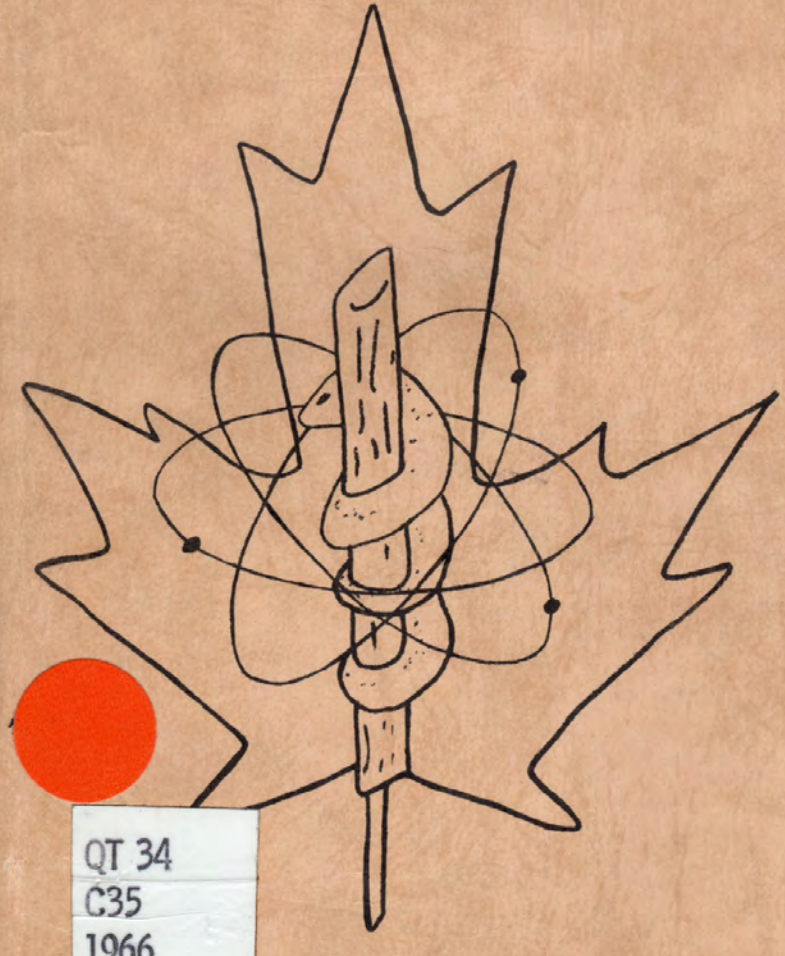
DIGEST OF THE 1ST CANADIAN MEDICAL AND BIOLOGICAL CONFERENCE



SEPTEMBER 8 AND 9, 1966

OTTAWA, CANADA

SPONSOR

Canadian Medical and Biological

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Acknowledgement

The financial assistance of the Medical and National Research Councils of Canada is gratefully acknowledged. The Conference is also indebted to the Radio & Electrical Engineering Division of NRC for the use of its building facilities.

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FOREWORD

This volume contains the digests of papers to be presented at the 1st Canadian Medical and Biological Engineering Conference to be held in Ottawa, Ontario, September 8th and 9th 1966. This conference is sponsored by the newly formed Canadian Medical and Biological Engineering Society.

The number and variety of papers submitted were such that we were unable to arrange the program to include all of them. Although we regret this restriction it was very encouraging to receive the response that we did to our request for papers. The papers contained herein will attest to the fact that Canada is making a very worthwhile contribution to the field of Bio-medical Engineering.

Off-set printing has been used in preparing this digest. No attempt was made to edit the papers presented. We would like to thank all of the authors who submitted papers for their kind cooperation and the assistance they have given in making this conference a success. In particular we would like to thank our lone American contributor. We hope that he finds our hospitality such that he will encourage his colleagues in the United States to present more of their work at subsequent conferences.

Extra copies of this digest are available for a nominal sum of $2.00 per copy by writing to me c/o Systems Study Group, University Hospital, Saskatoon, Saskatchewan, Canada.

For the Papers, Publications and Publicity Committees

B.A. Holmlund, Digest Editor

CONFERENCE SCHEDULE

OPENING SESSION

Thursday, 8th Sept., 9:00 a.m.

Auditorium, Radio & Elect. Bldg.

N.R.C., Montreal Rd.

Chairman: J.A. Hopps, President, Canadian Medical and Biological Engineering Society

9:00 a.m Welcoming Address: Dr. W.G. Schneider, Vice-President (Scientific), National Research Council.

9:15 a.m. Opening Address: Mr. L.E. Flory, Secretary General of the International Federation for Medical and Biological Engineering.

"Twenty Years of Medical and Biological Engineering".

SESSION I

Thursday 8th Sept., 10:00 a.m.

Auditorium

Chairman: J.A. Hopps, N.R.C. Ottawa.

10:00 a.m. Invited Tutorial Paper: Dr. J.H. Milsum, Professor of Control Engineering, McGill University.

"Biological Control System Modelling"

10:45 - 11:00 a.m. Coffee Break

SESSION IA

Thursday 8th Sept., 11:00 a.m.

Auditorium

"Modelling and Simulation"

Chairman: Dr. E. Llewellyn Thomas, Institute of Bio-Medical Electronics, University of Toronto.

11:00 a.m. (a) "Physiological Experimentation and Modelling of the Auditory System", C.A. Laszlo & J.H. Milsum, McGill University.

(b) "A Model of the Cardiac Electrophysiological System", F.A. Roberge & R.A. Nadeau, University of Montreal.

(c) "Digital Computer Simulation of Frog Sciatic Nerve", R.W.J. Ford & J.A. Tanner, N.R.C., Ottawa.

(d) "Measurement and Simulation of Handwriting Signals", J.S. MacDonald, U.B.C.

SESSION IB

Thursday 8th Sept., 11:00 a.m.

Conference Room

"Biomedical Instrumentation"

Chairman: O.Z. Roy, N.R.C. Ottawa.

11:00 a.m. (a) "Regional Hypothermia by an Extracorporial Arterial Shunt", T.R. Ringer & E.W. Peterson, N.R.C., Ottawa.

(b) "A Canadian Approach to the Problem of Haemodialysis Machinery", F.H. Siemonsen, Kingmed Ltd., Kingston.

(c) "Device for Assessing Intracranial Pressures", J. Wade, G. Wacker, J.W. Gerrard & B.A. Holmlund, University of Saskatchewan

(d) "An Automatic Blood Coagulation Timer", F. Engler, Litton Systems, Rexdale.

12:45 p.m. Lunch at the N.R.C. cafeteria (included in the registration fee).

SESSION II

Thursday, 8th Sept., 2:00 p.m.

Auditorium

Chairman: Professor R.N. Scott, Director, Bio-Engineering Institute, University of New Brunswick.

2:00 p.m. Invited Tutorial Paper: J.Foort, Director, Prosthetics Orthotics R & D Unit, Rehabilitation Hospital, Winnipeg.

"Lower Extremity Prostheses"

SESSION IIA

Thursday 8th Sept., 2:45 p.m.

Auditorium

"Prostheses"

Chairman: Professor R.N. Scott, University of New Brunswick.

2:45 p.m. (a) "Myoelectric Control", A. Stein, D. Lewis, F. Engler, & L. Thompson, Litton Systems, Rexdale.

(b) "The Human Operator in a Myo-Electric Control System", R.N. Scott, Univ of New Brunswick.

(c) "A Method of Providing Kinesthetic Feedback from Prostheses", D.S. Alles, M.I.T.

4:00 p.m. Coffee Break

4:15 p.m. (d) "Communication Systems for the Physically Handicapped", B.A.Holmlund, R.N. Kavanagh & A.E. Krause, Univ. of Saskatchewan, Saskatoon.

(e) "Programmed Speech Output System for the Handicapped", G.J. Huff, B.A.Holmlund, & A.E. Krause, Univ. of Saskatchewan, Saskatoon.

SESSION IIB

Thursday 8th Sept., 2:45 p.m.

Conference Room

"Mathematical Techniques"
and
"Physiological Control Systems"

Chairman: Dr. J.A. Tanner, N.R.C. Ottawa.

"Mathematical Techniques"

2:45 p.m. (a) "Computer Simulation of the Variability of the Normal SA Node Pacemaker Firing Frequency", H. Wolf, Dalhousie University.

(b) "A Mathematical Description of the EEG", W. Dietiker, N.R.C., Ottawa.

(c) "The Averaging Technique and Signals of Biological Origin", V.C. Abrahams, Queen's University.

4:00 p.m. Coffee Break

"Physiological Control Systems"

4:15 p.m. (d) "Paths Traced by the Visual Axis during Saccadic Eye-Movements", E. Llewellyn-Thomas & H. O'Beirne, University of Toronto.

(e) " A Technique for the Measurement of Pupillary Response in Conjunction with other Psychophysiological "Parameters", J.C. Lawless & F.R. Wake, National Defence Medical Centre, and Carleton Univ., Ottawa.

WORKSHOPS

Thursday 8th Sept. 8:00 p.m. - 10:00 p.m.

(i) Prosthetics - State of the Art To be held at the Beacon Arms Hotel.

(ii) Models of Biological Systems To be held at the National Research Council, where some special working demonstrations of Biological Models have been arranged by Dr. J. A. Tanner.

SESSION III

Friday 9th Sept., 9:00 a.m.

Auditorium

Chairman: Professor F. L. Thurstone, Bowman Grey School of Medicine, Visiting Professor, Queen's University.

9:00 a.m. Invited Tutorial Paper: Professor N.F. Moody, Dr. W. Paul & Mr. M. Joy. Institute of Bio-Medical Electronics, University of Toronto.

"Gamma-Ray Camera Techniques in Medicine".

SESSION IIIA

Friday 9th Sept., 9:45 a.m.

Auditorium

"Monitoring and Diagnostic Systems"

Chairman: Professor F. L. Thurstone, Bowman Grey School of Medicine.

9:45 a.m. (a) "An Improved Ultrasonic Brain Scanner", D. M. Makow & D. L. McRae, N.C.R., Ottawa, & Montreal Neurological Institute.

(b) "Biological Telemetry", O.Z. Roy & J.S. Hart, N.R.C. Ottawa.

10:35 a.m. Coffee Break

10:50 a.m. (c) "A Method of Monitoring Fetal Heart Sounds", F. Konopasek, Hargrave Research Corp., Winnipeg.

(d) "The Intensive Care Monitoring System of the Montreal Heart Institute", G. Warner, Montreal Heart Institute.

SESSION IIIB

Friday 9th Sept., 9:45 a.m.

Conference Room

"Electrocardiology"

Chairman: Dr. F. Berkman, University of Ottawa.

9:45 a.m. Invited Speaker: J.A. Hopps, Medical Electronics Group, N.R.C., Ottawa.

"Aspects of Electro-Cardiology"

10:10 a.m. (a) "Estimation of Cancellation Effects During Ventricular Depolarization", F. Berkman & O.Z. Roy, University of Ottawa and N.R.C., Ottawa.

10:35 a.m. Coffee Break

10:50 a.m. (b) "Feasibility of Analog Computer Techniques for Diagnostic Analysis in Electrocardiography"
R. Kieser, A. Wong, & P.M. Rautaharju, Dalhousie University.

(c) "Quality Control in Recording of Exercise Electrocardiograms", D.A. Winter, Dalhousie University.

11:45 a.m. Closing Address, Auditorium

12:00 p.m. Transportation leaves NRC for Conference Luncheon

1:00 p.m. Conference Luncheon - Banquet Room 'B', Beacon Arms Hotel

2:00 p.m. Annual Meeting, CMBES - Banquet Room 'B', Beacon Arms Hotel

2:30 p.m. Transportation leaves hotel for tour of NDMC

4:30 p.m. Transportation leaves NDMC for cocktail party

4:30 - 6:00 p.m. Cocktail party, 346 Mountbatten Avenue

We wish to apologize for the fact that the Table of Contents and page numbers have been omitted. However, the papers are bound in the order in which they are presented in the Conference schedule.

B.A. Holmlund

ON BIOLOGICAL CONTROL SYSTEMS MODELLING

J.H. MILSUM, DIRECTOR, BIOMEDICAL ENGINEERING UNIT
MCGILL UNIVERSITY

Introduction: Modelling is the fundamental concern of the scientific method, whether quantitative or not. Mathematical modelling is rapidly increasing in the biological systems area, both because of the disciplines needs and because of increasing interdisciplinary co-operation. Such models range from the physical size of the biochemical activities in a cell to the population dynamics of a large ecology. Each may involve the order of a hundred difference or differential equations, be highly nonlinear and have distributed-parameter effects, and yet still involve gross simplifications. However, the art of modelling necessarily demands such simplifications, and indeed simple models may be more revealing conceptually in many system dynamic's studies, even if they do not correctly incorporate the detailed physical phenomena.

Such biological models cannot always be easily made to fit the conventional framework of engineering control systems, because the feedback paths, comparators, controllers and controlled processes are not directly and uniquely identifiable. Nevertheless the multiple interactions involved necessarily produce various closed-loops, and generally overall system stability also. Such models may usefully be defined as incorporating "passive" control, in contrast to the "active" control of the many homeostatic and neuromuscular systems which recognizeably incorporate feedback transducers and therefore purposeful closed-loops.

Functional Nature of Problems: The general dynamic system model can be shown canonically as in Fig. 1(a); that is, a set of inputs x(t) act on the system H to produce a set of outputs y(t). The input set comprises both those inputs which can be manipulated by the experimenter, and those unpredictable and unmeasurable disturbance variables which cannot. The output set comprises those system variables of concern to the modeller, and are usually measurable or at least physically meaningful. Incidentally the powerful generalized method of system representation by state variables involves a further step in that the state of the system in terms of its state variables is determined by its initial conditions and the subsequent history of inputs, while in turn the outputs are specified as some function of these state variables, Fig. 1(b) [2]. This method permits equally efficient representation of continuous and discrete-time, and linear and non-linear systems.

Note that whether or not H represents a complete closed-loop system, eg. mammalian thermoregulation, or only a subsystem, eg. sweating, depends upon the experimenter's current interest. If it is a subsystem then generally some of the x(t) will be determined in turn by the output variables y(t) through the system's control action. This point is particularly relevant in biological experimentation where the closed-loop action cannot normally be broken temporarily and reversibly, unlike the easier situation in engineering systems.

Within this canonical framework three functionally different problems arise dependent upon which of x, y and H is the variable set to be solved for, and these are now briefly considered.

(a) Forward Analysis: In this standard problem x and H are given, and the output y is to be computed. In the modelling process such output predictions are usually made for various H-models to evaluate the best version. Of course, a "best" model can only be trusted with confidence in the dynamic domain for which it was found to be best, but nevertheless a good one can be valuable for predicting the outcome of physiologically dangerous or impermissible experiments. In this sense a fast, cheap, although potentially inaccurate probe is available to aid actual experimentation. Furthermore, some system variables of interest, which cannot be measured experimentally, can however be monitored continuously.

(b) Identification or Dynamic Discovery: This is the basic problem of modelling, namely that of determining H given x and y. However, determining H involves specifying both a structure and its parameter values, and a unique solution method is not available. In practice the analyst specifies the structure of H either from a priori physical knowledge or by arbitrary choice, and various steep-descent schemes are then implemented to obtain convergence on to some best set of the adjustable parameter values in the least-squares sense, as noted in (a). An important biological example is the on-line determination of human operator characteristics in a visual-manual tracking task involving a random signal as input. The general state-of-the-art restricts the number of parameters to be determined in a given structure. It should be noted that very crude linear models can be surprisingly good in representing complicated non-linear systems of the continuous up-dating can be fast, and that computationallythis is very economical.

(c) Input Discovery: Every instrument must be calibrated so that its actual input may be inferred from its output reading, the catheter correction problem representing a typical biological example. More generally the input discovery problem is not at all trivial; for example, in gastro-intestinal absorption studies by tracer techniques the flow rate across the gut into the venous system cannot be directly inferred from the plasma concentration because of the material's own disappearance dynamics through excretion etc. Fig. 2. However, these dynamics, the so-called "plasma curve" can be measured after impulsive intravenous injection of the same material, and constitutes the H of the larger problem. Since the measured plasma concentration after oral ingestion is the y, then the gut absorption rate is found by solving an input-discovery problem. In the example quoted [1], this can be done directly by inverting the problem to treat y as the input and (1/H) as the system. In general non-linear systems, however, this inversion task may require an iterative computing method.

Some Special Biological Aspects. In the emerging picture of biological control, certain phenomena seem noteworthy with regard to the modelling task:
(a) Energy acquisition is expensive for living organisms and in consequence adaptive optimization, on an energy basis, of such homeostatic subsystems as respiration seems to have conferred selective advantage.
(b) Neuromuscular skill seems to involve learning a preprogramming of the muscular innervation, so that the action can by-pass the richly-connected, but slow feedback structure, and hence can operate faster.
Certainly in the absence of significant system parameter changes and of disturbance inputs the open loop configuration can always be faster and more economical.
(c) Receptors enhance rates-of-change in feedback signals which they originate. They thus provide the derivative control action often necessary for closed-loop stability, and also anticipatory feed-forward to nullify undesirable disturbances. This rate sensitivity is often inherently unidirectional, which also confers certain dynamic advantages, although at first it appears to be a disadvantageous effect due to limitations in the physical processes involved, for example, that negative intervals between neuronal spikes cannot exist.
(d) Statistical averaging of neurophysiological data from single cells is necessary for meaningful analysis. Presumably the living organism instead obtains its own rapid output actions in response to stimuli by ensemble averaging of many cells, and hence also implements a mechanism which can still function even after individual cells may randomly have died.

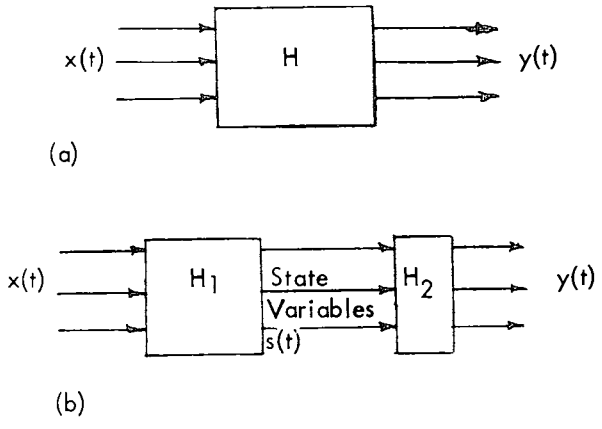

Fig. 1 The Basic Dynamic System

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(2) R.C. Dorf, Time-Domain Analysis and Design of Control Systems
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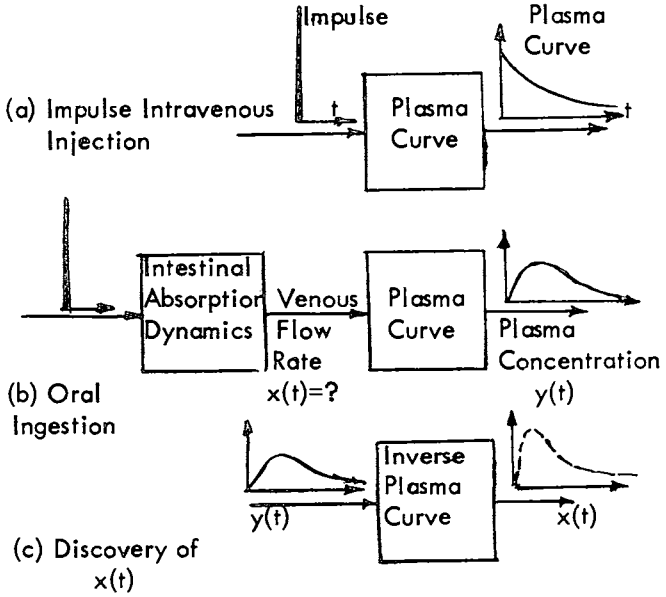


Fig. 2 The Input-Discovery Problem--For Gastrointestinal Absorption.

PHYSIOLOGICAL EXPERIMENTATION AND MODELLING OF THE AUDITORY SYSTEM

C. A. Laszlo*and J. H. Milsum**
* Otolaryngological Research Laboratories, McGill University, Montreal
** Director, Bio-Medical Engineering Unit, McGill University, Montreal

In auditory physiology only a few system parameters can be measured in a given type of experiment. Nevertheless, it is very important to keep track of the behaviour of the non-observed parameters as well. Toward this end the simultaneous monitoring of an appropriate model of the system in parallel with experimental observations of the measurable parameters will facilitate such correlation of observable and non-observable parameters. This then raises the important problem of constructing a suitable model of the auditory system.

Phenomena associated with the sensation of hearing (loudness, pitch, masking, etc.) seldom can be utilized in constructing physiologically compatible models because the underlying mechanical, neural and biochemical factors are not well understood. Physiologically observable objective performance parameters of the auditory system, at least on experimental animals, are:

(a) the motion of the ossicles and the mechanical transmission characteristics of the middle ear;

(b) the acoustic input impedance of the ear as measured at the eardrum;

(c) the motion of the basilar membrane;

(d) the electrophysiological response of the cochlea, in particular the cochlear microphonic potential;

(e) the neural pulse pattern of the acoustic nerve;

(f) the excitation pattern of the cortical and subcortical brain structures.

The description of sensory processes by a black box approach does require a special philosophy. Due to the complexity of the real system and the relative simplicity of the mathematical description, it is absolutely necessary to maintain a close connection between model and process. At all stages of study, the analysis of one must help in organizing the study of the other. Thus, a dual view must be maintained throughout: part physiological and part mathematical.

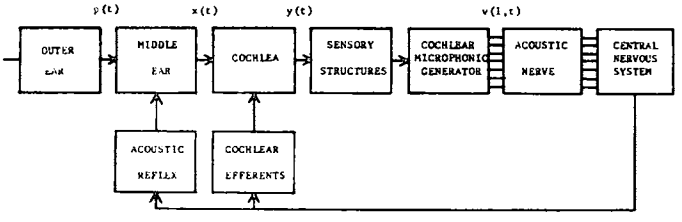
Since auditory physiologists have not so far used the classical technique of input-output analysis extensively their data for the description of the peripheral auditory system dynamics do not exist in a form suitable for this analytical technique. In any given experiment a choice must be made to select a suitable parameter for measurement. This choice is restricted by practical and theoretical considerations. The parameter in question must be easily measurable, yield reproducible measurements and exhibit few subject-to-subject variations.

Measurability is a very important factor in choosing an auditory system parameter for an experiment. The acoustic input impedance is measured with relative ease. Ossicular motion can be recorded with capacitative probes but the surgical procedures and the measuring apparatus excludes the measurement of other factors of interest. The motion of the basilar membrane, however, is very difficult to observe, since the surgical procedure is by necessity destructive, and hence, all known observations were done on cadavers. The neural pulse pattern cannot, as of yet, be observed in sufficient detail, due to the fact that present microelectrode observations are restricted to only a very small percentage of the total number of nerve fibres in any one preparation. Gross electrodes, on the other hand, record a spatio-temporal summation of the pattern, hence details are not distinguishable. On the cortical level similar measurement problems exist compounded by our sketchy knowledge of the anatomy of the nervous pathways.

Before we attempt to model the cochlear microphonic response of the auditory system, it is necessary to find models of the middle ear and the cochlea suitable for our purposes. Also, it is necessary to evaluate published data on the middle ear, the dynamics of the cochlear partition and especially the cochlear microphonics. We have integrated and reinterpreted the results of Bekesy, Zwislocki and Flanagan, to produce a preliminary model representing the guinea-pig ear. The middle ear is represented by a second order system and the cochlea model is found to be of the sixth order.

We also report on experimental work designed to fill existing gaps in available physiological data and measurements. This phase of the work concentrates on the measurement of the cochlear microphonic potential.

Finally, we discuss a practical plan for "interfacing" of model and experiment.



p(t) = Sound Pressure Level at the eardrum
x(t) = volume displacement of the stapes
y(l,t) = displacement of the cochlear partition 'l' distance away from the stapes
v(l,t) = cochlear microphonic potential measured 'l' distance from the stapes

Fig.2.1. Information flow diagram of the auditory system.

A MODEL OF THE CARDIAC ELECTROPHYSIOLOGICAL SYSTEM

F.A. ROBERGE, R.A. NADEAU

Department of Physiology, University of Montreal, Montreal

Under normal physiological conditions, the cardiac impulse originates in the region of the sinus node and propagates successively through the auricles, A-V node, and ventricles. The A-V node serves usually as a transmission pathway between the two chambers of the heart; yet, under certain conditions, it also acts as a pacemaker. In recent experiments by the authors (1, 2), evidence was provided to suggest that the A-V node would be an automatic center normally dominated by the faster sinus node and synchronized with it. Further studies (3) have indicated that feedback action from A-V node and ventricles to sinus node may also be important. These results imply that the two heart nodes would be reciprocally coupled automatic centers, and form an arrangement which should behave in many respects like a system of interacting oscillators.

The proposed model of the cardiac electrophysiological system consists of two electronic relaxation oscillators (SN and AVN) arranged in a closed-loop fashion, Fig.1. These oscillators have properties of threshold, graded excitability, absolute and relative refractoriness, and paradoxical excitation (4); and thus are adequate models of physiological pacemakers such as the heart nodes. The forward and feedback links between the two oscillators include pure time delays which simulate the transmission delays encountered between the two heart nodes. Although feedback action upon the sinus node may be other than electrical, there is little question that this retrograde action occurs after an appreciable time delay also.

Depending on the relative frequencies of the oscillators and the amount of coupling, various modes can be obtained in which the two units are either synchronized or desynchronized. Each mode may be shown to correspond to a specific normal or abnormal electrophysiological cardiac condition.

Sinus rhythm with constant PR interval. The rates of SN and AVN are constant, and SN is faster than AVN. Both oscillators are synchronized and the phase between the two outputs (which represents the PR interval) remains constant.

Passage from sinus rhythm to A-V nodal rhythm. Initially SN is faster than AVN and 1:1 synchronization exists. When the rate of SN is gradually reduced until it is below that of AVN the phase decreases progressively, passes through zero, and AVN becomes the leading oscillator. Thus the two oscillators are still synchronized, the phase has a stable negative value, and a condition equivalent to A-V nodal rhythm has been realized. A similar result may be obtained by gradually increasing the rate of AVN above that of SN.

Atrial acceleration during ventricular arrest. By properly adjusting the amount of coupling between the oscillators, the system may be made to adopt a synchronized rhythm which is intermediate between the free running rates of SN and AVN. Ventricular arrest is simulated by rapidly decreasing the frequency of AVN until it is completely silent. Thus SN is liberated and accelerates to its free running value. The magnitude of this escape depends, of course, on the difference between the synchronized rate and the free running rate of SN.

Atrial slowing upon return of ventricular activity. The return of ventricular activity is simulated by reactivating AVN to its initial free running level. As AVN begins to discharge, the rate of SN is seen to decrease. When AVN has recovered completely, the two oscillators resynchronize and adopt the initial synchronized rhythm which is slower than the free running rate of SN.

Wenckeback-Luciani cycles. Initially SN is faster than AVN and both oscillators are synchronized in a 1:1 ratio. If the rate of AVN is reduced gradually and slowly until desynchronization occurs, then Wenckeback phenomena are produced. The Wenckeback periods become shorter and shorter as the rate of AVN is decreased further. A similar result may be obtained by increasing the rate of SN instead of decreasing the rate of AVN.

Conduction blocks. Starting from 1:1 synchronization frequency ratios of 2:1, 3:1, etc, are obtained either by increasing the frequency of SN or decreasing that of AVN.

The above examples are only a few of the several phenomena which can be reproduced by a model in which both sinus and A-V nodes are represented by astable relaxation oscillators. Phenomena of frequency entrainment, concealed conduction, and the effects of coupled and paired pulse stimulation could be demonstrated also.

REFERENCES

(1) NADEAU, R.A. and T.N. JAMES, The behavior of A-V nodal rhythms following direct perfusion of the sinus node. Can. J. Physiol. Pharmacol., 44, 317-324, 1966.

(2) ROBERGE, F.A. and R.A. NADEAU, Simulation of sinus node activity by an electronic relaxation oscillator. Can. J. Physiol. Pharmacol., 44, 301-315, 1966.

(3) NADEAU, R.A., A.K. AMIR-JAHED and F.A. ROBERGE, Further studies on synchronization between the sinus and A-V nodes. Proc. Can. Fed. Biol. Soc., 9, 84-85, 1966.

(4) ROBERGE, F.A. and R.A. NADEAU, Analogies between relaxation oscillators and biological pacemakers. Proc. Can. Electron. Conf. IEEE Cat. No. F17, 64-65, 1965.

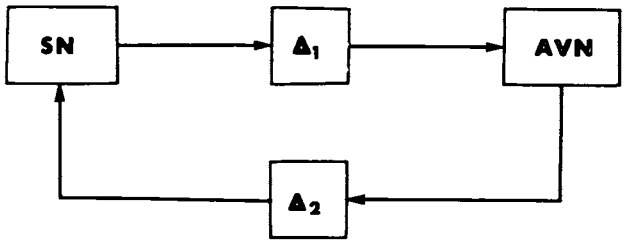


Figure 1

DIGITAL COMPUTER SIMULATION OF FROG SCIATIC NERVE

R.W.J. FORD and J.A. TANNER
National Research Council, Ottawa

Experimental studies of the responses of bullfrog (Rana Catesbeiana) sciatic nerve to high frequency electrical stimulation, of the order of kilocycles per second, revealed that the responses are essentially random, i.e. there is no correlation between the stimulus and the evoked response. Since the nerve is composed of a large number of fibers, each with its own conduction velocity, it was considered that the randomness of the nerve response might be more apparent than real. In other words, given a long enough recording interval, the response might become periodic. To study the structure of nerve responses produced by high frequency stimulation, and to ascertain the effect of randomness in the fiber thresholds, a mathematical model simulating the sciatic nerve was formulated in the form of a digital computer program.

The simulation of a nerve requires a definition of the nerve itself, a description of the nerve conduction process, and a definition of the experimental configuration.

In defining the nerve, certain assumptions were made regarding the number and size distribution of fibers in the nerve; their respective conduction velocities, thresholds to stimulation, time durations of absolute and relative refractory periods; and the size and shape of their action potentials.

Description of the nerve conduction process requires a definition of the stimulus used and the corresponding membrane response characteristics, a method of calculating the time at which an action potential is generated, and the time at which it is detected by a recording electrode. A method of arranging propagated action potentials in time is required so that they may be summed to give the total nerve response at the recording electrode as a function of time.

The experimental configuration consists of an isolated nerve having a pair of stimulus electrodes attached at one end, and a pair of recording electrodes at the other end. The innermost stimulating electrode is considered to be the cathode and the innermost recording electrode is termed "the recording electrode", the outer recording electrode being at a constant potential to insure monophasic recording.

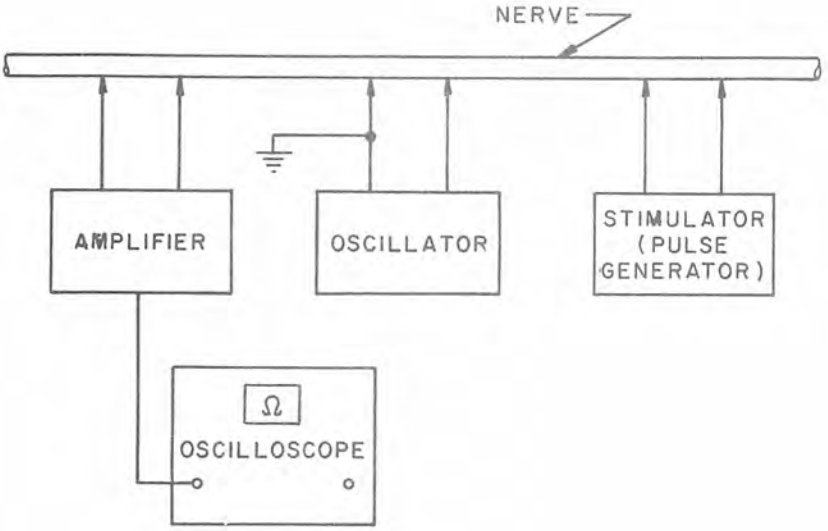
The model is designed to consist of 538 fibers with diameters ranging from 5 to 20 microns. Conduction velocities are assumed to be directly proportional to and the thresholds inversely proportional to fiber diameter. The thresholds are allowed to vary randomly about their mean values. Time durations of the absolute and relative refractory periods are assumed equal for all fibers. During the absolute refractory period fiber thresholds are considered infinite, and during the relative refractory period they are assumed to be elevated above their resting state values. Action potentials are described as triangular in shape with amplitude proportional to fiber diameter. This amplitude is reduced if the action potential is generated during a relative refractory period.

The stimulus in this model is a train of rectangular pulses with variable repetition rate, pulse amplitude and pulse duration. Arrangement is also made for the simultaneous application of two different stimuli.

Membrane response to a given stimulus is assumed to be that of a damped second order linear system. For a single stimulus pulse, the computed membrane response is a bell shaped curve which resembles an experimental recording of the change in membrane potential following a subthreshold stimulus.

If the membrane response of a fiber exceeds its threshold at time, t_1, an action potential is generated under the stimulating cathode and propagates along the nerve to the recording electrode, arriving at $t_2=t_1 + L/V$, where L is the nerve length between the stimulating cathode and recording electrode, and V is the conduction velocity. Total nerve response is calculated by summing the action potentials at the recording electrode as a function of t_2.

All of the nerve and stimulus parameters can be varied but to the present time, only variations of the stimulus parameters and of the range of random threshold variation have been studied. Typical computed nerve responses shown in Fig (2) can be compared with their living counterparts in Fig (1). The responses indicate that the model is a good first approximation to isolated frog sciatic nerve. That there are still large gaps in the knowledge of membrane characteristics is evident from the number of assumptions that were required in the formulation of the present model. Models such as this assist the experimenter in asking the appropriate questions of the living system.



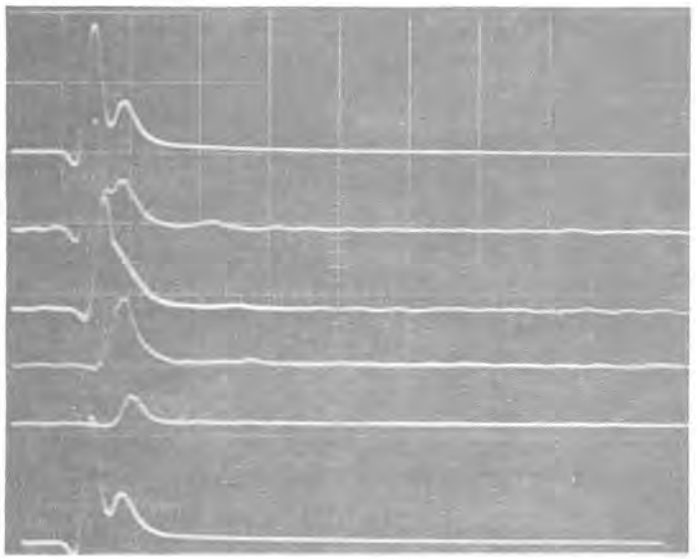


FIG. 1 AMPLITUDES OF THE A.C. ARE (FROM TOP TO BOTTOM): 0, 1.25, 2.73, 3.60, 4.30, AND 0 VOLTS. THE APPLIED A.C. DOES NOT APPEAR ON THESE RECORDS BECAUSE AN EQUAL AMPLITUDE, ANTI-PHASE 5-KC. SIGNAL WAS ADDED TO THE NERVE SIGNAL AT THE OSCILLOSCOPE INPUT. TIME SCALE 2 MSEC./DIV.

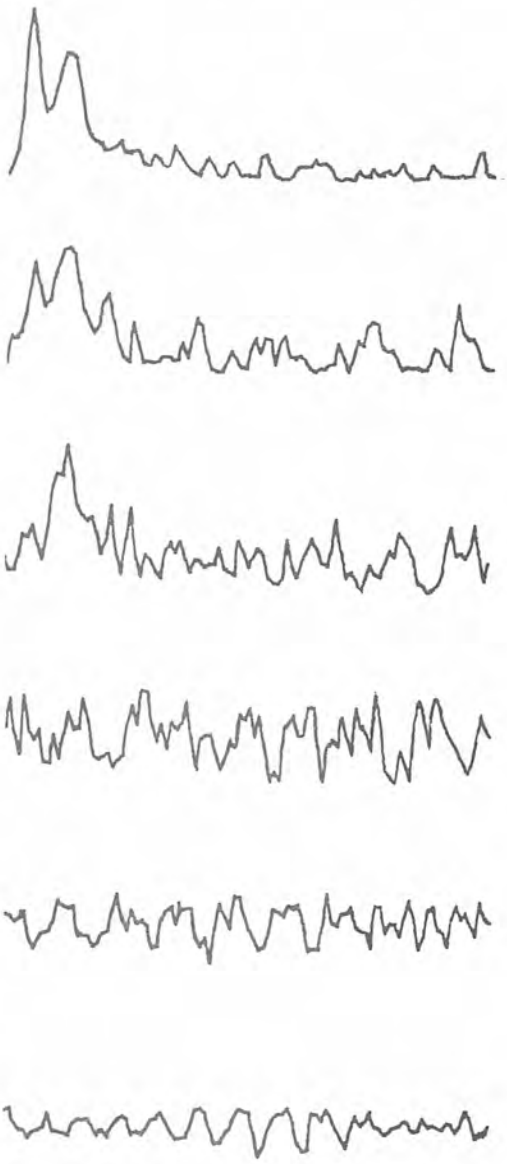


FIG. 2 NERVE MODEL RESPONSES CORRESPONDING TO THOSE OF FIG. 1. AMPLITUDES OF HIGH FREQUENCY STIMULUS (FROM TOP TO BOTTOM) IN ARBITRARY UNITS 2.00, 2.13, 2.88, 3.89, AND 5.89.

MEASUREMENT AND SIMULATION OF HANDWRITING SIGNALS

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University of British Columbia

INTRODUCTION

Lashley[1], in his paper on "The Problem of Serial Order in Behavior" in 1951, proposed that the execution of quick, accurate movements was governed by an effector mechanism which could be pre-set or primed to discharge in a given pattern. Denier van der Gon and his associates[2] later pointed out that these principles likely apply to handwriting movements. They further proposed that the control of handwriting movements is descrete, that is, the force in a muscle rises to a certain level and remains more or less constant for a length of time before attaining a new level; that is to say, the force waveform in handwriting motions is trapezoidal. The research reported in this paper seeks to test these hypotheses by direct measurement of the acceleration of handwriting movements, and the simulation of them by means of a simple electronic simulator.

MEASUREMENT OF HANDWRITING SIGNALS

Simple Motions

The system used to measure the acceleration of handwriting movements consisted of a displacement transducer in the form of a shallow electrolytic tank with appropriate amplifiers, followed by two differentiation and low-pass filtering operations. This system was first tested on simple to-and-fro motions of the hand. A typical result appears in Fig. 1. This figure shows the displacement, velocity, and acceleration of simple left-right movement about the wrist joint. The acceleration waveform is remarkably trapezoidal except for the sharp spikes which appear in coincidence with zero-crossings of velocity (i.e. extrema of displacement). These spikes are due to static friction, since they disappear when the writing stylus does not touch the bottom of the tank (Fig. 1b). Although friction is the biggest contributor to features which mask those aspects of the acceleration waveform directly attributable to the muscle forces, several others have been identified.

Handwriting

Figure 2 shows typical waveforms obtained from actual handwriting. Figure 2a shows the vertical displacement, velocity, and accleration together with horizontal acceleration, while Figure 2b shows horizontal displacement, velocity, and acceleration together with vertical acceleration. The two projections of the acceleration waveform appear twice. One pair of acceleration signals has trapezoidal approximations fitted to the waveforms (Figure 2a), while the other pair is left untouched so that one can see the detail of the waveform.

The dominant features of the acceleration waveforms are the abrupt changes which indicate significant alterations in the applied force. Using these dominant features as a guide, and neglecting the fine-structure and the effects of static friction, one can make good trapezoidal approximations to the acceleration waveforms. How much of the fine-structure, if any, is attributable to the function and control of the biological system, and how much is due to extraneous mechanical phenomena is a question that cannot be answered here. In making the trapezoidal approximation, we are assuming that the fine-structure is not important in handwriting, and all that really matters are the large abrupt changes in the acceleration. The simulations which have been carried out support this assumption. The static-friction spikes can hardly be called fine-structure, but they are easily identified (see below the arrow on the vertical acceleration of Figure 2), and are definitely due to a purely mechanical phenomenon.

Further evidence of the discrete nature of the control of handwriting movements is found in Figure 3. The record shown in this figure displays the acceleration occuring during simple to-and-fro motion of the fingers holding a pencil (middle trace), together with electromyographic records taken from surface electrodes. The sharp excursions in what would otherwise be flat tops and bottoms of the acceleration waveform are due to static friction. The electrodes responsible for the top trace (1) were placed on the palm of the hand, directly over the abductor pollicis brevis at the base of the thumb. The EMG in the bottom trace (3) was taken from the dorsal part of the forearm, approximately 4 inches below the elbow. The major superficial muscles contributing to this signal would be the extensor carpi ulnaris and the extensor digitorum.

Several of the features of the signals of Figure 3 are worthy of note: (i) The EMG signals are remarkably segmented, which gives strong support to the assumption that the controlling force waveform is discrete in nature. (ii) The onset of activity in the EMG signals coincides very well with changes in slope of the acceleration waveform. Several

of these are marked in the figure. The cessation of EMG activity also coincides with reproducible, recognizable features of the acceleration wave form. Since the two EMG records shown present an incomplete picture of muscle activity, there are many features of the acceleration waveform that are not accounted for. Unfortunately, equipment was not available to record more than two EMG's simultaneously.

SIMULATION OF HANDWRITING

The handwriting samples and associated waveforms presented above indicate that one should be able to simulate handwriting quite closely by using a simulation of a mass with viscous friction driven by a force waveform that is a trapezoidal time function. This is a modification of the scheme that was used by Denier van der Gon et al.[2] The two schemes differ in that the present simulation makes no assumptions about independent directions of motion or the number of discrete levels attainable by the trapezoidal function.

Simulation of the handwriting sample of Figure 2 is shown in Figure 4. In Fig. 4a, the top sample is the simulation, and the bottom one is the original. The waveforms in Fig. 4b and 4c are outputs from the simulator corresponding to those of Fig. 2a and 2b respectively. This simulation is typical of many that have been done, and shows the excellent results that can be obtained. All of the simulations were set up by a process of trial and error, adjusting the amplitude and timing of the trapezoidal "force" function to make the "acceleration" correspond as closely as possible to the measured waveform, and then trimming the settings to cause the "displacement" to match. One of the chief sources of error in the simulations is the fact that in the version of the simulator used here, all of the leading slopes of the trapezoidal waves are constrained to be the same. The same is true for all of the trailing slopes. Inspection of Fig. 2 reveals this to be clearly not the case in real handwriting.

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ACKNOWLEDGEMENT

The research reported here was made possible by support extended to the Massachusetts Institute of Technology, Research Laboratory of Electronics by the Joint Services Electronics Programs, National Science Foundation, National Institutes of Health, and the National Aeronautics and Space Administration.

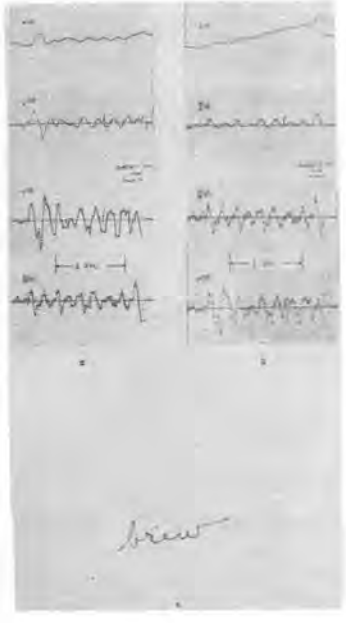

Fig. 2

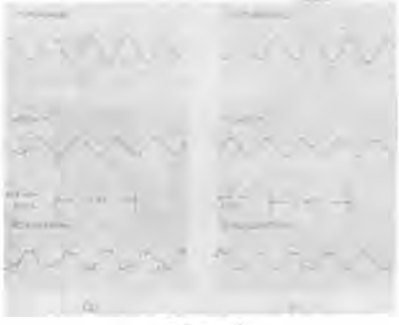

Fig. 1

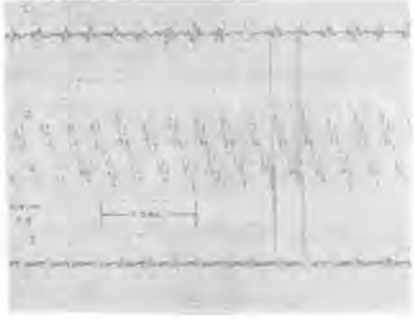

Fig. 3

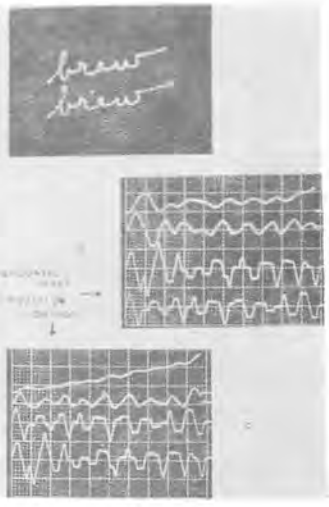

Fig. 4

REGIONAL HYPOTHERMIA BY AN EXTRACORPOREAL ARTERIAL SHUNT

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E. W. PETERSON, M.D., University of Ottawa, Ottawa, Ont.

Hypothermia may be induced by several methods for clinical application. Surface cooling either by immersion in an ice bath or by coolant blankets results in maximum cooling of the skin, the subcutaneous fat and muscles which require the protection of hypothermia much less than the vital organs. Core cooling, or blood stream cooling, has been an accepted practice since the early use of heart-lung bypass equipment. Blood cooling has been used with both open and closed chest techniques, with and without oxygenators. Until recently blood stream cooling has always required one or possibly two blood pumps in the extracorporeal circuit to overcome the resistance of the circuit if not to bypass the heart.

Selective hypothermia by blood stream cooling has been conducted for various organs including the brain, kidneys, or a limb. The systems include arterial shunts with a heat exchanger, a pump and the usual auxiliary apparatus, i.e. a filter and bubble trap, etc. In some cases isolated selective hypothermia has been conducted with a venous-arterial shunt by including a small oxygenator in the circuit.

The hemolysis of blood due to extracorporeal pumps is a well-known phenomenon which, while it is an undesirable effect, is an acceptable necessity for heart bypass. Where selective cooling of an organ other than the heart is desired, then the risk of hemolysis should be reduced.

In early 1963 the possibility of cooling the brain in a human, selectively without an extracorporeal pump, was first considered. Preliminary calculations showed that a high efficiency low resistance heat exchanger for blood stream cooling should be a technical feasibility.

The heat exchanger design was based on a single pass temperature reduction from 37°C to 5°C at the rated arterial flow of 250 ml. per minute while a pressure drop of 5 mm. Hg. across the extracorporeal circuit was allowed.

The prototype heat exchanger met the thermal and flow resistance requirements as shown by laboratory calibrations and experimental applications.

A number of improvements have been made in the mechanical design of the heat exchanger as a result of experimental use under operating room sterile procedures.

The heat exchanger together with the arterial cannulae and the connecting tubing provides a simple, safe procedure for selective brain cooling to profound temperatures, or it can be used for inducing a general moderate hypothermia. Reheat can be accomplished with the heat exchanger.

A CANADIAN APPROACH TO THE PROBLEM OF HAEMODIALYSIS MACHINERY.

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More than four hundred Canadians under the age of forty die annually of uremia. Of these, many could be maintained and live a useful existence on a program of haemodialysis.

Several machines are available to carry out haemodialysis, all of which use the same basic principle, and each of which seems to have its own peculiar advantages and attendant disadvantages. The two most prevalent types of machinery are the Kolff system and the Kiil system. In each method, the costs run presently between $7000 and $10,000 per patient per year.

Other systems such as peritoneal dialysis, dialysis of lymph fluid, blood filters, and resin exchange systems, have all undergone extensive investigation without evidence that these offer any significant advantage over haemodialysis techniques.

With the increasing load of uremic patients, a mechanical system seems to be the most effective to use, in terms of efficiency and safety of operation.

We propose to review the problem, the major deficiencies of the above mentioned two types of machinery and to combine their outstanding features into an optimal artificial kidney. Our approach will be from the engineering point of view, and our solution is substantiated from records kept during the operation of one such machine at the Kingston General Hospital, in Kingston, Ontario, under Doctors C.F.D. Ackman, P.A.F. Morrin, P. Honda, and A.W. Bruce, of the Departments of Medicine and Urology of Queen's University.

The Problem:

In simple terms, the kidney is a washing machine for blood. The problem is to replace this function mechanically with maximum safety, efficiency, patient comfort, and minimum cost and labour effort.

The major deficiencies of the Kolff system are:

(1) The cost of the disposable parts (approximately $60.00 per dialysis).
(2) Inadequate safety devices requiring close supervision.
(3) Cumbersome equipment.

The major deficiencies of the Kiil system are:

(1) High labour and assembly of membrane.
(2) Machine is not efficient in handling of acute renal failure.
(3) Dialysis time of twelve to sixteen hour per patient is too high.

The Optimal Machine features:

(1) The use of disposable coils
(2) Significant reduction of the bath volume required.
(3) Simplicity of machinery in operation and maintenance.
(4) Reliable fail safe devices.
(5) Simple hot water cleaning and storage.
(6) Competitive capital and operating costs.
(7) Availability of efficient ultrafiltration.
(8) Permits a favourable staff to patient ratio within a complete eight hour shift cycle.

The Kingmed Artificial Kidney machine approaches the optimal machine as follows:

(1) Disposable coils. Although currently using the regular Baxter disposable twin coil at $60.00 per unit, the system is designed to function with a disposable coil and tubing which will cost $20.00 or less. Work is underway, on the development of a coil to meet these stated requirments and it is expected that similar coils will be available on the American market this fall.

(2) Bath. The dialysis coil is mounted in a ten litre heated tank and bath is pumped through the coil at a rate of 20 litres per minute. Cold fresh bath fluid is added at a variable rate and the excess allowed to flow freely into the drain. In this manner, we have -

(a) reduced the total volume of bath required for a complete haemodialysis to about 1/3rd of the previous requirements.
(b) Accomplished adequate removal of waste products, such as urea and creatinine.
(c) Controlled the bacterial count since the bath is stored at room temperature and bacterial counts for six hours are usually well below 10,000 per millitre, as compared to warm bath counts of one to two million.

(3) Simple Machinery. The system is designed to function as a series of small independent dialysis units connected to a central tank, which feeds the cold bath to the entire system. The use of a bath concentrate in preparation of the bath obviates the risk of mixing errors and a simple test will confirm the correctness of bath concentrate. Blood is pumped by a roller pump with a variable speed control. As the blood requirements per patient have been similar to, or less than that of the Kiil system, our concern about potential haemolysis has largely been unfounded. The roller blood pump offers an additional safety factor as it seals the arterial circuit whenever it is stopped, thus preventing a lethal haemorrhage. The bath circulating pump is a standard unit which can be replaced in minutes. The in-flow of cold bath is individually controlled at each dialysis station and all heating is done at each station. Piping and fittings are standard items which can be repaired on the scene by routine hospital maintenance staff.

(4) Reliable fail-safe devices. Our major concern has been the monitoring of the pressure within the coil and we have so arranged our high and low pressure fail systems so that should the

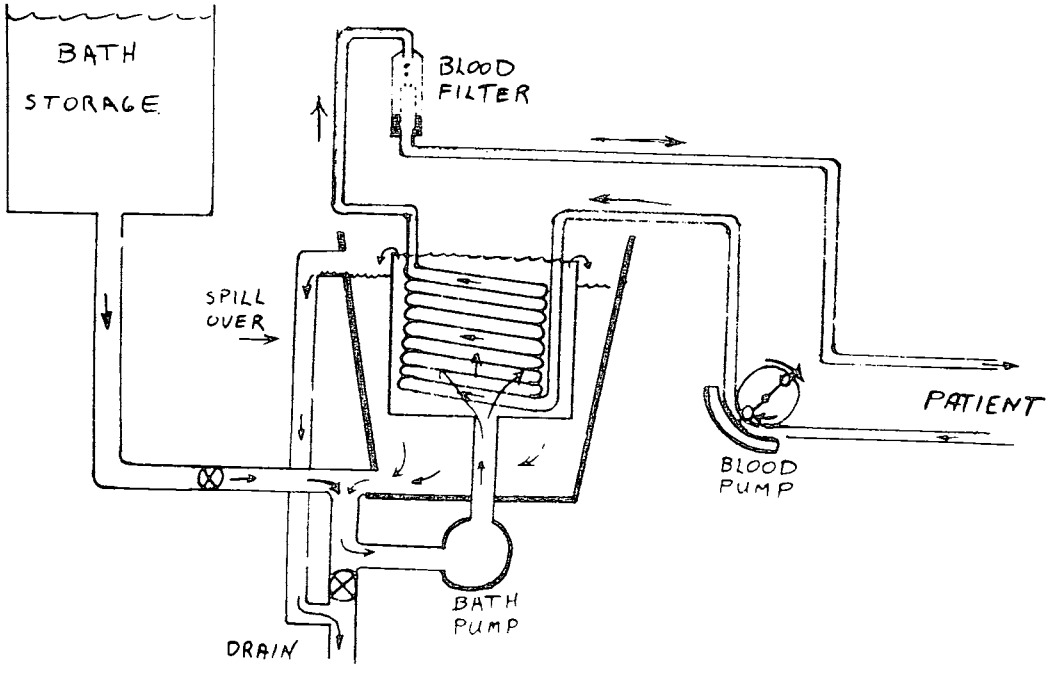
coil burst, the patient accidentally obstruct the flow, or even pull out his connection, the pump will stop within a few seconds and sound an alarm. A cardiac arrest or significant arrhythmia which would impair blood flow would also bring the machine to a halt and sound the alarm. These simple devices have lessoned the strain on the nursing staffs and has allowed them considerable freedom in operating the machine.

(5) Simple hot water cleaning. The machine is so designed that it can be washed down with 190°F water and will drain dry leaving the machinery virtually bacteria free.

(6) Economy. The capital cost of this equipment is competitive with existing machines. The potential saving of the system lies in the ability to run a large number of patients on the system at a very favourable staff to patient ratio.

(7) Availability of efficient ultrafiltration. Since a blood pump is incorporated, ultrafiltration of water from the blood can be achieved by increasing the pressure of the blood inside the coil, thus a weight reduction for a patient of twelve pounds per dialysis is practical.

(8) Permits a favourable staff to patient ratio within a complete eight hour shift cycle. The Kiil system presently requires between twelve to sixteen hours twice weekly to manage a majority of patients. Patients on the more efficient Kolff system can achieve comparable results in four to six hours twice weekly, however, these patients suffer a high incidence of nausea and vomiting associated with some more rapid biochemical shifts. It is felt that our system will provide a compromise between the two extremes in the dialysis of approximately six hours using a smaller bath volume which appears to be satisfactory without placing undue strain on the patient. Acceptance of our equipment has been further improved by having the machinery remote from the patient. Our present system is operated from start to finish in a single eight hour shift, which has proven most acceptable to both staff and patient.



SCHEMATIC REPRESENTATION OF THE KINGMED DIALYSIS SYSTEM.

A DEVICE FOR ASSESSING INTRA-CRANIAL PRESSURES

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University of Saskatchewan

The diagnosis and treatment of several disorders and diseases in infants would be considerably aided by a quantitative measure of the intra-cranial pressure. Simple hydrocephalus, for example, is accompanied by an increase in this pressure. Similarly, a knowledge of pre- and post-operative pressure variations in babies with myelomeningoceles, or in situations where ventriculo-caval shunt valves have been installed, would be most useful. The accepted method of monitoring pressure by head circumference measurements only reveals the long-term effects of pressure variations. Generally, the magnitude and variation of the intra-cranial pressure in normal and abnormal infants is not well known, largely because of the lack of a technique or device capable of determining this quantity. An instrument which will satisfy this need is described.

The design of the device is influenced by several factors peculiar to the situation. As a general criterion, the measurement must be performed simply and quickly without alarming or disturbing the infant since straining or crying will increase the intra-cranial pressure. A second difficulty is the short-term variability of the intra-cranial pressure which is apparent in the fontanelle pulsations of most infants. Figure 1 shows a strain gage recording of these fontanelle pulsations which were observed to correlate with respiration and pulse. Model studies indicate that the magnitude of the variations in a typical child may be as much as 60 mm of water which is significant since the normal range of average pressures lies between 70 and 210 mm of water. A third and most serious difficulty is the lack of direct contact with the cerebro-spinal fluid (CSF). Therefore, an indirect measure of the pressure is obtained through the soft membrane-like tissue of the anterior fontanelle on the basis of the applanation theory which is widely employed in tonometry.

The applanation theory and principle of operation of the transducer are illustrated in Figure 2. The lower half of the figure shows, in cross section, a curve representing the fontanelle membrane and the fluid-filled cylindrical transducer resting on the fontanelle. Clearly, there are three situations that can occur depending on the relative magnitude of the unknown intra-cranial pressure P_2 and the pressure applied inside the probe P_1. If P_1 is less than P_2 the fontanelle bulges up into the cylinder. If P_1 is greater than P_2 the fontanelle is depressed. Between these two extremes the unique situation occurs when P_1 is equal to P_2 and the membrane is flat. Under this condition, the intra-cranial pressure can be determined by simply measuring the identical applied pressure P_1. Briefly then, in a single determination of CSF pressure, the applied pressure P_1 is varied, the occurrence of membrane flatness is detected by the transducer, and the applied pressure is measured at that instant. P_1 is varied as illustrated in the upper half of Figure 2 which is accomplished by means of a solenoid-operated pump, a storage chamber, and an air escapement valve. The condition of flatness is detected by the electrical switching action of the conducting fluid in the transducer which just touches the center probe when applanation occurs. Besides providing a suitable switch, the fluid minimizes adverse effects of hair on the fontanelle and presents a flat surface for detection, and as well, it enables some amplification of membrane movement by a narrowing of the fluid channel as illustrated in Figure 3. Disadvantages of the fluid filled transducer are that it must always be used and handled in a near vertical position and that pressure readings can be taken only on the "make" portion of the cycle because of meniscus effects. Figure 4 shows the transducer in use.

The instrument is completely automated through the use of logic and control circuitry such that measurement is initiated by the placement of the transducer on the fontanelle, stops if the transducer is inadvertently removed, makes 20 separate determinations of the CSF pressure and provides a visual readout of the average of the readings, and then stops. A photograph of the complete instrument is shown in Figure 5. Rubber model studies indicate that the device is capable of determining pressures within $\pm$10 mm of water of the true average. While the accuracy of CSF pressure determinations is not expected to be this good because the fontanelle may not act as a true elastic membrane (which is a precept of the applanation theory), an estimated accuracy $\pm$20 mm of water is not unrealistic.

A normal range of pressure between 70 and 210 mm of water was observed in a study of 40 normals ranging in age from a few days to 10 months. (This compares favorably with the accepted range of 70 to 200 mm of water in the prone position as established by lumbar puncture.) It was also found that moderate long-term pressure variations may occur and that minor conditions, such as skin rash, can markedly affect the CSF pressure.

Some clinical results are shown in Figures 6, 7, and 8. Figure 6 shows the pressure variations encountered in the surgical repair of a myelomeningocele with post-operative development of hydrocephalus. Notice that the head circumference measurements at the beginning of the case study do not reflect the large pressure variations. Figure 7 illustrates the pressure changes accompanying removal of CSF by puncture of the anterior fontanelle. Figure 8 shows the normal

insitu pumping of a Holter valve on three separate occasions.

Our studies have shown that the instrument will measure intra-cranial pressures accurately, simply and quickly. Presently the device is being used as a clinical research instrument and it is hoped that it may be used as a routine screening device for the detection of abnormal pressure in newborns.

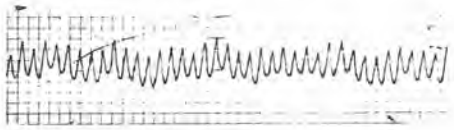

Fig. 1 Strain gage recording of fontanelle pulsations.

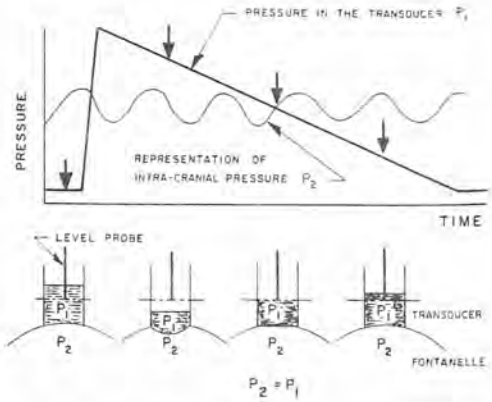

Fig. 2 Diagram illustrating the applanation theory and the principle of transducer operation.

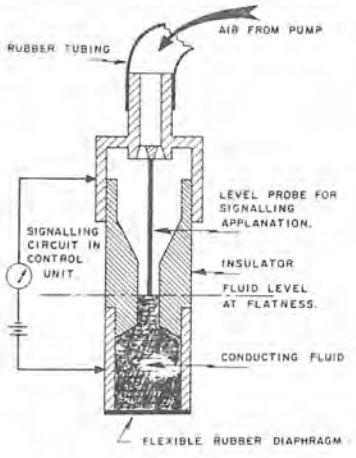

Fig. 3 Fluid-filled pressure transducer.

Fig. 4 Pressure transducer in use.

Fig. 5 Instrument for assessing intra-cranial pressure.

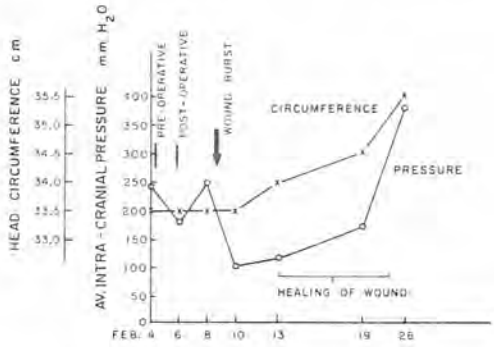

Fig. 6 CSF pressure variations following myelomeningocele repair with post-operative hydrocephalus.

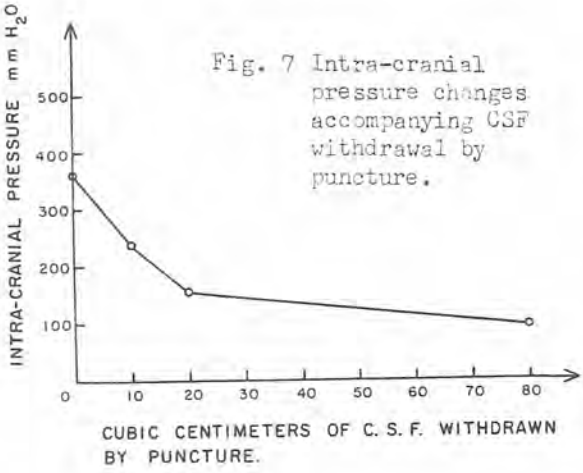

Fig. 7 Intra-cranial pressure changes accompanying CSF withdrawal by puncture.

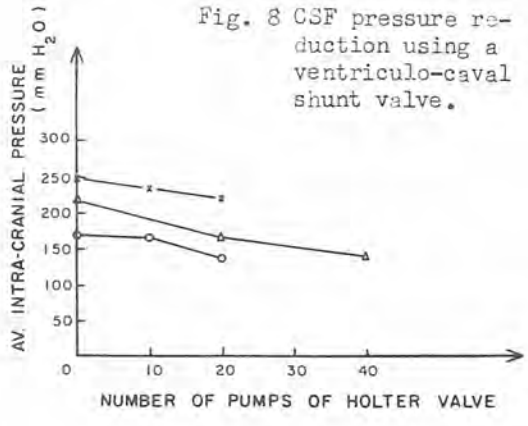

Fig. 8 CSF pressure reduction using a ventriculo-caval shunt valve.

AN AUTOMATED BLOOD COAGULATION TIMER

F. ENGLER

General

Blood coagulation is one of man's fundamental most complex protective mechanisms. Malfunctions of this mechanism are of considerable medical significance.

The "whole blood clotting time" (W.B.C.T.) is one of the clinical measurement techniques employed in the examination of patients with coagulation defects. Blood is extracted from a vein by means of a plastic or silicone coated glass syringe and a timing device is simultaneously activated. One cc samples of this blood are placed in three glass tubes housed in a waterbath, maintained at body temperature (37°C). The tubes are tilted at 30 second intervals and monitored for coagulation. When all three samples have coagulated the average clotting time is calculated. This determination normally requires between 5 and 10 minutes. Other defects in coagulation such as congenital haemophilia or those induced by anticoagulant therapy, prolong the blood clotting time to intervals which, on occasion are in excess of 60 minutes. A unit developed by Litton Systems (Canada) Ltd. and the Institute of Biomedical Electronics of the University of Toronto determines the WBCT automatically. Previously this determination was a manual operation. Since the coagulation reaction is temperature dependent, the removal of the tubes for varying periods for tilting and observation introduced a source of artefact. Also the coagulation mechanism is accelerated by contact with glass so that the speed and tilting angle also influence the result. The Litton Unit is an attempt to remove these sources of extraneous variation.

The machine is portable and a clotting time determination once initiated does not require constant attention so that a number of determinations can be made simultaneously by a single technician.

The Blood Coagulation Timer has been standardized so as to provide sets of results comparable to those obtained by a highly skilled technician using the manual method. The "whole blood clotting time" measurement employing glass tubes has been of little use in the detection of hypercoagulability. When clotting times are measured with plastic tubes, the surface activation of the coagulation mechanism is reduced and the normal clotting time is prolonged to an interval of 20 to 40 minutes. It has been suggested that a short clotting time, under these conditions, indicates an overactive coagulation mechanism. The Litton Unit permits this type of measurement.

Litton Systems Coagulation Timer

The development phase of the "normal" Litton Coagulation Timer was plagued with the difficulty of calibration samples. "Standard blood" does not exist so that even minor changes in mechanics or electronics required a lengthy sequence of comparative tests involving the machine versus the technician. After a period of trial and error tests a number of basic design rules for a reliable tilt mechanism were established:

a) The tilt angle of the sample holder should not exceed 85 degrees. As an angle of 90 degrees is approached the path of blood flow is increased, and with it the agitation, thus causing the coagulation process to be accelerated. A machine with an 88 degree tilt angle resting on a very slightly inclined surface (1 degree) was found to produce results which differed considerably from those obtained on a level surface.

b) It is necessary to accelerate the tilting action so as to produce centrifugal forces which cause the liquid blood to flow. Our original specification called for a very smooth tilting action. It was found that smooth tilting produced a convex bubble inhibiting blood flow and causing an eroneous indication of coagulation.

The Blood Coagulation Timer consists of four main functional components:

a) A tilt block array containing three test tube compartments surrounded by manganin heating wires which maintain the block at blood temperature. Neon tubes and an associated photoelectric sensor are situated on opposite sides of each compartment. (If coagulation has not occurred tilting causes a flow of blood to obscure the light source with respect to the sensor).

b) A motor assembly containing two small induction motors which control the timing and tilting operations.

c) A printed circuit board containing: the temperature control circuitry, three photoelectric amplifiers, and logic circuitry.

d) A power supply which provides the various voltages required for the unit's operation.

Operation

The unit is initially switched to the "heat" position causing the temperature control system to begin operation. In less than two minutes blood temperature is reached and an indicator light begins to flash, indicating that the unit is ready for use.

The unit is then switched to "time" simultaneously with the extraction of blood from the patient. The three counters on the front panel have at this point begun to count in decimal fractions of a minute.

The three test tubes containing the blood samples are placed into the compartments of the tilt block which is still stationary and the unit is switched to "tilt".

During the first tilt, the block is automatically aligned to the required angular position, the sample holder is then tilted at intervals of 30 seconds through an angle of 80 degrees. After passing through this angle it remains stationary for a 2 second interval and then returns to its initial vertical position. If blood coagulation has occurred the path between the neon light source and the detector is uninterrupted during the tilt. If this condition occurs it stops the counter associated with the container in which coagulation has occurred. When all three counters have stopped the test is completed and the average coagulation time can be calculated.

LOWER EXTREMITY PROSTHESES

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The term "prosthesis" means a mechanical device for the replacement of a missing body part. Because amputations of the leg were most commonly done through either the thigh or shank, prostheses for these amputations became known as ABOVE KNEE and BELOW KNEE prostheses. Other types were named after the approximate anatomical or medical description of the amputations.

A simple classification system linking the amputation to the prosthesis would indicate that amputation was through either a leg segment or a joint. Joint amputations are DISARTICULATION amputations. The basic nomenclature for prostheses would then be forefoot amputation prosthesis, ankle disarticulation prosthesis, shank amputation prosthesis, knee disarticulation prosthesis, thigh amputation prosthesis, and hip disarticulation prosthesis. This scheme is in partial use now for disarticulation prostheses.

Current practice is to custom build most of the prosthesis. In all cases except for some temporary prostheses, the stump container, called the socket, is always custom made. New materials and techniques, along with increased understanding of the significance of the constants and variables in body dimensions hold a promise for prefabrication of sockets. Meanwhile, plastics are replacing hand carved wooden sockets and molded leather sockets used in artificial legs.

At the floor end, the SACH foot (Solid Ankle Cushion Heel foot) is most commonly used, despite functional limitations. The SACH foot consists of a wooden beam projecting forward to serve as a forefoot, surrounded by molded polyurethane rubber. A wedge of softer rubber at the heel (the Cushion) absorbs shock and provides the appearance of ankle action. SACH feet are prefabricated in a variety of sizes, so that hand work is not required to prepare them for use. This is a large factor in their favor.

Extension of this approach to all other elements of prostheses is the task of designers. A big step forward will be a change from the exoskeletal type of prosthesis now used to the endoskeletal type now developing. This type, made up of tinker-toy-like elements can be quickly assembled from prefabricated mass produced parts, and can be made adjustable in length and alignment of the segments. A variety of joint control mechanisms can be interchangeably installed. Cosmetic shape can be provided by plug-on forms surrounding the structure, a procedure already used on the Hydracadence prosthesis.

The hip disarticulation prosthesis best illustrates the possibilities for this scheme. The socket is custom made now. By using a frame-like structure for the socket, this part could quite conceivably be prefabricated since it is the skeletal frame, predictably variable, to which the socket must connect. A good joint system, the Northwestern University hip joint is already available. This joint allows adjustment of socket position relative to the hip joint axis. Three stove bolts clamp the socket between a plate inside the socket and a spherical surface outside the socket. A mating surface carrying the hip joint shaft and joint stops connects to the spherical surface by a single bolt which locks it in the required position. We have designed an aluminum fork with bearings to connect to the Northwestern University hip joint. Beneath the fork we use a Winnipeg wedge-disc alignment unit. This extends the range of alignment adjustment possible, and provides a plug for attachment of an aluminum tubing thigh section. A second wedge disc alignment unit is attached to the bottom end of the tubing. This is bolted to the knee block of the Winnipeg knee-shank unit. Aluminum tubing of the required length is clamped onto the lower end of the knee-shank unit to make the shank of the required length. A third wedge-disc alignment unit is attached to the bottom end of the tubing. A SACH foot is bolted to this wedge disc alignment unit. Elastic control cords connect across the joins. Assembly and adjustments can be made in minutes. Parts can be easily disconnected for replacement.

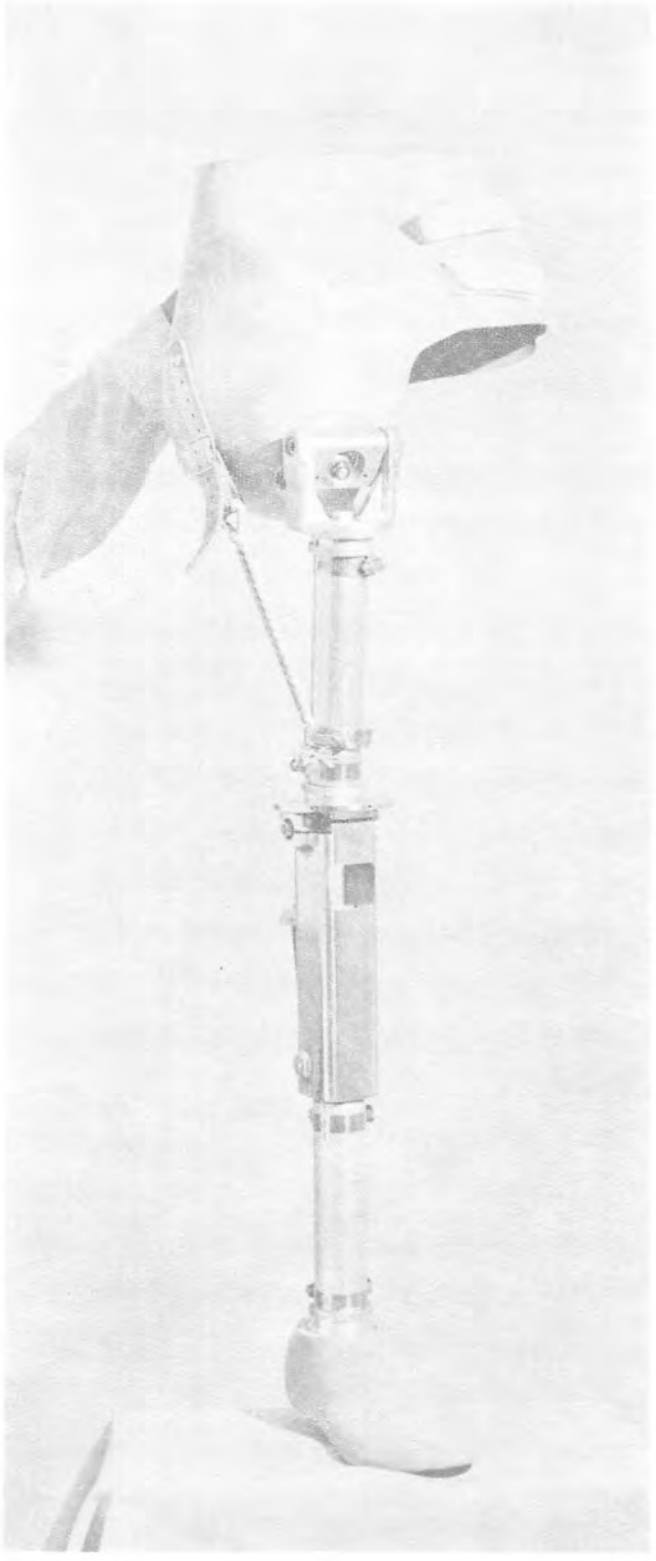
The scheme is adaptable to other prostheses. For the thigh amputee, the same knee-shank-foot assembly is used. A plastic receptacle into which the thigh socket plugs, is bolted to the wedge-disc alignment unit at the upper end. For the shank amputee, a plastic receptacle for the socket is attached to a wedge-disc alignment unit at the upper end of the shank tubing.

A comparable system must be developed for the knee disarticulation and ankle disarticulation prostheses. Currently, the knee disarticulation prosthesis has side-joints joining the socket to the shank. These are aligned, and the attachment straps are bent to fit the socket and shank by trial and error. No suitable swing control is available. It is difficult to set the two joints coaxially.

When a SACH foot is used on an ankle disarticulation prosthesis, considerable hand work is required to get correct alignment, height adjustment and to make the foot durable. Currently we are designing a metallic beam for molding into the SACH foot which could solve these problems.

A forefoot amputation prosthesis is merely an RTV Silastic foam rubber filler cast between foot stump and shoe. Its function is to maintain shoe shape and prevent the foot stump from sliding forward. It illustrates one important example of the use of modern materials.

With the plug-in, bolt-on tinker-toy approach, the use of prefabricated sockets, and the use of new materials which can be formed against the body, production, assembly, adjustments, repairs, salvage, and design refinements will be easy.



HIP DISARTICULATION PROSTHESIS

MYOELECTRIC CONTROL

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Introduction

The human central nervous system is the most versatile adaptive computer we are likely to have available in the foreseeable future. In most man-machine control systems only a minute fraction of man's computational and decision making mechanism is utilized. The channel capacity of the central nervous system far exceeds the bandwidth necessary to convey commands to the environment manipulating members: arms, fingers, feet. The redundant channel capacity rather than providing a margin of safety, results in tedium and a consequent deterioration in operator performance.

If we replace the human operator by a computer an obvious solution suggests itself: match the bandwidth of the computer to a number of auxiliary output devices, operating on a time multiplexed basis. In human terms, this analogy suggests that an operator be provided with additional arms and hands.

In military air and land vehicles it can be demonstrated, that a single human operator can frequently accomodate the decision and command rate assigned to a crew of three. The crew act as extra limbs; extensions of the vehicle commander's nervous system.

Myoelectric potentials offer a means of providing an interface between the central nervous system and external electro-mechanical devices.

Myoelectric Potentials

The myoelectric potentials are volitionally evoked by concious commands originating in the brain. A neurographic potential is transmitted from an axone of the anterior horn of the grey matter of the spinal cord. The dendrites of these motor nerves terminate on end plates of muscle fibres which in turn affect motion. Prior to the occurrence of movement a readily detectable muscle potential is produced. Heuristically one may regard the muscles as amplifiers of the electrical manifestations of mental commands.

The myoelectric project at Litton Systems (Canada) Ltd. is an investigation into the feasibility of evoking, training and combining these potentials for the control of external appliances.

Signal Processing

Electrodes

Surface electrodes although less sensitive than sub-cutaneous or intra muscular electrodes avoid the problems of infection and are easily applied and removed. Initially Beckman-silver-silver chloride electrodes were employed, however their relatively large surface area acted as a limiting factor in resolving adjacent myoelectric signal sources. To overcome this difficulty we developed a printed circuit process for preparing gold plated electrode pairs. The resulting electrodes have 3/16" contact area diameters and a spacing of 3/4" between the centres of these areas. Plastic washers are cemented to the surface of the printed circuit to provide a shallow well for electrode paste.

The electrodes have an impedance of 290 ohms (at 400 HZ), as opposed to the 160 ohms of 1/4" diameter Beckman electrodes.

Amplifiers

The amplifiers used in our apparatus have a gain of 10^4, an input impedance of 2M ohms, an output impedance of 5 ohms, a common mode rejection of approximately 100 db at 60 HZ. The frequency response falls 30 db from 5 HZ to 12K HZ.

The circuit configuration consists of transformer to provide a balanced input, and three feedback pairs, stable with respect to transconductance, current and transresistance.

Pulse Recognition Units

Since our apparatus makes direct use of the myoelectric pulses a definition in electrical terms of a "legitimate" pulse, is necessary.

The pulse recognition unit performs this function. It passes only those pulses whose amplitudes fall within preset high and low amplitude limits and whose durations lie within preset high and low limits. The device may be called a decision filter since it increases the signal to noise ratio, by deciding that certain of its inputs are noise rather than signal.

Frequency Threshold Unit

In order to effectively interface the myoelectric pulses with a mechanical device we require electrical pulse analogue of a reduction gear. The frequency threshold unit serves this function.

Frequency Threshold Unit (cont'd.)

It provides an output if, and only if, the number of pulses entering it within a fixed interval exceeds a preset threshold number.

Training Units

Two types of control termed discrete and con tinuous were investigated. Discrete control is defined as the ability to control the presence or absence of one or more myoelectric potentials. Continuous control is defined as the ability to control the frequency with which myoelectric pulses are generated over a selected time interval.

The discrete training unit generates a binary light display. The subject is connected to the training unit so that the presence of a potential on the electrodes causes a pattern of lights on the unit, to be illuminated. The subject attempts to match the light pattern generated by his potentials, to the one generated by the training unit. When he succeeds, a new light pattern appears. Counters in the unit monitor and record the time taken to achieve target matching and the number of errors occurring before matching.

The continuous training unit illuminates one of twenty lights on the circumference of a circle. The subject is equipped with two electrodes which enables him to rotate a pointer on the rotor of a stepping motor. He then attempts to bring the pointer into coincidence with the illuminated light in as short a time as possible. Upon doing so a new light is illuminated. The unit monitors and records the tracking parameters of the subject and permits a variety of tracking modes to be generated at the command of the experimenter.

Both of the above units provide the subject with a continuous auditory feedback signal, indicating his success in the control task.

Observations and Conclusions

The experimental subjects rapidly acquired the ability to score highly in terms of time and resolution on the continuous training unit. The discrete training unit tests were less conclusive. All of the subjects tested acquired some degree of control with three myoelectric sites while some achieved excellent control with four sites and were able to rapidly evoke any of the sixteen possible combinations. Some subjects were able to gain a precise control over the firing of single motor units.

Our experiments were largely confined to the facial muscles; the subjects located suitable muscle sites themselves. The variability of performance in discrete control may be accounted for by the success enjoyed by some subjects in locating sets of "uncoupled" muscle sites.

Manipulator Design and Control

A manipulator has been constructed which operates under myoelectric control. The manipulator consists of a gear arrangement which permits its horizontal and vertical motion in a cylindrical coordinate system about the origin, and extension and contraction of a radius vector. A hand capable of rotation, suppination and clenching will soon be added to the unit.

The manipulator employs the discrete and continuous control, modes. Three electrodes permit the operator and enable him to generate seven different binary patterns. Each pattern corresponds to the selection of a motor and its direction of rotation. A fourth electrode on the operator supplies pulses, whose frequency controls the rotation rate of the selected motor. The operator selects the motion desired and supplies pulses to the motor. The motor rotation is translated to an appropriate movement in direction and magnitude by the gear train.

When the arm reaches a designated point in space, the subject can generate a signal which causes control to be transferred from the arm to the hand. After the required hand motion has been completed the subject can return control to the arm.

Each myoelectric pulse is made to correspond to a definite quantum of motion by means of a digital servo mechanism. Pulses proportional in frequency to motor rotation are obtained from the motor shaft and are fed into the "down" input of an "up-down" counter whose "up" input receives myoelectric pulses. A resistor ladder provides an analogue voltage proportional to the instantaneous count resident on the counter. This voltage is in turn applied to the selected motor.

The relation between myoelectric pulse rate and motor velocity is of the form $\omega = K_2 M\ 1 + K_1 K_2$. By recording the signals supplied to the manipulator a programme can be prepared which would enable the manipulator to repetitively perform any task mastered by a human operator.

Our experience indicates that excellent control after only a few hours of practise can be achieved. The discrete mode of control manifests itself in jerky motions. In many, resembling the tentative arm movements of an infant.

Future Work

It is felt that the application of decision theory techniques will permit an efficient coding for motor selection states and a consequent improvement in smoothness of control. This approach resembles that of Tomovic and leads to a Markov process viewpoint of manual control and to a future control unit which takes the form of a likelyhood computer.

Finally by means of pressure pads mounted on the skin and strain gauges on the manipulator we hope to simulate the proprioceptive feedback afforded in normal limbs by the muscle spindle and the Golgi organ.

THE HUMAN OPERATOR IN A MYO-ELECTRIC CONTROL SYSTEM

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MYO-ELECTRIC CONTROL

A myo-electric control system is a man-machine system in which the control signal at the man-machine interface is the electric potential which accompanies voluntary contraction of human muscle. Such a system may be useful to persons who are physically handicapped, by providing them with replacement control modes [1, 2, 3]. Also, it may have significant military and industrial applications [4].

The present state of development is that several hundred hand prostheses using myo-electric control, manufactured in Russia, are now in use by patients, while systems designed in the western world are undergoing clinical trials or laboratory evaluation. These systems are intended for medical, rather than for military or industrial applications. Most, if not all, have been designed by empirical techniques, to provide "satisfactory" performance.

OPERATOR MODELS

Now that myo-electric control systems have been shown to be feasible, and since the crudely designed prototypes are working satisfactorily, it is natural that more sophisticated design procedures be sought in an attempt to produce systems which are more nearly optimum. Before a man-machine system can be designed by any formal technique, it is necessary that the human operator be represented by an appropriate mathematical model.

The science of model building for human operators in manual control systems is well developed. Obermayer and Muckler [5], state that "in the general area of manual control systems research, the most active single topic is the development of formal theoretical models for the description of human performance per se." However, the models which are devised for this purpose are of little practical value in the design of myo-electric control systems for the physically handicapped. It is my purpose in this paper to emphasize their serious limitations, and to suggest a reasonable approach to the problem for the system designer.

Adaptability

The human operator does not function in a constant or stationary manner, but rather adapts his performance to suit the task at hand and the characteristics of the external apparatus with which he must perform that task. McRuer and Krendel [6], state the plight of the designer of human operator models as follows, "the fact of human adaption makes description of the operator enormously complex when viewed in the large, and necessitates the set-up of simplified, constant situations with which one may have some hope of obtaining a simple behavioral model of engineering value." The situation is further complicated by the fact that the limits of human adaption are determined only partially by physiological factors, being greatly affected by psychological factors which vary significantly from day to day.

The research worker endeavouring to devise a mathematical model for the operator will usually consider a simple well-defined task, and will consider operators in good physical condition with controlled motivation. By contrast, the designer of a myo—electric control system for use by a physically handicapped individual in his daily activities must concern himself with a very wide range of activities, all of which must be performed safely and effectively despite a range of physical and emotional states much greater than that experienced by an individual who is not handicapped.

It is not feasible to design a simple model which will represent the operator satisfactorily throughout this variety of activities and moods. It is probable that a model of sufficient complexity to accomplish this would not be very useful to the designer.

Variability

The second characteristic of importance to this discussion is the variability of the human operator. By this I mean the differences in operator performance which are produced by day-to-day or hour-to-hour changes in motivation, by fatigue, or by illness. These represent a more serious problem

to the designer of models than the operator's adaptability, for these changes are not related to signals in the control system.

While these factors may be neglected in certain studies which relate to the performance of healthy individuals in well-defined tasks for limited periods of time, they may not be neglected by the designer of a prosthetic system. Moreover, they cannot be modeled.

Performance Index

Finally, the process of developing a mathematical model for the human operator requires that a suitable performance index be established. In a tracking task, of the sort which constitutes many military activities, it is relatively easy to establish such an index which is meaningful. It is not nearly so simple to define a valid performance criterion for a prosthetic system which will be used by a patient for the great variety of activities of daily living. There is little use in "optimizing" a system if the performance criterion is not meaningful to the patient.

SYSTEM DESIGN

My object in presenting this paper is not to criticize research concerning operator models, or to minimize the importance of this work in the design of man-machine systems. Rather, it is to point out certain limitations which are particularly severe in the design of prosthetic systems, and to suggest a design approach which makes the best of the existing situation. These recommendations are not at all radical, but may, it is hoped, provide some new insight into these limitations.

All designers, and particularly those working with prosthetic systems, should make use of the adaptibility of the human operator in order to achieve the greatest degree of engineering economy.

It is obvious that the designer of myo-electric control systems for prosthetic applications must consider the variability of operator performance. He may not find it possible to express this variability in quantitative terms, but he will ignore it at his peril.

For the designer of control systems for prostheses, the search for a realistic performance criterion may be never-ending. I would suggest that a useful criterion, although one which is probably impossible to determine quantitatively, is that the system should be designed to permit adequate performance of a wide variety of tasks with minimal operator fatigue.

At the present time, it would seem essential that the designer concentrate upon obtaining "adequate" system performance by empirical means, where no better means can be found. However, I would strongly recommend that one additional step be taken.

This step, which is almost invariably omitted, by the designer, is the determination of the characteristics of the external control system which he has created, and the inclusion of this information in any publication of design data. If this were done, the literature would contain an increasing number of examples of "satisfactory" control systems with their important characteristics tabulated for convenient reference. This, in itself, would be a considerable help to the designer of new systems for similar applications.

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A METHOD OF PROVIDING KINESTHETIC FEEDBACK FROM PROSTHESES

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INTRODUCTION

Myoelectric control is currently allowing amputees to operate their prostheses in a manner which is far more natural than with comparable cable operated equipment, and if the current trend continues these new prostheses will replace more functions than has been practical with cable operation. However, the improvement is not without penalty because by its nature myoelectric control removes the kinesthetic or position feedback which the amputee had with his cable operated prosthesis. This is due to the fact that the magnitude of the myoelectric signal (which is proportional to muscle tension) controls the velocity and/or force of its prosthesis where as in cable operated equipment body position is directly related to some prosthesis position. Therefore the burden of monitoring position with myoelectric control falls primarily on the eyes. As new degrees of freedom are added this task may become so difficult that smooth simultaneous operation of several joints -- as required in eating -- may be either impossible or require the amputee's complete attention.

In an attempt to anticipate this problem a research program was initiated last year, first to take a broad look at displaced kinesthetic feedback and second to make and test a practical system. It was decided to investigate a joint with a single degree of freedom to avoid unnecessary complication. The elbow was selected for this reason and because a myoelectrically controlled elbow* is under development in the Boston area so that evaluation of a feedback device may be incorporated in the arm development program.

The present discussion is intended to present our current results and proposed program in order to create interest in this problem at other centers and to stimulate suggestions for our own work.

* Being developed by Liberty Mutual Insurance Company, Massachusetts General Hospital, and Massachusetts Institute of Technology. For further information see Reference 5.

REQUIREMENTS FOR A KINESTHETIC FEEDBACK SYSTEM

In order to prevent the results of this work from being purely academic it was felt that human factors should be given primary consideration. Therefore our initial requirements became: 1) The system must not burden the senses used in normal daily living. 2) It must not draw undue attention to its user. 3) Its use and attachment should require a minimum of effort on the amputee's part. As a result of these requirements the tactile sense was chosen as the man-machine communication link, and the location of the tactile stimulators was chosen to be the area of the stump inside the socket so that the prosthesis and feedback system would be a single package.

CODING AND THE DISPLAY

The requirements for the feedback display were that it should be capable of indicating elbow position within 10 degrees. Although this is much less accurate than the physiological system it appeared to be a good compromise between complexity and accuracy. There has also been an attempt to maintain some spatial or temporal analogy between the joint angle and the display in the hope that this would facilitate learning. Below is a summary of several displays which are under consideration.

1. A direct mapping of elbow position on the upper arm through some mechanical linkage. In this a tactile stimulator is moved up and down the upper arm as the elbow is flexed and extended. This has the advantage of simplicity; however, if the stimulator is inside the socket, fitting problems may arise and unless the contactor is vibrated adaption may soon hide the stimulator's location.

2. The use of muscle receptors to indicate joint position. R. Alter[1] recently described an informal experiment in which an amputee having a long above elbow amputation (however, none of the joint remained) reported that he could sense a change in the position of his missing arm as his biceps and triceps muscles were moved by external pressure through the skin. The position would appear to remain constant when the muscles were held in one position, even though they were held long enough to allow the skin to adapt to this constant pressure. This would indicate that the position signal was originating from some deep, slow adapting receptor rather than the cutaneous receptors. Although current neurological theories do not indicate

that the muscle receptors play any role in the kinesthetic senses we shall attempt to repeat these results with a group of above elbow amputees this fall to determine whether or not this phenomena is common.

3. The use of a vibrator whose frequency varies with elbow position. Sherrick[6] has pointed out that in the physiological measuring system single thalmutic cells respond with an increase of impulse frequency as the joint is flexed or extended from some resting position. This type of display might therefore be easily accepted by the C.N.S. However, the work of von Békésy[2,3] and Gebhard[4] indicate that the tactile display may pose serious practical problems because of man's complex tactile frequency sensation which is effected not only by frequency but also by temperature, adaption, and intensity.

4. G. von Békésy's phantom sensation phenomena allows a vibration to be felt at any desired point on the skin between two vibrating stimulators. Since these stimulators may be fixed in the wall of the socket motion of the sensation may be achieved without any movement inside the socket, and elbow position may be indicated by the position of the sensation on the skin. This tactile phenomena is similar to directional hearing in the ear because the time difference between the onset of the two stimuli changes the apparent location of the stimulus. If both stimuli are equally "loud" and are presented at the same instant neither stimulus is felt directly, rather they will combine to form a phantom sensation midway between the two stimulators. Although our tests are not complete present results are promising for closely spaced stimulators. However, for separations of 8 to 10 inches the sensation becomes less distinct and as a result accurate sensation location is difficult and may require too much concentration on the part of the amputee.

6. Detection of elbow position by knowledge of its relative position. A tactile stimulus would indicate an incremental change of angle from some recent known position. The rationale being that the arm position will be optically monitored frequently enough to make the display of absolute position unnecessary.

EVALUATION

Preliminary evaluation of these displays will be made on normal subjects. They will operate a myoelectric elbow from their biceps and triceps muscles while their own elbow is immobilized. The position of the controlled prosthetic elbow will then be fed back to the subject through the display to be tested. The displays will be evaluated by comparing the subjects' performances on a tracking task. The best feedback device selected on this basis will then be tested on the group of amputees evaluating the myoelectric elbow.

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COMMUNICATION SYSTEMS FOR THE PHYSICALLY HANDICAPPED

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1. INTRODUCTION

A need exists to establish communication from people with severe physical handicap and no speech, but who possess normal or near normal intelligence. Systems have been designed which provide typewriter output from the handicapped, such as those of Perron (1965), Roy (1965) and the POSM system of Maling and Clarkson (1963). Other communication systems have been built which do not type or otherwise provide permanent output, such as that of Miller (1964).

In order to achieve satisfactory communication from these people, two areas of research exist. Firstly, it is necessary to construct transducer devices which are accurately and rapidly controlled by a person with poor motor control; and secondly, one must then make most efficient use of these transducer inputs by maximizing the amount of information transmitted by one movement. In the system discussed here, this is approached by making it possible to type an entire word as the result of one actuation of the input transducer.

2. TRANSDUCER CONFIGURATIONS

The movements exhibited by candidates for this type of aid vary widely in characteristics. Spasticity, poor directional control, temporal instability (difficulty in transmitting something like Morse Code) and weakness of movements make the task of the transducer designer difficult. Often, custom-built devices intended for use by one person are designed, such as the hand key made by Perron.

At the University of Saskatchewan an attempt was made to construct a more general device, based on the relative familiarity with keyboard-like transducers possessed by children at the Physical Restoration Centre in Saskatoon. A keyboard as shown in Fig. 1 was built, and it eliminated the need to depress mechanical switches by using photocells as the switching media. Insertion of a light into a hole tapered to provide guidance for a poorly directed movement triggered two detectors simultaneously, one for column and one for row detection. These devices provided an array of 100 "keys" and could be operated by lights held in the hand, feet or mounted on headsticks.

3. THE A.T.S. (AUTOMATIC TYPING SYSTEM)

The block diagram of a system constructed at the University of Saskatchewan is shown in Fig. 2. This device has a permanent vocabulary of approximately 400 letters, characters or words up to 8 letters in length. The words used in the machine were selected from studies of word frequency distributions in written compositions up to the Grade VIII level.

When a selection is made on the light-keyboard of Fig. 1, one of the words or characters is typed out on a remotely operated type writer. Each of the 100 "keys" has four potential meanings, and a selection is made by first selecting one of four keys representing the first, second, third and fourth word of each key, and then selecting the key itself.

This is achieved through the use of 2 stepping switches as the primary system for routing typing pulses to typewriter solenoids in sequence to form a word. An electronic pulse control system provides properly timed stepping and typing pulses, and selection logic is achieved by relays and diode matrices.

4. RESULTS

The above system has been installed at the Physical Restoration Centre in Saskatoon. An educational testing program intended to test vocabulary score before and after practice with the machine has been initiated to assess the effects on approximately 6 children. There is a great deal of enthusiasm and acceptance of the device by the children of the Centre.

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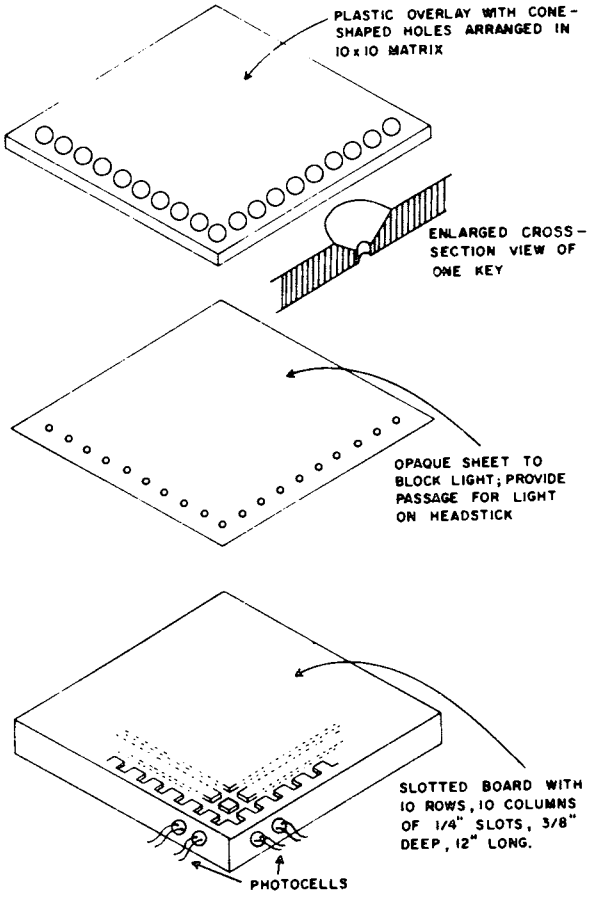

FIGURE 1 LIGHT SENSITIVE KEYBOARD

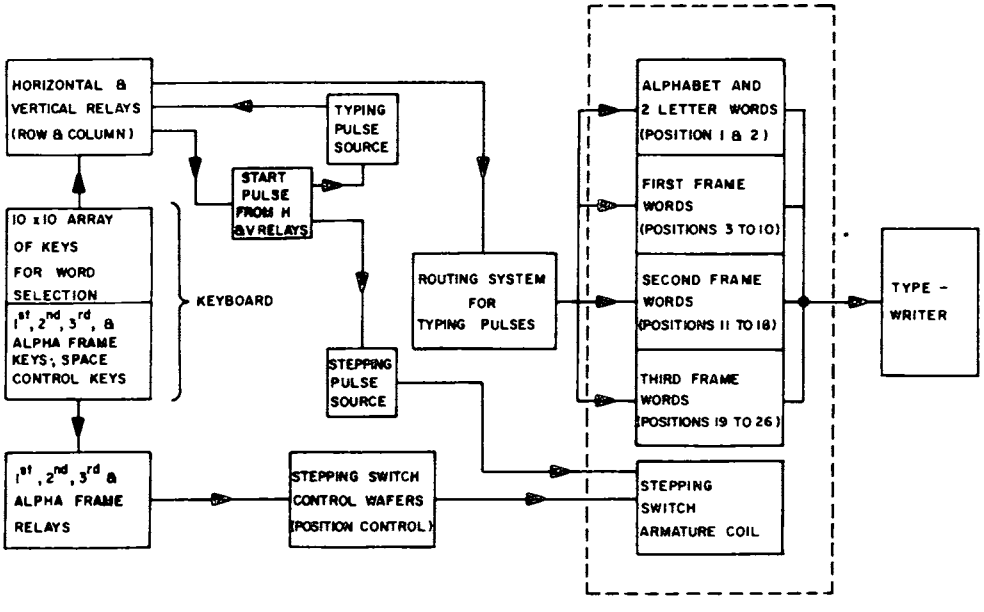

FIGURE 2 BLOCK DIAGRAM OF A.T.S.

PROGRAMMED SPEECH OUTPUT SYSTEM FOR THE HANDICAPPED

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1. INTRODUCTION

The Programmed Speech Output System described herein is an extension of work done by the Division of Bio-Medical Engineering at the University of Saskatchewan. This device, an "Automatic Typing System", a description of which appears in this digest, allows the physically handicapped to communicate via a typewriter. Whereas the output medium for the Speech Output System is a loudspeaker.

The input transducer requirements are exactly the same as those described in the "Automatic Typing System", and therefore this project was not concerned with the problems associated with input.

2. DESIGN CRITERIA

The overall objective in developing the "Programmed Speech Output System" was to design an inexpensive, wheel-chair portable communication device which offered a vocabulary of 400 words to victims of aphasia. Another essential requirement is that the output information rate not be limited by the apparatus itself but only but the operator's selecting speed.

3. PROGRAMMED SPEECH OUTPUT SYSTEM

The heart of the system is a two speed endless loop tape deck. The loop is about 7 inches across and 15 inches in circumgerence. The two speeds allow scanning at high speed - 15 inches per second - and playing out at low speed - 1.5 inches per second. These two speeds are controlled by magnetic clutches and result in an average access time of 1/2 second.

The input circuitry consists of 24, 1 bit buffers which could be considered to form a 10 x 10 x 4 matrix. Setting one buffer in each group constitutes an input.

Referring to Fig. (1), it can be seen that the 10 horizontal inputs and 4 level inputs uniquely define 1 of the 40 heads on the memory tape via their corresponding gates. Each head reads one track containing 10 words - thereby giving a total vocabulary of 400 words. This completes the selecting of the track on tape and it remains now, in order to uniquely define 1 of the 400 possible positions on tape, to select the longitudinal position.

The longitudinal position is sensed by the photo electric sensors on the edge of the memory tape. Pulses derived from these sensors are counted by the binary counter. By energizing one of the vertical inputs, its corresponding buffer will be set with a digital number. This number is converted to its binary equivalent and is compared to the contents of the binary counter by means of the coincidence detector. When the two numbers are identical the coincidence detector outputs a "1" to the "AND" gate. If the outputs of the "OR" gates that are connected to the horizontal and level buffers are also in the "1"'s state-indicating selections have been made, the output of the "AND" gate will also be in the "1"'s state and will enable the pulse gate. The pulse gate passes pulses derived from the trailing edges of the photo electric count holes. Since the leading edge of the holes is used to advance the binary counter, the tape deck cannot slow down and play out only a portion of the word. If coincidence is not detected within one hole length (approx. .1 in.) of the beginning of the word the tape must make one complete revolution and start at the beginning of the word.

When all the foregoing conditions have been fulfilled favorably, the pulse gate passes the pulse from the appropriate schmitt trigger to the clutch flip-flop which slows the tape down for play back. Play back is terminated when the binary counter is advanced by the next hole which destroys coincidence. This action is used, via the reset pulse amplifier, to reset all the input buffers to zero. The unit is now ready to accept another input.

4. PRESENT STATUS

The selecting sections of the electronics are working very well and have been operating without a failure since completion. The audio is, however, of rather low quality and there has been difficulty experienced in splicing the mylar tape and also in keeping the tape in contact with the head. It is expected, however, that these problems will be overcome shortly.

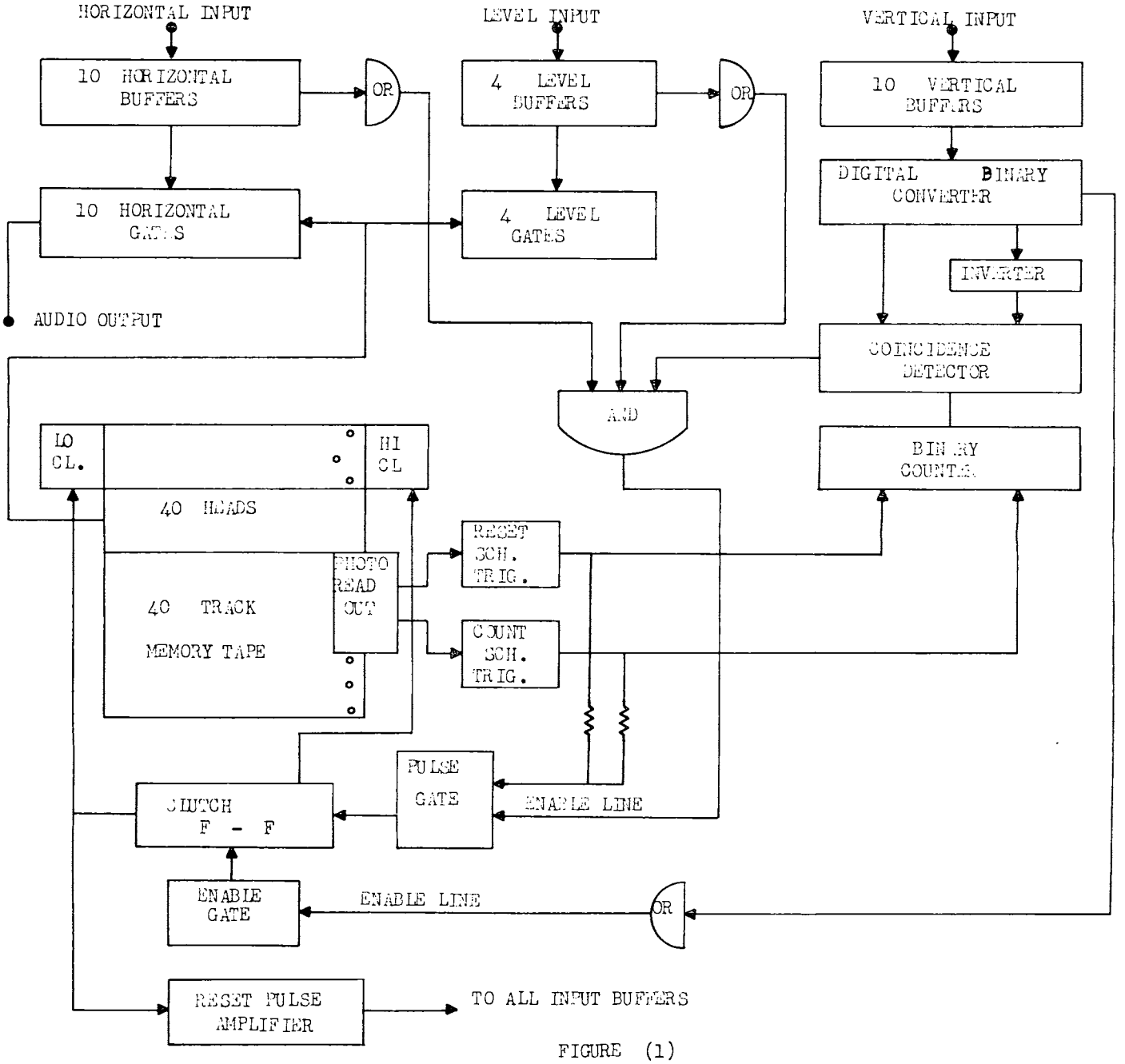
HORIZONTAL INPUT
LEVEL INPUT
VERTICAL INPUT
10 HORIZONTAL BUFFERS
OR
4 LEVEL BUFFERS
OR
10 VERTICAL BUFFERS
DIGITAL BINARY CONVERTER
10 HORIZONTAL GATES
4 LEVEL GATES
INVERTER
AUDIO OUTPUT
COINCIDENCE DETECTOR
AND
LO CL.
HI CL
BINARY COUNTER
40 HEADS
RESET SCH. TRIG.
PHOTO READ OUT
40 TRACK MEMORY TAPE
COUNT SCH. TRIG.
PULSE GATE
ENABLE LINE
SWITCH F - F
ENABLE GATE
ENABLE LINE
OR
RESET PULSE AMPLIFIER
TO ALL INPUT BUFFERS

FIGURE (1)

COMPUTER SIMULATION OF THE VARIABILITY OF THE NORMAL SA NODE PACEMAKER FIRING FREQUENCY

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In previous investigations it was observed that the beat to beat variability of the heart rate (the R-R interval) significantly decreases in conditions which result in an increase of the heart rate. A typical result is shown in Fig. 1 which depicts non-sequential R-R interval histograms of a subject at rest, and during exercise. The calculation of the first and second statistical moments gave, in this particular case, factors of variation (ratio of second and first moment) of .06 and .085 for exercise and resting respectively. For distributions that are not symmetric, the third statistical moment can be used as an index of the skewness (third moment is positive, if the difference (mean-mode) is positive). It was observed that the skewness changed from negative value in the resting condition to a positive value during exercise.

No significant variability was detected in P-R intervals. The P-R interval is an index of the conduction time from the sinus pacemaker region in the right atrium to the ventricles. It is therefore reasonable to assume that the variability of the heart rate is due to variations of the cell potential of the SA node cells.

Model experiments were carried out to study how some known physiological factors would influence the variability of the firing rate of the SA node cells. A typical action potential of these cells is shown in Fig. 2. The interval between two firings (I) is expressed as a function of three variables, the duration of the action potential (T), the pacemaker potential (h), and the slope of diastolic depolarization (s). These three variables are assumed to be subject to random fluctuations or "noise" such as might result from random variability in the ion fluxes across the membrane. If a normal distribution with a certain variance, and mean value equal to zero, is assumed for this noise, then the probability density function expressed by equation (1) can be calculated:

$$h(I) = \frac{1}{2\pi\sigma_T\sigma_h}\int_{-\infty}^{+\infty} \frac{m_s + m_h\left(\frac{\tau\sigma_s^2}{\sigma_h^2}\right)}{\sigma_h\left(1+\left(\frac{\tau\sigma_s}{\sigma_h}\right)^2\right)^{3/2}} \cdot \exp\left[-\frac{1}{2}\left\{\left(\frac{I-\tau-m_T}{\sigma_T}\right)^2 + \left(\frac{\frac{m_h}{m_s}-\tau}{\frac{\sigma_h}{m_s}}\right)^2 \cdot \frac{1}{1+\left(\frac{\tau\sigma_s}{\sigma_h}\right)^2}\right\}\right] d\tau \qquad (1)$$

where m_s = mean value of s,
m_h = mean value of h,
m_T = mean value of T,
σ_s = variance of s,
σ_h = variance of h,
σ_T = variance of T.

A graph of this density function is shown in Fig. 3 for a chosen set of parameters. The influence of the inherent random noise on the variability and skewness can now be estimated with aid of equation (1).

The remaining difficulty is caused by the fact that the SA node cell potentials are influenced by the vagus and the sympathetic nerves. In the resting state the vagus influence is dominant and the amount of vagus control varies with respiratory cycle in a characteristic way.

To obtain the R-R distribution from a situation where the vagus activity is approximately constant, a signal was derived from the ECG, which turned on a yellow light during three R-R intervals and a red light during the following three R-R intervals. This cycle was repeated and the subject then inspired and expired according to the lights. In this way respiration and pacemaker activity were locked in phase. Since every sixth R-R interval occurred at the same phase of the respiratory cycle and approximately the same state of vagus activity, it became possible to compute distribution curves for R-R intervals collected from a known phase of the respiratory cycle and thus minimize variability caused by vagus. The variance for each single interval, obtained from the calculated distribution curves, is, of course, less than that for the total resting ECG. What is surprising, however, is the fact that the variance decreases with increasing duration of the intervals. This is possibly due to different vagus activity at each interval.

An attempt was made to explain this decrease in variance by assuming a time variant amount of transmitter substance between endplate and pacemaker cell membrane. If each action potential in the vagus nerve releases a certain amount of transmitter (A) and the transmitter is bound according to

$$-\frac{dA}{dt} = \text{const.} \cdot A, \qquad (2)$$

then the amount of transmitter as a function of time is a decreasing exponential function, as shown in Fig. 4.

According to experiments done by Trantwein, the slope of depolarization and the duration of

action potential depends on vagus activity. If one now assumes that these variables depend on the amount of transmitter substance present in the gap between endplate and cell membrane, it is possible to calculate the R-R intervals for a given firing frequency of the vagus nerve. Two constants k_1 and k_2 were chosen to represent the influence of (A) on s and T.

$$s(t) = s_0(1-k_1 \cdot A(t) \qquad (3)$$

$$T(t) = T_0(1-k_2 \cdot A(t)$$

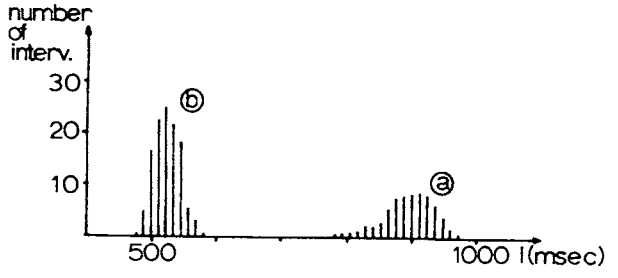

Fig.1 Nonsequential histogram of 100 R-R intervals during (a) resting and (b) exercise.

The interval distributions predicted using Eq. (3) are in close agreement with the experimental results, as far as variance is concerned. The simplified model used here does not adequately explain the observed skewness of the interval distribution and refined models must be sought and tested before the biophysical significance of the skewness can be adequately understood.

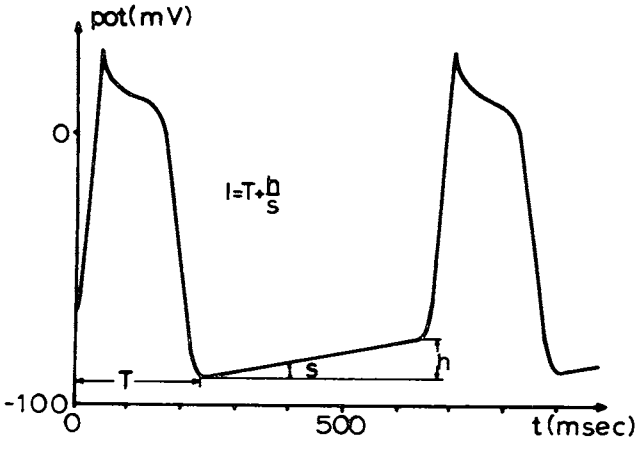

Fig.2 Cellpotential of a SA-node cell. (for symbols see text)

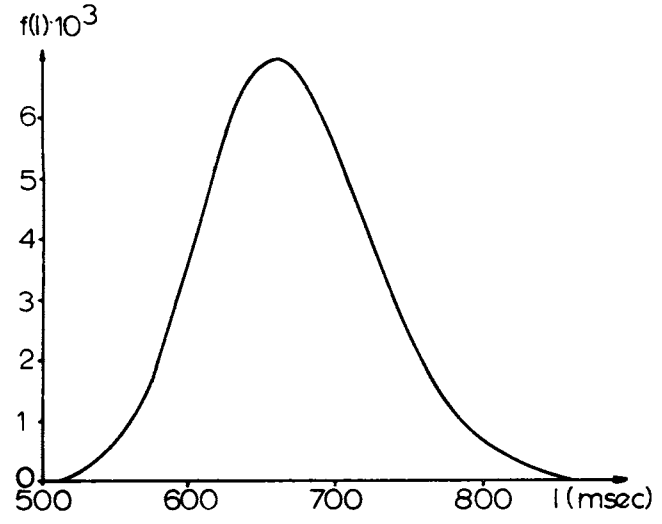

Fig.3 Density function calculated from Eq. (1), with parameters m_s=30 mV/sec, m_h=15 mV, m_T=200 msec, σ_s= 3 mV/sec, σ_h= 1 mV, σ_T= 25 msec.

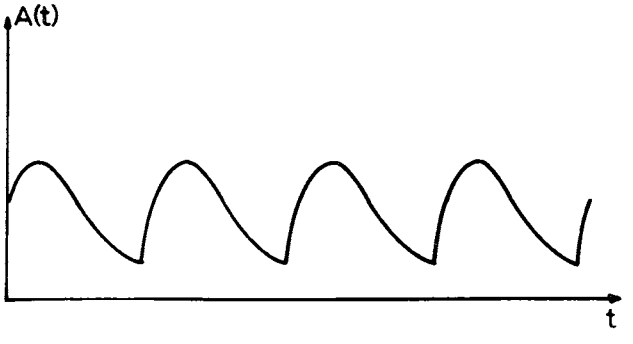

Fig.4 Amount of transmitter [A] between endplate and cell membrane as a function of time.

A MATHEMATICAL DESCRIPTION OF THE EEG

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The attempt in this study is to find a way in which a 10 to 30 second sample of electroencephalogram (EEG) could be described by an equation the coefficients of which could be adjusted to accommodate a wide range of normal EEGs.

The first requirement is to find a suitable mathematical expression or model that will have a reasonable chance of satisfying the requirements. Lowenberg (1959) showed that a number of EEG patterns can be synthesised by passing noise through a narrow band filter. Also in 1959 Weiss used the narrow band filter model to explain the beating effect that appears in the autocorrelogram of an EEG with a dominant alpha rhythm. He also suggested that the amplitude distribution of the EEG and its envelope be studied further as a means of validating the narrow band filter as a model. Saunders (1964) in his studies on the amplitude distribution of the alpha activity found that the density function of the envelope of the alpha rhythms had a Rayleigh distribution as does the output of the narrow band filter with a white noise input. With this background it seemed reasonable to use the narrow band filter as a model.

The approach taken was to adjust the coefficients of the known correlation function of the output of a narrow band filter to obtain the best fit to an autocorrelogram of EEG. This autocorrelation function is given by

$$\phi_{xx} = Ae^{-a\tau} \cos b\tau \qquad (1)$$

where A, a and b are the adjustable coefficients and τ is the time shift. It was found that this model adequately described a large number of EEGs recorded in the parietal-occipital region but was of little use in other regions. Visual inspection of the computed correlograms suggested that the addition of two more exponential functions to the model would provide a more universal model. With the limited tests made to date, this has proved to be true and the mathematical model is now as follows

$$\phi_{xx} = (1-X_3-X_5)e^{-x_1\tau} \cos X_2\tau + X_3e^{-x_4\tau} + X_5e^{-x_6\tau} \qquad (2)$$

where x_1 to x_6 are the parameters and τ is the time shift.

The fact that equation (2) provides a close fit follows directly from the work of Lanning and Battin who have shown that any autocorrelation function can be desribed by an exponentially damped cosine plus an exponential series.

If one wishes to assume a white noise input to a system whose output has an autocorrelation function described by the equation (2), then equation (2) is the square of the system weighting function. From this function one could derive the equivalent transfer function. The assumptions made in arriving at such a transfer function are that the random process is stationary and the system is a linear, constant parameter system operating for infinite time. While there can be serious questions as to whether or not anything useful can be gained by assuming such a transfer function to describe the dynamics of the brain, there is no reason why the autocorrelogram of a sample of EEG should not be represented by equation (2). Only by obtaining the coefficients of this equation for a large number of samples of EEGs recorded under various conditions can it be determined whether or not such a model will serve any useful purpose. The vast number of calculations required can best be done using a digital computer system.

The EEG to be sampled is usually available as a voltage signal from a magnetic tape recorder. However, it is quite possible to connect the subject directly to the computer through suitable amplifiers. A specified number of samples of the EEG are stored in the computer memory by using an analog-to-digital converter sampling at a preset rate. While the EEG is being sampled the first four moments (mean, variance, skewness and kurtosis) are calculated. These values are typed out and then the autocorrelogram is computed. When this is finished, the coefficients of the model are obtained using a hill-climbing method operating on a mean-squared error criterion. When a satisfactory fit has been achieved, the coefficients are typed out. If desired, the computed autocorrelogram and the fitted function can be plotted on an X-Y plotter. The whole process is then repeated for a new sample of EEG.

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THE AVERAGING TECHNIQUE AND SIGNALS OF BIOLOGICAL ORIGIN

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The last few years has seen the introduction into electrophysiology of a number of techniques made possible by advances in electronics. One of the most widely used of procedures is the signal averaging technique. Originally introduced into the physical sciences for the extraction of signals from noise, the technique is used in physiology for what is apparently a similar purpose. In most instances the signal is a potential change originating in the brain substance and recorded from the scalp (evoked potential), and the noise is the fluctuating electrical potential which is continuously recorded from the scalp and which derives from the brain (the electro-encephalogram). The problem that arises is due to the fact that both the electro-encephalogram and the evoked potential are of similar amplitudes. When averaging is performed by the usual process of signal summation, it is very easy to demonstrate the existance of stable waveforms following the application of repetitive brief sensory stimuli. These waveforms show great reproducibility even when recording periods are separated by intervals as long as a few months, and thus appear to provide a stable criterion for many types of testing of brain function.

More careful examination shows that the situation, although apparently simple, is actually a highly complex one, and one in which a change in the averaged signal may come about in unexpected ways. In this type of biological system the relationship between signal and noise is not a simple passive additive one, but may be a very complex one in which the noise actively modifies the signal. For example, if, when recording evoked potentials from the cerebral cortex signal to noise ratio is improved by cutting a hole in the skull and placing the recording electrode directly on the brain, then as Bindman, Lippold and Redfearn (1964) have shown, the amplitude and polarity of an evoked potential varies considerably and is largely dependent on small differences in the D. C. level of the cortex. We have demonstrated even more complex relationships to exist between evoked potentials and background electrical activity in the hypothalamus of the chloralose anaesthetized cat (Abrahams and Langworth, 1963), and by the use of a circuit developed by McDonald (1964) we have been able to quantitate the effect of this background activity on a computed average and have shown that it can be considerable.

The interpretation of changes in this type of averaged waveform in terms of neural functioning can thus be difficult. If such an interpretation is to be made, positive evidence must first exist to show that active interfering signals have been compensated for or excluded.

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PATHS TRACED BY THE VISUAL AXIS DURING SACCADIC EYE-MOVEMENTS

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SUMMARY

The eye-marker camera of Mackworth & Llewellyn Thomas (1) was modified so that the actual path traced by the visual axis of the left eye during a movement could be recorded. It was found that during saccadic movement these paths are seldom straight, and that the curvature appears to be a function of the direction and length of the movement. An experiment was performed in which movements of controlled magnitude and directions were stimulated in twelve subjects. The results suggest that the lateral rectus consistently contracts later than the five other extraocular muscles. A simplified model of the eye muscle system, incorporating this time delay, was programmed on a digital computer, and the calculated paths agree well with those found by experiment.

METHOD

Saccadic eye-movements were elicited by nine small neon lights, arranged in a circular pattern. Each stimulus represented a five-degree movement in one of eight directions, to or from the primary fixation position at the centre of the circular pattern. The stimulus was synchronized to the shutter of the eye-marker camera, which was slowed down to record about 1 frame every 750 milliseconds. The shutter remained open for about 333 ms after the application of each stimulus, and thus the resulting path of saccadic movement was nearly always recorded. It should be noted that the eye-camera partly obscures the field of view of the left eye, and therefore this experiment actually measured movements of the left eye in response to stimuli applied to the right.

Two stepping switches were used to program the sequence of stimuli. A pseudo-random sequence, 468 stimuli long, designed on a randomized block-plan, was wired into these stepping switches. This sequence was repeated exactly for each of the 12 subjects. Thus a total of 3525 complete saccadic paths was recorded and analysed.

ANALYSIS AND RESULTS

The paths of movement resulting from these stimuli were analysed using standard statistical techniques and were found to be consistent. Each direction of movement gave a distinctly different path, yet corresponding movements showed close agreement between subjects. Eight of the sixteen paths for the left eye had one feature in common: towards the end of the movement they showed an abrupt hook to the left. This hook implies that the extraocular muscles are not being innervated simultanesouly; either the lateral rectus muscle is consistently contracting late, or the medial rectus is consistently relaxing late. The remaining eight paths, which do not consistently hook left, all involve horizontal or diagonal movements to the left, where the lateral rectus is the prime mover.

There is an anatomical reason for suggesting that the lateral rectus is, in fact, contracting late. This one muscle is activated from nerve VI, the abducans nerve, the nucleus of which is located some 25 mm caudally to the nuclei of nerves III and IV, which activate the other extraocular muscles. Thus it is reasonable to expect a time difference between the innervation of the lateral rectus and that of the other five muscles.

A COMPUTED MODEL

A model of the extraocular muscle and eyemovement system, shown in Fig. 1, was programmed for an IBM 7094 computer using MIMIC language (2). A simple mapping equation was first used to convert the 2-dimensional field-of-view coordinates into 3-dimensional muscle coordinates. Three second order linear systems were assumed for the oculomotor responses, using $\zeta = 0.7$ and $\omega_n = 240$ rads/sec (3). The step input signal to the second order system representing the lateral rectus muscle was delayed by about 14 milliseconds. The resulting muscle position coordinates at 1ms intervals were then mapped back into the field-of-view coordinates. The computed paths

agree well with the paths recorded in the experiment.

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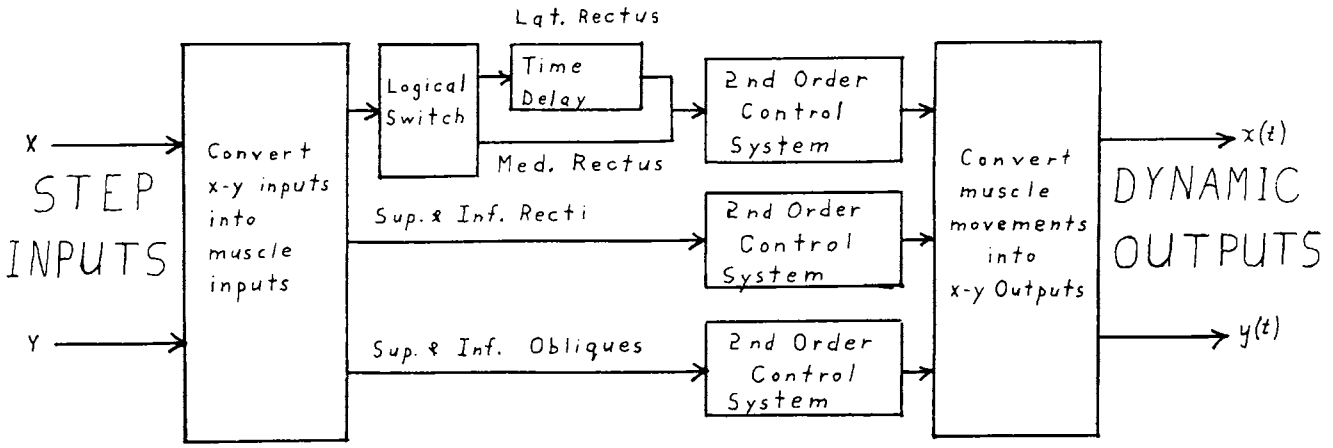

Fig. 1.

Saccadic Eye-Movement System Simulated on a Computer

A TECHNIQUE FOR THE MEASUREMENT OF PUPILLARY RESPONSE IN CONJUNCTION WITH OTHER PSYCHOPHYSIOLOGICAL MEASURES

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F.R. WAKE, Carleton University, Ottawa

The pupillary reflex has been under active investigation since the 1930's, but continuous and accurate recording with high-speed high definition is a development which has occurred only as a result of the technological advances of recent times. Reeves' conditioning studies (1918), Bender's pharmacological studies (1933), and Lowenstein's work (1942) are all pioneering efforts in this field. Since 1960, Hess and Polt of the University of Chicago have been studying the pupillary response as it relates to interest values and theirs were the first studies to make use of the high speed-fast film combination. Simpson and Paivio at the University of Western Ontario have recently developed a technique for use in learning studies.

The method presented here is felt to be an improvement on the Hess-Polt technique as it allows for greater relaxation of the subject while facilitating the recording of other parameters simultaneously. Utilizing the high definition qualities of the equipment, it also enables the recording of the pupil against a dark iris with as little difficulty as against a light iris.

The system makes use of an Arriflex 16mm. high speed motion picture camera fitted with a 28" Cine Tele-Zenar lens. The camera operates at 16 frames per second at f4, with the lens positioned 28" from the subject's eye. An Offner Type R four channel Dynograph fitted with a stimulus timer was used to measure heart rate, respiration, and galvanic skin response. Visual stimuli, consisting of 2 X 2 Kodachrome slides can be presented to the subject using rear-projection from a Kodak Carousel 35mm. automatic slide projector. Half of these stimuli are control slides having no stimulus value, and all are equated for light value.

The entire photographic arrangement is supported above the subject who is lying on a standard hospital bed. The pupil is photographed by means of an arrangement of mirrors above the subject's right eye, while a second series of mirrors enables stimuli to be presented in a horizontal plane above the subject. Front surface mirrors are used to eliminate distortion. All switching arrangements are led to a control panel some distance from the subject and out of his range of vision.

Several films were tested along with several different illuminants in order to obtain a clear delineation of the edge of the pupil. It was found that a proper balance of both light source and film speed could be obtained by using Kodak Linograph Pan film with a single #2 photoflood lamp, and that this balance could be maintained for both dark and light eyes by maintaining the illuminant at a 45 degree angle from the subject and displacing it 2 feet for dark eyes and 3 feet for light eyes. The subject's head is held steady in a padded head-clamp. The subjects report no discomfort from either the head-clamp or the proximity of the light.

The other parameters, heart rate, galvanic skin response, and respiration, are recorded in a standard manner leaving the subject's right hand free to control the remote stimulus switch. Stimuli can also be controlled automatically through switches on the projector with intervals of 5, 10, 15, or 20 seconds. However, subject control adds a further useful measurement, that of interest value, and for this reason the automatic timer is seldom used.

Film is processed in an automatic processing unit using normal D-72 developer. There is little doubt, however, that a very fine grain developer such as DK-20 would improve the contrast quality of the film. In this, some compromise was necessary as the processing equipment is the property of another hospital department. In this connection, the use of colour film was considered but rejected. Although it has the advantage of being grainless, it has the disadvantages of difficulty of processing, higher cost, and requires an illuminant which could do harm to the subject's retina.

Scoring of the film is achieved by mating a Kodak Analyst Stop Motion projector with a rear-projection box. The projector is capable of both backward and forward motion at various speeds and also of stopping at any frame without damage to the film. Set at a standard distance from the screen and again using mirrors, a magnification of four times is obtained. The scorer sits in front of the screen, controlling the projector from a remote control box and measures each pupillary image

vertically in millimeters using a movable plexiglass graticule. The screen is gridded at right angles so that the graticule is always vertical and the measurements are always made across the centre of the pupil.

Since the camera operates at a continuous speed of 16 frames per second, any interval of measurement can be selected. In this way, latency times are easily calculated and the behaviour of the pupil continuously monitored.

GAMMA-RAY CAMERA TECHNIQUES IN MEDICINE

N. F. MOODY, W. PAUL & M. JOY
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University of Toronto

The use of radioactive tracers in medicine is of ever growing importance and correspondingly great effort has been devoted to the design of instruments (cameras), which will map the areas of radioactive concentration on which diagnosis depends.

The first part of this paper attempts to bring together the appropriate isotope technology, and clinical problems, which are then used to evaluate successive developments in camera-like devices. Certain criteria are thus established for the ideal camera.

In the second part, attention turns to a specific class of camera based on image intensifiers, with which our group has been particularly concerned. The remainder of this abstract confines itself to a particular camera, under development, which incorporates the results of our researches. We believe it will go far towards meeting the ideal criteria.

The principle of operation is shown in Fig.1. The object under study (S) emits gamma rays (R) which are collimated in the usual way and impinge on a phosphor (P_1). A simple pin-hole (B) in the shielding material (L) is shown as the collimator, and a sodium iodide crystal mosaic is a suitable phosphor. Scintillations of light, created near the point of impact of each gamma ray, form the signals which will finally become picture elements.

As is well known, a large area of phosphor (e.g. 8"diam.) is needed to yield high counting efficiency, together with a picture of good definition. The image intensifier (I_1) uses a photocathode of this same area to convert each scintillation into a corresponding small pencil of electrons. The electron current of each such scintillation is to be enhanced by two image intensifiers, of which the second (I_2),produces the principal gain. For technical reasons intensifier I_2 is necessarily of small diameter (1"). One function of I_1, therefore, is to focus the electron ray pencils onto a 1" diameter output phosphor (P_2) whose scintillations, a replica of those on P_1, can be transmitted via the optical system (O) to the photocathode (P_2)of intensifier I_2. The light gain of intensifier I_1 is largely offset by the inefficiency of the lens coupling system, and intensifier I_2 must supply the major part of the gain which is sought (10^6, luminous intensity).

In an earlier design, described in Tokyo last year [1], the quantum gain of I_1 was supplied only by the acceleration of electrons between C_1 and P_2. This was found to give a total electron gain between the photocathodes C_1 and C_2 of less than unity, even with the best optics which we could devise. The present design, therefore, incorporates a multichannel plate (A) between C_1 and P_2 to provide electron multiplication, which should raise the gain 1000 times over the earlier tube. The tube I_2 in the previous design was of the Transmission Secondary Emission Type. This has proved to have poorer contrast than an alternative structure, the multistage type of intensifier which derives its electron gain from phosphor-photocathode sandwiches. In this type of intensifier, electrons leaving the photocathode C_2 of I_2 are accelerated towards an intermediate phosphor plate (D_1), whose reverse face is coated with a photocathode (Q_1). Each such phosphor-photocathode sandwich yields considerable electron gain, and several are cascaded within the envelope of I_2 to yield the final intensified image on output phosphor P_3. A single electron from the photocathode C_2 can, under favourable circumstances, yield a scintillation at the output phosphor P_3 of sufficient intensity to be photographed. A focus coil (F) allows the image on C_2 to be transferred to phosphor P_3 while retaining picture detail.

The simple system, as so far outlined, will not yield satisfactory pictures at the low isotope concentrations often used in clinical studies. Thermally generated electrons, for instance, to the number of perhaps 30,000/sec. leave the

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The use of radioactive tracers in medicine is of ever growing importance and correspondingly great effort has been devoted to the design of instruments (cameras), which will map the areas of radioactive concentration on which diagnosis depends.

The first part of this paper attempts to bring together the appropriate isotope technology, and clinical problems, which are then used to evaluate successive developments in camera-like devices. Certain criteria are thus established for the ideal camera.

In the second part, attention turns to a specific class of camera based on image intensifiers, with which our group has been particularly concerned. The remainder of this abstract confines itself to a particular camera, under development, which incorporates the results of our researches. We believe it will go far towards meeting the ideal criteria.

The principle of operation is shown in Fig.1. The object under study (S) emits gamma rays (R) which are collimated in the usual way and impinge on a phosphor (P_1). A simple pin-hole (B) in the shielding material (L) is shown as the collimator, and a sodium iodide crystal mosaic is a suitable phosphor. Scintillations of light, created near the point of impact of each gamma ray, form the signals which will finally become picture elements.

As is well known, a large area of phosphor (e.g. 8"diam.) is needed to yield high counting efficiency, together with a picture of good definition. The image intensifier (I_1) uses a photocathode of this same area to convert each scintillation into a corresponding small pencil of electrons. The electron current of each such scintillation is to be enhanced by two image intensifiers, of which the second (I_2),produces the principal gain. For technical reasons intensifier I_2 is necessarily of small diameter (1"). One function of I_1, therefore, is to focus the electron ray pencils onto a 1" diameter output phosphor (P_2) whose scintillations, a replica of those on P_1, can be transmitted via the optical system (O) to the photocathode (P_2)of intensifier I_2. The light gain of intensifier I_1 is largely offset by the inefficiency of the lens coupling system, and intensifier I_2 must supply the major part of the gain which is sought (10^6, luminous intensity).

In an earlier design, described in Tokyo last year [1], the quantum gain of I_1 was supplied only by the acceleration of electrons between C_1 and P_2. This was found to give a total electron gain between the photocathodes C_1 and C_2 of less than unity, even with the best optics which we could devise. The present design, therefore, incorporates a multichannel plate (A) between C_1 and P_2 to provide electron multiplication, which should raise the gain 1000 times over the earlier tube. The tube I_2 in the previous design was of the Transmission Secondary Emission Type. This has proved to have poorer contrast than an alternative structure, the multistage type of intensifier which derives its electron gain from phosphor-photocathode sandwiches. In this type of intensifier, electrons leaving the photocathode C_2 of I_2 are accelerated towards an intermediate phosphor plate (D_1), whose reverse face is coated with a photocathode (Q_1). Each such phosphor-photocathode sandwich yields considerable electron gain, and several are cascaded within the envelope of I_2 to yield the final intensified image on output phosphor P_3. A single electron from the photocathode C_2 can, under favourable circumstances, yield a scintillation at the output phosphor P_3 of sufficient intensity to be photographed. A focus coil (F) allows the image on C_2 to be transferred to phosphor P_3 while retaining picture detail.

The simple system, as so far outlined, will not yield satisfactory pictures at the low isotope concentrations often used in clinical studies. Thermally generated electrons, for instance, to the number of perhaps 30,000/sec. leave the

cathode of C_1: they result in random background scintillations at P_3, which fog the desired picture. Therefore, discrimination is introduced to eliminate low energy events such as these. When the discriminator is in use the intensifier I_2 is rendered inoperative (in the absence of picture signals) by bias on a gate electrode (G). The discriminator creates a pulse which overcomes the bias whenever a picture event is recognized. Thereby intensifier I_2 operates only for that short time (2 μsec.) required to pass the picture event.

Electron time of flight in the first image intensifier is far too short to permit signal identification processes. Discrimination against electron background is made possible only by virtue of delay introduced into the image intensifier chain by the decay time of the cascaded phosphors P_1 and P_2. The discriminator consists of a photomultiplier (M) which collects a little of the light scattered by the phosphor P_2, followed by a fast-acting amplifier and threshold sensing unit (E). The latter receives the photomultiplier output current, and, by testing its growth rate and amplitude, decides whether the signal is a useful picture element. If so, it switches on the image intensifier I_2 for an instant.

Some insight into the discriminator action is given by Fig.2. The relative ratio of charge released from the photocathode of C_1, due to a photopeak gamma quantum and a single electron, is in the order 100:1. Fig.2(a), (i) & (ii) shows the resultant (smoothed) photocathode currents of the photomultiplier (M). It is seen that the electron-generated waveform (ii) reflects only the simple decay of phosphor P_2 (type P.24), whereas the gamma event signal is convoluted by phosphors P_1 and P_2. These signals are not smooth waveforms as shown, but contain a strong granular structure arising from the statistics of the small electron population which creates them. The signals (i)-(ii) are now processed by an integrator followed by delay line differentiation, which performs the mathematical operation

$$i_{out} \propto \int_{t-\tau}^{t} i_{in}(t).dt$$

where τ is set to 50 nS by the delay line. The current delivered by this system now trips a tunnel diode discriminator, set to level (iii) of Fig,2, in order to reject thermal electron signals. The tunnel diode waveform is amplified and formed into waveform (b) which activates the image intensifier I_2. Very little of the signal energy is lost as a result of the small gating delays (shown in the waveform diagrams).

The system described in this paper has not yet been assembled, but its components will have essentially the physical appearance shown in Fig.3, which is an assembly of parts from the earlier camera. The picture shows the image intensifier arrangement, the shielding, phosphor and discriminator having been removed. Sodium iodide crystals of ¼" face by 2" long have been used experimentally for the phosphor mosaic, but the optimum dimensions depend on the gamma ray energy involved. Image intensifier I_1 (left) is a specially modified form of X-ray image intensifier, in which a channel amplifier has been added and the usual X-ray phosphor omitted. Image intensifier I_2 (contained outside the focus coil (F)) is an experimental device, developed by the 20th. Century Electronics Co. of England.

Our research was supported by the Ontario Cancer Treatment & Research Foundation; and a joint grant by the Foster Bequest, Allan F. S. Robertson Estate, and the James Franceschini Foundation. Special circuit researches were supported by grant A 1589 from the National Research Council of Canada.

(1) "High Speed Medical Gamma Ray Cameras", N. F. Moody & M. Joy, Digest of the 6th International Conference on Medical Electronics & Biological Engineering, Tokyo, 1965, pp 224 - 225.

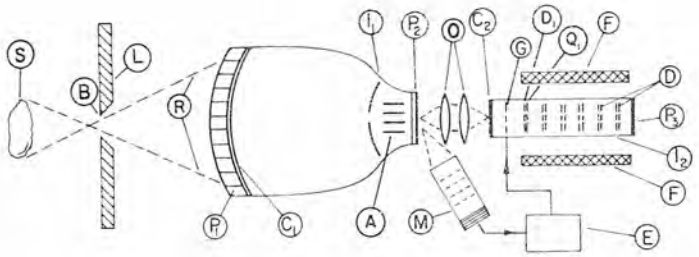

Fig. 1.
Schematic Diagram of Camera

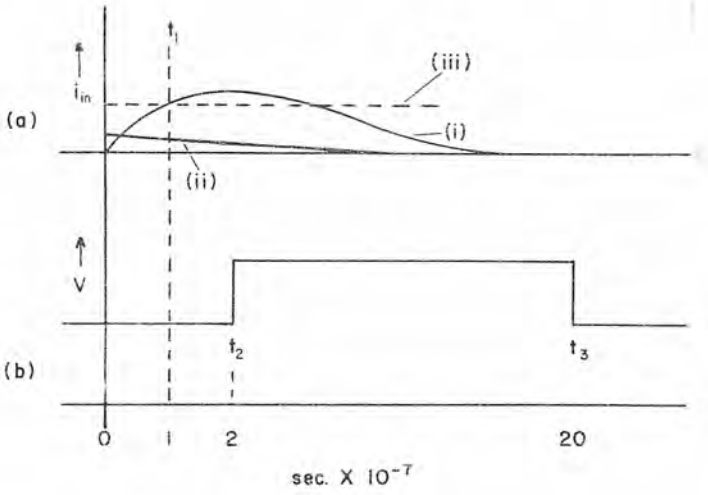

Fig. 2.
Discriminator Action Showing Gating Delay

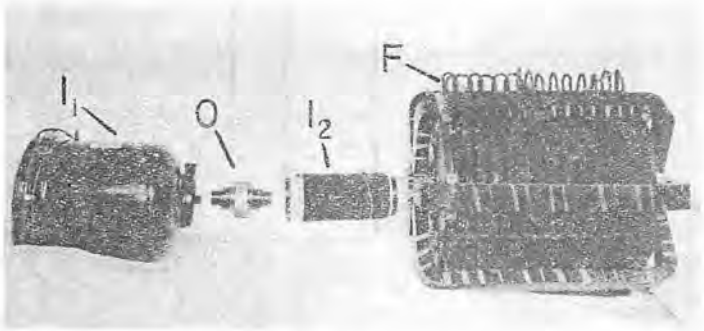

Fig. 3.
Exploded View of Camera
Showing Main Components

Fig. 1 : The new brain scanner in operation. The patient's head is immersed in water and is scanned by the rotating transducer. A true scale picture of a head section appears on the C.R.T. screen, which is then photographed.

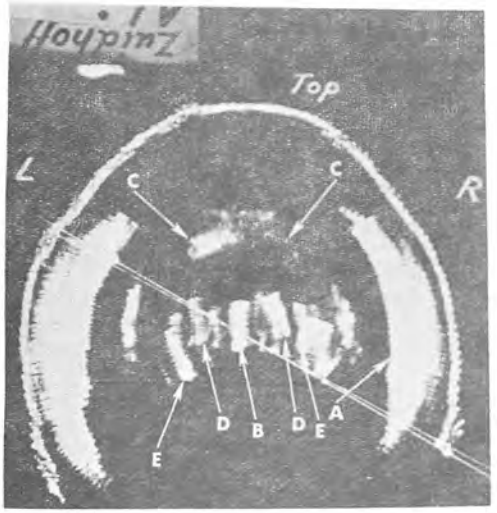

Fig. 2 : An example of a coronal scan of a normal 47 year old male. a) inner surface of the skull, b) third ventricle, c) roof of lateral ventricles, d) possible midbrain echos, e) trigones or temporal horns of lateral ventricles.

Fig. 5 : Scan of a patient with a large fronto-parieto-temporal bone defect under which there was no brain, only greatly distended ventricle. When scanned from the left, only a single image was seen, probably of a grossly shifted midline. When scanned from the right, a trail of echos was obtained which probably represented reverberations due to decreased attenuation caused by the absence of the bone in part of the left side of the skull.

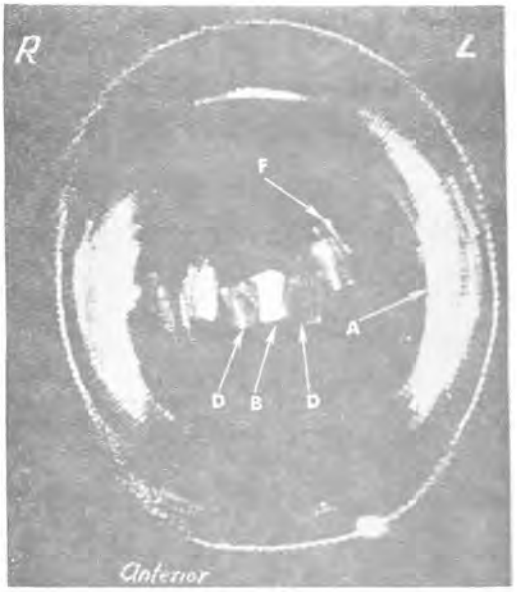

Fig. 3 : An example of a horizontal scan of a normal 47 year old male. a) inner surface of skull, b) third ventricle, d) possible midbrain echos, f) possible trigone of lateral ventricle.

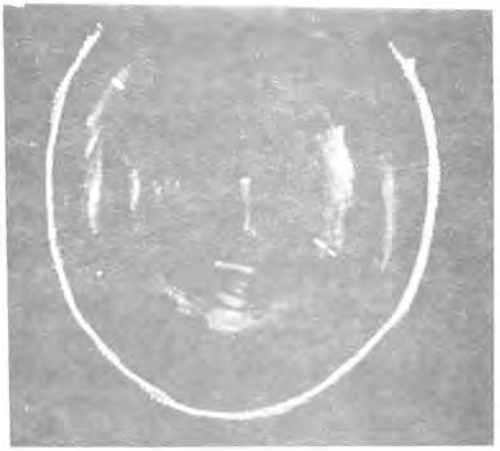

Fig. 4 : A shift of the midline echos to the right caused by a subdural hygroma on the left side.

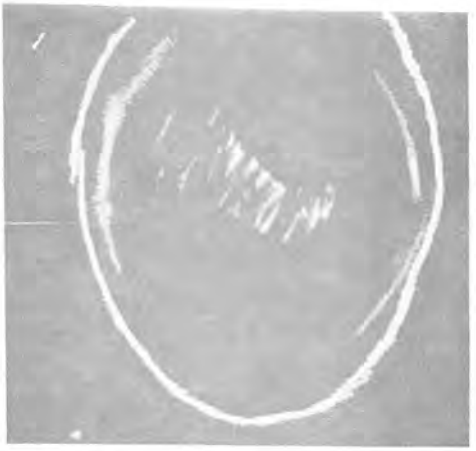

Fig. 5 :

BIOLOGICAL TELEMETRY

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The use of telemetry for the transmission of physiological data is an attempt to increase the understanding of various body functions while the subject is in a dynamic state, and preferably in his own familiar environment. The library of circuitry, transducers, and results obtained, is growing daily and it appears that every species known to man will eventually bear the burden of a transmitter.

We have been primarily interested in monitoring data from birds in flight. Our apparatus which will be described consists of a two channel transmitter which weighs 30 g (batteries included),operates at 230 Mc/s and has a range of 5 miles (Fig. 1). We are presently monitoring with the appropriate sub-carriers and transducers - temperature ecg, respiration rate and tidal volume, i.e. any two of these at a time.

All of this circuitry can be applied to human monitoring; howevever, we feel that a lot of grief can be eliminated by using the engineering know-how and technical skills of others in one of the more difficult aspects of the design, i.e. a stable transmitting section and receiver. This know-how is available in all discount stores in the form of hand-held transceivers as shown in Fig. 2. These transceivers operate in the citizen's band 27 Mc/s are crystal controlled, inexpensive, and do not require licensing. Crystals can be obtained from the manufacturer for operation in the industrial, scientific and medical band if one so desires.

Conversion of the transceivers to a system for telemetering ecg's is shown in Figs.3 & 4. In the transmitter, the speaker is replaced by a differential amplifier and subcarrier oscillator. The amplifier is a standard type of long tail pair with constant current feed and dc laddered to produce a gain of 500, an input impedance of 250 kilohms with a bandwidth from 0.1 - 200 cps. The amplifier output frequency modulates a sub-carrier oscillator. This is necessary in order to prevent detuning effects and signal strength variations from producing artifacts in the transmitted data, as well as allowing for multi-channel transmission. By choosing a subcarrier frequency within the audio range one can not only utilize the gain of the audio amplifiers in the receiver, but one can also listen to the modulated tone and hence the heart beat. This has found application where a patient is moved from one point in a hospital to another. In this case the porter carries the receiver and continuously listens to the patient's heart beat on an earpiece.

To look at the data transmitted, a demodulator at the receiver is necessary. This is shown in Fig. 4. It consists of a pulse shaping circuit and a pulse integrating or counting circuit. A signal at the input of the transmitter of 1 mV produces an output at the integrator of 50 mV, which is sufficient to be seen on most oscilloscopes, or fed directly into any electrocardiographic pen recorder.

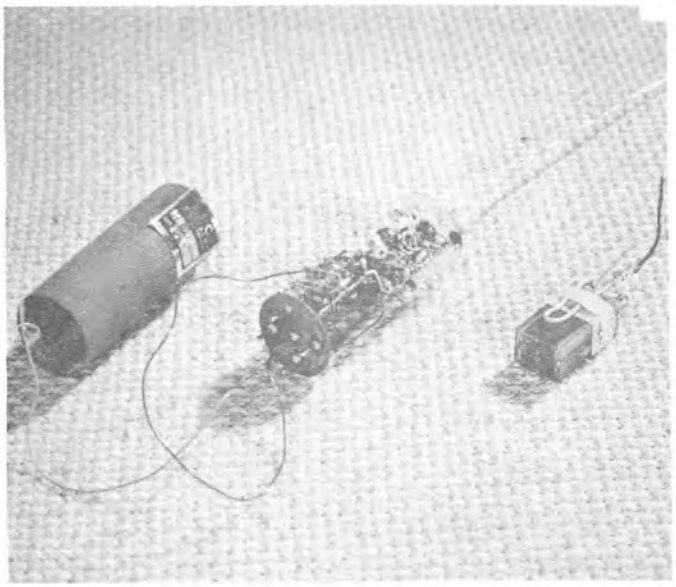

Fig. 1: Exploded view of bird transmitter;from left to right the magnesium case, the the transmitter with antenna and the 3 V battery pack.

Fig. 2: The "walkie talkie" ecg telemetry system. From left to right, transmitter, receiver and demodulator.

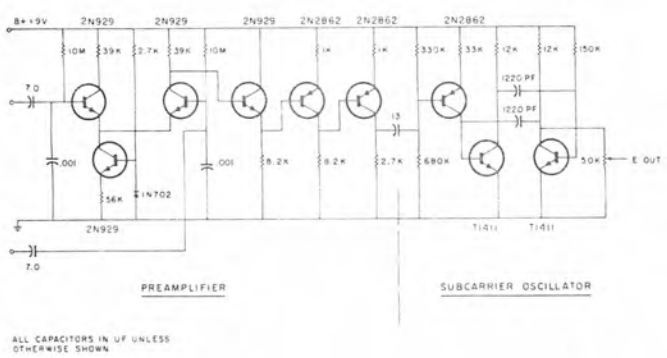

Fig. 3: Circuit diagram of ecg preamplifier and subcarrier oscillator.

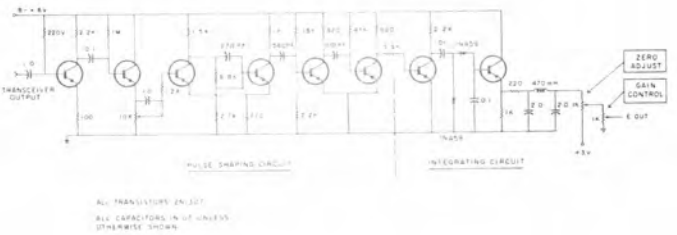

Fig. 4: Circuit diagram of the demodulator.

AN IMPROVED ULTRASONIC BRAIN SCANNER

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Ultrasonic pulses at a carrier frequency of 1Mc/s have been found to penetrate hair and the human skull. Reflected echos from certain intracranial structures such as the falx, lateral and third ventricles etc. can be received and positioned on a C.R.T. screen in a one to one geometrical relationship. The method is not unlike that used in radar, except that the position of the radiating source and moving target are interchanged and water is used as the transmitting medium.

Tumours and cerebral hemorrhages often destroy the symmetry of the brain, which can be shown on the image of a head section on the screen. This method, when perfected, may prove to be useful for diagnosis of brain disorders.

The design of the system is based on a compromise between the limited angular and depth resolution of the ultrasonic beam and the attenuation of ultrasonic energy. Further limitations are due to the heterogeneous composition and varying thickness of the skull bones (which refracts, diffracts and attenuates the beam) and the complex curvature of the ventricles and reverberation phenomena, particularly in the bone. The practical aspects of the design must provide for reasonable comfort for the patient during the examination and easy, speedy operation, preferably by non-professional personnel.

The new scanner (Fig. 1) incorporates several improvements in this direction over the system previously described (ref. 1, 2 and 3). The stretcher has been redesigned to permit a more secure and comfortable position of the patient over the water tank. The scanner consists of a stationary tank with a ring system which carries the transducer around the head one to three times. The transducer is raised or lowered to obtain scans at various horizontal planes. The rings can be tilted to obtain scans in the coronal and oblique planes. The increased speed of the scanner permits a superimposition of three scans to make up one picture, which may improve the reliability.

The display system provides a one to one picture of the head section. The latter is photographed on medical X-ray film to give a true scale photograph. A special focused transducer has an almost uniform beam width in the region occupied by the head. Together with improved receiver sensitivity, small echos can be displayed on the screen with greater ease than previously.

Pictures of the head sections obtained with the improved scanner on patients with brain disorders are expected to be available by the date of presentation of this paper. Preliminary results obtained on a normal person are shown in Figures 2 and 3. They represent coronal and horizontal scans obtained on a 47 year old male adult. Figures 4 and 5 show scans of abnormal patients that were obtained with the scanner previously described (ref. 2, 3).

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Illustrations next page

A METHOD OF MONITORING FETAL HEART SOUNDS

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WINNIPEG MANITOBA CANADA
Hargrave Applied Research Corp. Ltd.

Resume

A number of papers have been published on electronic techniques of fetal heart sound monitoring. The following describes an instrument which the author believes has a number of unique features and which has been tested extensively in several leading hospitals with very gratifying results. The instrument utilizes signal conditioning techniques which separate the fetal heart sound from maternal heart sounds, and in many cases allow the continuous monitoring of fetal heart rate throughout labor, fetal transfusions, etc.

The instrument has been designed with rechargeable nickel cadmium batteries in a small, portable, safe, low-cost package, useable on the operating table and in the ward by relatively untrained personnel. The instrument (Fetal Heart Monitor) is now in commercial production.

At the request of a Winnipeg Obstetrician, (Dr. M. Bruser), an investigation was started in 1963 to find a method of continuously monitoring fetal heart rates. A preliminary comparison of fetal ECG and fetal heart sound monitoring techniques showed the latter to be the most reliable method of following fetal heart rate. A major shortcoming of fetal ECG monitoring is the relative complexity of instrumentation, the aim being to produce a simple, lightweight, cheap, battery operated instrument. However, the ECG investigation was not abandoned, and has resulted in a somewhat more complicated instrument monitoring fetal heart rate of means of ECG signals. A future paper will describe this instrument which has advantages in certain circumstances. After the decision to concentrate on fetal heart sounds, a preliminary and very unsophisticated investigation of the sound energy at the surface of the mother's abdomen from the fetal heart seemed to show most energy to be at frequencies between about 20 and 200 c/s, the actual sound spectrum varying somewhat from patient to patient and with the position of fetus. The energy spectrum of maternal heart sounds at the mother's abdomen is very similar to that of fetal heart sound with perhaps somewhat less high frequency content. Thus frequency filtering helps very little in separating fetal from maternal heart sounds.

After considerable experimentation with various commercial piezo-electric transducers in contact with the mother's abdomen, we decided that this type of transducer was not suitable. Two major difficulties were frictional noise from contact with the mother's skin and the difficulty of devising a low noise, high input impedance field effect transistor input stage for the preamplifier, necessary for good low frequency response. 60 c/s power line interference was particularly troublesome with a high input impedance instrument. Instead, we decided to use a non-contacting dynamic microphone of our own design in conjunction with a current driven preamplifier. A short air column approximately 2mm long is used to transfer sound from the skin to the diaphragm of the microphone. The stiffness of the suspension of the coil, the length of the air column and the amount of damping are all adjusted to give a frequency response curve that is roughly centered over that of the energy-frequency curve of fetal heart sounds. Thus some preliminary frequency filtering is done right at the microphone. This preliminary filtering is most useful in that artifacts can often overload the following amplifier, making later filtering of little value. The use of a non contacting microphone avoids artifacts due to skin friction found with piezeoelectric or strain gauge transducers. The microphone is about 5cm diameter and about 2cm thick. It is affixed to the patients abdomen by means of "Stomoseal" adhesive pads. Alternatively three microphones are attached to an elastic belt strapped around the patient so that as the fetus moves, it can be followed by means of a switch on the instrument.

The output of the microphone is fed into an operational amplifier with a virtual ground input so that the microphone feeds into almost zero impedance. The gain of the amplifier is set so that a current pulse of peak value 0.2 microamps from the microphone, corresponding to a very weak fetal sound, gives an output pulse of 20 millivolts. The frequency response of the amplifier is centered at about 65 c/s by means of a band pass filter. The filter is adjusted empirically to give the best signal to noise ratio with -3db points at about 25 and 85 c/s (fig. 1). However, the frequency response of the microphone cannot be measured easily under working conditions so that the overall frequency response of the system is unknown. A simple bridged T, RC filter was tried to eliminate interference from 60 c/s power lines but in actual use we found that 60 c/s pickup was not a problem with a current driven amplifier.

If the signals from the microphone and amplifier are fed into a good audio amplifier, the resulting sound is a mixture of maternal and fetal heart sounds, and considerable low level noise from muscle movement, etc. The signal to noise ratio is considerably worsened by the fact that the human ear is very insensitive to low level sounds of low frequency so that even small amounts of high frequency or 60 c/s noise completely mask fetal sounds. An operator wearing earphones could usually pick out fetal rythms but we felt that earphones were too much of an encumbrance and a loudspeaker essential. We then tried

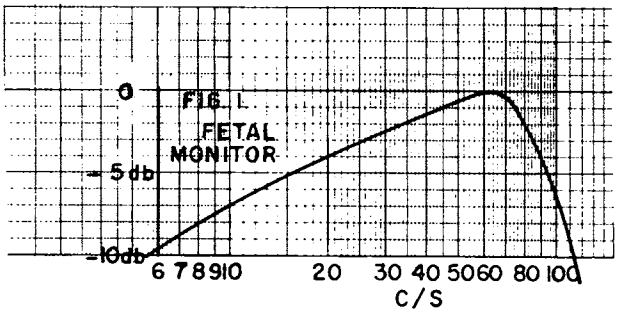
a discriminator which could be set by the operator to cut out all signals below a certain level.(fig. 2) Although this helped considerably, we were still troubled by the low audibility of the fetal sounds, by the presence of the maternal heart sounds, and also by acoustic feedback problems. Since the aim was to design a fetal heart rate monitor, we questioned a number of doctors as to whether the actual sound was necessary. Would a rate meter or flashing light do? The almost unanimous opinion was that a sound signal was mandatory. We tried three electronic sound sources modulated by the fetal and maternal signals; a 400 c/s tone frequency modulated, a 1 kc/s tone amplitude modulated, and a 1 kc/s beeper, all of which seemed acceptable under laboratory conditions but were too distracting and annoying in the operating and delivery rooms. We finally devised a simple pulse generator which gave a sound approximating to the true undistorted fetal heart sound, with high frequency components unfiltered so as to give maximum audibility. The pulse generator is triggered by the fetal and maternal signals. No adverse comments have been received about the artificiality of the sounds, in fact the instrument has sometimes been praised for the fidelity of reproduction. In order to remove maternal heart sounds, the difference in rates of the two was utilized. The maternal heart rate is unlikely to go above 120 pp/m, the fetal rate unlikely to go below 120 pp/m. The voltage on a capacitor C in fig. 2 is charged by the output of the pulse generator and switches off the gate for the discharge period of the capacitor. By setting the discharge period properly (approx. 1/10th sec) the pulse generator will tend to lock onto the faster of the two rates. Of course in the unlikely case that the maternal and fetal heart rates are almost equal, the circuit cannot discriminate, and the instrument will respond to both. Under normal conditions where the maternal rate is lower than the fetal rate, the fetal beats come through almost undisturbed except for an occasional dropped beat or extra beat of just after the fetal beat. In the presence of considerable motion artifacts such as during fetal transfusions, it takes a little practice to pick out the fetal beats, but in any case one is immediately aware of any sudden increase or decrease in rate. With signals below the triggering level, the pulse generator is designed to act as a linear amplifier so that by turning down the gain and using the recorder output, one can record phonocardiograms, though somewhat distorted by the filtering networks.

If the gain is turned up to trigger the pulse generator, the recorder trace shows a series of identical pulses, useful as a permanent record of fetal heart rate or the output can be fed to a simple rate meter.

This instrument has found particular use in the long term monitoring of patients during labor, delivery, and during fetal transfusions etc. Many hundreds of recordings have been made both on paper and on magnetic tape. A sample is shown in fig. 3.

The simplified circuit of the monitor is shown in fig. 2.

The complete instrument weighs approximately 2 lbs. and is useful for about 24 hours continuous operation without recharging. Recharging is accomplished by plugging the instrument into the 110V line overnight. The use of low voltage ensures safety in the operating theatre and complete portability. The monitor is small enough to be used while wheeling a patient on a stretcher or in an ambulance. A meter coupled to the pulse generator provides visual indication of fetal heart beats when the sound level is turned to zero. Output jacks are provided for remote speakers and for pen or tape recorders.



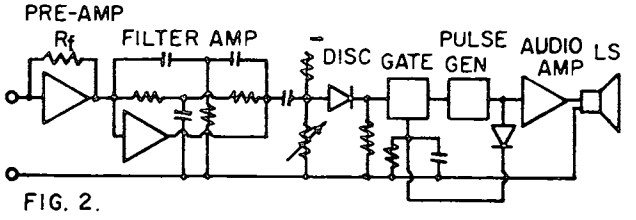
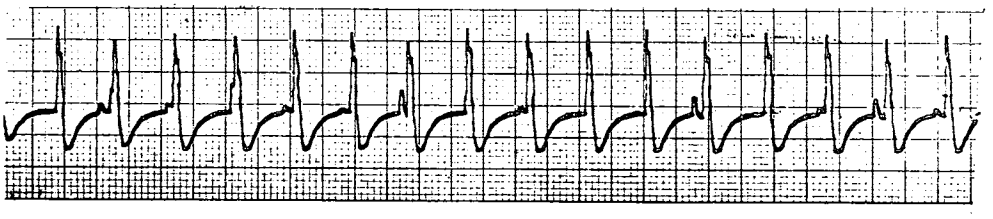


FIG. 2.



11:45 a.m. 170 ml. OF BLOOD HAVE BEEN INJECTED INTO PERITONEAL CAVITY OF FETUS. NEEDLE & CATHETER HAVE BEEN REMOVED. RATE ON SOUND TAPE 160.

FIG. 3

THE INTENSIVE CARE MONITORING SYSTEM OF THE MONTREAL HEART INSTITUTE

GLENFIELD WARNER

THE MONTREAL HEART INSTITUTE

Coronary Care Monitoring Equipment has one principal use at present, - to save the lives of patients who might suddenly suffer a fatal arrhythmia - i.e. cardiac standstill or ventricular fibrillation. Immediate treatment is mandatory in these cases. Only about 3 minutes is available to restore blood circulation. A cardiac monitor which measures the heart rate from the patient's electrocardiogram and which sounds an alarm when the heart rate falls below or rises above certain prefixed limits has proved extremely useful in alerting medical staff to the presence of these fatal arrhythmias in patients quickly enough to save their lives.

The number of cardiac monitors now on the market is quite large, but they have shown up one glaring deficiency - the lack of suitable performance standards for these instruments. On a number of monitors tested in our laboratory the common mode rejection ratio at 60 cps and with no unbalance in the differential input was found to vary from 80 db down to 53 db. Amplifier input time constants for monitoring varied from 250 m sec. to 1.5 seconds. The number and placement of controls also showed considerable variation, e.g. no horizontal position control and the horizontal width control inside the chassis. Adjustments like these, while they should not appear on the front panel should still be easily made from the rear. It should be noted that cardiac monitors are essentially nurse operated. This is why only the minimum number of controls essential for operating them should be on the front panel.

It would be desirable to make a number of laboratory measurements on a monitor and from these get a reasonable idea of what its performance might be on a patient. Standard measurements like the Common Mode Rejection Ratio, Sensitivity, Ratemeter integrator time constant etc. were useful, but it was found that perhaps the two most significant measurements were the input time constant and the ratemeter counting circuit bandwith.

The usefulness of a cardiac monitor as an alarm device falls to almost zero when it gives too many false alarms. When false alarms occur too often the alarm circuits are simply turned off by the nurses. Anyone who has seen a nurse come on the run to a cardiac patient whose monitor has sounded an alarm will understand why they do this.

We have found that monitors whose amplifiers have long input time constants can become blocked long enough by large artifact signals for a low rate alarm to be given.

Much of the movement artifacts and muscle noise picked up by the electrodes are triangular wave-like in form. The time rate of change of these voltages is important in determining if the signal will produce a count on the ratemeter. Ideally it would be preferable that the cardiotachometer respond only to voltage rates of change near that of a QRS complex of about 1 millivolt amplitude. In this way it would have minimum susceptibility to noise. A triangular wave of of fixed amplitude and variable frequency could be used to check the counting circuit response.

METHOD OF MEASUREMENT

We used a sine wave generator to check the ratemeter response, as a triangular wave generator was not available. It was found that the wider the bandwith around 9 or 10 Hertz, the more the high rate alarm could be triggered by noise. In the Intensive Care Ward it was found that these monitors with larger ratemeter bandwiths (0.35 Hertz-200 Hertz) produced more false alarms than those with small ratemeter bandwith (4 Hertz -42 Hertz). We are now investigating two different ways of preventing the ratemeter from counting noise.

Some cardiac monitors will count both positive and negative going ECG signals, others will count only one polarity. The single polarity monitors are clearly superior to the others in their ability to handle abnormal ECG signals without false counting (in this case counting more than one beat per heart cycle).

Following are a few specifications which we think a good monitor should incorporate:

(SEPARATE INPUT FOR RATEMETER)

MECHANICAL

Size: Should not exceed 15" X 9" X 16"

Weight: Should not exceed 30 lbs.

Construction: Should be modular with carrier chassis containing all necessary interconnections. Modules should be operable in any position in carrier chassis.

Controls: Minimum controls for operation on front panel. Maintenance controls easily accessible from rear. Only factory pre-set controls should be inside chassis.

ELECTRICAL

Common Mode Rejection Ratio: 80 db minimum at 60 Hertz, maximum 5 K ohms unbalance in input.

Sensitivity: Ratemeter should count with 100 micro-volt input signal reliably.

Bandwith: Scope - 0-100 Hertz

Ratemeter 2 mv sine wave input should not produce signal below 4 Hertz.

50 mv sine wave input should not produce signal above 45 Hertz.

Amplifiers: Scope and Ratemeter should have separate amplifiers, separately controlled. Circuits solid-stste (silicon) and protected against voltage overloads. (Defibrillator etc.)

Noise: With input short-circuited internal noise should not exceed 10 microvolts peak to peak.

Gating: Circuit should be unipolar with polarity switch.

Electrodes: Cup type - Ag-AgCl.

The Coronary Care Wards of about a dozen hospitals in a number of American cities were inspected during the past year and and half. Here again a good deal of variation was found. Each hospital had its own way of doing things depending on which manufacturer's equipment it had and the views of the medical staff. Most hospitals had a bedside monitor beside each patient, with varying amounts of equipment at the Nurses Station.

The Montreal Heart Institute will generally monitor three different class of patients.
1) Surgical patients - i.e. those who have had operations. These patients are confined to bed for most of their stay in the Surgical Intensive Care Ward. After 2 or 3 days they are fairly safe from sudden circulatory collapse. Each patient has a bedside monitor which connects to a central station where there will be one cardioscope ECG display per patient. Visual and aural alarms will signal alarm conditions and indicate which bed is involved. Heated stylus recorders will be provided which will be operated either manually or automatically. Arterial systolic pressure will also be displayed.here.

2) Medical patients - Coronary patients admitted with myocardial infarctions. The major incidence of fatalities in these patients occur in the first 72 hours. they will therefore be carefully monitored during this period. Facilities for these patients will consist of 4 or more beds. Each bed will have a bedside monitor connecting to the Nurses Desk. Here there will be an ECG display for each patient, alarms and an automatic ECG recorder. In some cases when the patient is permitted out of bed, he will be fitted with an ECG telemetry transmitter. These ECG telemetry systems have proven to be very useful in patient care. Facilities for monitoring blood pressure by cannulation will be provided.

3) Medical patients - non-coronary patients with external pacemakers, other kinds of heart trouble, children etc. These patients do not need to be kept as quiet or be as closely observed.

Each patient floor is provided with a mobile, completely equipped heart resuscitating station consisting of a DC defibrillator, a pacemaker, a cardioscope and a recorder mounted on a cart which also carries all of the necessary accessory equipment. Another cart carries the drugs and medical-surgical equipment. One electrical outlet per bed on the patient floors and all outlets in the surgical intensive care ward are connected to the hospital emergency electrical power system. All patient rooms are connected by conduit to the Nursing Station.

The investigation of cardiac monitors and planning for the Montreal Heart Institute Intensive Care facilities show that there is much room for progress in the patient care field. Cardiac monitors based on the ECG , while effective are not 100% reliable. It is possible for a patient to suffer circulatory collapse while his ECG will be good enough to produce a count rate within the specified limits on the monitor. Some signal more closely related to blood flow would be preferable as an indicator of a patient's condition. Earpulse and finger plythysmograph monitors indicate only peripheral circulatory conditions. The present generation of cardiac monitors for the most part indicate that something has happened (cardiac arrest, etc.) The next generation must indicate that something may happen. The monitor must take into account the trends of various physiological variables and decide whether to sound a warning or not. This means, if not a true computerized operation, at least some logical and memory functions in the monitor. We should, however, remember one thing - the best monitoring system is no substitute for a good, warm-hearted nurse - only a helpmate.

ASPECTS OF ELECTROCARDIOLOGY
Looking at Arrhythmias

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The evolution of various techniques for long-term monitoring of the electrocardiogram have resulted in an increased awareness of the high incidence of arrhythmias which occur in patients following cardiac infarcts. Kurland and Pressman[1] cited an incidence of 94% in patients monitored continuously for one to three weeks. Sixty-nine percent of their patients exhibited more than one arrhythmia. In a survey of over 300 cases, deaths following acute myocardial infarction increased up to 700% when arrhythmias occurred.

This report outlines some principles involved in studying variations in the heart beat and reviews a few procedures which have been applied for detection of specific arrhythmias, or for scanning long intervals of recorded electrocardiograms.

Most heart monitors are sensitive only to changes in rate, responding to extended periods of tachycardia, bradycardia or stand-still. Thus they only detect abnormalities associated with varying R-R intervals. In general their time constants are too long to detect random phenomena of short duration.

It has been observed that a sequence of premature beats may lead to ventricular tachy-cardia or to fibrillation. Warning of a short sequency of irregular beats could permit corrective measures before arrest ensues.

Since an ectopic beat originates at an irritable focus outside of the normal conduction system, its conduction velocity is reduced and the recorded R-wave generally is of extended duration - 0.12 second or longer. Also there is no signal cancellation from the opposite side of the heart, and the amplitude therefore may be increased. These characteristics may be exploited in differentiating an ectopic R-wave from the normal.

The block diagram of an experimental ectopic beat detector is illustrated in Fig.1, and the circuit of a pulse-width discriminator is shown in Fig. 2. Each R-wave triggers a multivibrator to turn off Q_1 and produce a -12 V pulse at the collector of Q_2. This pulse is isolated through Q_3 and is used to turn off Q_4. The input signal is also fed to Q_4. If its pulse duration exceeds 0.12 second, the differ-entiated tail will pass Q_4 and trigger Q_5, producing a negative pulse at the collector of Q_6. This pulse applies a small voltage step to an integrator circuit. A pre-determined number of recurring ectopic beats will turn on Q_8 and actuate a relay to switch off a tape recorder, actuate a pen writer, sound an alarm, etc.

Ectopic beats may derive from multiple foci on the heart, and may thus produce R-waves of differing polarity. It is necessary therefore in designing an ectopic beat monitor to incorpor-ate a pulse inverter, which must preserve the integrity of the R-wave amplitude and duration. The circuit will then detect all ectopic beats which follow the classic criteria. Unfortunately the premature beats will not predominate in all leads and some will inevitably be missed. The R-wave may indeed be smaller than the T-wave in some instances.

From these studies it appears that neither R-R interval nor pulse width/amplitude monitoring can detect all arrhythmias of the heart. Interval timing is probably the more versatile technique.

Scanning techniques for viewing the continuously-monitored electrocardiogram date back to the Smith and Stone cardioscope developed in 1949. This instrument provided vertical dis-placement of recurrent sweeps on a long-persistent screen to allow comparison of the configurations of several heart beats. The sweep was free-running, without facility for synchronizing repetitive R-waves.

In 1962, Webb[2] reported his "contourograph" method of displaying quasi-periodic physiological signals. For ECG presentation, alternate R-waves are used to sync the sweep. A vertical deflection which is linear with time is added to the vertical ECG signal to displace recurring sweeps downward. A third dimension illusion is added by intensity-modulation of the vertical signal. The result, when viewed on a long-persistence screen, is a raster which shows very clearly any rate-dependent arrhythmia.

Another system of scanning long-term ECG records was introduced by Holter[3] in 1961. The electrocardiogram is recorded on two channels at a slow tape speed for periods up to ten hours. One of these channels is advanced on the tape to provide a trigger for synchronizing the playback CRT display from each R-wave. The ECG signal can be displayed on the screen at 60 X speed or recorded on a pen writer for real-time playback. The arrhythmiagraph feature allows very fast scanning of a long record, or detailed inspection of specific episodes.

References

1. Circulation, Vol.XXXI, 834 (June 1965)
2. Proc. 15th Ann.Conf.Eng. in Med. (1962)
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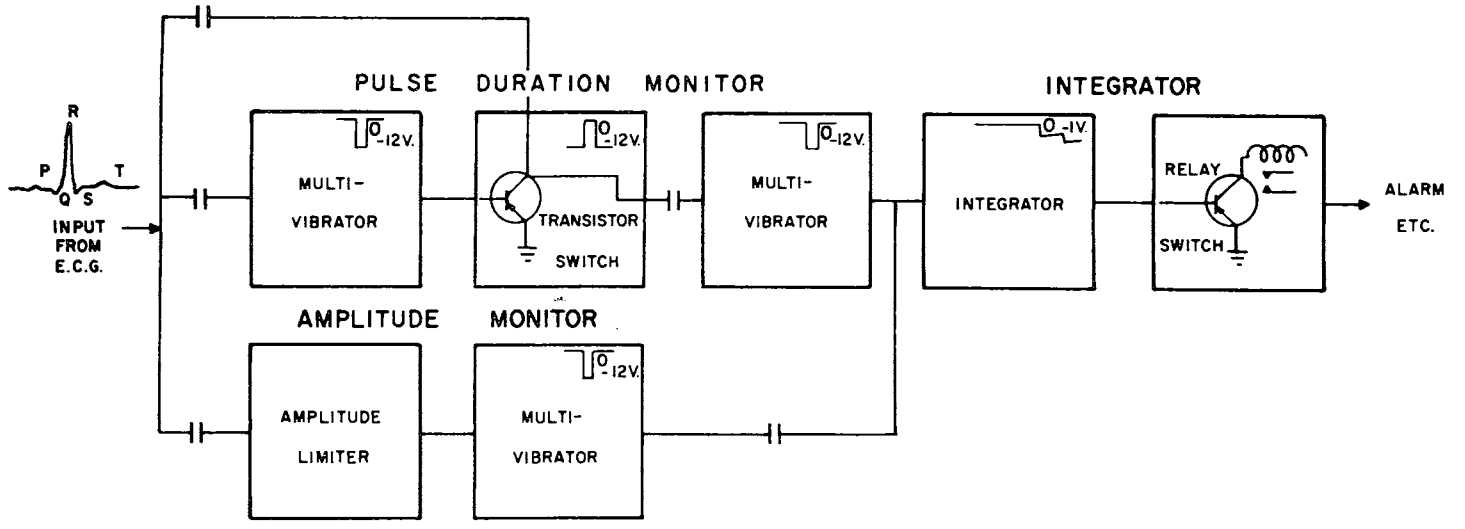

Fig. 1: Block diagram of a pulse amplitude/duration arrhythmia detector

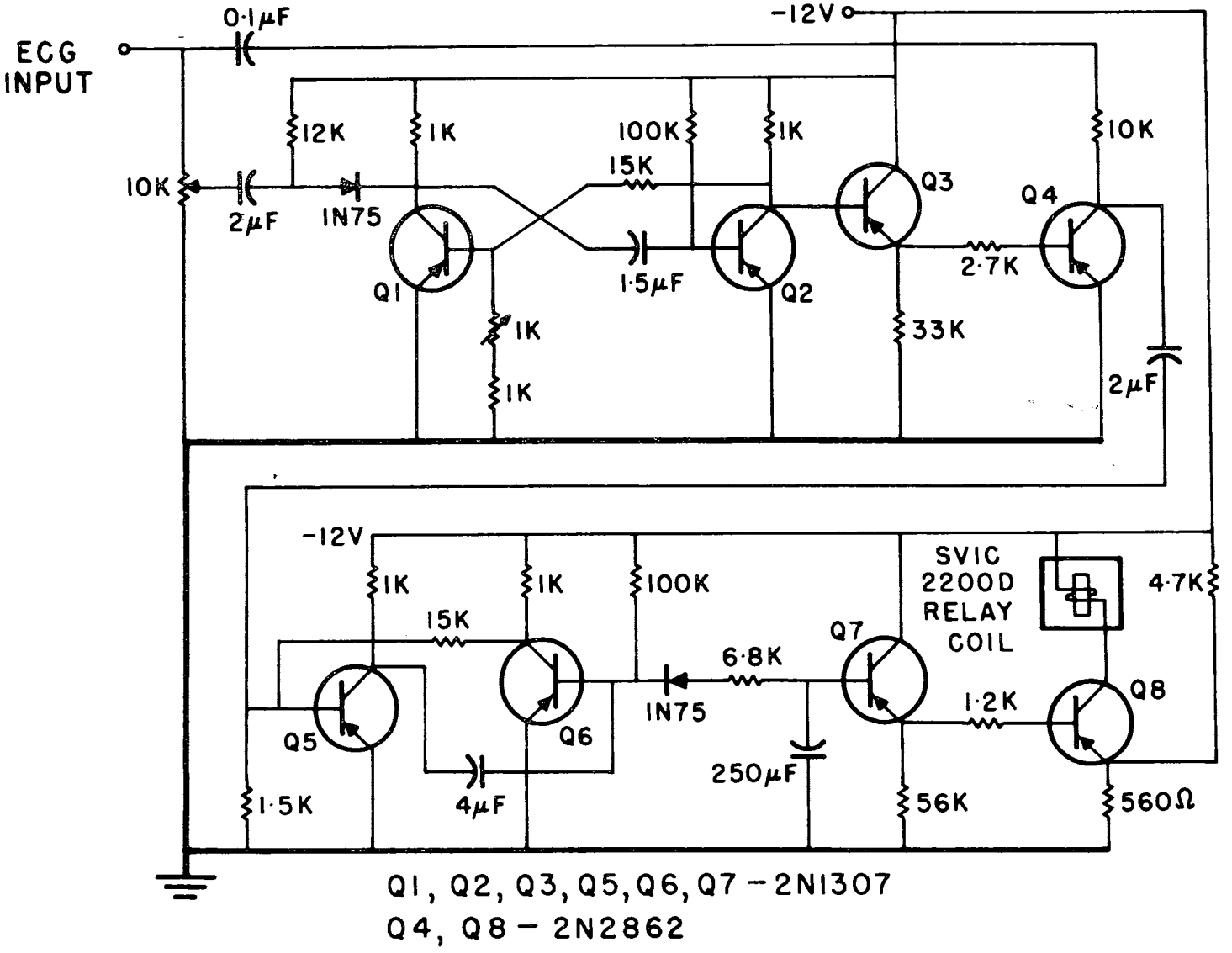

Fig. 2: Circuit drawing of a pulse duration arrhythmia detector. (Explanation in text)

ESTIMATION OF CANCELLATION EFFECTS DURING VENTRICULAR DEPOLARIZATION

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Electrocardiography has long been established as a most useful clinical diagnostic tool. The interpretation of electrocardiograms is still highly empirical in spite of considerable research since the turn of the century. The basic problem is to relate, in a meaningful way, the electrical information derived at the surface of the body to the wave fronts occuring in the heart during depolarization and repolarization. It has not been possible to establish this relationship, because an infinite variety of generators can give rise to the same type of surface information.

The actual cardiac generator consists of a convoluted lamina (a dipole layer) that is constantly changing in size, shape, and orientation from moment to moment. All portions of this lamina contribute to the electrical field. If portions of the lamina are oppositely directed in space, they contribute potential fields that are oppositely directed and hence tend to cancel out each other. At the surface of the body, we are dealing with only a portion of the electrical field actually generated by all portions of the heart. To facilitate understanding and mathematical treatment, it has been assumed that this electrical field is generated by a point source - "the equivalent cardiac dipole".

The left ventricle, including the interventricular septum, constitutes the major anatomical and electrical generating portion of the heart. In the normal heart, the wave of depolarization moves for the most part from endocardium to epicardium. A point, in the cavity of the left ventricle, is, therefore, unique in the body in that it faces the negative side of most dipoles during depolarization of the ventricles. At this point, there will be a summation of fields contributed by most portions of the ventricle, and this point therefore should reflect the total potential generated during ventricular depolarization.

The potential outside the heart, or, the "total outward potential", developed by the equivalent cardiac dipole can be determined from the orthogonal components which are derived from suitable surface leads.

Thus, an exploring electrode within the cavity of the left ventricle approximates the total potential generated during depolarization by the left ventricle, while the total outward potential or what is left over after cancellation of oppositely directed fields, can be derived from a corrected orthogonal lead system with appropriate calculations by an analogue computer.

The difference between these two potentials (the left ventricular intracavity and the spatial magnitude curve) is due to (1) the difference in proximity to the generator (i.e. the wall of the left ventricle - the intracavity lead being close, while the surface leads are more remote), (2) cancellation effects as noted above, (3) differences in conductivity of intracavity blood and extra cardiac tissues.

If we correct for proximity effects and differences in conductivity, the difference between the intracavity potential and the total outward potential is related entirely to cancellation of oppositely directed forces.

If at a particular instant of depolarization the cardiac generator approximates a plane, then the difference between the intracavity lead and the surface spatial magnitude curve (the total outward potential) is entirely related to differences in proximity from the generator and conductivity as noted above. It is likely that at the very beginning or at the very end of ventricular depolarization the generator closely approximates a plane. If we therefore subtract the surface spatial magnitude curve from the intracavity curve, gradually increasing the gain on the surface magnitude curve amplifier until we get a slight negative deflection on their difference curve, we will then have corrected for differences in proximity and conductivity. At that point, the difference curve will be an indication of the magnitude of the potential field lost through cancellation effects. This difference curve is also an indication of the multipolar nature of the cardiac generator.

To study the above theory, experiments were carried out using mongrel dogs. An exploring electrode was passed via the brachiocephalic artery to the root of the aorta and then across the aortic valve into the cavity of the left ventricle. The position was confirmed by fluoroscopic control and pressure recordings. This exploring electrode was used with Wilson's central terminal as a bipolar lead. The McFee lead system suitable for dogs was employed to record the corrected orthogonal surface leads. The appropriate calculations giving the spatial magnitude curve were done with an analogue computer. The X Y Z leads, spatial magnitude curve, and the intracavity lead were recorded on multichannel equipment (Electronics for Medicine). The intracavity left ventricular lead was then recorded on a single channel to permit maximum fidelity. The spatial magnitude curve was then recorded on a single channel and the difference in the size of the two curves was noted. The spatial magnitude curve was then subtracted from the intracavity curve and the gain slowly advanced on the spatial magnitude curve until a slight negative deflection occurred in the difference curve. At this point, the resulting curve was recorded on a single channel. This latter curve was considered to represent the amount of potential lost to the surface leads through cancellation effects. Its size indicated that the loss was considerable. This curve was also an indication of the multipolar nature of the cardiac generator.

SUMMARY

1. The theoretical basis for the existence of cancellation effects during ventricular depolarization has been reviewed.
2. A technique for the measurement of cancellation effects has been developed.
3. The instantaneous magnitudes of the potential field lost through cancellation effects throughout ventricular depolarization has been recorded in dogs.
4. It has been shown that a large portion of the electrical signal related to ventricular depolarization is not available at the body surface.
5. The presence of cancellation effects is an important reason for the present empirical nature of the interpretation of electrocardiograms.

FEASIBILITY OF ANALOG COMPUTER TECHNIQUES FOR DIAGNOSTIC ANALYSIS IN ELECTROCARDIOGRAPHY

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Extraction of diagnostic information from an exercise ECG requires advanced statistical techniques not only for diagnostic analysis but also for noise and data reduction. We have previously shown that fairly advanced statistical techniques can be implemented with the aid of an analog computer (Proceedings of the 1965 Atlantic Region Medical Engineering Symposium). The method proposed for on-line analysis of exercise electrocardiograms and vectorcardiograms has been developed subsequently to a stage where all manual adjustments previously required are controlled automatically. The selection of the baseline reset point, however, still occasionally requires manual adjustment, and visual monitoring for a reliable performance.

The program involves the following sequence of operations:

(a) Resolving.

The orthogonal X, Y and Z leads (Frank lead system) are resolved in the direction of the maximal R vector ($\vec{R}$) by solving for each cardiac period the transformation equation:

$$R(t) = \frac{\vec{R} \cdot \vec{V}(t)}{|R|}$$

where $\vec{V}(t) = X(t)\vec{X} + Y(t)\vec{Y} + Z(t)\vec{Z}$.

Alternatively, any other direction can be selected for resolving, like the direction of the maximum ST or T vector. See Fig. 1 and Fig.3.

(b) Sampling.

The scalar ECG is sampled at two points of the segment extending from the ST-T junction (J point) to the peak of the T wave. The length of the J-T segment (I) can be approximated as a linear function of the heart rate (HR). The duration of the J-T segment is normalised to unity in time and samples are taken at the 1/2 and the 3/4 point of the J-T segment. Any other pair of points can be selected in the present program.

(c) Smoothing.

Extensive noise in the exercise records makes smoothing of the samples taken from successive heart beats absolutely necessary. Exponentially mapped past is used for smoothing in the present program. Each sampled amplitude V_i is replaced by a smoothed value V_i^* according to the relation:

$$V_i^* = \alpha V_i + (1-\alpha)\, V_{i-1}^*$$

(d) T-Square analysis.

The sample values are then fed into a statistical program which weights them according to their position along the J-T segment. The coefficients involved are computed from the elements of the inverse covariance matrix of the amplitudes sampled from the ECGs of a group of 60 normal male subjects aged from 40 to 60 years. This template was then used to calculate the T-Square values of a test group consisting of 88 normal and 27 clinically abnormal subjects. The abnormal subjects were selected on the basis of clinical examination but without any knowledge about their ECG findings. A four-minute exercise test was carried out with the bicycle ergometer with a load of 600 kpm. The block diagram for the diagnostic program is shown in Fig. 2.

Results.

A screening value of T-Square corresponding to 80% specificity in the normal test group yielded a detection of 50% of the abnormal subjects. The analog computer program used here with only 2 sampling points produced the same sensitivity as the conventional post-exercise electrocardiogram analysed independently in the same group. A more elaborate signal representation with aid of 6 Chebyshev coefficients improves the analysis to 60%. Signal representation with a limited number of Chebyshev coefficients and subsequent T-Square analysis is entirely feasible with analog computer techniques, and this modification can be expected to improve the diagnostic power of the method. Fig. 4 shows a typical ST segment approximated by a varying number of Chebyshev coefficients.

The type of system used for ECG analysis depends largely on the facilities available to the investigator. One important consideration is that a fast digital computer can perform and even simulate practically any analog operation. Our present experience seems to suggest that the logical order of proceeding in developing a technique for the ECG analysis is to use initially a large scale general utility digital computer, in order to find the optimal method for the extraction of diagnostic information. Once the most suitable method has been found a special purpose analog system can be set up and is the method of choice in some circumstances, provided that qualified engineering assistance is available for design, maintenance and quality control. The versatility of analog systems is much improved if supplemented with a memory. A hybrid system may well be the ideal solution to this kind of analysis, if equipped with fast noise-free electronic switching for sampling and recirculating delay lines as storage media for smoothing by averaging.

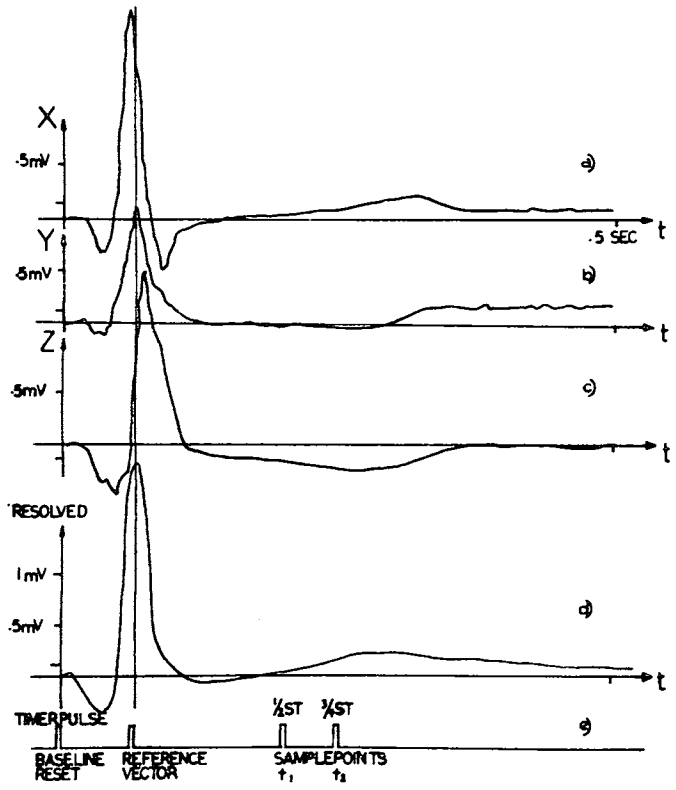

Fig. 1. The ECG response to exercise is analysed from a single scalar lead (d), obtained by resolving the orthogonal X, Y, Z components (a, b, and c) in the direction of the maximum QRS vector (occurring at the time indicated by the line). The duration of the J-T segment is predicted in a linear fashion as a function of the heart rate and normalised to unity. The samples are taken at time points indicated by the trigger pulses in (e).

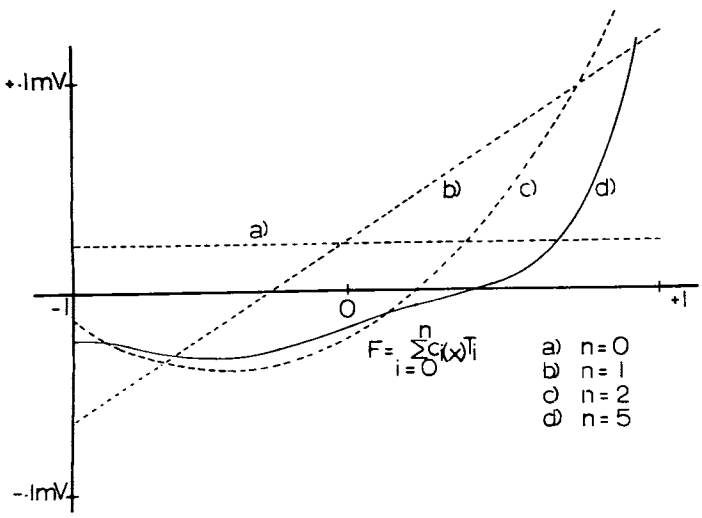

Fig. 4. A typical S-T segment can be accurately represented with aid of Chebyshev approximation using six terms (d). a, b, c show the improvement in fit when an increasing number of terms is used.

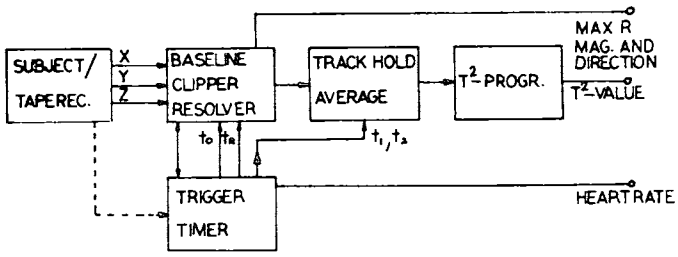

Fig. 2. Block diagram of the analog computer-program for on-line analysis of exercise ECG. The heart rate, its variance and the magnitude of the maximal R wave are also computed and displayed together with the T-Square value on a storage oscilloscope, a chart recorder or a digital voltmeter.

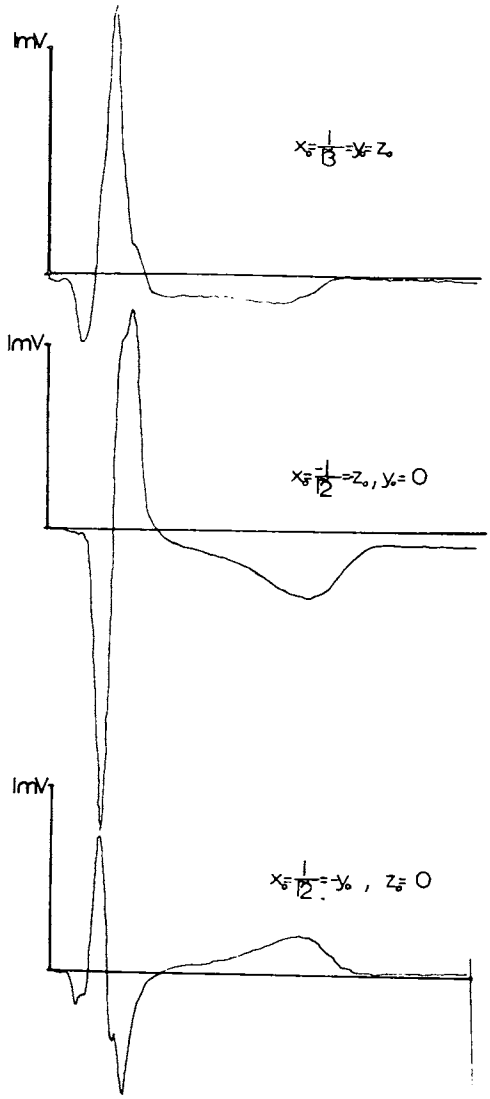

Fig. 3. The pattern of the ECG as a whole, as well as the shape of the S-T segment, depends to a great extent on the orientation of the "leads" used for recording. This is illustrated by resolving a normal VCG in various directions specified by unit vectors. Conventional criteria based on "J-amplitude" and slope of ST segment are of limited value unless spatial orientation is taken into consideration.

QUALITY CONTROL IN RECORDING OF EXERCISE ELECTROCARDIOGRAMS

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There has been increased interest in exercise electrocardiography. The full extent of the diagnostic information present in these exercise records has not yet been determined, partially because of the difficulty in obtaining records of satisfactory quality, especially during submaximal exercise stress conditions. The need for improved procedures in data acquisition has been stressed by several investigators and by the Research Committee of the International Society of Cardiology in April 1965. To a large extent, the quality of the signal depends on logistic factors such as electrode and skin preparation, and the choice of electrode, lead system and exercise test. What is required is some quantitative measures and criteria for judging the quality of these records.

FACTORS AFFECTING QUALITY OF ECG RECORD

The criteria chosen for quantizing the quality of any overall system are (i) electrode/skin junction impedance, (ii) bioelectric and electrode junction dc potentials, (iii) hum and (iv) noise. The first three are relatively easy to determine, but the effect and cause of the latter are far more difficult to evaluate.

Electrode impedance, polarization voltage, hum.

Electrode impedances were measured at 20 Hz; at this frequency the polarization and phase shift are negligible, and standard ac instruments may be used. The dc polarization potentials of each electrode pair (each to the common ground) were recorded during the entire exercise run. 60 Hz hum was discriminated down to the 1.5 μV level. This latter measurement required a line voltage reference signal to be recorded on a spare channel at the time of the exercise test; a cross correlation of this reference signal with each ECG signal gave the accurate measure of the average hum.

Noise.

Noise, in ECG exercise records, can approach the magnitude of the signal itself and can seriously interfere with the pattern recognition schemes, either visual or computerized. ECG "noise" can be defined as any unwanted potential, whether random or not, that adds to the signal. It is generated mainly by the electrode and skin movement, and myoelectric signals directly related to the stress test, and from other physiological sources such as respiration.

Determination of Noise Characteristics.

By sampling the signal at the same point in the ECG complex (in our case, in the S-T segment about 100 ms after an R wave trigger) we obtain a sample ensemble of the slowly changing signal plus additive noise. A description of the noise component was obtained from an autocorrelation of these samples. From the prominent exponential nature of the autocorrelation function the random component of the noise was seen to be Markoffian with a half power density frequency varying from .15 c/s to 2.3 c/s. In normal exercise records, about 85% of the total noise power was contained in this random component. However, strong periodic components were produced by respiration and the periodic physical activity. These effects were emphasized in special tests in which the subject was forced to breath in synchronization with his heart rate, whereby the subject inspired during 3 successive heart beats, and expired during the next 3 beats. Figure 1 shows the autocorrelation of the samples taken during such a resting run, and a very dominant periodic component of a period corresponding to six heart beats is evident. In addition, the same subject was required to bicycle in synchronization with his heart rate (one revolution of the ergometer for each two heart beats). Again, as indicated in Fig. 2, a strong periodicity is seen, this time with a period equal to the two heart beat intervals (and a lesser periodic component of 6 heart beats). Drift components were also shown to be present in a few of the correlations, but their magnitude remained small mainly due to the ac coupling of the recording amplifying system.

Based on the four criteria described above, a pilot study was carried out to compare three commercially available electrodes, and four bipolar lead systems. The final choice would also take into account the characteristics of the recording system, and possibly the signal processing and pattern recognition systems to be employed.

Precautions in Statistical Filtering.

Common techniques of noise reduction are exemplified by average transient computing, which can be usefully employed only in the reduction of random components. For example, 60 Hz hum cannot be reduced as random noise because the hum signal affects the triggering point (usually the R wave) required for such averaging methods. The time relationship between the cardiac pacemaker and hum is essentially random. However, partial correlation is produced when the ECG trigger is perturbed by a periodic interference such as hum. The same can be said for baseline periodicity such as that due to bicycling. This partial synchronization of the trigger point will not only reduce the effect of noise reduction, but can permanently distort the averaged ECG signal. A special situation also exists with regard to the respiratory component,

which would normally disappear if there were no correlation between heart rate and respiration. However, many investigators have noted the frequency modulation effect of respiration on heart rate. This correlation would mean that baseline shifts due to respiration could never be reduced to zero by averaging techniques, but would permanently bias the average.

On-Line Quality Control

Measures of the overall noise content can be estimated from the complete record by digital or analog computer techniques. Typical is the autocorrelation described previously, which infers that the signal (at the sample point) is the mean, and remains unchanged throughout the record. Slowly changing signals will appear as a drift component in the samples and may unduly bias the noise measure as indicated by correlation function. Filtering techniques can be applied to estimate this slowly varying signal, and thereby determine the noise content as being any deviation from this estimate. Such filters include moving average types which weight both future and past data, either linearly or exponentially. However, all of these filters require extensive computer facilities not normally available for on-line recording.

A proposed beat-to-beat indicator of the noise content is shown in Figure 3. Using the samples of noise and signal, f(t), calculating the first, second and third order exponential filtered values, a prediction of the signal is made assuming a quadratic model for the signal. This expected value of the signal, $\hat{f(t)}$, is continuously updated by new samples, and is used as a reference for measurements of noise deviations, $f(t) - \hat{f(t)}$. The magnitude of these beat-to-beat deviations can be used as an on-line measure of noise to operate a meter, a threshold warning, or any editing control device. An integration of $|\hat{f(t)} - f(t)|$ over the exercise period also provides a good overall measure of the noise.

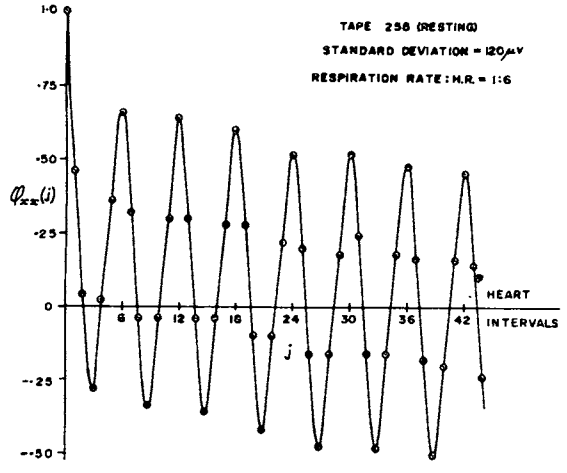

Figure 1. Autocorrelation of noise samples of resting ECG record in which subject inspired during 3 successive heart beats and expired for the next 3 beats.

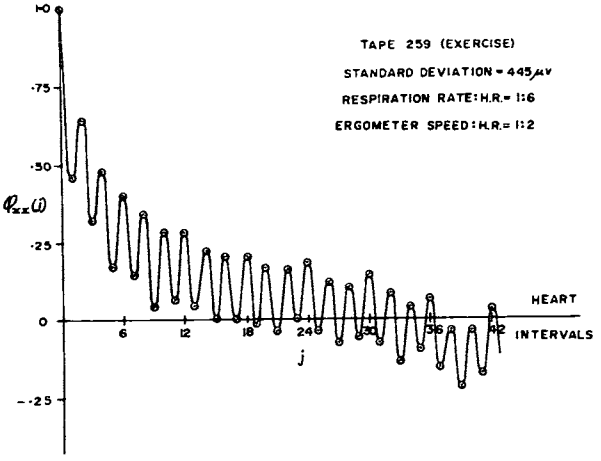

Figure 2. Autocorrelation of noise samples of exercise ECG record in which bicycling speed was 1/2 heart rate, and respiration rate was 1/6 heart rate.

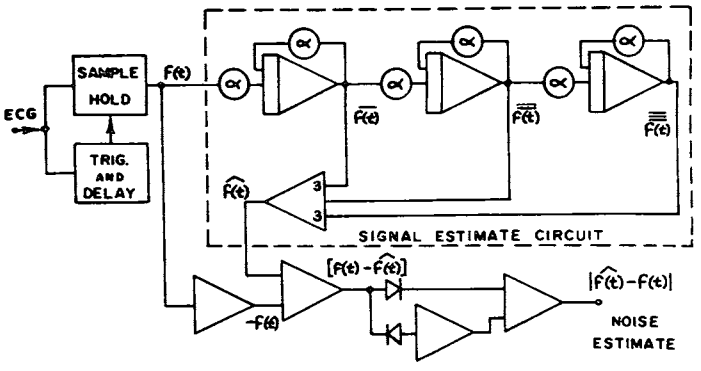

Figure 3. Circuit to measure beat-to-beat indication of noise.