsing with DI-water / ethanol and subsequent drying
9. Sealing of cavity by deposition of 1 µm LTO [17]
10. Opening of contacts to poly-Si and substrate
11. Sputtering and patterning of AlSiTi metallization
12. Deposition of PECVD nitride passivation layer
13. Opening of bond pads and membrane area
14. Spin coating of thick resist as dicing protection

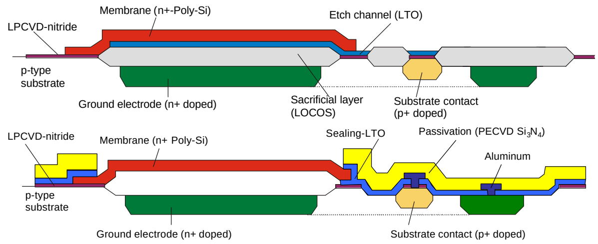

Fig. 3 Cross section of sensor device before sacrificial layer etching (top) and finished (down)

The result of that simple process flow is shown in Fig. 3, where the area of the sensor without electrode feed-through is depicted. Advantageous are the well controllable, simple etching and drying process without sticking problems and the highly reproducible one-layer membrane. The main disadvantages are the non-dielectric isolation of the lower electrode by a reverse biased pn-junction and the somewhat higher offset capacitance variation because of the low uniformity of the LTO-layer on top of the thermal oxide. The sensor device has to be coated with a protective coating against influences (e.g. humidity) from the surrounding. This is especially important in in-vivo applications.

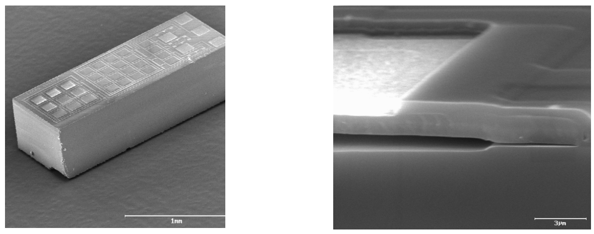

Fig. 4 SEM of capacitive pressure sensor (left), close up of view of LTO sealed etching-channel (right)

In Fig. 4 a complete sensor (0,8 mm x 0,5 mm x 2 mm) and cross section of a single sensor membrane is shown.

The layout of the pressure sensor is shown in Fig. 5. The sensor capacitance C_x is constructed by 16 square elements each 75 µm x 75 µm). An offset compensating capacitance array C_0, which is independent from pressure p, with $C_0 = C_x$ (p=0) is build up by leaving all layers on top of the polysilicon membrane layer, which stiffens the structure mechanically.

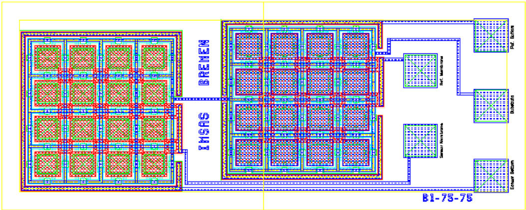

Fig. 5 Pressure sensor layout

Sensor modeling

An analytical HDL-A model of the sensor, verified by FEM simulations, has been set-up for co-simulation and interdependencies with the read-out electronics [17-19]. This model includes the influence of geometrical data, material properties gained by especially designed process monitoring structures [20].

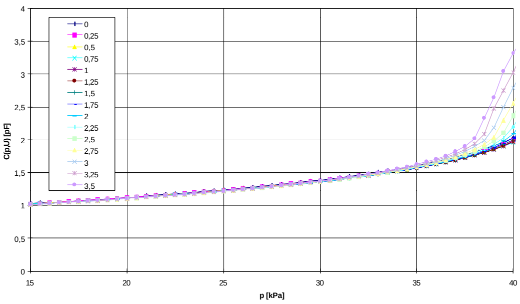

Fig. 6 Electrostatic influence on sensor characteristics vs. applied voltage of read-out circuitry

The pressure induced load deflection of the membrane and the resulting capacitance is calculated analytically. The electrostatic influence of the read-out electronics as shown in Fig. 6 is taken into account. The simulations showed, that the electrostatic force has to be taken into account when using the full deflection range of the sensor membrane.

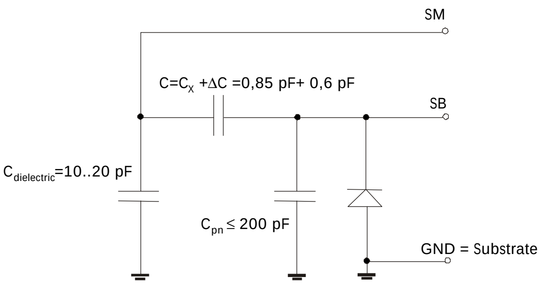

Fig. 7 Simplified electrical model of pressure sensor

The parasitic electrical properties of the three electrode configuration like the stray capacitances of the sensor element due to the reverse biased pn-junction isolation of the lower electrode respectively the dielectric capacitance of the membrane electrode to the substrate are taken into account in the electrical model (Fig. 7).

Read-out electronics

The read-out interface ASIC was designed and fabricated in a commercial 0,7 µm CMOS process with analog extensions (poly-Si-substrate capacitors, high-ohmic poly-Si resistors and low-V_T PMOS transistors).

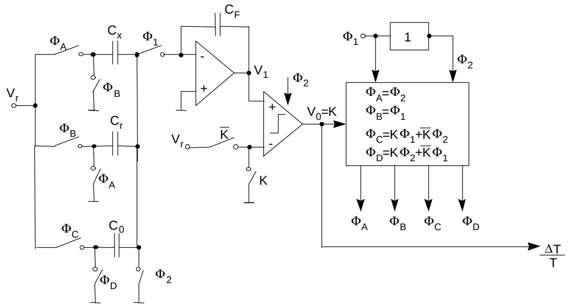

Fig. 8 Schematic of SC-relaxation oscillator

The interface shown in Fig. 8 is based on a simple low-power relaxation-oscillator circuit [21-24] realized in switched-capacitor technology.

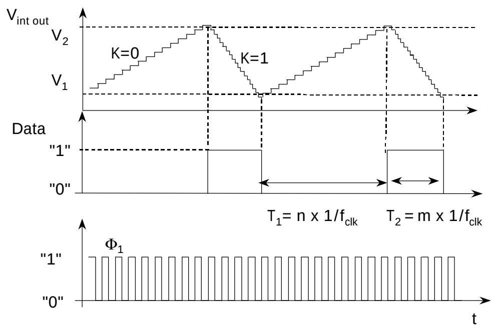

Fig. 9 Working principle of SC-relaxation oscillator

The principle functioning of the circuit is illustrated in Fig. 9. The sensor capacitance C_X, offset compensating capacitance C_0 and reference capacitance C_r are charged and discharged alternating and the sum (or difference) of charges is integrated onto C_F until the upper reference voltage level V_2 is reached. The direction of integration is reversed by the output signal k of the comparator, which is also the PWM output. The capacitance C_F is then discharged until the lower reference voltage V_1 is reached. The discrete integration times T_1 and T_2 result from the capacitance dependent small voltage steps. The minimum number of voltage steps in each integration cycle determines the theoretical resolution of the oscillator. The transfer function is nearly independent of the value of the reference voltages, but they must be kept stable during one conversion cycle.

The ASIC is capable of reading out the capacitance change $\Delta C = C_x - C_0$ of the pressure sensor with a resolution of 10 bit = 0,1 % at a sampling speed of approx. 100 samples/s. It is powered with a single 3,5 V supply and provided with a clocking of 125 kHz by the telemetry ASIC. The sensitivity and measurement range of the oscillator circuit are determined respectively limited by the value of the on-chip capacitance C_r, the resolution is set by the ratio of the integrating capacitance C_F and C_r , which are both located on the signal processing die.

The pulse width modulation (PWM) output data is

$$dT(C_X) = \frac{\Delta T}{T} = \frac{T_1 - T_2}{T_1 + T_2} = \frac{C_X - C_0}{C_r} = \frac{\Delta C}{C_r}$$

(1)

The system architecture ensures offset compensation of C_X on the sensor side by building the difference of pressure dependent sensor capacitance C_X and pressure independent offset compensation capacitance C_0. Because of the varying sensitivity of sensor and circuitry a final external calibration is still needed for the targeted biomedical overall measurement accuracy of 0,1 % (≅ 100 Pa = 1 mbar) pressure reading.

The oscillator circuit consists of transmission gates as analog switches, a voltage reference network followed by adaptive biasing operational amplifiers for charging the parasitic capacitances at both nodes of sensor and reference capacitance, an operational amplifier in the SC-integrator and a two-phase offset compensating comparator [25]. A few additional digital cells form the switch control logic and provide the proper two-phase clocking. The comparator determines if the lowest possible phase jitter of the oscillator can be reached. This phase jitter limits the reachable resolution while running the oscillator with maximum measurement frequency.

Telemetry

The telemetry chip [26] is performing data reduction for the sensor PWM signal, encoding, and modulation. It supplies a stabilized 3.5 V voltage supply and 125 kHz clocking for the read-out electronics extracted from the external transmitter carrier frequency of 4 MHz. The chip is connected to two planar Au-electroplated micro-coils for data and energy transmission.

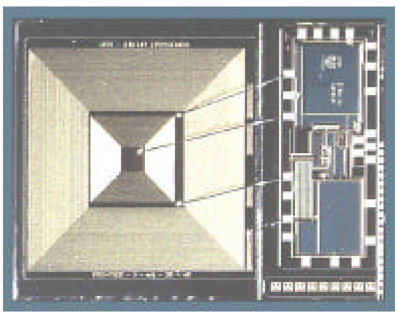

Fig. 10 Electroplated Au micro-coils (5 mm x 5 mm) for data & energy transmission, telemetry ASIC

Mounting and assembly

The two chips are both processed on one side of the silicon substrate only.

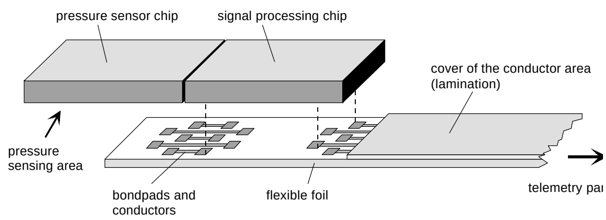

Fig. 11 Schematic of flip-chip bonding

Therefore an advanced chip to carrier substrate mounting by flip-chip bonding (Fig. 11) is possible. The flip-chip bonding ensures an one-step electrical and mechanical interconnect. Five chip to chip interconnects are necessary: The two capacitances C_x and C_0 are connected to the reference voltage network on the signal processing ASIC. Because of the pn-junction isolation the connection of the two substrates is necessary, too. The signal processing needs four external connections to the telemetry device: CLK, V_{DD}, V_{SS} and DATA. The fifth pad can be used e.g. for switching the device in test-mode.

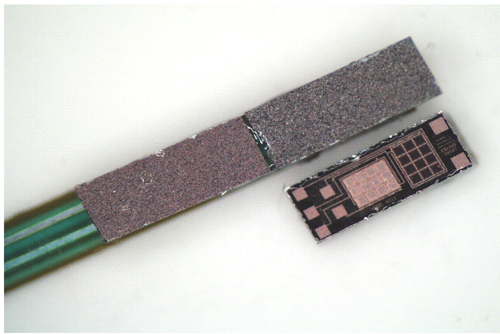

Fig. 12 Flip-chip configuration of pressure sensor & read-out ASIC and top view of pressure sensor (right)

The substrate for the biomedical application depicted in Fig. 12 is a flex-foil carrier with thin, photolithographically patterned conductor lines. Because these conductors are etched with nearly the same precision as the on-chip metal lines, the associated additional offset capacitance is reproducible and does not lower the offset capacitance matching of the sensor device. The mounting of the single chips can be performed by bonding them face-down onto the conductor lines.

A conventionally wire-bonded sample, which is used for prototyping purposes is shown in Fig. 13. The complete sample is coated by a moisture resistant barrier and a bio-compatible silicone based coating . This procedure does not reduce the mechanical sensitivity of the sensor membrane significantly and ensures a long-term implantation time. The bio-compatibility was enhanced by a special treatment of the standard Cu-based flex substrates and the use of Au-electroplated micro-coils.

Fig. 13 Coated implant with pressure sensor (left), telemetry ASIC and micro-coil (right)

MEASUREMENT RESULTS

The basic sensor characteristic and variation predicted by the simulations has been verified by extensive measurements and is shown in Fig. 14. The sensor temperature coefficient of sensitivity is $TK_S = -240$ ppm/°C, which is one order lower than for uncompensated piezoresistive pressure sensors.

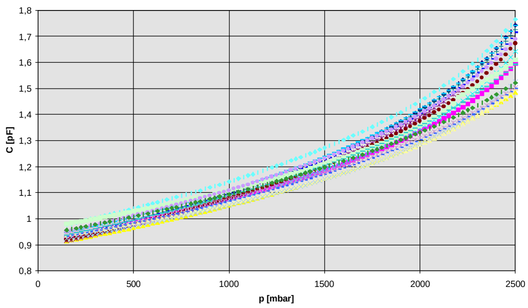

Fig. 14 Measured variation of sensor capacitance sensitivity and offset

The sensor has been tested successfully together with a discrete prototype of the interface electronics (Fig. 15), together with the custom designed read-out ASIC and with a commercially available oscillator-IC for non-medical applications.

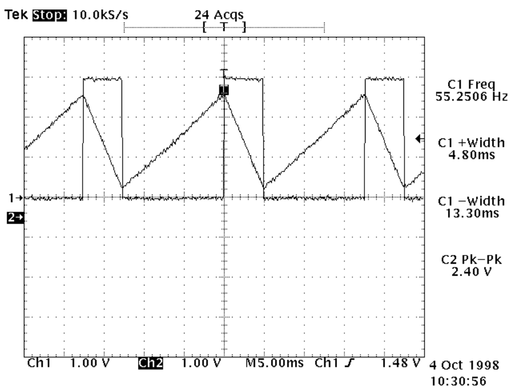

Fig. 15 PWM Output of sensor + interface prototype

The measured 10 bit sensor output resolution gives an externally calibrated accuracy of approx. ± 1 mbar which is needed for a variety of medical research and diagnosis purposes. The sampling rate is ≥ 30 samples/s at the moment, which is sufficient to resolve blood pressure curves dynamically. A transmission distance of approx. 5 mm could be successfully demonstrated.

The power consumption of below 350 µW enables the pressure sensing system to be powered telemetrically for the targeted biomedical application.

CONCLUSIONS

A highly miniaturized, simply constructed two-chip pressure measurement system is presented. Its low power consumption will allow the desired telemetric operation. Because of the applied advanced mounting and assembly technology the hybrid integration of a capacitive sensor and a read-out circuitry is possible despite of the small capacitances to be read out. Due to small size the system is going to be used in a telemetrically powered passive intra-ocular implant in the near future. Using advanced mounting and assembly technology the hybrid integration approach seems to be promising for successful commercialization of microsystem technology in the area of biomedical telemetric implants. These results of this work are of comparable performance to previously presented work, but because of the limited number of systems needed in the medical application field the hybrid approach seems to be more promising for successful industrialization.

ACKNOWLEDGEMENTS

Parts of this work have been carried out under grand number 16SV128/7 of the German Ministry for education and research (BMBF) 1995-1998.

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