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ADVANCED SENSORS TECHNOLOGY SURVEY

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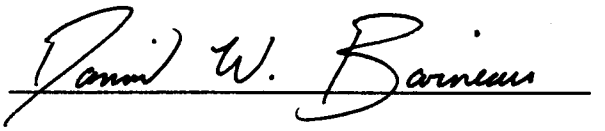

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EXECUTIVE SUMMARY

PROJECT OBJECTIVES

To assess the state-of-the-art in advanced or "smart" sensors technology for NASA Life Sciences research applications with an emphasis on those sensors with potential applications on the SSF

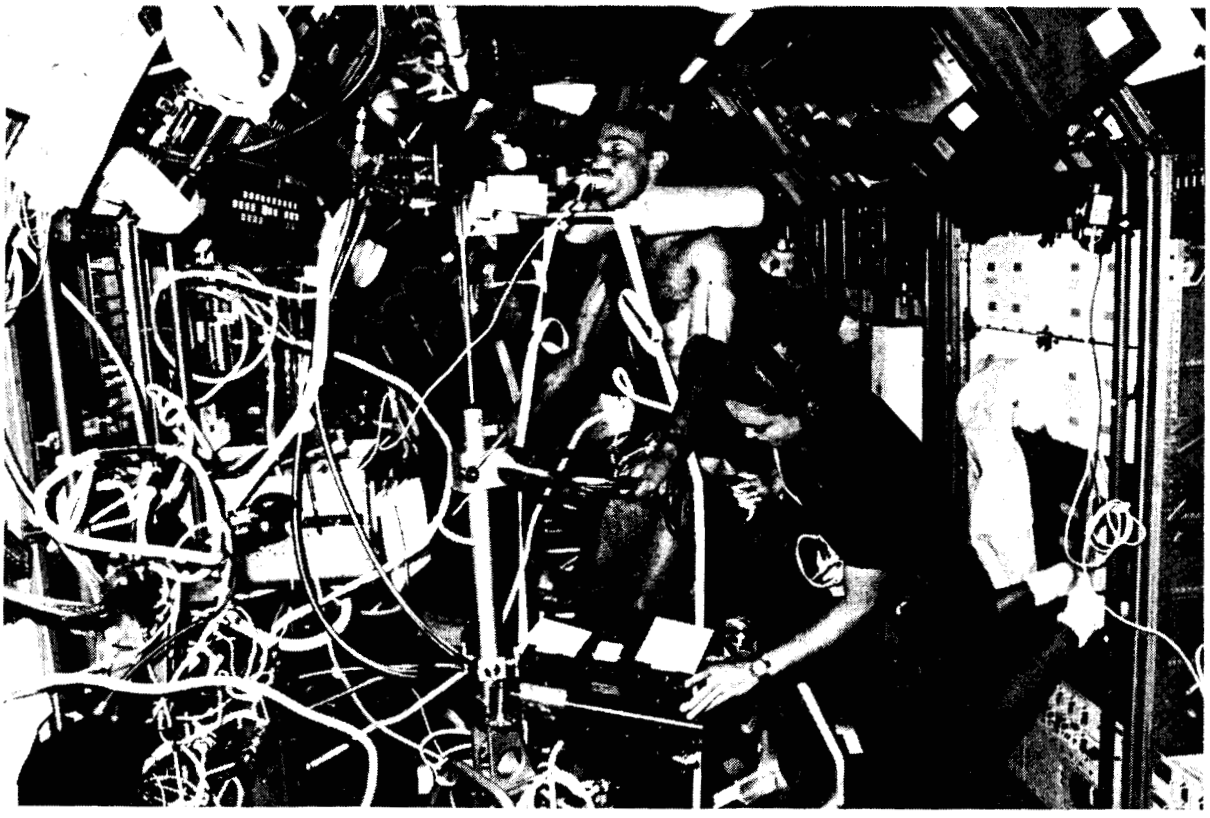
- Conduct literature reviews on relevant advanced sensor technology
- Interview various scientists and engineers in industry, academia, and government who are knowledgeable on this topic
- Provide viewpoints and opinions regarding the potential applications of this technology on the SSF
- Provide summary charts of relevant technologies and centers where these technologies are being developed

PROJECT METHODOLOGY

- A team of experienced biomedical technologists in both industry and academia was selected to provide comprehensive coverage of the topics
- Each expert covered his area of expertise using literature reviews, interviews, and personal opinions
- The written reports were combined, compiled, and edited for submittal as a final report

BACKGROUND

- Various biomedical sensors, with considerable overlap in function, have been used by NASA in the past
- Each sensor has had its special-purpose processor circuitry attached which prompted an unnecessary proliferation of hardware configurations
- Life Sciences research-intensive installations, such as Spacelab, have demonstrated the need for systems planning to simplify future sensor deployment activities



- The SSF represents an opportunity to integrate a new systems approach to multiple sensors and to their signal processing interfaces
- "Smart" sensor technology holds a promise for solving life science research instrument deployment problems on spacecraft of the future

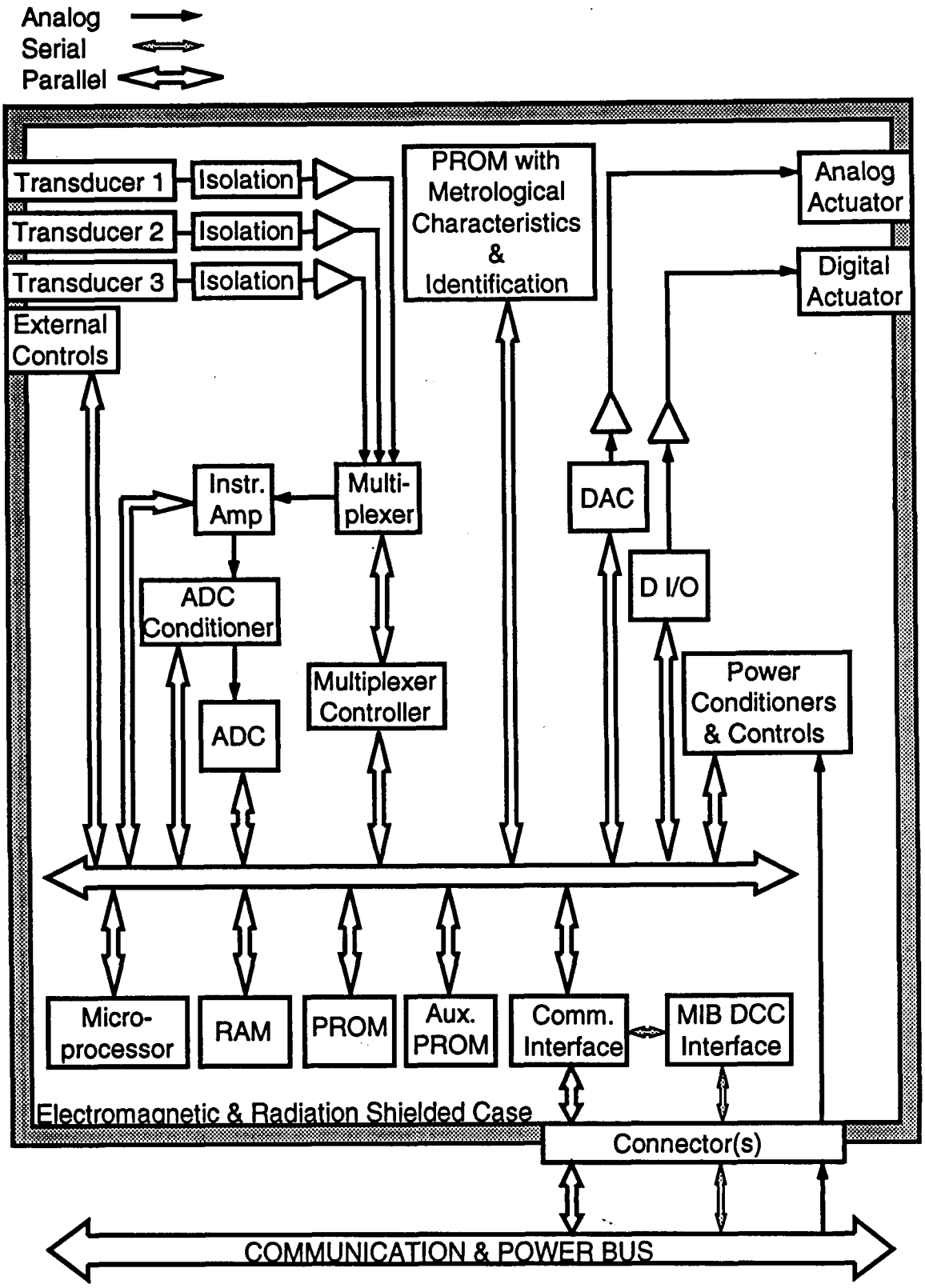
WHAT IS A "SMART" SENSOR?

A "smart" sensor is a device which contains, within the same enclosure, both sensing and signal processing functions and, perhaps, also contains controls and actuators.

Sensor-incorporated functions include:

- Environmental Compensation
- Signal Conversion
- Diagnostics
- Communications

GENERALIZED BLOCK DIAGRAM OF A "SMART" SENSOR



SYSTEM ADVANTAGES DERIVED FROM USE OF "SMART" SENSORS

- Improved performance, accuracy, and reliability
- Minimized setup and calibration time by the user
- Distributed signal processing and system intelligence
- Reduced transducer inventory due to the multifunctionality and adaptability provided by programmability
- Reduced central processor workload
- Reduced communication network traffic
- Improved operational reliability through self testing and fault isolation
- Increased compatibility with new object-oriented programming methods

Features:

- Enhanced interchangeability of sensors
- Programmability of signal processing parameters
- Re-configurable for various input measurements
- Re-configurable outputs for interface with various subsequent devices
- Built-in self-calibration and self-diagnosis
- Real-time data reduction, sampling, and computation of derived measures

TYPES OF "SMART" SENSORS WITH SSF APPLICABILITY

- Electrophysiological Sensors
  - Biopotential Sensors
  - Impedance Sensors
  - Biomagnetic Sensors

- Physical Sensors
  - Temperature
  - Pressure
  - Acceleration
  - Flow
  - Sounds

- Chemical/Bioanalytical Sensors
  - pH
  - Blood Analytes
  - Urine Analytes
  - Hydroponic/Bioprocessing Analytes
  - Water Quality Analytes
  - Air Quality Analytes

ADVANCED SENSOR TECHNOLOGY
SSF APPLICABILITY MATRIX

Near-Term (1992-1995)	Mid-Term (1996-2000)	Far-Term (2000 +)
Electrophysiological		
Wet/Dry Electrodes	Insulated Electrodes	SQUIDs
Programmable Signal Conditioners	DSP	
Physical		
Silicon, Hybrid	Silicon, Monolithic	Silicon, "Hyperintegrated"
Silicon Fused Bonding		
Surface Micromachining	Bulk Micromachining	Micromachines/Actuators
Fiberoptic Pressure Sensors		
Chemical/Bioanalytical		
Ion-Selective Electrodes	Semiconductor Microsensors	Surface Plasmon Resonance
Visible Spectroscopy	IR & NIR Spectroscopy	Micro-Mass Spectroscopy
Amperometric/Potentiometric	Thermoelectric	Raman Spectroscopy
Photochemical	Optrodes	NMR Spectroscopy
	Surface Acoustic Wave	

MAJOR SOURCES OF "SMART" SENSOR SYSTEM INFORMATION

- SMART HOUSE Program
- National Laboratories (Sandia, Los Alamos, Lawrence Livermore)
- Sensors 2000! Program
- Patient Monitor/Ambulatory Monitor Industry
- Air Force Flight Dynamics Laboratory
- Geophysical Exploration Industry

CONCLUSIONS

- Smart sensor **technology development** has outpaced smart sensor **utilization**
- Silicon sensor technology is ideal for the development of integrated smart sensors due to the processing techniques which are shared with microelectronics circuitry
- Major commercial applications of smart sensors over traditional sensors involve requirements for:
  - long-term performance without maintenance
  - greater accuracy
- Smart sensors providing self-test, calibration, control, communication, and decision-making capabilities are going to be prevalent in the 1990s
- Whether a device is a smart sensor or an instrument depends more upon its role within a system, its dependence on other physical devices, the level of integration within the sensor package, and its mode of communications with the user
- A systems design approach (involving communications, software, display, human factors, etc.) must be used to maximize the effectiveness of smart sensors
- Digitally programmable signal processing in external signal conditioners is a good first step toward the integration of advanced sensors into future Life Sciences research instrumentation
- Increased data rates and intensive experiment control requirements associated with SSF place increased demands on the human interface required for control of advanced sensors and their data
- Smart sensors have numerous ideal applications on SSF and future space biomedical research programs
- Direct digital signal processing holds a promise for increased system **flexibility** in the far-term period

RECOMMENDATIONS

- Develop an expertise in "universal" programmable signal conditioners
- Develop smart sensor platform concepts to provide a wide range of measurement and control functions, depending upon the sensor installed
- Adopt a systems approach to biomedical data acquisition that incorporates the features of the Medical Information Bus (MIB)
- Follow the development of commercial-off-the-shelf versions of ambulatory monitors for early adoption of cost-effective smart sensors
- Utilize RF and IR telemetry to monitor and control instrumentation containing smart sensors
- Assess the impact of smart sensors on the feasibility of expanding inflight biomedical data analysis
- Consider the use of "smart" cards for numerous purposes on SSF
- Develop a microminiature sensor capability to support future Life Sciences research which will expand at the cellular level, where the emphasis shifts toward increased requirements for biochemical measurements

ACRONYMS AND ABBREVIATIONS

A/D	Analog-to-Digital
AAMI	Association for the Advancement of Medical Instrumentation
ABPM	Ambulatory Blood Pressure Monitor
AC	Alternating Current
ADC	Analog-to Digital Converter
AHA	American Heart Association
AI	Artificial Intelligence
ANSI	American National Standards Institute
ASIC	Application-Specific Integrated Circuit
BCC	Bedside Communications Controller
BMAC	Biomedical Monitoring and Countermeasures
BP	Blood Pressure
bpm	beats per minute
CAE	Computer-Aided Engineering
CDUSS	Chemical Defense User Safety System
COTS	Commercial Off The Shelf
CRT	Cathode Ray Tube
CSIC	Customer-Specific Integrated Circuit
D/A	Digital-to-Analog
DC	Direct Current
DCC	Device Communications Controller
DEMUX	Demultiplexer
DMS	Data Management System
DSP	Digital Signal Processing
ECG	Electrocardiogram
ECU	Electronic Control Unit
EEG	Electroencephalogram
EEPROM	Electrically Erasable Programmable Read-only Memory
EMG	Electromyogram
EOG	Electrooculogram
EPROM	Erasable Programmable Read-only Memory
EVA	Extravehicular Activity
F.S.	Full-scale
FDA	Food and Drug Administration
FEM	Finite Element Modeling
FET	Field Effect Transistor
FFT	Fast Fourier Transform
FIR	Finite Impulse Response
FM	Frequency Modulation
GSR	Galvanic Skin Response
high-Q	Resonant circuit with low losses
HPSAPMD	High Pressure Stand-Alone Pressure Measurement Device

HUD	Heads up Display
HVAC	Heating, Ventilating, and Air Conditioning
Hz	Hertz (cycles per second)
IEEE	Institute of Electrical and Electronics Engineers
IFPDAS	Inflight Physiological Data Acquisition System
I/O	Input/Output
IR	Infrared
JFET	Junction Field Effect Transistor
LAN	Local Area Network
LCD	Liquid Crystal Display
MCU	Microcomputer Unit
MDDL	Medical Device Data Language
MDM	Multiplexer/Demultiplexer
MIB	Medical Information Base
MIDR	Modular Intelligent Data Recorder
MOSFET	Metal-Oxide Semiconductor Field Effect Transistor
MUX	Multiplexer
NIR	Near Infrared
NIST	National Institute of Standards and Technology
nm	nanometer
PC	Personal Computer
PCG	Phonocardiogram
pCO_2	Partial pressure of carbon dioxide
PM	Payload Manager
PMDM	Payload Multiplexer-Demultiplexer
pO_2	Partial pressure of oxygen
PROM	Programmable Read-only Memory
PTCM	Programmable Transducer Control Module
PWM	Pulse-width modulation
RAM	Random Access Memory
REM	Rapid Eye Movement
RMS	Root Mean Square
S/N	Signal-to-Noise Ratio
SAPMD	Stand-Alone Pressure Measurement Device
SEI	Space Exploration Initiative
SPDS	Safety Parameter Display System
SQUID	Semiconductor Quantum Interference Device
SSF	Space Station Freedom
SSPIDR	Solid State Physiological Inflight Data Recorder
VAC	Volts, Alternating Current
VDC	Volts, Direct Current

1.0 INTRODUCTION

1.1 OBJECTIVES

The primary objective of the *Advanced Sensors Technology Survey* was to conduct a rapid assessment of the state-of-the-art in advanced or "smart" sensor technology for NASA Life Sciences research applications. Although this report contains the complete results of the survey, one of its major objectives is to emphasize those sensors with major potential applications in future NASA Life Sciences Programs, including Space Station Freedom (SSF) and the Space Exploration Initiative (SEI). The detailed objectives of the study plan were as follows:

- Conduct literature reviews of relevant advanced sensor technology.
- Interview various scientists and engineers in industry, academia, and government who are knowledgeable on this topic.
- Provide viewpoints and opinions regarding the potential applications of this technology on SSF.
- Provide a summary chart of the relevant technologies and centers where these technologies are being developed.

1.2 METHODOLOGY

Since the topic of advanced sensors and their systems design implications is extremely diverse, it is best covered by utilizing the backgrounds and experiences of several scientists and engineers working with various institutions. One complication of the study of this topic is the reduced availability of information due to the proprietary nature of work that is being conducted. Five experienced consultants in industry and academia were placed under subcontract to provide their insights and perspectives on advanced sensor technology. These professionals were selected to provide a well-rounded, comprehensive coverage of the topic. They were given overall guidance on the objectives of the project, but they were not restrained on the content or style of their research and reports; thus, they were allowed to emphasize those topics that they felt were most important.

The contributors conducted comprehensive literature reviews of the current state-of-the-art in advanced or "smart" sensor technology, with an emphasis on those biomedical applications in their particular areas of expertise. Leaders in both research and industry were contacted for information regarding the status of recent developments. By using this approach, the network of access to information was broadened significantly. Interviews were conducted with numerous scientists and engineers in industry, academia, and government who are knowledgeable on the topics and sub-topics of the survey. Special efforts were made to include interviews of NASA personnel and technologists at the various government laboratories and facilities.

The contributors were asked to provide their insight and perspective on the potential SSF applications of the technology, including a systems overview of integrated and programmable approaches to signal conditioning and data acquisition/control. Each contributor was allowed approximately six weeks to complete the survey and submit a written report containing the findings that they considered most important.

This report is a combined, edited, and expanded compilation of the written results submitted by the study team.

1.3 SCOPE

Within the scope of this effort, it was impossible to review all the data acquisition and data processing issues involved with advanced sensors; thus, the technology identified and described in this report must be viewed as representative rather than exhaustive. The topics emphasized in this report were those that the contributors judged to be the most important biomedical sensor technologies in relation to potential SSF/SEI usage.

The technology described in this report was not, in general, limited by flight-worthiness criteria. Obviously, there are rigorous standards for materials and engineering required to flight qualify biomedical instrumentation for space flight; and, at some point, candidate biomedical devices must be evaluated for their potential to meet these standards.

The body of this report begins with a set of descriptions/categories, that cover the types of "smart" sensors that are currently most prevalent. The descriptions are followed by a listing of the advantages of utilizing smart sensors, as well as some specific examples. The report then relates these technologies to those areas which are most applicable to NASA Life Sciences usage. Incorporated in this analysis is the identification of those NASA systems with which the smart sensor technologies must interface. These include the SSF core communication and data transmission systems, the interaction of the sensors with the user/astronaut, and the utilization of the generated data.

The subsequent section describes a number of commercially available technologies. These descriptions detail a variety of hardware/software items that could be quickly integrated into the available NASA inventory. Finally, a set of recommendations are made which will enable NASA to make full use of the developments in this field.

The tutorials contained in this report (Appendix C) were inserted for those readers desiring a more detailed description of the background technology.

Boldfaced type is used to identify those thoughts that are particularly relevant to SSF/SEI applications.

2.0 WHAT IS ADVANCED SENSOR TECHNOLOGY?

2.1 DEFINITIONS

Advances in solid state sensor technology, microelectronics, micromachining, and electronics packaging technology have made it possible to incorporate virtually all data acquisition functions into a single, small package. Such highly integrated systems have become known as "smart" sensors.

The term "smart" sensor can be defined as a device which contains both sensing and processing functions, within the same enclosure and, perhaps, controls and actuators. The sensing function, as illustrated in Figure 2.1-1, is considered to consist of:

- One or more sensors for measuring physical or chemical phenomena; the ideal sensor being one which has standardized transduction coefficients.
- Signal conditioning to maximize signal-to-noise ratio (e.g., amplifiers, filters, etc.).
- Information specific to the sensor(s) (e.g., type, sensitivity, offset, linearization constants, correction factors, etc.) stored in Programmable Read-only Memory (PROM).

The processing function includes:

- Power supply/control for all components.
- A computational device, typically a microprocessor, with associated memory and peripheral devices (A/D, D/A, digital I/O, etc.) for performing scaling, corrections, and information extraction.
- Communications interfaces for networks, telemetry, control actuators, etc.; digital interfaces providing protocols for communication to, as well as from, the sensor.

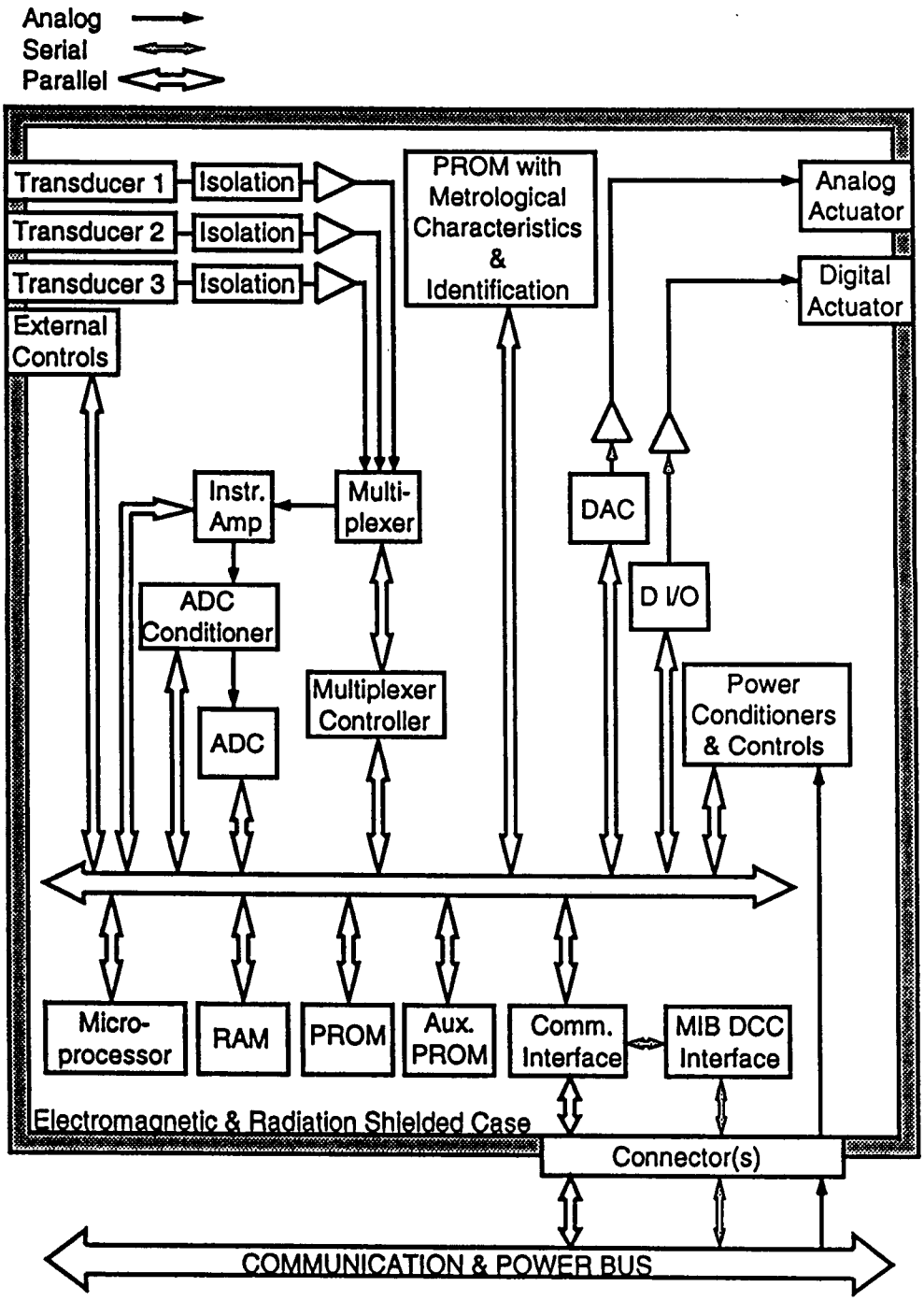


Figure 2.1-1 Generalized Block Diagram of a Smart Sensor

The control portion may include:

- Power amplifiers, switches, etc.
- Electromechanical devices, such as solenoids, solenoid valves, heaters, micromotors, motors, etc.

The software installed in the smart sensor defines the device functionality. Smart sensors are expected to incorporate:

- Self testing of functional sections in addition to processor and memory testing; error messages will be provided (e.g., sensor continuity, power supply voltages, etc.).
- Self-calibration and correction for environmental conditions for which the sensors are not standardized and internally compensated. (e.g., temperature compensation).
- Signal processing and analysis which extracts the desired information from the measured phenomena (e.g., real-time data reduction, sampling, and computation of derived measures).
- Communication with, and control by, the host system in addition to its assigned independent measurement and control function; artificial intelligence (AI) techniques can now be ported into the microprocessor environment and used to selectively determine the configuration of the sensor/recording system or to interpret the output signal (see Smith & Abdelrahman, 1991).
- Programmability; e.g., signal conditioning parameters such as gain and filtering characteristics can be set at run-time; the same sensors can thus be configured for recording a variety of measures, depending on the scientific interest of the moment.
- Reconfigurable outputs to allow a given sensor to interface with a variety of devices implementing subsequent stages of data processing.

2.2 DESCRIPTIONS

Most advanced and future biomedical sensors will differ from today's commercially available sensors in what could be described as packaging.

Decreased size, weight, and power requirements, as well as the need for tethering will improve the usability of the sensors. Present and emerging microprocessor technology requires less power and is smaller than conventional recording systems required to achieve similar functionality. Moreover, since the same sensor can be reprogrammed for various applications, the savings in size, weight, and stowage capacity is multiplied. Smart sensors implemented in a wireless system will minimize the degree to which the human subject needs to be tethered to recording or data processing instrumentation, thereby reducing cabling that can be cumbersome to the subject or a hazard to others in the microgravity environment. At some point, most of today's sensors interface with computers. Thus, adding intelligence to the sensor allows it to assume some of the signal processing functions that would normally be done by the computer. Ever increasing electronic circuit densities and high speed microprocessors have prompted the integration of the sensor (transducer) with preconditioning circuits, analog-to-digital (A/D) conversion circuits, and microprocessors for high speed analysis. With capability for more than just a single channel of one physiological measure, the trend will be to analyze multiple channels simultaneously, preferably with parallel processing architecture. This is a significant step in sophistication, but remains a step short of the, so called, "digital transducers." The output of such transducers is a formatted binary code in lieu of a continuous analog signal proportional to a physical phenomena being observed. The names being used in the sparse literature for this data acquisition technique include "programmable transducer," "smart" sensor or transducer, and "artificial intelligent sensor."

Smart sensors provide a number of advantages and enhancements over existing measurement systems. In addition to the broad advantages of distributed processing and intelligence, other advantages include:

- Improved performance, accuracy and reliability.
- Minimization of setup and calibration time by the user.
- Reduced transducer inventory due to the multifunctionality and adaptability provided by programmability.
- Reduced external central processor workload.
- Reduced communication network traffic.
- Improved operational dependability through self testing and fault isolation.
- Increased compatibility with new object-oriented programming methods.

Improvements in manufacturing processes and quality control have produced remarkable advances in transducer uniformity and made possible interchangeable sensors for many applications, such as the measurement of blood pressure. As a result, the same sensor

package can be "reused" for a variety of applications, thereby reducing the number of elements needing to be maintained for a given set of applications. Uncontrollable, and often very small, intrinsic variations in transducers limit the ultimate accuracy of sensors. These accuracy-limiting variables (zero drift, sensitivity changes) are often functions of environmental conditions such as temperature, pressure, etc. Non-linearity and hysteresis are often functions of the design and materials used. These sources of inaccuracy are particularly troublesome when readings of interest are only a small fraction of the sensor's full-scale value. Self-calibration, self-diagnosis, and capabilities for real-time artifact rejection and correction can produce more usable data and less down time. In addition, by implementing at least the first stage of amplification in proximity to the sensor/transducer, artifacts that are normally introduced by electrode wire movement and some environmental noises will be avoided.

While sensor-to-sensor variability will always exist due to these causes, many of the sources of variability are well-behaved and can be characterized for individual sensors. Correction factors can be derived from the calibration results, which can be used with other measurements to produce more accurate values. This "intelligence" that can be built into smart sensors will enable considerable selectivity in the recording of data. It may no longer be necessary to store large volumes of data that are later discarded. In the past, however, this has been a laborious task for both people and data acquisition systems.

Smart sensors can provide solutions to this problem by virtue of *in-situ* environmental sensing, internal memory, and computational capability. Calibration and correction constants for the device can be stored in its memory as part of the calibration process. During operation, the measured signal is corrected using the stored values and environmental data from the other transducers in the smart sensor before transmission of the signal to the user. Since this is done in the sensor, the correction process does not load the rest of the data acquisition system and, for all intents and purposes, the process is transparent to the user.

2.3 TWO "SMART" SENSOR EXAMPLES

Many of the benefits listed in the previous section are seen in the following cases. The improvements in accuracy provided by smart sensors are illustrated by Table 2.3-1, which is excerpted from Favennec (1987). Table 2.3-1 contains the metrological characteristics of a differential pressure transducer when used as a conventional sensor and when embedded in a smart sensor. When used as part of a smart sensor, the signal is corrected for the temperature and static pressure effects on zero and span as well as those due to sensor non-linearity, hysteresis, and repeatability. Values are listed for both full-scale and for one-sixth full-scale.

Table 2.3-1 Comparison of Conventional And Smart Sensor Accuracy

	Conventional Sensor		Smart Sensor	
Error	%F.S.	%F.S./6	%F.S.	%F.S./6
Non-linearity	0.1	0.1	0.05	0.05
Hysteresis	0.1	0.1	0.02	0.02
Repeatability	0.05	0.05	0.01	0.01
Temperature Effect from 60°C Change				
Zero Drift	0.3	1.8	0.2	0.2
Span	0.3	0.3	0.2	0.2
Static Pressure for 70 Bar Change				
Zero Drift	0.2	0.2	0.1	0.15
Span	0.2	0.2	0.1	0.15
Quadratic Cumulation of Errors	0.53	2.2	0.22	0.26

As can be seen from the table, significant accumulation of error due to temperature and pressure occurs in the conventional sensor, especially when operated well below full-scale. Conversely, the smart sensor configuration, has over twice the full scale accuracy which is only slightly degraded when operated well below full-scale.

This example does not include the effects of ambient noise on accuracy which can occur when low level output sensors are widely separated from their signal conditioning electronics. Smart sensors, with their sensing, signal conditioning, and their processing components properly selected and enclosed within the same well-designed case, can be more immune to ambient electromagnetic noise. While ambient noise can be a problem with digital communication lines, full-duplex transmission with error checking can identify corrupted transmissions and provide for retransmission of data. It should be pointed out that error checking of this type is only effective on those digital signals past the RAM.

Another area where a majority of "smart" sensor benefits can be seen is in the field of bioanalysis. There has been a virtual explosion in world-wide research efforts directed toward bioanalytical sensors. This dramatic increase is evidenced by an annual quintupling

of articles produced in the scientific literature over the last four years, tripling of attendance at research conferences over the last two years, and a steady increase in the number of patents awarded in the field. With all the possible permutations of techniques and materials, and with new approaches occurring every few months, it is difficult to assess the status of the field and even more difficult to predict the eventual success of current investigations. Yet, a few definite trends seem to be apparent. The trends are generally in the areas of sensitivity, sample requirements, miniaturization, and materials.

The number of compounds targeted for sensing applications continues to increase. This trend is generally a byproduct of the availability of immunochemical reactions for a wider variety of biologically important analytes. Conventional electrochemical detection methods were previously restricted to the detection of ionized species occurring naturally or by enzymatic reaction. Understanding of the electrochemical activity of polymer materials and electronic coupling techniques has opened electrochemical techniques to more complex analytes.

The sensitivity and detection limits of bioanalytical sensors continues to improve. This is due in large part to the emergence of specifically synthesized bioactive materials, advances in instrumentation, and adoption of advanced signal processing techniques enabled by microcomputer devices. Bioanalytical sensor research will continue to evolve toward the analysis of more chemically complex samples. In general, the availability of immunochemical reactants, selective biomaterials, and selective membrane technologies will increase the effectiveness of microseparation techniques which will allow the analysis of compounds in unprepared samples such as whole blood and untreated waste water. Bioanalytical sensor systems will move closer to the subject systems and the labor-intensive need for withdrawing samples for analysis will be reduced. At the same time, *in situ* sensors will evolve away from the subject systems producing less invasive devices. In medical sensor systems, required sample volumes will continue to decrease, with accurate results requiring samples on the order of 100 microliters.

Miniaturization and integration of sensor functions will be the focus of intense investigation. However, the addition of subsystems to basic sensors cannot fully compensate for every fundamental problem. The addition of more functional subsystems, material interfaces, and chemical reactants will produce new problems in reproducibility of sensor devices. Design and synthesis of custom bioactive materials is an area which can potentially reduce the number of interfering variables in sensor systems. This effort will be a key factor in the success of any future sensor systems.

Direct spectroscopic techniques might allow significant reductions in functional subsystems. Infrared, near IR, mass spectroscopic, and magnetic resonance techniques will probably supersede more complex approaches to chemical analysis as the range, sensitivity, and instrumentation requirements of these techniques continue to improve.

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3.0 NASA PROGRAM APPLICATIONS AND INTEGRATION ISSUES

3.1 BACKGROUND

NASA's approach to recording biomedical indices in space has thus far been a relatively conventional one of transducing, telemetering, and/or and storing the raw data for later analyses. Traditional electrodes, pressure transducers, temperature sensors, and optical sensors have passively sensed and transduced the signals of interest, with analog amplification and signal conditioning as necessary. In most instrumentation, the amplified signals have been stored along with time stamps on analog tape cartridges or cassettes. On occasion, microprocessor-based systems have been used to digitize and store the raw data in solid-state memory or the analog data have been telemetered to the ground and stored there. In either case, data reduction and analysis have been performed off-line, usually by ground-based personnel, and usually post-flight. This approach has been dictated, of course, by available technology and the constraints of size, weight, and materials imposed by space missions and the microgravity environment.

Discussions with NASA personnel in several research centers suggest that, in contrast to the very sophisticated use of sensors for launch control and monitoring aeronautical variables, there has been little use to-date of advanced sensors in biomedical instrumentation or other scientific research applications. Moreover, there is little indication of such instrumentation being in the "pipeline" of equipment being prepared for future space flights.

One notable exception to this finding is a technology development program called Sensors 2000! (2K!) conceived at the NASA Ames Research Center (ARC). The program was initiated in Fiscal Year '88 with the objective "to design, develop, and evaluate biomedical sensor systems for application to NASA spaceflight programs." (See Hines, 1989) The goal of the program, as stated in its first status report, is "to create a process by which animal (and some human) sensor systems can be efficiently developed and transitioned to applicable NASA flight programs for making high-priority life science measurements (Hines, 1989)." Appendix B contains the Executive Summary from the first year's report. Technical details on the results of this program, to date, have not been widely disseminated.

In order to take full advantage of the smart sensor technologies described in this report, NASA should take a system perspective, complementing the enhanced capabilities of the sensor/transducer with associated instrumentation for monitoring transducer output, intelligent utilization of signals by onboard AI, and archiving derived measures in an efficient manner. The design of the SSF infrastructure remains in flux, being at the mercy of budgetary constraints and engineering differences of opinion. Thus the present situation offers considerable room for innovation. Smart sensor applications would seem to be readily justified since they offer the potential for providing the advantages described above while reducing operating costs.

Sensors which are a component in a fixed, permanently-mounted, single-function instrument having a processor or controller would probably not provide significant advantages if re-designed as smart sensors. In such instances, much of the function of the sensors intelligence would be unprofitably redundant to that of its controlling processor.

The primary application of smart sensors should be in those instruments which make similar measurements but in different contexts. Similarly, they will be useful in those cases where different information is extracted from a given signal depending upon the application.

They will also be useful in those situations where extensive preprocessing of a signal is needed for extraction or compression of information, or in certain instances where information (e.g., an abnormally high heart rate) is transmitted infrequently.

3.2 SMART SENSOR USAGE IN THE BMAC PROGRAM

Using the SSF Biomedical Monitoring and Countermeasures (BMAC) Equipment List as representative of the NASA Life Sciences sensor needs in the future, three types of smart sensors can be identified. These are:

- Smart Electrophysiological Sensors.
- Smart Physical Transducer Sensors.
- Smart Chemically Active Sensors.

3.2.1 Smart Electrophysiological Sensors

The most notable opportunity for smart sensor applications is electrophysiological recording and monitoring. As illustrated in the example used earlier, a smart electrophysiological sensor could be used singly or in combination with others to provide a wide range of experimental and crew monitoring functions.

For instance, the BMAC Equipment List calls out for an Electrocardiographic Recording Device (ECG), an Electroencephalogram (EEG), an Electromyograph Recording Device (EMG), and an Electro-oculograph Recording Device (EOG). The electrophysiological measurements are all basically differential, low-level, AC voltage measurements. They differ only in the type and placement of electrodes, voltage range, and filtering bandwidth.

Since programmability of gain and bandwidth are now standard features of microprocessor devices as is real-time digital signal processing, it is practical to perform all these measurements with a single device.

The smart electrophysiological sensor might consist of a high impedance, AC-coupled differential input amplifier (low frequency cutoff of 0.01 Hz to accommodate diagnostic ECG), followed by one or more program-controlled gain and offset amplifier stages.

Automatic gain adjustment is needed because electrophysiological signals have a relatively wide dynamic range both within and between individuals. Offset control is needed to avoid motion artifacts and electrode potential imbalances from saturating the gain stages of the analog circuits. An antialiasing amplifier couples the analog circuitry to an A/D converter connected to the microcontroller of a digital signal processing circuit. The processor performs digital bandpass filtering appropriate to the measurement being made as instructed by the PMDM.

In addition to the similarities in measurement, electrophysiological signals usually require some form of signal analysis to extract the information of interest; a function best done in the sensor to avoid overloading the communications network. ECG signals may be passed through R-wave detectors for determination of heart rate and dysrhythmias or pattern recognition routines to identify abnormal beats such as pre-ventricular contractions. Pattern recognition routines may also be used to identify epileptiform EEG patterns. The EEG power spectral density coupled with relative wave band power densities are often used for sleep staging. Evoked response studies require time averaging of stimulus synchronized EEGs, followed by delay time analysis to produce very small sets of output data from very large sets of input data. EMGs, especially, are rarely recorded in their raw form; the RMS or integrated power usually being the parameter of interest in myographic studies.

Even in those cases where the original signal is the desired information, the data desired for capture is usually a finite length record selected to contain specific events. This selection can often be most efficiently done by the smart sensor's processor.

All of these signal processing methods are currently performed in small battery-operated instruments. It is, therefore, reasonable to expect that they could be implemented in a smart electrophysiological sensor. The routines could all be stored within the sensor memory or downloaded from the PMDM as needed.

The smart electrophysiological sensor could also be readily adapted to other purposes with the addition of a self-generating transducer. Addition of a microphone would produce an electronic stethoscope for collection and processing of phonocardiographic data. Use of an accelerometer could produce either a Wrist Activity Monitor or Motion Analysis Monitor. Finally, the smart electrophysiological sensor, when used in conjunction with a constant current oscillator, could satisfy the BMAC requirement for a Bioimpedance Analyzer.

3.2.2 Smart Physical Transducer Sensors

Many physiological measurements use sensors, such as thermistors and pressure transducers, which require a well-regulated excitation voltage. Further, the measurements are usually DC and referenced to some baseline level (e.g., ambient pressure). Smart sensors for these variables would differ from the electrophysiological smart sensor by the inclusion

of a regulated power supply for the sensor and a DC-coupled input amplifier. They would also require provision for auto-zeroing, which could require interaction with the user.

Perhaps the greatest distinguishing feature common to most smart sensors is the use of auxiliary sensors to provide compensation for environmental conditions. Most common of these will be a temperature sensor to provide correction of temperature affects on offset and sensitivity. Auxiliary sensors will require their own signal conditioning circuits and require the use of a MUX or multiple A/D converters. This will increase the complexity and space required in the sensor.

The digital processing in smart transducer sensors will differ little from that in the smart electrophysiological sensor. Peak detection (e.g., systolic pressure) and mean values will be the most common types of values extracted. It is expected that smart transducer sensors will commonly be used as part of a control loop. It is reasonable to expect that the controlled device will either be included within the same package as the sensor or in close proximity to it. The smart sensor then will need provision for interfaces between its D/A and digital I/O ports to allow it to perform local control of processes.

Again, using the BMAC equipment list, a basic smart transducer sensor could be used in a variety of applications using different transducers. A smart transducer sensor fitted with a thermistor could be used to satisfy the Telethermometer requirement; supplied with a variety of thermistor probe configurations could make it suitable for measuring oral, rectal, or ear temperature as well as ambient dry air temperature and wet bulb temperatures for humidity determination. As a smart controller, it could become a component of the refrigerated and incubator centrifuges, freeze drier, freezers, incubators, and freezing point osmometer.

It is important to note that, in all of the control applications noted, the smart temperature sensor would perform continuous stand-alone control of the process. In addition, it would have the important task of informing the Payload Manager whenever the process was out of control tolerance which might indicate a system malfunction. The periodic polling of the cognizant PMDM provided by the MIB protocol would help assure that the smart sensor was itself functioning.

A smart transducer sensor fitted with a 0-300 mmHg pressure transducer and temperature sensor could also find application as a monitor/controller in a number of NASA Life Science instruments. Among those in the BMAC list are the Blood Flow and Plethysmography System, Blood Pressure Instrumentation, and Carotid Sinus Baroreceptor Stimulator. It could also be used as the controller for the Lower Body Negative Pressure garment in conjunction with the Blood Pressure Instrumentation to reduce the possibility of syncope.

3.2.3 Smart Chemically Active Sensors

The third type of smart sensor which has some application at this time, but which will become of more importance in the future, is a chemically active sensor. A general trend, established over the past decade, in physiological research has been toward increased emphasis of cellular biology/biochemistry. **As more and more physiological studies are carried out at the cellular level, the importance of advanced microminiature sensors will increase significantly.** "It will be necessary to develop sensors of both physical and chemical variables that can measure concentrations, gradients, forces, displacements, electrical properties, etc. within cells and organelles. New technologies will have to be developed for such sensing systems, and problems of packaging, signal transmission, interaction with biologic host, long-term reliability, reproducibility, and economic manufacturing will have to be worked out" (Neuman, et al., 1989). **This concept is exemplified by the thought process behind the development of new techniques for flow cytometry at NASA/JSC.** Other examples in the BMAC list include pH and ion specific analyzers, pO_2, and pCO_2. In the near future, sensors will become available for measuring glucose and other biochemicals important to biomedical research and health care. Chemically active sensors will require a wide range of DC or AC excitation at different voltages and currents.

While chemically active sensors have many similarities to physical sensors, several characteristics make them generally incompatible with typical transducer instrumentation. They typically have high internal impedance, produce electrode potentials which are sensitive to temperature and ion concentration, and are often highly nonlinear. Smart sensors incorporating chemically active sensors might, therefore, require exceptionally high input impedances and exceptionally stable amplification and signal conditioning. Multiple sensors (e.g., temperature, ion concentration, etc.) will be needed. Finally, many of the sensors will be degradable and, in some cases, single use. This will require provision for periodic replacement of sensing elements. These factors can be expected to result in more complex designs and larger packages than those using electrodes and conventional transducers.

However, given the wide range of chemically active sensors that may be required for future missions, development of a smart chemically active sensor would be warranted as a means of reducing the number of specialized instruments that must be inventoried.

A smart sensor can have its compensation algorithms reprogrammed to fit the specific sensor and can use predictive algorithms, such as are used with pCO_2 electrodes, to reduce measurement time. A smart sensor can also use information (e.g., "Is it blood or water?") input by the user or DMS to select the proper calibration constants.

Chemical sensors also have applications in more global situations within the space program including life support systems monitoring, crew health care, experimental analysis, and bioproduction systems monitoring. Applications in propulsion and power systems will not be considered within the scope of this report. Listed on the following pages are compounds identified by Krug International Life Sciences Division as being probable target analytes for space applications. Table 3.2-1 describes the relative requirements of various sensor systems for the these applications.

Life support systems monitoring will include monitoring of air and water supplies. Typically, these systems will require sensors with very high stability over long periods (months) and very low detection limits for most compounds. The sensor data system will require a relatively high order of redundancy and a relatively complex network of sensors to monitor these systems at several locations along the supply and recycling pathways.

Sensors for crew health care applications will be typified by long shelf lives, short breakout-to-use times, high accuracy, and reliability. Sensor systems for this type of application may benefit from requirements for relatively short term (days) and intermittent use which could allow recalibration.

Sensors for bioproduction monitoring applications could include monitoring and control of hydroponic or bioreactor systems. These sensors would require a moderate term of use (weeks) and accurate results. The nature of these systems would probably allow operators to cross check and verify results without critical damage to the products, so reliability requirements may be reduced with respect to other applications.

In addition to the above specialized applications, sensors in space will be needed to perform typical analytical functions commonly found in chemical laboratories. In general, these applications may use currently available sensor systems with modifications required for the space environment. Systems for routine analysis will require a high degree of variability with respect to the type and complexity of the sample and with respect to the analytes detected.

Table 3.2-1 Qualities of Sensor Systems for Space Applications

	Term	Reliability	Type of use	Sensitivity
Life Support Monitoring	months	very high	continuous	high
Crew Health Care	days	high	intermittent	moderate
Bioproduction Monitoring	weeks	high	continuous	moderate
Analytical Chemistry	hours	high	intermittent	variable

Table 3.2-2 contains lists of applications for chemical and bioanalytical sensors in space and Table 3.2-3 is an applicability chart for the various technologies.

Table 3.2-2 Applications for Analytical Sensors in Space

Life Support Systems Monitoring

Water Quality	
alcohols	0 - 0.5 mg/L
ammonia	0 - 0.5 mg/L
arsenic	0 - 0.001 mg/L
barium	0 - 1.0 mg/L
cadmium	0 - 0.005 mg/L
calcium	0 - 30 mg/L
chlorine	0 - 200 mg/L
chromium	0 - 0.05 mg/L
copper	0 - 1.0 mg/L
cyanide	0 - 0.001 mg/L
iodine	0 - 15 mg/L
iron	0 - 0.3 mg/L
lead	0 - 50 mg/L
magnesium	0 - 0.05 mg/L
mercury	0 - 0.005 mg/L
nickel	0 - 10 mg/L
nitrate	0 - 10 mg/L
phenols	0 - 0.001 mg/L
potassium	0 - 340 mg/L
selenium	0 - 0.01 mg/L
silver	0 - 0.05 mg/L
sulfate	0 - 250 mg/L
sulfide	0 - 0.05 mg/L
zinc	0 - 5.0 mg/L
Air Quality	
carbon dioxide	0 - 5% v/v
carbon monoxide	5 - 200 ppm
oxygen	15 - 25% v/v

Blood Electrolytes	
chloride	65 - 140 mmol/L
calcium	10 - 120 mg/L
magnesium	0.05 - 13 mg/L
phosphorous	0.5 - 13 mg/L
potassium	1 - 11 mmol/L
sodium	95 - 300 mmol/L
Blood Gases	
pH	7.0 - 7.8
carbon dioxide	20 - 80 mmHg
oxygen	50 - 250 mmHg
Blood Enzymes	
aspartate aminotransferase	4 - 950 U/L
alanine aminotransferase	3 - 950 U/L
amylase	5 - 900 U/L
creatine kinase	20 - 1600 U/L
creatine kinase MB isoenzyme	1 - 300 U/L
lactate dehydrogenase	100 - 1750 U/L
alkaline phosphatase	15 - 1500 u/L
Blood Metabolites	
albumin	1 - 6 g/dL
bilirubin, direct	0.1 - 20 mg/dL
bilirubin, total	0.1 - 20 mg/dL
cholesterol	35 - 425 mg/dL
HDL cholesterol	1 - 110 mg/dL
creatinine	0.1 - 17 mg/dL
glucose	20 - 450 mg/dL
lactic acid	0.5 - 2 mmol/L
total protein	2 - 11 g/dL
urea nitrogen	1 - 110 mg/dL
uric acid	0.3 - 16 mg/dL
triglycerides	15 - 400 mg/dL
Urine Analysis	
creatinine	0.1 - 630 mg/dL
glucose	20 - 650 mg/dL
potassium	2.5 - 1.74 mmol/L
sodium	1 - 50 mmol/L
urea nitrogen	1 - 200 mg/dL
calcium	1 - 18 mg/dL
total protein	2 - 11 g/dL
phosphorous	2.5 - 10 mg/dL
amylase	5 - 900 U/L

Table 3.2-3 Applicability of Bioanalytical Sensor Technologies to Space Station Freedom

Technology	Expected Applicability	Availability	Technical Hurdles
Silicon microsensors	high	5 - 10 years	reagent leakage, frequent recalibration
IR & NIR direct analysis	high	5 - 7 years	detection limits, sensitivity to monatomic compounds
Ex situ ISE analyzers	moderate	available	frequent recalibration, sample transport
Ex situ optical analyzers	moderate	available	sample preparation, transport & separation
In situ ISE's	moderate	available	frequent recalibration, chemical interferences
In situ optrode sensors	high	some available	stable indicators for trace metals, CO, detection
Electroacoustic devices	moderate	3 - 7 years	poor sensitivity to small molecular weights
Mass spectroscopy	high	7 - 15 years	design of systems specifically for space environment
Thermoelectric sensors	moderate	5 - 10 years	poor detection limits, limited analyte potential

3.3 WIRELESS SENSOR AND CONTROL PACKAGES

Other qualitative enhancements would be provided by creating completely wireless physiological monitoring systems. **In such systems the circuitry for amplification, signal conditioning, and telemetry would be integrated into each individual sensor, with an array of interchangeable sensors telemetering data to a transceiver on the body or directly to a base station.** Such a system can be seen as combining some features of the present NASA portable amplifier/signal conditioning units, the CDUSS, and helmet-mounted electrode/amplifier units. The advantages of such an approach would likely be decreased susceptibility to environmentally induced artifacts and less cumbersome cabling such as that presently used to tether the subject to body-worn or off-body amplifiers, signal conditioners, or data recording instrumentation.

The feasibility of a harness-less, multi-sensor telemetry system has been proven on a recent Navy project. The initial Navy objective was the monitoring four temperatures (one "core", three skin) concurrently from divers by transmitting these combined data to a base station located nearby. An accompanying objective was to have the individual temperature sensors, whether core or skin, coupled to their own short range telemetry transmitters with their combined signals being received by a body-worn, scanning receiver/retransmitter (transceiver). This would speed sensor installation and reduce encumbrance of personnel being instrumented. A further objective was operation of such a system in a metallic-walled enclosure, such as a boiler room. A complete system with multiple temperature transmitters was delivered and can be used to evaluate advanced software for evaluation of harness-less capabilities. Preliminary tests indicated that temperature sensors on the ankles, wrist, rectum, thorax, and axillae, as well as transmissions from an ingestible pill can be received by the CDUSS transceiver. Unfortunately, advanced microprocessor capabilities (i.e., with more EEPROM capacity) were not available in a timely fashion for this project so the unit could not be tested in the mode of receiving multiple sensor transmissions simultaneously. An appropriate chip has since become available.

New designs in commercially available patient monitoring systems are emphasizing complete two-way data communications technology that might have future applications on the SSF. For example, SpaceLabs (see Appendix D) has recently introduced a system that makes it possible to monitor patients in **different locations at the same time** and claims to **integrate more information with less interaction** to enhance **staff productivity. These goals are precisely those that one might imagine for remote control and supervision of life sciences experiments on SSF where it might not be desirable to have full-time personal supervision for each experiment while it is in progress.** In the SpaceLabs system, the "Cordless Remote Keypad" lets the clinician remotely control bedside monitors from up to 20 feet away. Infrared telemetry is used to "suspend or adjust waveform size, initiate recordings, or perform any other monitor function." **Perhaps such a controller could eventually be built into a wristwatch-type device for SSF crewmembers.**

3.4 SYSTEMS INTEGRATION ISSUES

In environments like SSF, for which the data processing infrastructure is still being designed, it is advisable to take a systems perspective in considering the implementation of smart sensor technology. While smart sensors will typically be capable of stand-alone operation and decision making, the information they produce will need to be integrated with other information for assessment of the system being monitored and controlled. This requires communication with other devices, usually via a digital communication network. Ideally, I/O standards should be established so that sensor systems can be interchangeably plugged into the infrastructure, allowing data from different sensor/transducers to be networked, made available to common sensor fusion algorithms, and archived efficiently. The resulting data and information collected by NASA Life Sciences smart sensors will typically be input through SSF payloads, facilities, and EVA systems; it is even conceivable that a smart sensor might be designated as a payload.

The latest restructuring of SSF proposes a payload network architecture shown in Figure 3.4-1 (Whitelaw, 1991). A schematic of the SSF Data Management System (DMS) physical configuration and hookup of a typical payload is shown in Figure 3.4-2 (Foster, 1991). Each payload rack will be equipped with an optical disk drive for local storage of information and a Payload Multiplexer-Demultiplexer (PMDM) for providing communication connection with the payload network ring. The PMDM will contain all core MDM functions with the exception of handling the Local Buss. This will include all MDM services software, all sensor and effector (analog, discrete, and solenoid drivers) I/O, and serial/parallel/digital communications. It will also accommodate unique user I/O buses and bus cards, and user payload application software. It will provide an interface to the DMS network for high throughput communications to the Payload Manager resident in SDP-7 and for telemetry via the IRGW interface to the Communications and Tracking System.

The command and control of payloads will be distributed between the Data Management System (DMS), the Payload Manager (PM), and the Payload. The DMS will provide generic services to applications in the SDP and MDM for communications, establish mode-dependent telemetry, and inhibit classes of actions that are incompatible with the station mode. It will notify the Payload Manager of station mode changes. The PM will acknowledge the changes in station mode, and either notify the payloads to change their mode-dependent operation and telemetry or act on behalf of the payloads when the PM is collecting telemetry.

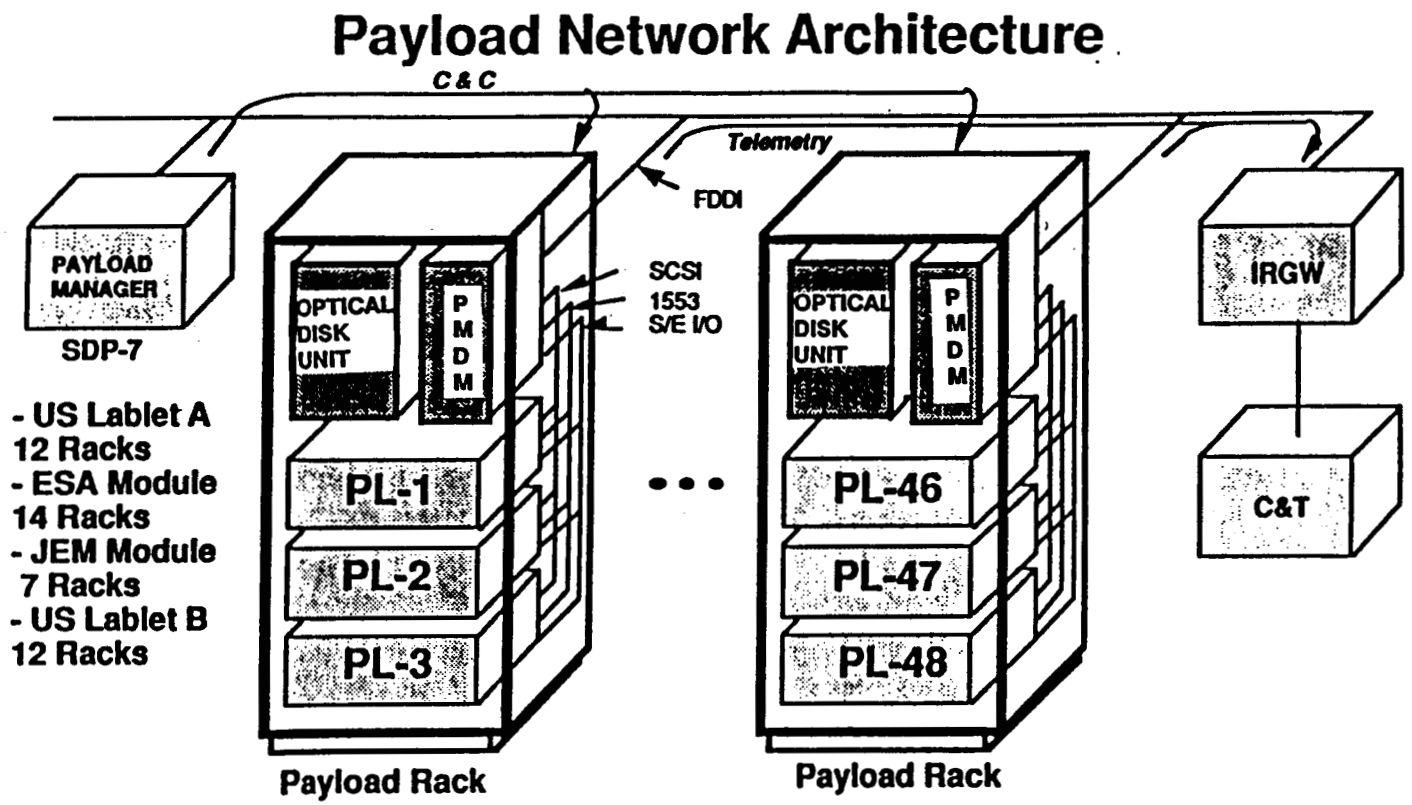

Figure 3.4-1 SSF Payload Network Architecture

- PMDM - Proposed Payload MDM with FDDI and SCSI Interfaces
- OPTICAL DISK UNIT - Proposed WORM Optical Disk Unit with SCSI, 1553, and Serial Interfaces

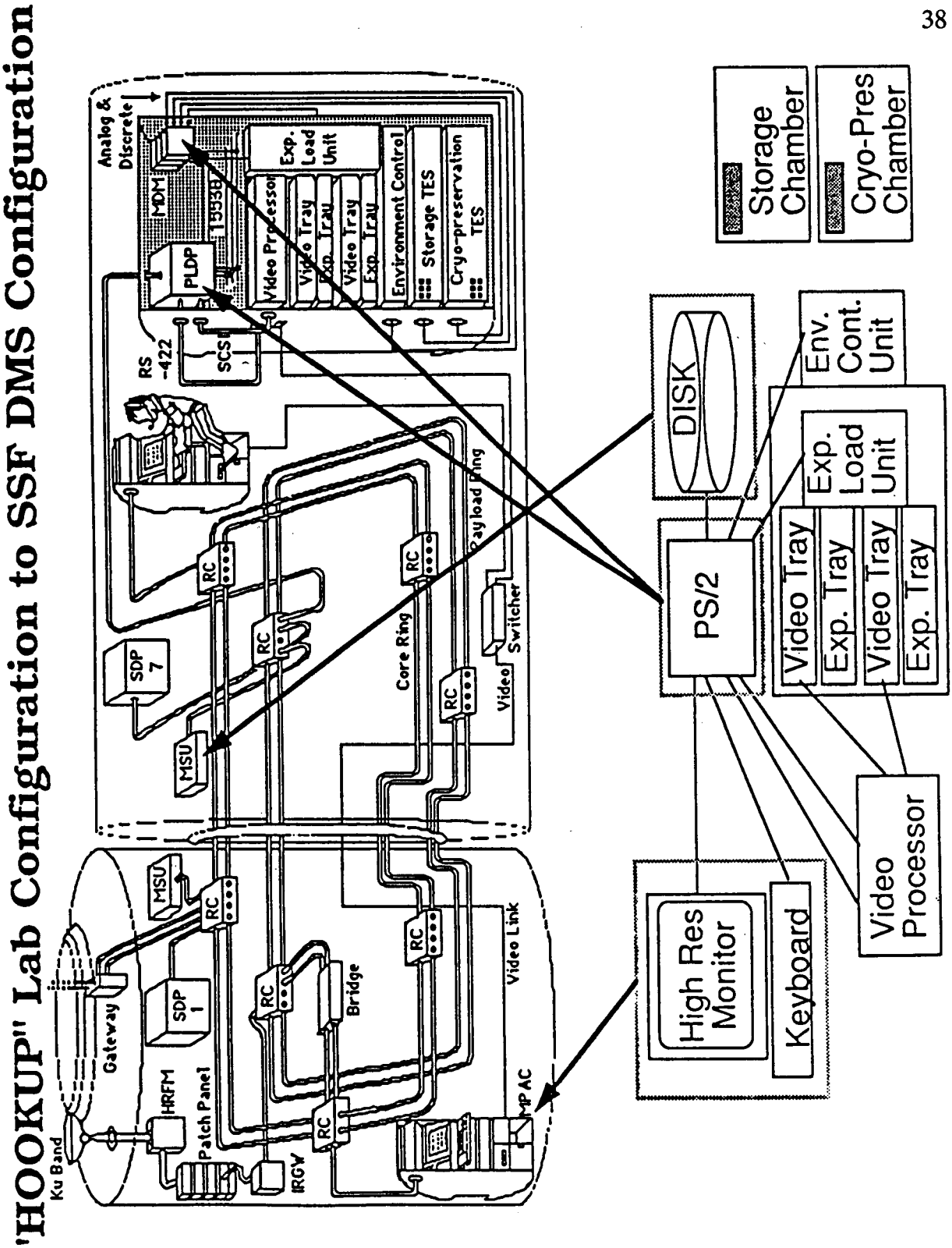

Figure 3.4-2 SSF DMS Architecture and Typical Payload Hookup Schematic

The payload, acting through the PMDM, will respond to commands from the PM, crew or ground as well as perform any independent activities assigned to it which do not require communications. It will provide data as needed to satisfy requests from the PM, crew, and ground. The payload will pre-define its data, commands, and operations which may have potential conflicts with station modes. Although payloads may not all be mode-dependent, many will be required to pre-define their mode-dependent telemetry requirements. Finally, when empowered by the payload manager, the payload will monitor and control its activity locally although making use of the PM and DMS services if coordination with other activities are needed.

As mentioned earlier, smart sensors will either be included within a payload or possibly be considered as a payload themselves. As such, smart sensors will require design features which make them compatible with the SSF Payload Network Architecture. Highly integrated smart sensors will require only power and digital communication interfaces to the payload. Since power regulation will probably be included within the sensor, power interfacing requirements should be minimal. The principal design requirement, then, will be associated with the communication interface.

Fortunately, the proposed PMDM includes significant flexibility in its payload interfaces. Most notable of the available interfaces are 16 channels of serial data ports and accommodation of significant user application software. These resources make possible the use of the P1073 series Medical Information Bus (MIB) standards being developed by the IEEE. This standard, although designed for interconnecting medical equipment in hospitals, should satisfy the requirements for interconnecting NASA Life Sciences smart sensors with the SSF DMS. In addition, it should satisfy the unique requirements for life science data documentation and management.

3.4.1 Medical Information Bus (MIB)

The Medical Information Bus (MIB) is proposed by the Institute of Electrical and Electronics Engineers (IEEE) as an international standard for a local area network (LAN). The proposed IEEE P1073 MIB standard describes a layered network architecture that is compatible with the International Standards Organization (ISO) Open Systems Interconnection specifications. It is designed for use in automating information recording, monitoring changes in system status, and automating control of devices within the system. Further, the MIB design provides for the specific needs of systems which monitor, record, and control biomedical information and control systems.

The specific objectives are:

- Enable vendor-independent interfacing of host computers to medical devices in hospitals.
- Ensure suitability for acute and continuous patient care monitoring.
- Provide high transmission accuracy and low network downtime.
- Accommodate frequent equipment changes and reconfigurations.
- Simplify user interfaces.
- Accommodate a wide range of host computer system topologies and complexities
- Operate cost effectively.

Briefly, the MIB is a star-connected, local-area network having three logical nodes as shown in Figure 3.4-3. Individual bedside devices communicate to the network through a Device Communications Controller (DCC). The DCC may either be integral to each device or may be a separate device to which the bedside device attaches. The individual DCCs in an area connect to a Bedside Communications Controller (BCC).

The BCC, as the primary node, coordinates and collects data produced by the bedside devices with each device being polled at least once, but no more often than 10 times per second. The BCC connects to a host computer via a host interface node; the host interface node is a software interface that controls the collection of data by the host computer over a conventional local-area network (LAN).

Physical connection between the DCC and the BCC consists of either an MIB-specified cable or data generators or receivers conforming to serial data transmission specifications. The MIB cable provides three, foil-shielded pairs of conductors with shield drain: one pair for time synchronization, one for serial data transmission (RS485), and one for power for the DCC. For safety reasons, the MIB also specifies a unique connector that is incompatible with all existing connector systems.

The unique feature of the MIB that differentiates it from other local area networks is that it provides automatic system reconfiguration through automatic detection of device connections, identification of the devices, and establishment of communication with added devices. The only user interactions in this process are to plug the device into the network and to turn on the device.

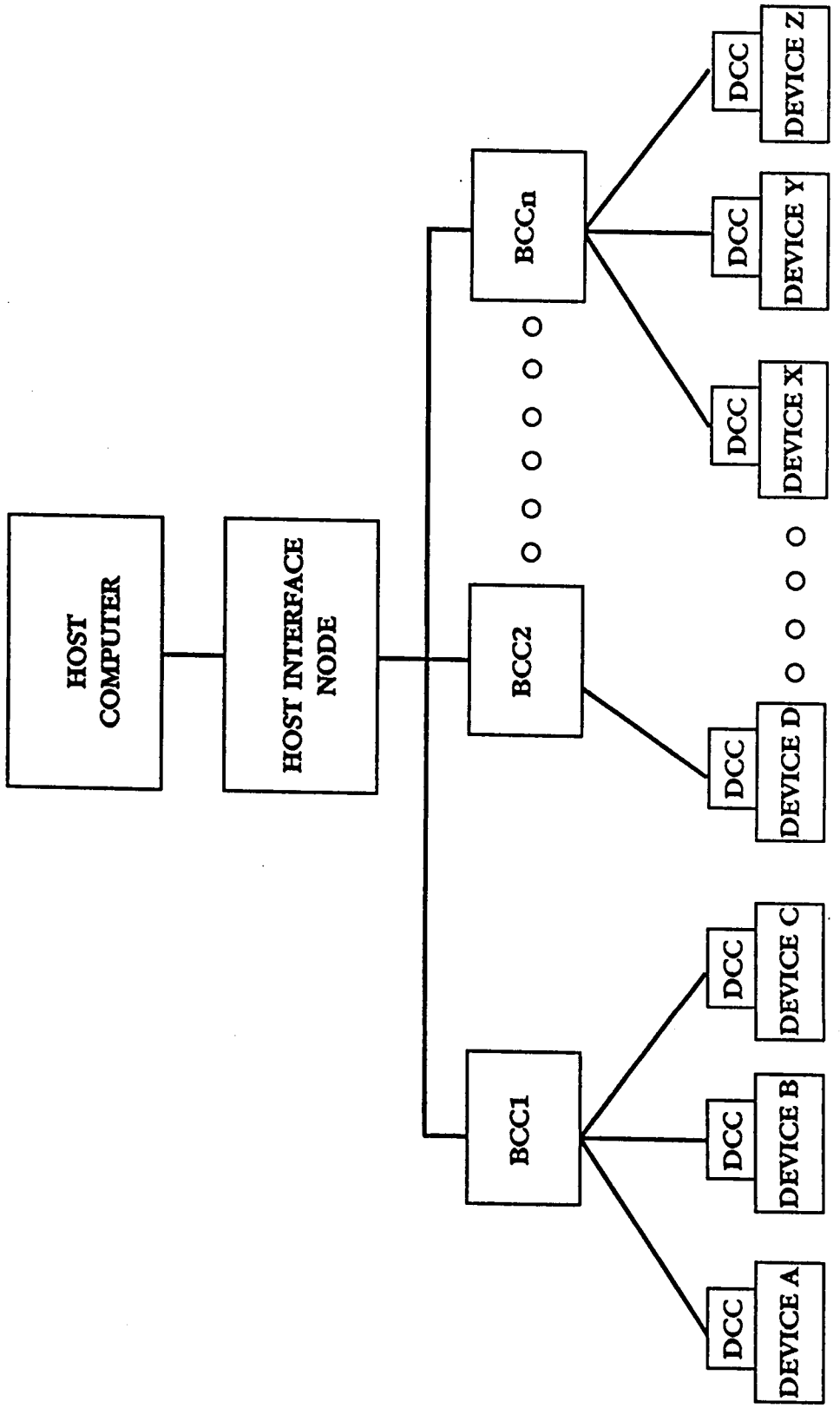

Figure 3.4-3 IEEE Medical Information Network Nodes Structure

A second attribute of specification P1073 is that it provides for measuring the data load that a given bedside will place on the network. The amount of data each device places on the network is expressed in terms of "MIB loading units." The processing capacity of the BCC's and host computer are likewise expressed in MIB loading units. As devices are added or removed from the network, the total loading units can be monitored and compared against the loading unit processing capacity available to the system. This has utility in providing the Payload Manager with network and telemetry communication requirements.

The third attribute of the MIB that recommends its use for NASA Life Sciences devices and smart sensor systems is its specification of a Medical Device Data Language (MDDL). The MDDL provides a dictionary of acceptable words (terminal symbols) consistent with biomedical monitoring and control, along with a set of rules (grammar) for combining words and data (syntax) into understandable sentences (terminal symbol strings). These provide a common language for communication between devices. An example of a communication between a bedside monitor and a central nurses station using an MDDL could be the terminal symbol string (sentence):

Heart Rate = 150 beats per minute.

The terminal symbols (words) "Heart", "Rate", "beats", "per", and "minute" are all included within the MDDL dictionary.

The proposed MDDL goes far beyond this simple example in providing a communication language capable of transmitting data as well as instrument identification, status, and control information. Further, it includes the capability of communicating the context in which the data is collected; contextual information might include the measurement units, site of collection, method of collection, and the time of measurement.

The inherent flexibility of smart sensors in space will demand that the context of their use be included in their reporting of information. Consider, for example, a hypothetical smart electrophysiology sensor. Several existing ambulatory ECG monitors could be adapted and expanded to produce such a device. This device would be capable of monitoring and processing any one of a number of signals, such as ECG, EEG, EOG, and EMG. **Programmable analog and digital gain, bandwidth and electrode summation methods would adapt the basic differential voltmeter to the parameter being measured.** The packaging of the sensor would be such that it could be installed in space suits, worn by free ranging astronauts inside the vehicle, or connected to a rack mounted experiment package.

A few of the ways this smart physiological sensor might be used singly or in multiples in a mission include:

- Monitoring pilot EOG and forearm EMG during launch and landing; reporting eye blink intervals in seconds; and integrated EMG activity for man-machine interface studies.
- Continuous monitoring of ECG in two astronauts following different microgravity countermeasure regimens with storage of mean and standard deviation of heart rate along with diagnosis and examples of ectopic beats.
- Heart rate monitoring during EVA with reporting of heart rate upon command, alarm messages when heart rate exceeds set limit, and storage of heart rate profile for the entire EVA.
- Continuous monitoring and reporting of heart rate to astronauts for target heart rate regulated exercise.
- Monitoring EOG for REM and sleep ECG power spectral density for later transmission to the ground for sleep stage analysis.

Similar measurements (e.g., heart rate, eye-blink rate, REM frequency) are obtained in these applications, but, the actual values are meaningless unless they are identified and other information is reported. For example, the number "150" would be meaningless unless its is known to be heart, rather than eye-blink, rate with units of bpm. Similarly, the significance of a heart rate of 150 bpm is decidedly different in a novice astronaut at launch, one engaged in an exercise stress test evaluation, or one in the process of dislodging a jammed payload from the Shuttle bay. Also, the quality of measurement is dependent upon the conditions of measurement. ECG bandwidths, for instance, are different depending upon whether one is making diagnostic recordings, doing ambulatory monitoring, or simply monitoring heart rate under strenuous exercise. Without bandwidth information, later interpretation of ectopic beats would be difficult if not impossible.

The MDDL, being specifically formulated for biomedical measurements, is ready made to deal with the problem of measurement context. It includes both medical and physical vocabularies that make near natural language declaration of information context possible.

Applying the MIB to smart sensors for use in NASA Life Sciences has a number of advantages. While only a draft standard, MIB provides a communication architecture consistent with the SSF DMS while providing specificity for distributed processing and management of life science data. The standard development is being supported by National Institute of Standards and Technology (NIST), major manufacturers, the major biomedical engineering societies and other international organizations; thus, it can be expected to reach fruition.

3.4.2 Smart Sensor/MIB/SSF Relationships

Smart sensors may either be stand-alone devices or an extension of another device. The choice of configuration will depend upon the function of the sensor and the memory and interface hardware that can be installed in the available sensor volume. **Regardless of the configuration used, smart sensors should be designed for compatibility with the RS485 interface and be provided with software providing DCC functionality. This will provide connectivity with either the PMDM or an intermediate device.**

In transferring the MIB concept to an SSF DMS architecture, "bedside" can be synonymous with such words as "payload," "space suit," "experiment," "facility," and "rack." Based upon the current definition of the Payload Network Architecture, the PMDM appears to be a logical choice to provide the MIB BCC function. In general, life science instrumentation and control is associated with one of these physical locations. Even ambulatory recording devices will transmit data to a central receiver or be periodically connected to a specific facility communication port for downloading of stored data.

In some cases (e.g., experiments and facilities) a single payload may use a large number of devices or perform continuous measurement and control requiring little interaction with the SSF network. In such instances, the BCC function could be included within a payload. However, the payload would still need to have the DCC functionality so that it is seen as a device by the PMDM. The host interface node function of the MIB is consistent with the role of the SSF Payload Manager.

3.5 HUMAN FACTORS ISSUES

Maintaining a systems perspective is crucial when designing smart sensors which will provide user-compatible data. The scientific personnel on SSF missions will be faced with the requirement to monitor and control numerous ongoing experiments, including those outside their own areas of expertise. Because of these supervisory and process control requirements, there will be some of the same cognitive demands, and dangers of information overload, on these personnel as on the crewmembers tasked with flight operations and maintaining safety. Moreover, the crews may be made up of multi-national investigators, raising issues of wording and terminology in the use of equipment labels, displays, and documentation. The onboard research instrumentation may vary to some extent from flight to flight and will certainly be supplied by various vendors. Even if each component of this system is well designed from a human factors standpoint, incompatibilities may arise as equipment and operating procedures are integrated into a common workplace.

It may be possible to mitigate some of these information processing demands with smart sensor technology. Derived measures and interpretive reports can be displayed to the scientific supervisory personnel rather than, or in addition to, raw data. Modern display technology encourages extensive use of graphics, icons, and symbology. However, the full

potential of this environment will be realized only if sound human factors engineering principles are adhered to in providing means for the crewmembers to reprogram sensors, monitor ongoing experiments, and interpret the data being produced.

To some extent, the human factors design issues raised here are shared with numerous other man-machine system environments. These issues concern such matters as:

- Display of system status so that it can be interpreted "at a glance."
- Input devices that promote efficient data entry and control actuations.
- Workstations designed for both operator comfort and operational efficiency.
- Standards or conventions in the use of color, labeling, terminology, and control-display layouts.

Fortunately, there is industry guidance available on these topics, some of it developed by NASA specifically for spaceflight.

Operating in the microgravity environment poses special challenges. New technology (not only hardware but also software environments) has become available so quickly, that the appropriate research into preferred design practices and optimal implementation procedures have not kept up. Research into closed-loop implementations of adaptive automation are a good example of this. Voice interfaces are another example. To some extent, the needed guidance pertaining to user interactions with multi-tasking software environments, graphical user interfaces, pull-down and pop-up menus, and on-line help for the user, are only now being developed. Much more remains to be learned about human performance in microgravity, including issues of optimizing cooperative team performance, minimizing the effects of fatigue and space motion sickness, and individualizing work-rest cycles. Thus, it will be important in considering the development of smart sensor technology to focus not only on the design of the hardware but on the user-system interface with machine intelligence.

3.5.1 Human Factors Guidelines and Standards

The need to delineate human factors design guidelines has been recognized by a number of groups and institutions, spanning a wide range of man-machine task environments. Guidelines for military systems are set forth in Mil-Standard 1472D (1986) and for the nuclear power industry in NUREG-0700 (1981). The Association for the Advancement of Medical Instrumentation (AAMI) has recently published human factors guidelines and preferred practices for medical device manufacturers (1988). Some organizations such as SMART HOUSE have seen fit to customize industry guidance for their own applications (1988). One of the more ambitious efforts of this sort has been conducted by NASA in compiling Man-System Integration Standards (NASA STD-3000) specifically for spaceflight and microgravity environments.

Other recent publications have nicely summarized current issues and knowledge in the human factors of software design (Anderson & Olson, 1985), human performance measurement (Boff, et al., 1986; Boff & Lincoln, 1988; Elkind, et al., 1989), human-computer interaction (Card, et al., 1983; Helander, 1988; Shneiderman, 1987), and supervisory control (Sheridan & Hennessy, 1984). Smith & Mosier (1986) have pulled together a wealth of information and practical guidance on the design of user-system interfaces, with an emphasis on software. The Air Force now maintains the Crew System Ergonomics Information Analysis Center, an extensive data-base and bibliographic retrieval service.

Despite these laudable efforts to establish human factors guidelines and standards, one striking aspect of this literature is that there are few absolutes. The choice of optimal screen design, input device, user-system interaction protocol, and error recovery scheme is dependent on the task at hand. This is not to say that optimal design is impossible or that useful guidance is not available. It is to say that a thorough task analysis should be the first step in deciding which parts of the available research literature and design guidelines are pertinent. Then, user needs and wishes should be taken into account from the earliest stages of a design and throughout the life-cycle. Techniques such as modeling, simulation, and prototyping should be utilized to minimize the possibility of non-optimal end products or costly changes in the final stages of test and evaluation.

3.5.2 User-System Interfaces and Input Devices

Recent developments in graphical user interfaces, "point and shoot" input devices, voice input technology, and integrated displays of system status offer means of meeting some of the challenges described above. In fact, the same microprocessor and AI technology that allows the envisioned enhancements in sensor technology can also be applied to the data processing and interpretation of multi-sensor output. Approaches such as automated pattern recognition, adaptive neural networks, simulation, and analysis by synthesis, offer the prospects of considerably automating the process control of scientific experiments and analysis of the resulting data. Presenting these results to the human supervisor in ways that can be more readily encoded and understood is always a challenge. A brief overview of some of the emerging technologies for enhancing man-machine interactions relevant to smart sensor systems is provided in the following paragraphs.

Graphical User Interfaces

A graphical interface (as opposed to a text-oriented approach) offers advantages in many applications (Brems & Whitten, 1987). The advantages can be visual (less fatiguing), motor (speeded responses because choices are more self-evident), and learning (easier to learn) (Shneiderman, 1987). Much research has been done on this topic, starting with the Xerox Star system. There is still significant work being done on windowing displays, pull-down menus, and iconic symbology (see Davies, et al., 1985; Tullis, 1986). Most of

this research has not yet been pulled together into industry guidelines, but considerable guidance could be gleaned from this literature for a given application (see Card, et al., 1983; Elkind, et al., 1989; Helander, 1988; Shneiderman, 1987; Smith & Mosier, 1986).

Extensive discussion can now be found in the literature about User Interface Management Systems (UIMS) (e.g., Olsen, et al., 1984; Foley, et al., 1988). The idea here is to build the interface separately from the application software and, in fact, make the application somewhat transparent to the user. This provides for interchangeability and integration of applications and promotes ease of learning and use by the user. Some of the UIMS approaches are based on formal logical descriptions of the user-system interface and derived models (Foley, et al., 1989; Hurley & Siebert, 1989; Kieras & Polson, 1985).

Point and Shoot Devices

Coupled with the development of graphical user interfaces has been the development of innovative means to "point-and-shoot," rather than typing at a keyboard or keypad. The choice of input device depends on the nature of the material being input or chosen (see Shneiderman, 1987). Some well-known pointing devices are:

- Mouse.
- Light Pen.
- Touch Screen.
- Trackball.
- Joystick.

Others are more developmental:

- Data Glove -- a glove instrumented with sensors to determine the articulation of the arm/hand/fingers and their position in 3-D space, coupled with a display which gives the user the sensation of control by moving his hand in space.
- Gestural Interfaces (e.g., Rhyne & Wolf, 1987) -- the idea here is to use naturalistic gestures (entered via touch screen, digitizing tablet, etc.) to accomplish certain operations. For example, one might delete text by marking it out in word processing application, rather than moving the cursor to the beginning of the text, making an input, moving to the end and making another input.
- "Look and Shoot" Interfaces -- this involves the use of an oculometer to monitor the direction of gaze. The operator would simply look at the item to be selected and press a single button (as with a mouse, but without having to first position the cursor manually). Oculometers that measure the direction of an operator's visual siting have gotten very precise, and are getting cheaper (although still somewhat costly). The Air Force has done some research on this type of interface for aircraft cockpits (Calhoun, et al., 1984). There are prosthetic devices for quadraplegics that allow the person to "type" by looking

at the desired item and then blinking the eyes to activate it (see Beringer & Scott, 1985).

Voice Interfaces

The possibility of using automated voice recognition to support an interface in which the operator can control the system by simply speaking has been a long-time dream of designers in many man-machine environments. McCauley (1984) provides a useful, if somewhat dated, review of the human factors concerns in this developing technology. Speaker-dependent systems with limited vocabularies are now available and the research thrust is to increase the vocabulary and gain speaker independence. Speech recognition algorithms are being developed to include the ability to recognize a given speaker when the frequency make-up of his voice is altered by stress. The prospects of implementing a hands-off control mode in the SSF, even as a supplement to more conventional manual control, is particularly attractive given the biomechanical difficulties of operating in microgravity. The fact that personnel are likely to be instrumented with microphones anyway, and the advantages that such an approach would provide in reducing operator workload indicate that voice-activated interfaces might be viable.

3.5.3 Display Technology for Monitoring

Complementary to the above technology for user input are a number of new approaches for displaying information in a highly integrated, user-compatible manner. These display modes lend themselves to systems with smart sensor capabilities because they are themselves dependent on highly processed data and artificial intelligence.

Hierarchical Display Systems for Monitoring System Status

The need for a human operator to get a continually updated display of system status is shared by many complex systems and will certainly reach an acute status on SSF. Given that an operator frequently has other activities that time-share with this system monitoring function, it is best to provide at least a high-level overview of the system in a way that conveys system status "at a glance." More detailed information should then be available upon request to allow the operator to diagnose and troubleshoot system problems. Networks of smart sensors backed by interpretive artificial intelligence offers the prospect of providing such a hierarchy of information displays.

The particular display format which best conveys the necessary information may be somewhat system specific. Polar graphics, in which system parameters are displayed at the nodes of a geometrical figure, are particularly interesting. When some system parameter goes awry, the deviation is seen as a distortion of the figure, a change which can be picked up "at a glance." Deviation bar charts offer some of the same advantages. Of course, such overview displays must be augmented with more detailed information about parameters

within a given system or subsystem, probably down to the raw data coming from individual sensors. However, this more detailed information need only be available to the operator. Continuously scanning through such detailed information in order to determine system status is unnecessarily burdensome.

Guidance can be gleaned from the nuclear power industry which, since the Three Mile Island accident, has mandated that all plants implement a Safety Parameter Display System (SPDS). Most plants have chosen to implement their SPDS as a CRT-based hierarchical display system, with considerable use of multi-sensor inputs and error-checking algorithms. A few plants are now supplementing this system with artificial intelligence that provides interpretive input to aid the operator in quickly diagnosing or mitigating plant problems. This application seems completely analogous to the likely requirements for process control of scientific experiments on the SSF.

Integrated Displays

A major thrust in aerospace display design is in the implementation of integrated displays; i.e., displays which efficiently convey numerous types of sensor information in one picture, often with various measures contributing to a given display element. Heads-up displays projected on the windscreen and helmet-mounted displays projected on the visor are two manifestations of this technology. The goal, of course, is to allow the pilot to take in information without looking down at the instrument panel. One telepresence system that links a remotely located human operator with a robotic device operating in hazardous environments (e.g., a system developed by the Naval Ocean Systems Center) uses a helmet with two miniature video cameras displaying the remote scene transmitted from two cameras on the robot (in order to create a 3-D view).

Such display approaches may be of interest in considering the workstations or helmet/suit designs to be used on the SSF. However, it should be noted that these displays have been plagued with problems related to visual fatigue and eye strain, given the distance at which the eyes must accommodate (e.g., see Norman & Ehrlich, 1986). Aircraft heads-up displays (HUDs) are still controversial and have been implicated in several military plane crashes. The optics are collimated to allow the eyes to focus at optical infinity, but objective measurements suggest that people using HUDs accommodate at intermediate distances, causing them to misjudge the distance of objects (and the ground) in their physical environment (Iavecchia, et al., 1988).

Data Fusion

Data fusion describes the machine intelligence that is required to automate interpretation of multi-sensor data and to present a potentially overburdened operator with only that information which he needs at a given time. This field is now exploiting the self-learning features of neural network approaches to signal extraction and signal recognition. It is beginning to combine such signal processing techniques with more conventional rulebased

expert systems (see Johns Hopkins Applied Physics Lab, 1991). Some of the severe real-time demands of fighter aircraft environments that are driving this work may not be present in the SSF environment. The processing of smart sensor data and display technologies that emerge from this area of inquiry over the next few years will be well worth noting during the design of user-system interfaces for SSF.

3.5.4 Closed-Loop Systems Applications

In addition to the quantitative enhancements that smart sensors would provide, it should be emphasized that they may enable qualitatively new possibilities in the study of dynamic man-machine interactions. **For example, when embodied with capabilities for real-time data processing and artificial intelligence (AI), this technology will support studies of task allocations between man and machine and enable studies of human supervisory control of semi-automated systems--a topic of vital practical importance to the scientists on SSF.** Physiological measures, in that they offer a relatively objective, continuous, and non-intrusive means of determining operator functional status, can play an important role in determining the decision rules by which automated aids are activated or deactivated, in order to cooperatively maintain the human in the loop at an appropriate level of vigilance without cognitive overload. The envisioned systems would be closed-loop, since measures of operator status and system performance would be fed back onto the subsystems controlling automated aids or automated subsystems, thereby determining when this automation is invoked.

Such adaptive aiding is now being designed into the Advanced Tactical Fighter and other military aircraft and being considered in other environments such as air traffic control and supervisory control of telecommunication systems. For space missions, this concept may be useful, not only for crew support of in-flight operations and for control of robotics, but also for onboard scientific personnel's process control of ongoing experiments. Such adaptive automation, let alone the considerable research required to effectively implement it, would not be possible without the real-time processing that smart sensors can provide.

3.6 IMPEDIMENTS TO SMART SENSOR DEVELOPMENT

The component technology for implementing the sort of smart sensor technology that is envisioned here for biomedical applications is, by and large, available. The primary impediment to moving forward with the required R&D is likely to be budgetary. **Custom integrated/hybrid circuitry of the sort required will be expensive to prototype, perhaps prohibitively so if applications of the resulting instrumentation are limited to the SSF or to manned space missions. The key to fostering the needed development is likely to be in finding commercial applications for the same instrumentation in clinical medicine or operational environments.** This should not be difficult, in that there would seem to be numerous uses for such instrumentation for ambulatory monitoring of patients, monitoring

personnel working in hazardous environments (military, nuclear power plants, underwater, fire-fighting) to ensure their safety, and for tracking and improving athletic performance.

It should also be noted that, while the present charge was to think in terms of applications of smart sensors for facilitating scientific research in space, many of the biomedical applications that meet this qualification are also of potential value for monitoring crew safety. While fulfilling the requirements for safety-critical instrumentation would further encumber smart sensor devices with many additional specifications and reliability requirements, it may also help justify their development. In other words, development efforts that would be difficult to justify as cost effective for scientific research support might be more readily accepted if their possible future contributions to flight safety and crew welfare were recognized.

Theoretically, therefore, the concepts of smart sensors can be applied to almost any NASA Life Science sensing function. However, certain considerations should be made before committing to this approach. The attributes normally associated with smart sensors can only be achieved with considerable study of the total system. The following factors must be analyzed for system impact to prevent the advantages from becoming disadvantages.

Sensor Package Volume - Smart sensors can require more volume than their conventional counterparts to house the signal conditioning and processing components. Memory and batteries, if included, will be significant volume consumers and significant volume may be needed for crosstalk suppression and shielding. This increased volume may result in the smart sensors being too large or having inappropriate form factors for their intended application.

Mass Increase - While sensors generally require little protective shielding from radiation, the electronics in smart sensors will require shielding. Additionally, the added electronics will produce small, but proportionally large, increases in sensor mass. Therefore, the individual sensor mass as well as that of the total instrumentation may increase over those built with separate electronics systems. An increase in the total sensor inventory could be offset by multifunctional sensors which could reduce the number of sensors needed.

Power Requirements - The inclusion of signal conditioning and processing components, especially memory, in each sensor may add to the total operational and, perhaps, peak power requirements. The communication interface (DCC) needed for each sensor will also add significantly to power use. Processing power use can be offset through power management by the BCC and by the use of processors having "sleep" states.

Component Availability - Space-rated versions of the devices and technologies best suited for smart sensor development may not be available. Development of selection, reliability, quality assurance, and qualifying methodologies for Commercial Off-The-Shelf (COTS) technologies could result in a wider range of component choice.

Development - While reprogrammable, smart sensor hardware modifications, retrofits, and "patches" are more difficult, if not impossible. "Scarring" opportunities will also be reduced because of their cost in wasted volume. Careful mission and design studies leading to detailed device definition and specification will be needed. Custom components from several suppliers may be needed since solid state sensor firms have limited product lines and are not often associated with integrated circuit suppliers. Application of modern "product team" and concurrent engineering management methods should be made to insure that mission requirements, user needs, technology advances, and manufacturing constraints are rapidly assimilated during development.

Manufacturability/Maintainability - Fabrication of highly integrated ("single chip") smart sensors will require significant tooling cost and willing manufacturers may be difficult to find. Hybrid fabrication technology, while more readily available, will also be relatively costly. Sensors will have limited life expectancy, some due to inherent sensor degradation (e.g., O_2 sensors) and some due to simple wear and tear. In many cases, degradable sensors can be designed for replacement of sensor elements. In most cases, however, smart sensors will not be repairable. Therefore, provision for a continuing supply of replacements should be included in the original procurement. Suppliers should have, and apply, flexible manufacturing technologies. Designs based on automated, flexible manufacturing technologies should help contain replacement cost.

Adaptive Modularity vs. "One Size Fits All" - One smart sensor may be capable of serving many needs; for example a smart pressure sensor may be used for measuring direct and indirect blood pressure and respiratory pressures as well as controlling suit pressures, freeze drier pressures, and pressures in an aquatic experiment tank. While it is possible that a single package design could be developed to fit all mission needs, such a design would necessarily include compromises and attributes that may reduce its efficiency for a given application. **In designing smart sensors, the emphasis should be placed upon development of a sensor/electronics module that can be used in a number of different packages or as a component in another device.**

4.0 CURRENT AND FUTURE HARDWARE/SOFTWARE

4.1 GENERAL

As discussed previously, the broad definition of a smart sensor includes functions of sensing, signal conditioning, signal processing, data analysis, decision making, control, and communication. Given this, smart sensors can have all the attributes of an "instrument" and should be considered as such, if operating independently of control by a host and with direct data display to the user. **Whether a device is a smart sensor or an instrument, then, depends more upon its role within a system, its dependence on other physical devices, the level of integration within the sensor package, and its mode of communication to the user.**

A number of devices currently on the market exemplify this duality as smart sensors and instruments. Dive computers provide recreational divers with numerous functions. In addition to providing measurements of water temperature, tank pressure, and depth, the integral computer provides the diver with the time he has remaining at depth with his remaining air supply. Some provide decompression scheduling as well.

Despite the apparent lack in space biomedicine of existing smart sensor hardware having all the advantages listed previously, there are a number of technologies that can be seen as precursors to the full implementation of advanced sensor capabilities. These existing technologies include ambulatory monitoring equipment, biotelemetry systems, solid-state recorders with programmable signal conditioning and signal sampling characteristics, and a variety of other sensing technologies. Most of these systems are not smart sensors/transducers in the sense of having reprogrammable microcircuitry in proximity to the sensing element; however, they can be viewed as "smart" biomedical systems if one takes into account the capabilities embedded in microprocessor-based units to which the sensors are tethered or to which they transmit their output. These "smart systems" serve to illustrate the capabilities that increasingly can be migrated to the sensor itself.

4.2 DATA PROCESSING

4.2.1 Digital Signal Processing

Direct digital signal processing (DSP) has received considerable engineering attention in the past decade as the digital capability of microchips has increased and the cost has fallen. Almost 50 million floating point operations per second can be accomplished by today's 32-bit DSP chips. Applications for this amount of processing power should grow rapidly. Ideal applications tend to be algorithm-oriented and support signal processing, such as digital filters, Fast Fourier Transforms (FFTs), and matrix manipulation (Weiss, 1991). However, DSP has never really been considered a viable technique for electrophysiological recording due to the unique signal conditioning normally required for interface to biopotential electrodes, especially the impedance transformation that is

required. Since the introduction of the Gould 24-Bit A/D converter (see Appendix D), some visionaries foresee the advantages, potential, and realism of the application of DSP to biomedical signal processing. One of the major advantages to the use of digital signal processing is the **flexibility** in function--it is easy to modify or upgrade a DSP-based system by simply interchanging an EPROM or otherwise modifying the software.

The flexibility in system configuration allows the user to specify how a channel will be used without changing the hardware. The flexibility of the system even extends to filtering, where changes can also be made without modification of the hardware. The implication of the development of this technology for use in space-qualified instrumentation is significant.

One of the first medical products to use this approach with electrophysiological signals is the recently introduced CATH 2000 cardiac catheterization system (Erlandson, 1991). In this case, DSP was used to replace special signal conditioners for strain-gages, surface ECG, intracardiac electrograms, and thermistors. The reduction in inventory and special-purpose circuitry in this product is striking. **DSP is likely to have a big impact on the way electrophysiological signals are acquired in the next decade.**

DSP chip design has moved from the dedicated function chip (Intel 2920) to the programmable DSP processors (Texas Instruments TMS 320C10 and C20) and recently to the DSP multiprocessor chips (TI's TMS 320C35 and C40, Motorola's DSP 69002, etc.)(see Weiss, 1991). These DSP chips can calculate in both integer and floating-point in parallel, examine instruction addresses, and store results from a previous calculation in the same single-cycle. Four major factors which contributed to these changes include (1) shrinkage to the submicron regime in production of complex, fast-integer or floating-point chips (2) development of super computer performance levels (3) development of high-level-language programming support to transition rapidly from bench to production, and (4) extensive manufacturer education of the engineering community on design and programming of DSP chips.

In the near future, improvements in lithography systems should increase available chip circuit area by a factor of two or three, thereby producing superchips containing from 20 to 40 million transistors in a one-square-inch surface area. At the same time, the clock frequencies may exceed 50 MHz. Shaul Berger, director of applications at the DSP Group indicated that "Designers eye the floating-point DSP chips as the ideal target, since much of the software development done on workstations and large computers employ 32-bit floating-point computation and algorithms and could be directly migrated. The algorithms would not have to be scaled to compensate for the more limited dynamic range or resolution of an integer processor." High-level languages are excellent for development and simulation of an application but hand-coding at the assembly-language level will remain the key for critical subroutines. Existing tools of today provide part of the solution to assembly-level programming. In the future, programs will be available that can

automatically translate the higher-level-language final algorithm into a program listing in assembly-language for the particular DSP chip.

Several new medical ultrasound and imaging devices are based on the computation-intensive DSP chips, but DSP chips are not the entire solution to the development of smart sensors. Since signals from the body are products of continuous biological processes, analog signal processing (acquisition, amplification, and noise suppression by common-mode-rejection, filtering) is usually used prior to converting the analog signal to some digital representation. Once the analog signal is digitized, the data may be further filtered for a particular range of frequencies.

The possibility exists for using the DSP chip to increase the usefulness of a sensor by changing its range of interest. For example, a montage of electrodes (placed about the temporal and frontal areas of the head) could be used to obtain electrocardiograms (ECGs), electrooculograms (EOGs), electromyogram (EMGs), and electroencephalograms (EEGs). The DSP could be programmed remotely to execute program "A" with specified digital filtering and processing, then to execute program "B" etc. Program "A" might set the filter cutoffs at 3.0 Hz (high-pass) and 50.0 Hz (low-pass) and the processing algorithms would detect the peak of the R-wave and determine the interval of time between peaks. The results could be processed in several parallel paths. One processing path might average "n" number of intervals and output the average heart rate every "x" number of seconds. The other path would accumulate a new wave form which could be analyzed in either time or frequency for respiration rate from the respiratory sinus arrhythmia. Program "B" might set the digital filter parameters so that the high-pass cutoff would be 0.05 Hz and the low-pass cutoff would be 5.0 Hz. The process would then include DC-offset compensation, a fast-phase identification and, perhaps, a removal algorithm followed by a slow-phase reconstruction subroutine. The data could be analyzed by curve fitting subroutines or by transformation into the frequency domain for spectral estimations. The same sensor could then be reprogrammed for EEGs and/or EMGs. It is also possible to use a multiprocessor DSP chip with two parallel processing ports to process "A" (ECG) and "B" (EOG) simultaneously, if the data have been digitized at the proper sampling rates.

4.2.2 Portable Signal Conditioning Units

NASA presently uses small, interchangeable, but not reprogrammable, amplifier/signal conditioning units which are preset for bandwidths characteristic of ECG, EMG, or EOG. These units, each slightly larger than a matchbox, consist of hybrid circuitry with I/O connectors on the housing and are powered by two stacks of coin-cell batteries. Conventional electrodes can be plugged into these amplifier/signal conditioning units and the resulting data can be fed to analog or digital recording devices on-board the spacecraft or telemetered to the ground in analog form. These units can be mounted on the body or in a flight suit and have been successfully used to monitor physiological signals during

extravehicular activities. Plans are underway to replace these units with programmable, reconfigurable signal conditioning units which take better advantage of current technology.

4.3 DATA ACQUISITION

4.3.1 Ambulatory Monitoring

Ambulatory blood pressure monitors (ABPM), such as the CMI ABPM-630 and SpaceLabs 90207, have similarly evolved from systems which simply tape record microphone and cuff pressure signals. Current commercial ABPMs typically take and display blood pressures, store the results, provide optional alarms, and provide event marking. They also contain communication ports for reporting data to specialized printers or host computers.

Both ambulatory ECG and BP monitors are programmable in terms of measurement conditions (start and stop times, frequency of measurement, abnormalities to be detected, etc.) and include self test and power monitoring with attendant user messaging. They also have the capability of providing treatment instructions to the users, although this function is not presently included due to medico-legal reasons.

The only items differentiating these monitoring instruments from the concept of the smart sensor is their construction and the limited time spent connected to a centralized data management system. All are built using commercially economical standard printed circuit board techniques and discrete components. Even without large scale integration of components onto single chips or extensive hybridization, these products are contained in shirt pocket sized packages in which most of the volume is devoted to batteries and displays.

Inflight physiological recordings have been conducted for some time, with varying degrees of success, using modified laboratory equipment and whatever ambulatory monitoring technology that was state-of-the-art at the time. Early inflight recordings involved custom-designed hardware which, while impressive for their time, offer modest capabilities by today's microcomputerization standards. Examples of such systems include those which supported recordings during parachute drops, aircraft carrier landings, during combat over Vietnam (e.g., Roman, et al., 1967; Lewis, et al., 1967; Austin, et al., 1967), and the system that was used by NASA for sleep monitoring on Skylab (e.g., Frost, 1975).

The more immediate forerunners of current technology were the Bio-Pack system developed by the Navy (see Horrigan, et al., 1970) and the Inflight Physiological Data Acquisition System (IFPDAS) developed by the Navy and Air Force (see Call, et al., 1982). The IFPDAS provided up to 32 channels of physiological and engineering parameters for flight durations up to four hours. Recordings of heart rate, skin temperature, and respiration rates were supported and the data were stored on cassette tape. The problems with this system included tape speed fluctuations when exposed to aircraft acceleration forces, maintaining transducer integrity, fixed data sampling rates, excessive maintenance efforts, and incompatibilities between the inflight processor and ground-based playback units

which made data retrieval problematic (see Banta, 1988). Several present-day systems, such as the Oxford Instruments Medilog and Del-Mar Avionics Neurocorder/Physiograph, embody this same basic design, although with some improvements in reliability and usability. Data retrieval on these systems continue to be dependent on playback units that are relatively expensive and limited in their capabilities for data selection and analysis. Another approach is to telemeter the transduced data to a base station where it can be stored on relatively inexpensive media such as video cassette recorders. These systems are not particularly "intelligent", since they use conventional passive sensor technology and store continuously recorded data in a raw, unprocessed, form.

Several efforts are ongoing to embed physiological sensors in flight suits and helmets. A system devised by motion sickness/space sickness investigators at NASA Ames involves a sensor package which provides measures of ECG, EEG, respiration, skin conductance, vasodilation, and skin temperature. Conventional sensors are used, but they are integrated into an elastic halter which is worn under the flight suit. Cabling from the sensors extends to belt-mounted amplifiers. Data are stored in analog form, on an FM tape drive incorporated in the belt-mounted package, but a digital module also provides some real-time data processing and display to a wrist-mounted LCD display. This package has been used successfully on Space Shuttle missions.

The Neuropsychology Lab at UCLA is reportedly using a custom-designed sensor/amplifier package to record EEG inflight. The package, designed by TRW Corporation, contains EEG electrodes and amplifiers molded into a flight helmet. A research group in the Crew Systems Branch at Brooks Air Force Base is likewise developing a similar concept. By locating the amplification stage close to the sensor, these approaches minimize the motion artifacts caused by the movement of electrode wires carrying low-voltage signals.

4.3.2 Stationary Monitoring

An example of a smart sensor containing both sensing/signal conditioning, and actuation can be found in a recently introduced, non-invasive, continuous blood pressure monitor (Colin Medical Instruments, CBM 3000). This device uses radial artery tonometry calibrated by periodic oscillometric cuff measurements to display the radial artery waveform and beat-to-beat blood pressures and heart rate. The key to successful arterial tonometry is precise placement of a small pressure sensor in the center of the aplanated region of a partially flattened vessel. To accomplish this, the tonometer uses an array of pressure transducers which are automatically positioned over the radial artery using an intelligent, robotic positioner.

The radial pressure is sensed by an array of thirty-two, 150-micron pressure sensors deposited in a 7-mm-long trench micromachined into a silicon substrate. Completion and compensating resistor networks, along with temperature sensors, are deposited on the substrate. The sensor is hybridized with power regulation, amplifier, and multiplexer chips for

the pressure signals. The hybridized sensor is mounted on a regulated, air-pressurized bellows which presses the transducer array against the skin. The sensor/bellows is installed on a miniature motor driven lead screw and track within a wristwatch-sized package that straps to the wrist as shown in Figure 4.3-1.

After notification that the sensor is strapped to the wrist, the processor pressurizes the bellows while monitoring the output of the pressure transducers. The bellows is pressurized until the underlying vessel is flattened but not collapsed. Waveshape and relative amplitude criteria are used to evaluate the initial position of the sensor. If the position is not optimum, the bellows pressure is released and the sensor/bellows position changed by commands to the stepper motor. This process is repeated until one of the centermost transducers is over the center of the flattened artery. The measured waveform is then calibrated against the standard blood pressure cuff measurement; care is taken in closely correlating the waveform parameters temporally to the cuff derived parameters.

After placement and calibration, the calibrated waveform is displayed on a monitor. Measurement quality parameters are continuously monitored and displayed and, when deemed necessary, the sensor is repositioned and recalibrated.

Had the CBM sensor not been designed as a smart sensor, the cable connecting the wrist sensor to the monitor would have required a thick, very stiff cable containing over 40 wires and the air line. The integration of the signal conditioning and multiplexing into the sensor resulted in the patient cable being less than 1/8 inch in diameter and containing only 8 wires and an air line.

4.3.3 Biotelemetry Technology

One particularly relevant application of biotelemetry technology is the Chemical Defense User Safety System (CDUSS) developed for the Army. The CDUSS is a physiological data telemetry system for use by untethered soldiers wearing chemical defense gear. It is designed to provide immediate warning of thermal stress and other hazards to human safety during field studies of chemical defense prophylactic drugs and protective clothing. The CDUSS contains instrumentation for transducing not only core temperature but also for ECG, respiration, and activity. These signals are obtained from individual infantrymen in the field and telemetered, via a body-worn transceiver, to a base station that will accept input from a number of such individuals. Each body-worn monitor contains two eight-bit microprocessors dedicated, respectively, to 1) control of data acquisition and storage of data in digital memory and 2) telemetry of selected data and hazard levels to the base station. In addition, there is an on-board, 16-bit processor which can be used for real-time computation of current hazard level, based on data input, with auditory feedback to the user.

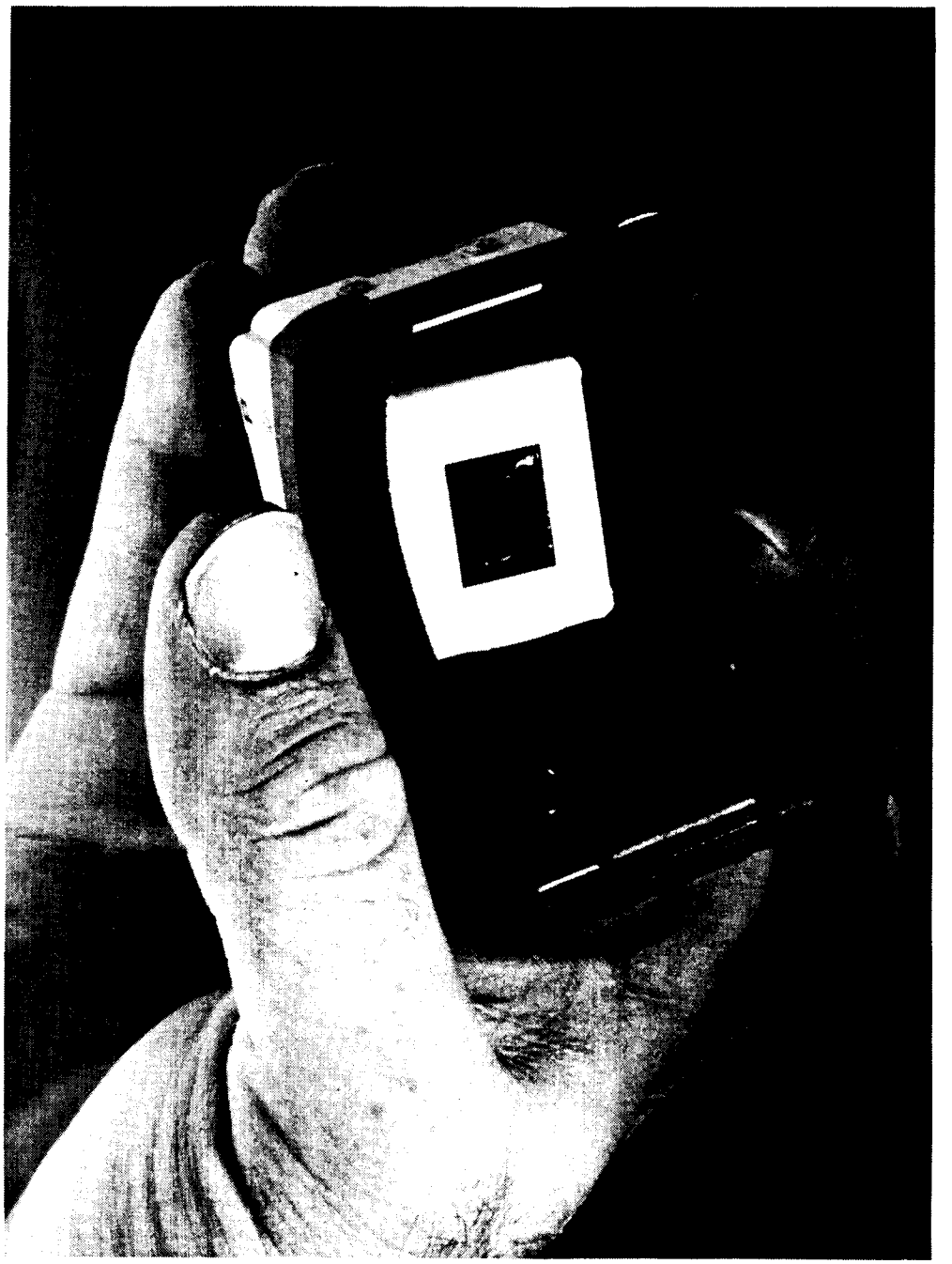

Figure 4.3-1 Wrist-Mounted Tonometer

The microprocessor, which tunes to and tracks the temperature pill, also performs data reduction of both the pill signal and other external physiological data acquired by the system. This microprocessor periodically changes the transceiver from the pill tracking (receive) mode to a transmit mode in which the reduced physiological data is sent to the remotely located base station. The CDUSS does not allow for the transmission of high-speed, multi-channel data and, although it can both transmit and receive data, it cannot perform both functions simultaneously.

Another basic design of ingestible biotelemetry sensors involves the use of very low-power radio-frequency signals which are modulated by a transducer or sensor (e.g., pressure, pH, or temperature). The battery and electronics are enclosed in a capsule and coated with an impermeable plastic. The pill is swallowed and eventually excreted in the feces, usually within 7 - 70 hours. While in the gastrointestinal tract, the signal is picked up by a body-mounted antenna and radio receiver, and demodulated into useful data. Recently, two pill designs have been advanced which represent major technical improvements, especially in terms of signal consistency, accuracy, and battery life.

One design was recently cleared by the FDA for marketing and clinical usage. The draft design was developed in 1987 for NASA Goddard. It emits a continuous wave at about 262 kHz whose frequency is modified by temperature effects on the internal crystal. The receiver consists of an inductive coil antenna and a counter which determines the incoming frequency, converts it to temperature, and stores the data at 30 second intervals in solid-state memory. The temperature is also displayed on a belt-worn receiver unit. Because of the relatively low frequency and inductive link, this pill may be sensitive to noise in proximity to computers, heavy vehicles, and other dense electromagnetic fields.

Yet another design had its origins in a pill developed by Konigsberg Instruments in 1972 and used by both NASA and the Naval Medical Research Institute. Rather than a continuous high-frequency wave, these pills emit brief bursts or pulses of a radio signal in the FM band. The time interval between pulses conveys the temperature information to a belt-worn radio receiver with decoding software. Because of this design, pills with differing frequencies are free from cross-talk, and can be used in several subjects at the same time. Furthermore, this pill emits two brief pulses in rapid succession, followed by a long pause. The long interval is modulated by temperature, while the short pulse-pair are characteristic for each pill. Thus receiving/decoding software can discriminate the pill in spite of varying amplitude and ambient noise. This permits more accurate tracking, reduced dropout, and resistance to interference.

4.4 DATA STORAGE

New biomedical recording devices and approaches bear witness to the progress that has been made in recent years in devising systems based on advances in microprocessor technology and data analysis techniques. It is clear that increasing intelligence is being migrated to small, body worn units or units that can otherwise be located in proximity to

the sensor/transducer while still communicating effectively with on-line data processing instrumentation. These capabilities foreshadow the near-term future development of true smart sensor technology. It is likely that we will soon see increasing capabilities built into the sensing unit itself, as well as multi-sensor outputs being networked together and integrated with artificial intelligence for the purpose of interpreting system status. The advantages of smart sensors that have been noted earlier for other applications (e.g., Voecks & Seshan, 1991) certainly pertain to biomedical monitoring systems and promise to increase reliability, flexibility (i.e., user programmability), and data storage efficiency while decreasing size, weight, power requirements, and obtrusiveness of the physiological sensors.

4.4.1 Smart Sensors and Biomedical Recording

Magnetic tape ambulatory ECG recorders are beginning to be replaced by solid-state ambulatory ECG monitors. The microprocessors in these devices perform real-time data compression to allow solid-state storage of the ECG waveform data. Since data compression inevitably produces data loss, the solid state monitors also perform real time ECG analysis to detect abnormal beats. Abnormal beats are then stored without data compression. Solid state ECG monitors also compute and store heart rate, accept event mark input from the user, and some provide optional alarms when specified rates are exceeded or other abnormalities occur. Some of these new devices also provide report formatting and control of the record so that they can connect directly to a printer. Others make use of special-purpose computers for data analysis and display, but most use serial or parallel communication connections to personal-computer-based data acquisition systems. Manufacturers typically provide interfaces for modem transmission as well as direct cable connection.

4.4.2 Solid-State Data Recorders

Other recent systems have begun to utilize on-board microprocessors and have recognized the advantages of storing data in non-volatile, solid-state memory. The Vitalog offered the first widely used multi-channel recorder with digital storage capabilities. It was user-programmable to some extent and provided facilities for monitoring ECG, temperature, respiration and GSR. However, it had fairly severe limitations in its frequency response/sampling rate (e.g., would not accommodate EEG), data storage capacity (256 KB), and data retrieval/data analysis capabilities. Other solid-state physiological "data loggers", such as the Grant Squirrel, are even more severely limited in their sampling rates and data processing capabilities, although they provide a very cost effective alternative for recording parameters, such as temperature, which can be sampled fairly infrequently. Several ambulatory ECG "Holter monitors" are now available with solid-state memory.

The Solid State Physiological Inflight Data Recorder (SSPIDR) developed by Systems Research Laboratories for the Navy probably represents the state-of-the-art over the last few years. Developed as a successor to the IFPDAS, this portable system (see abstract by Call, et al., 1988) offers sixteen data channels and several megabytes of solid-state RAM,

providing capabilities for recording relatively high frequency physiological indices such as EEG and EMG. The on-board microprocessor allows for user selection of variable sampling rates and gains. Data acquisition can be switched on and off based upon programmed criteria such as G-loads sensed by accelerometers built into the unit.

The present state-of-the-art in solid state data recorders, and a system which perhaps foreshadows true smart sensor applications, is the Modular Intelligent Data Recorder (MIDR) offered by GMS Engineering Corporation. The MIDR, which has just recently been introduced commercially, is a general-purpose, data recorder, but was developed with physiological signals and dynamic man-machine system applications in mind. The MIDR system, in its minimum configuration, consists of a power supply module, a primary analog input module, a digital interface module, a computer & communications module, and an extended memory module.

The minimum configuration can be expanded by installing additional modules. Up to a total of four primary analog input modules (each with a maximum of 16 channels) can be added, yielding a maximum of 64 analog input channels, each with programmable software and independently specified amplification, filtering, baseline adjustment, and sampling rate. Up to eight extended memory modules can be added to the system, yielding a maximum of 256 megabytes of solid-state memory. An auxiliary power module can be used to change ("swap") extended memory modules during data acquisition, without interrupting system operation; this permits virtually infinite memory storage capacity. The minimum configuration system additionally has eight auxiliary analog inputs, two 12-bit resolution analog outputs, four digital level inputs, four digital level outputs, two event (interrupt) inputs, and two programmable clock (pulse width modulator) outputs, providing a great deal of flexibility in interfacing with both sensor/transducers and other computing devices.

The MIDR obtains part of its enormous flexibility from its open software architecture and contains, albeit in a "black box," much of the flexibility and user-programmability that is foreseen for smart sensors. Gains, bandpasses, filtering characteristics, and sampling rates are all user-programmable on a per channel basis (allowing for the concurrent recording of a variety of different physiological signals or a variety of equipment/system indices). Analog signals are sampled with a 16-bit digitizer. Incoming data can be processed in real-time, allowing derived measures to be stored along with, or instead of, raw data; data storage can be triggered in a variety of ways (e.g., based on time, external signals, or real-time processing of input data); and control signals can be output to other instruments contingent on the processed input.

4.4.3 Activity Loggers

Another relevant ambulatory recording technology, while not providing physiological data *per se*, is the Mini-Motionlogger Actigraph offered by Ambulatory Monitoring, Inc. The Actigraph is a wrist-watch-sized accelerometer/recorder. The sensor is a piezoelectric beam with the capability of detecting motion in all three axes. The unit is powered by a

lithium coin-cell battery and will run certain data collection protocols for more than a week without recharging. An on-board 8-bit microprocessor controls the data acquisition protocol, with programs being downloaded from a PC via a serial port. Digitized data are stored in a 32K RAM and can be uploaded to the PC for analysis using supplied data analysis software which implements both zero-crossing and time-above-threshold measures. The Actigraph and its predecessor have been used for studies of wake-sleep cycles, circadian rhythm, and the effects of jet lag. A related device incorporates the functionality of the Actigraph along with a linear light sensor, and stores both light and motion data along a common time-base. These devices provide good examples of the sort of flexibility and processing power that can that can be provided in proximity to specific sensors.

4.4.4 Smart Cards

A "smart" card has been defined as "a credit-card-sized plastic card, with an integrated circuit embedded into it, with a microprocessor on board, in addition to the memory" (Seidman, 1991). Such cards may have contacts or they may be contactless and they are being cast in a variety of shapes, such as badges, keys, buttons, etc. (Legg, 1991). A recent configuration has also included an integrated power supply, keyboard, and display. One vendor has even shown a prototype of a smart card with a 30-megabyte optical disc in the center and the entire card is spun in the reader/writer (Seidman, 1991).

Although "smart" cards have been around for many years, they have not lived up to the dreams of their earlier proponents. However, these cards, now involving some level of computational power rather than just data storage (which can be done less expensively with magnetic-stripe cards), are poised for many new uses.

The smart card device configuration is likely to appeal to many uses on SSF. Smart cards could be used to program smart sensors for special applications, or they could be used for extending the computational power of a sensor/signal conditioner. Re-programming of on-board instrumentation might be carried out with smart cards. Data security and access control might also be provided through their use by individual crewmembers.

4.5 ADVANCED PHYSICAL SENSORS

One of the primary technologies for advanced physical sensors is based on silicon processing or micromachining, which has proven its utility in the success and maturity of pressure sensing products available today. Fiberoptic and piezoelectric film technologies are competitive in features with silicon microsensors in some applications, such as chemical and temperature sensors (fiberoptic) and accelerometers (piezoelectric film). However, the advantages of silicon sensors in terms of cost, size and material characteristics make silicon generally the technology of choice. Although over a million transistor equivalents can currently be integrated on a single silicon chip, this number is expected to expand by several orders of magnitude over the next decade. Such progress will make very sophisticated sensors cost effective for numerous new system applications.

The nascent field of micromachining, especially, holds promise of even greater advances in miniature motors and actuators combined with chemically active sensors. Combining these technologies provides the promise of producing small, low-power devices with sensing, conditioning, processing, and control elements on a single chip. The integration of sensors and actuators using microfabrication techniques promises to provide a vital link between monitoring and control applications (Wise and Najafi, 1991). Such devices can act either as stand-alone devices or as elements of a distributed, networked system.

To date, many smart sensors have been developed by integrating components from these silicon technologies into hybrid devices. This is due in part to incompatibilities in manufacturing processes but more to a lack of market demand for "single chip" devices that would justify their development cost. Hybridization, it should be noted, is not necessarily a disadvantage. As a general rule, sensors must be in contact or as close to the environment they are measuring as possible. Harsh environments may require physical separation of the sensor from the rest of the system. In other cases, sensors may require periodic replacement making separate packaging desirable. Further, spatial separation is sometimes defined by the measurement (e.g., mass and heat transfer measurements and field measurements such as ECG).

Silicon fabrication technologies have been applied in recent years to large-scale production of low-cost, standardized pressure transducers. Initially spurred by sensor needs for emission control systems in the automotive industry, disposable, standardized silicon pressure transducers are routinely used for intraoperative monitoring of direct blood pressure. Silicon pressure transducers have also made possible low-cost automated sphygmomanometers for hospital and home indirect blood pressure monitoring.

This section is primarily focused on silicon sensors used for physical measurements and on the convergence and integration of these measurement devices with integrated circuit technology to form "smart" sensors. Biomedical applications for these sensors have often followed automotive and heating, ventilating, and air conditioning (HVAC) applications and have benefitted from the manufacturing economies of scale associated with consumer products.

4.5.1 Micromachined Silicon Technology

Micromachining is derived from a silicon engineering process in which small mechanical components are manufactured by chemical shaping or etching techniques. The manufacturing process utilizes deposition, lithography and etching techniques which were developed for integrated circuit production along with some specialized techniques peculiar to sensors. These processes provide the advantage of using established low cost batch processing. For example, one manufacturer utilizes a four-inch silicon wafer which contains 6,600 sensors. The wafers are processed in cassettes which carry 25 wafers or 165,000 sensors. In a manufacturing run using 30 cassettes, several million sensors can be produced. The typical wafer processing cycle is only three to six weeks. In volume pro-

duction (> 1 M units per year), silicon pressure sensors cost less than $2 each. This low processing cost for silicon sensors is a result of ten years of production experience in which silicon sensors followed the cost/learning curve for integrated circuits.

The mechanical characteristics of silicon make it attractive for use in sensor fabrication. It has a high modulus of elasticity, the same as steel, while having the density of aluminum. Its tensile strength is greater than steel and its hardness the same as quartz. Due to its crystal structure, it has essentially perfect elasticity with no mechanical hysteresis. It has the thermal coefficient of expansion of quartz and is available commercially in a very pure form which eliminates fatigue effects. It is micromachinable to submicron geometries. It is a semiconductor, thus allowing for integration of electronics directly with the sensor mechanism to form a monolithic "smart" sensor.

It is important to note some of the market conditions which enabled the introduction of low-cost, high-volume sensors into the medical and automotive markets. In the mid-1980s, the government started imposing cost containment guidelines on the medical industry. This caused the selling price for disposable blood pressure transducers to drop and forced sensor manufacturers to accelerate price learning curve efforts to reduce costs. The disposable transducer was also viewed as lower in cost, safer, more accurate, and more reliable than the reusable transducer it replaced.

The market for silicon sensors in automotive applications is growing rapidly and is becoming a driving force in new sensor development, increased manufacturing efficiencies, and unit cost decreases. Several factors are responsible for the increase in electronics and sensors in automobiles. Government mandates for fuel economy, passive safety restraints, and emission controls are certainly a primary factor. The availability of inexpensive digital microcontrollers and displays and the subsequent trend towards increased self diagnostics plays a part in the increased demand. Additionally, the need for improved performance and increased luxury and convenience are also factors. Silicon sensors are one of the primary technologies benefitting from being incorporated into the automotive market. Low-cost, high-reliability, rugged construction and small size are the primary factors for the success of the silicon sensor in this market.

The automotive market utilizes about 20 million silicon pressure sensors per year whereas medical applications involve about 8 million per year. Additionally, silicon accelerometers which are used for airbag controllers are experiencing increased usage.

4.5.2 Pressure Sensors

An early example of a "smart" pressure sensor adapted from a conventional sensor is shown in Figure 4.5-1. The Validyne Model P305D pressure transducer is based upon Validyne's standard variable-reluctance pressure transducer. The oscillator and demodulator electronics, as well as the regulated power supply, needed for this type of transducer have been miniaturized and mounted on the same baseplate as the transducer itself.

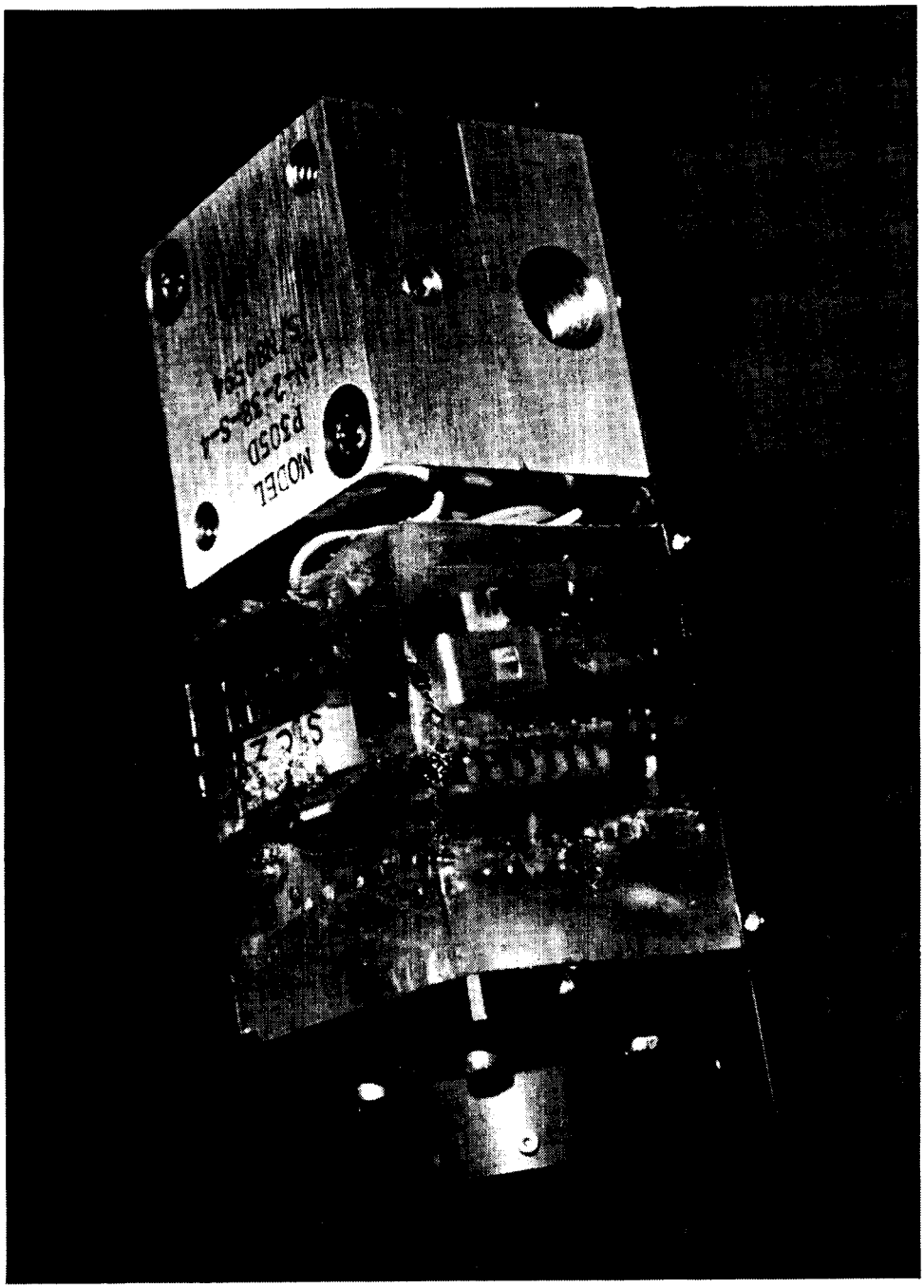

Figure 4.5-1 Early "Smart" Pressure Transducer

Conventional electronic components mounted on four printed circuit boards are used. As shown in the photograph, the printed circuit boards are potted in silicone rubber to provide environmental protection and separation of the boards. A protective cover (not shown) is used to provide further protection of the electronics package. The potting, along with the physical separation of the electronic sections, is needed to reduce crosstalk within the circuitry. This illustrates the fact that the limits of micro-miniaturization can often be defined by the degree of isolation that is required and the means selected for providing that isolation.

While not particularly advanced, this design overcomes the major problems often associated with variable-reluctance transducers. By placing the modulation and demodulation near the transducer, crosstalk and signal loss occurring when using the long cables are eliminated. Further, when viewed at the connector, the transducer appears and behaves like more commonly used resistance strain gage pressure transducers. This allows its use with conventional recording instrumentation without the need for intermediate preamplifiers.

Further advances in pressure sensing technology can be divided into the following three categories, based on the sensing method:

- Piezoresistive.
- Capacitive.
- Fiberoptic.

Piezoresistive Silicon Pressure Sensors

Piezoresistive silicon pressure sensors are the most common type of advanced sensors available today. They are fabricated in large volume using micromachining and semiconductor batch processing to achieve low cost. They are typically laser trimmed in production to adjust their performance to meet specifications.

One of the innovations which makes silicon sensors attractive is the development of computer aided engineering (CAE) systems for analysis of stresses in these small mechanical sensors. These CAE systems perform finite element modeling (FEM) using three dimensional models of the sensors. This analysis capability allows the chip designer to locate the piezoresistive elements at optimal locations for stress transfer. It also allows modeling of temperature effects. The effect of this analysis and modeling capability is shorter design cycles, higher yields and enhanced performance of sensor products. In consumer markets this capability can be critical where market windows and product life cycles are short. It can also be critical where sensor performance is an issue.

The integration of electronics into the piezoresistive sensor is feasible. At least one company, Motorola, offers a standard product (MPX5000 Series) with an interface amplifier integrated onto the same silicon wafer as the pressure sensor. These sensors are designed to interface directly with A/D converters without additional circuitry. They operate from a 5-volt supply and have a full scale output voltage of 0.5 to 4.5 volts. These sensors are a proven step towards "smart" sensors which have more extensive circuitry.

Capacitive Silicon Pressure Sensors

Capacitance-based silicon pressure sensors are one alternative to piezoresistive silicon pressure sensors. These sensors detect the deflection of diaphragms or other microstructures which cause a change in capacitance between conductive areas on these structures. The silicon microstructures are fabricated using the previously described micromachining techniques. The claimed advantages of capacitive sensors over piezoresistive sensors are improved temperature and long term stability and increased sensitivity.

The interface circuitry for converting capacitance changes to a voltage or frequency suitable for displaying or recording is an integral part of the capacitive sensor measurement system. Integration of the capacitive sensor and interface electronics could improve the performance and manufacturing efficiency of these sensors. Application-specific integrated circuit (ASIC) technology has been used to develop front-end circuitry for capacitive sensors. The interface circuitry usually employs either an oscillator whose frequency is modulated or a switched capacitor integrator whose output voltage amplitude is a function of the capacitance change.

Another feature of capacitive sensors involves a unique technique for built-in self testing. In some microstructure configurations, an electrostatic force may be generated between the plates of the sensor capacitor. This force may be used to test the pressure sensitivity of the sensor. This self test feature may have potential applications for checking sensors in remote locations which are virtually inaccessible, such as on a space vehicle or inside the body.

Capacitive silicon pressure sensors have been overshadowed by piezoresistive sensors and have not had widespread commercial use. However, at least one manufacturer (Kavlico, Corp.) has produced a capacitive silicon pressure transducer. The fabrication of the silicon microstructures required for capacitive sensors is still a relatively new area of development. As the efforts to integrate micromachining with electronic interfaces advance, there may be a potential for a commercial, low-cost capacitive sensor with self- test capabilities.

Fiberoptic Pressure Sensors

Fiberoptic pressure sensor technology is used for catheter-mounted pressure sensors in medical applications. The conversion of the physical input stimulus to a measurable light

signal is usually achieved by a miniature structure at the tip of the fiberoptic catheter. Light is transmitted down a fiber to the force sensing structure where it is modulated. The sensing structure is usually a mirror or a conical diaphragm which reflects light to receiving optical fibers in proportion to a physical input. The sensing mechanism may be a chemical indicator which changes absorption or fluorescence as a function of pressure. The modulated light is returned to an electronic monitor which converts it to a voltage or digital value.

The optical fibers in these sensors are suitable for low-cost, high-volume production; however, the sensing structures and the mechanism for coupling them to the optical fibers appear to be limiting factors. Another disadvantage in the medical market is the complex and expensive electronic interface module required to convert the modulated light in the fiberoptic fibers to an electronic signal.

The fiberoptic pressure sensor has two advantages over the silicon sensor. It is inherently electrically isolated for patient safety. And it can be constructed to have better sensing media compatibility. The piezoresistive elements and compensation circuitry for the silicon sensor must be protected from corrosive environments. Typically, they have a silicone gel coating which is very soft and flexible so as not to affect the performance of the sensor. These gels are not impervious to corrosive environments or moisture over long-term exposure.

Fiberoptic pressure sensors are available from several manufacturers. Camino Labs has a disposable catheter-tip pressure transducer for monitoring intracranial pressure as does Ladd Research Industries. Fiberoptic Sensor Technology offers fiberoptic pressure sensors for commercial applications.

4.5.3 Accelerometers

Silicon accelerometers are fabricated using micromachining and processing techniques similar to those used for silicon pressure sensors. The immediate potential for high volume, low-cost units in the automotive industry has resulted in intense development efforts for silicon accelerometers.

The design of silicon accelerometers is based on measuring the acceleration-induced inertial forces exerted on a seismic mass anchored to the silicon chip by an elastic suspension. The motion of the seismic mass or the strain on the suspension may be measured to determine the acceleration. The strain on the suspension is measured by piezoresistive elements while the mass motion is measured by capacitive sensing. Accelerometers using this type of sensing are referred to as open-loop designs. Silicon accelerometers using force-balance designs have been constructed. They require complex electronics and usually use capacitive sensing. A third type of silicon accelerator is based on detection of frequency shifts in pairs of resonant microbridge strain gages embedded in the seismic

mass suspension. These resonant accelerators require complex microfabrication techniques.

A number of challenging design problems must be overcome to achieve a viable silicon accelerometer. Such devices must be capable of withstanding over-shocks hundreds of times their normal operating range. This protection is accomplished by designing mechanical stops to limit deflection of the seismic mass. The combination of the seismic mass and suspension mounting spring form a high-Q mechanical system which must be damped. Air trapped in the cavity between the seismic mass and deflection stops is used to provide this damping. Another problem for moveable microstructures is the interference of particles. This problem is addressed by using the deflection stop structures to form a seal around the seismic mass. Also, all processes prior to sealing are performed in a clean room environment.

There are additional problems with these sensing techniques. The piezoresistive sensors are temperature dependent and must be compensated. The capacitive sensing technique requires complex circuitry which adds cost. The force-balance type silicon accelerometer usually utilizes capacitive sensing to eliminate the temperature effects associated with piezoresistive sensing. These designs typically combine the sensing circuitry with the electrostatic force feedback control circuitry in an application-specific integrated circuit (ASIC).

One advantageous characteristic of these silicon accelerometer designs is the relative ease of adding self testing capability. By metal plating one surface of the seismic mass and a parallel surface on the deflection stops, a parallel plate capacitor is formed. By applying a voltage across these plates, an electrostatic force is generated simulating a known acceleration. This self test feature can be used to calibrate the sensor in production and to compensate for drift and temperature effects in use.

An example of a smart sensor design applied to accelerometers has been presented by Analog Devices (Core, et al. 1991). This silicon accelerometer design is for automotive airbags, smart munitions, and inertial guidance. It integrates the silicon acceleration sensor and the sense, control, and interface electronics into a monolithic silicon wafer occupying an area of 10 mm^2. The accelerometer employs capacitive sensing and can be operated as an open-loop or closed-loop force balance system with self-test capability. It operates from a 5-V supply and has a high level analog output which has a 5% accuracy over a +/-50-g range. It employs low-powered, integrated circuit technology capable of 24-V operation and is described as industry's first surface micromachined accelerometer with on-chip signal conditioning and self-test circuitry.

Accelerometers have not been used extensively in medical applications. One potential application is to measure the acceleration of the heart wall (ballistocardiogram). Another potential medical application is as an activity monitor.

4.6. MISCELLANEOUS "SMART" TECHNOLOGIES

4.6.1 A/D Converters

Integrated schemes with a multiplexer and a single A/D converter work well when the signals to be acquired have the same frequency and magnitude ranges. If the problem is to collect various physiological signals with differing frequency and/or magnitude ranges, then the A/D conversion must take into consideration the frequency ranges of the acquired signals. Thus it might be possible to combine signals such as, ocular movements via infrared sensors, electrocardiograms via a 3-lead configuration, respiratory and valve sounds via a piezoelectric microphone, electroencephalograms via a 10/20 electrode configuration (montage), and core temperatures via infrared sensing from the ear drum. In this case, parallel processing may be advantageous over serial processing. In turn, the decision to use parallel or serial processing affects the selection of the microprocessor.

A few years ago, Gould Electronics released for marketing a two-channel, 24-bit, 34-pin, dual in-line package, A/D convertor that can handle signals of ± 10 volts, dc-to-410 Hz (See Data Sheets, Appendix D). When operated with a 2.048 MHz clock, the converter will sample at a rate of 1,000 samples per second and output serially. Additionally, the converter may be paired with the finite impulse-response (FIR) filter which operates with a 20 MHz clock. Although this converter has a very large dynamic range (120 Db), it has a rather low input impedance (60 kilohms). Unless the sensor output impedance is very low (60 to 600 ohms), the convertor may load the output of the sensor and a buffer amplifier must be placed between the sensor and the A/D converter chip.

The silicon sensor market is heading towards new sensors with increased integration, according to a recent paper by Randy Frank, Motorola. The development of low cost silicon sensors in parallel with the development and availability of single chip microcomputer units (MCUs), digital signal processors (DSP), ASICs, and customer-specific integrated circuits (CSICs) offers the opportunity for integration to smart sensors. One of the challenges of increased integration is packaging and interconnection of these devices at the wafer level. The packaging requirements for silicon sensors and microelectronics are very different in that stress isolation is a primary objective for sensors while wire bonding and adequate pin outs are primary concerns for digital processors. Mr. Frank feels that the integration of sensors and microelectronics will progress in stages.

In support of this stepwise integration, Motorola has introduced the first commercially available silicon pressure sensor suitable for direct interface to an A/D converter (MPX-5000 Series). These efforts at integration improve the interface between the sensor and the intelligence process; however, they do not provide any self-calibration, decision-making, communication, or control functions associated with smart sensors.

Digital compensation is an application of intelligence to the sensor interface to improve system performance (Ansermet, et al., 1990). This technique involves characterizing the

sensor performance and converting it to a compensation function or polynomial and has shown significant performance improvement in laboratory tests.

4.6.2. Modelling/Artificial Intelligence

A more sophisticated approach to digital processing of sensor information is presented by Kroschel and Wernz (1991). They describe a method for detection, localization, and identification of sensor failure or performance degradation such as offsets, drift, or loss of sensitivity. In this approach, the process which the sensor monitors is mathematically modeled to generate statistical estimates of measured quantities. The system or process faults are suppressed by an adaptive decorrelation algorithm. The system detects sensor faults by differences in measured output versus estimated output. It can localize the fault to a particular sensor and statistically characterize the error to correct for it. This type of intelligence could be incorporated into an integrated sensor-controller system.

Voecks and Seshan (1991) describe the possible use of *in situ* sensors intelligently integrated into a control architecture for continuous operation of highly autonomous life support systems for humans in space exploration. Desirable functions of the intelligent integrated sensor system require operation in "a hierarchical network involving distributed data bases, prediction models, rule-based reasoning modules for hypothesis generation and verification, controllers and final control elements." Abundant literature is available on gas or chemically sensitive sensors or infrared detectors on silicon chips. The widest applied sensors by the aerospace, automotive, and medical industries are the silicon-based, monolithic, pressure transducers. Additionally, there is renewed interest in fiber optic sensing techniques of gasses and chemicals to evaluate levels of concentrations.

The next logical step in the development of these sensors is the evolution from a smart programmable sensor to a sensor with artificial intelligence. With the addition of faster microprocessors and greater memory, it follows that advances in the next generation should emphasize software rather than hardware to emulate human manipulation of knowledge, reasoning, and decision making. Smith, et al. (1991) categorize the volumes of raw material to be processed into four general types of data processing space; i.e., *data, information, knowledge, and intelligence.* The space increases in complexity and sophistication as the raw *data* are organized and refined into more efficient representations from data processing to *intelligence* processing. Most of today's computer operations are in the *data* processing category and are capable of *information* processing. Smith, et al. (1991) propose that the "data become structured as information when they are linked by semantic and syntactic relationships." The linking of information items and symbolically manipulating them is termed "knowledge processing." The final processing space is *intelligence* processing, which the authors contend does not exist to date. Current artificial intelligence applications fall into the knowledge processing space. Popular forms of artificial intelligence include knowledge-based systems, artificial neural systems, and fuzzy logic control. "These AI technologies are all model-free estimators," implying that no models,

algorithms or statistical manipulations are used to derive or compute the correct solution (Smith, et al., 1991).

Artificial Neural Systems try to emulate parallel processing in biological systems; biological neurons facilitate a process after many iterations (training). These networks can be trained to perform a task. Training strategies include reward (by increasing the weighting functions along a path) and punishment (by decreasing the weighting functions). After repeated trials, the system will learn correct solutions to new problems on its own.

Further advances in chaos theory and fuzzy logic may see their way into the advanced intelligent transducer application field. According to Smith, et al. (1991), "fuzzy logic contends with uncertainty and imprecise knowledge, meaning, and inference in control and decision-making applications."

4.6.3 Smart House

The electrical distribution system of the SMART HOUSE can serve as a conceptual model for systems integration. It provides generic ports throughout the house into which electrical, telephonic, audio, video, and high speed data processing devices can be interchangeably connected (see Moore, 1986; Gilmore, 1988). These devices, which themselves adhere to certain "smart system" standards, handshake with the infra-structure in a way which both identifies the device to the system and provides interconnects which enhance safety and energy efficiency. The unified wiring network replaces the multiple wiring systems found in homes today. The SMART HOUSE core wiring system is based on a hybrid cable design consisting of a flat, ribbon-cable construction containing conductors for 120-volt AC power, 12-volt DC uninterruptible power, telephone, control functions, and coaxial cable. These hybrid cable branches terminate at attachment points called "convenience centers" and "switch/sensor outlets." The homeowner can interchangeably plug any compatible appliance, telecommunications, data processing, or entertainment device into these ports, and competing vendors can supply proprietary devices for use in the house as long as they adhere to certain publicized standards. Numerous vendors have been licensed by the SMART HOUSE organization, a subsidiary of the National Association of Home Builders, and several prototype houses are now in operation. **Although the core wiring system of the SMART HOUSE is quite different from the electrical, telecommunications, and data processing features needed on SSF, the concept of an infrastructure that supports generic ports and interchangeable smart devices is worth considering for SSF.**

4.6.4 Chemistry Analysis

Another example of a smart sensor system in a medical application has recently been announced by PPG Medical Systems. They have introduced a new product called the StatPal Blood Gas Analysis System which measures pH, pO_2 and pCO_2 in 60 seconds at the bedside. This system combines a low cost disposable sensor-syringe with an intelligent

monitor. The disposable Sensyr™ syringe combines the sampling syringe, sensing electrodes, calibration solutions, cable and connector into one package. The Sensyr™ is first connected to the monitor and a self test/calibration cycle is initiated by pressing the plunger to expose the amperometric and potentiometric chemical electrodes to the calibration solution. The monitor compares the electrical response of the electrodes with bar code calibration values on the disposable packaging. After 60 seconds the monitor indicates whether or not the Sensyr™ is performing normally. A blood sample is then taken and a reading is obtained in seconds. The monitor is portable (3 lbs), battery-operated, and costs about $3,000.

For several years, researchers at Sandia National Laboratories have been developing surface acoustic wave (SAW) devices that can be built into a variety of sensors. In general, two transducers are formed on a piezoelectric substrate such that when one of them is excited with an alternating current, it launches an acoustic wave which travels along the surface. The alternating mechanical strain is converted back into an electrical signal by the other transducer. The wave interacts with a thin film applied to the surface during the course of the travel. The polymer film is selected for its chemical absorption characteristics. Thus, the sensor has the promise of becoming a "mass spectrometer on a chip" which could improve the speed and reduce the cost of chemical sensing.

Sansen, et al. (1989) describe what they term a monolithic "smart" sensor for voltammetric measurement of glucose or oxygen concentrations. The sensing element consists of a silver/silver-chloride reference electrode and an active silver electrode for measurement of pO_2. The electrodes connect to a current-to-voltage converter which consists of a switched capacitor integrator followed by a sample-and-hold circuit. Sansen's smart sensors are not in agreement with the terminology referred to by Coats, et al. (1989), that "smart" is the designation given to electronic devices which contain a microprocessor.

4.6.5 Multi-Feature Sensors

Two examples of stand-alone smart sensors used in space flight applications are the Stand-Alone Pressure Measurement Device (SAPMD) and its derivative, the High Pressure, Stand-Alone Pressure Monitor (HPSAPMD). The SAPMD flew on the maiden voyage of Space Shuttle Discovery. This small (6.0 X 1.5 X 0.4 inch) data acquisition system monitors the aerodynamic pressures on external surfaces of the Space Shuttle for several minutes after launch. Containing pressure (0-15 psi) and acceleration sensors, signal electronics, A/D converter, microprocessor, EEPROM, and a serial port for postflight data download, the SAPMD is packaged to endure the rigors of up to 10 space flights when mounted under the heat resistant tiles. The SAPMD, having been preprogrammed with a nominal launch date, monitors the acceleration sensor to determine the actual start of launch. It then begins acquiring and storing pressure data for the duration of launch and then shuts itself down. The HPSAPMD is mounted inside the nozzle of booster rockets to record pressure developed within the nozzle during test firing. It is essentially the same as the SAPMD except that a higher range (0-1000 psi) pressure gage

is used along with a heavier (2000 psi proof test) case. This more rugged construction is required to protect the sensor from not only the high pressures and temperatures in the nozzle but also from the solid fuel residue or slag which may completely cover it. The HPSAPMD is mounted in the nozzle at assembly where it remains until disassembly after the test. Its resident programming senses initiation of the rocket burn and then records 160 seconds of pressure data.

In applications, such as for long term ambulatory electrocardiogram monitoring, the smart sensor may consist of a montage of electrodes integrated with impedance matching circuitry, multiplexing circuitry, anti-aliasing filters, analog-to-digital converters, a microprocessor, memory, EPROM with software for analysis, and telemetry circuitry to output data. The complementary receiving situation must demultiplex the data for subsequent processing, analysis and display. An example of this technology is demonstrated by the works of Fitzgerald, German, Hensley, Denton, and Tatman directed and mentored by Dr. M. Kabrisky, in the development of an array of microelectrodes for studies of the visual cortex (Tatman, 1979; Fitzgerald, 1980; German, 1981; Hensley, 1982). Each microelectrode is connected to a junction field-effect transistor (JFET) which is switched on or off by an electrode drive circuit. The signal from an electrode is input to a source-follower amplifier (single transistor, unity gain amplifier) which serves as impedance matching circuitry. A pulse-width-modulation (PWM) scheme was selected over an entire A/D converter, because the PWM circuitry required less components, less substrate area, and had simpler circuitry when compared to an A/D converter. Data were transferred transcutaneously via an internal RF link (Ko and Newmann, 1967).

Hartl, et al. (1991) have described an example of a smart sensor system which is used for airbag control in automobiles. A silicon accelerometer is integrated into a microprocessor-based electronic control unit (ECU). This integrated package operates as an independent unit for crash sensing and airbag activation. The unit is intended to replace distributed systems where several remote crash sensors are linked to a central controller. The primary function of the ECU is to deploy the airbag during a vehicle crash. It also performs continuous system monitoring and failure detection. It has self test capability for the silicon sensor, warning lamps, and firing circuits and provides a warning indicator to the driver if it detects a failure. The ECU hardware has several features for enhanced reliability, including a reserve energy supply for maintaining operation during a battery loss which might occur during a crash. It has a watchdog timer for monitoring the operation of the microprocessor and for warning of improper operation. Redundant mechanical g-switches in series with the airbag actuator firing circuit are used to prevent inadvertent inflation. A communication interface to external diagnostic testing devices is also provided. The performance and reliability enhancements incorporated in this system are an example of the type of integration that smart sensor technology is intended to achieve.

Talmadge and Appley (1991) have fabricated a programmable transducer microchip. The chip provides conditioning/processing of sensed input signals, including impedance matching, filtering, gain selection, integration/differentiation, temperature compensation, and

calibration. The chip is designed to interface with specific sensing elements. Piezoelectric and piezoresistive transducers convert vibration, shock, pressure, and direct force signals across the two opposite faces of the crystals which are proportional to the applied acceleration, pressure, or force. The resistive elements are passive (bridge circuitry, strain gage) and active (piezoresistive). Additionally, the microchip generates output codes that permit identification of the sensor and test signals for evaluation. The microchip is operated through a programmable transducer control module (PTCM) which supplies power to the chip and provides the serial interface for the transducer signals to a PC or data acquisition equipment.

5.0 CONCLUSIONS AND RECOMMENDATIONS

The concept of a smart sensor combining sensing, signal conditioning, signal processing, and digital communication and control has many potential benefits to the NASA Life Sciences Program. **Development of a relatively small number of smart sensor platforms can provide a wide range of measurement and control functions depending upon the sensor installed.** The high level of autonomy provided by intelligence at the sensor level should reduce communication network load and control demands. However, when included in a robust architecture such as the Medical Information Bus, smart sensors will provide efficient and rapid response to DMS and mission information needs. The design and development of smart sensors should to be done carefully to insure that maximum flexibility of sensor platforms is obtained while providing for mission-specific requirements. Planning and procurement should include provisions for guaranteeing long-term supplies of sensors and incorporation of advances in the state-of-the-art.

Smart sensor technology has not been well-utilized to date for biomedical applications in space or otherwise. Somewhat surprisingly, there is little evidence of such technology being in the "pipeline" of instrumentation being prepared for near-term scientific research on the Space Shuttle or Space Station. However, a number of currently available biomedical devices implement aspects of smart sensor technology and can be seen as foretelling future engineering developments as the component technologies become more cost effective. These precursor physiological monitoring technologies include such devices as the portable amplifier/signal conditioning units presently used by NASA, radio-frequency transmitting temperature pills, biotelemetry systems with some degree of local data processing, certain ambulatory monitoring systems, solid-state data recorders with programmable "front-ends," motion-sensing activity monitors, and sensor packages built into flight suits and helmets.

The advantages of smart sensors that have been noted for other applications also pertain to biomedical monitoring systems. The promises to increase reliability, flexibility (i.e., user programmability), and data storage efficiency while decreasing size, weight, power requirements, and obtrusiveness can best be observed at the system level, since they may not be applicable to individual sensors. Moreover, aside from such quantitative enhancements, smart sensors enable new possibilities in the study of dynamic manmachine interactions. For example, by providing capabilities for real-time data processing and artificial intelligence (AI), this technology will support studies of adaptive function allocation between man and machine and enable studies of human supervisory control of semi-automated systems. Wireless physiological monitoring systems may be possible, with the circuitry for amplification, signal conditioning, and telemetry being integrated into each individual sensor.

Particularly in an environment like SSF for which the data processing infrastructure is still being designed, it is advisable to take a systems perspective in considering the implementation of smart sensor technology. Ideally, I/O standards should be established so that

sensor systems can be interchangeably plugged into the infrastructure, allowing data from different sensor/transducers to be networked, made available to common sensor fusion algorithms, and archived efficiently. The electrical distribution system of the SMART HOUSE may serve as a conceptual model in this regard, in that it provides generic ports throughout the house into which electrical, telecommunications, and data processing devices can be interchangeably connected. These devices, which themselves adhere to certain "smart system" standards, handshake with the infra-structure in a way which both identifies the device to the system and provides interconnects which enhance safety and energy efficiency.

An important part of a systems perspective is providing user-compatible means for the users, in this case the crewmembers, to interact with the information provided by smart sensors. Thus, the full potential of this information will be realized by sound human factors engineering in providing means for the crewmembers to reprogram sensors, monitor ongoing experiments, and interpret the data being produced. The microgravity environment poses special challenges in this regard. However, a number of design guidelines are available, many developed specifically by NASA for spacecraft applications. Moreover, recent developments in graphical user interfaces, "point and shoot" input devices, voice input technology, and integrated displays of system status offer means of meeting these challenges. In fact, the same microprocessor and AI technology that allows the envisioned enhancements in sensor technology can be applied to the data processing and interpretation of multi-sensor output. Such data fusion (sensor fusion) approaches as automated pattern recognition, adaptive neural networks, simulation, and analysis by synthesis, offer the prospects of considerably automating the process control of scientific experiments and the analysis of the resulting data, as well as presenting these results to the human supervisor in way that can be more readily encoded and understood.

Integrated sensors and microsystems have widespread biomedical applications. Some of the major findings in silicon sensors are as follows:

- Silicon sensors are a prime technology for development of integrated smart sensors due to similar processing technologies shared with microelectronics.

- The parallel development of low-cost, high-volume silicon sensors and single chip microprocessors (and related devices) offer an opportunity for producing programmable low cost smart sensors.

- There is intense activity in the U.S., Europe, and Japan for development of new sensor technologies using advanced micromachining and processing techniques.

- Self-testing and self-calibration are two primary features being designed into silicon smart sensors to ensure performance and reliability.

- Silicon sensors have developed faster than other technologies due to the close relationship with silicon microelectronics which has research funding estimated at $250B annually. The effect of this shared experience in processing is illustrated by the cost/learning curve of silicon pressure sensors over the past 10 years.

- Silicon has mechanical properties which make it an ideal material for micro-structure fabrication.

- Smart sensors, providing self-test, calibration, control, communication and decision making capabilities, are going to be prevalent in new products in the 1990s.

- Catheter-based sensors are a trend in the medical applications of advanced sensors. Their impact will be determined by cost effectiveness.

The challenge for applications of chemical and bioanalytical sensor research in Space Station applications is to provide systems with long shelf life, low maintenance, high sensitivity, and very high reliability for the detection of a wide variety of analytes including gases, trace metals, electrolytes, metabolites, and proteins. No single currently available technology offers a solution to this problem. When asked to select currently available technologies for this application, the majority of a panel of experts recommended reduction and miniaturization of currently available intermittent *in vitro* analysis systems as the only feasible approach. These systems may however present limitations with respect to required manpower and functionality in the space environment.

The development status of selected chemical and bioanalytical sensors can be summarized as shown in Table 5.0-1.

Table 5.0-1 Development Status of Selected Bioanalytical Sensors

Product Phase
(1992+)
* pH electrodes
* Oxygen electrodes
* Ion selective electrodes
  - sodium
  - chloride
  - potassium
  - calcium
  - ammonia
  - nitrates
* *In Vitro* Diagnostic Systems
  - glucose
  - alkaline phosphatase
* Bedside Analyzers
  - blood gases
  - electrolytes
    - sodium
    - chloride
    - potassium
    - calcium
* Intra-circulatory Sensors
  - oxygen electrodes
  - miniature Ph electrodes
  - pH optrodes
  - carbon dioxide optrodes

Emerging Phase
(1995+)
* oxygen optrodes
* blood gas optrodes
* hand-held blood gas analyzers
* non-invasive glucose analyzers

Research and Development Phase
(1997+)
* enzyme electrodes
* IR and NIR spectral analysis
  - *ex situ*
  - *in situ*
* semiconductor sensors
* electroactive polymer electrodes
* surface plasmon res. devices
* electroacoustic devices
* refractometry devices

Market forces which are currently promoting the development of sensors for medical, environmental, and bioprocessing applications will stimulate the most likely sources for sensors in space. However, the labor, power, bulk, reliability, shelf life, and maintenance requirements of these sensor systems are likely to be less stringent than those for space applications.

The development of custom synthesized chemical reactants for sensors is a new and important field of investigation. These materials, modelled after less durable natural compounds, may provide the stability and reliability required for sensors systems in space. Artificial enzymes, electroactive polymers, and synthetic bioreceptors are three examples of these efforts which may provide future solutions.

Direct spectroscopic measurements of chemical species in subject systems are the most likely future sources for sensor technologies. The chemical simplicity and potential for analytical diversity of these techniques will provide the ability to solve many analytical problems with a minimum of developmental effort. IR and NIR analysis will probably be the first available technology for the widest range of analytical problems. The potential advantage of using the space environment for mass spectroscopic systems is also appealing. The feasibility of using nuclear magnetic resonance (NMR) to make measurements in space is also being studied.

Table 5.0-2 is a summary of advanced sensor research activities in bioanalytical sensing.

Table 5.0-2 Selected Centers of Competence and Research Programs in Bioanalytical Sensing

Academic Programs		
U. of Iowa	ISE's, IR & NIR	Dr. Mark Arnold
VA Commonwealth U.	Optical immunosensors	Dr. W. Greg Miller
State U. of New York	Bioreceptor sensors	Dr. George Ayoub
Iowa State U.	Photochemical instrumentation	Dr. Glenn Bastiaans
U. of Texas	Electroactive polymers	Dr. Adam Heller
U. of Maryland	Pharmacological sensors	Prof. Mohyee Eldefrawi
U. of New Orleans	Electroacoustic sensors	Prof. George Guilbault
U. of Cincinnati	Bioactive polymers	Dr. William Heineman
Rensselear Polytech. Inst.	Whole cell biosensors	Dr. Charles Keese
Washington State U.	Neural cell sensors	Dr. Bernard Van Wie
University of Kansas	Immunosensors	Dr. George Wilson
Tufts University	Polymers for optical sensors	Dr. David Walt
Colorado State U.	Electroactive polymers	Dr. Charles Martin
U. of New Hampshire	Refractometric sensors	Dr. Rudy Seitz
U. of Michigan	Electrochemical sensors	Dr. Mark Meyerhoff
U. of New Mexico	IR & NIR analysis	Dr. Reese Robinson
U. of Pittsburgh	Immunosensors	Dr. Jerome Schultz
U. of Washington	Phase transition optrode	Dr. Paul Yager

Corporate R&D Programs	
Abbott Laboratories	Blood gas optrodes, IR & NIR analysis
The BOC Group	Gas electrodes and optrodes
ORD Corporation	Ex situ optrodes
E.I. Du Pont	Electroactive polymers, ex situ diagnostics systems
Optex Biomedical, Inc.	Blood gas optrodes
Puritan-Bennet FOxS Laboratories	Blood gas optrodes
Orion Research	Enzyme electrodes
Roche Diagnostic Systems	Immunochemical sensors
Pharmacia Biosensors	Immunochemical sensors
Hewlett-Packard Laboratories	IR & NIR analysis
Covalent Associates, Inc.	Bioactive materials
Ingold Electrodes, Inc.	Enzyme electrodes
Miles Inc.	Enzyme electrodes
Allied Signal, Inc.	Bioprocess monitoring
Ciba-Corning Diagnostics	Blood electrolytes & metabolites
Eastman Kodak	In vitro diagnostic systems
PPG Sensors	Hand held blood gas analysis
General Motors Corp.	IR & NIR analysis, carbon monoxide

Governmental Programs		
Sandia National Labs	IR & NIR analysis	Dr. David Haaland
Sandia National Labs	Semiconductor microsensors	Dr. R.C. Hughes
National Institute of Standards and Technology	Immunosensors	Dr. Anne Plant
NASA-Ames Research Center	Life signs detection	Dr. Sjeord Bonting
Naval Research Lab	Life support systems	Dr. Frances Ligler
U.S. Dept. of Agriculture	IR & NIR analysis	Dr. Marjorie Medina
Lawrence Livermore National Laboratory	Environmental monitoring	Dr. Allen Northrup
Foreign Programs		
Imperial College, London, UK	Immunosensors	Dr. Gurteet Athwal
U. of Cambridge, Cambridge, UK	Surface plasmon resonance	Dr. David Cullen
Institute fur Spektrochemie, Dortmund, Ger.	IR & NIR analysis	Dr. H. Michael Heise
National Research Council, Montreal, Canada	Biosensors	Dr. John Luong
Fisons Applied Technology, Cambridge, UK	Surface plasmon resonance	Dr. Denise Pollard-Knight
Cranfield Institute of Technology, Cranfield, UK	Immunochemical sensors	Dr. Anthony Turner

Table 5.0-3 contains a list of centers for advanced sensor systems with potential applications on NASA SSF programs.

Table 5.0-3 Centers of Advanced Sensor Technology

Temperature Pill	
	Johns Hopkins Applied Physics Lab, Laurel, MD
	Konigsberg Instruments Inc., Pasadena, CA
	Lovelace Scientific Resources, Houston, TX
Biotelemetry systems	
	Konigsberg Instruments Inc., Pasadena, CA
	UFI, Morro Bay, CA
	Wilton Systems Inc., CT
Activity Loggers	
	Ambulatory Monitoring Inc., Ardsley, NY
Physiological Ambulatory Monitoring Equipment	
	Stuart Medical Inc., Owings Mills, MD
	Ambulatory Monitoring, Inc., Ardsley, NY
	Oxford Instruments Inc., Clearwater, FL
	Delmar Avionics Inc., Irvine, CA
	Vitalog Inc., Palo Alto, CA
Solid State Data Recorders	
	GMS Engineering Corporation, Columbia, MD
	Navy Air Test Center, Patuxent River, MD
	Vitalog Inc., Palo Alto, CA
Electrical System with Interchangeable Outlets and Handshaking with External Devices	
	SMART HOUSE Inc., Upper Marlboro, MD

Physiological Sensors Integrated into Flight Gear

NASA Ames Research Center, Moffett Field, CA
Brooks Air Force Base, San Antonio, TX
Neuropsychology Lab, UCLA, Los Angeles, CA

Continuous Non-Invasive Blood Pressure Sensors

Southwest Research Institute, San Antonio, TX
Stanford Research Institute, Palo Alto, CA
Cortronics, Ronkonkoma, NY
Ohmeda, Madison, WI

Piezoresistive Pressure Sensors

Motorola, Phoenix, AZ
Nova Sensor, Fremont, CA
IC Sensors, Mipitas, CA
SenSym, Sunnyvale, CA

Capacitive Pressure Sensors

Wen Ko, Case Western Reserve Univ.
Kavlico, Moorpark, CA

Fiberoptic Sensors

Fiberoptic Sensor Technologies, Ann Arbor, MI
Camino Labs, San Diego, CA
Ladd Research Industries

Accelerometers

Analog Devices, Norwood, MA
Nova Sensor, Fremont, CA
IC Sensors, Milpitas, CA
Motorola, Phoenix, AZ

SUMMARY CONCLUSIONS

- Smart sensor **technology development** has outpaced smart sensor **utilization**
- Silicon sensor technology is ideal for the development of integrated smart sensors due to the processing techniques which are shared with micro-electronics circuitry
- Major commercial applications of smart sensors over traditional sensors involve requirements for:
  - long-term performance without maintenance
  - greater accuracy
- Smart sensors providing self-test, calibration, control, communication, and decision-making capabilities are going to be prevalent in the 1990s
- Whether a device is a smart sensor or an instrument depends more upon its role within a system, its dependence on other physical devices, the level of integration within the sensor package, and its mode of communications with the user
- A systems design approach (involving communications, software, display, human factors, etc.) must be used to maximize the effectiveness of smart sensors
- Digitally programmable signal processing in external signal conditioners is a good first step toward the integration of advanced sensors into future Life Sciences research instrumentation
- Increased data rates and intensive experiment control requirements associated with SSF place increased demands on the human interface required for control of advanced sensors and their data
- Smart sensors have numerous ideal applications on SSF and future space biomedical research programs
- Direct digital signal processing holds a promise for increased system **flexibility** in the far-term period

RECOMMENDATIONS

- Develop an expertise in "universal" programmable signal conditioners
- Develop smart sensor platform concepts to provide a wide range of measurement and control functions, depending upon the sensor installed
- Adopt a systems approach to biomedical data acquisition that incorporates the features of the Medical Information Bus (MIB)
- Follow the development of commercial-off-the-shelf versions of ambulatory monitors for early adoption of cost-effective smart sensors
- Utilize RF and IR telemetry to monitor and control instrumentation containing smart sensors
- Assess the impact of smart sensors on the feasibility of expanding inflight biomedical data analysis
- Consider the use of "smart" cards for numerous purposess on SSF
- Develop a microminiature sensor capability to support future Life Sciences research which will expand at the cellular level, where the emphasis shifts toward increased requirements for biochemical measurements

APPENDICES

A. References

B. Executive Summary, Sensors 2000! Program

C. Sensor Tutorials

D. Product Data Sheets and Brochures

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APPENDIX A--REFERENCES

AAMI. (1988). Human Factors Engineering Guidelines and Preferred Practices in the Design of Medical Devices. Association for the Advancement of Medical Instrumentation.

Allen, HV; Terry, SC; De Bruin, DW. Accelerometer systems with self-testable features. Sensors and Actuators 1989; 20: 153-160.

Anderson, NS; Olson, JR (Eds.) (1985). Methods for designing software to fit human needs and capabilities. Washington, D.C.: National Academy Press.

Andrade, JD; Lin, JN; Hlady, V; Christensen, D; Kopecek, J (Dep. Bioeng., Univ. Utah, Salt Lake City, UT, USA). Immunosensors: remaining problems in the development of remote continuous, multichannel devices. In: Biosensor Technology, [Proceedings of the International Symposium] 1989, 219-39, (Eng). Buck, RP (ed). Dekker, NY, NY.

Ansermet, S; Otter, D; Craddock, RW; Dancaster, JL. Cooperative development of a piezoresistive pressure sensor with integrated signal conditioning for automotive and industrial applications. Sensors and Actuators 1990; A21-A23: 79-83.

Austin, FH, et al. (1967). Aerospace Medicine, 38, 593-596.

Barth, PW. Silicon fusion bonding for fabrication of sensors, actuators and microstructures. Sensors and Actuators 1990; A21-A23: 919-926.

Beringer, D; Scott, J (1985). The long-range light pen as a head-based user-computer interface: Head-mounted "sights" versus head positioning for computer access by the disabled. Proceedings of the Human Factors Society -- 29th Annual Meeting, Baltimore, Md., 114-118.

Bernard, Steven M; Walt, David R (Dep. Chem., Tufts Univ., Medford, MA 02155, USA). Chemical sensors based on controlled-release polymer systems. Science (Washington, D. C., 1883-), 251(4996), 927-9 (Eng) 1991.

Boff, KR; Lincoln, JE (1988). Engineering Data Compendium: Human Perception and Performance, Wright-Patterson Air Force Base, OH: CSERIAC.

Boff, KR; Kaufman, L. & Thomas, JP, (Eds.) (1986). Handbook of Perception and Human Performance, New York: Wiley Interscience.

Brems, DJ & Whitten, WB (1987). Learning and preference for icon-based interface. Proceedings of the Human Factors Society -- 31st Annual Meeting, New York City, 125--129.

Bryzek, J et al. "Silicon Sensors and Microstructures in the Health Care Industry," Proceedings of Sensor Expo International, Cleveland, OH, pp. 303A1-303A10, Sept 1989.

Bryzek, J et al. Silicon Sensors and Microstructures. Chapter 8, NovaSensor, Fremont, CA. 1990.

Budimir, MV; Sak-Bosnar, M.; Kovac, S.; Duic, LJ. (Fac. Agric., Univ. Osijek, Osijek 54000, Yugoslavia). The application of zero-current potentiometry in chemical synthesis and characterization of polypyrrole using electrochemical sensors. Synth. Met., 39(3), 359-65 (Eng) 1991.

Calhoun, GL; Arbak, CJ; & Boff, KR. (1984). Eye-controlled switching for crew station design. Proceedings of the Human Factors Society -- 28th Annual Meeting, San Antonio, TX., 258-262.

Call, DW; Kelly, DM; Robertson, DG (1982). A self-contained, man-bourne biomedical instrumentation system in the flight testing of naval weapons systems. In M. L. Frazier & R. B. Crombie (Eds.), Proceedings of the Workshop on Flight Testing to Identify Pilot Workload and Pilot Dynamics, Edwards Air Force Base: AFTEC-TR-82-5, 318-321.

Call, DW; Miller, WF. (1987). Aviation, Space and Environmental Medicine, 58(5): 508 (Abstract).

Cammann, Karl; Lemke, Udo; Rohen, Anja; Sander, Juergen; Wilken, Hildegard; Winter, Babette (Inst. Chemo-Biosensorik, Univ. Muenster, Muenster S-4400, Fed. Rep. Ger.). Chemical and biological sensors. Principles and applications. Angew. Chem., 103(5), 519-41 (See also Angew. Chem., Int. Ed. Engl., 1991), 30(5), 516-39) (Ger) 1991.

Card, S.; Moran, T.; & Newell, A. (1983). The Psychology of Human Computer Interaction. Hillsdale, N.J.: Lawrence Erlbaum.

Carl, RT. (Nicolet Spectrosc. Res. Cent., Nicolet Instrument G.m.b.H., Offenbach/Main D-6050, Fed. Rep. Germany). Quantification of the fat content of milk using a partial-least-squares method of data analysis in the near infrared. Fresenius. J. Anal. Chem., 339(2), 70-1 (Eng) 1991.

Carroll, JM & Olson, JR, (Eds.). (1987). Mental Models in Human-Computer Interaction. Washington, D. C.: National Academy Press.

Cho, ST and Wise, KD. A self testing ultrasensitive silicon microflow sensor. Sensors Expo Proc. 1991: 208B-1 through 208B-4.

Coats, MR et al. "Smart Transmitters in Temperature Measurement," Proceedings of Sensor Expo International, Cleveland, OH, pp. 204C1-104C8, Sept. 1989.

Core, T; Payne, RS; Quinn, D; Sherman, S; and Tsang, WK. Integrated, complete, affordable accelerometer for airbag applications. Sensors Expo Proc. 1991: 204B-1 through 204B-4.

Crary, SB; Baer, WG; Cowles, JC and Wise, KD. Digital compensation of high-performance silicon pressure transducers. Sensors and Actuators 1990; A21-A23: 70-72.

Cunningham, S and Porter, AL: Communication networks: A dozen ways they'll change our lives, The Futurist, Jan./Feb., 1992, pp 19-22.

Davies, SE; Bury, KF; & Darnell, MJ. (1985). An experimental comparison of a windowed vs. a non-windowed operating system environment. Proceedings of the Human Factors Society -- 29th Annual Meeting, Baltimore, MD., 250-254.

Elkind, JI; Card, SK; Hochberg, J., & Huey, BM. (Eds.) (1989). Human Performance Models for Computer-Aided Engineering. Washington, D. C.: National Academy Press.

Erlandsen, J. (1991). DSP--digital signal processing--maximizes cath-lab flexibility. Medical Electronics, Oct., 104-105.

Favenec, JM, "Smart Sensors in Industry," Journal of Physics E: Scientific Instrumentation, Vol. 20, 1987. p. 1088.

Fitzgerald, GH. "The Development of a Two-Dimensional Multielectrode array for Visual Preception Research in the Mammalian Brain," MS Thesis, AFIT, WPAB, OH, Dec. 1980 (ADA100763).

Foley, JD; Gibbs, C., Kim, WC., & Kovacevic, S. (1988). A knowledge-based user interface management system. Human Factors in Computing Systems, Proceedings of CHI'88.

Foley, JD; Kim, WC; Kovacevic, S; & Murray, K. (1989). Defining interfaces at a high level of abstraction. IEEE Software, 25-32.

Foster, Mark, "Data Management System Payload Study - Payload-to-DMS Interfaces," Presentation to SSSAAS, July 22, 1991.

Frank, R. Pressure sensors merge micromachining and microelectronics. Sensors and Actuators A. 1991; 28: 93-103.

Frost, JD Jr. NASA Technical Report, NTIS No. N75-27747/5.

Gakkestad, J and Jakobsen, H. A front end CMOS circuit for a full-bridge piezoresistive pressure sensor. Sensors and Actuators A. 1991; 25-27: 859-863.

German, GW. "A Cantically Implantable Multielectrode Array for Investigating the Mammalian Visual System," MS Thesis, AFIT WPAFB, OH, Dec. 1981.

Giachino, JM. Smart sensors. Sensors and Actuators 1986; 10: 239-248.

Gilmore, VE. (August, 1988). Tomorrow's house. Popular Science, 41-46, 82-83.

Goodenough, F. "Fast 24-bit ADC converter handles dc-to-410 Hz input signals," Electronic Design, Oct. 1987.

Gould Electronics. Enhanced Delta Modulation Encoder (EDME), Gould Technical Note 0141A0860.

Grace, R. Enabling sensor technologies for automotive applications review and overview. Sensors Expo Proc. 1991: 106C-1 through 106C-10.

Grace, RH. "Silicon Sensors and Microstructures Technology, Trends and Applications, WESCON 1987 Professional Program Papers, WESCON 87 Conference, 18-20 November, 1987, San Francisco, CA. pp. 24/0, 1-9.

Hanneborg, A and Ohlckers, P. A capacitive silicon pressure sensor with low TCO and high long-term stability. Sensors and Actuators 1990; A21-A23: 151-154.

Hartl, A; Franz, J; Vogt, R. Application of an electronic accelerometer for a single point sensing airbag electronic control unit. Sensors Expo Proc. 1991: 204C-1 through 204C-12.

Haskard, MR. An experiment in smart sensor design. Sensors and Actuators A. 1990; 24: 163-169.

Helander, M. (Ed.) (1988). Handbook of human-computer interaction. Amsterdam: North-Holland.

Hensley, RW and Denton, DC. "The First Cortical Implant of a Semiconductor Multielectrode Array: Electrode Development and Data Collection," MS Thesis, AFIT, Dec. 1982.

Hesketh, PJ and Maclay, GJ; Micromachining for chemical and biosensors. Sensors Expo Proc. Oct. 1991: 102C-1 through 102C-6.

Hesketh, PJ; Madou, MJ; Otagawa, T; Joseph, J and Nikolchev, JN. The application of micromachining to biomedical sensors. Sensors Expo Proc. 1990: 305D-1 through 305D-8.

Hines, JW: Sensors 2000! Program: Status and Accomplishments Fiscal Year 1989, NASA Ames Research Center, Moffett Field, CA, Dec, 1989.

Horrigan, DJ, Jr., et al. (1970). Naval Air Test Center Report, TR-70-61(U).

Howe, RT; Muller, RS; Gabriel, KJ; and Trimmer, WS. (1990). Silicon micromechanics: Sensors and actuators on a chip. IEEE Spectrum, 29-35.

Hughes, RC; Ricco, AJ; Butler, MA.; Martin, SJ. (Microsensor Div., Sandia Nat. Lab., Albuquerque, NM). Chemical microsensors. Science, 254, 74-79, (Eng.) 1991.

Hurley, WD & Sibert, JL. (1989). Modeling user interface-application interactions. IEEE Software, 71-77.

Iavecchia, JH.; Iavecchia, HP; & Roscoe, SN (1988). Eye accommodation to heads-up virtual images. Human Factors, 30, 689-702.

Iida, Katsumi; Hidoh, Osamu; Fukami, Junichi; Kajiwara, Masahiro (Dep. Med. Chem., Meiji Coll. Pharm., Tanashi 188, Japan). Direct monitoring by carbon-13 nuclear magneticresonance spectroscopy of the metabolism and metabolic rate of carbon-13-labeled compounds in vivo. Chem. Pharm. Bull., 39(1), 210-13 (Eng) 1991.

Inagaki, N; Tasaka, S; Kobayashi, M. (Fac. Eng., Shizuoka Univ., Hamamatsu 432, Japan). Application of plasma films prepared from manganese acetylacetonate for CO gas sensor device. Polym. Bull. (Berlin), 25(2), 273-8 (Eng) 1991.

Janata, Jiri. (Cent. Sensor Technol., Univ. Utah, Salt Lake City, UT) Solid state potentiometric sensors. In: Biosensor Technology, [Proceedings of the International Symposium] 1989, 219-39, (Eng). Buck, R. P. (ed). Dekker, NY, NY.

Janata, Jiri; Principles of chemical sensors. (Cent. Sensor Technol., Univ. Utah, Salt Lake City, UT). Plenum Press, 1990.

Johns Hopkins University Applied Physics Laboratory. (1991). Proceedings of the Data Fusion Symposium.

Jones, K. Caring sensors. Sensor Review July 1990: 111-114.

Jost, Jean Pierre; Munch, Olivier; Andersson, Thomas (Friedrich Miescher Inst., Basel CH-4002, Switz.). Study of protein-DNA interactions by surface plasmon resonance (real time kinetics). Nucleic Acids Res., 19(10), 2788 (Eng) 1991.

Kieras, DE; & Polson, PG. (1985). An approach to the formal analysis of user complexity. International Journal of Man-Machine Studies, 22, 365-394.

Kjesmo, A; Hanneborg, A; Gakkestad, J; and Von Der Lippe, H. A CMOS circuit for a capacitive pressure sensor. Sensors and Actuators 1990; A21-A23: 102-107.

Klainer, SM; Thomas, JR; Dandge, DK; Frank, CA; Butler, MS; Arman, H; Goswami, K (FiberChem, Inc., Las Vegas, NV 89119, USA). *In-situ* monitoring for hydrocarbons using fiber optic chemical sensors (FOCS). Proc. SPIE-Int. Soc. Opt. Eng., 1434 (Environ. Sens. Combust. Diagn.), 119-26 (Eng) 1991.

Ko, Wen H. Capacitive pressure sensors. Sensors Expo Proc. 1991: 208C-1 through 208C-4.

Ko, WH; and Newann, MR. "Implant Biotelemetry and Microelectronics," Science, 156: 351-359, 21 April 1967.

Kooyman, Rob PH; Lenferink, Aufried TM; Eenink, Rob G; Greve, Jan (Dep. Appl. Phys., Twente Univ., Enschede 7500 AE, Neth.). Vibrating mirror surface plasmon resonance immunosensor. Anal. Chem., 63(1), 83-5 (Eng) 1991.

Kroschel, K and Wernz, A. Sensor fault detection and localization using decorrelation methods. Sensors and Actuators A. 1991; 25-27: 43-50.

Lee, Peter S; Majkowski, Richard F; Perry, Thomas A. (Dep. Physics, General Motors Research Lab., Warren, MC). Carbon dioxide laser spectroscopy for isotope analysis - detection of isotopic carbon monoxide in exhaled breath. IEEE Trans. Biomed. Eng. 38(10), 966-972, (Eng.) 1991.

Leung, Larry K.; Komplin, Norma J.; Ellis, Arthur B.; Tabatabaie, Ned (Dep. Chem., Univ. Wisconsin, Madison, WI 53706, USA). Photoluminescence studies of silver-exchanged cadmium selenide crystals: modification of a chemical sensor for aniline derivatives by heterojunction formation. J. Phys. Chem., 95(15), 5918-24 (Eng) 1991.

Lewis, CE, et al. (1967). Aerospace Medicine, 38, 581-592.

Lieberman, Robert A; Wlodarczyk, Marek T; Editors (USA). Proceedings of SPIE-The International Society for Optical Engineering, Vol. 1368: Chemical, Biochemical, and Environmental Fiber Sensors II, 19-21 September 1990, San Jose, California. (SPIE-The International Society for Optical Engineering: Bellingham, Wash.), 274 pp. (Eng) 1991.

Mallon, Jr JR, Pourahmadi, F; Peterson, K; Barth, P; Vermeculon, T; and Bryzek, J. Low pressure sensors employing bossed diaphragms and precision etch stopping. Sensors and Actuators 1990; A21-A23: 89-95.

McCauley, ME. (1984). Human factors in voice technology. In F. A. Muckler (Ed.) Human Factors Review: 1984, Santa Monica, CA: The Human Factors Society.

McClelland, S. Europe heads for the smarter sensor. Sensor Review 1989; Vol 9(4): 207-208.

Medical Information Bus Draft IEEE Standard 1073.1, Revision J, August 1988, p.7.

Medical Information Bus-Bedside Communications Subnet, P. 1073.2, Draft 13, February 21, 1990. pp. 91-96.

MIB MDDL LANGUAGE SPECIFICATION, Version 1.05PD (Preliminary Draft), September 23, 1988. pp. 6-8.

MIL-STD-1472D. (1986). Human Engineering Design Criteria for Military Systems, Equipment, and Facilities, Washington, DC: Department of Defense.

MIRC, Fiber Optic Sensor Markets in Medicine; July 1988: 170 pgs.

Moore, T. (November, 1986). The SMART HOUSE: Wired for the electronic age. EPRI Journal, Palo Alto, CA: Electric Power Research Institute.

Murray, Royce W; Dessey, Raymond, E; Janata, Jiri; Seitz, W Rudolf (ed.). Chemical sensor and microinstrumentation. ACS Symposium Series #403, American Chemical Society, (Eng) 1989.

NASA Man-System Integration Standards. (1987). Washington, DC: NASA STD-3000.

Neuman, MR; Brill, AB; Gibbons, DF; Greatbach, WF; Mates, R and Rushmer, RF (1989). Research directions in biomedical engineering, IEEE Engineering in Medicine and Biology Magazine, 18-25.

Norman, J & Ehrlich, S. (1986). Visual accommodation and virtual image displays: Target detection and recognition. Human Factors, 28, 135-151.

NUREG-0700. (1981). Human Factors Guidelines for Detailed Control Room Design Reviews, Washington, DC: Nuclear Regulatory Commission.

Olsen, D, Buxton, W, Ehrich, R, Kasik, D, Rhyne, J, and Sibert, J. (1984). A context for user interface management. IEEE Computer Graphics and Applications, 4, 33-42.

Ormond, T: Inexpensive sensors provide precision. EDN, Nov. 7, 1991, pp.99-104.

Peterson, K; Brown, J; Vermeculon, T; Barth, P; Mallon Jr, J and Bryzek, J. Ultra-stable, high-temperature pressure sensors using silicon fusion bonding. Sensors and Actuators 1990; A21-A23: 96-101.

Portais, JC; Pianet, I; Allard, M; Merle, M; Raffard, G; Kien, P; Biran, M; Labouesse, J; Caille, JM; Canioni, P. (IBCN, Univ. Bordeaux II, Bordeaux 33077, Fr.). Magnetic resonance spectroscopy and metabolism. Applications of proton and carbon-13 NMR to the study of glutamate metabolism in cultured glial cells and human brain *in vivo*. Biochimie, 73(1), 93-7 (Eng) 1991.

Pourahmadi, F; Barth, P; and Peterson, K. Modeling of thermal and mechanical stresses in silicon microstructures. Sensors and Actuators 1990; A21-A23: 850-855.

Ramsey III, M. Blood pressure monitoring: automated oscillometric devices. Journal of Clinical Monitoring January 1991; Vol. 7: 56-67.

Rhyne, JR & Wolf, CG (1987). Gestural interfaces for information processing applications. Proceedings of the Second International Conference on Human-Computer Interaction.

Roman, J; Older, H; & Jones, WL. (1967). Flight Research Program, VII: Medical monitoring of Navy carrier pilots in combat. Aerospace Medicine, 38, 133-139.

Sansen, W. et al. "A Smart Sensor for Biomedical Applications," Proceedings of the Annual International Conference of the IEEE Engineering in Medicine and Biology Society, Seattle, WA., ll:1088-1089, Nov. 1989.

Schultz, Jerome, S (Center for Biotechnology and Bioengineering, Univ. Pittsburgh, Pittsburgh, PA, USA). Biosensors. Sci. American, 64-69, (Eng.) 1991.

Segebarth, C; Grivegnee, AR; Longo, R; Luyten, PR; Den Hollander, JA. (Unite Reson. Magn. Nucl., Hop. Erasme, Brussels B1070, Belg.). In vivo monitoring of fructose metabolism in the human liver by means of phosphorus-31 magnetic resonance spectroscopy. Biochimie, 73(1), 105-8 (Eng) 1991.

Seidman, S: Alternate card technologies, Memory Card Systems and Design, Nov./Dec., 1991, pp. 38-40.

Sheridan, TB & Hennessy, RT. (1984). Research and modeling of supervisory control behavior: Report of a workshop. Washington, DC: National Academy Press.

Shneiderman, B. (1987). Designing the user interface: Strategies for effective human--computer interaction. Reading, MA: Addison-Wesley Publishing Co.

SMART HOUSE. (1988). System Level Specification, SH-1200 -- Human Factors Guidelines, Upper Marlboro, MD: SMART HOUSE, Inc.

SMART HOUSE. (1988). SMART HOUSE electrical systems: An introduction, Upper Marlboro, MD: SMART HOUSE, Inc.

Smith, ML and Abdelrahman, M. "The Impact of AI on Sensing Technology," Sensors, Sept. 1991, pp 16-21.

Smith, SL & Mosier, JN. (1986). Guidelines for designing user interface software. U. S. Air Force Electronic Systems Division, Report ESD-TR-86-278.

Stenberg, Esa; Persson, Bjoern; Roos, Haakan; Urbaniczky, Csaba (Pharm. Biosens. AG, Uppsala S-751 82, Swed.). Quantitative determination of surface concentration of protein with surface plasmon resonance using radiolabeled proteins. J. Colloid Interface Sci., 143(2), 513-26 (Eng) 1991.

Sundgren, H; Winquist, F; Lukkari, I; Lundstroem, I. (Dep. Phys. Meas. Technol., Linkoeping Inst. Technol., Linkoeping S-581 83, Swed.). Artificial neural networks and gas sensor arrays: quantification of individual components in a gas mixture. Meas. Sci. Technol., 2(5), 464-9 (Eng) 1991.

Talmadge, R; Appley, KE. A programmable transducer microchip. Sensors, June 1991: 14-19.

Tatman, JA. "A Two-Dimensional Multielectrode Microprobe for the Visual Cortex," MS Thesis, AFIT, Wright-Patterson AFB, OH. Dec 1979 (ADA080378).

Taylor, Gerald, "The Space Station Freedom Biomedical Monitoring and Countermeasures (BMAC) Project - Inflight Equipment Derivation," February, 1990.

Terry, S; Eckerle, JS; Kornbluh, RD; Low, T; and Ablow, CM. Silicon pressure transducer arrays for blood-pressure measurement. Sensors and Actuators 1990; A21-A23: 1070-1079.

Tullis, TS. (1986). A system for evaluating screen formats. Proceedings of the Human Factors Society -- 30th Annual Meeting, Dayton, OH., 1216-1220.

Turner, Anthony, J; Karube, I, Wilson, George, S. (ed.). Biosensors: fundamentals and applications. Oxford University Press, 1987.

Voecks, GE & Seshan, PK. (1991). Real-time *in situ* sensors and control integration for life support systems. Presented at the 21st International Conference on Environmental Systems, July, 1991, SAE Technical Paper 911356.

Ward, KJ; Haaland, DM; Robinson, MR; Eaton, RP. (Sandia Nat. Lab., Albuquerque, NM). Quantitative infrared spectroscopy of glucose in blood using partial least squares analyses. Proc. SPIE Vol. 1145, Fourier Transform Spectroscopy, 607-608, (eng.) 1989.

Weiss, R: 32-bit floating-point DSP processors, EDN, Nov. 7, 1991, pp.127-146.

Whitelaw, V., "Review of SSF Data Management and Communication Systems," Presentation to SSSAAS, July 22, 1991.

Wilson, P. Tutorial: Silicon micro-machining. Sensor Review July 1991: 178-181.

Wise, KD and Najafi, K: Microfabrication techniques for integrated sensors and microsystems, Science, Nov. 29, 1991, pp. 1335-1342.

Yang, Fuzi; Bradberry, GW; Sambles, JR. (Dep. Phys., Univ. Exeter, Exeter/Devon. EX4 4QL, UK). The study of the optical properties of obliquely evaporated nickel films using IR surface plasmons. Thin Solid Films, 196(1), 35-46 (Eng) 1991.

Yun, Weijie and Howe, RT. Silicon microfabricated accelerometers: a perspective on recent developments. Sensors Expo Proc. 1991: 204A-1 through 204A-8.

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APPENDIX B--EXECUTIVE SUMMARY, SENSORS 2000! PROGRAM

I. EXECUTIVE SUMMARY
Sensors 2000!

Introduction

Sensors 2000! (S2K!) is a technology development Program conceived at the NASA Ames Research Center (ARC) to design, develop, and evaluate biomedical and biochemical sensor systems for application to NASA spaceflight programs. The S2K! Program was initiated in Fiscal Year (FY) '88 and is managed from the ARC Systems Engineering Division, Electronic Systems Branch. Core funding for the Program is provided by the NASA Headquarters (HQ) Advanced Technology Development (ATD) Program Office and administered by the ARC Space Life Sciences Payloads Office (SLSPO).

Program Strategy

The S2K! Program strategy for the development of advanced sensor systems was initially defined during symposium and workshop activities sponsored in FY'88 by HQ Life Sciences ATD. A symposium was held to review the preliminary S2K! Program with representatives from NASA life sciences, engineering and management groups, as well as university, industry and non-NASA government agency participants. Immediately following the symposium, a workshop was held to assess current technologies and evaluate a strategy by which the Program could conduct collaborative development of sensor technologies with industry and university groups. These events were similar to a series of Technology Assessment Workshops conducted at the NASA Johnson Spaceflight Center (JSC) in FY'87. In those forums, the ability of current emerging sensor technologies to satisfy the measurement requirements for the cardiovascular and calcium homeostasis science disciplines was assessed. The measurement requirements reviewed during the JSC workshops were for maintenance of human health and conduct of biomedical research on future spaceflight missions. The ARC events focused on sensor systems development aspects and were organized by functional sensor and bioinstrumentation technology topics covering multiple science disciplines.

The participants in these ARC FY'88 activities favorably endorsed the S2K! Program Plan but recommended additional emphasis on the following areas: i) identification of high priority biomedical measurements, ii) technology assessment and iii) identification of funding resources beyond NASA ATD funding. Based on these recommendations, plans were made to explore collaborative and resource leveraging opportunities with industry, university, and ARC flight program groups (and subsequently other Centers) and to utilize the proposed NASA HQ Science Discipline Working Groups for definition/ranking of biomedical and biochemical measurement requirements.

Scope

Since its initiation in FY'88, the S2KI Program has developed a more focused implementation plan for biomedical sensor development to facilitate the transition of sensor prototypes to ground-based and flight research programs. Briefly, the Program -

1) works with existing NASA Science Discipline Working Groups to determine measurement requirements,

2) surveys available sensor technologies for prototype candidates aided by the compilation of a sensor technology database,

3) selects specific biomedical sensors for development which satisfy the high priority measurement requirements,

4) develops prototype sensor systems to meet flight hardware requirements, and

5) conducts biosensors tests and evaluation to ascertain functional performance characteristics.

In addition to the overall management of sensor system development, the S2KI Program also -

- serves as a point-of-contact for inquiries on sensor technologies through production and dissemination of symposia reports and of papers on specific sensor technologies, and
- pursues collaborative sensor development with industry and university groups, as well as flight programs, as a means of leveraging its resources

Major Program Tasks and Accomplishments

The following specific Research and Technology Projects, originally proposed and funded by NASA HQ/ATD in FY'88, are planned to continue into FY'90 and beyond. The project summaries given below outline the direction taken by the Program and the subsequent accomplishments within each.

Biomedical Sensor Systems Development
UPN: 199-80-42-01.

Whereas biomedical or biochemical "sensor" typically refers to physiological sensors and other transducers of biological interest, the development of *sensor systems* includes the design and interaction of all components for a complete measurement configuration. Biomedical/biochemical sensor technology is but one component of the instrumentation development required. In addition, development is needed for associated signal conditioners, communication

strategies, data systems, as well as design optimization for component modularity and biocompatibility. The S2KI Program assesses all of these engineering aspects as it coordinates the development of sensor systems for life science experiments using primarily, rodent, squirrel and rhesus monkeys, as well as some human subjects. Once sensors and technologies are developed, rigorous testing and evaluation are required to determine and validate performance capabilities and hardware readiness levels.

Development of sensor systems has been initiated to satisfy the required science measurements projected for near-term, mid-term and far-term flight missions. Currently available sensor technologies can satisfy many of the basic near-term measurement requirements as identified by various science disciplines, but the majority of existing technologies require additional development prior to direct application to flight systems. Work has begun to improve several biopotential, hemodynamic, imaging and biochemical sensor systems for this purpose.

To facilitate this development process, the Program is compiling a Macintosh-compatible database of state-of-the art sensor technologies. This repository of available technologies will be surveyed for applicability to sensor systems development. The S2KI Program also intends to assemble and implement a "clean room" laboratory designed for the contaminant-free assembly, calibration and testing of miniaturized biomedical/biochemical sensor instrumentation. A biomedical engineer and a biomedical test/systems engineer have recently joined the S2KI staff, thereby broadening the ability of the Program to support this development process.

The S2KI Program is presently a participating member of the NASA ARC Rhesus Research Facility (RRF) team with responsibility for development of sensors, biotelemetry and related instrumentation subsystems. By direct transition of available sensor systems, S2KI will support near-term demonstrations for planned RRF flight missions and supporting ground-based studies. Prototype development of two mid-term development biomedical sensor systems are scheduled for demonstration in FY'90. Included is an EMG/tendon force sensor system for muscle physiology measurements and an improved cardiovascular sensor system resulting from a S2KI collaboration with the Laboratory for Aeromedical Cardiovascular Research (LACR) at Brooks Air Force Base.

The S2KI Program is also participant in other flight programs including Pathfinder Advanced Life Support, the 1.8 M Centrifuge Team and the U.S./U.S.S.R. Cosmos Biosatellite project, in addition to its role in the RRF project, substantiating the Program's ability to interface with current flight programs.

Biosensor Technology Consortium/Technical Focus Groups.
UPN: 199-80-42-02

Four subtasks have been identified in this category which represent a continuation of activities initiated at ARC and JSC in FY'87/88. These activities focus on sensor technology assessment and contribute to the Program's role as a point-of-contact for biosensor activities.

Biosensor Technology Consortium - a pilot group of representatives from various ARC flight projects and related research activities have been invited to collaborate for the purpose of consolidating resources and for defining requirements and areas of emphasis for the S2KI Program. This consortium will be an ongoing means for information exchange and will provide scientific and engineering support in this area to flight projects and, if appropriate, to the proposed NASA HQ Science Discipline Working Groups as well.

"White Paper" Technology Status Reports - two position papers (on sensor-based biotelemetry systems and chemical/biological sensor technologies) have been solicited from technology experts to serve as guides for further structuring of the Program, for determining priorities, and for developing specific Technology Focus Groups.

Initiation of Technical Focus Groups - the development of bioinstrumentation and biomedical sensors should be closely coordinated with the on-going definition, refinement and interaction of requirements and technologies outlined by the science sector. Thus, Engineering Advisory and Working Groups will be focused around specific "white paper" topics and assembled to address technical issues, such as feasibility, development options, costs and schedules for sensor and sensor-systems development.

End-of-Year Status/Accomplishments Report - a report which includes a review of FY'88 activities as well as those activated in FY'89 and projected for FY '90 was prepared for distribution to interested parties. A more detailed version focusing on sensor technical descriptions and accomplishments will be issued during FY'90.

Biotelemetry Systems Development.
UPN: 199-80-42-03

Since the expected time line and duration of current and projected flight mission scenarios can range from 1-12 months or even longer, further development of biotelemetry systems has been initiated to accommodate chronic and long-duration measurements, to minimize the associated experimental artifacts, and to satisfy ground-based and Flight Program requirements. This Program task is to develop and evaluate prototype biotelemetry system configurations for use with available sensors or those currently being developed under S2KI. The eventual goal is to develop a biotelemetry system to support untethered and

unrestrained animal subjects. A variety of telemetry technologies and configurations will be evaluated in support of these requirements.

Two biotelemetry system configurations have been initially selected for development, but alternative configurations are under consideration. One system includes distributed, individually-matched transmitter-receiver pairs for transmission at video band rates. This configuration will be an external system designed with percutaneous leads or surface electrodes for use outside of the body. The other system is a multichannel, inductively-coupled digital system capable of bidirectional communications and will be initially evaluated in a backpack mode. Both of these biotelemetry systems are potentially applicable to human (external sensor) as well as animal (implanted sensor) measurement requirements. An EMG/Tendon force sensor and signal conditioner configuration are being specifically optimized for inclusion in the two-channel digital biotelemetry system. This biotelemetry system development is for applications with rhesus monkeys, and delivery of the initial prototype isscheduled for early FY'90.

A biotelemetry development and evaluation test laboratory is being established in order to investigate: i) modular packaging and improved biocompatibility as a means to increase the usefulness and longevity of systems while the potential for sensor implantation is being evaluated and ii) the application of a standardized reference test for the evaluation of existing biotelemetry systems such as those described above.

Chemical and Biological Sensor Technology Development.
UPN: 199-80-42-04

At present, no reliable sensors exist for making real-time chemical and biological measurements during flight - a measurement requirement identified within many life science disciplines. "Real-time" sensor systems, designed for the temporal measurement of ion gradients and partial pressure levels, are being developed to lessen the need to collect inflight biological samples for postflight analysis. This capability would also significantly reduce resources required from the spacecraft, flight hardware and crew.

This task will focus on the development and evaluation of chemical and biological sensors for measuring fluid and chemical concentrations, in vitro or in vivo. Methods for collaboration and co-development with industry and academia are being utilized to maximize the available resources. Feasibility studies for a silicon calcium microsensor and for a fiber-optics calcium sensor are being conducted by a private company and university, respectively. The S2KI Program has assumed the responsibility for instrumentation development, testing and evaluation, and will support sensor prototype design and development. The Program has defined procedures for the testing and evaluation of in vitro measurements utilizing developed calcium sensors. Expansion of this technology development is being projected for measurement of ions in addition to calcium and of other analytes and for environmental measurements required by Advanced Life Support Systems.

Near-Infrared Sensor Development
UPN: 199-80-52-03

Near-Infrared (NIR) Spectroscopy is a non-invasive method for monitoring physiological parameters such as blood oxygen saturation, tissue hydration, hematocrit and pH, as well as detection of anomalous physiological conditions such as breast tumor growth. The potential wide-spread application of this technology to spaceflight missions, for conduct of biomedical research and life support system development, is being actively reviewed by the S2KI Program. The acquisition of a biomedical engineer with specific expertise in NIR Spectroscopy has resulted in the investigation of potential collaborations with industry, the design of a long-term plan for NIR applications, and a technology assessment of applicable NIR technologies. Immediate plans for NIR applications include:

- Identification of measurement requirements for a NIR glucose sensor,
- conduct of a feasibility study for application of NIR to ion measurements,
- baseline determination of NIR spectra for basic components of serum, plasma and urine,
- development of measurement validation procedures, and
- development and evaluation of specific NIR prototype systems.

Sensor Applications

In addition to the ATD biosensor systems development tasks summarized above, another important aspect of the S2KI Program is designated Sensor Applications. These applications areas represent an outreach and link of the S2KI Program with other NASA programs, via leveraged arrangements, which complement specific S2KI ATD elements and subtasks. Applications consist of developing sensor prototypes for current NASA flight projects and, with the addition of this funding, significantly expanding the S2KI Program resources and capabilities. Three application areas currently being supported are:

- Flight Program Development
- Advanced Life Support Systems
- Collaborative Development Projects.

In support of Flight Programs, the S2KI Program develops sensors and bioinstrumentation for application in the RRF, tests and evaluates sensors systems prior to certification of flight hardware under development for the Cosmos Biosatellite missions, and assists in instrumentation and sensor planning for upcoming projects on Space Station Freedom. Technology developments in the area of chemical and biological sensors and in near-infrared spectroscopy are being further examined by the Program for expansion to Advanced Life Support Systems, specifically, elements of Physical/Chemical Closed Loop Life Support Systems and Monitoring and Control Instrumentation Development. The S2KI Consortium and Technical Focus Group activities, as

well as the Program's sensor database development capabilities, also contribute to Program support of the above activities.

Finally, the development of sensor systems, biotelemetry systems and chemical and biological sensors are being pursued by the S2KI Program in a number of Collaborative Development efforts. As previously noted, a Fiber-optic calcium sensor, a microsensor for calcium and other ions, and a cardiovascular sensor will result from co-development in a university consortium, with the commercial sector and with the Air Force, respectively.

Impact

It is the goal of the S2KI Program to create a process by which animal (and some human) sensor systems can be efficiently developed and transitioned to applicable NASA flight programs for making high-priority life science measurements. The efficacy of the S2KI Program lies in its ability to characterize clearly these specific science requirements, identify applicable technologies as available within NASA and the private and academic sectors, and develop emerging and existing technologies for application to spaceflight missions. The progress of the S2KI program in FY'89 has been demonstrated by the initiation of biomedical and biochemical sensor systems transition to Flight Programs, Advanced Life Support Systems development , and by the leveraging of available resources through new collaborative programs in sensor development. The success of the S2KI Program will be determined by its ability to produce state-of-the-art sensor systems applicable to flight missions in a timely and efficient manner.

Program Contacts

For additional information on the Sensors 2000! Program at ARC, interested parties are encouraged to contact the following:

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APPENDIX C--SENSOR TUTORIALS

A review of various methods for the expansion of man's senses to acquire vital life signs under a specific scenario leads to several viewpoints from which to approach the subject of biomedical instrumentation. One approach is from the physical phenomena that are to be measured, such as, bioelectric potentials, biomechanical sound waves, electromagnetic, and thermal radiations from the body. Another approach is from the measurement techniques of each physiological system. For vital life signs measurement and assessment, this reduces to the three principal systems: cardiovascular, pulmonary, and nervous. From a physician's viewpoint, the latter may be more familiar and therefore preferred; whereas, from an engineer's viewpoint, the former is more familiar and can be associated with the principles of transduction; i.e., resistive, inductive, capacitive, piezoelectric, mechanical force, pressure, etc.

This tutorial section is approached from the biomedical engineers' view; i.e., the application of engineering science and technology to form solutions to medical problems. The sections which follow, therefore, will be presented in the order of physical phenomena to be measured. This style of presentation allows measurement techniques which are applicable to more than one physiological system to be discussed in the same section. This will be the case where the same transducer may be used for more than one system (in special cases, the measurements may be simultaneously obtained).

ELECTRODES

Charles S. Lessard, Ph.D.

In a living organism the functioning of many systems is often accompanied by a pattern of electrical signals [11]. These potentials can be monitored, measured, and/or recorded if some type of interface is introduced to transport the signal from the body to the electronic measuring apparatus [10]. Such a signal may then be used to monitor a particular physiological process. The conventional means for carrying out this function are biopotential electrodes [10]. Webster's Dictionary states, "an electrode acts as a conductor to establish electrical contact with a nonmetallic portion of a circuit," in this case the human body [4]. In essence, the electrode operates as an electrochemical transducer to change ionic current into an electronic current [2,10]. This current is a feasible mode by which the physiological monitoring of a particular body function, specifically the electrical activity of the heart, can be obtained. The conventional method for recording voltage variations associated with the beating of the heart is the conducting electrode pair [7,11]. "Despite the many configurations and names applied to electrodes used to measure bioelectric events, there are basically two functional types, extracellular and intracellular" [13]. There are two basic types of extracellular electrodes; i.e., nonpolarizable and polarizable. Perfectly nonpolarizable electrodes freely permit current to pass across the electrode-electrolyte interface without requiring energy to make the transition. An electrolyte is a conducting medium containing readily available ions; e.g., water with salts or acids, gels with salts, etc. On the other hand, perfectly polarizable electrodes are those in which no actual charge crosses the electrode-electrolyte interface when a current is applied. The transfer of ionic charge which does pass is a displacement current and the electrode actually functions as a capacitor [10]. When electrodes are applied to a subject a galvanic cell is created and the pair of electrodes act as transducers [12,13].

The particular charge distribution that occurs when an electrode comes in contact or close proximity with an electrolyte or the subject causes the electrode to acquire a potential [13]. The electrical stability of an electrolyte is related to the stability of the region of charge gradient. The region of charge gradient is the charge layer that exists at the interface between the electrode and the subject under observation. Stabilization of this interface and prevention of movement artifacts is of major concern in design consideration. The interface includes not only an electrode-electrolyte interface (if used) but also the skin and its underlying tissue fluids [12]. Therefore, "distortionless insertion of the event into a recording apparatus requires special consideration of the electrodes and the input impedance of the amplifier" [12].

Because of their unsatisfactory performance in long-duration experiments, conventional "wet" electrodes are not feasible for extended use. During long-term use, increasing variations in resistance readings occur if the skin surface is damaged. Irritation as well as infection can result from the electrode.

The desire to reproduce the time varying biological phenomena as a distortionless signal requires that current into the preamplifier be zero [1]. The input impedance of most biological preamplifiers is high [12] in order to avoid the loss of amplitude and distortion of the electrical waveform.

Geddes recommends that the input impedance of the amplifier be 100 to 1000 times the impedance of the electrode-subject circuit [13]. When the electrode-skin impedance magnitude is much smaller than that of the amplifier input impedance, distortion and noise effects on the biopotential reading are negligible [2].

Wet Electrodes

Various wet electrodes have been used reliably for quite some time. One of the oldest electrode designs is the suction electrode. This electrode employs a rubber cup that is easily and quickly applied, especially to a wet surface, but only for a short period of time because the negative pressure alters the capillary pressure gradient [6]. This electrode requires an electrolyte. The trend in clinics has been to use recessed electrodes with a conducting gel. The recessed electrode moves the metal disc a short distance away from the subject. This method results in movement artifact free readings because the electrolyte absorbs the relative movement between the surface of the skin and the electrode. On the other hand, the electrode impedance is higher than direct contact electrodes. Two promising concepts in electrode design and fabrication are the pressed pill and the disc [1]. The main design goals are a stable, reversible, low-impedance electrode. The pressed pill electrode consists of metal-powder/metal salts that are combined under pressure. Often plant hydracolloids are included because they afford plasticity, adhesion, and texture. This technique also provided a good safety feature because the membrane protecting the electrode is homogeneous with the matrix and can be penetrated (scratched) or damaged. Hydrophillic colloid incorporated with silver/silver chloride was found to be relatively free from polarization effects and afforded extremely low potentials [1].

Geddes contends that the silver/silver chloride electrode is most suitable because of its low cell potential, stability, and longevity [1]. The silver/silver chloride electrodes are popular because they can be made electrically stable. The silver/silver chloride electrode has also been fabricated with a gelatin coat, and tests indicate that this process adds to the electrode's reversibility by keeping out protein and other poisoning molecules while being very compatible with human skin [1]. Nevertheless, wet electrodes are not without their disadvantage, as they all require an electrolyte which can be contaminated in a chemical environment. Silver/silver chloride is photosensitive, reacts with light, and the chloride coating can be degraded by abrasion [12].

Dry Electrode/Active Electrodes

Another type of electrode introduced by the USAF School of Aerospace Medicine, is the anodized aluminum plate (2.5 x 2.5 cm) electrode or dry electrode. The anodized metal

disc relies on capacitive coupling with the skin [3]. A major disadvantage of aluminum oxide is that it can be corroded by saline [8]. The Air Force has also developed a lithium impregnated balsa wood electrodes. This electrode was designated as a dry electrode since no electrolyte was necessary. Its design requires that the intermolecular air spaces in the balsa wood are first emptied and then filled with lithium chloride solution [14].

An active electrode is an electrode packaged with its amplifier into one single unit, which neither uses an electrolyte nor requires skin preparation [5]. Presently, there are two types of active electrodes; insulated and dry [5,13]. Insulated electrodes contain no metal-electrolyte interface. Theoretically, the only current is due to capacitor coupling [7,10,12]. One plate of this capacitor is the electrode itself while the other plate includes the insulating covering on the electrode and the dry outer layer of skin (stratum corneum) [10,13]. This insulating covering is usually a surface oxide film of the metal that has been grown on the electrode plate by vapor deposition [10]. Because no electrode/electrolyte interface is used, there are no artifacts from motion or electrode polarization; however, "displacement on the skin will change the capacitive coupling and hence alter the charge distribution and potential" [3]. Whether this technique causes major problems has not yet been determined. On the other hand, the dry electrode incorporates the use of metallic or other conductive contact between the body and the input of the electrode amplifier. Both of these electrodes are excellent prospects because: (1) electrolytes are not needed, (2) patient safety is improved since little DC current is passed, (3) long-term application is possible, and (4) reduction of electrode area is feasible [5]. The capacitive electrode shows much promise in the noninvasive measurement of physiological signs, particularly the electrocardiogram. One must take a couple of points into consideration. Since the output impedance of the electrode is very high, it is necessary to supply an impedance matching circuit in conjunction with the electrode to be used along with the usual clinical monitoring equipment [3]. Also, a field effect transistor (FET) or a metal-oxide semiconductor field effect transistor (MOSFET) is useful in cleaning up the signal from an insulated electrode [3]. These devices help prevent electrostatic puncture of the insulator. However, this extra precaution is not as necessary when an isolated power supply, such as a battery, is used [13]. Another means of compensating for the high impedance output involves putting ultra-thin films of insulating materials having high dielectric constants and strength on the surface of the electrode [13].

A few types of insulated electrodes are being employed today. A recent paper by Griffith, et al. [7] delineates a lighter weight, more compact electrode configuration through the use of hybrid integrated-circuit technology. Their electrode consists of such a circuit adhered to a capacitor. The capacitor has been formed by "sputtering thin films of tantalum pent-oxide dielectric on a circular silicon substrate 0.176 cm in radius" [7]. However, if a larger substrate had been used, they could have used a more efficient impedance matching circuit. One drawback of this electrode is that there must be no gap between the subject and the capacitor of the electrode. Any misconnection creates a series capacitance that seriously affects results [5]. The capacitor is housed in a plastic disc that is attached to an electrode housing that contains a hybrid impedance matching circuit. The capacitor is

linked directly into this circuit. This is particularly advantageous because this enables capacitors to be exchanged without changing the impedance matching circuit. In order to meet the ideal operational amplifier requirement of no significant DC offset at its output, the largest chip resistor, 100 Megohm was used in the buffer circuit [5,9]. The conductor circuit was printed onto a circular ceramic substrate (1.35 cm in diameter, 0.064 cm thick) with a conducting glaze fired on top. Then the operational amplifier and resistor chips were attached. The electrode was finally integrated to the input of the impedance matching circuit. The ensemble was then placed in a National Aeronautics and Space Administration (NASA) plastic electrode housing. Studies were done on the electrode's ability to function if dirty or damaged and showed no change in step response until the film had been heavily scratched. The signal from the active electrode is superior and quite readable although a little noisy upon overall comparison with that of conventional wet electrodes. The main problem at present occurs when gross separation between the subject and the capacitor happens which causes capacitance changes [5].

Ko, et al. [3] used a silicon substrate rather than a ceramic substrate. The insulating layer is thermally grown silicon dioxide instead of tantalum-peroxide. Their chip of N-type silicon (6 x 6 mm , 0.23 mm thick) incorporates a MOSFET in place of a JFET. Results of their studies show that a thicker dielectric layer is not really necessary as the true dielectric layer is the thickness of the stratum corneum plus the thickness of the insulating film [3,13]. The silicon may be specially etched for a desired circuit; however, this technique is costly regardless of the superiority of design.

Problems to Overcome

The main problem appears to be perspiration or moisture -- the lack of it, the presence of it, the unknown amount of it. Sweat can corrode the insulating layer and/or change the impedance, etc. [12,13]. Therefore a thin, waterproof, insulating coating is suggested.

Electrode size should also be considered. The basic capacitance theory does relate size to actual capacitance which affects impedance. The smaller the electrode, the higher the interface impedance [12]. High input impedance amplifiers have made dry electrodes very feasible with records of equal quality to wet electrodes [1,3].

References

1. Day, J. L., and C. K. LaPinta. The Evolution of Long-term Systems for Electrocardiography on Manned Space Fight, pp. 351-355. In:Biomedical Electrode Technology-Theory & Practice. Miller, H. A., and D. Academic Press, Inc., 1974.

2. Olson, W. H., D. R. Schemincke, and B. L. Henley. Time and Frequency Dependence of Disposable ECG Electrode-Skin Impedance. Med Instrum, 13:269-272 (1979).

3. Ko, W. H., M. R. Neuman, R. N. Wolfson, and E. T. Ton. Insulated Active Electrodes, IEEE Trans Ind Elec Control Instrum, 17:195-198 (1970).

4. Gove, P. B., (Ed.). Webster's Third New International Dictionary. Springfield, Mass.: G&C Merriam Co., 1976.

5. Ko, W. H., and J. Hynecek. Dry Electrodes and Electrode Amplifiers, pp. 169, In: Biomedical Electrode Technology-Theory & Practice. Miller, H. A., and D. Academic Press, Inc., 1974.

6. Fraden, J., M. R. Neuman, and R. Rick. A Dry Electrode Monitoring System, pp. 36-39. Engineering Design Center and Francis Payne Bolton School of Nursing, Case Western Reserve University, Cleveland, 1979.

7. Griffith, M. E., W. M. Portnoy, and L. J. Stotts. Improved Capacitive Electrocardiogram Electrodes for Burn Applications. Med Biol Eng & Comput, 641-646 (1979).

8. Ruggera, P. S. An EMC Test Procedure for an Electroencephalograph Using Human Subjects and Simulated Electronic Patients. IEEE Internat Symp Electomag Compat, 200-205 (1980).

9. Johnson, D. E., and V. Jayokumar. Operational Amplifier Circuits: Design and Englewood Cliffs, NJ: Prentice-Hall, Inc., 1982.

10. Webster, J. G. Medical Instrumentation, Application and Design. Boston: Houghton Mifflin Co., 1978.

11. Mackay, R., and J. Stuart. Biomedical Telemetry: Sensing and Transmitting Information from Animals and Man. New York: John Wiley & Sons, Inc., 1968.

12. Geddes, L. A., and L. E. Baker. Principles of Applied Biomedical Instrumentation. New York: Wiley-Interscience Publication, 1975.

13. Geddes, L. A. Electrodes and the Measurement of Bioelectric Events. Wiley-Interscience Publication, New York, 1972.

14. Salter, M. G. Fabrication of Lithium Chloride-Balsa Wood Electrodes for Electrocardiographic Monitoring. USAF School of Aerospace Medicine, AFSC, Brooks Air Base, TX, SAM-TR-69-51, Aug 1969.

ELECTROMECHANICAL (SOUNDS)

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Another family of noninvasive physiological measurement systems are the electromechanical devices which transform mechanical energy into electrical energy. A large number of medical instruments use strain gage transducers. Most recently self- generating piezoelectric transducers are being used in medical instruments. Some examples of these applications are:

(1) microphones for detection of sounds from the body, i.e., respiratory and cardiac,
(2) accelerometers for motion and tremor measurements, and
(3) ultrasonic measurements [35].

This section will discuss the use of microphones for detection of respiratory sounds and basic problems associated with the use of sounds as a means to diagnose the condition of the respiratory system.

Background

Laennec established the clinical relationship between respiratory sound and gross pulmonary pathology [21] and introduced auscultation of respiratory sounds by stethoscope [32] in the early 19th century. Since he established auscultation of respiratory sounds as a means of diagnosis of the lung's condition in 1819, auscultation in respiratory medicine has not undergone a great deal of development. This lack of advancement is due to (1) the confused usage of the subjective terminology, (2) the limitations of the instrumentation process of analysis of the sounds, (3) the incompleteness of understanding of the mechanism by which the sounds are generated, and (4) the lack of specificity of the location of the source of the sounds.

Limitations of Detection Instruments and Human Hearing

Although the stethoscope being used by physicians today is substantially different from the original version invented by Laennec, the underlying working principle remains the same. The entire range of heart and respiratory sound is transmitted, but the frequency response is uneven. Some frequencies are amplified while others are attenuated [12]. In addition, selection of the chest piece to be used and the variation in pressure used in applying it to the chest wall affect the pitch of the sound heard. For a period of time there was a question as to which chest piece was best. It was decided that both the diaphragm and the bell were necessary for the auscultation of the heart. The difference in frequency transfer of the bell and diaphragm and the effect on frequency response of varying the application pressure can be demonstrated by the auscultation of heart sounds. Low pitched heart sounds are heard best with the bell resting lightly on the chest. On the other hand, firm pressure with the bell or diaphragm amplifies the higher frequencies and suppresses low pitched frequencies. Thus, the faint diastolic murmur of aortic reflux which is composed

mainly of high frequencies and the mitral diastolic murmur which is composed mainly of low frequencies can be selectively heard.

Both the stethoscope and the ear have limitations in their use as instruments for the evaluation of respiratory sounds. Respiratory sounds have a wide spectrum. There is no significant concentration of energy in a particular frequency band except during wheezing. However, this frequency band will differ from one patient to another. In order to compare such relative frequency intensities within a particular sound spectrum, the measuring instrument must not contribute variations in intensity as does the conventional stethoscope. An additional complication in sound evaluation is the nonlinearity of the human auditory system. The ear recognizes very small differences in pitch, but its sensitivity to intensity variations decreases logarithmically as the intensity increases. In addition, the ear's perception of intensity falls off at both ends of the frequency spectrum [14]. The ear also has limitations in its ability to distinguish short sound bursts. A burst shorter than 3 ms will be heard only as a click irrespective of the frequency [16]. With the introduction of advanced electronic microphones, amplifiers, and filters, the human factor and instrumentation shortcomings can be easily overcome [10]. Electronic instruments can be designed to exhibit a flat frequency response over the entire range of respiratory sound spectrum. It is the lack of complete understanding of the mechanisms and sources from which respiratory sounds are generated, however, which still poses obstacles in the acceptance and advancement of using respiratory sounds as a major clinical tool in pulmonary measurement.

Nevertheless, since Laennec's time, it has been known that pulmonary pathology can cause changes in respiratory sounds. It is not surprising that differences exist between normal and pathological respiratory sounds. Changes in the pulmonary system can be revealed by comparing pathological versus normal respiratory sounds.

References

1. Banaszak, E. F., R. C. Kory, and G. L. Snider. Phonopneumography. Am Rev Resp Dis, 107:449-455 (1973).

2. Bullar, J. P. Experiments to determine the origin of respiratory sounds. Proc Roy Soc London, 37:411-423 (1884).

3. Bunin, N. J. and R. G. Loudon. Lung sound terminology in case reports. Chest, 76:690-692 (1979).

4. Bushnell, G. E. The mode of production of the so-called vesicular murmurs. JAMA, 77:2104-2106 (1921).

5. Cabot, R. C. and H. F. Dodge. Frequency characteristics of heart and respiration. JAMA, 84:1793-1795 (1925).

6. Cegla, U. H. Some aspects of pneumosomography. Prog Resp Res, 11:235-241 (1979).

7. Charbonneau, G., J. L. Racineux, M. Sudraud, and E. Tukchais. An accurate recording system and its use in respiratory sounds spectral analysis. J Appl Physiol, 55:1120-1127 (1983).

8. Dekker, E. The transition between laminar and turbulence flow in the trachea. J Appl Physiol, 16:1060-1064 (1961).

9. Dosani, R. and S. S. Kraman. Lung sound intensity variability in normal man. Chest, 84:628-631 (1983).

10. Druzgalski, C. K., R. L. Donnerberg, and R. M. Campell. Techniques of recording respiratory sounds. J Clin Eng, 5:321-330 (1980).

11. Druzgalski, C. Breath sounds in pulmonary diagnosis. IEEE Frontiers of Eng in Health Care;, 383-385 (1981).

12. Fahr, G. The acoustics of the bronchial respiratory sounds. Arch Int Med, 39:286-302 (1927).

13. Fletcher, H. and W. A. Munson. Loudness, its definition, measurement and calculation. J Acoust Soc Amer, 5:82-108 (1933).

14. Forgacs, P., A. R. Nathoo and H. D. Richardson. Breath sounds. Thorax, 26:288-295 (1971).

15. Forgacs, P. Lung Sounds. London: Bailliere Tindall, 1978.

16. Gavriely, N., Y. Palti, and G. Alroy. Spectral characteristics of normal breath sounds. J Appl Physiol, 50:307-314 (1981).

17. Hardin, J. C. and J. L. Patterson, Jr. Monitoring the state of the human airways by analysis of respiratory sound. Acta Astronautica, 6:1137-1151 (1979).

18. Kraman, S. S. Determination of the site of production of respiratory sounds by subtraction phonopneumography. Am Rev Respir Dis, 122:303-309 (1980).

19. Laennec, R. T. H. A Treatise on the Disease of the Chest and Mediate Auscult Translated from the French edition by John Forbes. New York: Samuel Wood and Sons, 1935.

20. Leblanc, P., P. T. Macklem, and R. D. Ross. Breath sounds and distribution. Am Rev Resp Dis, 102:10-16 (1970).

21. Martini, P. and H. Muller. Studies on bronchial breathing. Deutsche Arch F Klin Med, 143:159-173 (1923).

22. Martini, P. The mechanism of production of respiratory sounds. Arch Int Med, 32:313-322 (1923).

23. McKusick, V., J. T. Jenkins, and G. N. Webb. The acoustic basis of the chest examination. Am Rev Respir Dis, 72:12-34 (1955).

24. O'Donnell, D. M. and S. S. Kraman. Vesicular lung sound amplitude mapping by automated flow-gated phonopneumography. J Appl Physiol, 53:603-609 (1982).

25. Pedley,T. J.,P. C. Schroter, and M. F. Sudlow. The prediction of pressure drop and variation of resistance within the human bronchial airways. Resp Physiol, 9:387 (1970).

26. Ploysongsang, Y., R. R. Martin, and W. R. D. Ross. Breath sounds and Region. Am Rev Resp Dis, 116:187-199 (1978).

27. Ploysongsang, Y., P. T. Macklem and W. R. D. Ross. Distribution of regional ventilation measured by respiratory sounds. Am Rev Respir Dis, 117:657-664 (1978).

28. Powell, A. Theory of vortex sound. J Acoust Soc Am, 36:177-195 (1964).

29. Sakula, A. RTH Laennec 1781-1826 -- His life and work: a bicentenary appreciation. Thorax, 36:81-90 (1981).

30. Schroter, R. C. and M. F. Sudlow. Flow patterns in models of the human bronchial airways. Resp Physiol, 7:341 (1969).

31. Skoda, J. A Treatise on Auscultation and Percussion. English ed. London: Highly & Son, 1853.

32. Welkowitz, W. and S. Dentsch. Biomedical Instruments, Theory and design. New York: Academic Press, 1976.

33. West, J. B. and P. Hugh-Jones. Patterns of gas flow in the upper bronchial tree. J Appl Physiol, 14:753-759 (1959).

37. Wong, W. C. Correlation of Respiratory Flow Rate with Frequency Spectrum of Respiratory Sound at Trachea of Normal Young Adults. Unpublished Master of Science Thesis, Texas A&M University, August 1984.

PHONOCARDIOGRAPHY

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Physicians use the acoustical stethoscope not only to examine a patient's respiratory sounds, but also to examine cardiac sounds for murmurs. Some clinics obtain phonocardiograms during medical examinations. A phonocardiogram (PCG) is the graphic representation of the combined sounds of the heart and great vessels. The vibrations that are produced within or about the heart or great vessels propagate outwards through the various tissues. These vibrations appear continuously on the surface of the body.

In the cardiovascular system, low-frequency (infrasonic) mechanical waves are produced and propagated to the skin surface in greater amplitude than the higher frequencies. Consequently, measurements at the skin contain infrasonic components of far greater magnitude than the audible sounds. The tremendous disparity in signal amplitude presents a technical problem in recording if the relative amplitude of all waves are displayed. Recording systems with sufficient sensitivity to detect the low-amplitude, high-frequency sounds will be saturated by the large-amplitude, low-frequency events. Reducing the sensitivity of the recording system to accommodate the large amplitude of the low-frequency waves may result in loss or nondetection of the relatively small-amplitude, high-frequency sounds from the data. The infrasonic waves may be eliminated by filtering; then the instrument sensitivity may be increased to permit measurement and interpretation of the audible waves. The use of electronic filtering may distort the signal. Distortion introduced by sharp bandpass filters may obscure splitting of heart sounds making it impossible to discern where a heart sound ends and a murmur begins. The "ideal" set of filters cannot be optimized simultaneously for both heart sounds and murmurs. Van Vollenhoven, et al. [1] suggest that the best compromise for obtaining the most information for clinical diagnosis is to use lower order high-pass filters which have cut-off frequencies in the low-frequency range and gradual slopes of attenuation.

Most PCGs have at least two filter settings. One range results in moderate filtering of low-frequency sounds, thereby reproducing more effectively low-frequency events. The second filter range setting results in greater filtering of low frequencies. It is designed to accentuate high-frequency events such as diastolic murmur of aortic insufficiency. American National Standards Institute (ANSI) recommends that the input signal be networked for sound measurement. Webster [2] suggests that a solution to the problem of filtering is to pre-emphasize the heart sound frequencies by inserting a frequency-compensation network in the amplifier. The basic system for sound recording consists of a microphone, an amplifier, and an oscilloscope. The function of the microphone is to convert mechanical energy (sound vibrations) into electrical energy. There are many types of microphones, but the following two are the main types. The first is the crystal, or piezoelectric microphone. Certain crystalline materials generate electric energy when subjected to the pressure changes of sonic vibrations. This electric energy occurs when pressures are applied to them in such a way as deform the molecular lattice structure. Since the phenomenon is molecular, it is possible to obtain large electric signals from very

small displacements. For this reason, low-frequency sound waves may be faithfully reproduced [3]. The second type is the dynamic or electromagnetic microphone. In this apparatus, a diaphragm is connected to a movable coil, and the latter is located in a magnetic field. Sonic vibrations set the diaphragm into an oscillating motion, which in turn moves the coil back and forth in the magnetic field. This movement generates electrical energy in direct proportion to the displacement velocity of the coil; therefore, the strength of the electric signal is dependent on both the intensity and frequency of the vibrations. As a result, the system is less sensitive to lower frequencies than the piezoelectric system. On the other hand, its signals are easier to amplify for sound recording and less subject to the pickup of extraneous noise [1].

Microphones should have the general characteristics of being easily applied, comfortable to the patient, and stable in contact with the chest wall. Microphones may make contact with the skin either through bell-shaped endings, which use air coupling between skin and sensing element, or by flat pieces, shaped like a diaphragm, which allow conduction through a solid material directly to the sensing element. Most researchers and clinicians prefer the bell-shaped microphone since it appears to reduce extraneous background noises. The microphones are usually held in place with an elastic strap or suction [1].

Travel [1] states that there is little to be gained by attempting to standardize the amplification of sound waves, because of the variation in patient-to-patient from the contour of the chest and the sound intensity actually reaching the chest wall. It is better to adjust the gain of the instrument to demonstrate what the clinician is seeking. For best sound recording, it is generally desirable to increase the amplitude as much as possible, while maintaining a relatively smooth baseline.

References

1. Travel, M. E. Clinical Phonocardiography and External Pulse Recording. Chicago: Yearbook Medical Publishers, Inc., 1978.

2. Webster, J. G. Medical Instrumentation. Boston: Houghton Mifflin Co., 1978.

3. Geddes, L. A. and L. E. Baker. Principles of Applied Biomedical Instrumentation. New York: John Wiley and Sons, Inc., 1968.

4. Longhini, C., F. Portaluppi et al. The Fast Fourier Transform in the Analysis of the Normal Phonocardiography. Jpn Heart J, 20:333-339 (1979).

AUSCULTATORY MEASUREMENT OF BLOOD PRESSURE

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The measurement of arterial blood pressure is an important indicator of vascular disease or physiological problems. Blood pressure is commonly used in clinical screening and in monitoring of acutely ill patients, because it reflects the effects of changes in cardiac output peripheral vascular resistance, and other hemodynamic or physiological changes. The indirect measurement of blood pressure by a standard sphygmomanometer and stethoscope is not accurate when compared with direct intra-arterial measurements. If the need is to establish whether or not a person's arterial blood pressure is in the normal range, then the accuracy of the measurement has limited consequences. But if the need is for accurate diagnosis of a murmur or serial observations during anti-hypertensive therapy, then the errors in indirect blood pressure measurements must be reduced or avoided [1]. The accuracy of arterial blood pressure measurements by the indirect auscultator from normal subjects with normal circulatory status were found to be poor in some situations.

Direct blood pressure measurements are generally obtained clinically from patients with an unstable cardiovascular system by inserting a 20-gauge catheter into a radial artery and recording the signals from a linear pressure transducer. The advantages of the direct or invasive method of measuring arterial blood pressure are:

(1) permits continuous observation of blood pressure,
(2) provides access to the arterial system for sampling blood, gases, electrolytes, and hematocrit/hemoglobin [5].

The disadvantages of direct measurement of intra-arterial blood pressure are technical problems of inserting the catheter, including air bubbles in the fluid-filled pressure lines, thrombus formation, ischemic damage, and cerebral embolization from irrigation of radial artery lines.

The advantages of the indirect measurement methods of arterial blood pressure are improved patient safety, reduced risk of side effects, and ease of noninvasive measurement. The disadvantage of the indirect method is the accuracy of the measurement.

The accuracy of indirect arterial blood pressure measurement by the auscultatory (Korotkoff) method is dependent on: (1) observer errors; i.e., observer bias, carelessness, inattention, influence by prior readings, (2) differences between observers, (3) errors because of the observed position and conditions under which the measurement is obtained, and (4) instrument error; e.g., incorrect size of cuff [1,7,8].

Equipment For Indirect Measurement Of Blood Pressure

The mercury sphygmomanometer remains the standard instrument for measuring arterial blood pressure in medical practice. The sphygmomanometer consists of a compression

bladder enclosed in an unyielding cuff, an inflating bulb or pump, a controllable exhaust valve, and a manometer. In addition a standard medical stethoscope is necessary to obtain the indirect measurement of blood pressure [1]. Geddes and Whistler [8] contend that the relationship of the width of the bladder to the size of the member to which it is applied is the most important factor in the accuracy of the measure. In an effort to reduce the indirect measurement errors resulting from incorrect cuff size to no more than 5%, the American Heart Association (AHA) recommended that the cuff width should be 40% of the arm circumference. The arm circumference measurement is obtained at the midpoint of the arm--half the distance from the acromion to the olecranon [1]. The bladder length should cover 80% of the circumference of the arm. In summary, arm circumference is the basis for proper cuff and inflation bladder combination, not the age of the patient [1].

Manometers

The most commonly used manometers to register pressure are the gravity mercury manometer and the aneroid manometer. The mercury manometer consists of (1) mercury in a vertical tube and reservoir and (2) a calibrated linear vertical scale. Aneroid manometers make use of a metal bellows which elongates with applied pressure and mechanically transmits the movement to the indicator needle [1].

In the last decade, numerous automatic noninvasive arterial blood pressure measuring systems were developed to eliminate errors resulting from the observer and to extend the range of measurement [7-12]. Silas, et al. [7] evaluated the Dinamap 845 automated blood pressure recorder against a Hawksley random zero sphygmomanometer which was used as the standard. They did not use direct intra-arterial pressure readings as a basis for comparison. Hunyor, et al. [9] evaluated four automated devices, two manual models and the standard mercury sphygmomanometer against direct intra-arterial blood pressure of the brachial artery to which the blood pressure cuff was applied. The findings suggested that all seven types of sphygmomanometers (automated or manual) were not accurate when compared with simultaneous direct readings. The subjects for this study were nine patients who were receiving various drug treatments for hypertension. In all cases the indirect systolic blood pressure measurements were lower than the intra-arterial systolic pressure. In all indirect diastolic blood pressure measurements, with the exception of those taken with one system, the readings were higher than direct readings [9].

In summary, indirect blood pressure measurements whether taken manually or with an automated system are inaccurate when compared with direct intra-arterial measurements. Edwards et al. [12] conclude that "even allowing for an unbiased approach of the observer and a perfectly still and rested subject, the results are no more satisfactory than those of a standard mercury-in-glass sphygmomanometer and stethoscope with moderate care. The sphygmomanometer has the advantage of cheapness and many years of user experience."

References

1. Ramsey, M. Noninvasive Automatic Determination of Mean Arterial Pressure. Med Bio Eng & Comput, 17:11-18 (1979).

2. Collins, V. J., and F. Magora. Sphygmomanometry: The Indirect Measurement of blood pressure. Anesth & Analg. 42:443-452 (1963).

3. Cohn, J. N. Blood Pressure Measurement and Shock. JAMA, 199:118-122 (1967).

4. Paulus, D. A. Noninvasive Blood Pressure Measurement. Med Instrum, 15:91-94 (1981).

5. Downs, J. B., et al. Hazards of Radial-Artery Catheterization. Anesthesiol 38:283-286 (1973).

6. Silas, J. H., A. T. Baker, and L. E. Ramsey. Clinical Evaluation of Dinamap Automated Blood Pressure Recorder. Br Heart J, 43:202-205 (1980).

7. Geddes, L. A., and S. J. Whistler. The Error in Indirect Blood Pressure Measurements with the Incorrect Size of Cuff. Am Heart J 96:1, 4-10 (1976).

8. Hunyor, S. N., J. M. Flynn, and C. Cochineas. Comparison of Performance of Various Sphygmomanometers with Intra-Arterial Blood Pressure Readings. Br Med J, 2:159-162 (1978).

9. Polk, B. F., B. Rosner, R. Feudo, and M. Vandenburgh. An Evaluation of the VitaStat Automatic Blood Pressure Measuring Device. Hypertension, 2:221-227 (1980).

10. Berkson, D. M., I. T. Whipple, L. Shireman, M. C. Brown, and W. Raynor. Evaluation of an Automated Blood Pressure Measuring Device Intended for General Public Use. Am J Publ Health 69:473-479 (1979).

11. Edwards, R. C., R. Bannister, A. D. Goldbery, and E. B. Raftery. The Infrasound Blood Pressure Recorder: A Clinical Evaluation. Lancet, 398-400 (1976).

PHOTOPLETHYSMOGRAPHY

Charles S. Lessard, Ph.D.

The use of electromagnetic radiations from the body in the infrared spectrum of light energy provides a passive, noninvasive method of measuring and recording a subject's relative blood volume change. Changes in blood volume and the blood volume pulse may be measured by observation of the peripheral vascular bed. Historically, blood volume measurements of the body were performed by submerging the body (or body part) in water or in a tightly enclosed air container. These volume measurements and recordings are referred to as plethysmography.

Photoplethysmography is a simple method of recording blood volume changes. The photoplethysmograph instrumentation is not complicated and the transducer can be applied to many external parts of the body. It is important to note that a plethysmograph is capable only of recording relative changes in blood volume and their temporal relationships. The photoplethysmographic method is based upon the large difference between the extinction coefficient of whole blood and that of tissue. The extinction coefficient is also referred to as the absorption coefficient. Tissue is comparatively transparent to the red part of the visible spectrum, whereas blood absorbs most of the red light. Thus, the extinction coefficient of blood is much higher than that of tissue, especially in the near-infrared region of the spectrum (800-900 nm). Therefore, light in this spectral region undergoes variations in intensity as it passes through the tissue. The variations in intensity are indicative of changes in the amount of blood in the tissue, since lower intensity of infrared light transmitted implies a greater amount of blood present. Two basic methods of photoplethysmograph implementation are: (1) measurement of the light transmitted through the tissue (transmitted mode), and (2) measurement of the light reflected from the tissue (reflective mode). In either method, it is standard to record an upward pen reflection as an increase in opacity [1].

The photoplethysmograph transducer consists of two distinct parts: a light source and a photodetector. The most commonly used light sources are small incandescent lamps, which cover the 700-900 nm band with sufficient intensity. The size and intensity of the lamp are limited by heat emission, which may alter the phenomena to be measured [1]. In spite of the great difference between the extinction coefficients of blood and tissue, only small fluctuations in light intensity result because the blood volume is only a small part of the total tissue volume. In turn, the blood volume pulse is also a small part of the blood volume. An average finger has a tissue volume of 4.5 cm, a blood volume of 100 mm, and a blood volume pulse with an amplitude of 2.8% of the blood volume. Calculations made by Weinman [1] show that only 1% of the incident light in the infrared region will be transmitted by the finger phalanx, and that the blood volume pulse will have an amplitude of only 1% of the transmitted light. Thus, the photodetector must be sensitive enough in this spectral region to register the small amount of infrared light reaching it. In addition, the inherent noise level of the photodetector must be 50 to 100 times smaller than the signal intensity level. Two photocells meet these criteria: the photomultiplier, and the

photoconductive cell. Weinman suggests the use of the photoconductive cell since it is much smaller than the photomultiplier. The behavior of the photoconductive cell is such that the conductance of the cell varies proportionally to the fluctuations in light reaching it. An increase in incident light causes an increase in the photocell conductance, and a resulting increase in the current.

The response time of the photoconductive cell is a function of the incident light intensity; i.e., long response time for low light intensities and shorter for higher intensities. Weinman suggests that a time constant of 10 milliseconds is sufficient for blood volume pulse measurements. A high signal-to-noise ratio is also an important characteristic in the selection of a detector, since the lowest detectable signal is limited by the noise level of the photoconductive cell. The signal-to-noise ratio of the photoconductive cell is a function of conductance. At large conductances (high intensities of light) the signal-to-noise ratio is high and deteriorates as the conductance decreases. The conductance of a cell is about 6.7 mhos at typical photoplethysmographic intensities. The signal-to-noise ratio of a 903 Clairex cell is about 10,000 to 1.

The conductance of the photoconductance cell is not only a function of light intensity, but of light history. After application of the transducer, 5 to 10 minutes elapsed time is required to allow the effects of the exposure of the transducer to room lighting to diminish. A typical transducer may contain two light sources and a photocell. It may be used in either the transmitting or reflective mode. The transducers are usually small enough to be applied to a variety of locations. However, since the transducer must be held in place with a minimum amount of pressure to avoid signal distortion, it is very sensitive to motion artifacts. Thus, the subject must remain quite still during measurement recording [1].

An alternative to the system discussed above is a recently developed LED-transistor plethysmograph. The LED's spectral emission is 940 nm with a bandwidth of about 40 nm. The silicon photo-transistor used is sensitive to radiation only in this range. This infrared system overcomes many problems associated with the miniature tungsten lamp/photocell systems, but presents other problems. The first of these is the question of how far the oxygen content of blood influences the measurement with the LED-transistor system. It is well known that the intensity of light transmitted through whole blood or reflected from it is a function of its oxygen saturation. If such is the case in photoplethysmography, no distinction could be made between light intensity changes caused by varying oxygen saturation and blood volume fluctuations. The oximetry of whole blood is based on the great difference between the distinction coefficients of saturated and reduced blood in the red region of the spectrum (650 nm). However, in the infrared region, the extinction coefficient is independent of blood oxygen content. By using infrared light, the influence of oxygen saturation of blood on photoplethysmographic records is removed [1].

Another disadvantage of the photocell is that its output is very temperature-dependent, whereas the phototransistor exhibits temperature drift of 4 millivolts/degree C which is insignificant for most applications. Furthermore, the LED will not create as much heat as the miniature lamp. Test results indicted that the small amount of heat produced by the LED was either too small to be detected, or was dissipated too quickly to have any effect. A further deficiency of the photocell which was mentioned earlier is the dependency of the DC signal on light history. The photo-transistor tested in this respect showed remarkable stability under drastically changing light conditions, indicating that the output of the LED-transistor plethysmograph is independent of prior light exposure [2].

Hocherman [3] fabricated a functional LED-transistor infrared photoplethysmograph coupler. He found that its response was extremely stable and more consistent with the spectral response of blood volume pulse than previous systems.

References

1. Weinman, J. Photoplethysmography. A Manual of Psychophysiological Methods, pp. 187-217. Venables, P.H. and I. Martin (Eds). New York: John Wiley & Sons, Inc., 1967.

2. Tahmoush, A. J., J. R. Jennings, A. L. Lee, S. Camp, and F. Weber. Characteristics of a Light Emitting Diode - Transistor Photoplethysmograph. Psychophsiology, 13:357-362 (1976).

3. Hocherman, S. and Y. Palti. Correlation between Blood Volume and Opacity Changes in the Finger. J Appl Physiol, 23:157-162 (1967).

4. D'Agrosa, L. S. and A. B. Hertzman. Opacity Pulse of Individual Minute Arteries. J Appl Physiol, 23:613-620 (1967).

5. Uretzky, G. and Y. Palti. A Method for Comparing Transmitted and Reflected Light Photoelectric Plethysmography. J Appl Physiol, 31:132-135 (1971).

6. Weinman, J., A. Hayat, and G. Raviv. Reflection Photoplethysmography of Arterial Blood Volume Pulses. Med Biol Eng Comput, 15:22-31 (1977).

7. Ochoa, W., I. Ohara. The Effect of Hematocrit on Photoelectric Plethysmography. Tohoku J Exp Med, 132:413-419 (1980).

8. Wouda, A. A. Raynaud's Phenomenon. Acta Medica Scandinavica, 201:519 1977.

9. Wasserman, D., W. Carlson, S. Sameuloff, W. Asburry and T. Doyle. A Versatile Simultaneous Multifinger Photocell Plethysmography System for Use in Clinical and Occupational Medicine. Med Instrum. 13:232-234 (1979).

CHEMICAL SENSOR TECHNOLOGY

David J. Costello

Chemical sensors are devices which provide output that can be used to predict the concentrations of chemical compounds in the proximity of the sensor. Bioanalytical sensors include both chemical sensors (which are used to determine the properties of the sources, metabolic processes, and products of living systems) and biosensors. Biosensors are those which use biochemical compounds, organelles, tissues, or organisms as a functional part of the sensor. Most current research in the field of chemical sensors is related to the development of bioanalytical sensors. This work is often applicable to the measurement of non-biological and inorganic processes. For example, an oxygen sensor developed for blood gas sensing may be useful in monitoring the use of oxygen by fuel cells.

The observation field of a sensor is the volume of space in which chemical compound may interact with the sensor. An analyte is the chemical compound to be measured, existing in some unknown concentration in the observation field of the sensor. The chemical background includes all of the other chemical compounds existing in the observation field. The sensor environment includes the non-chemical energy producing elements such as radiation sources and potential movement of the subject system. The subject system may be a pipeline, process container, or a human body.

A means for separating the analyte from the chemical background is sometimes required. This is generally provided by a selectively permeable membrane. Separation may also be performed before the test sample is brought in contact with the sensor by centrifugation or chromatography. Reactants are chemical compounds provided as part of the sensor which react with the analyte to produce a product. Mediators, substrates, and catalysts control the reactions. The analyte itself may be a mediator, substrate, or catalyst. Cascading reactions may be employed to eventually produce a change in an indicator which is the product, or effect of a reaction which influences the flow of energy from the sensor. Conduction means carry the modified energy from the sensor for detection and analysis. Conductors may also be required to provide energy to the sensor.

Additional transducers may be needed to transform the sensor energy into electronic form to simplify analysis. Preamplifiers increase the magnitude of the sensor output to reduce the effects of subsequent noise sources, and transform the signal into a form, generally a voltage, which can be handled by subsequent devices. Signal conditioning devices are used to reduce or subtract noise. Digital conversion of the signal generally takes place at some point either before or after conditioning. Signal analysis converts the signal into a prediction of chemical concentration of the analyte.

Interferents are chemical compounds capable of interacting with the sensor to produce errors in the predicted analyte concentration. Noise sources are those capable of producing energy errors in conduction of the sensor signal. These sources of error are often

reduced by the use of a reference sensor which responds to the error source but not to the analyte. The evolution and development of chemical sensors are headed toward the elimination and integration of these elements to produce simplified systems and performance improvements.

Design and Performance Issues

Sensitivity is the most important performance criterion of a chemical sensor system. Sensitivity can be defined as the change in sensor output given a known change in analyte concentration (dS/d[a]). High sensitivity in a sensor will reduce the effects of all error sources which are not directly related to the reaction chemistry, improving most of the other performance measures. Sensitivity is generally measured by linear regression of the system output values against samples of known analyte concentration. Sensing range is the analyte concentrations between which the sensor is expected to yield reliable results.

The signal-to-noise ratio (S/N) of a chemical sensor system is the ratio of the output produced by a sensor at a known concentration of analyte, to the standard deviation of the signal produced in the absence of analyte, given that the absence of analyte is expected to yield an output of zero. For quantities measured on a non-zero scale, such as Ph, the ratio of change in signal over a known range to the average deviation of the signal at a constant concentration may be used.

The detection limit is the minimum concentration of analyte which produces a signal that can be reliably differentiated from the signal produced in the absence of analyte. This is generally the amount of analyte which produces a signal change three times greater than the noise level. The resolution is the minimum change in analyte concentration within the sensing range which will produce a reliable change in signal.

Linearity is the change in sensitivity of a sensor with respect to changing analyte concentration. If sensitivity is constant over the entire sensing range, the sensor is linear. If the sensor is non-linear, error measurements and specifications made or quoted in one part of the range will not hold true in other parts.

Calibration is the process of determining the mathematical constant terms of the equations used for predicting analyte concentration from signal output. Most sensors are calibrated on an individual basis prior to use by exposing them to two or more samples of known concentration. Some sensors may be manufactured with such a high degree of similarity that the calibration constants will be nearly identical. These may be calibrated by statistical sampling of a manufactured batch. Some sensors use a combination of these techniques.

Accuracy is a measure of the prediction error of a sensor represented by the average difference between measurements of analyte concentration made in a number of samples,

and the known true concentration of analyte in the samples. Often a second sensing system will be used to determine the "true" concentrations of the samples, and this measure will be called the bias of the first system with respect to the second. Precision is the variation in results when a sample with the same concentration is measured repeatedly. Precision may also be the variation in the difference between measurements and true concentrations of samples.

Drift is the change in accuracy of a sensor system over its time of use. The materials and compounds used in most sensors will degrade over some time period, producing drift. Sensors which drift beyond a useful accuracy within a short period must be recalibrated often.

Sensor systems use several sampling methods. Intermittent sampling involves the occasional or periodic withdrawal of a sample from the subject process, transporting the sample to the sensor, and possibly pre-treating the sample by separating components or removing components by chemical reaction. Continuous flow sensing is similar but involves less sample interaction. These sampling techniques may be automated to reduce operator interaction. *In situ*, indwelling and *in vivo* sensors reside within the subject or process stream and generally provide continuous, or nearly continuous, output.

Reversibility is the property of the reactants of sensors which allows them to measure both increases and decreases in analyte concentration. Some reactions can proceed in only one direction, as reactants are used up and cannot be regenerated. Sensors which are not reversible are nonetheless useful as threshold monitors or alarms, or disposable intermittent measurement devices.

Response time is the time which a sensor takes to reach chemical equilibrium in response to an abrupt change in analyte concentration, response time is generally used in reference to a continuous sensing system. Measurement time is the time that a sensor system takes to produce one measurement. Cycle time is the minimum time period between sequential samples.

Selectivity is the measure of how well a sensor works in complex solutions. A highly selective sensor can work well in complex solutions like blood, seawater, and sewage. Poorly selective sensors require additional sample preparation. Specificity refers to the ability of a sensor to respond only to the analyte of interest and not to any interferents which may be present in the chemical background.

Chemical sensors are basically created by the combination of a chemical recognition element, or reaction mode, and a transduction mode. Table C-1 presents a selection of the more commonly used elements. Currently investigated transduction modes for bioanalytical sensors include electrochemical, photochemical, piezoelectric, mass spectroscopic, and magnetic resonance techniques. Each mode has specific advantages for different applications.

Table C-1 Basic Bioanalytical Sensor System Components

Transduction Modes

* Electrochemical
    - Faradaic
        - Potentiometric
        - Amperometric
    - Non-Faradaic
        - Conductive
        - Impedimetric
        - Capacitative
    - Semiconductor
* Photochemical
    - Photoabsorption
    - Fluorescence
    - Emissive Lifetime
* Photophysical
    - Refractometric
    - Reflectometric
    - Surface Plasmon Resonance
* Thermometric
* Electroacoustic

Reaction Modes

* Exchange Membranes
* Enzymes
    - Micro-organisms
    - Tissues
* Immunochemicals
* Bioreceptors
* Photochemicals
    - Electroactive Polymers
    - Amphoteric Dyes
    - Fluorescent Labels
* Ionophores

Electrochemical and photochemical techniques are somewhat similar on an analytical level in that they both depend on molecular interaction and electron transfer reactions. Sensors based on an electrochemical mode possess the greatest body of research and are available in product form for a variety of analytes. Electrochemical modes include Faradaic techniques; potentiometric (charge distribution between electrodes), and amperometric/-polarographic (current between electrodes), and non-Faradaic techniques; conductometric (resistance of a sample between electrodes), and capacitative (capacitance of a sample between two conductors). Changes in these properties are imparted to the sensor by the analyte or reactant products. The most basic amperometric sensor is the Clark electrode based on the cathodic reduction of oxygen on the surface of a charged platinum electrode. The concentration of reduced oxygen changes the resistance of an electrolyte between the platinum and reference electrodes which is measured at constant voltage. The Clark electrode and the potentiometric pH electrode form the foundation of electrochemical measurement techniques. Electrochemical sensors will generally exhibit shorter reaction times because they can be measured on the basis of a surface response with junction capacitance or reaction rates being the controlling variable. Electrochemical sensors often exhibit greater noise levels, thermal drift, and inaccuracy, with decreasing size because of an exponential increase in electrical impedance and because of the electronic instability of reactants at very low concentrations. Electrochemical sensors typically possess very favorable power consumption requirements and inexpensive signal analysis instrumentation. Electrochemical sensors are often susceptible to interferences from electromagnetic sources in surrounding environment. Advances in electronic manufacturing processes have made it possible to effectively address many of the weaknesses of electrochemical sensors, stimulating a variety of new research efforts.

Photochemical sensors are based on the observation of the absorption, reflection, or refraction of light, or the production of a fluorescent emission by the analyte or reactants. The reactants which exhibit the observed optical effects are called indicators. Optical sensors may be as simple as a transparent container through which light is projected. Photochemical sensor systems typically use waveguides, such as optical fibers, to conduct light both to and from the sensor. Most optical sensing techniques allow control of the signal output and some noise characteristics of the sensor, since they are generally dependent on the signal input which can be varied by selection of the light source. Some laser and incandescent light sources currently employed in optical sensor systems may possess qualities of power consumption, weight, and heat which would be prohibitive in space applications. Photochemical sensors often exhibit slow reaction times relative to electrochemical sensors because the sensitivity will often depend on the concentration of the indicator, and bulk reactions which require diffusion of the analyte are generally employed. Surface plasmon resonance, evanescent wave, and refractometric techniques exploit photometric effects at the surfaces of materials and may reduce reaction times. The dependence of these techniques on surface area typically requires relatively larger sensors or more complex instrumentation to achieve sensitivity. Optical sensors exhibit relatively low drift rates and have often been used in continuous monitoring applications. Optical sensors are not affected by electromagnetic noise but may be susceptible to degradation

from alpha and gamma radiation. A few photochemical sensing systems for blood gases and glucose are currently on the market. In addition, colorimetric microplate readers which are in common use in analytical laboratories are classified as optical sensors. The success of these products, and current research in light sources, integrated optics, and optical materials, is stimulating a rapid growth in this area. Historically, optical sensors have been restricted to the use of peak height analysis or fluorescent lifetime measurements as a primary analytical technique, prohibiting the use of the infrared (IR) and near infrared (NIR) spectra. The use of IR and NIR spectra is advantageous since the molecular structures of most biochemical compounds could be observed directly. The use of multivariate spectral analysis techniques and emerging electro-optical hardware technology makes NIR technology the fastest growing area for investigation in bio-analytical sensing.

Acoustic wave sensors are based on the mediation of the acoustical resonance of a piezoelectric crystal by interaction with the analyte or reactants. These sensors depend on the change in mass of the crystal as analyte or reactant is absorbed or deposited on its surface. These sensors exhibit high sensitivity and very good stability, but poor selectivity since small changes in the mass of compounds deposited on the crystal will be detected as analyte effects. Many piezoelectric sensor systems depend on the precipitation of a reaction product. These sensors will be adversely affected by microgravity conditions. Acoustic wave sensors are generally investigated for use in the detection of large complex molecules and the use of immunochemical reactions provides good specificity. Acoustic wave sensors typically exhibit long shelf lives but require frequent recalibration in use so long detection cycles may prohibit use in continuous monitoring applications.

Investigation and research efforts to produce sensor systems using mass spectroscopy have declined recently because of the emergence of alternative techniques. The major disadvantage of mass spectroscopy has been the need for complex equipment to produce pure vacuum and the effects of gravity. The space environment may present an ideal setting for progress in the use of this highly sensitive technique. Mass spectroscopy is limited, however, by the requirement for detection of small, charge sensitive molecules.

Semiconductor sensors are produced by combination of a field effect transistor with an electro-active chemical reaction. Field effect transistors respond to electron transport into and out of their silicon gate region. In general a metallic conductor is placed in contact with the gate region on one side and an immobilized reactant on the other. Reaction with the analyte produces electron transport across the conductor and changes in the capacitance in the gate. The major advantages of semiconductor sensor devices lie in the potential to mass produce a large number of essentially identical sensors. However the small size of these sensors and their dependence of their response on a very limited number of analyte molecules has produced poor reproducibility. Semiconductor sensors also share the favorable response time dynamics of electrochemical sensors. Poor selectivity produced by sensitivity to small electrochemical state changes in the sensor chemical and thermal background have been exhibited by most semiconductor sensors. In

addition, the electrical transduction of CHEMFET signals has also been a significant problem. These problems have recently been addressed by the integration of solid state circuitry within the chemically sensitive device. Integrated circuits have included thermo-compensation devices, preamplifiers, signal analysis, filtering, and analog to digital conversion circuits which dramatically enhance the fidelity of semiconductor sensors.

Transduction modes may be combined to produce sensor systems with attractive properties. For example, a recent innovation has been the production of biosensors based on optical interrogation of a biologically mediated silicon phototransistor. This design incorporates the stability and safety of optical transduction, the speed of electrochemical reactions, and the potential for microintegration provided by silicon electronics.

Generally, when selecting a reaction type, the more complex the reactants, the less stable the sensor. Bioanalytical sensor research has, for the most part, evolved toward simpler reactant schemes and the use of immobilized reactants and catalysts. Sensors that require water as a reactant will be subject to increased hydrolysis and/or oxidation which will shorten shelf life if the sensor is packaged in a hydrated state, and lengthen preparation time if hydration of the sensor is required before use. Oxidation can often be controlled by appropriate packaging. Typical reaction types include ion exchange, reduction/oxidation reactions, immunochemical reactions, photometric reactions (fluorescence, absorbance, and fluorescence quenching), enzymatic catalysis, natural chemoreceptors and the metabolic processes of living cells. These reaction processes may be combined in a variety of ways to produce the energetic changes comprising a sensor response.

The pH electrode is the simplest and earliest bioanalytical sensor. Its response is based on the electrical potential developed between two electrodes when one is maintained under controlled electrochemical conditions and the other may donate electrons to neutralize hydrogen ions in the detection field. Ions in the proximity of the sensor are provided by an electroactive membrane which exchanges ions in the chemical environment for ions in the vicinity of the electrode. This basic paradigm has motivated the development of many materials which exchange different types of ionic compounds. The sensitivity of an ion exchange reaction depends on the charge and concentrations of the ions being detected. Ion exchange materials are generally rugged and have very fast response times since the ions are exchanged at the surface of the membrane and no diffusion is involved. A wide variety of potentiometric sensors have been created by producing ionic compounds as the by-products of other reaction types and detecting these ions with potentiometry or by subsequent reactions. The major weakness of exchange membrane techniques is the maintenance of constant conditions on the sensor side of the membrane, so drift is often a problem. Membrane biofouling is also a significant challenge since free ion flow at the membrane surface is required for reliable sensor operation.

Enzymes are proteins which work as catalysts of biological reactions. Enzymes facilitate or inhibits the conversion of a target compound, called the substrate, into a product. The product may be combined with another substrate or released. A huge variety of enzymes

exists in the membrane and cytoplasm of cells. Each enzyme type produces a unique and specific reaction. These compounds are generally simple globular proteins. Enzyme based biosensors generally work by detection of the products or sources of enzymatic reactions. The most widely investigated example of an enzyme sensor is based on the detection of oxygen consumption by glucose oxidase which catalyses the oxidation of glucose. In this sensor the consumption of oxygen is detected with an amperometric or optical sensor. Enzymes are generally chemically instable when stored under wet conditions, or when exposed to extremes of temperature and radiation. Thus, enzyme sensors may possess serious disadvantages with respect to sterilization, lifetime, and storage. Chemical immobilization of enzymes has been used as a method for improving stability. Stabilization with respect to hydrolysis may also be improved by adsorbing the enzyme into an anhydrous membrane, or by requiring rehydration of the sensor immediately before use. The reaction time of enzymatic reactions may increase the response time of a sensor by factors on the order of a second. Recent investigations of synthetic enzymes (with increased stability and reduced dependence on environmental conditions) promise to expand the applicability of enzyme-based sensors.

Immunochemical compounds have produced a large potential for the specific detection of chemicals. Immunochemicals consist of specifically paired sets of reactants, an antibody and an antigen. An antibody is generally a large globular protein produced by cells in response to the presence of an immunogen and possessing a specific chemical binding site for the target antigen. Antigens may be complex molecules, proteins, drugs, cells or whole microorganisms. The chemical and structural determinants of an antigen's ability to bind to an antibody are represented at specific locations called epitopes. The degree of complimentary similarity between the epitopes of an antigen and an antibody determine the binding affinity between the two. Chemically similar antigens may possess different degrees of affinity producing crossreactivity. The formation of a immunogenic complex by the union of an antigen and an antibody may be detected by the change in mass of the compound using immunochemicals immobilized on an electroacoustic sensor. Immunochemicals may be modified by the addition of a photometric indicator component. Changes in the chemical structure of the labeled compound may be designed to produce observable changes in the indicator. In competitive binding techniques a radioactive or photometrically labeled antigen is exposed to the antibody either simultaneously or in sequence with a sample containing an unknown concentration of antigen. After washing the sensor, the proportion of labelled antigen present will be dependent on the binding affinities of the two antigens and the concentration of unknown antigen in the sample. In some cases, the immunocompounds themselves may be photometric indicators.

Biological chemoreceptors, or bioreceptors, are biologically produced chemical structures which, unlike antibodies, are not produced in response to the presence of other chemical compounds. Bioreceptors are generally found in the structure of neural and digestive tissue cells. These compounds function by increasing the electrical or chemical conductivity of the cell membrane upon target molecule binding. Cellular response to bioreceptor binding has been found to represent the highest order of chemical selectivity and

sensitivity. The isolation, purification and synthetic production of bioreceptor compounds is a relatively new field and techniques for the use of these compounds generally have far to go before producing methods robust enough for commercial sensor applications. The current use of bioreceptors is generally performed through the observation of the cells in which they are found. The metabolic performance of whole cell biosensors is observed by the consumption or production of metabolites as detected by more conventional electrochemical and optical techniques. The use of whole cells provides possibilities for improved stability of enzymes and chemoreceptors in the cell since the intracellular conditions are actively maintained. The relatively labile nature of bioreceptors and cells presents a significant problem to progress.

Photometric reactions are produced by changes in the potential energy of the chemical bonds of molecules. All poly-atomic molecules are potential photometric indicators and there is a theoretical basis for detection of these compounds directly. Application of this theory has proven impractical for several reasons. First, emphasis in bioanalytical sensing has been placed on the detection of simple mono-atomic and ionic species. Perhaps this is because of the availability of electrochemical sensing techniques, but also because these simple molecules provide the basis for cellular metabolism. Second, most compounds absorb light poorly across a wide wavelength spectrum. The instrumentation and data analysis techniques for distinguishing low concentrations of poorly absorbing compounds have not existed until recently. Instead of direct photometric analysis of analytes, techniques have been developed to produce secondary reactions with strongly absorbing indicators. The photochemical properties of indicators include not only their absorbance of light, but their emission of absorbed energy as light (fluorescence). In addition the photophysical properties, such as the index of refraction or reflectance, of some compounds may change upon reaction with an analyte. Bioanalytical sensors based on photometric reactions have generally taken the form of an indicator system immobilized on either the distal portion of an optical fiber, or on the surface of a clear polymer plate. The indicator may be immobilized or adsorbed into a polymer media which is in turn attached to the conducting means. The reaction is monitored by directing a light beam into the indicator via the plate or fiber, and observing changes in light reflected or transmitted through the indicator complex. The chemical stability of photometric reactions varies widely and is determined by the identity of the indicator used. A common problem for many photochemical systems is the degradation of the indicator compound produced by light absorption. In addition, photophysical degradation of the light conducting media and the immobilization media contribute to long term changes in performance.

PIEZORESISTIVE SILICON PRESSURE SENSORS

Tommy G. Cooper

A piezoresistive silicon sensor is usually constructed in two layers. The top wafer is etched to form a thin diaphragm, typically 0.001" thick. The bottom wafer which is either silicon or glass is attached to the top wafer via anodic bonding, glass fit sealing or silicon-to-silicon bonding. The cavity between the two layers may be sealed or a hole may be laser drilled or chemically etched through the constraint wafer. With this cavity sealed under a vacuum, the sensor could be used for absolute pressure measurements. The hole in the constraint wafer allows pressure to be applied to either side of the thin sensing diaphragm. In a typical gage pressure transducer application, the hole is used to vent the sensor to atmospheric pressure.

The conversion of mechanical pressure to an electrical signal is achieved by diffusing piezoresistive strain gage elements at specific locations on the surface of the thin diaphragms. The diffusion process is a standard integrated circuit technique by which impurities are diffused into the crystal lattice of the silicon to form functional electronic components. Piezoresistive strain gage elements are similar to wire strain gages in that they change their resistance in response to mechanical deformation. The piezoresistive element has a much higher sensitivity to mechanical stress due to the "piezo" effect of deforming the crystal lattice of the silicon. These piezoresistive elements are configured in the form of a resistive wheatstone bridge which is a conventional circuit for transducers.

The mechanical (or physical) input is converted to an electrical signal in the following way. The pressure exerted on the thin diaphragm causes it to deflect. The piezoresistive sensing elements are strained by this deflection, causing a change in their resistance. This change in resistance is converted to an electrical voltage in the bridge circuit and is detected by a bridge amplifier. Bridge circuits are ratiometric; that is, the output voltage is proportional to the change in the resistive elements and the voltage which excites the bridge.

The piezoresistive silicon pressure sensors are not calibrated and are temperature sensitive after etching the silicon and depositing the piezoresistive elements. They require adjustment to compensate for these deficiencies. This adjustment is achieved by laser trimming resistive components located on the silicon around the diaphragm or on a separate ceramic chip carrier. The zero offset and sensitivity of the sensor is adjusted by applying a calibrated physical input and trimming the output voltage to a specified value. Temperature compensation is achieved by process design or by off-line characterization of the temperature errors of the sensor then laser trimming resistive components to computed values. This laser trimming process is accomplished at high speed to achieve cost targets. This laser trimming process results in a calibrated pressure sensor trimmed to within tight tolerances.

APPENDIX D--PRODUCT DATA SHEETS AND BROCHURES

The following product data sheets and brochures contain examples of relevant smart sensor technology.

Fast 24-bit ADC converter handles dc-to-410 Hz input signals

An oversampling sigma delta a-d converter in a pair of hybrids converts dc-to-410-Hz, 120-dB dynamic-range signals into 24-bit digital words.

FRANK GOODENOUGH

Just when you think you've seen the last bit of resolution wrung out of an analog-to-digital converter (ELECTRONIC DESIGN, Sept. 3, p. 65), along comes one from Gould to set a new world record. At 24 bits, a new a-d converter outdoes the 22-bit unit reported earlier, as well as anything else on the market. What's more, it handles ±10-V 500-Hz input signals with 100% overrange capability.

The oversampling a-d converter is formed by a trio of hybrids that represent a two-channel system *(Fig. 1)*. Gould calls the two Model 860 chip-and-wire devices on the photo's left enhanced delta modulation encoders (EDME). The surface-mounted model 863 hybrid on the right is a dual-channel, finite impulse-response (FIR), digital, antialiasing (low-pass) filter. Compared to most systems that employ from a few to often hundreds of channels, the dual-channel approach permits filter parts to be shared between channels and significantly reduces system size and cost. In addition, a single-channel system is also available.

The 860/863 combo converter was originally developed as the front end of digital signal processing (DSP) systems that detect earthquake-masked underground nuclear explosions. Its prime capability, separating or resolving small ac signals riding on top of large ones, quickly attracted the attention of the seismic wing of the petroleum exploration industry, which also has a strong interest in detecting underground explosions.

A second basic advantage: the converter can physically and/or electrically separate its two major blocks of sensed signals and transmit a relatively low-speed bit stream between them. Other typical applications range from systems analyzing the vibration in heavy machinery (such as locomotive diesels) to systems capturing minute biological potentials.

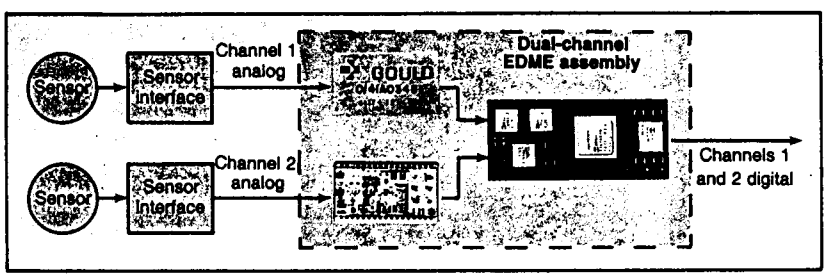
A BRICK WALL

The converter samples a ±10-V (7.07-V rms) input 256,000 times/s, or about 256 times the Nyquist rate of its maximum usable input frequency, 500 Hz. Its output is a serial, 1-bit digital representation of the input—a stream of ones and zeros. The brick-wall FIR filter takes this bit stream, rolls it off at an unbelievable 390 dB/octave, converts it to a 24-bit parallel word, and decimates the oversampled input by a factor of 256. It finally converts the 24-bit word into either a 3-byte integer mode or a 4-byte instantaneous floating-point output format ready for use by a microprocessor, VAX, or Cray. New words are available at a 1-kHz rate.

The integer mode format presents the data to the computer bus in three two's-complement bytes, with the fourth byte being a repeat of the first. The first three bytes of the instantaneous floating-point mode format are similar and represent the mantissa; the fourth byte represents the exponent. At any time the user can select data from channel 1 or 2 in the desired format and the desired byte.

Before delving further into the operation of the converter, let's look at its specifications. This very uncommon converter's specifications are similarly uncommon. Basically, the specifications are in the yet-to-be-standardized genre of specifications employed for digital signal processing.

When the converter samples ground theoretically at 250 kHz, input noise runs a maximum of 462 nV/$\sqrt{\text{Hz}}$ (330 nV/$\sqrt{\text{Hz}}$ typical), and system bandwidth (with the FIR filter) is 410 Hz. Dropping the



1. Detecting small signals buried in large ones, this trio of hybrids from Gould forms a two-channel, 24-bit a-d converter that can handle signals greater than 400 Hz.

sampling rate to 128 and 64 kHz raises the noise 3 dB each step. But since the bandwidth is also halved at each step, effective noise and dynamic range is unchanged. The instantaneous dynamic range (the largest input change the converter can handle from one sample to another) is 120 dB at all sampling rates. Harmonic distortion—defined as the level of the highest harmonic relative to a full-scale input signal of 7.07 V rms (20 V pk-pk)—is typically 0.00025%, or -112 dB.

A more familiar parameter, gain (full-scale) accuracy, is a minimum of ±0.056% at 25 °C. Gain stability is ±0.2% at 25 °C and double that from -55 to +85 °C. The converter also has a virtually constant offset voltage of 250 mV. It reaches that value within one second after power is applied. Both gain and offset errors are easily removed with software by occasionally sampling an external reference (gain) and ground (offset). The converter's analog circuits operate from typical ±15 V supplies, with 3 mA at 5 V needed for digital logic. It is specified to operate from ±10 V, typically drawing 9 mA, however. Thus in remote applications it can operate a long time from a pair of 12-V storage batteries.

The FIR filter's true brick-wall performance will make classic filter designers green with envy. Rolling off at 390-dB/octave, it looks like a 65-pole filter, with absolute linear phase response. In digital filter terms, it is an 11,456-coefficient filter. When sampling at 256 kHz, its output is down 6 dB at 450 kHz, down 40 dB at 500 Hz, and down 135 dB at just 535 Hz. Unlike a classic filter, it has no upward cusps in the spectrum at higher harmonic frequencies. And in-band ripple is typically a low ±0.5 dB. All inputs and outputs are TTL- and CMOS-compatible, taking 100 mA from the 5-V rail.

You don't get something for nothing. So what do you give up in converter performance as a result of this, at first glance, seemingly perfect filter? Well, like all filters, it has a delay. What's more, as might be expected from a 65-pole filter, the signal delay is 22.375 ms. That is, 22.375 ms pass from the time an impulse is applied to the input of the converter until an output digital word truly represents the peak of the impulse regardless of the magnitude of the impulse. This delay in no way interferes with the high-frequency response of the converter (410 Hz) in gathering data for future analysis. Careful system design is required to put this a-d converter in a feedback controlled loop.

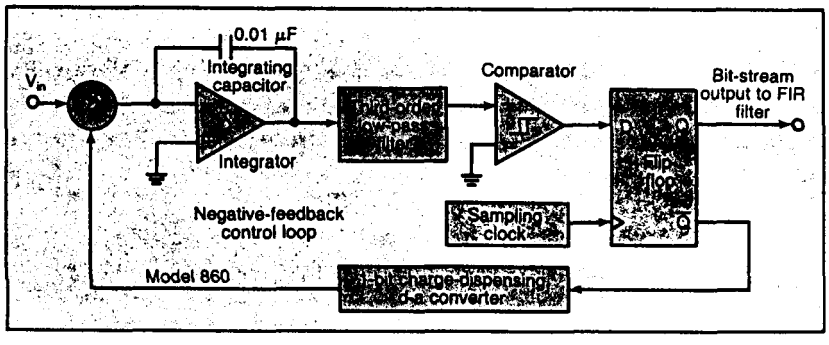
The converter is essentially a sigma delta modulator *(Fig. 2)*. Its performance is enhanced by the insertion of a spectrum-shaping, low-pass filter in the forward path of the control loop. This third-order filter offers high gain in the passband of interest and rapid attenuation in the stop band. As a result, it reduces the quantization noise of the d-a converter in the signal's passband, forcing it instead to rise rapidly beyond the passband.

PRICE AND AVAILABILITY

The EDME goes for $720 apiece. In similar quantities, the Model 863-02 single-channel FIR filter goes for $943 each, and the dual-channel Model 863-01 goes for $1,286 each. Thus a complete two-channel system, consisting of a pair of Model 860-04s and a Model 863-01, can be put together for $2726. Small quantities are available from stock.

Oil exploration application of the Gould converter is restricted to Geosource Inc. and the Western Geophysical Division of WAI, Inc. through a worldwide exclusive license. This restriction does not apply to other industrial applications.

Gould Inc., Ocean Systems Div., 10500 Richmond, Suite 201, Houston, TX 77042; Paul Madeley, (713) 783-3575.



2. The low-pass filter between the integrator and the comparator of this sigma delta modulator reduces the quantization noise in the pass band when it is used with an FIR filter to form an oversampling a-d converter.

JUST ONE BIT AT A TIME

Here is how the circuit works. The integrator operates on the sum of the input signal and the output of the 1-bit, d-a converter. The negative-feedback control loop continuously tries to minimize that difference by injecting precise quanta of either positive or negative charge into the integrator as required to reduce the charge on the integrator's capacitor. The comparator (a 1-bit a-d converter) checks the polarity at the output of the low-pass filter. If

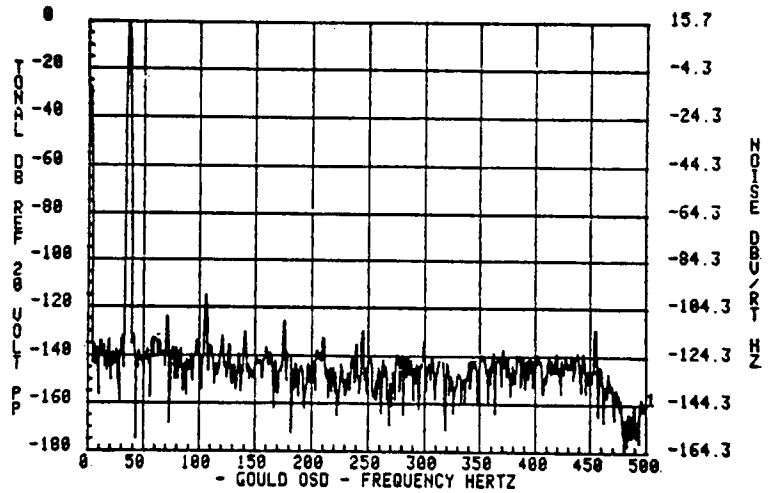
the polarity is positive, a logic one is clocked into the bit stream at the flip-flop's Q output; if the polarity is negative, a logic zero is clocked. Similarly, the $\overline{Q}$ output of the flip-flop decrees either the positive or negative hunk of charge from the d-a converter to the integrator input.

When operating as specified with a 2.048-MHz input clock, the sampling clock runs at 256, 128, or 64 kHz under software control to set signal bandwidths at 410, 205, and 102.5 Hz, respectively, when used with the FIR filter. The filter consists of a two-stage cascaded circuit. The first stage with 448 coefficients decimates the input bit stream by 64, supplying a low-pass filtered 24-bit word stream at 4000 samples/s. The second stage with 172 coefficients decimates the word stream by four, reducing the sample rate to 1000 samples/s, and further low-pass filters the data.

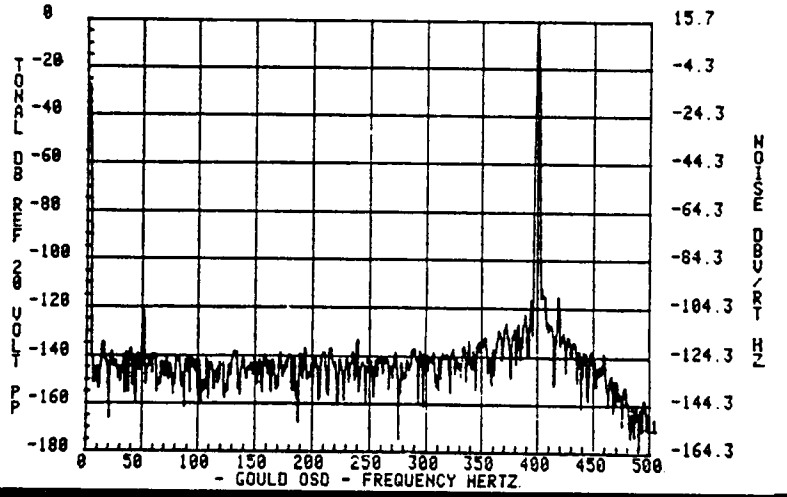
The FIR filter operates from a 20-MHz clock. In a two-channel system just three digital lines connect a pair of EDMEs to a FIR filter, a data clock, and the two bit streams. Besides the 2.048-MHz clock, the converter needs a handful of bypass capacitors and one very stable one for the integrator. For most applications that capacitor should be a foil-wrapped, Teflon device with a value of 0.01 μF.

The 860/863 chip-and-wire 24-bit a-d converter is housed in a 34-pin, triple-width, hermetically sealed metal package. The FIR filter is composed of multiple, surface-mounted, ceramic leadless chip carriers mounted on a ceramic substrate. The substrate consists of a 64-pin DIP that is 1.4-in. wide (between the pin rows).□

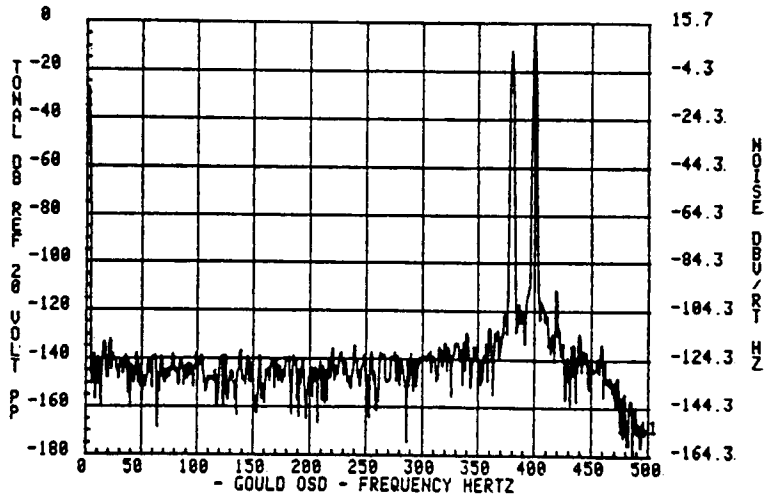
SINGLE TONE TEST
35Hz @ 0dB (20V P-P)
1000 s/s, 2048 POINT FFT



TWO TONE TEST
400Hz @ 0dB (20V P-P)
50Hz @ -120dB
1000 s/s, 2048 POINT FFT



TWO TONE TEST
400Hz @ 0dB (20V P-P)
380Hz @ -10dB
1000 s/s, 2048 POINT FFT



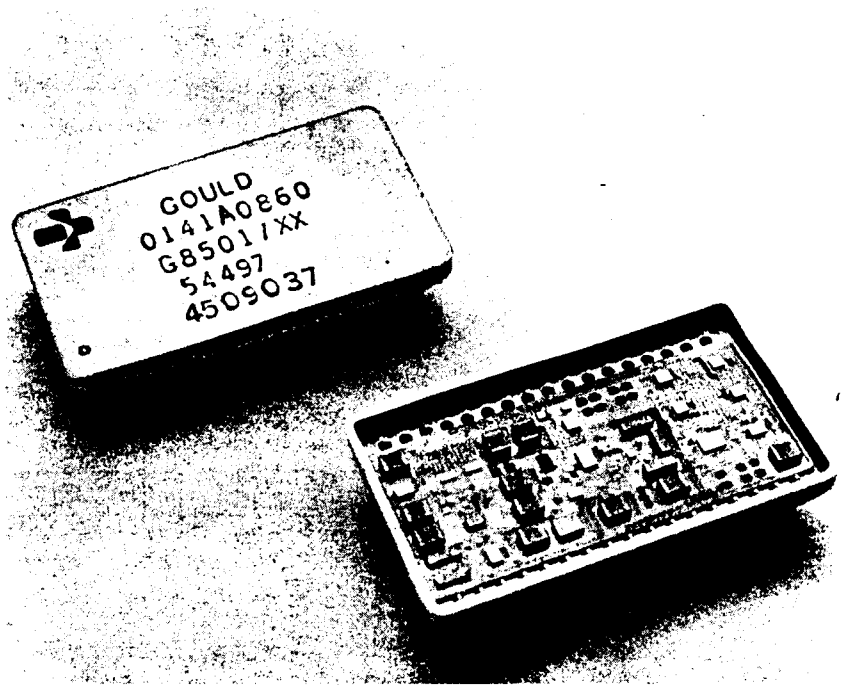
ENHANCED DELTA MODULATION ENCODER (EDME)* 0141A0860

FEATURES

- 120 dB broadband dynamic range
- A/D resolution in excess of 20 bits
- 0.00025% THD typical

APPLICATIONS

- Seismic measurements
- Oil exploration instrumentation
- Digital signal processing
- Imaging and graphics



PRODUCT DESCRIPTION

The Enhanced Delta Modulation Encoder (EDME) is a high accuracy, wide dynamic range Analog-to-Digital Converter (ADC) based on delta modulation. The EDME output is a serial, single-bit representation of the input at a high sample rate. Only seven external components are required to perform a 24-bit conversion when paired with the 0141A0863 FIR Filter.

Operating on ±10 volt supplies, the EDME will accept a full 20 volt peak-to-peak signal with all specs guaranteed. Three sample rates are available, externally programmable.

The EDME is available with an optional DC offset cancellation circuit (-01 version).

The device is manufactured using thick film hybrid microcircuit technology and is packaged in a 34-pin dual in-line package.

ORDERING INFORMATION

Suffix	Description
01	EDME W/DC Cancellation Circuit
04	EDME W/O DC Cancellation Circuit

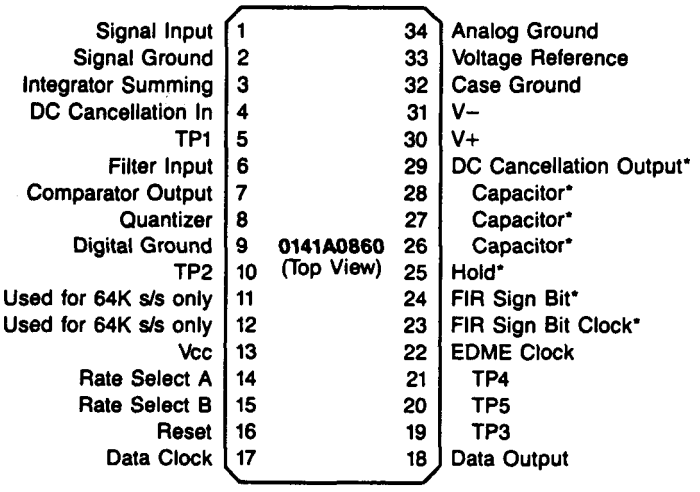
* *U.S. Patent No. 4,509,037, Canadian Patent No. 1,184,660*

ABSOLUTE MAXIMUM RATINGS

(CAUTION: Stresses above those listed may cause permanent damage to the device).

V+ to Analog Gnd	– +15V
V– to Analog Gnd	– –15V
Vcc to Digital Gnd	– +5.5V
Maximum Signal Input	– 20Vpp (+17 DBV)
Digital Inputs	– (GND–0.3V) to (Vcc+0.3V)
Power Dissipation	– 250 MW

PIN CONFIGURATION *(Not to scale)*



* *Pins are unused in the Suffix-04 EDME*

SPECIFICATIONS

T=–53°C to +85°C, V+=10V, V–=–10V, Vcc=5V unless otherwise indicated (See Test Circuit, Figure 1)

PARAMETER	MIN	TYP	MAX	UNITS	CONDITIONS	NOTES
INPUT NOISE (ground input)		–129.6 –126.6 –126.6	–126.7 –124.7 –122.7	dBv/√Hz dBv/√Hz dBv/√Hz	256K samples/sec 128K samples/sec 64K samples/sec	3-500 Hz* 3-250 Hz* 3-125 Hz* } DC cancellation circuit disabled
DYNAMIC RESOLUTION (instantaneous dynamic range)	–120			dB	all sample rates	w/10 dB SNR above input noise for 2 tones separated by specified level
HARMONIC DISTORTION		.00025		%	w/+17 dBV input signal	Level of highest harmonic referenced to the input signal
GAIN ACCURACY			±.056	%	+25°C	
GAIN STABILITY			±.2 ±.4	% %	+25°C at –53°C and at +85°C	
ANALOG INPUT IMPEDANCE		60		KΩ		
DC OFFSET	99.6			% of final value	measured at 500 msec after power up, 25° only, with final offset voltage voltage at one second	DC cancellation circuit disabled
DC OFFSET DC OFFSET SETTLING			–90 –74	dB dB	ref 17 dBv 20 seconds after power applied w/20MV signal.	DC cancellation circuit enabled. Applies to version -01 only.
HOLD CIRCUIT DRIFT RATE			1.0	mV/sec	measured 60 seconds after HOLD enabled.	

* When used with the 0141A0863 FIR Filter, system bandwidths are 3-410 Hz, 3-205 Hz, and 3-102.5 Hz, respectively.

SPECIFICATIONS

T=−53°C to +85°C, Vcc=+5V, D.C. ELECTRICAL (DIGITAL)

PARAMETER	MIN	TYP	MAX	UNITS	CONDITIONS	NOTES
Input Leakage Current	−1.0		+1.0	μA	GND≤VI≤Vcc	
Input/Output Leakage Current	−1.0		+1.0	μA	GND≤VIO≤Vcc	
Input Low Voltage	−0.3		0.8	V		
Input High Voltage	2.4		Vcc+0.3	V		
Output Low Voltage			0.4	V	IO=3.2 mA	
Output High Voltage	2.4			V	IO=−1.0 mA	
Input Capacitance			8.0	pF	VI=Vcc or GND f=1 MHz	
Input/Output Capacitance			10.0	pF	VIO=Vcc or GND f=1 MHz	
Supply Currents						
V+		9.2		mA	V+=+10V	
V−		8.3		mA	V−=−10V	
Vcc		2.9		mA	Vcc=+5V	

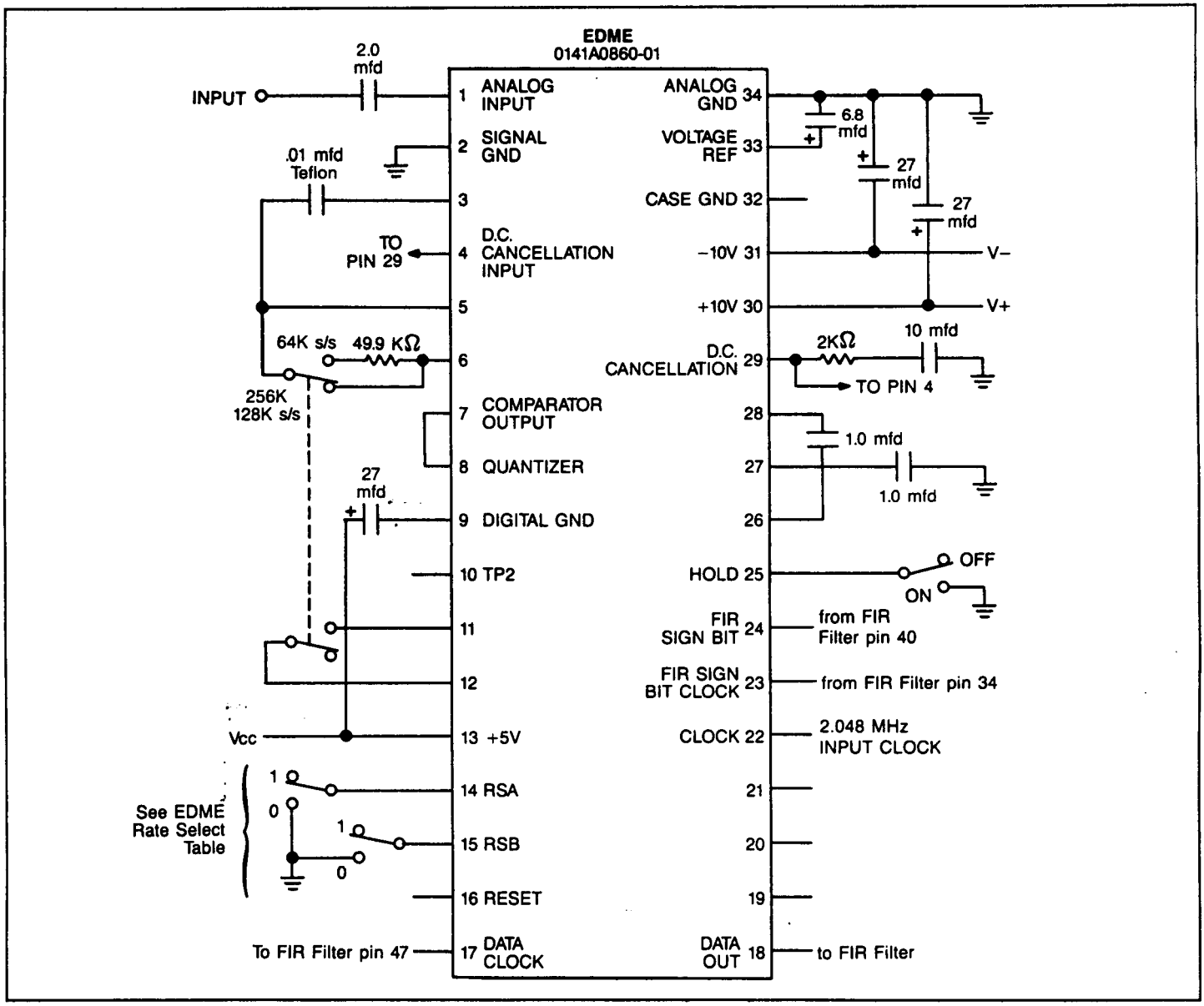


FIGURE 1. EDME TEST CIRCUIT W/ DC OFFSET CANCELLATION

EDME INPUT SIGNAL SPECIFICATIONS

Input Signal	Hybrid Pin No.	Specifications
V+	30	+10V ±5% at 20 mA Capacity, ±0.5% Load Regulation, Noise and Ripple less than 1 mV_{RMS}
V−	31	−10V±5% at 20 mA Capacity, ±0.5% Load Regulation, Noise and Ripple less than 1 mV_{RMS}
Vcc	13	+5V±10% at 5.0 mA Capacity, ±0.5% Load Regulation
Signal Input	1	$20V_{PP}=7.07V_{RMS}$ sine wave max., 3 Hz to 500 Hz at 256K s/s, 3 Hz to 250 Hz at 128K s/s, 3 Hz to 125 Hz at 64K s/s.
Clock	22	2.048 MHz±0.02%, 5V CMOS Logic Levels, min. 20 ns Pulse Width
Reset	16	Active Low, 5V CMOS Logic levels, min. 180 ns Pulse Width
Rate Select (A, B)	14, 15	Active High 5V CMOS Logic Levels. Refer to EDME Rate Select Table
FIR Sign Bit*	24	Active High, 5V CMOS Logic levels, min. 50 ns Pulse Width
FIR Sign Clk	23	Active High, 5V CMOS Logic levels, min. 20 ns Pulse Width
Hold*	25	Active Low, 5V CMOS Logic Levels

*Suffix -01 version only.

EDME RATE SELECT TABLE

RSA	RSB	SAMPLE RATE
0	0	64K samples/sec
0	1	128K samples/sec
1	0	Not Used
1	1	256K samples/sec

THEORY OF OPERATION

The Enhanced Delta Modulation Encoder (EDME) constitutes an improved method of analog-to-digital conversion based on delta modulation. The basic encoder has been enhanced by the addition of a filter which reduces the in-band noise. The output of the EDME is a serial, single bit representation of the input at a high sample rate.

A functional block diagram of the EDME circuit is shown in Figure 2. It consists of a DC offset cancellation circuit for suppressing DC, an integrator, a filter, a comparator, a flip-flop, and a current pulse generation circuit. The analog input signal is applied to the integrator together with the DC cancellation (bias) signal current and the feedback pulse current. The feedback pulses change polarity as required to keep the output of the integrator bounded to ±1 volt typical at a 256K sample rate. The integrator feeds a low pass filter which, because of its placement within the feedback loop, acts to reduce in-band noise. The filtered output is compared to the integrator output and a high/low decision is made by the comparator. The comparator's output is time sampled by the flip-flop. A new current pulse of the appropriate polarity is produced after the flip-flop is clocked. If the selected sample rate is 128K s/s, then two current pulses are injected into the summing junction of the integrator for every flip-flop clock. If the 64K s/s rate is selected, then four pulses are injected for every flip-flop clock. The new output bit is available with every flip-flop clock.

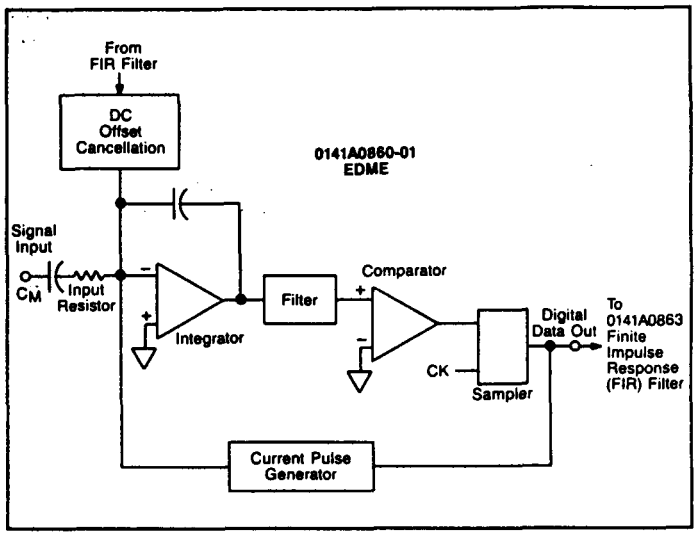


FIGURE 2. EDME BLOCK DIAGRAM

The optional DC offset cancellation circuit suppresses DC offset voltages in the EDME. The circuit requires the externally applied sign bit and sign clock from the 0141A0863 Finite Impulse Response (FIR) Filter, and results in reducing the DC offset voltage to −90 dB referenced to full scale input signal. The HOLD function disables the circuit and holds the output at a fixed DC voltage, allowing sensitive DC information to be obtained through the EDME. The DC offset cancellation circuit and hold function is available in the suffix-01 version of the EDME. It should be noted that optimum noise performance is achieved with the DC offset cancellation circuit disconnected.

I/O PINOUT DESCRIPTIONS

1 SIGNAL INPUT. Maximum allowable input signal on this pin is 20 volts peak-to-peak.

2 SIGNAL GROUND. Analog input signal ground.

3 A pin for the connection of an external capacitor.

4 INTEGRATOR INPUT. This pin is normally shorted to pin 29, the output of the DC cancellation circuit.

5 TP1. A pin for the connection of an external capacitor. Also normally connected to pin 6.

6 This pin is normally shorted to pin 5. When running the EDME at 64K samples/sec rate with full input signal, an external resistor is added between pins 5 and 6.

7 COMPARATOR OUTPUT. This pin normally connects directly to pin 8. It may be disconnected for trouble-shooting.

8 QUANTIZER INPUT. This pin normally connects directly to pin 7. It may be disconnected for troubleshooting.

9 DIGITAL GROUND.

10 TP2. Make no external connection.

11 This pin is shorted to pin 12 for the 64K sample rate only. Normally, there is no connection.

12 This pin is shorted to pin 11 for the 64K sample rate only. Normally, there is no connection.

13 Vcc (+5±0.5 VDC). An external bypass capacitor is usually connected between this pin and pin 9.

14 RSA. This pin allows for multiple sample rate selection. If an active high is applied to this pin and to pin 15 (RSB), then the EDME will operate at 256K samples/sec. Other sample rates are obtained by different combinations of logic levels applied to the pins.*

15 RSB. This pin allows for multiple sample rate selection.*

16 RESET. An active low on this pin will cause the EDME to reset. The signal must be asserted low for a minimum of 180 nanoseconds to reset the circuit. Operation resumes when this input goes high.

17 DATA CLOCK. This output clock is synchronized to the EDME data bit stream and is used to clock data into the FIR Filter on the rising edge.

18 DATA OUTPUT. This pin provides the encoded serial bit stream data output from the EDME.

19 TP3. Make no external connection.

20 TP5. Make no external connection.

21 TP4. Make no external connection.

22 CLOCK. This is the input pin for the 2.048 MHz clock.**

23 FIR SIGN BIT CLOCK. This input is the clock for reclocking FIR SIGN BIT data into the DC cancellation circuit. The clock originates at the FIR Filter.

24 FIR SIGN BIT. This input pin returns FIR SIGN BIT information from the FIR Filter, providing constant error correction to the DC cancellation circuit.

25 HOLD. An active low on this pin will cause the DC cancellation circuit to maintain its present state independent of the FIR SIGN BIT information.

26 A pin for the connection of an external capacitor.

27 A pin for the connection of an external capacitor.

28 A pin for the connection of an external capacitor.

29 DC CANCELLATION CIRCUIT OUTPUT. This pin is normally connected to pin 4, the input to the integrator. This pin also is used for the connection of an external resistor and capacitor.

30 +10±0.5 VDC. An external bypass capacitor is normally connected between this pin and pin 34.

31 −10±0.5 VDC. An external bypass capacitor is normally connected between this pin and pin 34.

32 CASE GROUND.

33 VOLTAGE REFERENCE. A test point for monitoring the internal 5 VDC level.

34 ANALOG GROUND. Power supply ground for the +10 VDC and −10 VDC supplies.

* Refer to EDME Rate Select Table, page 4.

** Refer to EDME Input Signal Specifications, page 4.

CONFIGURATION FOR 64K SAMPLES/SEC (FIGURE 3)

When the EDME is sampling at the 64K s/s rate, two additional connections are required for proper operations:

1. Short pins 11 and 12 together.
2. Add a 49.9KΩ resistor between pins 5 and 6.

By shorting pins 11 and 12, the stability of the filter is insured at the lowest sampling rate. The 49.9KΩ resistor between pins 5 and 6 decreases gain at the filter input preventing the filter from signal overload.

If the EDME is used in a multi-sample rate system, the use of relays is recommended to provide these connections. Analog switches are not recommended for this application as they will cause distortion. Keep all relays as close to the EDME as possible to minimize noise pickup.

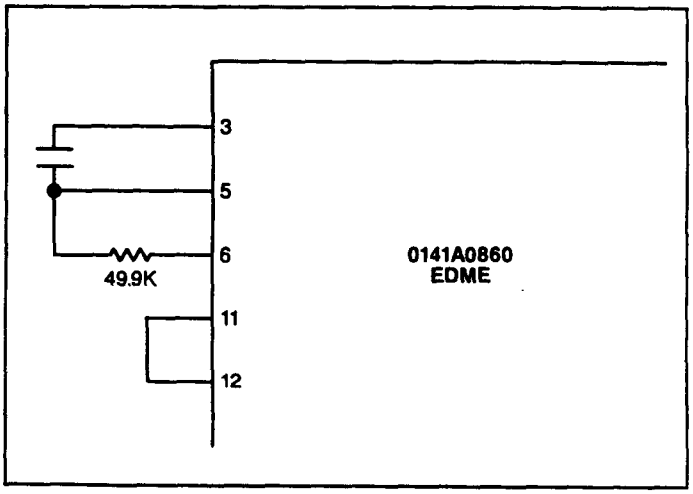


FIGURE 3. ADDITIONAL CONNECTIONS REQUIRED FOR 64K s/s OPERATION

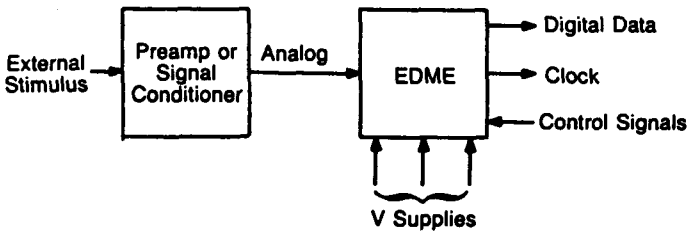
APPLICATION NOTES

For proper low noise, low distortion operation of the EDME, these recommendations should be followed:

1. Choice of integrator capacitor (pins 3 and 5) should be of the highest quality exhibiting low leakage and low dielectric absorption. Film capacitors such as teflon, polystyrene, polypropylene, and polycarbonate are suitable providing operating temperature and tempco requirements are met. Foil wrap are the best choice, followed by polypropylene.
2. A small, low leakage, solid tantalum filter capacitor should be connected between pin 33 (V_{ref}) and ground. Typically, a 6.8 μfd should be used, although values greater than 3 μfd with a working voltage rating ≥15 VDC should yield satisfactory results. Omission of this capacitor may result in degraded noise performance.
3. If the EDME signal input is AC coupled, a high quality capacitor, such as a polycarbonate, should be used. Note that this input capacitor is in series with the internal input resistor (60 K typical), forming a high pass filter. This filter could affect certain bandwidth applications of the EDME.
4. Proper grounding is paramount to low noise performance. Printed circuit boards should maintain as much ground plane as possible. Component lead and track lengths should be as short as possible. External components, particularly the integrator capacitor, should be located as close to the EDME as possible. The integrator

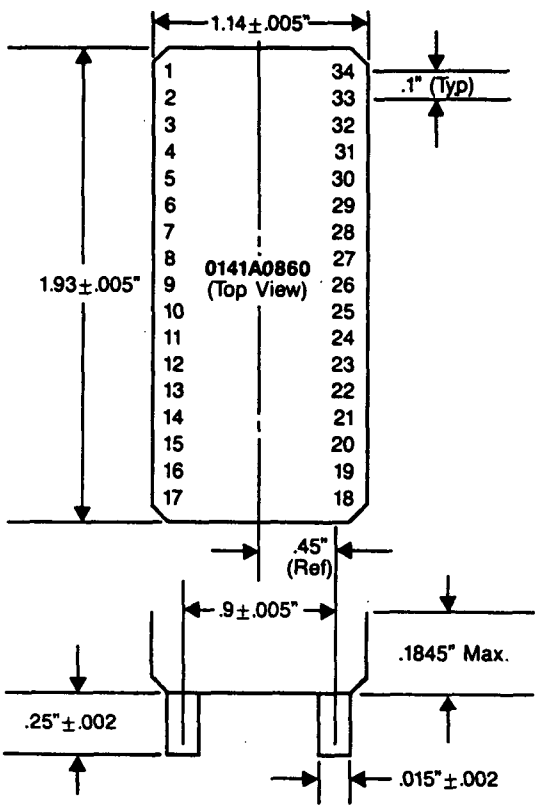
capacitor should have a ground plane underneath and should be kept away from digital signal paths.

5. The general recommended signal flow should be as shown below.

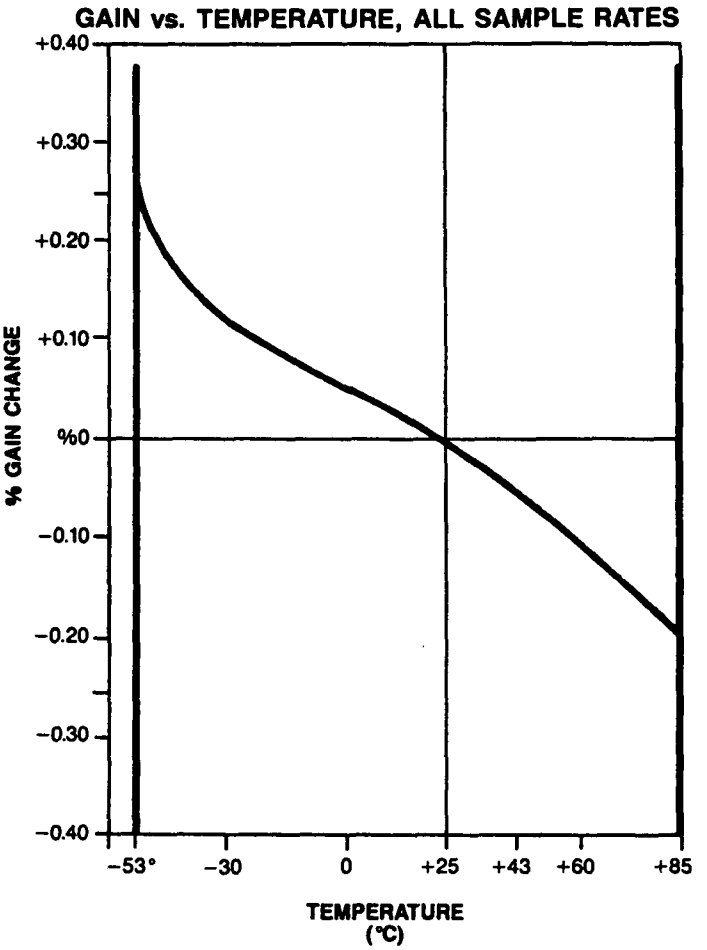


6. Power supply decoupling should be done as close to the EDME supply inputs as possible. Typically, 27 μfd tantalums are used here.

PACKAGE DIMENSION



TYPICAL PERFORMANCE CHARACTERISTICS



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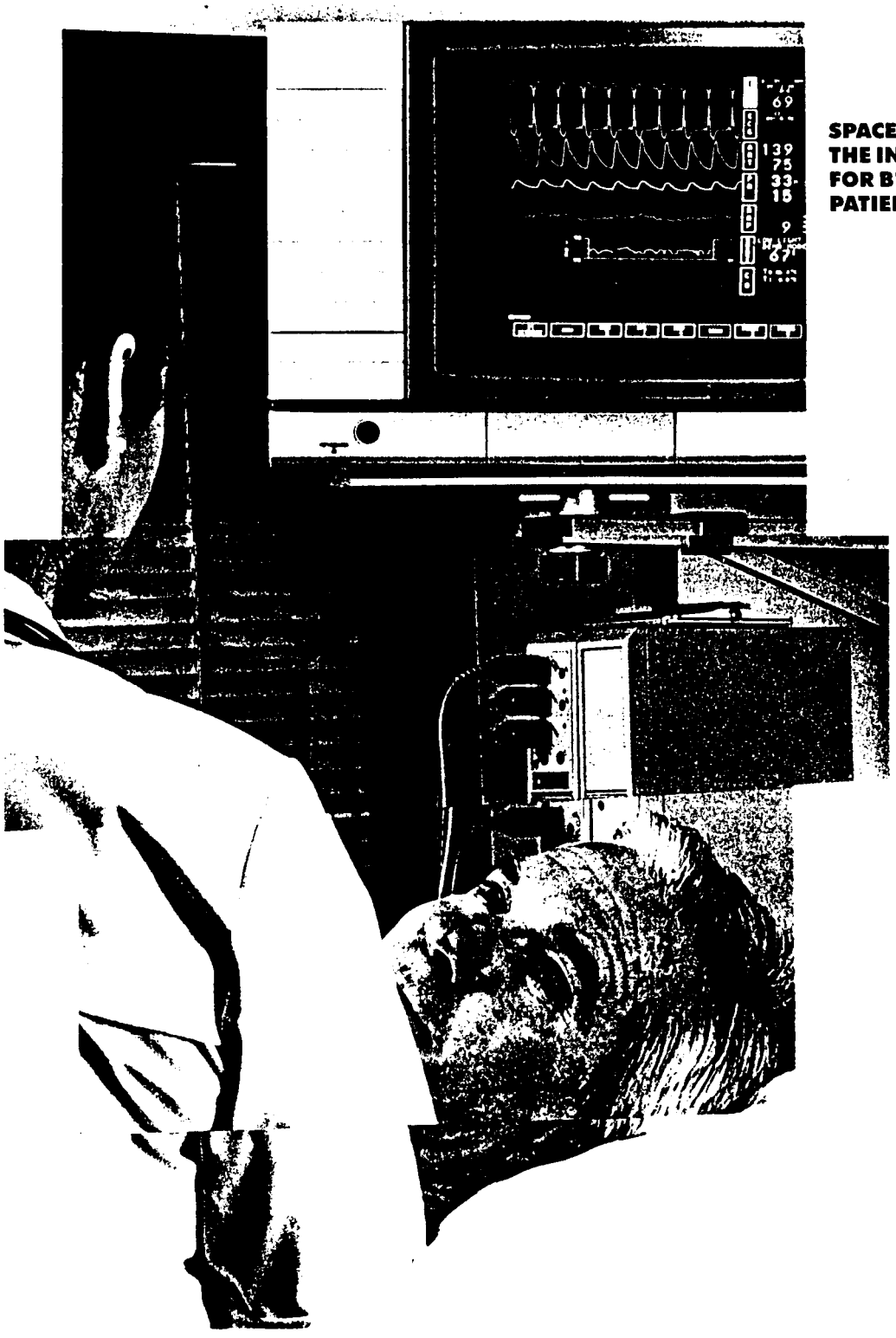

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90360
Option 01 Cordless Remote Control Keypad

Option 02 Receiver

The 90360 SpaceLabs PCMS Cordless Remote Keypad is an option for all SpaceLabs PCMS bedside monitors that lets the clinician remotely control the monitor from up to 20 feet away. The convenient, easy-to-use keypad, similar in size to a TV remote control, accesses both hard and soft keys on the monitor. The clinician can use the keyboard to suspend or adjust alarms, access graphic trends, adjust waveform size, initiate recordings, or perform any other monitor function. And because the keypad communicates with the monitor via infrared technology, it requires no cords or cables.

Special function keys on the unit permit rapid access to all monitor menus and functions. A zoom function enlarges menu soft keys, making them easy to read at distances of up to 20 feet.

The 90360 consists of two components: the keypad itself and a receiver mounted on the side of the monitor.

Option 01–Keypad

The keypad contains a splash-proof transmitter, powered by a 9-volt alkaline or mercury battery. A silicon rubber key panel allows easy cleaning.

Option 02–Receiver mounted on the side of the monitor

The receiver is powered by a coil cable that plugs into the keyboard plug on the monitor.

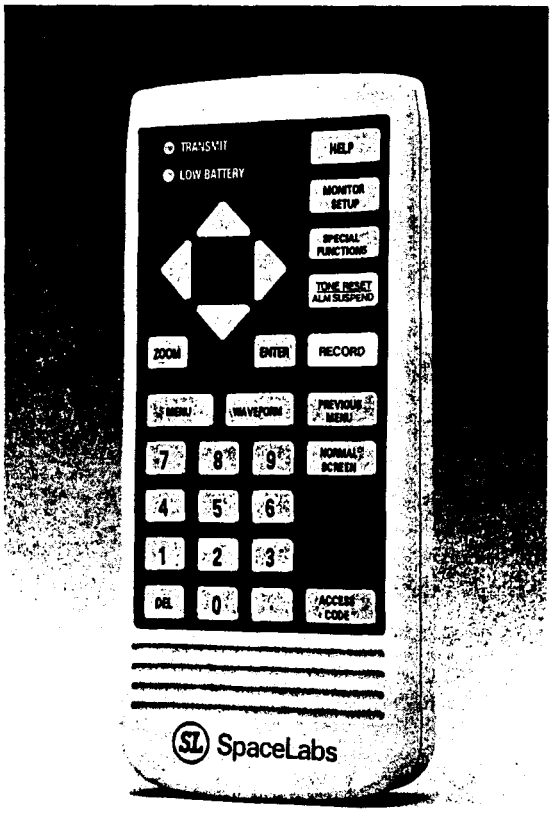
User Programmable

The keypad can operate with or without the use of transmission codes. If several monitors are located in close proximity, the use of transmission codes ensures that commands go to the selected monitor. With 32 user-programmable codes available, the keypad can distinguish among as many as 32 monitors. For rooms with just one monitor, the keypad can also be operated without transmission codes. In either case, the unit's unique transmission methods prevent it from being affected by other infrared remote control units, such as TV controls.

Field Upgradeable

Existing monitors can be field-upgraded to take advantage of the new remote control unit.

- ☐ Keypad allows PCMS monitors to be remotely controlled from up to 20 feet away
- ☐ Infrared technology allows monitor communication without cords or cables
- ☐ Small size (similar to a TV remote control) makes unit convenient to carry
- ☐ Keypad permits rapid access to all monitor menus and functions
- ☐ Zoom key enlarges menu soft keys for easy readability
- ☐ 32 user-programmable security codes ensure communication is directed to designated monitor



90360 Cordless Remote Keypad and Receiver

Specifications

System

TRANSMISSION METHOD – Infrared.
OPERATING RANGE – 20 ft.
OPERATING ANGLE – 45 degrees off center line.
SECURED MODE – 32 user-programmable access codes available in transmitter and receiver.
UNSECURED MODE – Receiver can be programmed to be open to any 90360 transmitter.
MENU ZOOM – Bottom of screen menu line can be zoomed to be readable at 20 ft.

Transmitter

KEY-PAD – 28 keys in a sealed silicon rubber pad.
POWER – 9V alkaline or 8.4V mercury battery. Good for 150,000 key strokes.
STATUS INDICATORS – Transmit and low-battery lamps.
CONSTRUCTION METHOD – Gasketed and splash-proof.
DIMENSIONS –
DEPTH – 5.5 in. (13.97 cm)
WIDTH – 2.4 in. (6.10 cm)
HEIGHT – 0.9 in. (2.29 cm)
WEIGHT – approx. 5 oz. (142 g)
ORDER SEPARATELY – 90360-01

Receiver

DISPLAY – 2-digit LED readout for access code.
POWER – Powered by host PCMS monitor via keyboard connector.
AUXILIARY CONNECTOR – Includes auxiliary connector for SpaceLabs PC Mode keyboard, so keyboard and keypad can be connected to monitor simultaneously.
MOUNTING METHOD – Bracket and 3M Dual Lock fastener provided.
DIMENSIONS –
DEPTH – 5.8 in. (14.73 cm)
WIDTH – 1 in. (2.54 cm)
HEIGHT – 1.4 in. (3.56 cm)
WEIGHT – approx. 6 oz. (170 g)
ORDER SEPARATELY – 90360-02

WARRANTY

SpaceLabs warrants their equipment for 12 months from the date of installation for products requiring installation or 12 months from the date of delivery if installation is not required.

All specifications are subject to change without notice.

SpaceLabs Inc.
4200 150th Ave. N.E.
P.O. Box 97013
Redmond, Washington 98073-9713
(206) 882-3700

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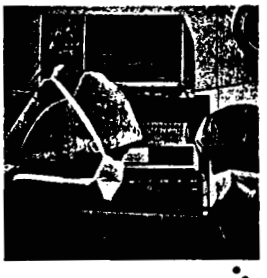
PCMS NETWORK

INFORMATION WHERE AND WHEN IT'S NEEDED, FOR QUICK RESPONSE TO CHANGING CLINICAL DEMANDS



SpaceLabs is proud to lead the way in clinical communication capabilities for healthcare. For example, SpaceLabs technology makes it possible to monitor patients in *different* locations at the *same* time. "Remote view" and "remote alarm watch" capabilities mean you can quickly respond to one emergency even while handling another—thereby saving precious time and steps.

In fact, the SpaceLabs Patient Care Management System (PCMS) acquires and integrates *more* information with *less* interaction, leaving more time for direct patient care. PCMS enhances staff productivity, and—ultimately—the quality of patient care by linking important data centers throughout the hospital into a single network.

The Ethernet™ Local Area Network offers more capacity and greater speed than any other network. Manual processing of data is dramatically reduced, because raw data from a variety of sources is automatically converted to useful information. And cost-containment is further enhanced by the flexible design of the SpaceLabs PCMS, which allows the system to expand economically as your needs change and grow. Modularity permits easy, economical integration of new products to your existing PCMS system and protects your long-term investment.

With SpaceLabs, you get the confidence of working with a specialist in healthcare and clinical information that has more than 32 years of monitoring experience, as well as the largest installed base of networked monitors.

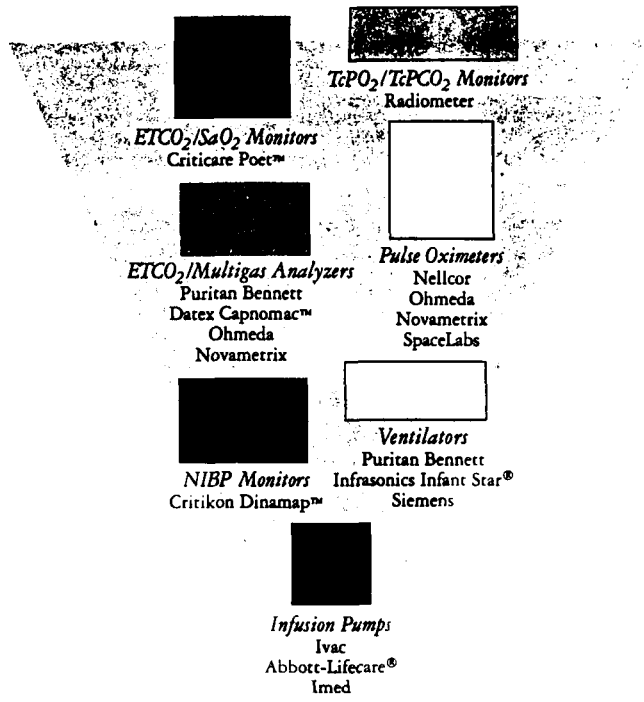
INFORMATION INTEGRATION

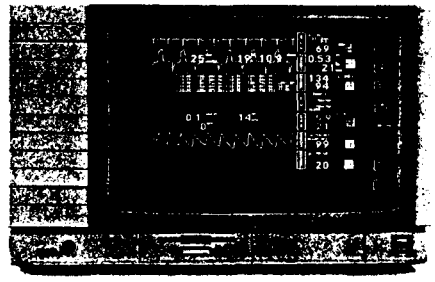
UNIQUE CAPABILITIES PROVIDE A MORE COMPREHENSIVE PICTURE OF EACH PATIENT'S STATUS

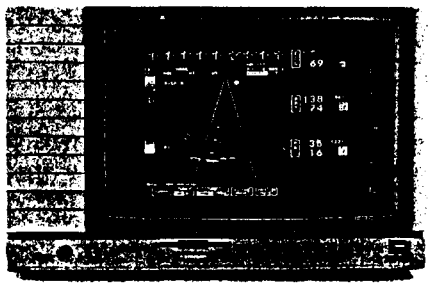
FLEXPORT™ integrates vital patient information from many sources for quick review, better decisions

Excellent patient care demands smooth, efficient integration of all relevant data and observations. *All* SpaceLabs monitors are designed to integrate data at bedside as well as at the central station monitor. In addition, Flexport serial interfaces integrate data from ancillary patient care devices with the SpaceLabs Patient Care Management System (PCMS); this data from other devices is accepted by the system just as data from SpaceLabs modules, creating a more complete picture of patient status. As a result, the relationship between different monitored parameters becomes much easier to determine—and clinical decisions become much easier to make.

The Flexport information integration benefits don't stop there, however. Flexport improves your return on investment by allowing you to use existing stand-alone devices. On the other hand, as new technological advances are introduced, Flexport can swiftly and easily accommodate them—and you—to further enhance data management capabilities.







ULTRAVIEW™ option integrates ultrasound images and vital sign data on a single display

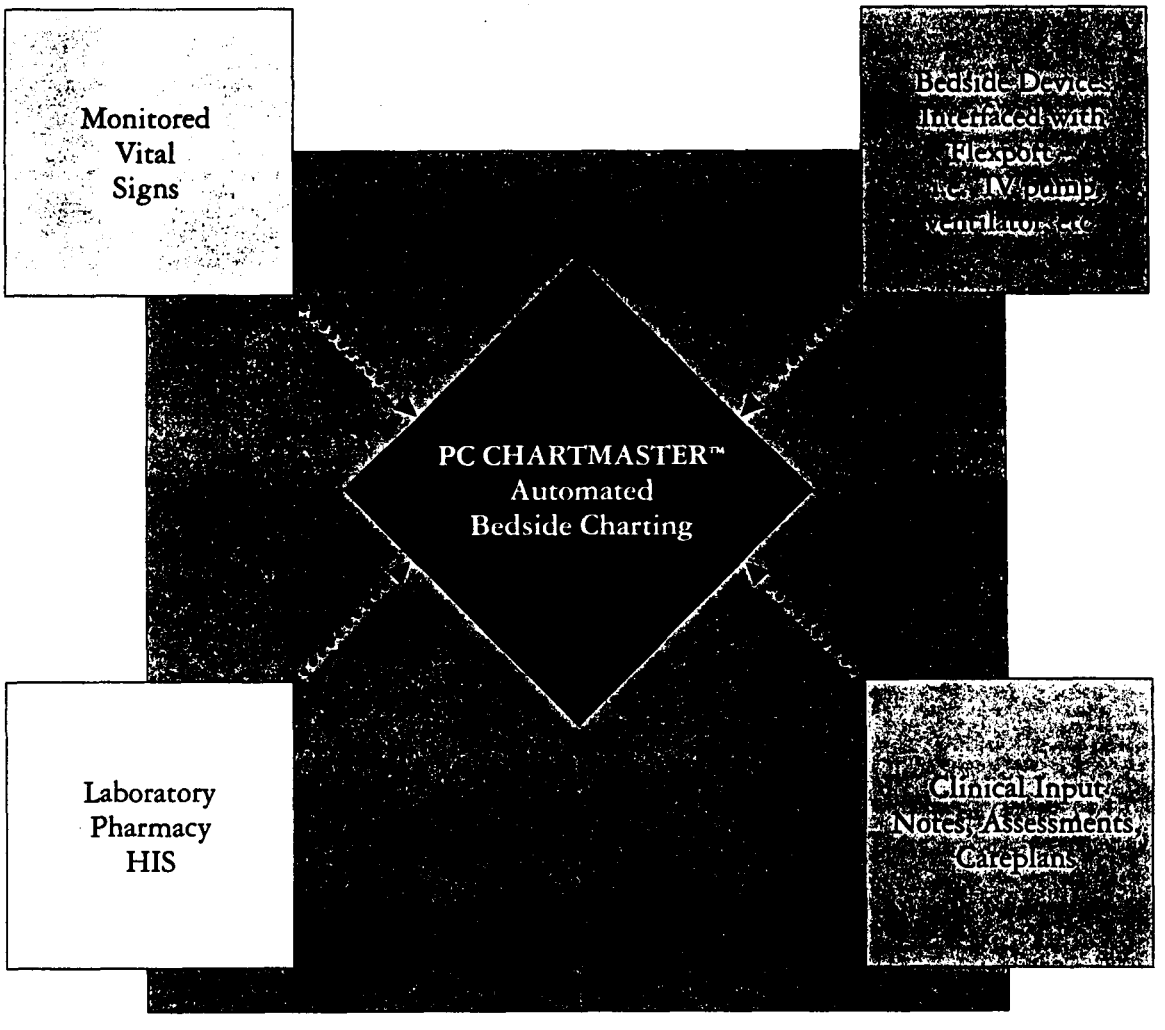
SpaceLabs' exclusive UltraView option allows the anesthesiologist and the surgeon to view real-time ultrasound/echo images and vital sign data on the patient monitor *and* remote display during surgery. Designed to support the growing interest in TEE as a monitoring modality in the OR, UltraView demonstrates the power and advantages of SpaceLabs' PCMS architecture. It is compatible with any ultrasound system with standard video outputs, and its convenient touchscreen controls assure easy setup.

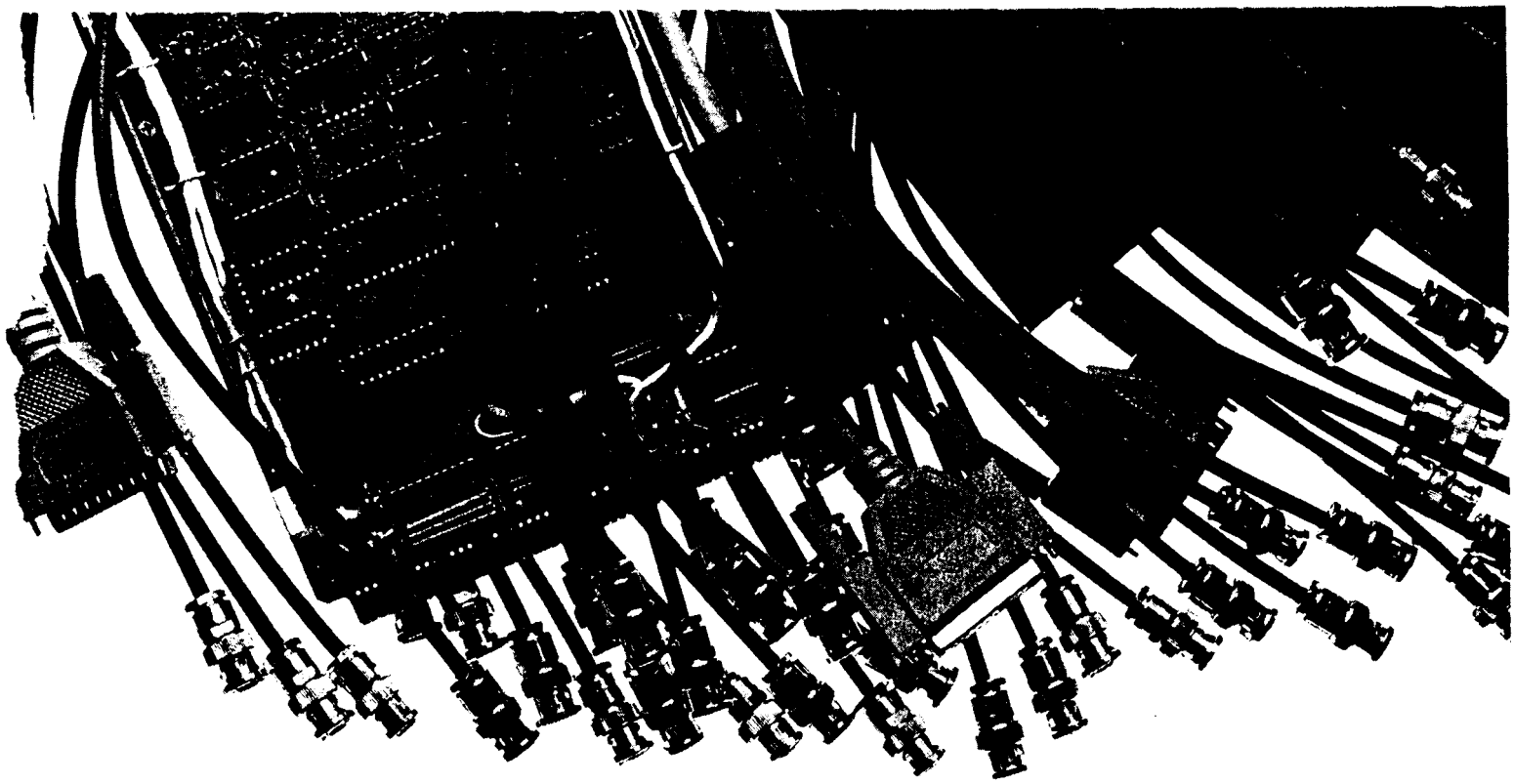
AUTOMATIC, COMPREHENSIVE CLINICAL INFORMATION SYSTEMS BACK UP CLINICAL DECISIONS WHILE REDUCING PAPERWORK BURDEN

In the critical care environment, access to accurate and timely information is essential for optimum patient care. SpaceLabs' Clinical Information System meets these needs in a number of ways.

First, it integrates information from patient monitors, other bedside devices, laboratory, pharmacy and other ancillary areas of the hospital—providing the comprehensive information clinicians need when making patient care decisions. And then, it automates much of the patient chart, reducing nurses' clerical chores and allowing them more time for direct patient care. The Chartmaster high performance workstations bring speed, accuracy and convenience to the charting process, enhancing nursing care and quality assurance.

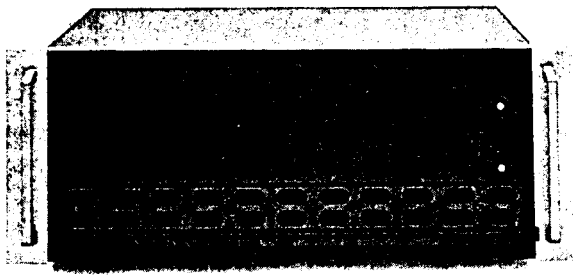
SpaceLabs' PC Chartmaster is an integrated system that can automatically acquire data from the four major sources of patient information.





YOU CAN SAVE A BUNDLE BY PUTTING AMPLIFIERS, FILTERS AND A/D'S IN THE SAME BOX.

Save on cabling, rack space and slots in your backplane. Save on engineering and installation, too. We integrate 128 channels of amplifiers, filters, SSH and A/D's into one Precision 128 DAS. You get one system, one software package, resolution to 16 bits and throughput to 2.6 MHz. Auto-zero standard. End-to-end calibration and diagnostics optional. A great value any way you look at it.



Precision 128 DAS

You can have up to 256 channels of data and control via a single-slot bus adapter card for VME, VXI, Q-bus and Futurebus. Expansion for larger systems with multiple bus adapter cards.

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Circle 94

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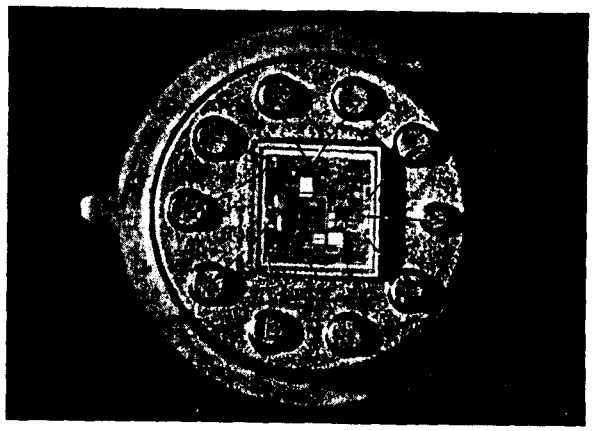
Surface-micromachined acceleration sensor includes on-chip signal conditioning

The ADXL-50 is a surface-micromachined acceleration sensor and signal conditioner. The accelerometer combines a silicon membrane formed by chemical etching with thin-film piezoelectric resistors connected as a bridge circuit. In operation, acceleration exerted on the device deforms the membrane. This deformation results in a change in the resistance of the piezoelectric resistors and produces a small output from the bridge circuit.



The device operates within a force/balance electronic control loop. Acceleration destroys the balance. On-chip circuitry amplifies, demodulates, and filters the differential signal to produce a 0.25 to 4.74V output proportional to the acceleration.

The device operates over the −55 to +125°C temperature range, and its capacitive sensing is essentially immune to temperature variations. This sensing allows operation over the frequency range of dc to 1 kHz. The monolithic device has a guaranteed accuracy to within 5% over its ±50g range. A self-test feature assures the user that the accelerometer is functional.

The ADXL-50 comes in a 10-pin, TO-100 metal can. The device costs $23 (100); in automotive OEM quantities, it costs $5.

Analog Devices, 181 Ballardvale St, Wilmington, MA 01887. Phone (617) 937-1428.