A MICROMACHINED INTEGRATED SENSOR WITH ON-CHIP SELF-TEST CAPABILITY

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ABSTRACT

A batch-fabricated multielectrode microprobe for extracellular biopotential recording in the central nervous system is described. The probe consists of thin-film conductors insulated with CVD dielectrics and supported on a micromachined silicon substrate defined using a deep boron etch-stop. Probe shank widths as narrow as 20μm are possible. Interface circuitry for eventual use on the probe offers per-channel amplification and multiplexing of the recorded data onto a single output lead. A 12-channel version of this circuitry dissipates 4mW from a single 5V supply and occupies an active area of 1.75mm² in 6μm E/D NMOS technology. Additional on-chip circuitry for on-line electrode testing is also described. This circuitry requires no additional leads and only a minimal increase in die area.

INTRODUCTION

A wide variety of integrated solid-state sensors are now being developed to extend microcomputer-based control into new areas such as health care, transportation, and automated manufacturing. These devices typically employ the full range of integrated-circuit process technology for their realization in addition to specialized processes for selectively shaping the silicon substrate (micromachining), packaging, and the deposition of special sensing materials. The use of on-chip interface circuitry with these transducing structures is now finding its way into production devices, offering the promise of reliable, bus-compatible, addressable sensors of wide applicability. In some cases, these integrated sensors offer higher reliability, higher system performance, and/or lower system cost than previous discrete or hybrid designs; however, in other cases the combination of small size with on-chip signal conditioning is allowing instrumentation not previously possible. Whatever the application, issues such as full process compatibility, the choice of an interface-circuit technology, partitioning the system electronics, selection and standardization of output signal formats, addressing modes, testability, and packaging are still largely unresolved.

This paper will comment on the above issues and illustrate them by examining one particular integrated sensor — a multichannel multiplexed intracortical microelectrode array capable of recording the activity of single neurons in the central nervous system. This probe should allow the simultaneous recording from many neurons separated in depth through the cortex and thus allow new insights into the signal processing techniques employed by neural structures. Over the longer term, it offers the possibility of implementing closed-loop control in a variety of neural prostheses. Early forms of such microprobes [1,2] were among the first integrated sensors, and the present applications demand state-of-the-art solutions to a number of the problem areas mentioned above.

EXTRACELLULAR SINGLE-UNIT RECORDING

The use of metal microelectrodes (usually in the form of sharpened pins) has been one of the principal techniques by which physiologists, over many years, have studied the nervous system at the cellular level. Much has been learned about the operation of single neurons, but relatively little is known about their organization as functional circuits. In order to explore neural signal processing at the cellular level, simultaneous recording from many points through the circuit is certainly required; however, previous electrode technologies have not permitted the realization of multielectrode arrays small enough to avoid excessive tissue damage in use. The need to conduct such studies on a chronic basis further complicates the situation. Similar requirements hold for prosthetic applications.

Figure 1 shows a typical extracellular recording situation. When the neuron receives sufficient stimulus, its cell membrane depolarizes causing ionic currents to flow in the vicinity of the cell. In order to record the resulting voltages, the probe must be small enough to approach the cell closely with a minimum of tissue damage. Typical signal levels are a few hundred microvolts at best, with a frequency spectrum extending to perhaps 6KHz. Empirically, acceptable neural recordings require electrode impedances of 5-15MΩ at 1KHz, which for a metal electrode represents a recording area of 50-100μm² and a series electrode capacitance of 10-20pF.

Figure 2 shows a typical probe structure. A silicon substrate supports an array of thin-film conductors which are insulated above and below by deposited dielectric layers. Openings in the upper dielectric are used to define recording sites. On-chip electronics is needed to reduce the recording impedance levels, minimize the output leads, prevent crosstalk, and amplify the microvolt signal levels. These requirements are common to many sensing appli-

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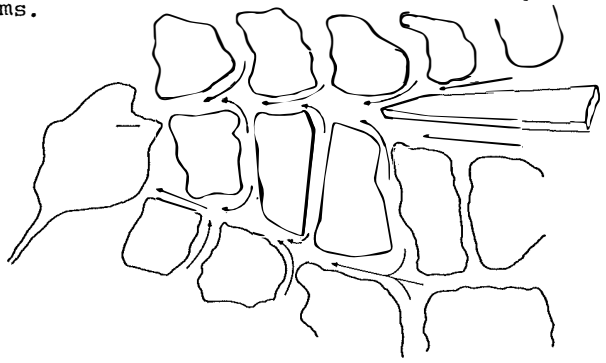
cations. Lead minimization is especially important here, and since total long-term insulation of the flexible output leads is difficult, on-chip circuitry is necessary. Circuit area must be minimized, and the total chip power dissipation should be less than 5mW to limit the temperature rise on the probe. The structure must be strong enough to penetrate the pia arachnoid over the brain and must be realizable with a process capable of high yield. In the past, the lack of such a process has prevented lithographically-defined probes from being generally available. Recent advances, particularly in the silicon micromachining processes needed for shaping the silicon substrate, have overcome these problems.



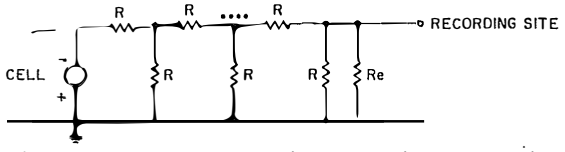


Fig. 1: Extracellular Single-Unit Recording

PROCESS COMPATIBILITY

Figure 3 shows a summary of the process for a passive probe. Fabrication begins with a p-type (100)-silicon wafer of standard thickness and doping. The wafer is first oxidized and patterned to define the intended probe areas. Next, these areas are subjected to a deep boron diffusion (1175°C for 15 hours) to define the probes. The field oxide is then stripped and a combination of thermal oxide, CVD silicon nitride, and silicon dioxide is deposited to form the lower dielectric. Conductors of tantalum or polysilicon are next deposited and patterned, followed by the deposition of the upper oxide-nitride-oxide dielectrics. These layers are patterned using a plasma process to open the recording sites and bonding areas. With the resist still in place, the exposed conductor surface is ion milled to remove any oxide, and gold is inlayed in the recording sites using ion-beam depostion. Lift-off is then used to remove the gold from everywhere except these sites and the bonding areas. The gold recording sites are thus self-aligned. The field dielectrics outside the intended probe areas are now removed using a plasma etch. Finally, the wafer is subjected to an unmasked etch in ethylene diamine-pyrocatechol-water (EDP) to separate the individual probes. This final etch dissolves the wafer and stops on the p+ probe substrate. It does not attack any of the other materials used. The completed probe chips are removed from the etch, ready for bonding.

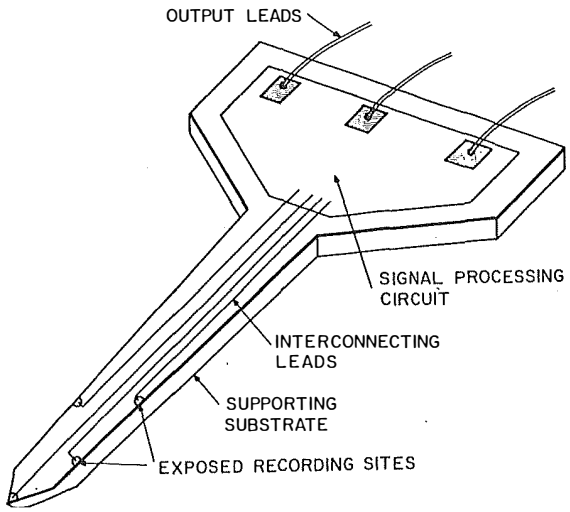


Fig. 2: A Multichannel Multiplexed Intracortical Recording Array

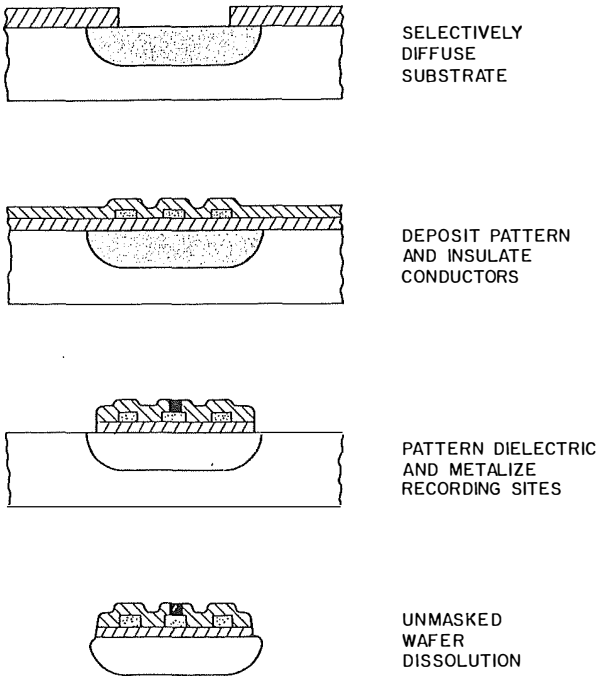


Fig. 3: Fabrication Process for a Passive Probe

This process is capable of high yields, results in very small structures, and yet requires only single-sided processing on wafers of normal thickness. All etching steps are highly selective and self-stopping. All probe features can be controlled of the order of ±1μm or better. The silicon substrates can be as thick as 15-20μm and of arbitrary two-dimensional shape. The probe edges are slightly rounded, reflecting boron outdiffusion into the field regions of the wafer. These structures are strong yet flexibile and are capable of repeated insertion through pia arachnoid without damage. A minor process modification can be used to form a self-aligned support rib down the probe shank to increase its thickness without loss of dimensional con-

trol. Finally, full circuit compatibility is achieved by doping only the perimeter of the rear probe area during the deep boron diffusion, leaving the center area lightly doped for the circuitry. The passive probe process requires only four masks, two of which are shared with the circuit process when on-chip circuitry is used.

Figure 4 shows two views of a passive probe tip, while Fig. 5 shows neural activity recorded by such a structure. Figure 6 shows a probe containing active test circuitry used to establish process compatibility.

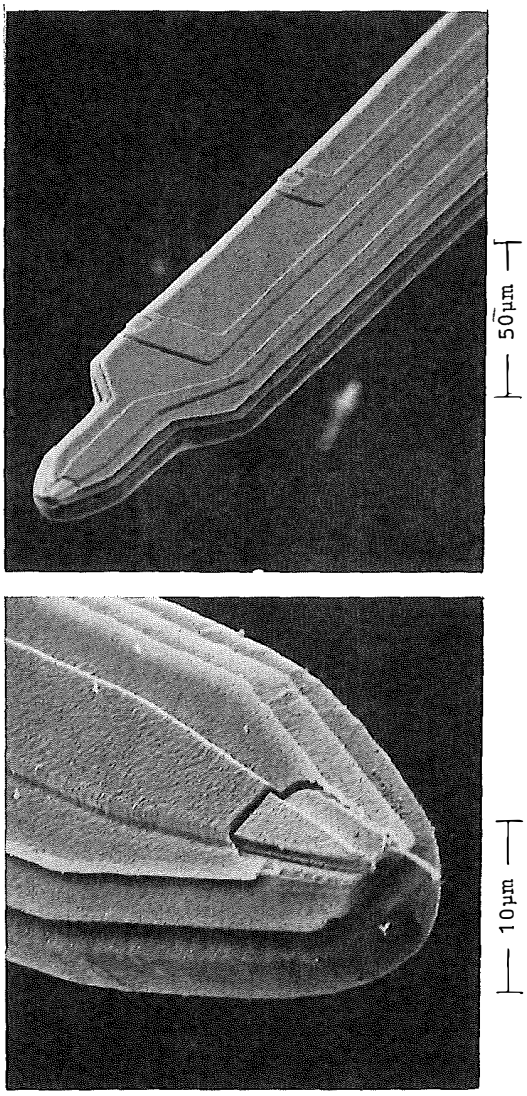


Fig. 4: Two SEM Views of the Tip Area of a Multichannel Probe

Protection of this probe structure in vivo represents a particularly significant challenge since the sensor must work in saline for extended periods and there is no room for a conventional package. Deposited dielectrics or other films must be used to ensure sensor integrity. While these packaging problems are not yet solved, some recent results are encouraging. On the probe shanks themselves, the use of CVD oxynitride films themselves may be adequate, particularly since we are primarily concerned with the ac integrity of the

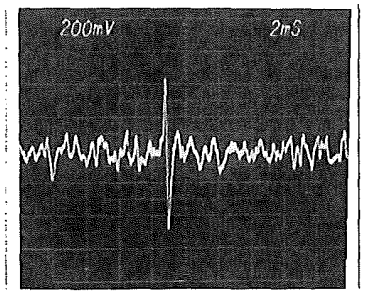


Fig. 5: Activity from Single Cortical Neurons Recorded with a Multielectrode Probe. The vertical scale is about 100µV/div.

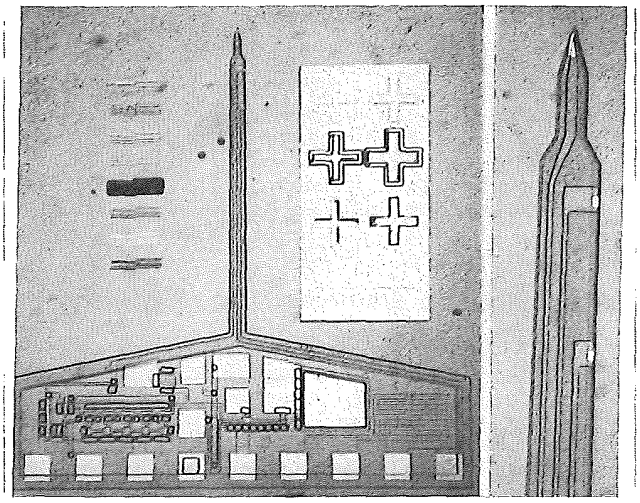


Fig. 6: A Three-Electrode Microprobe with On-Chip Test Circuitry for Studying Process Compatibility. The circuit technology is silicon-gate NMOS. An enlargement of the tip area is shown at the right.

dielectrics (100Hz - 6KHz) and the interconnect area is relatively small (0.01mm²). Over the circuitry (where appreciable DC voltages are present), as well as over the flexible output leads, thicker dielectrics can be used since they need not be photoengraved. In these areas, polymeric films appear attractive. Hughes [3] has recently reported leakage of less than 400pA/cm² for 12.5µm-thick films of Pyralin 2555 (a polyimide) after more than 250 days in saline, and the University of Missouri [4] has reported promising results with parylene. While much work remains, the packaging of sensor chips for chronic implantation is the subject of efforts in a number of laboratories, and the solutions to these problems promise considerably broader impact on the overall sensor community.

CIRCUIT DEVELOPMENT

Because of the low signal levels and high impedance associated with this sensor, on-chip circuitry is essential. While perhaps less obvious, similar situations exist for many other sensors and sensing applications. Figure 7 shows the electronics being developed for the probe. Per-channel amplifiers are

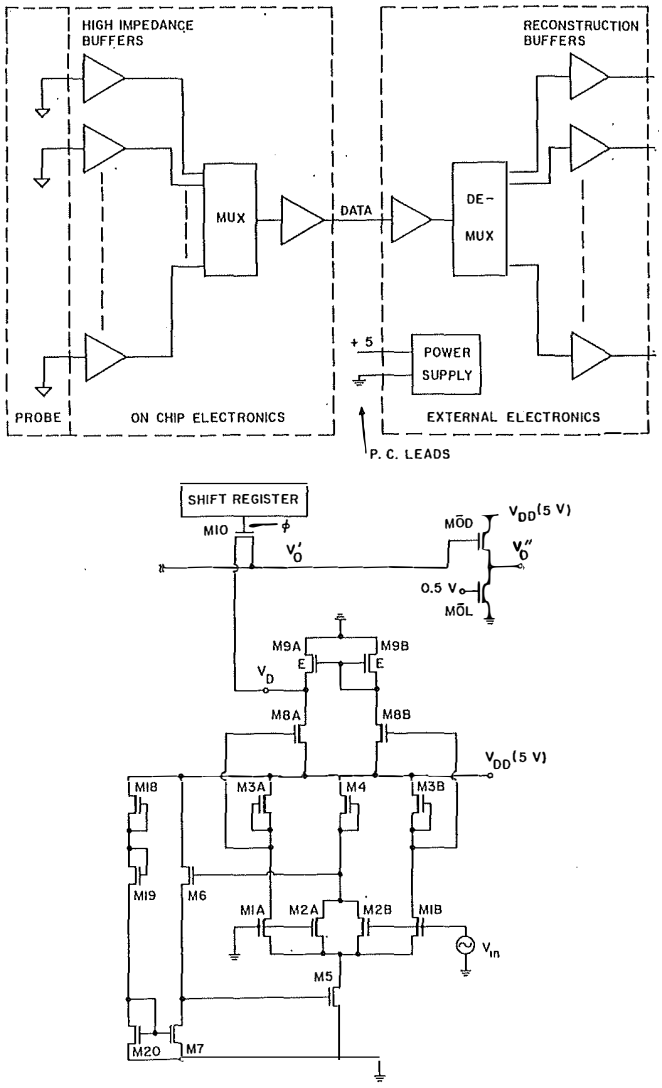


Fig. 7: Diagram of the Overall System Electronics to be used with the Multiplexed Probe and a Schematic of the On-Chip Analog Circuitry.

used for each of eleven recording electrodes. These amplifiers drive a 12:1 analog multiplexer and broadband output buffer. A total of three leads are thus required by the probe (VDD, GND, and DATA). The circuitry must operate without substrate bias. The multiplexer is driven by an on-chip two-phase dynamic shift register and a 200KHz clock to produce a per-channel bandwidth of 6KHz. A twelfth channel is used for a synchronization pulse which is decoded externally and used to regenerate the sample clock to control the demultiplexer. The external circuitry has been fully implemented and shown capable of handling 20μV neural signals with up to 40 data channels.

The on-chip circuitry to be eventually included on the probe has been implemented using a silicon-gate E/D NMOS LOCOS process as shown in Fig. 7. The input signal from the electrode is applied to one side of a differential pair of depletion-mode transistors (M1-2) capable of handling input signals near ground. This input signal is amplified and then presented to a differential-to-single-ended converter (M8-9) before being passed to the analog multiplexer (M10) and output buffer. The allowable amplifier gain is limited to about 100 by achievable input offsets. Figure 8 shows a prototype of the full twelve-channel signal processor. The die size is 2mm x 2mm in 6μm features, including pads and a variety of peripheral and per-stage test devices and probe points. Deleting these test devices, the active die area is about 1.75mm². The measured signal gain is 20 and the total chip power dissipation is 4mW at VDD= 5V.

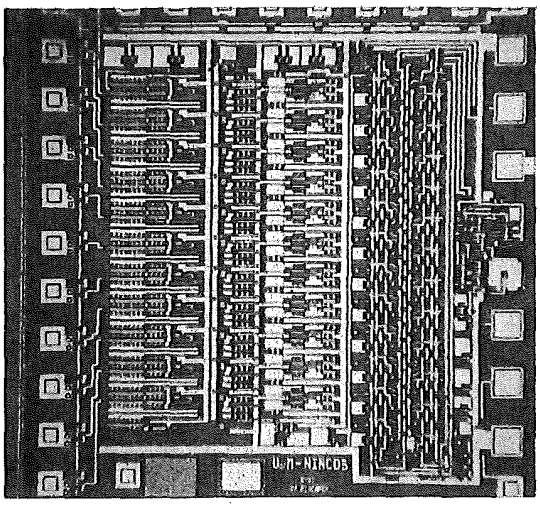


Fig. 8: A Twelve-Channel Signal Processor Prototype of the On-Chip Probe Electronics.

TESTING CONSIDERATIONS

One of the difficulties associated with the use of on-chip circuitry is that it may block the direct test access to the transducer which is available in nonintegrated system implementations. In our application here, the ability to test initially in vitro and then periodically in vivo is of major importance in evaluating device integrity in the face of long-term exposure to saline. With passive microelectrodes, the impedance magnitude is the most reliable known indicator of electrode recording ability and is the only parameter allowing electrode changes to be distinguished from physiological changes near the recording site. Thus, it is essential to preserve electrode testability in any integrated sensing structure and to do so without sacrificing normal operating performance.

Discrete electrode impedance is typically measured by injecting a known 1KHz current (e.g., 1nA) between the electrode and the recording amplifier. The resulting electrode voltage is a measure of the recording site impedance in parallel with any shunt input capacitance. Figure 9 shows the additional circuitry required to preserve testability in this integrated sensor. When the supply voltage is raised above its normal 5V level, the testing mode is enabled. The periodic strobe-channel enable signal is used as input to a four-stage counter, which produces a 1KHz clock for use in impedance testing. This clock is attenuated to a 50mV level and coup-

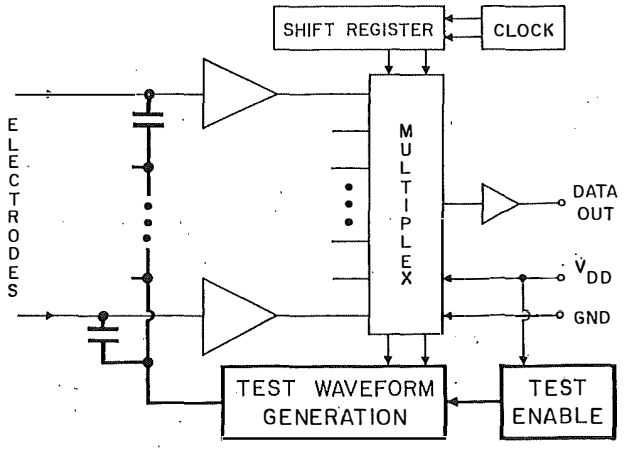


Fig. 9: Block Diagram of the On-Chip Electronics Showing the Additional Circuitry Required to Allow Self-Testing

led to the input leads using 50fF metal-to-poly crossover capacitors. The injected charge develops a signal level on the probe input lines of nominally 250μV, depending on the recording impedance of the particular electrode. This induced voltage is amplified and multiplexed out in the normal way.

This self-testing circuitry preserves the ability of the electrodes to be tested at any time on demand (in vitro or in vivo), requires no additional leads and only a minimal increase in die area. Being derived from the on-chip clock and passed through the normal recording channels, it also offers an additional functionality test for virtually all of the on-chip circuitry. Used in vitro with no immersion of the probe, it also offers a potential means for calibrating the gains of the individual recording channels before use.

In this particular application, great precision in the signal processing circuitry is relatively unimportant since most information is contained in the presence, general amplitude, shape, and time occurance of each neuronal event. In many other sensing applications, great precision is required not only in the transducer but also in the signal processing circuitry [5]. In most cases, frequency- or time-multiplexed digitally-encoded output formats will be used. In these applications, as in the present one, process compatibility, circuit partitioning, high-performance analog circuit design, and packaging are central issues which are certain to occupy the sensor community for some time to come. As these issues are clarified and the problems overcome, however, entirely new application areas for integrated electronics will certainly emerge and become practical.

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