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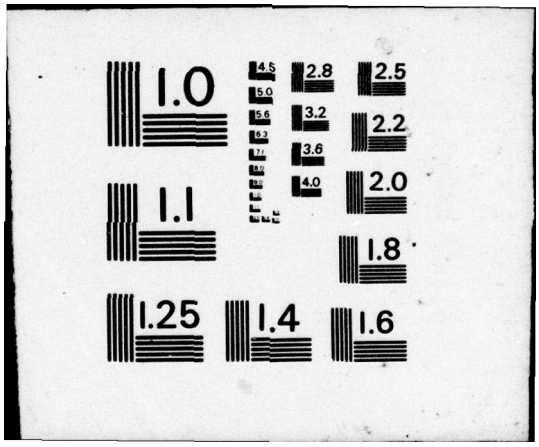
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MULTI-ELECTRODE TIME-MULTIPLEX TELEMETRY

Sid Deutsch
College of Medicine and Dentistry of New Jersey
Rutgers Medical School
P.O. Box 101
Piscataway, N.J. 08854

409 071

1 January 1977

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Naval Aerospace Medical Research Lab.
Box 29407
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20. ABSTRACT (Continue on reverse side if necessary and identify by block number)

Two multi-electrode time-multiplex telemetry units and their decoders are described. First, a 15-electrode totally-implanted unit that has a basic cycle rate of 400 Hz. Second, a 7-electrode partially-implanted unit that has a basic cycle rate of 4000 Hz. A remote 7-electrode stimulator-monitoring unit is briefly mentioned.

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Multi-Electrode Time-Multiplex Telemetry

This final reporting of research supported under ONR Contract #76-C-0241 is broken down into three sections: 1) a 15-electrode unit constructed for research at Rutgers Medical School; 2) ongoing research; and 3) units constructed specifically for the U.S. Navy.

1) 15-Electrode Unit Constructed for Research at Rutgers Medical School

This work is reported in the accompanying paper, "A 15-Electrode Totally Implanted Time-Multiplex Telemetry Unit", by S. Deutsch, IEEE Transactions on Communications, vol. COM-24, no. 10, October 1976.

2) Ongoing Research

We have completed the breadboard design and construction of a totally implanted remote stimulator-monitoring unit. This enables one to remotely stimulate one or any combination of seven electrodes. At the same time, the seven electrodes are continuously monitored. This will enable the experimenter to study simultaneous evoked potentials in seven selected areas of the monkey's brain. Evoked potentials may be a more effective way of assessing brain damage than the study of normally occurring EEG's.

3) Units Constructed Specifically for the U.S. Navy

During the year beginning with October 1, 1975, the Bio-engineering Section of CMDNJ-Rutgers Medical school designed and constructed a series of 6 biotelemetry units for monkeys for Dr. Marc Weiss and his colleagues at Michoud Station in New Orleans. The design and construction philosophy are similar

to that reported in the 15-electrode paper. Therefore, only a brief discussion of the U.S. Navy units and decoder is given here.

Only 7 active electrodes are used. The units were designed to be partially implanted, with an external 3 volt battery. To make it possible to record from single neural units, relatively high switching frequencies are used. The basic cycle rate is 4000 Hz, and the channel rate is 32,000 Hz. The channel 8 input consists of a 2000 Hz 5 mV peak-to peak reference signal, and a 16,000 Hz sync signal is added to all channels. The unit is 1.4 X 2.5 X 4.2 cm in size, weighs 20 grams, and draws 10 mA. Low-frequency cutoff is at 0.2 Hz. The output is FM with a peak deviation of 75 KHz of a 30 MHz carrier at 5 mV peak channel input. The circuit is depicted on the "Navy Transmitter" sheet.

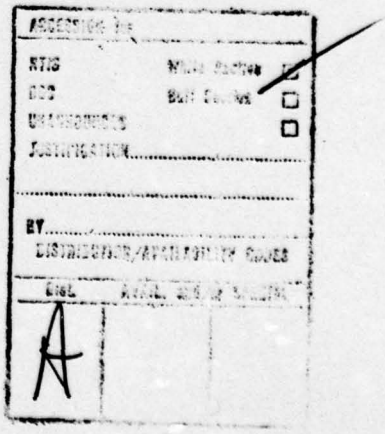
In the 15-electrode unit, the channels are first sampled at a 6400 Hz rate and then amplified in a single wideband operational amplifier. For the U.S. Navy unit, because of the much higher sampling rate, a single amplifier would have insufficient bandwidth. Therefore, each channel signal first goes to a unity-gain operational amplifier (for high input impedance) and it is then amplified in a second operational amplifier that has a nominal voltage gain of 300. Finally, the channels are sampled at a 32,000 Hz rate and fed, without further amplification, to the base of the FM oscillator.

Since each channel uses two operational amplifiers, a total of 14 is needed. This is supplied via four National Semiconductor LM 124 F flat-packs. Each of these contains four operational amplifiers, as indicated in the "Navy Transmitter" schematic.

The external battery supplies +3 V. Internally, the RF output is half-wave rectified to supply - 3 V. This arrangement is more convenient than an external center-tapped 6 V battery.

The decoder circuit is shown on the "Navy Decoder" sheet. The circuit is somewhat simpler than that of the 15-electrode decoder because the U.S. Navy unit only has 7 electrodes. Two additional drawings, "Navy Decoder Card No. 1" and "Navy Decoder Card No. 2", give the details for the two plug-in cards.

Instructions for setting the decoder knobs are given on the "Instructions for Rutgers Medical School Decoder" sheet.



A 15-Electrode Totally Implanted Time-Multiplex Telemetry Unit

SID DEUTSCH, FELLOW, IEEE

Abstract—A 15-electrode totally implanted time-multiplex telemetry unit is described that is powered by two mercury cells in series. Transcutaneous radio frequency (RF) turn-on and magnetic turn-off are used to conserve battery power. The basic cycle rate is 400 Hz, channel rate 6400 Hz. The channel 16 input consists of a 200-Hz 5 mV peak-to-peak (p-p) reference signal, and a 3200-Hz sync signal is added to all channels. The unit is 1.5 × 2.5 × 8 cm in size, weighs 41 g, and draws 3.5 mA. Low-frequency cutoff is at 0.2 Hz. The output is FM with a peak deviation of 75 kHz of a 20 MHz carrier at 5 mV peak channel input. The equivalent noise input is 2.8 μV root mean square (rms). The decoder automatically locks on the reference signal square wave with the aid of an exclusive-OR gate.

INTRODUCTION

We now have amplifiers and CMOS digital integrated circuits that can operate on as little as 5 V. This makes it possible to design a low-noise 15-electrode totally implanted time-multiplex [1]-[6] telemetry unit that is powered by two mercury cells in series (2.7 V). Radio frequency (RF) rectification is used to generate 5 V. A design is described that features transcutaneous RF turn-on and magnetic turn-off in order to conserve battery power [7]. The signal-to-noise ratio of the system is high because amplification yields a relatively high degree of frequency modulation (FM) of the RF carrier. A 400-Hz basic cycle rate is used to get an electroencephalogram (EEG) or electrocardiogram (EKG) idealized response extending to 200 Hz. The channel 16 input, used for initial synchronization, consists of a 200-Hz 5-mV peak-to-peak (p-p) reference signal. In addition, a 3200-Hz square-wave synchronizing signal is added to all channels.

The research goal is to obtain an integrated picture of electroencephalographic activity, either in widely separated regions (especially those involving bilateral symmetry) or in a local region, such as the visual or auditory cortex, of socially interacting epileptic monkeys [8]. The advantage of a totally implanted unit, of course, is that the animal can be completely ambulatory, with only a small protuberance to reveal the presence of the transmitter [9].

Including batteries that can operate 100 hours before replacement, the unit is 1.5 × 2.5 × 8 cm in size and weighs 41 g. Although it was designed for subcutaneous implantation in monkeys, it can of course be used for EEG and/or EKG monitoring of any animal that can carry 41 g.

The implant unit is designed to handle the signals picked up by relatively gross extracellular electrodes that are embedded in brain or heart tissue, up to 2 mV peak [10], [11]. This is larger, by an order of magnitude, than the signals picked up by relatively remote surface electrodes.

Ideally, the physiological signal bandwidth should extend to dc. Experience shows, however, that each input may contain a relatively large spurious dc component due to 1) "battery" action if the electrodes and common ground strip are made of dissimilar materials, and due to differences in surface layer even if they are made of the same material; and 2) a few millivolts due to integrated circuit rectification of the relatively strong RF field that surrounds the FM transmitter coil. Physically small 1-μF capacitors are therefore used in series with each channel input to block dc components. With an effective input resistance per channel of 1.6 MΩ, the nominal low-frequency cutoff is 0.1 Hz.

FM is used in order to get a receiver output that is independent of RF signal strength [12]. Since it is convenient to use commercial equipment, the unit described below is adjusted to have a peak deviation of 75 kHz of a 20-MHz carrier at 5-mV peak channel input. A frequency converter is used to beat down to the commercial IF of 10.7 MHz.

BLOCK DIAGRAM OF IMPLANT UNIT

A block diagram of the implant unit is depicted in Fig. 1. Standard readily available integrated circuit (IC) flat-packs are employed for maximum flexibility. There is a total of 5 flat-packs. Although some reduction in volume could be achieved via a special-purpose single flat-pack, an irreducible minimum volume is dedicated to the batteries, terminal board, input blocking capacitors, and RF transmitter coil. The epoxy layer should be at least 2 mm thick.

One-half of a CD4001AK acts as a 6400-Hz astable multivibrator clock (C). The waveforms are shown in (1) through (7) of Fig. 2. A CD4024AK binary counter supplies $C/2$, $C/4$, ···, $C/32$ square waves. (The counter actually has 7 stages, but the $C/64$ and $C/128$ outputs are not used.) The $C/2$, $C/4$, and $C/8$ signals [waveforms (3), (4), and (5)] cause two CD4051AK flat-packs to sequentially close input electronic switches. The $C/16$ signal [waveform (6)] turns on the first CD4051AK unit, which covers the reference channel plus channels 1 through 7, and $C/16$ inverted turns on the second unit, channels 8 through 15.

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The author is with the Bioengineering Section, Surgery Department, CMDNJ-Rutgers Medical School, and Electrical Engineering Department, Rutgers University, Piscataway, NJ 08854.

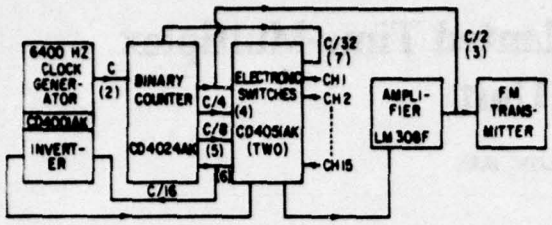

Fig. 1. Block diagram of implant unit.

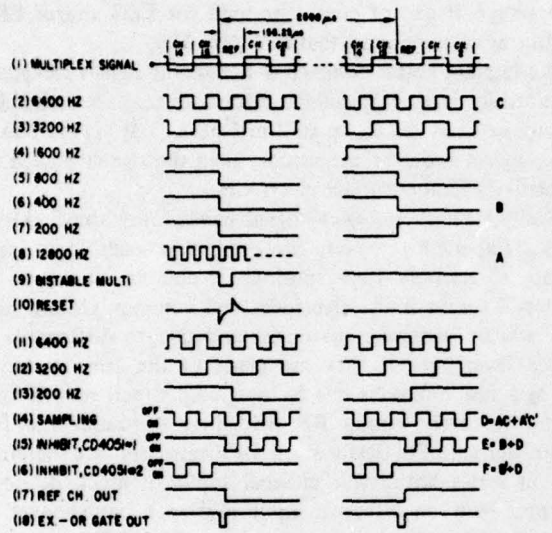

Fig. 2. Waveforms.

As previously mentioned, a $C/2 = 3200$ Hz square wave sync signal is added to all channel outputs, and the reference signal consists of a $C/32 = 200$ Hz square wave. These are shown in the "multiplex signal" waveform (1) of Fig. 2. The reference signal alternates between the −5 mV level during one basic cycle and the 0 V level during the next, 2500 μs later.

Amplification is obtained by means of a National Semiconductor LM308F operational amplifier flat-pack.

CIRCUIT DIAGRAM OF THE IMPLANT UNIT

The circuit diagram of the implant unit is shown in Fig. 3. The design philosophy here is, of course, to minimize volume, weight, and power requirements by using a minimum number of components.

Each half-period of the 6400-Hz clock, 78 μs, is approximately determined by the time constant of the 750 kΩ–100 pF combination. The 750-kΩ value is nominal; we actually use a value that gives a clock frequency within 5 percent of 6400 Hz.

With + and −2.5 V dc supplies, the counter generates 5 V p-p square waves. The 200-Hz wave is reduced to a 5-mV p-p reference signal by means of a 1 MΩ-1 kΩ voltage divider.

The bank of 15 1-μF capacitors, and the 22-μF capacitor, are low-voltage tantalum types, only 0.1 inch in diameter by 0.25 inch long.

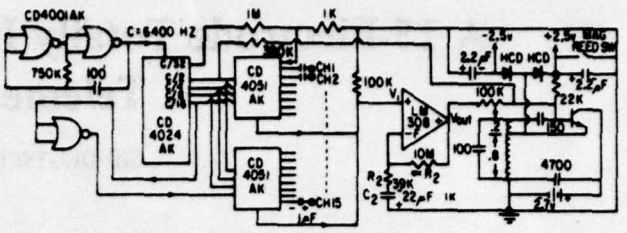

Fig. 3. Circuit diagram of implant unit.

Assuming that the resistance seen looking into the inverting input terminal of the amplifier is infinite (which only introduces appreciable gain calculation error at dc), the dependence of voltage gain on feedback is given by [7]

$$\frac{V_{out}}{V_1} = \frac{1 + j\omega\tau_2(1+\alpha)}{1 + j\omega\tau_2}$$

where the symbols are as defined in Fig. 3, with $\tau_2 = R_2C_2$. Starting with an idealized closed-loop voltage gain of 1 at dc, the gain starts to rise at a 6 dB/octave rate at a break frequency of $\omega = 1/[\tau_2(1+\alpha)]$. The gain levels off at the value $(1+\alpha)$ at a break frequency of $\omega = 1/\tau_2$. This is the nominal low-frequency cutoff contributed by feedback. With the values shown in Fig. 3, it occurs at 0.2 Hz. The nominal closed-loop voltage gain is 251 or 48 dB. Corresponding to 5 mV peak V_{in}, this yields 1.25 V peak V_{out}, the maximum obtainable without saturation with the given supply voltages.

With the relatively large amount of feedback used here, the resistance seen looking into the noninverting input terminal of the amplifier is over 1 MΩ. A physical 100-kΩ resistor is used to properly load the CD4051AK's. Because each input capacitor sees the 100-kΩ resistor for only 1/16 of the time, the effective input resistance per channel is 1.6 MΩ, and the input coupling time constant contributes a low-frequency cutoff of 0.1 Hz. Since this is substantially lower than the value contributed by feedback, the nominal low-frequency cutoff remains as 0.2 Hz.

With a typical source resistance of 50 000 Ω the LM308F specifications give an equivalent noise generator input voltage of 0.05 μV/(Hz)$^{1/2}$. Assuming a system bandwidth of 3200 Hz, the equivalent noise input is 2.8 μV root mean square (rms). This is lower than the minimum significant signal level. (Because sample-and-hold circuits in the decoder respond to wide-band noise, one must use system rather than individual channel bandwidth in the above calculation.)

Negative and positive supplies are derived from half-wave rectifiers driven by the oscillator coil. "Hot carrier" HP 5082-2835 diodes yield appreciably more dc output, at less lowering of the Q, than silicon or germanium diodes.

The positive rectifier output supplies base bias, through a 22-kΩ resistor, to the oscillator transistor. The mercury cells connect directly to the collector. To turn the unit on, an external 20-MHz oscillator is held against the animals' skin, opposite the implant coil. It induces sufficient voltage in the implanted coil so as to turn the transistor base "on," at which point oscillations begin and the unit can supply its own bias.

To turn the unit off, an external magnet is held near the implant unit. It causes a magnetic reed switch to close, thereby short-circuiting the oscillator bias supply. This turns the transistor base "off" without wasting battery power. The magnetic reed switch is 0.1 in in diameter by 0.3 in long, and is positioned towards the top of the unit.

There is a battery drain of about 3.5 mA, which gives a life of some 100 hours with RM-630 mercury cells which are 5/8 in in diameter by 1/4 in thick. With an average battery shelf life of 2 years, there is no compelling reason to use the unit as soon as it is implanted, in contrast with devices that cannot be turned on and off. When the cells are used up, a hacksaw is employed to sever them from the remainder of the unit. New cells are attached, positioned in the mold, and covered with paraffin and epoxy. The new and old epoxies adhere satisfactorily.

The transistor type has been omitted from the schematic because there are many physically-small silicon n-p-n (or p-n-p) types that can be used for the oscillator. We pretest the transistor in a 100-MHz Colpitts oscillator circuit.

The actual frequency used for the RF carrier is a compromise based on the following four considerations [7].

1) The advantage of operating at a relatively low frequency, for an implanted device, is that tissue losses are reduced, so that the oscillator Q can be correspondingly higher. One can show that the Q, based on eddy current losses, is inversely proportional to frequency.

2) Another advantage of low frequencies is that depth of penetration increases, so that one can work with larger animals. For a central-body implant, the animal's diameter can be as much as 22.5 cm (9 in) at 100 MHz, and 50 cm (20 in) at 20 MHz.

3) An argument for operating at higher frequencies is that the antennas become more effective.

4) There is an important advantage of operating at 100 MHz in that one can use standard low-cost FM receiving equipment.

Our experience is that the improved reliability and stability of an implanted oscillator at 20 MHz as compared to 100 MHz is the most important consideration. We use a standard FM receiver in conjunction with a converter. The converter contains a 30.7 MHz tunable local oscillator which provides a 10.7 MHz IF output. The latter is fed to the IF input of the FM receiver.

As shown in Fig. 3, the amplifier output feeds through a 100-kΩ resistor to the oscillator transistor base, and 3200-Hz channel sync is added via a 390-kΩ resistor. Here they vary the base current, which in turn varies the base-charging capacitance, thus yielding FM. The resistance values are nominal; we actually use values that correspond to a peak deviation of 75 kHz with 5 mV peak channel input.

The physical layout of the implant unit is outlined in Fig. 4 and a photograph is shown in Fig. 5. The unit is curved to better conform to the curvature of the animal's body. "Cordwood" construction is used in which the electronic components are held between two parallel strips of standard circuit board that have holes spaced 0.05 in apart. The assembly is first embedded in paraffin for waterproofing and then in epoxy. These are subjected to a partial vacuum to remove gases and moisture. The technique follows the methods that are employed in the heart pacemaker industry.

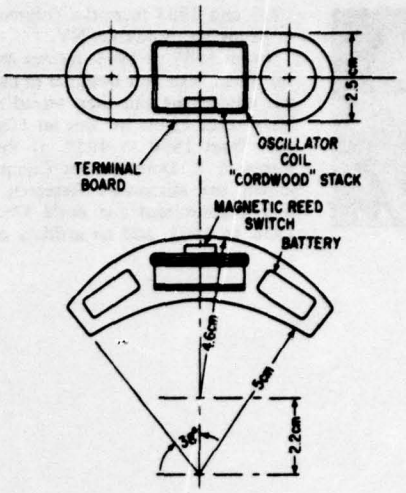

Fig. 4. Physical layout of implant unit.

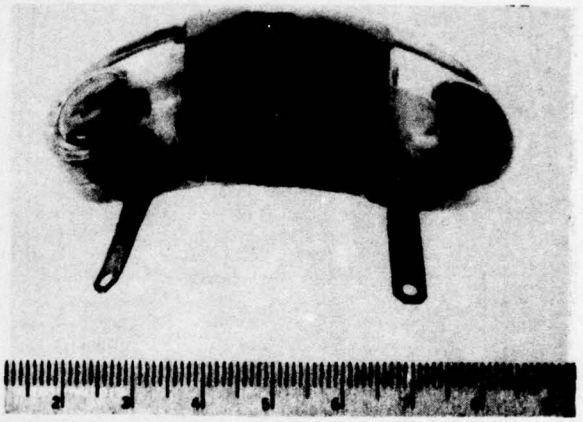

Fig. 5. Photograph of implant unit. Units shown on ruler are centimeters.

BLOCK DIAGRAM OF THE DECODER

The multiplex signal out of the FM receiver must, of course, be decoded into 15 channel outputs. This is more difficult than in a conventional time-multiplex communication system because, as the animal moves around the cage, some combinations of location and head orientation are such that the received RF signal becomes practically zero, so that synchronization is lost.

The block diagram, waveforms, and circuit diagrams are shown in Figs. 6, 2, and 7, respectively.

The philosophy used here is to electronically switch from a "search" mode when the multiplex signal is absent or first appears, to a "synchronized" mode when the multiplex signal is "captured." A phase-locked loop (PLL) is employed for synchronization.

1947 and 1955 from the Polytechnic Institute of Brooklyn, Brooklyn, NY.

From 1935 to 1944, he was an Electric Motor Technician and Designer of electro-mechanical equipment and then served with the U.S. Navy until 1946. He was an Electronics Engineer from 1950 to 1954, at the Polytechnic Research & Development Company,and then joined the Mircowave Research Institute. He was a consultant for Budd Electronics from 1958 to 1961, and an affiliate of the Rockefeller Institute from 1961 to 1964. He was a Physics Instructor at Hunter College during 1943–1944, was a radio and television instructor from 1946 to 1950, and taught at the City College of New York from 1955 to 1957. From 1951 to 1972, he was at the Polytechnic Institute of Brooklyn, most recently as Professor of Electrical Engineering and Bioengineering. He is now Professor of Bioengineering at Rutgers Medical School and an Adjunct Professor of Electrical Engineering at Rutgers University.

Dr. Deutsch is the author of *Theory and Design of Television Receivers* and *Models of the Nervous System*. He is a Fellow of the Society for Information Display.

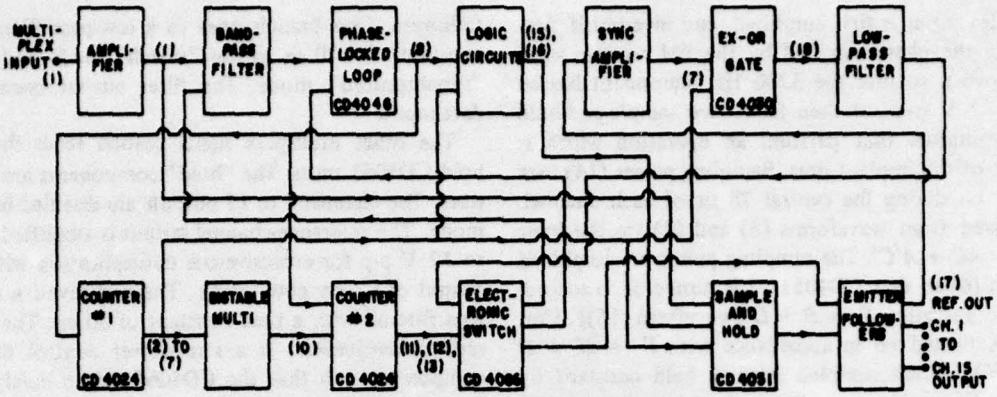

Fig. 6. Block diagram of decoder.

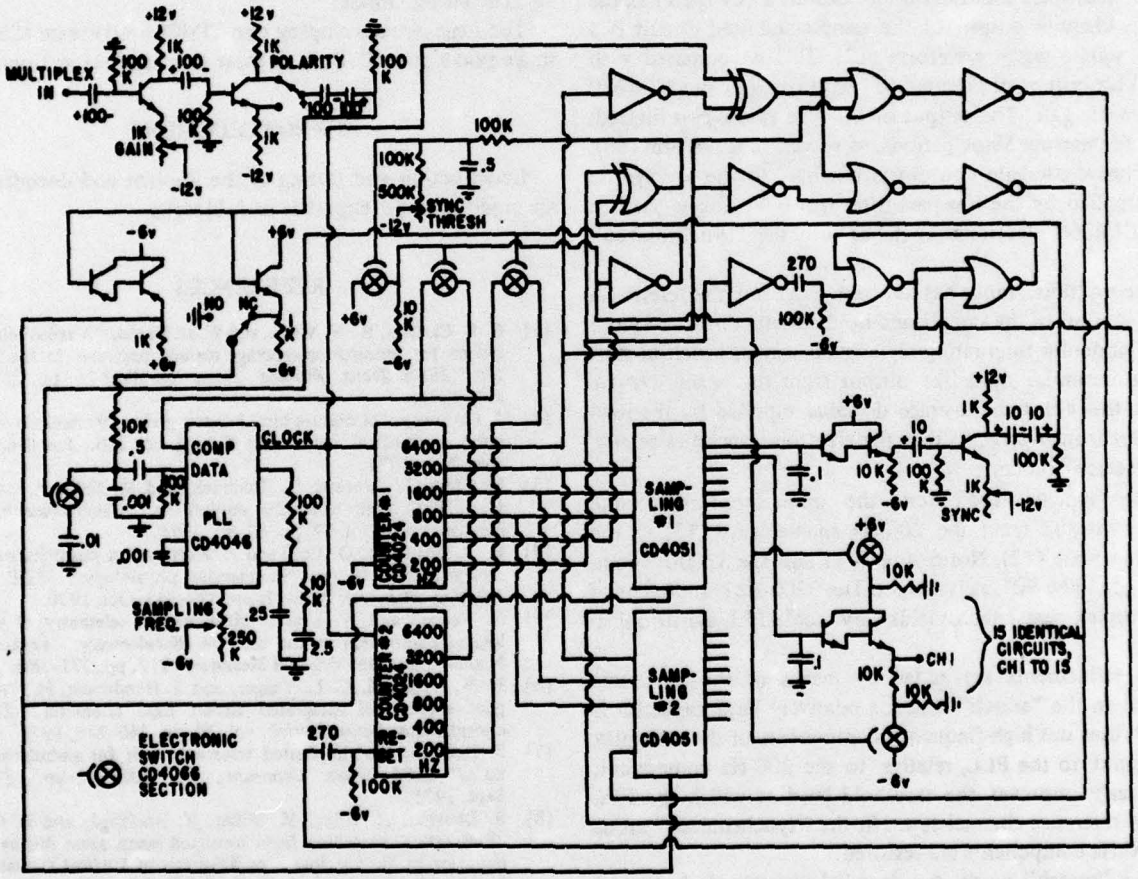

Fig. 7. Circuit diagram of decoder.

The CD4046 PLL compares two inputs, the data and comparator signals. Ideally, the comparator input should lag the data input by 90°. In the "search" mode, the comparator input is the 200-Hz waveform (13). The first portion of the waveform displays a negative-going step that occurs in the center of the negative-going reference level of waveform (1). The second portion of (13) shows a positive-going step that coincides with the center of the positive-going reference level of (1). This satisfies the 90° phase lag requirements. To generate (13), we first feed the 12 800 clock output of the PLL, waveform (8), to CD4024 counter #1. The latter supplies waveforms (2)-(7). The negative-going edge of (7) triggers a bistable multivibrator, causing its output to go negative, waveform (9). The bistable multi is returned to a positive output by waveform (2). The net result is that (9) is 78 μs wide. Differentiation of (9) gernerates a positive reset pulse for CD4024 counter #2. The latter supplies waveforms (11), (12), and (13). The 78-μs period is the time delay needed to give the "search" mode comparator signal (13) a 90° phase lag with respect to the reference channel data input.

The multiplex input is first amplified, and inverted if necessary (because the phase supplied by the FM receiver is, in general, unknown), so that the 3200 Hz component has an amplitude of 2 V p-p. It then feeds two sample-and-hold CD4051 commutators that perform an operation which is inverse to that of the implant unit. Sampling pulses (14) turn the CD4051's on during the central 78 μs of each channel. They are derived from waveforms (8) and (2) via the logic operation $D = AC + A'C'$. The sampling pulses are combined with waveform (6) so that CD4051 #1 is turned on in accordance with logic operation $E = B + D$ [waveform (15)]. Unit CD4051 #2 is turned on in accordance with $F = B' + D$ [waveform (16)]. Each sampled level is held constant by means of 0.1-μF capacitors until the next sampling period, 2500 μs later.

If the multiplex input has been "captured" by the PLL, the reference-channel output of the sample-and-hold circuit is a 200-Hz square wave, waveform (17). This is compared with the 200-Hz output of counter #1, waveform (7), in a CD4030 exclusive-OR gate. The output of the gate is low-pass filtered. Except for narrow 39-μs periods, as shown in waveform (18), the exclusive-OR gate output is positive, so the average dc value supplied by the low-pass filter is +6 V. This is used to throw CD4066 electronic switches into the "synchronized" mode.

If the multiplex input has *not* been captured, the reference-channel output of the sample-and-hold circuit consists of more or less randomly fluctuating rectangular waves, which in turn leads to a similar noise-like output from the exclusive-OR gate. In this event the average dc value supplied by the low-pass filter is near zero, so that the electronic switches remain in the "search" mode.

In the "synchronized" mode, the comparator input for the PLL is switched from the 200-Hz square wave (13) to the 3200 Hz square (12). Notice that (12) lags the 3200-Hz component of (1) by 90°, as it should. The 3200-Hz component of the multiplex signal then yields very stable PLL synchronization.

Two refinements are added by means of the electronic switches: In the "search" mode, a relatively large capacitor is used to filter out high-frequency components of the multiplex signal input to the PLL, relative to the 200-Hz component. This greatly improves the threshold level at which the PLL locks on reference-channel sync. In the "synchronized" mode the 3200-Hz components are restored.

In the "search" mode, the decoded outputs of channels 1 to 15 are inhibited by opening $B+$ and $B-$ supplies of emitter-follower output stages. Since the EEG output can look very much like a random-noise signal, it is important to inhibit the output unless bona-fide synchronization exists.

CIRCUIT DIAGRAM OF THE DECODER

Most of the details of the decoder circuit, Fig. 7, have already been covered in discussing the block diagram.

The multiplex input is amplified in a single stage. The signal then goes to a phase inverter. After the proper phase is selected by the polarity switch, it feeds to isolating emitter followers. One branch goes to a low-pass filter with a time constant of 110 μs in the "search" mode and 10 μs in the "synchronized" mode. The filter output becomes the PLL data input.

The other multiplex signal branch feeds the sample-and-hold CD4051 units. The "hold" components are 0.1 μF capacitors. The channel 1 to 15 outputs are disabled in the "search" mode. The reference-channel output is amplified to bring it up to 12 V p-p for exclusive-OR multiplication with the 200-Hz output of binary counter #1. The exclusive-OR output is low-pass filtered with a time constant of 50 ms. The "sync threshold" potentiometer is a screwdriver control that adds a dc component such that the CD4066 switch barely closes when the exclusive-OR low-pass filter output is +6 V. It is the closing of this electronic switch that places the decoder in the "synchronized" mode.

The logic circuits employ two CD4001 NOR gate IC's and a single ex-OR gate IC. Each of these IC's has four sections.

ACKNOWLEDGMENT

Construction and testing of the implant and decoder units are credited to the expertise of J. Morgan.

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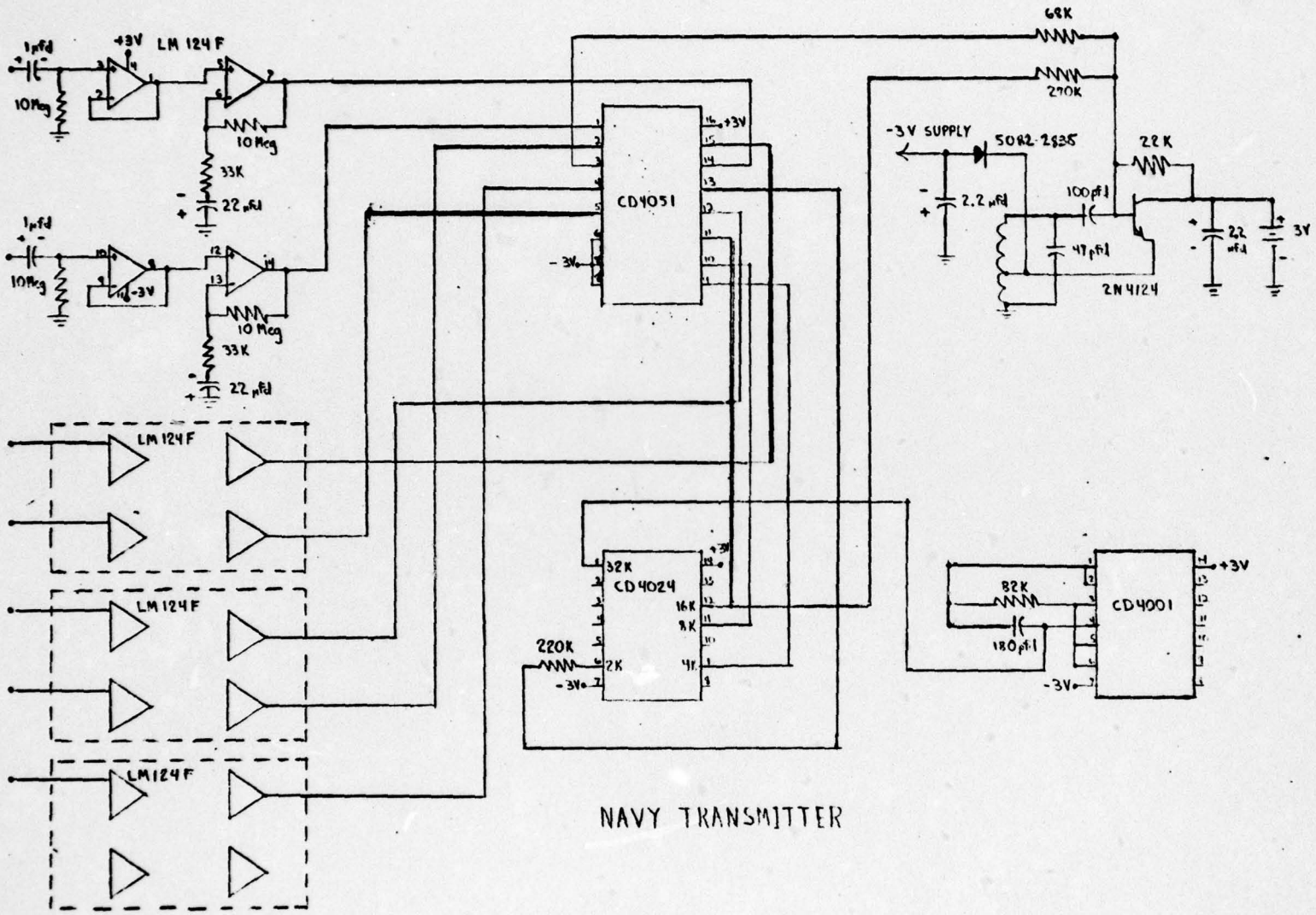
★

Sid Deutsch (A'44–M'55–SM'69–F'75) was born in New York, NY on September 19, 1918. He received the B.E.E. degree in 1941 from Cooper Union, New York, NY, and the M.E.E. and D.E.E. degrees in

1947 and 1955 from the Polytechnic Institute of Brooklyn, Brooklyn, NY.

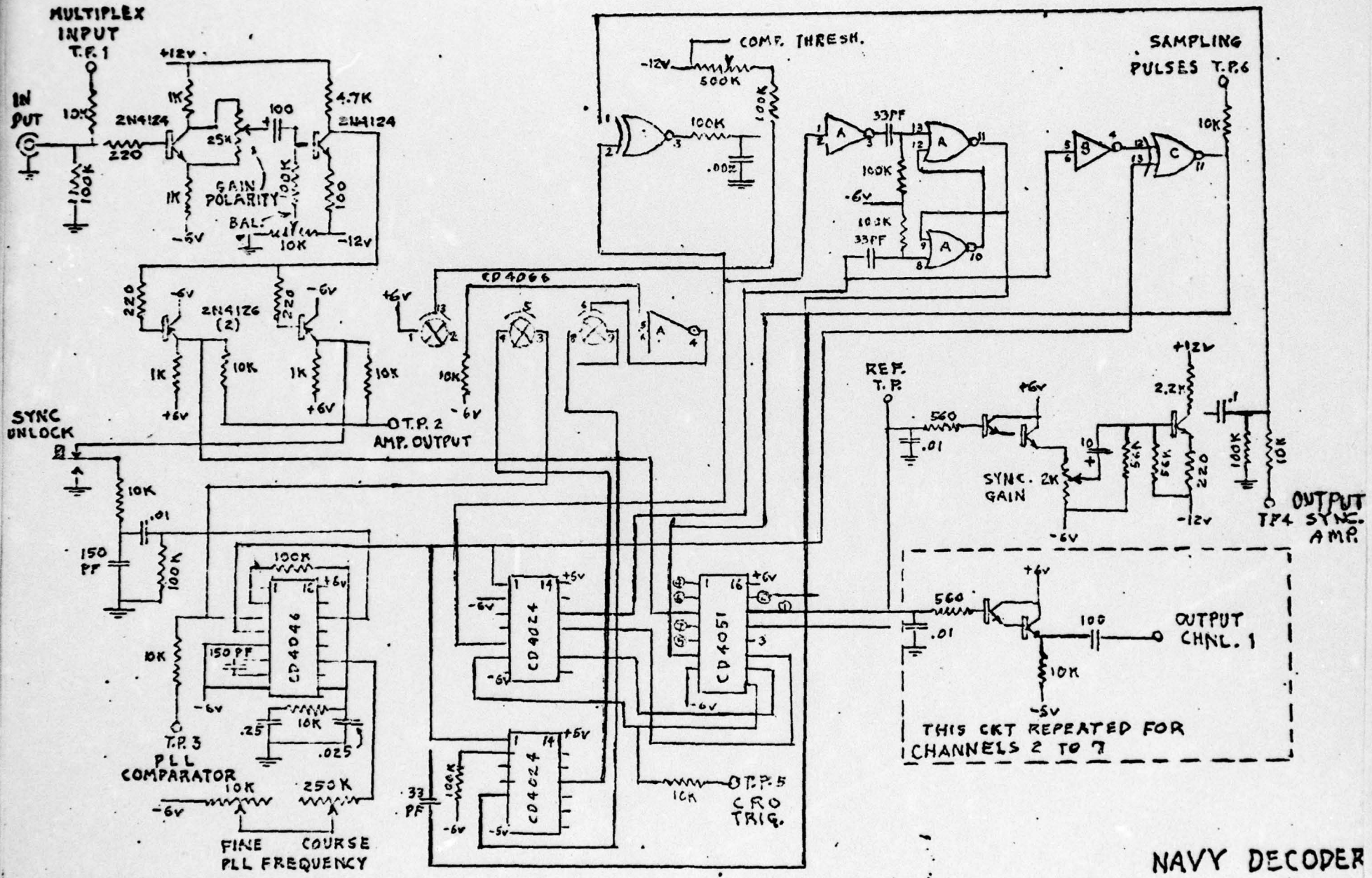
From 1935 to 1944, he was an Electric Motor Technician and Designer of electro-mechanical equipment and then served with the U.S. Navy until 1946. He was an Electronics Engineer from 1950 to 1954, at the Polytechnic Research & Development Company,and then joined the Mircowave Research Institute. He was a consultant for Budd Electronics from 1958 to 1961, and an affiliate of the Rockefeller Institute from 1961 to 1964. He was a Physics Instructor at Hunter College during 1943–1944, was a radio and television instructor from 1946 to 1950, and taught at the City College of New York from 1955 to 1957. From 1951 to 1972, he was at the Polytechnic Institute of Brooklyn, most recently as Professor of Electrical Engineering and Bioengineering. He is now Professor of Bioengineering at Rutgers Medical School and an Adjunct Professor of Electrical Engineering at Rutgers University.

Dr. Deutsch is the author of *Theory and Design of Television Receivers* and *Models of the Nervous System*. He is a Fellow of the Society for Information Display.



NAVY TRANSMITTER

J.P.
1/26/76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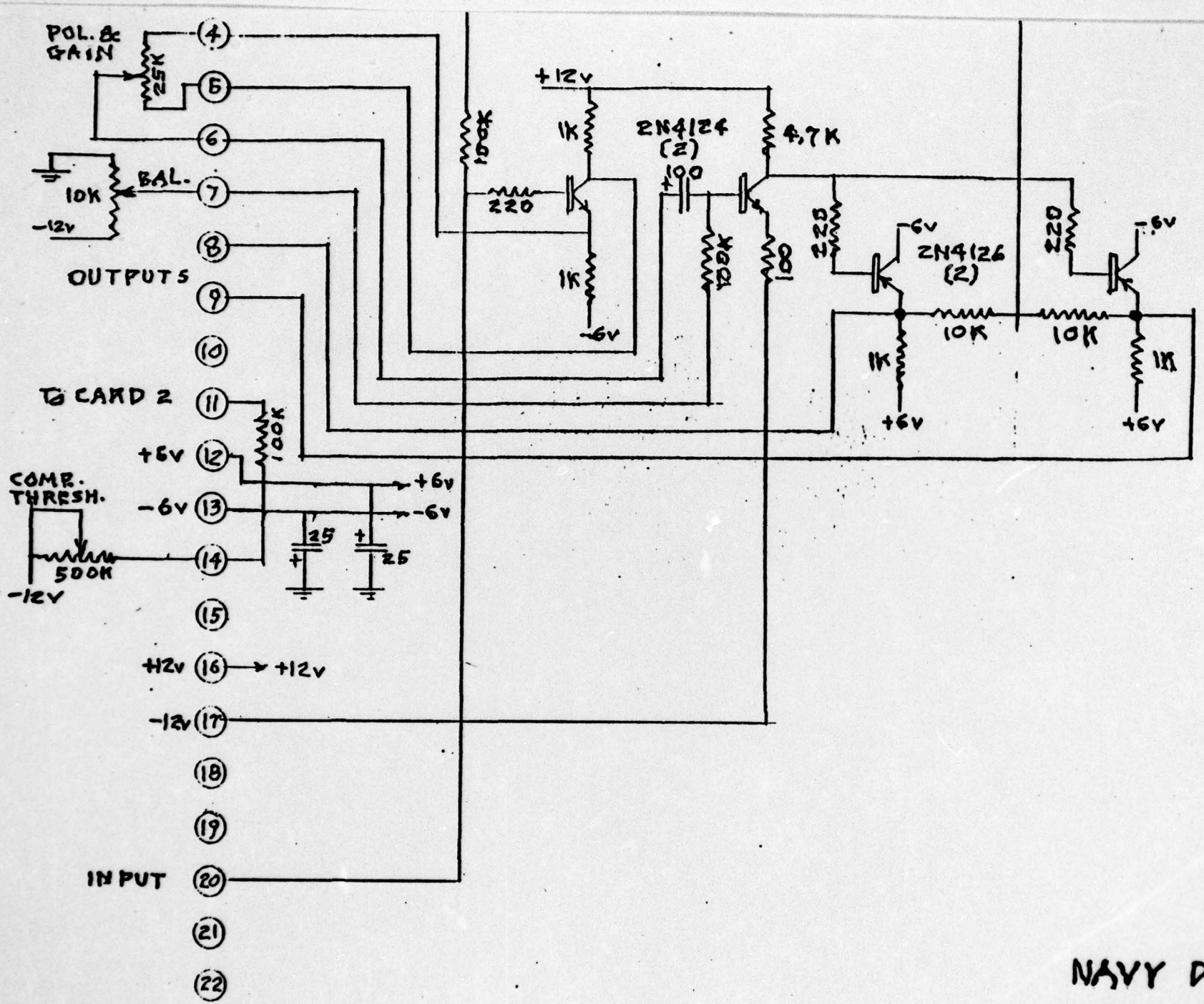
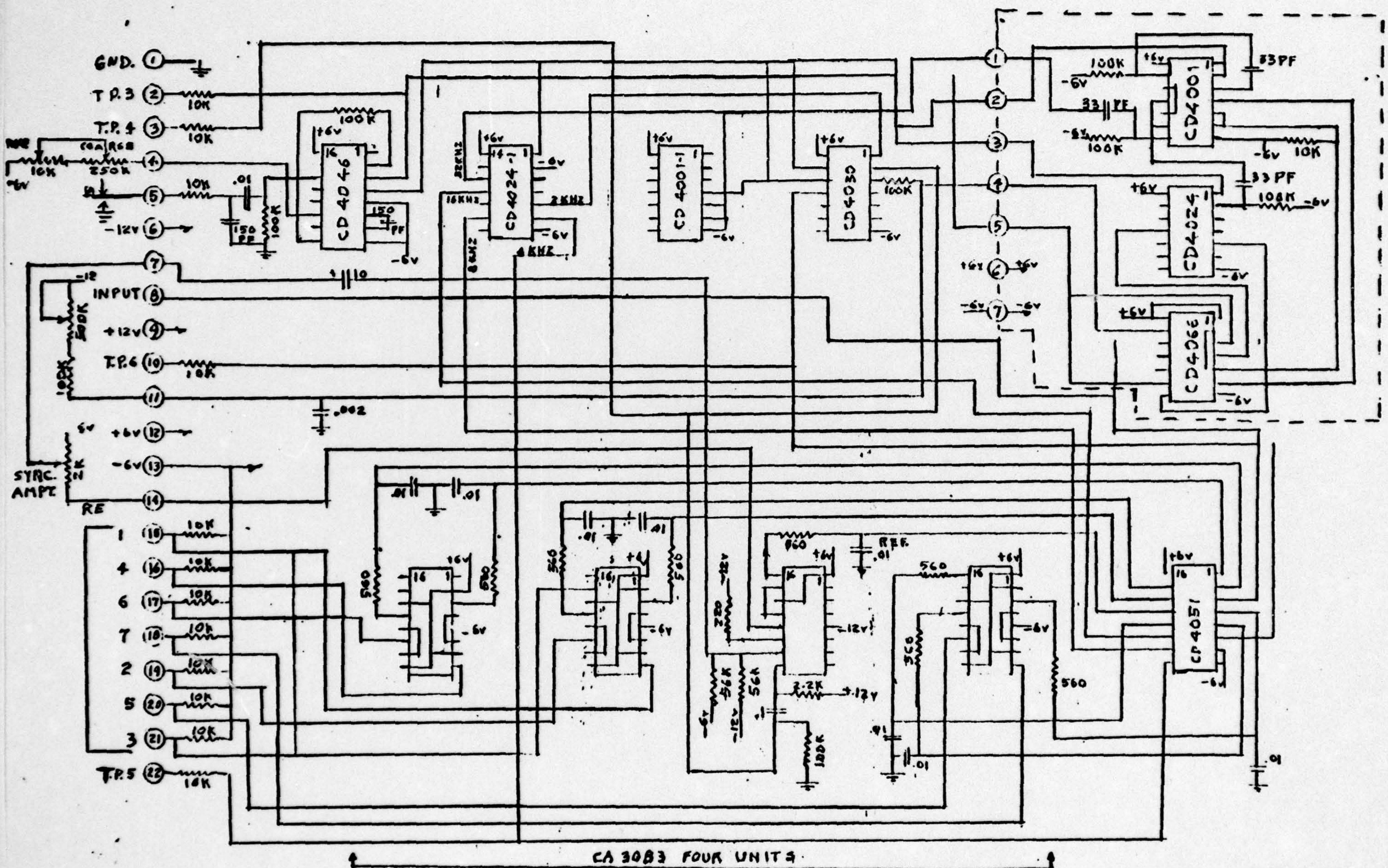
100K
GAIN POLARITY
BAL.
10K
-6V
-12V
10
COMP. THRESH.
-12V
500K
100K
100K
.002
33PF
100K
-6V
10.3K
33PF
SAMPLING PULSES T.P.6
10K
CD4066
+6V
-6V
2N4126 (2)
220
1K
10K
SYNC UNLOCK
O T.P. 2 AMP. OUTPUT
REF. T.P.
560
.01
+6V
2.2K
SYNC. 2K GAIN
10
5.6K
3.3K
220
100K
101
-12V
-6V
OUTPUT SYNC. AMP.
150 PF
100K
CD4046
150 PF
.25
10K
.025
T.P. 3 PLL COMPARATOR
10K
250K
-6V
FINE PLL FREQUENCY
COURSE
CD4024
CD4051
33 PF
100K
-5V
O T.P. 5 CRO TRIG.
560
100
OUTPUT CHNL. 1
10K
-5V
THIS CKT REPEATED FOR CHANNELS 2 TO 7
NAVY DECODER
SEPT. 1975



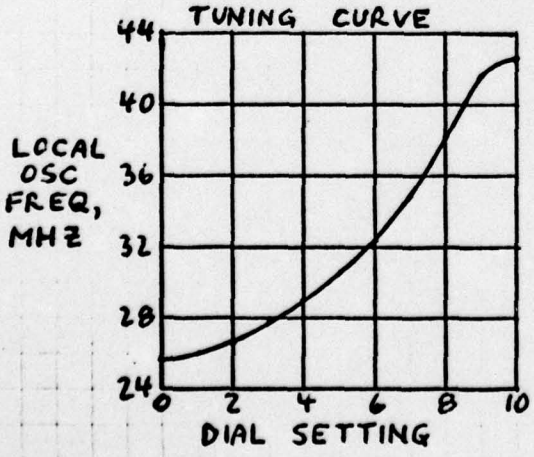
NAVY DECODER
CARD No. 1



NAVY DECODER
CARD No. 2

Instructions for Rutgers Medical School Decoder

1. Turn "Comparator Threshold" and "Sync Amp Gain" fully clockwise. Monitor "Amplifier Output" TP. Set "Balance" for zero DC output with no signal input.
2. Connect 30 MHz from transmitter to rear "RF In"of Decoder. Connect rear "IF Out" of Decoder to FM receiver "IF In". Connect Discrim. output of FM receiver to rear "Multiplex In" of Decoder.
3. Monitor "Multiplex Input" TP. Set "Convert Frequency" fine tuning to center (dot on top). Set coarse tuning to 10.7 MHz above carrier per tuning curve below. Tune for clean, noise-free multiplex signal. Check for zero DC offset.
4. Monitor "Amplifier Output" TP. Adjust "Amplifier Gain, Polarity" for 6V p-p Multiplex signal (16 kHz square wave). Set CRO controls for Reference Signal overlap (+ and - Reference Signals in same time slot as shown in Fig. 1). Polarity should be such that channels on each side of Reference Signal are positive-going. Set CRO controls so that 8 Divisions = 8 channels of multiplex signal.
5. Monitor "CRO Trigger" TP. Depress "Sync Unlock" button. Set "PLL Frequency" fine control to center (dot on top). Set coarse control so that one cycle of square wave = 8 divisions on CRO.
6. Use "CRO Trigger" TP to externally trigger CRO. Monitor "Amplifier Output" TP and "CRO Trigger." Reference signal should start at negative-going edge of 4 kHz square wave. Verify 8 channels = 1 cycle of square wave.
7. Monitor "Sync Amp Out" TP. Adjust "Sync Amp Gain" for 12V p-p.
8. Monitor "PLL Comparator" TP. The signal should be a 2 kHz square wave. Adjust "Comparator Threshold" so that 2 kHz square wave barely switches over to a clean 16 kHz square wave.
9. Monitor "Reference" Channel Out. The signal should be a 2 kHz square wave.
10. Monitor "Multiplex Input" and "Sampling Pulses". The negative-going sampling pulses should occur within each channel pulse.



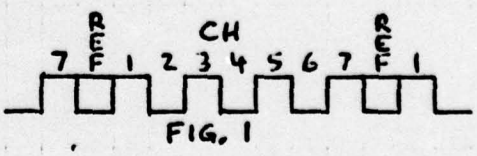
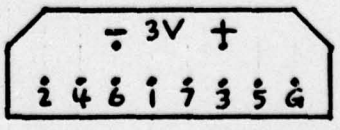


FIG. 1



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