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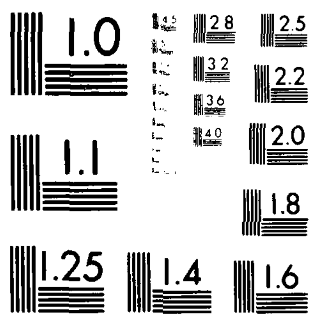

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OPTICALLY ISOLATED ECG AMPLIFIER WITH BASELINE STABILIZATION

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An electrocardiographic (ECG) amplifier that is optically isolated and suitable for use in all environments has been developed and tested in 100 experiments involving immersed humans during exercise and rest. The amplifier has less than a 10 μA current leakage and employs an improved surface electrode placement technique. The frequency bandwidth of the amplifier easily can be varied to yield the optimum signal fidelity and signal-to-noise ratio for a given monitoring situation. The system incorporates low-power integrated circuits, which facilitate construction, provide inherent high reliability, and increase

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SECURITY CLASSIFICATION OF THIS PAGE (When Data Entered) the useful lifetime of the battery power pack.

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OPTICALLY ISOLATED ECG AMPLIFIER
WITH BASELINE STABILIZATION

W. H. Mints, Jr. and W. E. Long

INTRODUCTION

Design of physiological instrumentation suitable for monitoring a diver's well-being is complicated by increased pressure, wetness, and various breathing gas mixtures, all elements of the diving environment. Of particular importance in monitoring the working diver is the detection of cardiac arrhythmia. The electrocardiograph (ECG) is most useful in detecting cardiac arrhythmia, but it requires a high degree of signal fidelity (Kanwisher, Lawson, and Strauss 1974; Unsworth, Williams and Tayler 1969) that is difficult to obtain in the wet environment. When available commercial ECG amplifiers and standard techniques for applying electrodes were used, the task of monitoring was difficult because of water penetration into electrode sites, muscle activity, and electrode movement artifacts, along with 60-Hz interference from AC power supplies. To avoid these problems, we have developed in this laboratory a miniature ECG amplifier and improved techniques (Hoar, Langworthy, Mints, Long and Raymond 1976) for placing surface electrodes. Through the use of this amplifier and the improved electrode placement techniques, more accurate ECG recordings from an immersed diver are possible.

METHODS

Our first attempts to monitor the ECG on active divers gave unreliable data because of the shifting base line and noise artifacts. At that time, ECG amplifiers available commercially were inadequate to fit the task because of high leakage current, large physical size, poor noise rejection, low input impedance, unstable base line, and high power consumption. These inadequacies necessitated the design of our own amplifier, tailored to the specific requirements of the working diver.

As we began the design of our amplifier, techniques for limiting leakage currents were investigated first, because safety of the human subject is always the primary concern. The method we initially selected (because of its simplicity and reliability) employed diode clamping and a series current-limiting resistor that limits supply leakage current to less than 10 µA, the value suggested by the American Heart Association (Pipberger 1975).

This clamping method offers safe current limits from the DC supply voltages of the amplifier, but does not provide adequate protection against the most common electrical shock, 60-Hz AC, which can be avoided only by breaking the AC ground path to the subject. Two forms of electrical ground isolation were considered, transformer and optical isolation. Optical isolators, the more recent innovation, were selected because they provide true electrical isolation with no feedback path, regardless of frequency. They are simpler to implement and can be made smaller in size and weight than their transformer counterparts; both features are important advantages for portable ECG monitors.

The major design problem of optical isolators is the nonlinear transfer function, which was corrected by incorporating matched optical devices into the input and feedback circuits of an operational amplifier. Linearity of better than 1% from DC to 1 KHz is realized in this configuration. The circuit utilizes MCD2 photodiode optoisolators (Monsanto Commercial Products, Cupertino, CA) (Fig. 1), which provide electrical isolation up to 1500 volts. The device has a DC resistance of 10^{14} ohms and a low coupling capacitance of 0.6 pF, which provides excellent electrical isolation from amplifer input terminals to its output terminals.

The input stage of the amplifier consists of two low-power supergain operational amplifiers, LM212 (National Semiconductor Corp., Santa Clara, CA). These amplifiers provide a true differential connection that gives an input resistance greater than 10^{10} ohms without the use of high-resistance feedback circuitry. The configuration facilitates resistor matching, which is required for good common-mode rejection (CMR), the ability to reject noise common to both input leads. Only four precision resistors ($\pm 0.1\%$) are necessary to obtain a CMR better than 80 dB.

The remaining three integrated circuits in the device are programmable operational amplifiers Model LM4250 (National Semiconductor Corp.). The low bias current, set at 1.0μ A for minimizing power consumption and increasing input resistance, also lowers the gain-bandwidth product and results in decreased amplifier noise. At 1.0μ A the LM4250 has a standby power consumption of only 330μ W; the entire ECG amplifier has a quiescent power consumption of only 40 mW (National Semiconductor 1973). These features provide a life expectancy of 100 h for the two U10 batteries (Burgess Division, Gould Inc., St. Paul, MN).

A low-pass two-pole active Butterworth filter is used to improve base-line stabilization (Melsheimer 1967; Scott 1966). The output from the low-pass filter A4 and the DC coupled differential amplifiers A1, A2, are applied to a difference amplifier A3, where the base-line tracking signal is subtracted from the composite ECG signal. Amplifier A3 also acts as a low-pass filter and imparts a gain of 20 to drive the light-emitting diode

(LED) of the optical isolator. All frequencies from DC to an upper value (f_L), which is a compromise between the opposing demands of signal fidelity and base-line stability, are resistor-selectable so that they can be switched easily, or permanently set for a specific application. For the monitoring of a working diver, an F_L of 0.5 Hz provides adequate fidelity for diagnosis of cardiac arrhythmia while maintaining base-line artifact at an acceptable level. This is not an adequate low-frequency response for precise S-T segment measurements, which can require an f_L as low as 0.05 Hz (Berson and Pipberger 1966).

RESULTS AND DISCUSSION

Fig. 2 illustrates the effect of low-frequency filtering on the ECG signal. The subject was sitting at rest in the laboratory at 24°C room temperature. The low-frequency cutoff, f_L, was changed by resistor selection of the two-pole active filter A4. Incremental values of f_L were selected between DC and 10 Hz with no apparent change in waveform for values for F_L less than 1.0 Hz, with 2.0 Hz showing slight alterations and 10 Hz showing extreme distortions.

The ECG amplifier described has been used successfully for several years under a variety of environmental conditions, including hypothermic immersion studies, in which the subject was both working and shivering (Fig. 3). The amplifier has provided ECG tracings with minimal electrical base-line shifts and little or no noise artifact. The fidelity of the device allowed adequate identification of all components of the ECG waveform. The low-frequency cutoff can be selected to provide accurate measurement of the ECG S-T segment, which is an important consideration for monitoring exercise effects or cardiac function (Berson and Pipberger 1966). The small size of the device enables the circuit to be incorporated into the ECG lead junction block, which reduces high-impedance lead length to a minimum and greatly diminishes its susceptibility to noise pickup.

A number of methods for data transmission from immersed humans have been devised (Kanwisher et al. 1974; Slater, Bellet, and Kilpatrick 1969); each has its advantages and disadvantages. In our particular application, a hardwire umbilical technique was selected for continuous subject monitoring. Because the diver was tethered, we were able to avoid the complexity of ultrasonic telemetry (Unsworth et al. 1969). Regardless of the data transmission path, this ECG amplifier can be utilized with slight modification because it is essentially a universal instrumentation amplifier with true electrical isolation. The circuit was also used successfully in another study as an electroencephalogram (EEG) amplifier for recording brain waves of dogs; it gave good quality EEG recordings in a high electrical noise environment.

Since the conception of our design, several small hybrid amplifiers have come on the market: the Analog Device Model 284J (Analog Devices, Inc., Norwood, MA) and the Burr Brown Model 3652 (Burr Brown, Tucson, AZ) are two such amplifiers. These commercial amplifiers have excellent electrical characteristics and easily can be incorporated into an ECG amplifier design. The complete amplifier will be larger and have a higher power consumption than the design described herein, but it may be an attractive alternative if space and low power consumption are not critical requirements.

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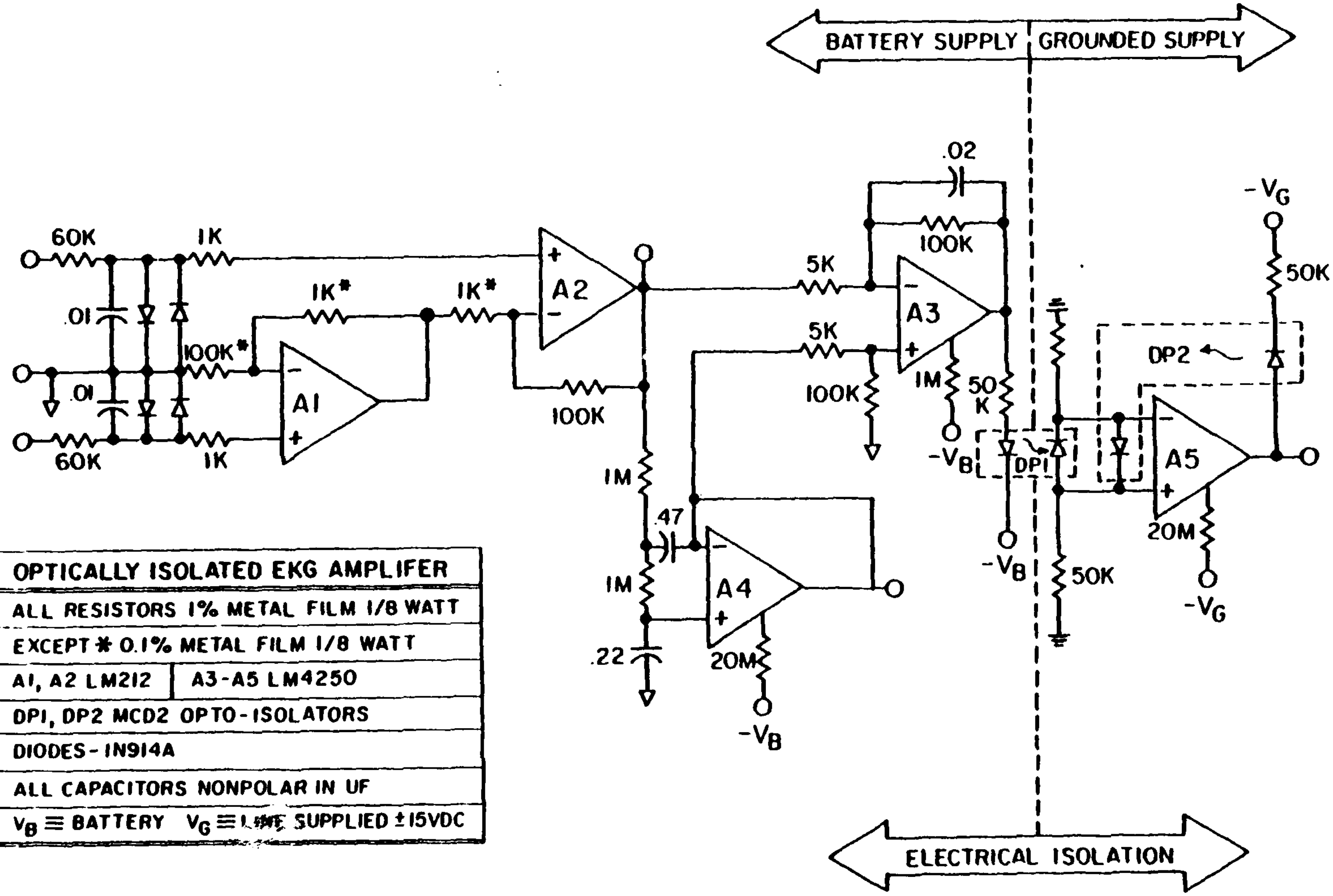
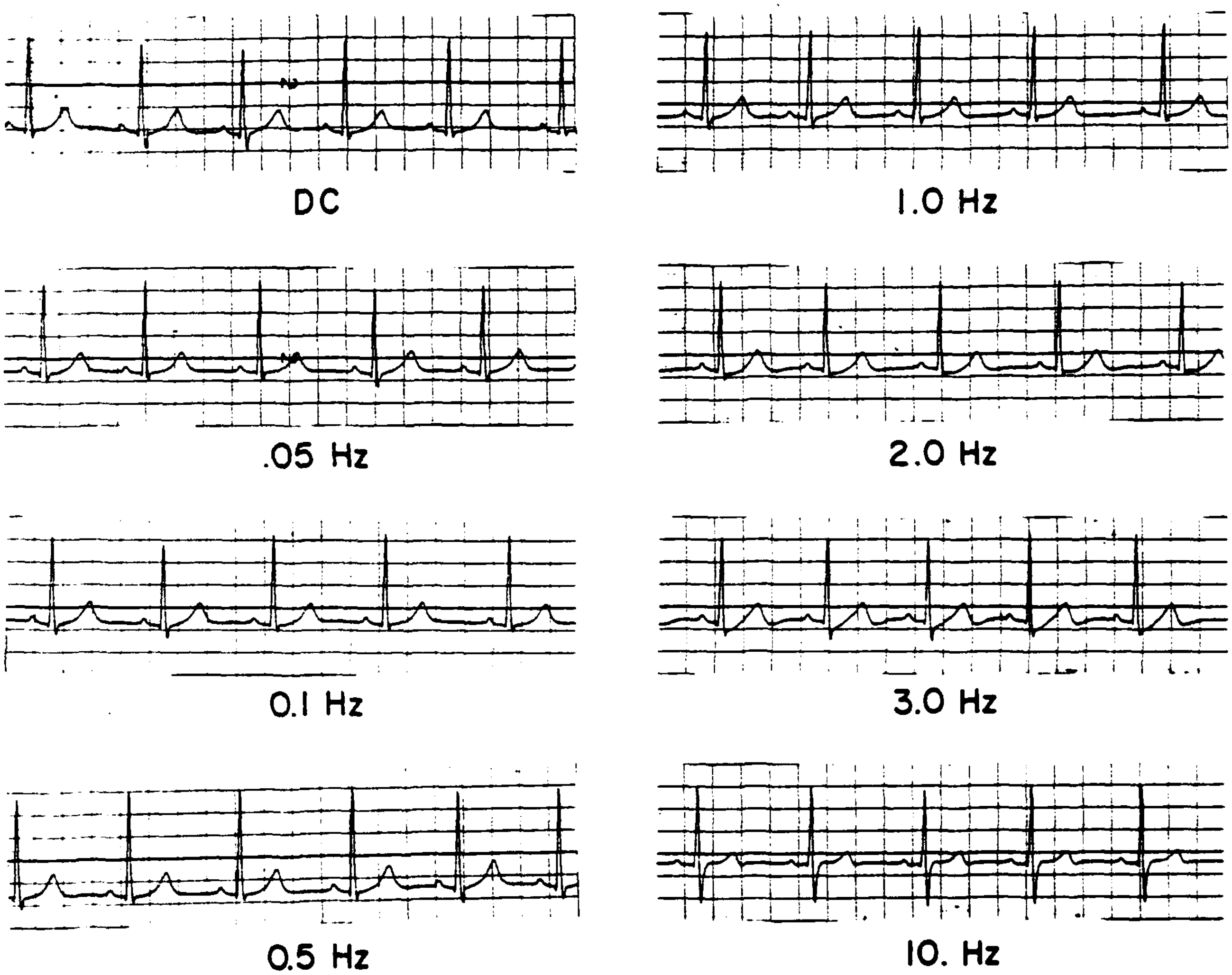


EKG CIRCUIT

Fig. 1. Schematic diagram of the ECG amplifier.



EFFECT OF LOW FREQUENCY CUTOFF "f_L"
ON THE EKG SIGNAL

Fig. 2. Effects of low-frequency cutoff (f_L) on the ECG waveform are most apparent in the S-T segment. No noticeable change can be detected for F_L 1.0 Hz.

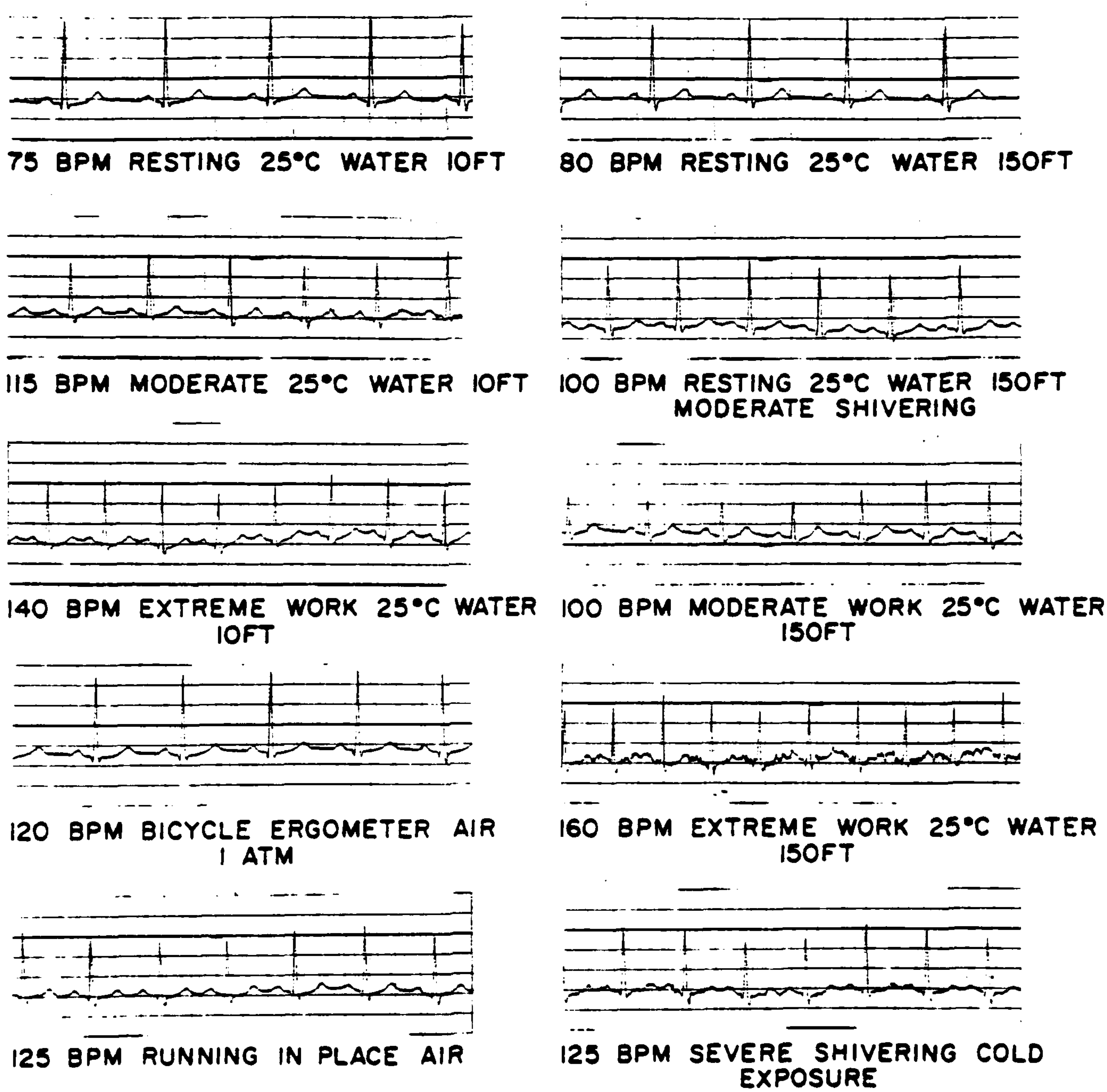


Fig. 3. Typical ECG tracings are shown under a variety of conditions. Tracings were made on a Gould Model-220 strip-chart recorder. The recorder settings were for all recordings: chart speed--25 mm/s; sensitivity--1 mV/cm.

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