US00RE35122E

United States Patent [19]

Corenman et al.

[11] E **Patent Number: Re. 35,122**

[45] **Reissued Date of Patent: Dec. 19, 1995**

[54] **METHOD AND APPARATUS FOR DETECTING OPTICAL PULSES**

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[73] Assignee: **Nellcor Incorporated**, Pleasanton, Calif.

[21] Appl. No.: **840,437**

[22] Filed: **Feb. 24, 1992**

Related U.S. Patent Documents

Reissue of:

[64] Patent No.: **4,911,167**
Issued: **Mar. 27, 1990**
Appl. No.: **175,152**
Filed: **Mar. 30, 1988**

U.S. Applications:

[63] Continuation-in-part of Ser. No. 742,720, Jun. 7, 1985, Pat. No. 4,802,486, which is a continuation of Ser. No. 718,525, Apr. 1, 1985, abandoned.

[51] **Int. Cl.**[6] **A61B 5/00**

[52] **U.S. Cl.** **128/633**; 128/665; 356/41

[58] **Field of Search** 128/632–633, 128/637, 667–668, 670, 671, 687–690, 696, 700, 706, 708

[56] **References Cited**

U.S. PATENT DOCUMENTS

Re. 27,042	1/1971	Jorgensen et al.	127/700
2,827,040	3/1958	Gilford .	
2,933,081	4/1960	Passannante .	

(List continued on next page.)

FOREIGN PATENT DOCUMENTS

102816	3/1984	European Pat. Off. .
104722	4/1984	European Pat. Off. .
104771	4/1984	European Pat. Off. .
0104771	4/1984	European Pat. Off. .
2089999	6/1982	United Kingdom .

OTHER PUBLICATIONS

Goodlin et al., "Systolic Time Intervals in the Fetus and Neonate *Obstetrics and Gynecology*", vol. 34, p. 295 (Feb. 1972).

Goodlin, *Care For The Fetus*, p. 101 (Masson, 1979).

Schotz et al., "The Ear Oximeter as a Circulating Monitor," Anesthesiology, vol. 19, p. 386 (1958).

Cohen et al., "Self-Balancing System for Medical and Physiological Instrumentation," *IEEE Trans. Bio–Med. Eng.*, vol. BME–18, p. 66, (1971).

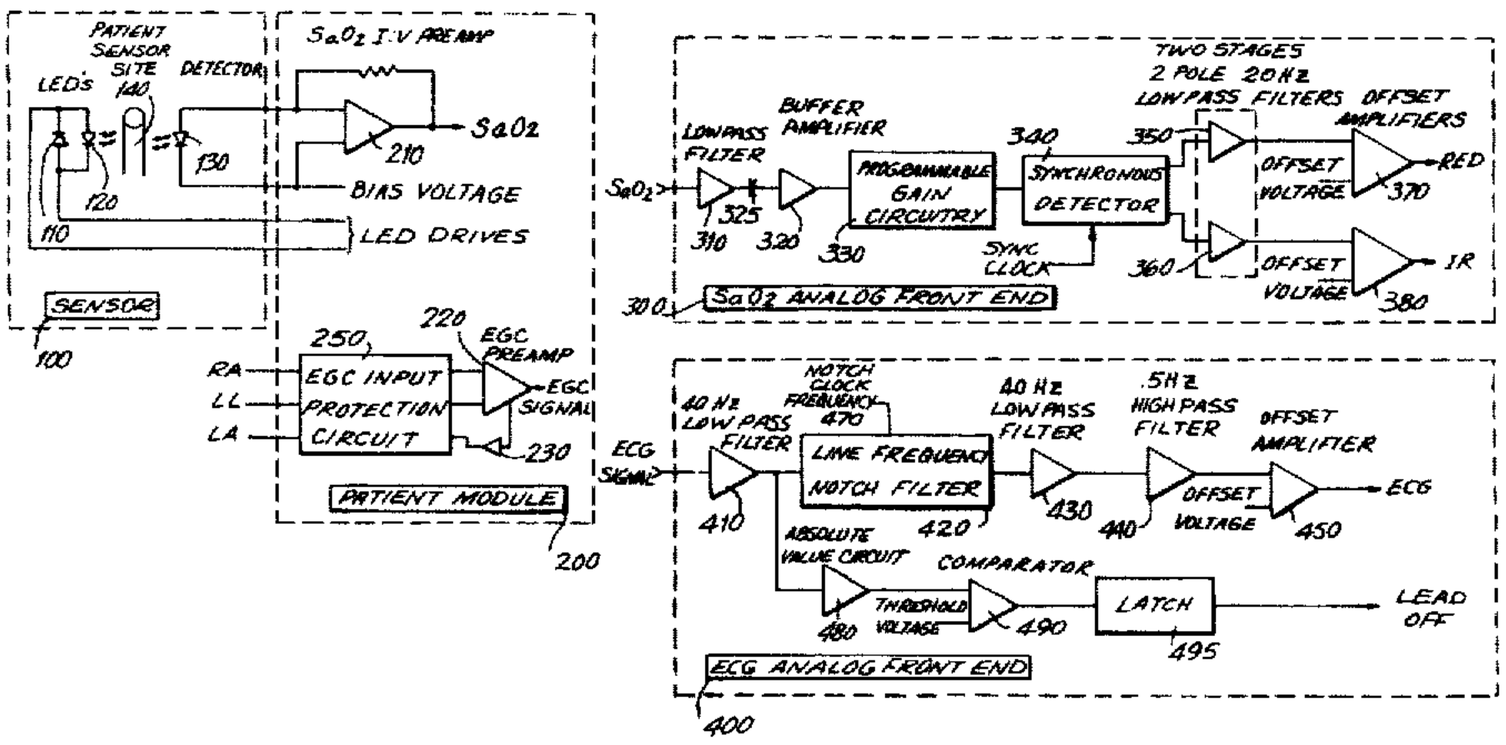
Goodlin, "Interpartum Fetal Heart Rate Responses and Plethysmographic Pulse", *Amer. J. Obstet. Gynec.*, vol. 110, p. 210 (1971).

Primary Examiner—Lee S. Cohen
Assistant Examiner—Robert L. Nasser, Jr.
Attorney, Agent, or Firm—Townsend and Townsend and Crew

[57] **ABSTRACT**

A method and apparatus for improving the calculation of oxygen saturation and other blood constituents by non-invasive pulse oximeters. The method and apparatus permit more accurate determination of blood flow by collecting time-measures of the absorption signal at two or more wavelengths and processing the collected time-measure to obtain composite pulsatile flow data from which artifacts have been filtered. The processing may occur in the time domain or in the frequency domain. In the preferred time domain embodiment, successive portions of periodic information are weighted and added together in synchrony to obtain the composite pulse information. In the preferred frequency domain embodiment, the time-measure is Fourier transformed into its spectral components to form the composite information. A new method and apparatus for correlating the heartbeat and optical pulse is provided whereby a product of the ECG R-wave and optical pulse signals corresponding to the same heartbeat is obtained, and one signal is time shifted relative to the other until a maximum waveform product corresponding to the heartbeat is determined.

34 Claims, 22 Drawing Sheets



U.S. PATENT DOCUMENTS

3,318,303	5/1967	Hammacher	128/687 X
3,412,729	11/1968	Smith, Jr. .	
3,520,295	7/1970	Kelly .	
3,590,811	7/1971	Harris .	
3,608,545	9/1971	Novack et al.	128/700
3,618,592	11/1971	Stewart .	
3,651,806	3/1972	Hirshberg .	
3,658,060	4/1972	Eklof .	
3,704,706	12/1972	Herczfeld et al. .	
3,734,086	5/1973	Phelps, Sr. .	
3,742,938	7/1973	Stern	128/687
3,773,033	11/1973	Rodbard et al. .	
3,948,248	4/1976	Zuckerman et al. .	
3,994,284	11/1976	Voelker .	
3,994,285	11/1976	Doll .	
3,998,550	12/1976	Konishi et al.	356/39
4,000,461	12/1976	Barber et al. .	
4,023,563	5/1977	Reynolds et al. .	
4,023,564	5/1977	Valiguette et al. .	
4,030,485	6/1977	Warner .	
4,036,211	7/1977	Veth et al. .	
4,053,911	10/1977	Hudspeth et al.	364/415
4,063,551	12/1977	Sweeney .	
4,086,915	5/1978	Kofsky	128/22
4,154,230	5/1979	Lee	128/661
4,181,134	1/1980	Mason et al.	128/689
4,216,779	8/1980	Squires et al.	128/680
4,266,554	5/1981	Hamaguri	128/633
4,325,384	4/1982	Blaser et al.	128/696
4,353,998	12/1982	Sawa .	
4,402,325	6/1983	Sawa .	
4,406,658	9/1983	Lattin et al.	128/802 X
4,407,290	10/1983	Wilber	128/633
4,422,458	12/1983	Kravath	128/671
4,425,921	1/1984	Fujisaki et al.	128/700
4,425,922	1/1984	Conti et al.	128/691
4,440,176	4/1984	Miodownik	128/708
4,446,868	5/1984	Aronson	128/708
4,446,871	5/1984	Imura .	
4,450,838	5/1984	Miodownik	128/204.23
4,545,387	10/1985	Balione .	
4,573,478	3/1986	Arnold et al. .	
4,577,639	3/1986	Simon et al. .	
4,586,513	6/1986	Hamaguri .	
4,589,420	5/1986	Adams et al. .	
4,617,937	10/1986	Peel et al. .	
4,653,498	3/1987	New, Jr. et al.	128/633
4,802,486	2/1989	Goodman et al.	128/666 X
4,824,242	4/1989	Frick et al. .	

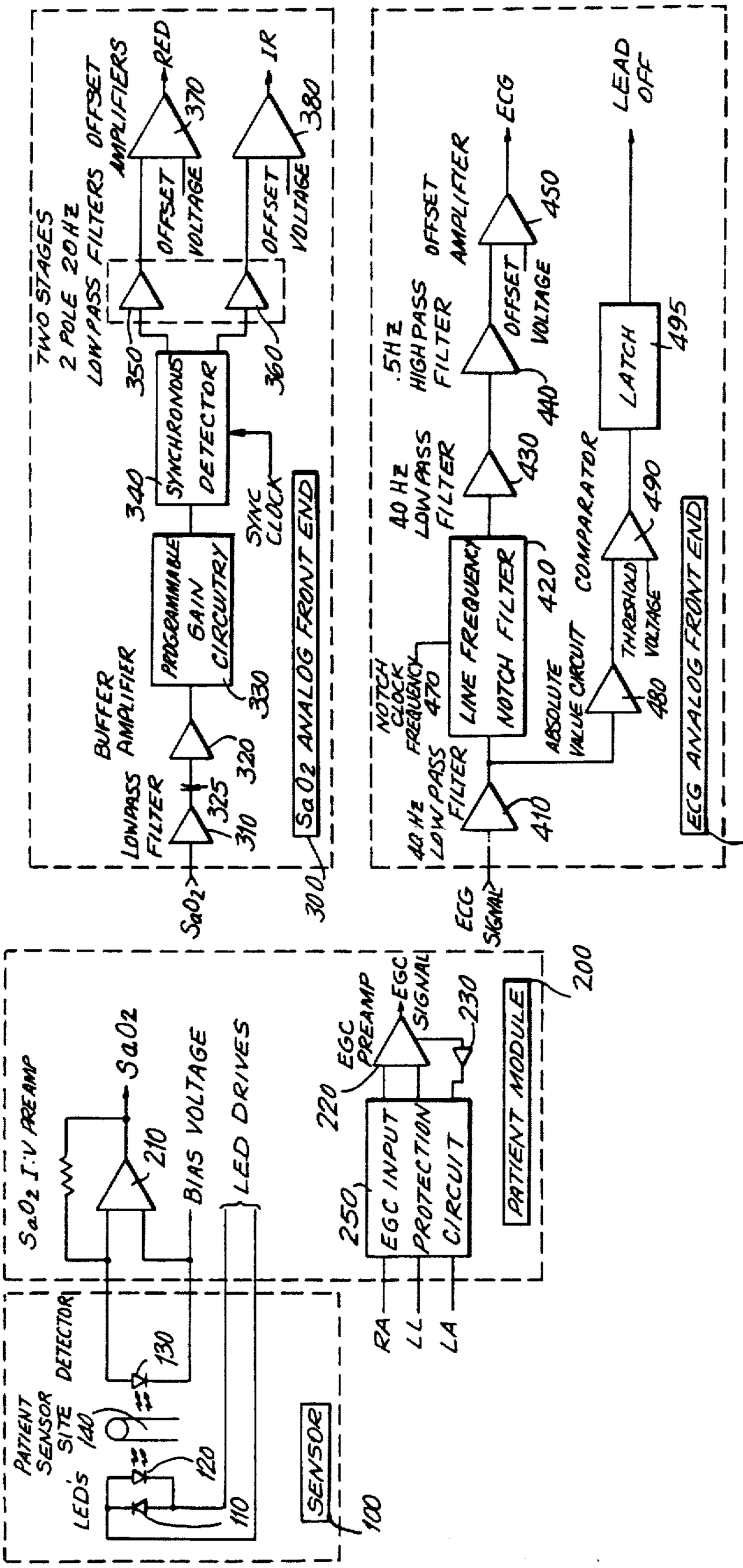

PATIENT SENSOR SITE
LED'S
140
120
110
130
DETECTOR
SENSOR
100
SaO_2 I:V PREAMP
SaO_2
210
BIAS VOLTAGE
LED DRIVES
BUFFER AMPLIFIER
PROGRAMMABLE GAIN CIRCUITRY
320
330
LOWPASS FILTER
325
310
SaO_2
300
TWO STAGES 2 POLE 20Hz LOWPASS FILTERS
OFFSET AMPLIFIERS
RED
370
IR
380
OFFSET VOLTAGE
OFFSET VOLTAGE
340
SYNCHRONOUS DETECTOR
350
360
SYNC CLOCK
SaO_2 ANALOG FRONT END
220
EGC PREAMP
EGC SIGNAL
230
250
EGC INPUT PROTECTION CIRCUIT
PATIENT MODULE
200
RA
LL
LA
ECG SIGNAL
40 Hz LOWPASS FILTER
410
.5Hz HIGHPASS FILTER
OFFSET AMPLIFIER
ECG
450
OFFSET VOLTAGE
440
40 Hz LOWPASS FILTER
430
NOTCH CLOCK FREQUENCY
470
LINE FREQUENCY NOTCH FILTER
420
ABSOLUTE VALUE CIRCUIT
COMPARATOR
THRESHOLD VOLTAGE
480
490
LATCH
495
LEAD OFF
ECG ANALOG FRONT END
400

FIG. 1A

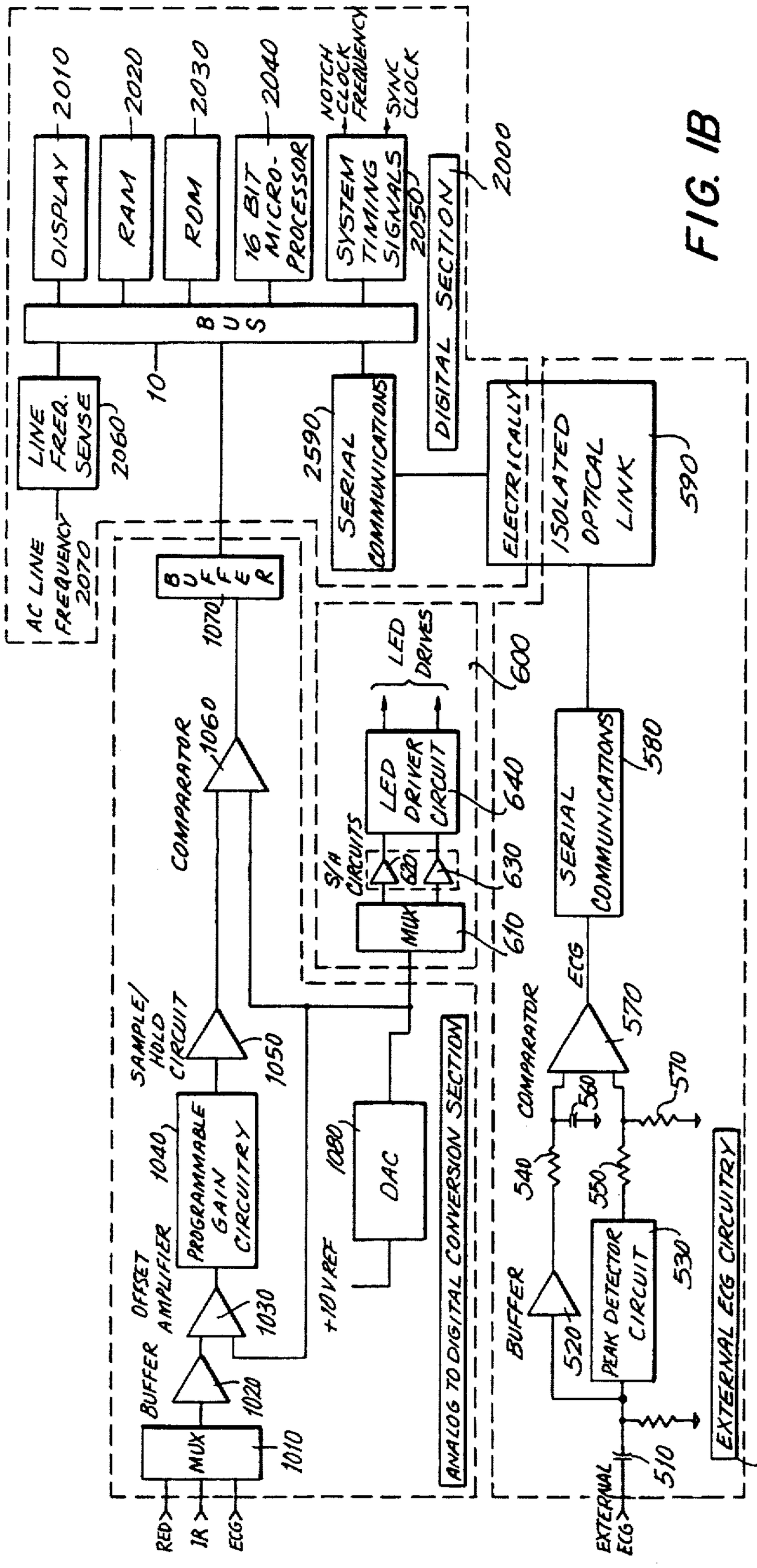

DISPLAY 2010
RAM 2020
ROM 2030
16 BIT MICRO-PROCESSOR 2040
NOTCH CLOCK FREQUENCY
SYNC CLOCK
SYSTEM TIMING SIGNALS 2050
SECTION 2000
BUS
LINE FREQ. SENSE 2060
10
AC LINE FREQUENCY 2070
BUFFER 1070
SERIAL COMMUNICATIONS 2590
DIGITAL
ELECTRICALLY ISOLATED OPTICAL LINK 590
LED DRIVES
600
COMPARATOR 1060
S/H CIRCUITS
LED DRIVER CIRCUIT 640
620
630
MUX 610
SERIAL COMMUNICATIONS 580
SAMPLE/ HOLD CIRCUIT 1050
1040
PROGRAMMABLE GAIN CIRCUITRY
1080
DAC
ANALOG TO DIGITAL CONVERSION SECTION
ECG
COMPARATOR 570
560
570
540
550
OFFSET AMPLIFIER 1030
BUFFER 1020
MUX 1010
+10 V REF
BUFFER 520
PEAK DETECTOR CIRCUIT 530
EXTERNAL ECG CIRCUITRY
500
RED
IR
ECG
EXTERNAL ECG
510

FIG. 1B

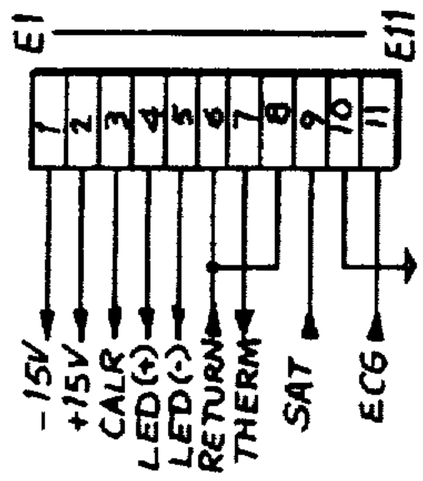

J1
1 2 3 4 5 6 7 8 9
-15V
+15V
CALR
LED(+)
LED(-)
THERM
DETECTOR RETURN
DETECTOR SHIELD
C9 0.1uF
C10 0.1uF

FIG. 2A

E1
E11
1 2 3 4 5 6 7 8 9 10 11
-15V
+15V
CALR
LED(+)
LED(-)
RETURN
THERM
SAT
ECG
SAT
+15V
LM317L
VR1
R7 2.67K
R6 464Ω
C4 1.0uF
C8 10pF
R13 249K
U2 LF442
DETECTOR
DETECTOR SHIELD
R8 1K
C5 0.22uF
SaO2 PREAMP
210
C3* 0.1uF
DS1
R12 2M
C11 0.01uF
R11 200K
U2 LF442
CR6 IN4148
CR5 IN4148
230
ECG
R9 2K
R10 2K
C7 0.1uF N/C
U1 101HP
R5 1.33K
C6 0.1uF
ECG PREAMP
220
RLD
250
CR2 IN4148
CR1 IN4148
CR4 IN4148
CR3 IN4148
R4 100M
R3 100M
R2 100K
R1 100K
DS3
DS2
C1* 0.1uF
C2* 0.1uF
RA
LA
INPUT PROTECTION CIRCUITRY
250

FIG. 2B

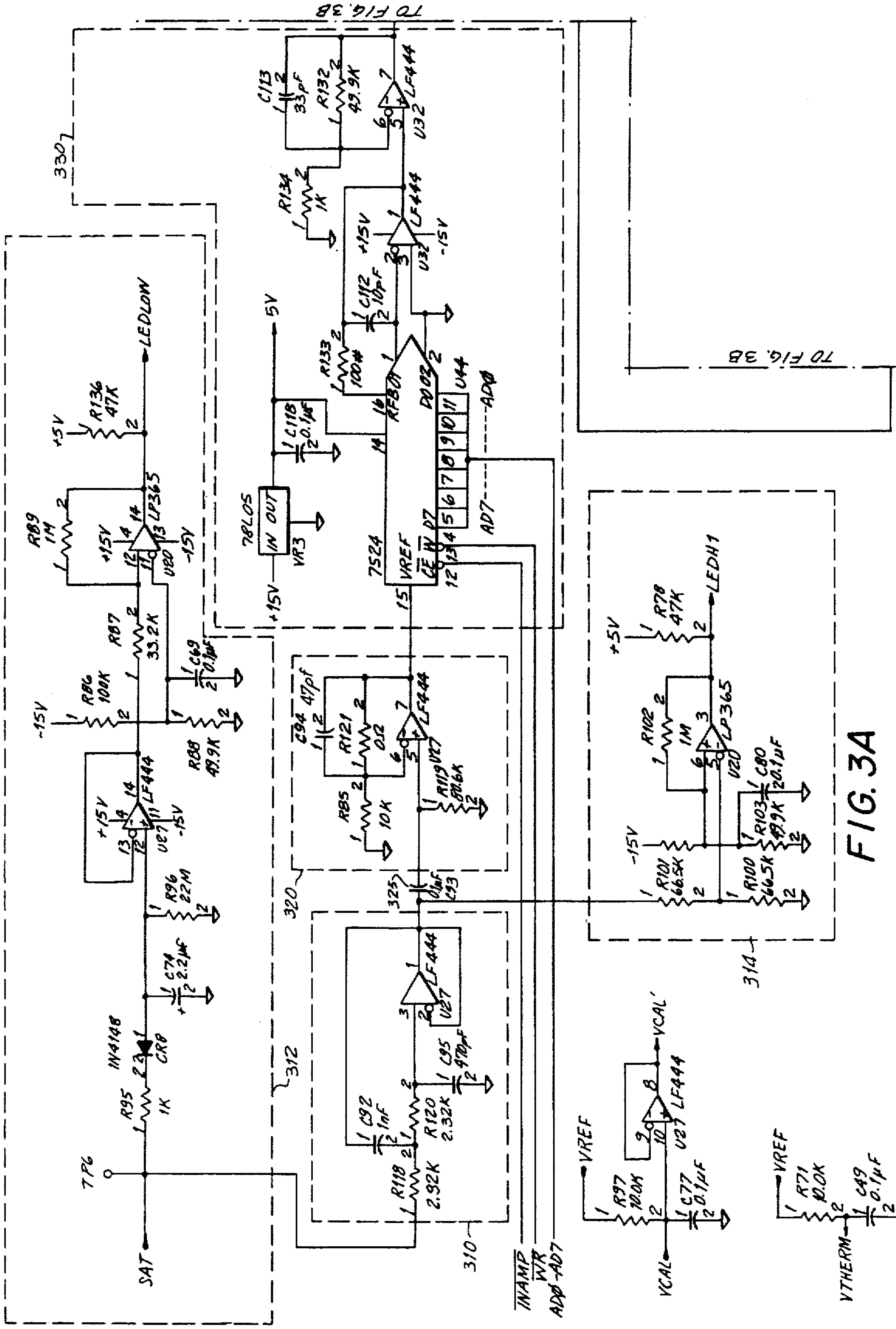
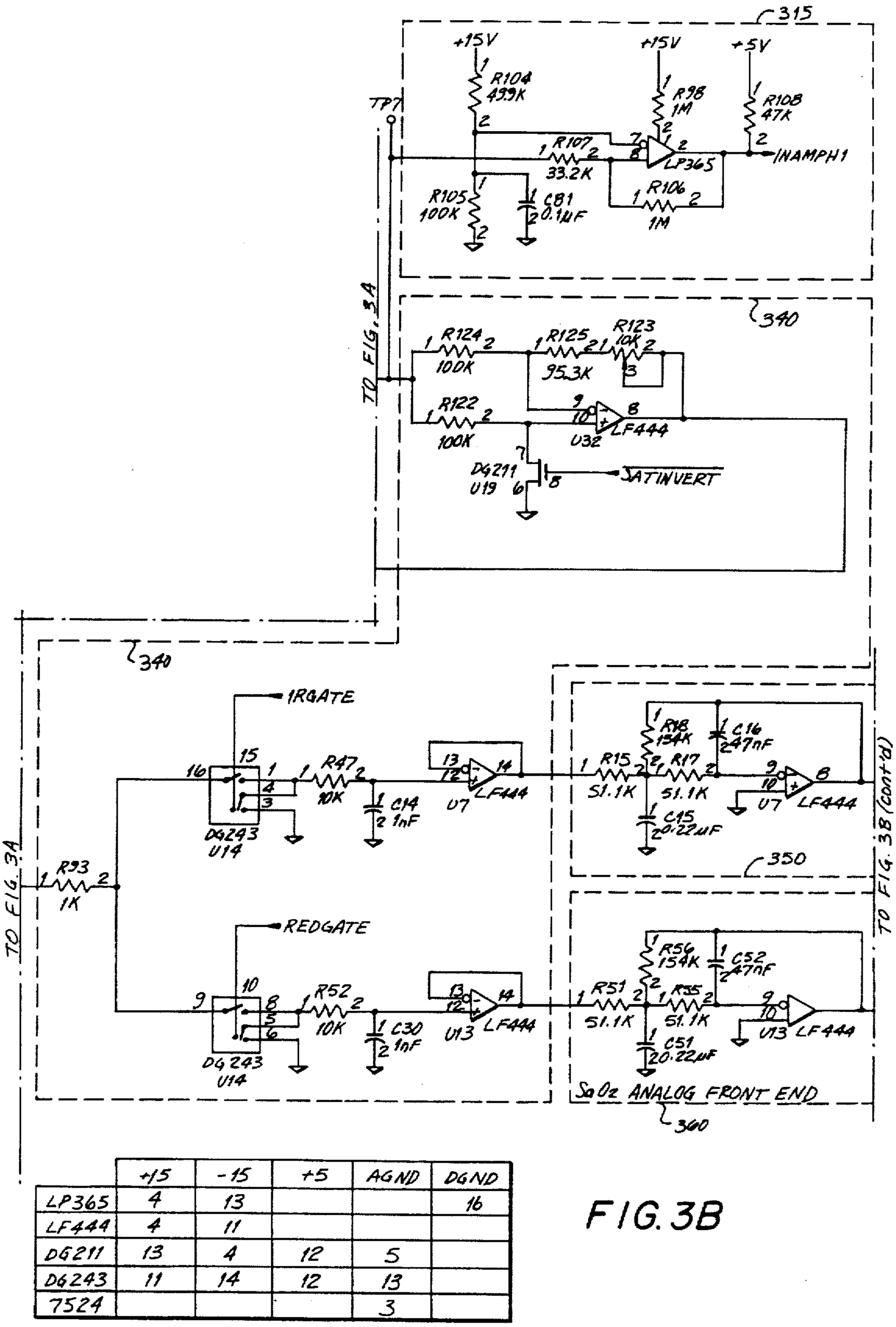

TO FIG. 3B
TO FIG. 3B
330
312
320
325
310
314
C113 33pF
R132 49.9K
LF444
U32
R134 1K
+15V
-15V
C112 10pF
R133 100#
5V
C118 0.1µF
78L05
IN OUT
VR3
7524
VREF
CE
WR
RFB
D0
D7
U44
AD0
AD7
LEDLOW
R136 47K
+5V
R89 1M
LP365
U20
R87 33.2K
R86 100K
C69 0.1µF
R88 49.9K
U27
R96 22M
C74 2.2µF
1N4148
CR8
R95 1K
TP6
SAT
C94 47pf
R121
R85 10K
R119 80.6K
C93 0.1µF
C92 1nF
R120 2.32K
R118 2.92K
C95 470pF
LEDH1
R78 47K
R102 1M
C80 0.1µF
R103 49.9K
R101 66.5K
R100 66.5K
INAMP
WR
AD0-AD7
VREF
R97 10.0K
C77 0.1µF
VCAL
VCAL'
R71 10.0K
C49 0.1µF
VTHERM

FIG. 3A



	+15	-15	+5	AGND	DGND
LP365	4	13			16
LF444	4	11			
DG211	13	4	12	5	
DG243	11	14	12	13	
7524				3	

FIG. 3B

FIG. 3B(cont'd)

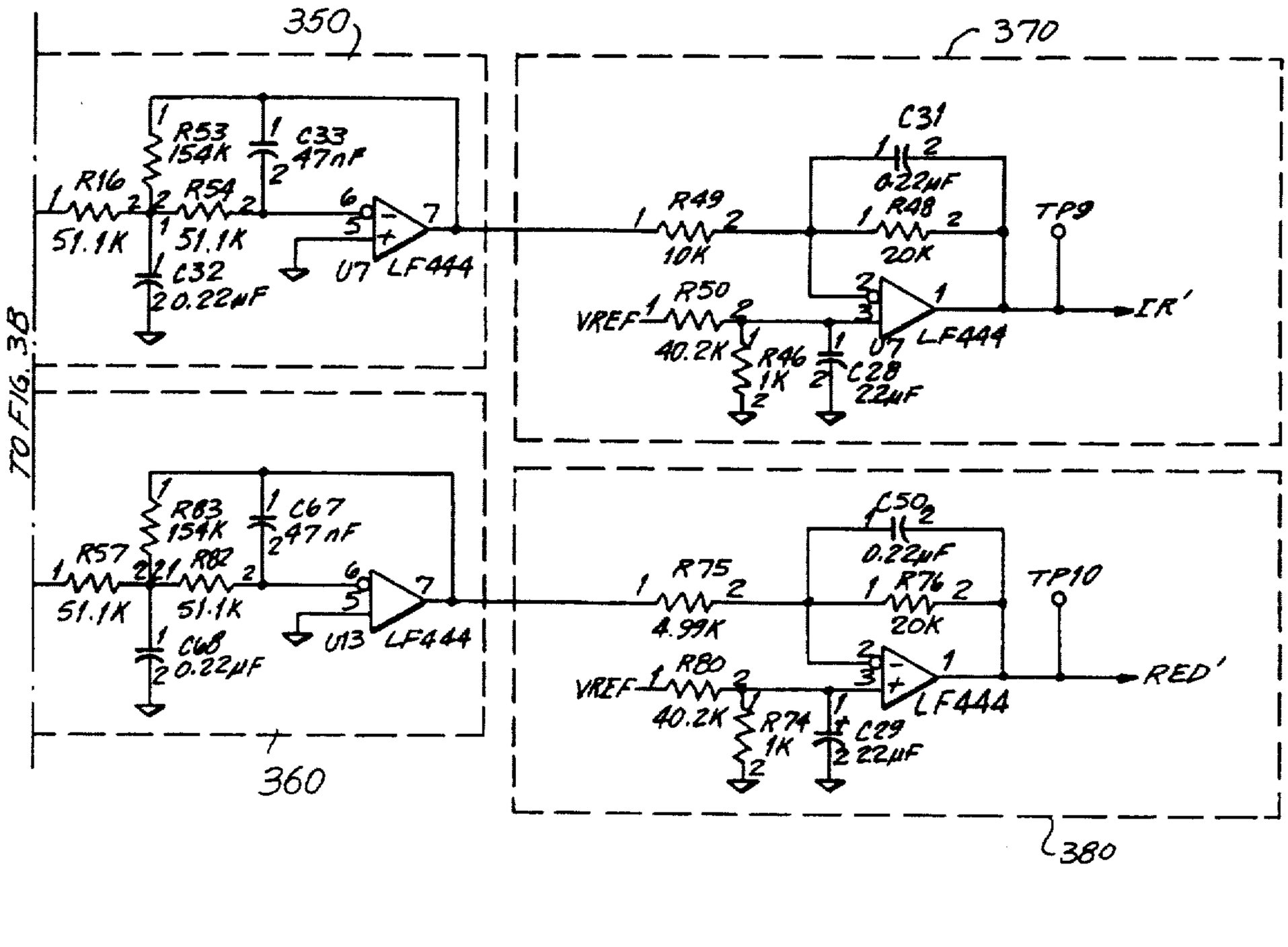
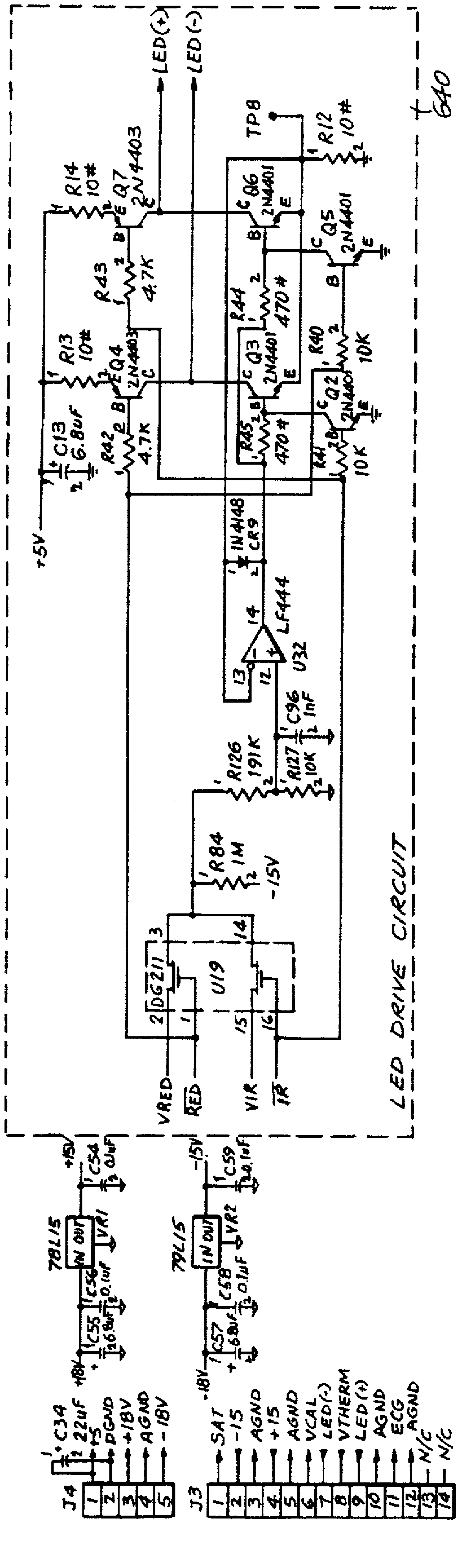

350
370
TO FIG. 3B
R53 154K
C33 47nF
R16 51.1K
R54 51.1K
C32 0.22μF
U7 LF444
C31 0.22μF
R49 10K
R48 20K
TP9
R50 40.2K
R46 1K
C28 22μF
VREF
IR'
R83 154K
C67 47nF
R57 51.1K
R82 51.1K
C68 0.22μF
U13 LF444
C50 0.22μF
R75 4.99K
R76 20K
TP10
R80 40.2K
R74 1K
C29 22μF
RED'
360
380

FIG. 4


LED(+)
LED(-)
TP8
R12 10#
R14 10#
Q7 2N4403
Q6 2N4401
Q5 2N4401
R43 4.7K
R44 470#
R13 10#
Q4 2N4403
Q3 2N4401
R40 10K
C13 6.8uF
R42 4.7K
R45 470#
Q2 2N4401
R41 10K
+5V
1N4148 CR9
LF444
U32
C96 1nF
R126 191K
R127 10K
R84 1M
-15V
DG211
U19
VRED
RED
VIR
IR
LED DRIVE CIRCUIT
640
+15V
C54 0.1uF
78L15
IN OUT
VR1
C56 0.1uF
C55 6.8uF
+18V
-15V
C59 0.1uF
79L15
VR2
C58 0.1uF
C57 6.8uF
-18V
J4
C34 22uF
+5
DGND
+18V
AGND
-18V
J3
SAT
-15
AGND
+15
AGND
VCAL
LED(-)
VTHERM
LED(+)
AGND
ECG
AGND
N/C
N/C

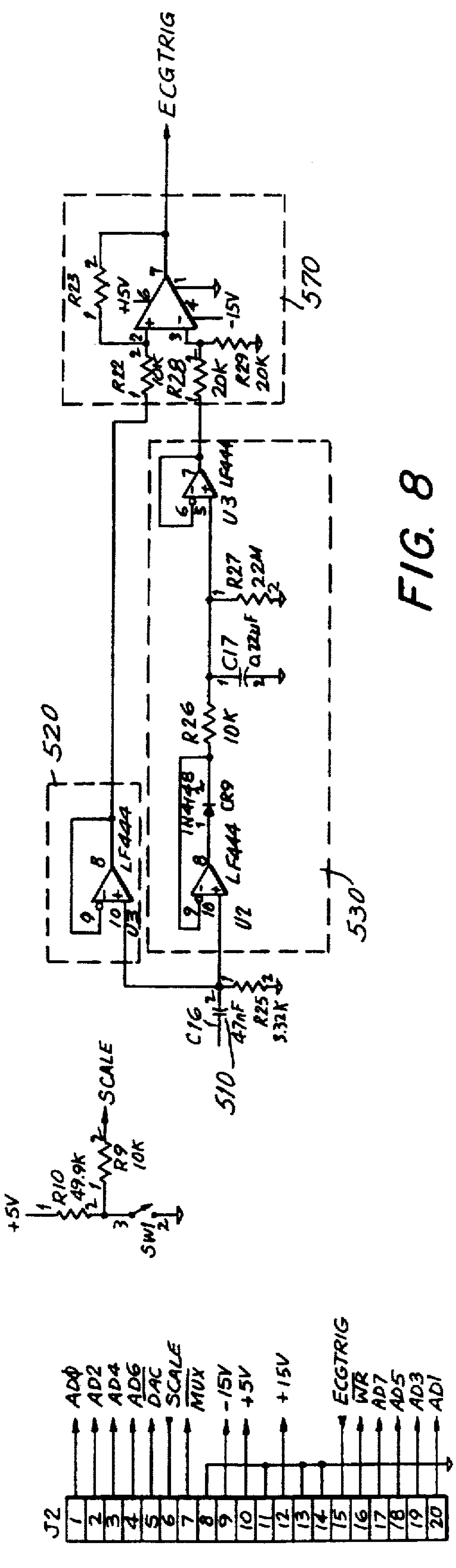

ECGTRIG
R23
+15V
-15V
570
R22
R28 20K
R29 20K
U3 LF444
R27 22M
C17 0.22uF
R26 10K
1N4148 CR9
LF444
U2
530
520
LF444
U3
C16 47nF
R25 3.32K
510
SCALE
R10 49.9K
R9 10K
+5V
SW1
J2
AD0
AD2
AD4
AD6
DAC
SCALE
MUX
-15V
+5V
+15V
ECGTRIG
WR
AD7
AD5
AD3
AD1

FIG. 8

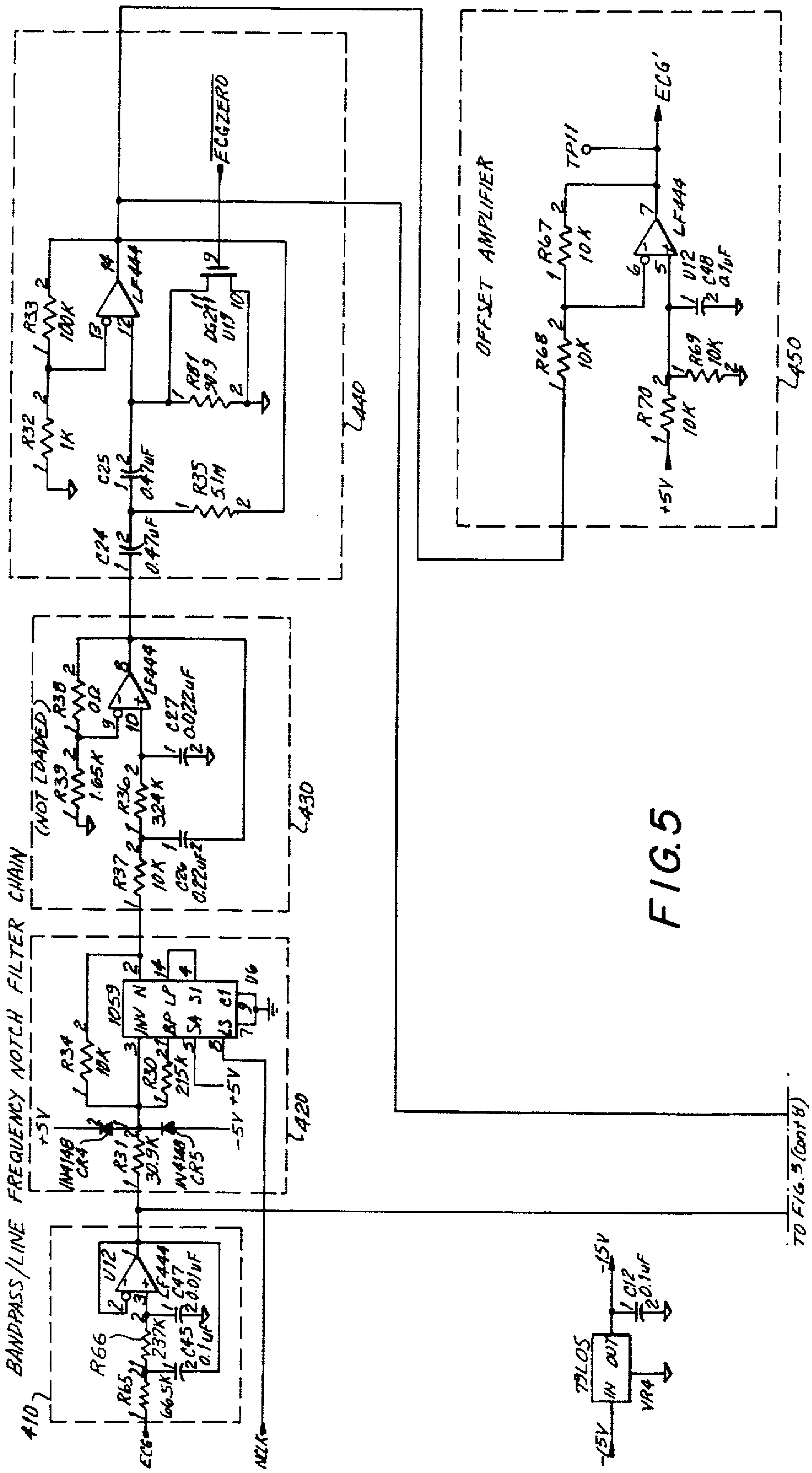
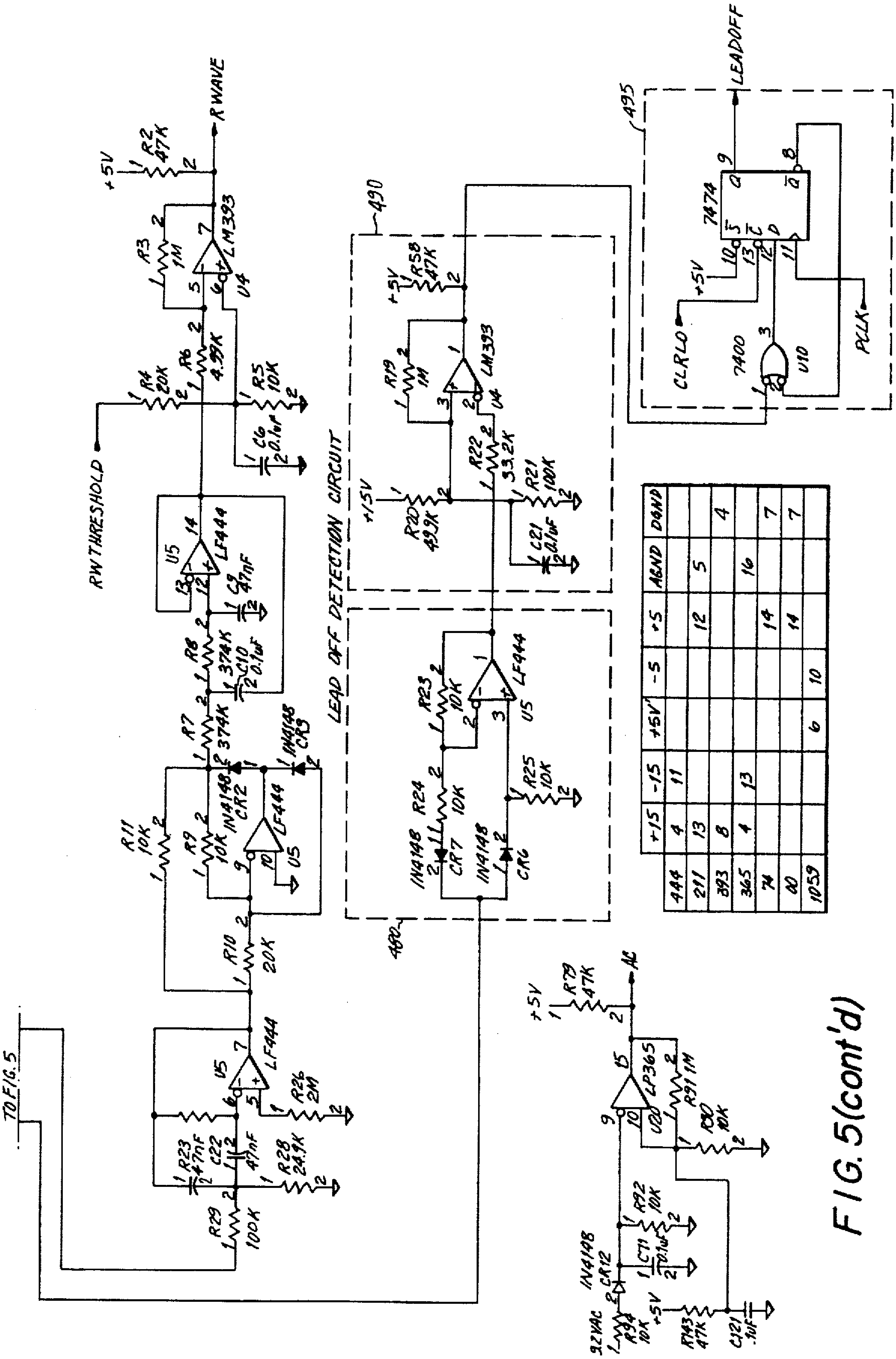

BANDPASS/LINE FREQUENCY NOTCH FILTER CHAIN
OFFSET AMPLIFIER
410
420
430
440
450
ECG
NCLK
ECGZERO
ECG'
TP11
R65 66.5K
R66 237K
C45 0.1uF
C47 0.01uF
U12 LF444
R31 30.9K
1N4148 CR4
1N4148 CR5
R34 10K
R30 215K
1059
U6
R37 10K
C26 0.22uF
R36 324K
C27 0.022uF
R39 1.65K
R38 0Ω
(NOT LOADED)
C24 0.47uF
C25 0.47uF
R35 5.1M
R32 1K
R33 100K
R81 90.9
DG211
U19
R67 10K
R68 10K
R69 10K
R70 10K
C48 0.1uF
+5V
-5V
-15V
C12 0.1uF
79L05
VR4
TO FIG. 5 (Cont'd)

FIG. 5



	+15	-15	+5V'	-5	+5	AGND	DGND
444	4	11					
211	13				12	5	
393	8						4
365	4	13				16	
74					14		7
00					14		7
1059			6	10			

FIG. 5(cont'd)

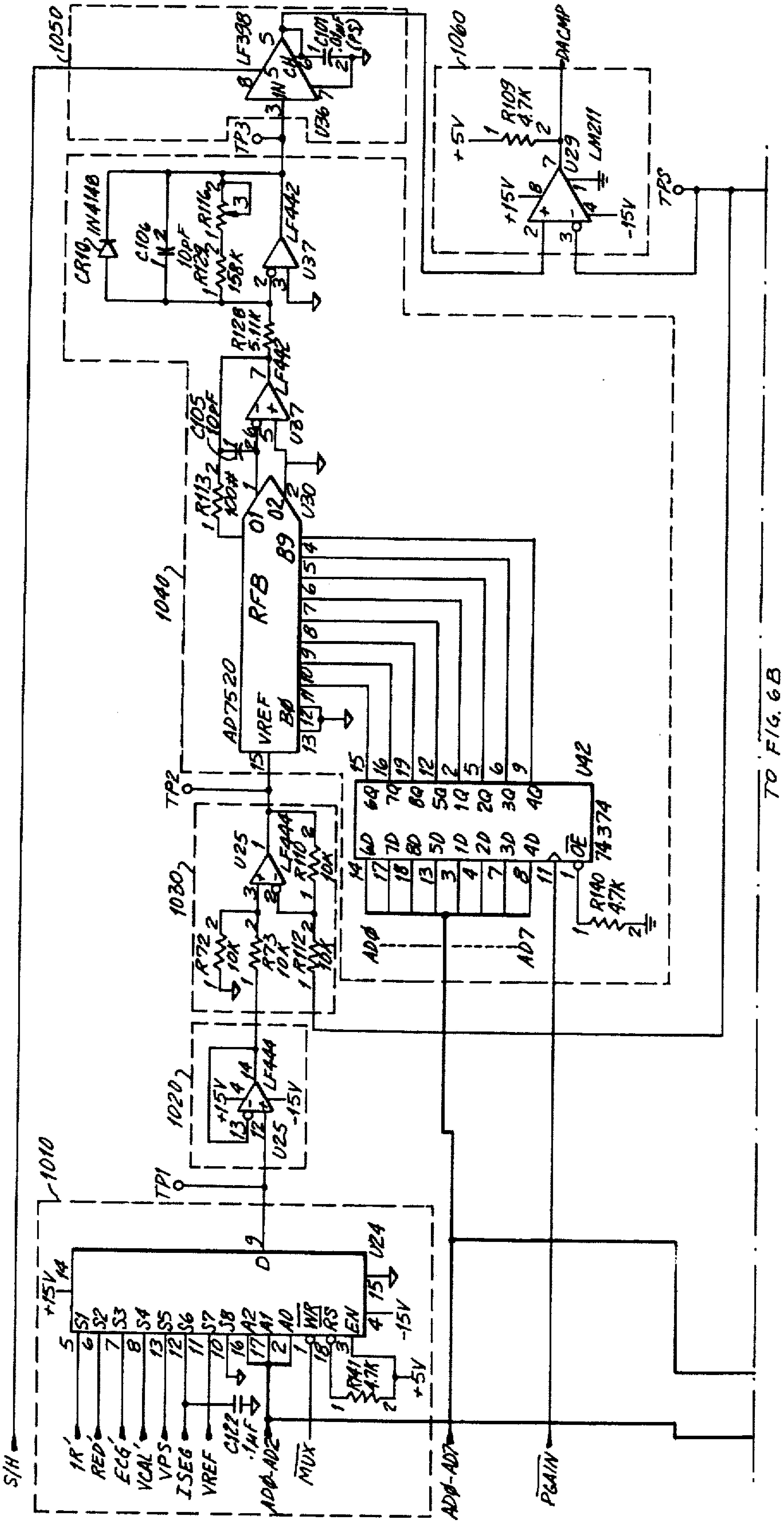
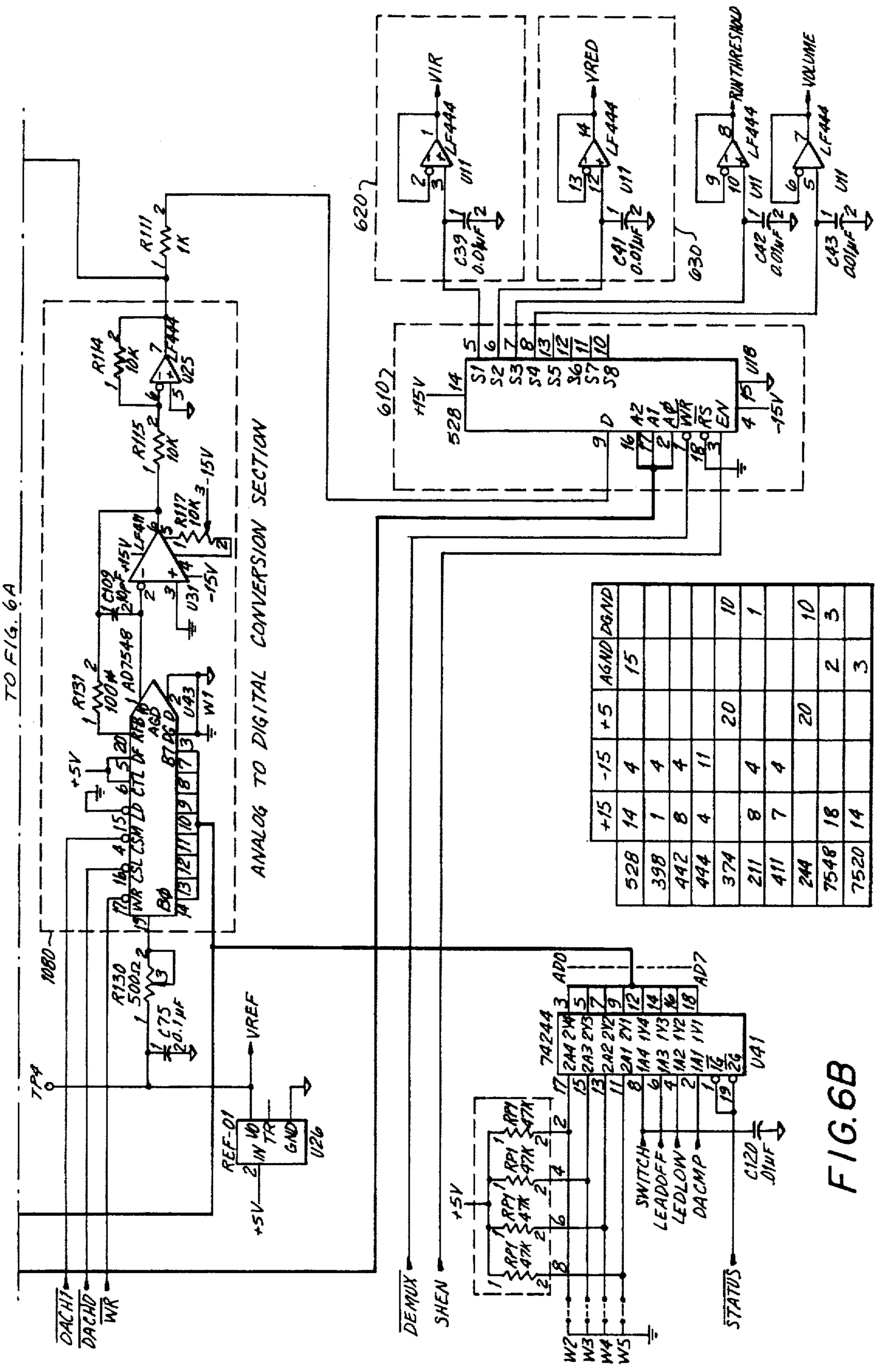

1050
LF398
U36
TP3
1060
DACMP
R109
4.7K
+5V
U29
LM211
+15V
-15V
TPS
CR10
IN4148
C106
10pF
R124
R116
158K
LF442
U37
R128
5.11K
C105
10pF
R113
U30
1040
AD7520
VREF
RFB
TP2
U25
LF444
1030
R72
10K
R73
R110
R112
U42
74374
R140
4.7K
AD0
AD7
1020
TP1
1010
U24
+15V
S/H
1R'
RED'
ECG'
VCAL'
VPS
ISEG
VREF
C122
.1µF
AD0-AD2
MUX
R141
4.7K
AD0-AD7
PGAIN
TO FIG. 6B

FIG. 6A



	+15	-15	+5	AGND	DGND
528	14	4		15	
398	1	4			
442	8	4			
444	4	11			
374			20		10
211	8	4			1
411	7	4			
244			20		10
7548	18			2	3
7520	14			3	

FIG. 6B

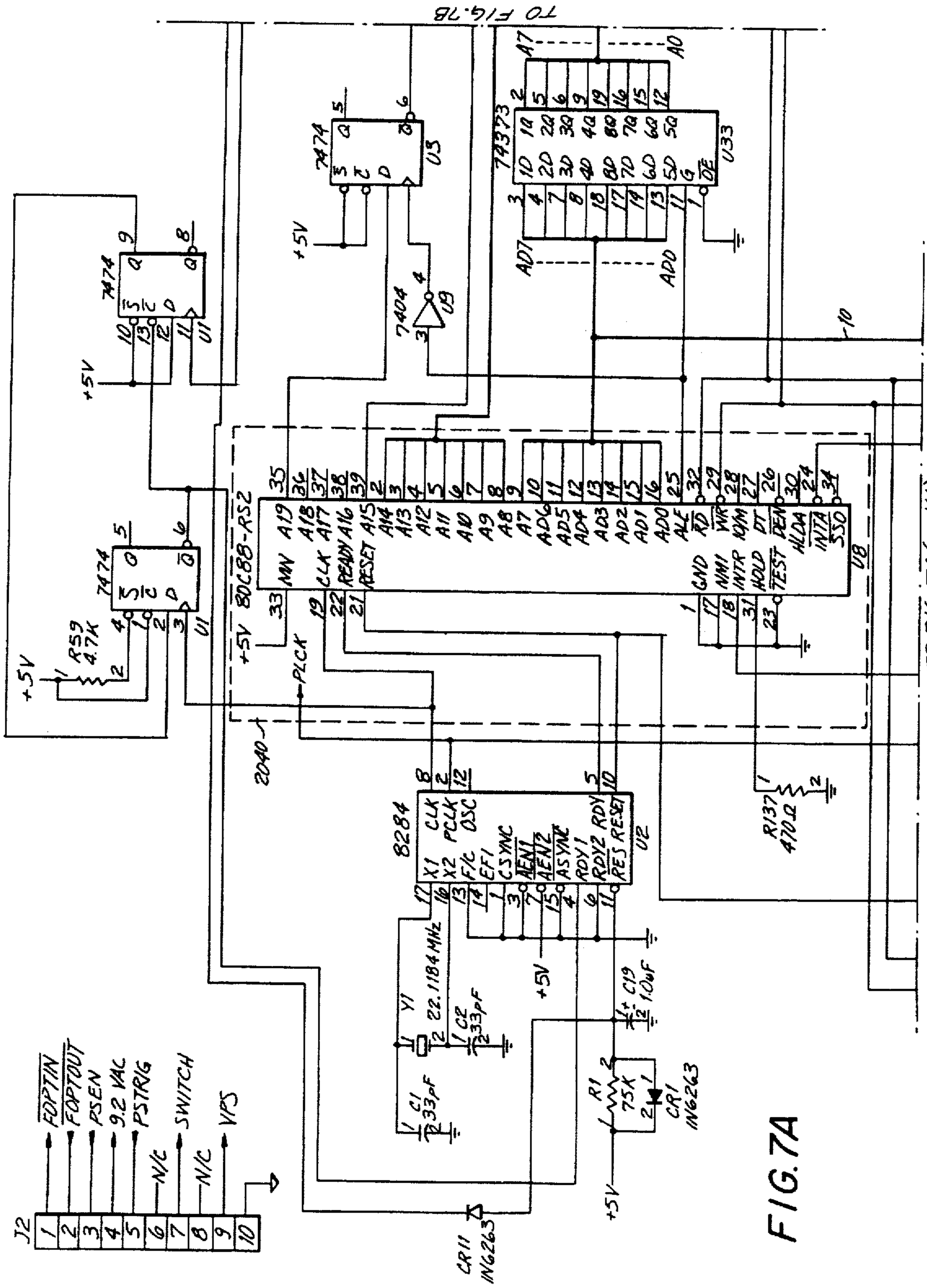

TO FIG. 7B
TO FIG. 7A (cont'd)
+5V
RS9 4.7K
7474
7404
74373
8DC88-RS2
8284
Y1 22.1184 MHz
C1 33pF
C2 33pF
C19 1.0μF
R137 470Ω
R1 75K
CR1 IN6263
CR11 IN6263
2040
PCLK
J2
FOPTIN
FOPTOUT
PSEN
9.2 VAC
PSTRIG
N/C
SWITCH
N/C
VPS

FIG. 7A

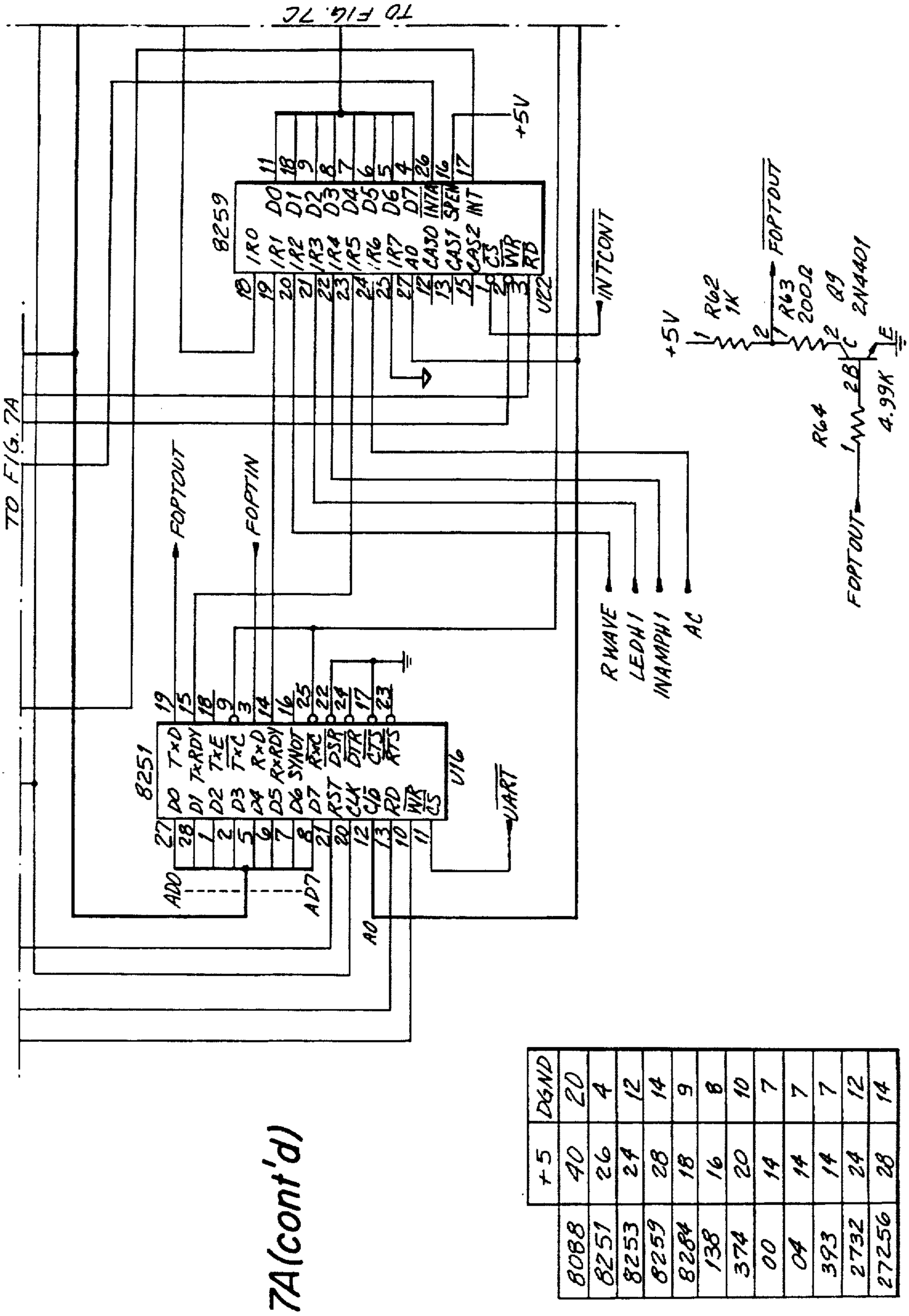
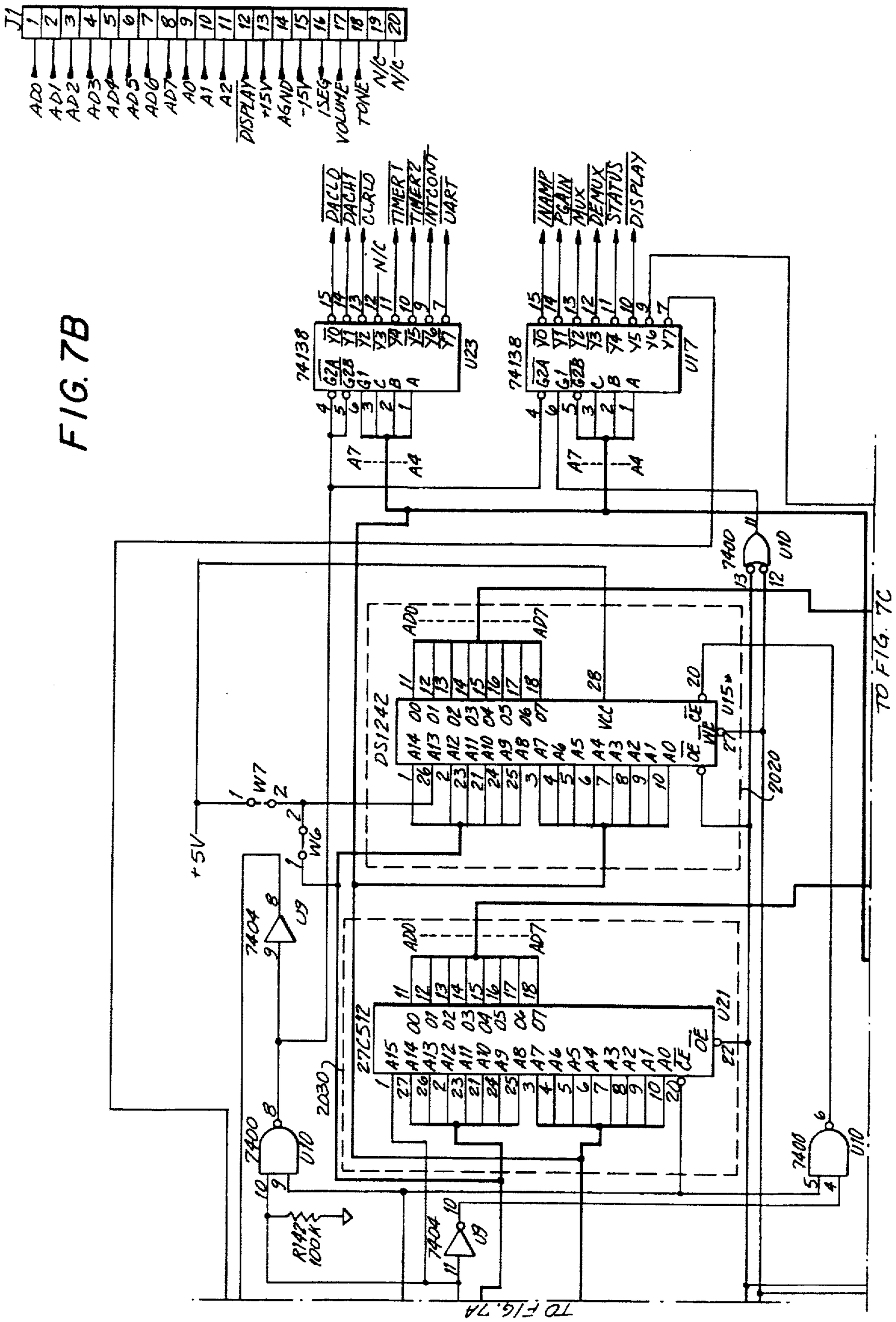


FIG. 7A(cont'd)

	+5	DGND
8088	40	20
8251	26	4
8253	24	12
8259	28	14
8284	18	9
138	16	8
374	20	10
00	14	7
04	14	7
393	14	7
2732	24	12
27256	28	14

FIG. 7B


J1
AD0
AD1
AD2
AD3
AD4
AD5
AD6
AD7
A0
A1
A2
DISPLAY
+15V
AGND
-15V
ISEG
VOLUME
TONE
N/C
N/C
74138
U23
DACLD
DACHT
CLRLD
N/C
TIMER1
TIMER2
INTCONT
UART
74138
U17
INAMP
PGAIN
MUX
DEMUX
STATUS
DISPLAY
A7
A4
7400
U10
DS1242
2020
VCC
U15
+5V
W7
W6
7404
U9
DS2512
2030
R142
100K
TO FIG. 7C
TO FIG. 7A

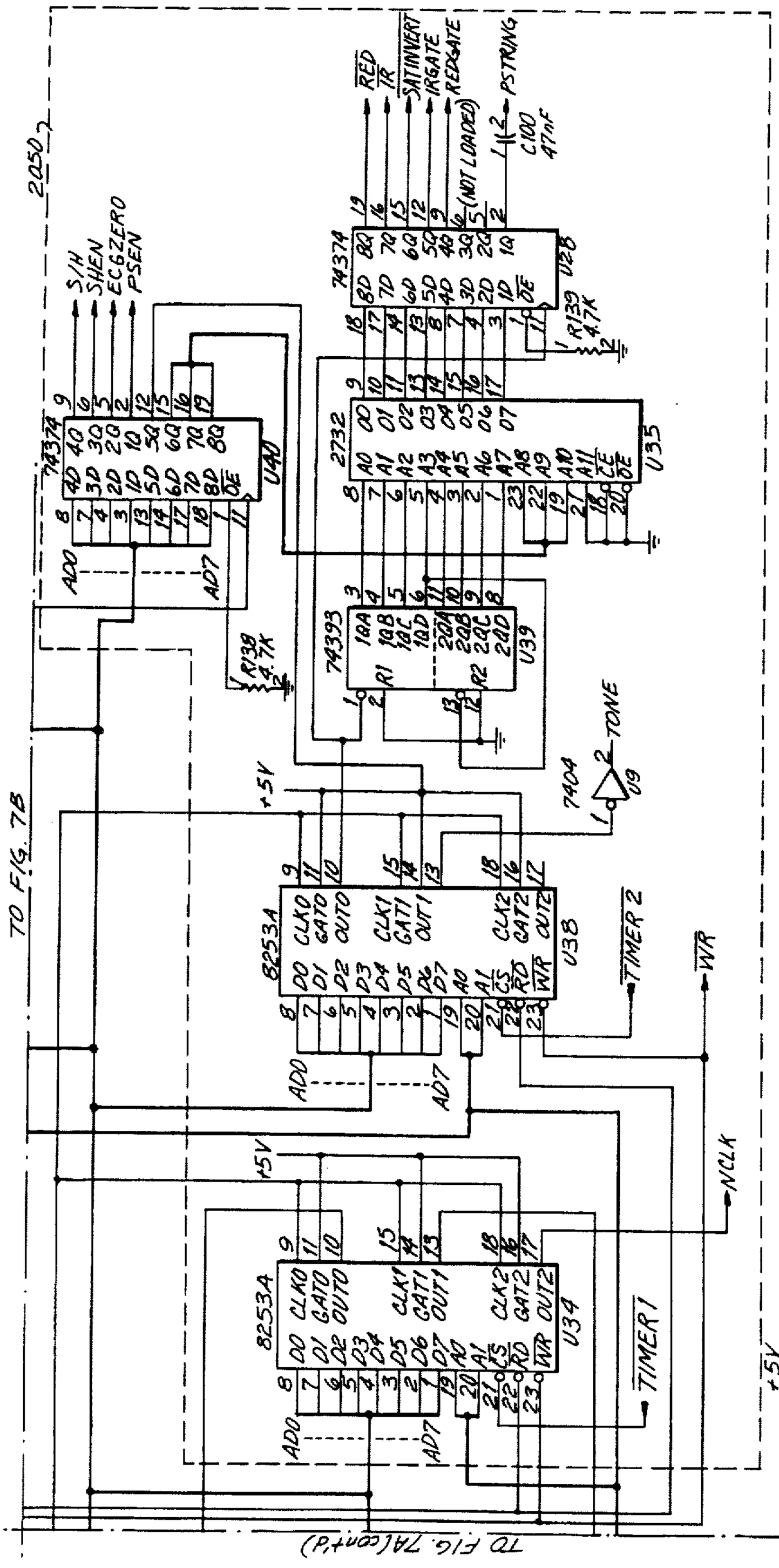

RED
IR
SATINVERT
IRGATE
REDGATE
(NOT LOADED)
PSTRING
C100
47nF
L12
2050_2
S/H
SHEN
ECGZERO
PSEN
74374
U40
U28
2732
U35
74393
U39
R139
4.7K
R1
R2
AD0
AD7
+5V
7404
U9
TO FIG. 7B
82534
U38
U34
TIMER 2
TIMER 1
WR
NCLK
FOPTIN
R60
1.54K
R61
10K
Q8
2N4401
TO FIG. 7A(cont'd)

FIG. 7C

FIG. 9a

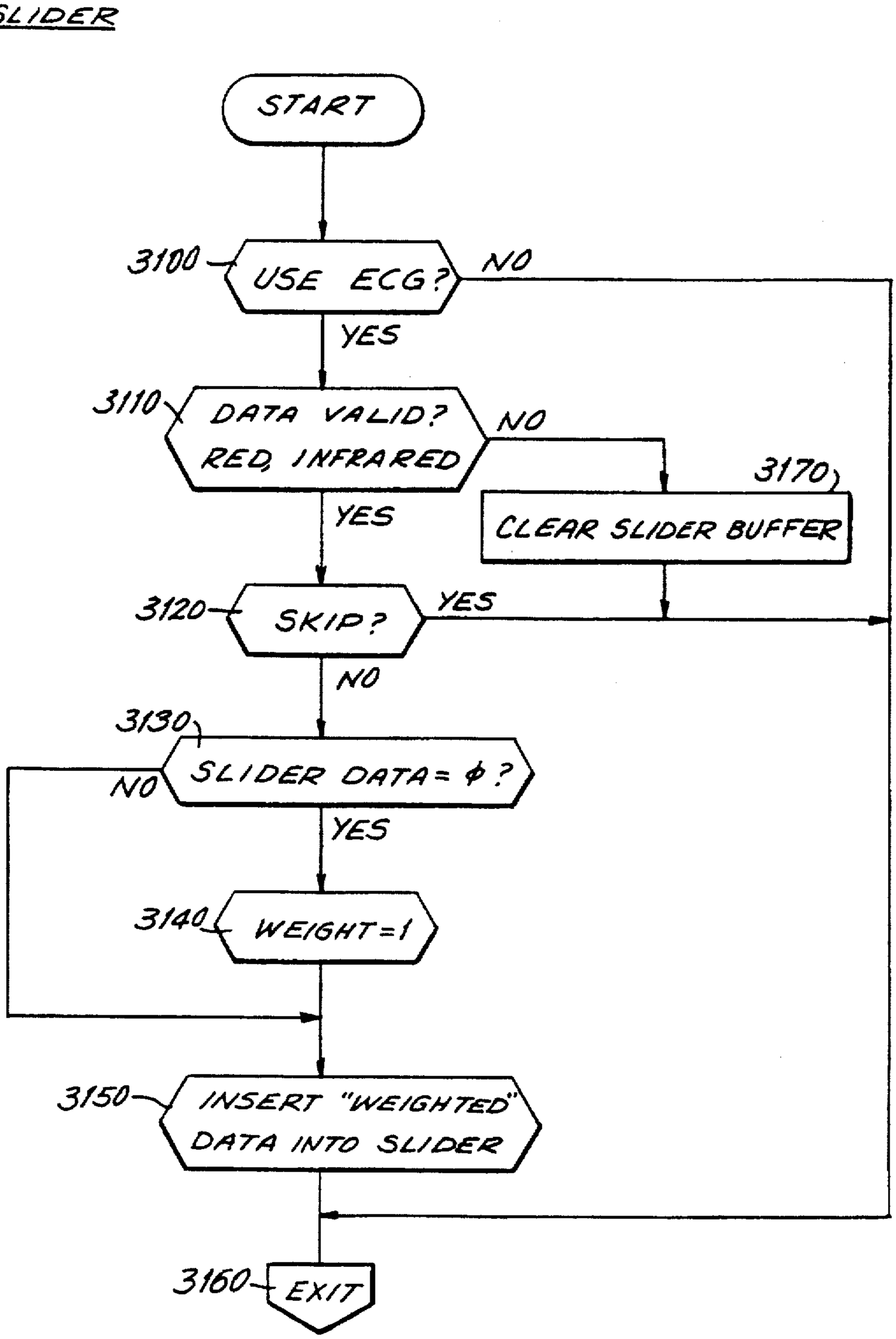
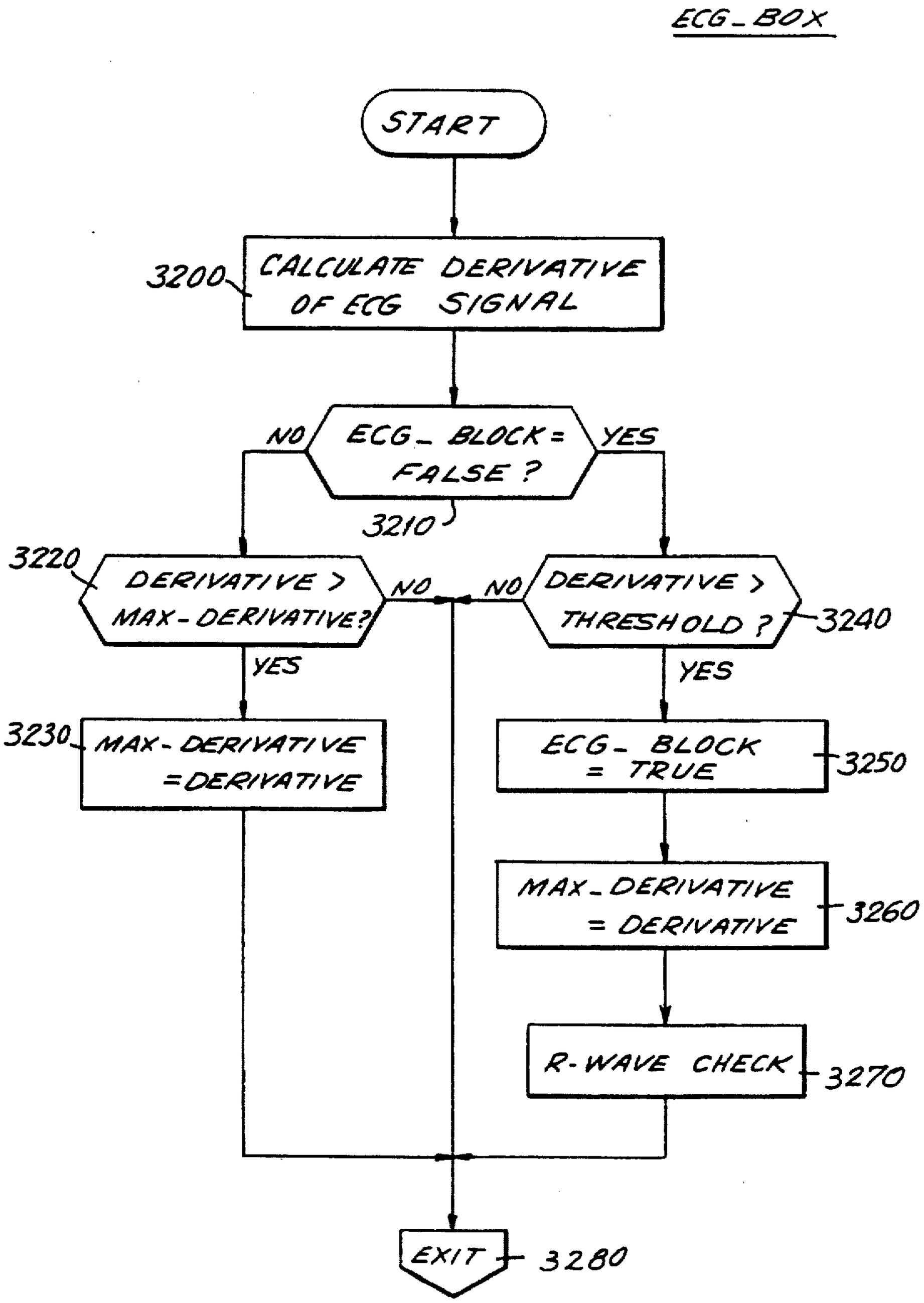

SLIDER
START
3100
USE ECG?
NO
YES
3110
DATA VALID?
RED, INFRARED
NO
YES
3170
CLEAR SLIDER BUFFER
3120
SKIP?
YES
NO
3130
SLIDER DATA = ϕ ?
NO
YES
3140
WEIGHT=1
3150
INSERT "WEIGHTED"
DATA INTO SLIDER
3160
EXIT

FIG. 9b


ECG_BOX
START
3200
CALCULATE DERIVATIVE OF ECG SIGNAL
NO
ECG_ BLOCK = FALSE ?
YES
3210
3220
DERIVATIVE > MAX-DERIVATIVE?
NO
NO
DERIVATIVE > THRESHOLD ?
3240
YES
YES
3230
MAX-DERIVATIVE =DERIVATIVE
ECG- BLOCK = TRUE
3250
MAX_DERIVATIVE = DERIVATIVE
3260
R-WAVE CHECK
3270
EXIT
3280

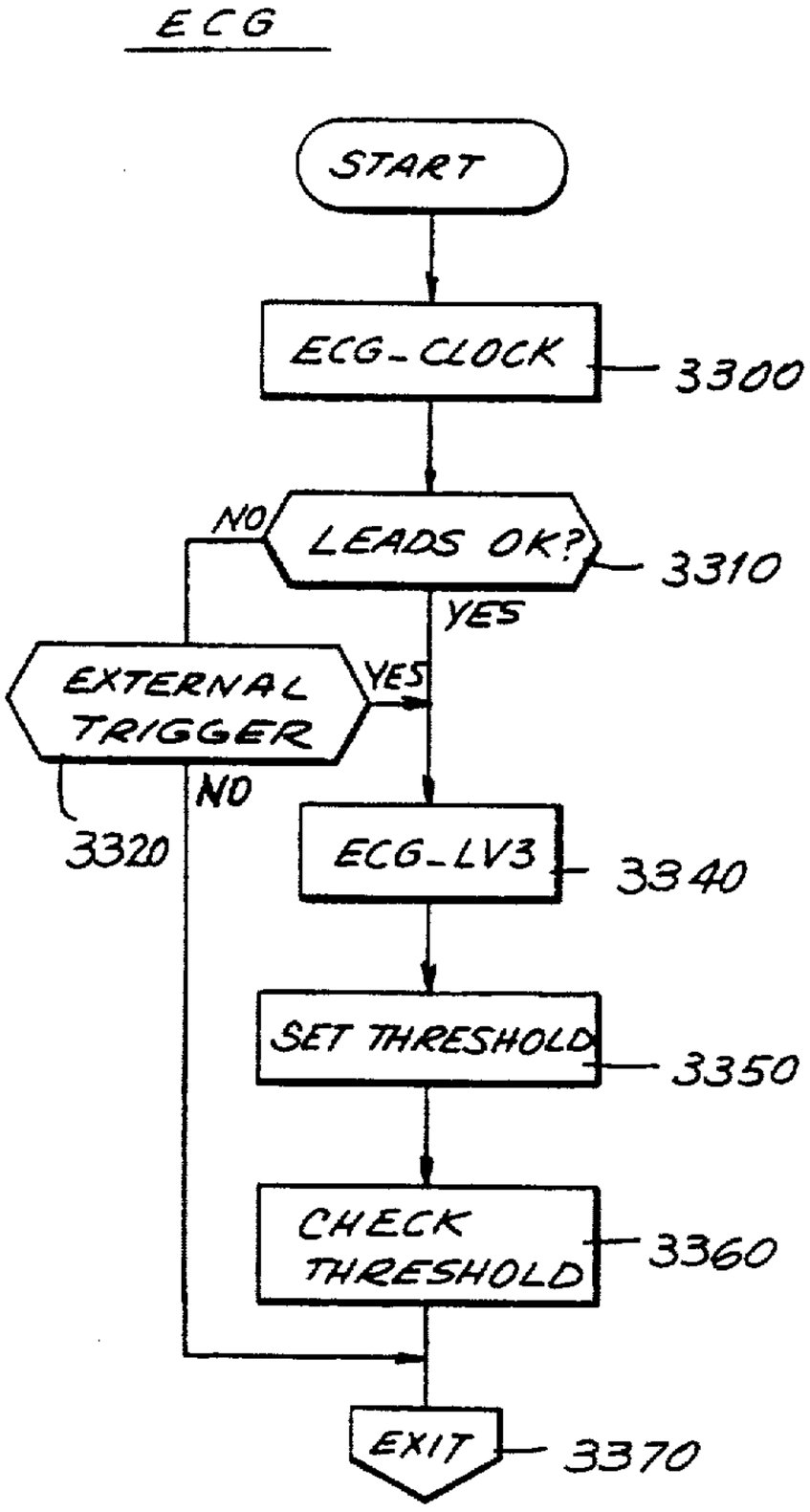

ECG
START
ECG-CLOCK
3300
NO
LEADS OK?
3310
YES
EXTERNAL TRIGGER
YES
NO
3320
ECG-LV3
3340
SET THRESHOLD
3350
CHECK THRESHOLD
3360
EXIT
3370

FIG. 9c

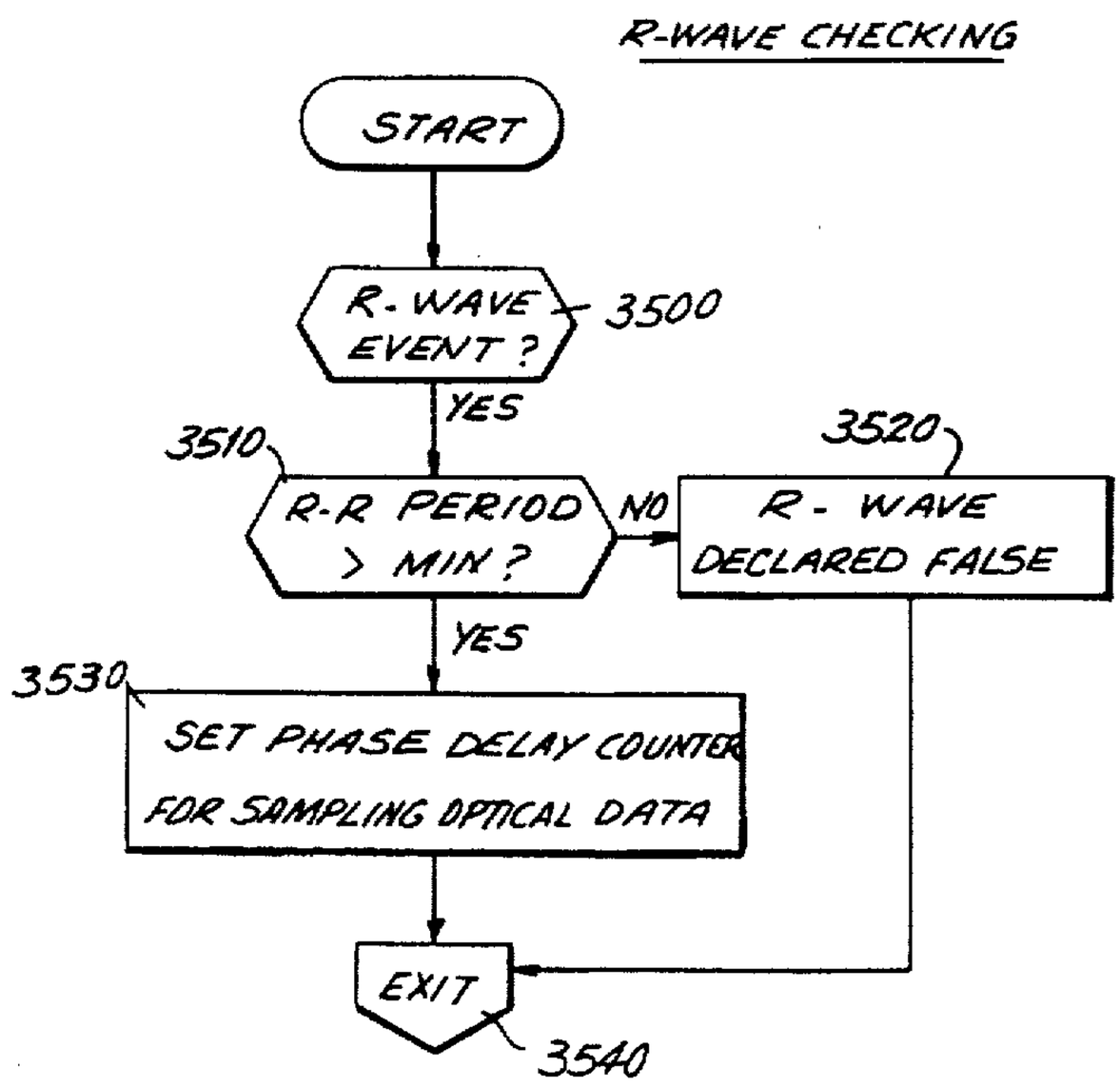

R-WAVE CHECKING
START
R-WAVE EVENT ?
3500
YES
3510
R-R PERIOD > MIN ?
NO
3520
R-WAVE DECLARED FALSE
YES
3530
SET PHASE DELAY COUNTER FOR SAMPLING OPTICAL DATA
EXIT
3540

FIG. 9e

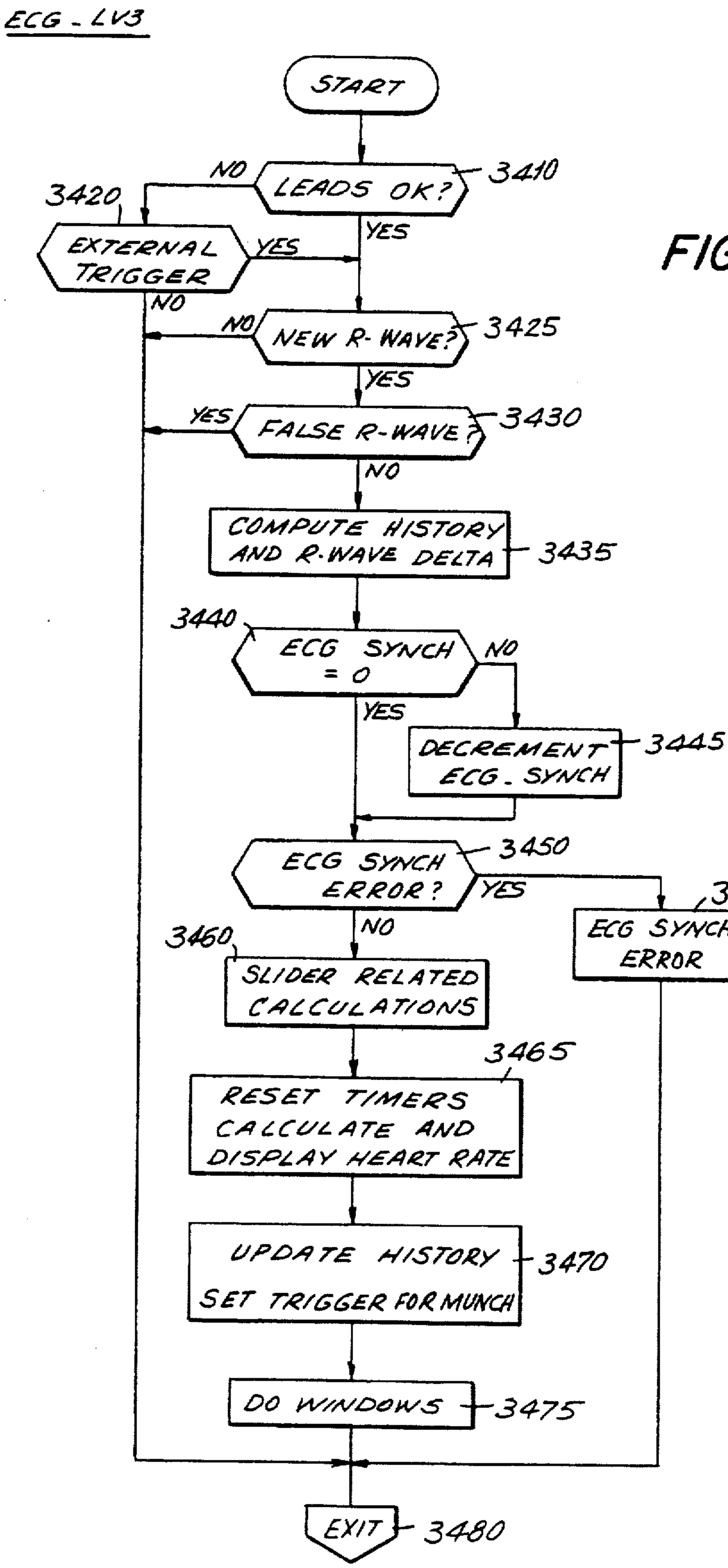
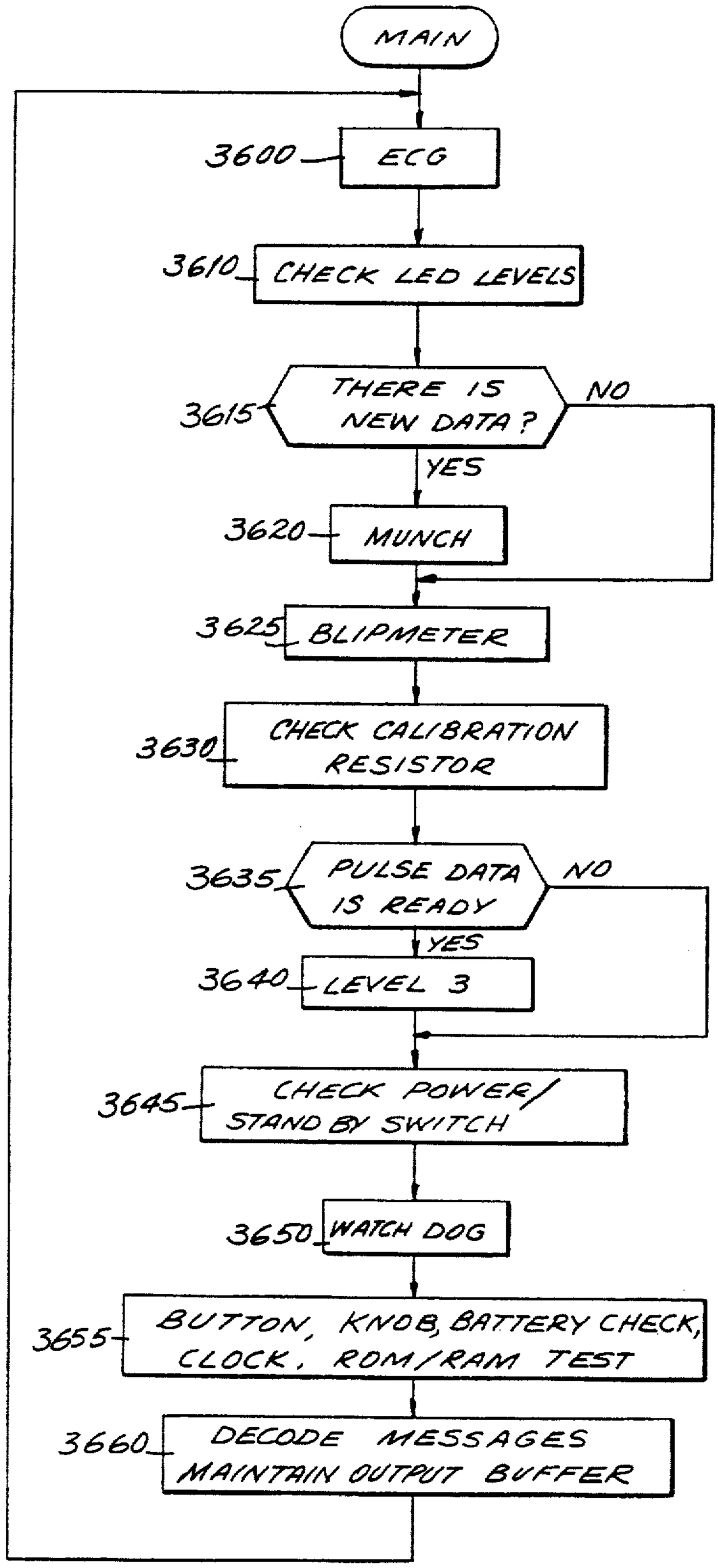
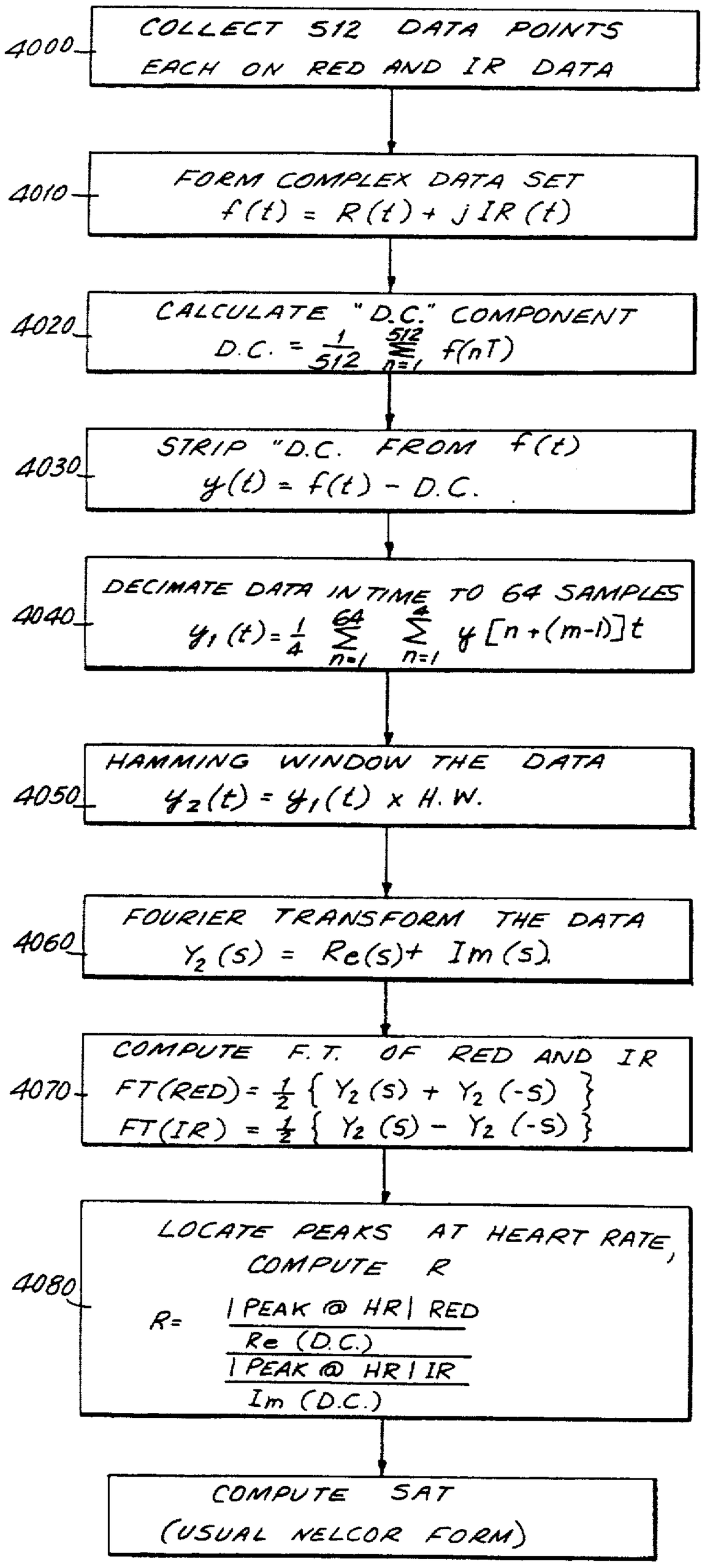

ECG - LV3
START
NO
LEADS OK?
3410
3420
YES
EXTERNAL TRIGGER
YES
NO
NO
NEW R-WAVE?
3425
YES
YES
FALSE R-WAVE?
3430
NO
COMPUTE HISTORY AND R-WAVE DELTA
3435
3440
ECG SYNCH = 0
NO
YES
DECREMENT ECG-SYNCH
3445
ECG SYNCH ERROR?
3450
YES
3455
ECG SYNCH ERROR
NO
3460
SLIDER RELATED CALCULATIONS
3465
RESET TIMERS CALCULATE AND DISPLAY HEART RATE
UPDATE HISTORY
SET TRIGGER FOR MUNCH
3470
DO WINDOWS
3475
EXIT
3480

FIG. 9d

FIG. 9f


MAIN LOOP PROCESSING
MAIN
3600
ECG
3610
CHECK LED LEVELS
3615
THERE IS NEW DATA ?
NO
YES
3620
MUNCH
3625
BLIPMETER
3630
CHECK CALIBRATION RESISTOR
3635
PULSE DATA IS READY
NO
YES
3640
LEVEL 3
3645
CHECK POWER/ STAND BY SWITCH
3650
WATCH DOG
3655
BUTTON, KNOB, BATTERY CHECK, CLOCK, ROM/RAM TEST
3660
DECODE MESSAGES MAINTAIN OUTPUT BUFFER

FOURIER OXIMETER FLOWCHART FIG. 10



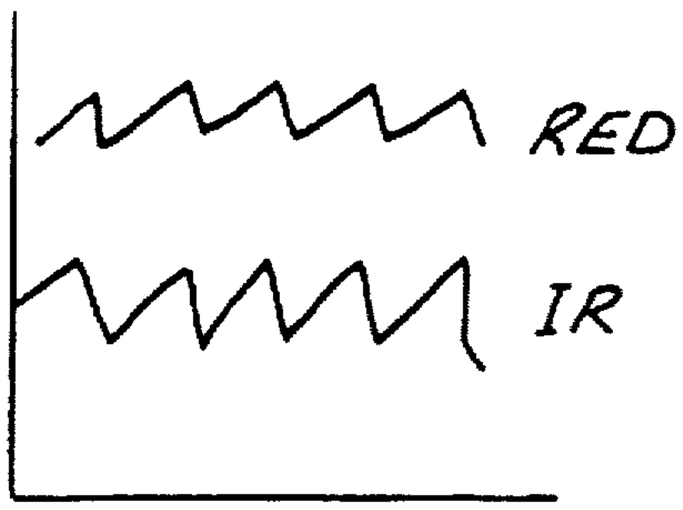


FIG. 10a

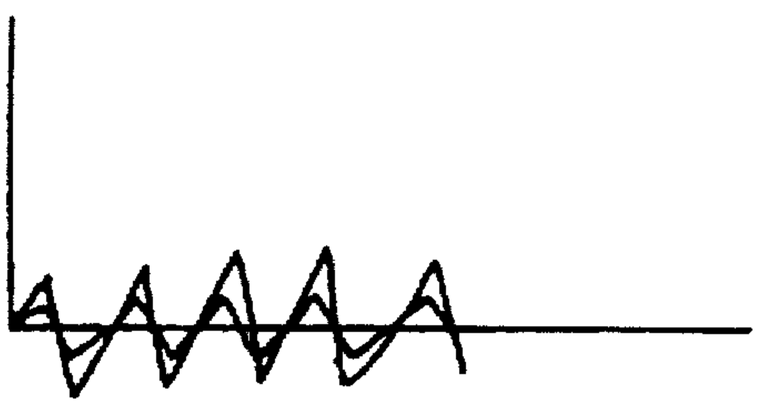


FIG. 10b

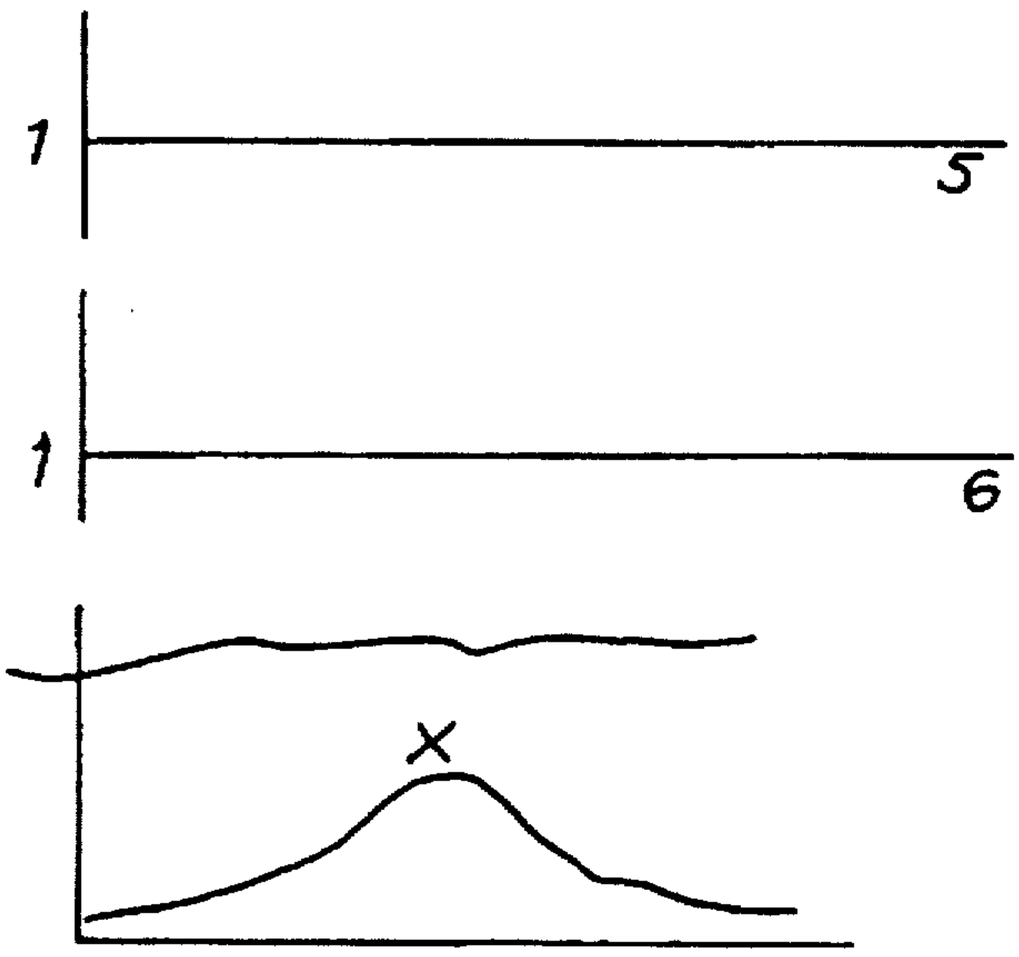


FIG. 10c

FIG. 10d

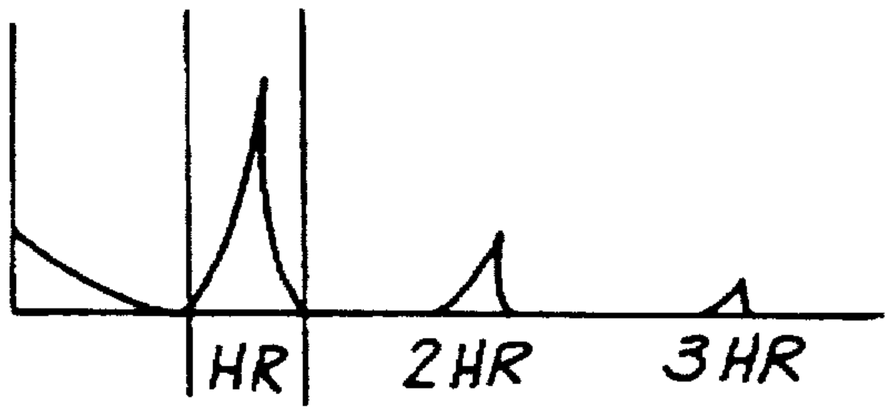


FIG. 10e

METHOD AND APPARATUS FOR DETECTING OPTICAL PULSES

Matter enclosed in heavy brackets [ ] appears in the original patent but forms no part of this reissue specification; matter printed in italics indicates the additions made by reissue.

CROSS REFERENCE TO RELATED APPLICATIONS

This application is a continuation-in-part application of and commonly assigned U.S. application Ser. No. 742,720, now U.S. Pat. No. 4,802,486, entitled Improved Method and Apparatus For Detecting Optical Pulses, filed Jun. 7, 1985 in the names of James E. Corenman and David E. Goodman, which is a continuation of U.S. application Ser. No. 718,525, entitled Improved Method and Apparatus For Detecting Optical Pulses, filed Apr. 1, 1985 in the names of James E. Corenman and David E. Goodman, now abandoned.

This invention relates to non-invasive pulse oximetry and specifically to an improvement on the method and apparatus for photoelectric determination of blood constituents disclosed in U.S. applications Ser. No. 742,720 and 718,525. This specification is accompanied by software appendices A and B.

BACKGROUND OF THE INVENTION

Non-invasive photoelectric pulse oximetry has been previously described in U.S. Pat. Nos. 4,407,290, 4,266,554, 4,086,915, 3,998,550, 3,704,706, European patent application No. 102,816 published Mar. 13, 1984, European patent application No. 104,772 published Apr. 4, 1984, and European patent application No. 104,771 published Apr. 4, 1984. Pulse oximeters are commercially available from Nellcor Incorporated, Hayward, Calif., U.S.A., and are known as, for example, Pulse Oximeter Model N-100 (herein "N-100 oximeter").

Pulse oximeters typically measure and display various blood flow characteristics including but not limited to blood oxygen saturation of hemoglobin in arterial blood, volume of individual blood pulsations supplying the flesh, and the rate of blood pulsations corresponding to each heartbeat of the patient. The oximeters pass light through human or animal body tissue where blood perfuses the tissue such as a finger, an ear, the nasal septum or the scalp, and photoelectrically sense the absorption of light in the tissue. The amount of light absorbed is then used to calculate the amount of blood constituent being measured.

The light passed through the tissue is selected to be of one or more wavelengths that is absorbed by the blood in an amount representative of the amount of the blood constituent present in the blood. The amount of transmitted light passed through the tissue will vary in accordance with the changing amount of blood constituent in the tissue and the related light absorption.

For example, the N-100 oximeter is a microprocessor controlled device that measures oxygen saturation of hemoglobin using light from two light emitting diodes ("LED's"), one having a discrete frequency of about 660 nanometers in the red light range and the other having a discrete frequency of about 925 nanometers in the infrared range. The N-100 oximeter microprocessor uses a four-state clock to provide a bipolar drive current for the two LED's so that a positive current pulse drives the infrared LED and a negative current

pulse drives the red LED to illuminate alternately the two LED's so that the incident light will pass through, e.g., a fingertip, and the detected or transmitted light will be detected by a single photodetector. The clock uses a high strobing rate, e.g., one thousand five hundred cycles per second, to be easily distinguished from other light sources. The photodetector current changes in response to the red and infrared light transmitted in sequence and is converted to a voltage signal, amplified, and separated by a two-channel synchronous detector—one channel for processing the red light waveform and the other channel for processing the infrared light waveform. The separated signals are filtered to remove the strobing frequency, electrical noise, and ambient noise and then digitized by an analog to digital converter ("ADC"). As used herein, incident light and transmitted light refers to light generated by the LED or other light source, as distinguished from ambient or environmental light.

The light source intensity may be adjusted to accommodate variations among patients' skin color, flesh thickness, hair, blood, and other variants. The light transmitted is thus modulated by the absorption of light in the variants, particularly the arterial blood pulse or pulsatile component, and is referred to as the plethysmograph waveform, or the optical signal. The digital representation of the optical signal is referred to as the digital optical signal. The portion of the digital optical signal that refers to the pulsatile component is labeled the optical pulse.

The detected digital optical signal is process by the microprocessor the N-100 oximeter to analyze and identify arterial pulses and to develop a history as to pulse periodicity, pulse shape, and determined oxygen saturation. The N-100 oximeter microprocessor decides whether or not to accept a detected pulse as corresponding to an arterial pulse by comparing the detected pulse against the pulse history. To be accepted, a detected pulse must meet certain predetermined criteria, for example, the expected size of the pulse, when the pulse is expected to occur, and the expected ratio of the red light to infrared light of the detected optical pulse in accordance with a desired degree of confidence. Identified individual optical pulses accepted for processing are used to compute the oxygen saturation from the ratio of maximum and minimum pulse levels as seen by the red wavelength compared to the maximum and minimum pulse levels as seen by the infrared wavelength.

Several alternate methods of processing and interpreting optical signal data have been disclosed in the patents and references cited above.

A problem with non-invasive pulse oximeters is that the plethysmograph signal and the optically derived pulse rate may be subject to irregular variants in the blood flow, including but not limited to motion artifact, that interfere with the detection of the blood flow characteristics. Motion artifact is caused by the patient's muscle movement proximate to the oximeter sensor, for example, the patient's finger, ear or other body part to which the oximeter sensor is attached, and may cause spurious pulses that are similar to pulses caused by arterial blood flow. These spurious pulses, in turn, may cause the oximeter to process the artifact waveform and provide erroneous data. This problem is particularly significant with infants, fetuses, or patients that do not remain still during monitoring.

A second problem exists in circumstances where the patient is in poor condition and the pulse strength is very weak. In continuously processing the optical data, it can be difficult to separate the true pulsatile component from artifact pulses and noise because of a low signal to noise ratio. Inability to reliably detect the pulsatile component in the

optical signal may result in a lack of the information needed to calculate blood constituents.

It is well known that electrical heart activity occurs simultaneously with the heartbeat and can be monitored externally and characterized by the electrocardiogram ("ECG") waveform. The ECG waveform, as is known to one skilled in the art, comprises a complex waveform having several components that correspond to electrical heart activity. The QRS component relates to ventricular heart contraction. The R wave portion of the QRS component is typically the steepest wave therein, having the largest amplitude and slope, and may be used for indicating the onset of cardiovascular activity. The arterial blood pulse flows mechanically and its appearance in any part of the body typically follows the R wave of the electrical heart activity by a determinable period of time that remains essentially constant for a given patient. See, e.g., Goodlin et al., "Systolic Time Intervals in the Fetus and Neonate", Obstetrics and Gynecology, Vol. 39, No. 2 February 1972, where it is shown that the scalp pulse of fetuses lag behind the ECG "R" wave by 0.03–0.04 second, and U.S. Pat. NO. 3,734, 086.

In prior U.S. application Ser. No. 742,720, copending and commonly assigned, the disclosure (including the software appendix, of which is hereby expressly incorporated by reference, and in corresponding International PCT Application publication No. WO 86/05674 published Oct. 9, 1986, also commonly assigned, there is disclosed an invention for measuring the patient's heart activity and correlating it with the patient's detected blood flow signal to calculate more accurately the patient's oxygen saturation and pulse rate. The correlation includes auto- and cross correlation techniques to enhance the periodic information contained in each individual waveform as well as determine that time relationship of one waveform to another.

Correlating the occurrence of cardiovascular activity with the detection of arterial pulses occurs by measuring an ECG signal, detecting the occurrence of the R-wave portion of the ECG signal, determining the time delay by which an optical pulse in the detected optical signal follows the R-wave, and using the determined time delay between an R-wave, and using the determined time delay between an R-wave and the following optical pulse so as to evaluate arterial blood flow only when it is likely to present a true blood pulse for waveform analysis. The measured time delay is used to determine a time window when, following the occurrence of an R-wave, the probability of finding an optical pulse corresponding to a true arterial pulse is high. The time widow provides an additional criterion to be used in accepting or rejecting a detected pulse as an optical pulse. Any spurious pulses caused by motion artifact or noise occurring outside of that tie window are typically rejected and are not used to calculate the amount of blood constituent. Correlating the ECG with the detected optical pulses thereby provided for more reliable measurement of oxygen saturation.

That application and publication refers to a modified N-100 oximeter (the "enhanced N-100 oximeter") whereby the device is provided with an additional heart activity parameter in the form of a detected R-wave from the patient's ECG waveform, in addition to the N-100 pulse oximeter functions, and the microprocessor is modified to include software and memory for controlling and processing the optical signal and heart activity information.

The additional heart activity parameter is independent of the detection of peripheral arterial pulses, e.g., ECG signals, ultrasound, ballistocardiogram, and maybe, accelerometers, nuclear magnetic resonators, electrical impedance techniques, and the like, and provides an identifiable and detectable signal to response to each heartbeat for use by the signal

processing of the oximeter.

It is an object of this invention to provide for improved processing of the detected optical signal containing periodic information corresponding to arterial pulsatile blood flow and aperiodic information corresponding to noise, spurious signals, and motion artifact unrelated to the beating heart and arterial pulsatile blood flow, to improve further the reliability and accuracy of the determination of blood constituent, particularly oxygen saturation of hemoglobin by a non-invasive oximeter device.

It is another object of this invention to provide an improved method and apparatus for collecting successive portions of detected optical signals encompassing periodic information for more than one heartbeat and processing the collected portions to attenuate and filter therefrom aperiodic signal waveforms to provide enhanced periodic information from which the patient's blood constituent can be accurately determined.

It is another object to maintain the enhanced periodic information updated by continuing to add new portions of detected optical signals as they are obtained. It is another object of this invention to create enhanced periodic information by collecting and processing successive portions of detected optical signals wherein the periodic information corresponding to the optical pulses have been added together in phase, synchronized to the occurrence of the patient's ECG and preferably the R-wave signal.

It is another object of this invention to add synchronized periodic information in a weighted fashion so that the most recent portion of detected optical signal is accorded a greater weight in the collected sum than any one prior portion of periodic information data.

It is another object of this invention to create the enhanced periodic information by adding together a predetermined number of the most recent successive portions of detected optical signal, whereby each portion corresponds to a heartbeat event and is given a weight according to its relative age so as to emphasise the newest information in the resultant weighted collective sum.

It is another object of this invention to correlate the periodic information with the ECG R-wave by using a waveform product technique to identify the occurrence of the heartbeat and the optical pulse corresponding to that heartbeat.

It is another object of this invention to evaluate the collect periodic information for a predetermined number of successive portions of the detected optical signal corresponding to a predetermined number of heartbeats in the frequency domain to obtain enhanced periodic information.

It is another object of this invention to Fourier transform a time-measure of detected optical signals including periodic information for N heartbeats to determine the relative maxima at the fundamental frequency N and the minima at the zero frequency for use in determining the light modulation ratio for the amount of blood constituents.

It is another object of this invention to correlate the Fourier Transform of the time-measure of detected optical signals with the Fourier Transform of a time-measure of the ECG signal, and more particularly the R-wave events of the ECG signal, to determine the maxima at the fundamental heart frequency.

It is another object of this invention to correlate the periodic information in a time-measure of the detected optical signal with a time-measure of the detected heart activity, preferably in the form of the ECG signal and more preferably in the form of the R-wave of the ECG signal, to define a predetermined number of samples in a data set and

use frequency domain analysis techniques to evaluate the collected predetermined number of sample data sets to determine the relative maxima at the fundamental frequency.

SUMMARY OF THE INVENTION

This invention provides enhanced periodic information with improved rejection of noise, spurious pulses, motion artifact, and other undesired aperiodic waveforms and thereby improves the ability of oximeters to accurately determine amounts of blood constituents.

The present invention provides methods and apparatus for collecting a time-measure of the detected optical signal waveform containing a plurality of periodic information corresponding to arterial pulses caused by the patient's heartbeat and aperiodic information unrelated to pulsatile flow, and processing the collected time-measure of information to obtain enhanced periodic information that is closely related to the most recent arterial pulsatile blood flow. The time-measure may comprise a continuous portion of detected optical signals including a plurality of periodic information from successive heartbeats, or a plurality of discrete portions of detected optical signals including a corresponding plurality of periodic information.

By updating the time-measure of information to include the most recently detected periodic information, and processing the updated measure collectively, an updated enhanced periodic information is obtained (including the new and historical data) from which aperiodic information (including any new aperiodic information is attenuated. In some embodiments, the updating process includes subtracting detected signals older than a certain relative time from the collected time-measure. Applicants have discovered that by collectively processing a time-measure including successive periodic information to obtain the enhanced periodic information, and using the enhanced periodic information as the basis for making oxygen saturation calculations, the accuracy and reliability of oxygen saturation determinations can be significantly increased. Applicants also have discovered that the time-measures may be collectively processed in either the time domain or the frequency domain.

The amount of a blood constituent, for example, oxygen saturation, can be then determined from this enhanced periodic information (also referred to as composite signal information) by determining the relative maxima and minima in the enhanced periodic information for the respective wavelengths for use in determining the modulation ratios of the known Lambert-Beers equations.

In the preferred embodiment, the detected optical signals are conventionally obtained by passing red (660 nanometers) and infrared (910 nanometers) light through a patient's blood perfused tissue, detecting the transmitted light which is modulated by the blood flow, and providing red and infrared detected optical signals that are preferably separately processed and optionally converted from analog to digital signals, for example, as described above for the Nellcor N-100 oximeter. Portions of the corresponding red and infrared digital signals are then collectively processed in accordance with the present invention and the light modulation ratios are determined based on the resulting enhanced periodic information and used to calculate oxygen saturation.

In the time domain analysis embodiment, the invention provides a method and apparatus for adding together a plurality of successive portions of the detected optical signal waveform whereby one portion of the detected optical signal waveform is added to the following selected portion so that their respective periodic information is added in synchrony, i.e., in phase. The synchronized sum thus forms a composite portion of detected optical signal information having enhanced periodic information. The following portion is then added to the composite portion so that the new periodic information is added to the prior composite periodic information in synchrony, forming an updated composite portion with update enhanced periodic information. Thereafter, subsequent successive portions of detected optical signal are added to the prior updated composite portion, one at a time, so that the composite and enhanced periodic information are update with each new portion and corresponding heartbeat event.

Weighting functions are applied to the two portions before they are added each other. This provides a scaled or weighted sum that can be adjusted, by selection of the respective weighting functions, to more closely reflect the patient's current condition, rather than the historical condition. In the preferred embodiment, the weighting functions are fractional multipliers which sum to one to provide a stable filter, and are discussed in greater detail below.

The periodic information (optical pulse) generally has the same pulse shape, height, and duration from heartbeat to heartbeat and, as is described in U.S. Ser. No. 742,720, follows heart activity by a determinable period of time.

Applicants have discovered that by synchronizing the occurrence of successive R-waves, it becomes possible to add the corresponding successive portions of the detected optical signal together so that the periodic information (optical pulses) corresponding to the arterial pulse in each portion will add in phase. The weighted magnitude of the new periodic information is reinforced by the existence of the weighted enhanced periodic information at the same time location in accordance with the degree of synchrony. If the new optical pulse is identical to the composite pulse, then the updated result is a composite optical pulse having the same magnitude. If the magnitudes differ, the additive result will differ according to the relative weights.

As a result of the collected, synchronized additive process, any aperiodic information that may be present in the portions of the detected optical signals also are weighted and added to the weighted composite portion waveform. However, because aperiodic signals differ in pulse shape, duration, height-, and relative time of occurrence within each portion, and are not synchronous with heart activity, they do not add in phase. Rather, they add in a cancelling manner whereby their weighted sum is spread across the relative time frame of the composite portion.

Applicants have discovered that by processing portions including the periodic information collectively, aperiodic information is attenuated by the absence of any corresponding historical aperiodic signal in the prior composite portion or any subsequent aperiodic at that relative time following heart activity. Further, because the new information can be given a small weight when compared to the absolute weight given the prior composite (as distinguished from the effective lesser weight given to any single prior portion of optical signals as explained below) new aperiodic information is quickly and effectively attenuated, and thus filtered out of the resultant additive portions.

To the extent that any aperiodic information would overlap and thereby obscure some periodic information in a portion, then that aperiodic information would be reinforced by the existing periodic information in the prior composite portion; but only to the extent there was overlap. Thus, the collective processing does not lose optical pulse information hidden by an artifact. Subsequent periodic information lack-

ing "identical" aperiodic information would attentuate any overlapping aperiodic pulse over time.

The collective additive sum having synchronized periodic information waveform thus presents enhanced periodic information that is a composite data set that corresponds to a composite optical pulse from which noise, spurious signals, and motion artifact, have been filtered out. By weighting the collective additive process to favor the most recent information and processing this weighted composite portion as it is update, an accurate estimated optical pulse (enhanced periodic information) that closely reflects the actual conditions is maintained. Basing oxygen saturation determinations on this enhanced optical pulse as it is updated thus provides a more accurate measure than was available by conventional and prior processing techniques.

As discussed in application Ser. No. 742,720, the determinable time period between the R-wave and the optical pulse makes it possible to determine a time window whose time length is long enough to include any likely periodic information, and short enough to exclude detected optical signals that are not of any significant or clinical use in making the determination of the selected blood constituent. A time window can be used in the present invention, following the occurrence of heart activity, to select a portion of detected optical signals for processing in accordance with this invention to reduce the amount of detected optical signal information that must be processed, to improve the rejection of aperiodic signals not proximate to the optical pulse, and to improve the resolution of the oximeter. The timing of the portion can be selected empirically, by considering the time length of the heartbeat pulse and how long it takes for the pulse to travel to the optiCal detection site so that the window is opened before the optiCal pulse maximum occurs at the optical detection site. In the preferred embodiment, the portion of signal is portion that begins 40 ms after the detection of an R-wave event, based on experimentation, and ends after the relative minimum of the optical pulse is detected, which ending time can vary from portion to portion, and may be, for example, about 230 ms after the R-wave event.

The time domain processing of collective weighted portions of the detected optical signal waveforms synchronized by the R-wave of the ECG waveform provides the equivalent of an optimal filter in the frequency domain, whose band-pass elements are those of an ideal heartbeat for the patient under examination. All frequencies which are found in a normal heartbeat are passed with weights of one, and all nonsynchronous frequencies are rejected with attenuation depending on the degree of asynchrony, and the time length of the filter (the effective number of portions processed collectively). As the weight of the periodic information corresponding to the current heartbeat is decreased, greater rejection of the low-frequency aperiodic artifacts occurs, but the delay in reporting the most accurate arterial pulsatile flow increases.

The weighting functions also assure that the new periodic information is not absorbed into the time and amplitude average of the old data. Using fractional weights provides scaling of the new and old composite information sum, and when the fractional weights add to one, stable performance of the filter is assured. Repeated multiplication of the old data by weights less than one accomplish the effective removal of older data, thus limiting in effect the number of periods processed collectively.

In the preferred embodiment, the detected optical signal information is processed in digitized form. Because the successive digitized information is weighted and added, the amount of digital computer memory required to contain the historical and updated composite periodic information only need be as long as the time period for a relative typical heartbeat, so that it can contain the entire time for a selected portion including an optical pulse. This simplified oximeter operation.

In the preferred embodiment, applicants have found that optimal performance occurs when the most recent information is accorded a weight of ⅙ and the historical weight-averaged composite information is accorded a weight of ⅚. Weights which are in powers of 2, e.g., ½, ¼, ⅛, etc., are attractive to use with binary digital computers because they require simpler mathmatical operations, however, they do not necessarily provide the optimal time and noise attenuation tradeoff in selecting weighting functions.

The resultant enhanced periodic information is a weighted composite optiCal pulse that is evaluated in the same manner that prior oximeters evaluated individual pulses they determined were appropriate optical pulses for determining blood constituents, whether or not the criteria included use of a time window. The relative maxima and minima for each of the red and infrared composite optical pulses are separately determined and used in the modulations ratios for determining amounts of blood constituents, e.g., in the modulation ratio R of the Lambert-Beers equations that are commonly used to determine oxygen saturation of arterial hemoglobins as described below. As additional data sets are taken, the collective set of periodic information is updated. Consequently, the most recent waveform data representing the actual amount of blood constituent is included in the updated composite optical pulse form which the updated oxygen saturation can be determined and displayed. Although the foregoing and following discussions generally discuss only a detected optical signal, if should be understood that both the red and infrared signal are separately obtained and processed by these techniques, except as indicated.

In another embodiment of the time domain embodiment, the time-measure of the detected optical signal is collected in a different manner. The digitized portions of information that are to be weighted, synchronously added together, and processed collectively are accumulated in a memory device having sufficient memory locations for storing separately the raw data for a predetermined number of portions of the detected optical signal. The time of occurrence of the R-wave also may be stored in memory as a pointer for the raw data. This filter embodiment permits assigning a different weighting function to each raw data set corresponding to a different heartbeat in the memory, to improve the attenuation of artifacts and reduce the time needed to estimate the actual arterial pulsatile flow in the detected optical signal.

In this embodiment, for N predetermined heartbeats, the average value of the detected optical signal for those N heartbeats is computed by assigning a weight to each data set and adding the weighted data for each heartbeat synchronously into a buffer with a weight of 1, then dividing by N. After each computation, the data set from the oldest stored heartbeat is subtracted from the buffer. As a new R-wave is detected, the incoming data is added to the buffer, and the result is divided by N for computation of relative amplitudes of the two wavelength (red and infrared) periodic information. Thus, the equivalent delay in determining the arterial oxygen saturaton is N/2 times the heartbeat interval. The stored R-wave pointer may be used to correlate the weight-

ing function with the raw data so that the oldest data is given the smallest weight and the most recent data is given the greatest weight, and after each composite heartbeat computation, the oldest data set can be subtracted from the buffer before the following newest data is added.

In an alternate embodiment of the time domain analysis techniques, the ECG and periodic information can be correlated by using a waveform product of the ECG signal and the detected optical signal to determine the location of the optical pulse from which oxygen saturation and heart rate values may be computed. The R-wave of the ECG has the largest slope component within the ECG waveform. In the optical pulse waveform in the portion of detected optical signal, the largest slope is created when the heart contacts to expell blood and thereby produce the arterial pulse. Thus, because of the determinable time interval between the ECG R-wave and the appearance of the optical pulse at the detection site, the detected optical signal can be moved backwards in time, relative to the ECG waveform, an amount equal to the determined time interval so that the portions of maximum slope in the two signals will be aligned, and their product will be at a maximum.

A periodic signals, such as motion artifact, having high slopes will not occur synchronously on the ECG and the detected optical signals. Therefore, once the two periodic waveforms are aligned, the largest slope product of the two will occur at the heart rate interval. Detection of the maximum slope product can be used to pinpoint the occurrence of a heartbeat, and the portion of the detected optical signal that is associated with that maximum slope product can be sued for calculating oxygen saturation.

In this embodiment, the time interval between the R-wave and the optical pulse can be determined by collecting a predetermined time measure history of optical and ECG waveform data comprising n seconds. The time interval must be long enough for the samples to include at least one R-wave and one pule respectively, given that the heartbeat may vary from 20–30 beats per minute at the slowest rates. A measure of six seconds is acceptable. The samples are conventionally digitized and stored in memory. An array of sample-to-sample slopes is obtained for each n second sample of the ECG waveform and for the second half of each optical pulse waveform sample. The first half of the optical pulse sample is discarded so that when the first optical pulse in the second half is slid backwards, the first R-wave peak it will come upon will be its corresponding R-wave, and also so that the most recent heartbeat data is detected. An optical pulse in the first half of the sample could miss its R-wave.

The number of slope values in the second half of the optical waveform, i.e. the number of data points minus 1 at the given sampling rate of 57 samples per second (every 17.5 msec), is taken as m, which corresponds to n/2 seconds of data.

A slope product is obtained by multiplying each element of the optical slope array by its corresponding three and one-half points in the ECG slope array (the ECG signal is sampled every 5 msec) and summing the products. This process is repeated for each of the m optical sample points as the optical waveform slope array is moved backwards relative to the ECG slope array, one optical waveform sample at a time. The backwards slide terminates when the first sample of the ECG waveform is aligned with the first sample of the second half of the optical waveform.

The maximum slope product is found to occur after the optical waveform slop array has been slid x optical sample points backwards. This establishes the time interval t between the detection of the ECG R-wave and the detection of the optical pulse produced by the same heart contraction. This time interval t is expressed in terms of a number of waveform samples, and is used in the determination of heartbeat occurrence.

Computation of the aligned waveform slope product will yield a slope product value for each optical waveform sample. A percentage of the maximum slope product produced during the establishment of the time interval component can be used to compute a maximum product threshold. For example, a percentage of 75% of the maximum slope product may be used. Thus, when the ECG and optical signal waveform slope product exceeds the maximum product threshold, it is likely that a true pulse has been located.

The "true" pulse, as it appears the detected optical signal, can then be validated and processed using known techniques for calculating oxygen saturation from detected optical signals. For example, the slope product could be used to synchronize the R-wave events so that the corresponding periodic information can be added in phase in accordance with the preferred embodiment of this invention, or the slope product could be used as an additional criterion for accepting the corresponding optical pulse as valid, and the oxygen saturation determination could be based only on the corresponding optical pulse. Alternately, the waveform product for the maximum and minimum values for the red and infrared waveforms could be used as the maximum and minimum values in calculating saturation. Also, one could integrate some portion of the selected waveform product waveform and compare the area of the change to the area of the total signal to obtain the relative transmittance for use in determining saturation. For example one could integrate the portion above a selected threshold and compare that area to the integral of the entire pulse.

Qualified "true" pulses are then used to update the slope product threshold value so that it will change as the patient's condition changes or as the quality of the received signals changes.

Applicants also have discovered that a time-measure of detected optical signals containing a plurality of periodic information corresponding to successive heartbeats can be collectively processed and analyzed using frequency domain techniques. These frequency domain techniques utilize the synchronous nature of the heartbeat and the asynchronous characteristics of noise, spurious signals, and motion artifacts.

In the frequency domain, the optical signals for a given wavelength corresponding to the pulsatile arterial blood flow have spectral components including a zero frequency at the background intensity level, a fundamental frequency at the frequency of the beating heart, and additional harmonic frequencies at multiples of the fundamental frequency. Noise, spurious signals, and motion artifact that appear in the detected optical signal have frequencies that spread across the spectrum. Transient changes in the average background intensity level have frequencies that appear spread out between the zero frequency and the fundamental frequency.

The frequency domain embodiment of the present invention provides a method and apparatus for collecting a time-measure of detected optical signals including a predetermined number of optical pulses, converting the collected detected optical signals into the frequency domain, and analyzing the spectral components of the frequency spectrum thereby to determine the red and infrared relative maxima intensity at the fundamental frequency, and relative minima at the background intensity zero frequency, for use as maxima and minima in the percentage modulation ratio for calculating oxygen saturation.

Applicants have discovered that if the digitized values of the time domain detected optical signals are stored in memory for a period of N heartbeats, and the stored data set is transformed into the frequency domain using Fourier Transforms, the amplitude of the fundamental heartrate is summed for the N heartbeats and appears in the frequency spectrum at a location of N cycles. In contrast, the amplitude of asynchronous signals is 1/m where m is the number of data pints in the digitized stored data set, and appear spread across the frequency domain spectrum. The average intensity of the detected optical signal background intensity appears at the spectral line corresponding to zero cycles and corresponds to the average background intensity for that wavelength.

If the detected optical signal for the red and infrared signals is considered as a single complex data set, i.e., having real and imaginary components, only a single Fourier transform is required to analyze the spectral contents of the collective time-measure of the two signals. If F(s) represents the Fourier Transform of the complex data set f(t)=Red(t)+jIR(t)(for Red(t) being the red detected optical signal and IR(t) being the infrared detected optical signal), the Fourier Transform of the real component of f(t) is found by

$$F\{Re[f(t)]\}=\tfrac{1}{2}\{F(s)+F^*(-s)\}.$$

Similarly, the Fourier transform of the imaginary component of f(t) is found by

$$F\{(Im[f(t)]\}=\tfrac{1}{2}\{F(s)-F^*(-s)\}.$$

F*(−s) is the complex conjugate of F(s) with the indexes reversed.

The relative amplitudes of the red and infrared fundamentals at the heartrate has been found to be equivalent to the foregoing time domain techniques for computation of arterial oxygen saturation. The amplitude data may be found by searching the frequency spectrum in the region of expected heart rates for a relative maximum and insuring that this is the fundamental by determining the existence of another relative maximum at twice this rate. This provides a technique for obtaining relative modulation data to calculate arterial oxygen saturation without the need to identify the heart rate independently, e.g., by detecting the ECG. Alternately, the amplitude data at the fundamental may be found by the use of independent heart rate determining mechanism such as ECG or phonoplethysmography or the like to determine a heart rate. However, unlike the time domain techniques, the precise time of occurrence of each heartbeat need not be determined and the optical signal and a heart rate parameter need no be correlated to obtain accurate saturation values. Rather, it is sufficient to obtain an approximate indicator of heart rate, which will facilitate identification of the fundamental frequency and improve saturation reliability.

The number of spectral lines computed is preferably optimized to include the expected range of clinically applicable heartbeats (from 20–250 beats per minute), while the length of the data set is selected by the allowable equivalent delay in displaying measured arterial oxygen saturaton. A time-measure of data of, for example, 9–10 seconds represent delays of only 4–5 seconds in the display of computed saturations, and, depending upon the computational speed of the oximeter microprocessor, the time-measure can be updated in a timely fashion every 1 to 2 seconds.

In the preferred embodiment, the optical signal is digitized at 57 samples per second for each red and infrared signal. When 512 data points are accumulated, the data is Fourier transformed, and the red and infrared fundamental maxima are located. The percentage modulation ratio (red/infrared) is computed by dividing the energy at each maxima by the zero cycle background intensity for that wavelength, then dividing the red modulation by the infrared modulation. The resultant ratio, R, is the used in the manner set forth in the Lamber-Beers equations for calculating arterial saturation of hemoglobin. The collective data can be updated so that new data points replace the oldest data points by using a push down stack memory or equivalent so that the transform, evaluation and saturation calculation could be made after each new data set was obtained.

An alternative embodiment of the frequency domain analysis technique includes sampling the real tim ECG waveform and the real time detected optical signal at high rates, e.g., 1000 samples per second. By examining the ECG wave, the time of occurrence for each heartbeat and the appropriate sample rate to obtain m samples during that heartbeat could be determined. Thus, the data set for each heartbeat can be selected to contain the same number of m samples, where each sample is a fraction of the heartbeat period and N heartbeats contains mxN samples. Taking the Fourier transform of this mxN data set and processing the spectral components of the transform in the same manner as described previously, results in a spectral analysis having several additional advantages. First, the fundamental maximum would always occur at the spectral line for N cycles in "heartbeat" space. Second, any signal present in the data set which did not remain synchronous with the heart, including noise, artifact and transient background intensity changes, would be spread over the heartbeat spectrum. Third, the enhancement in signal-to-noise would be the same for all heart rates. Fourth, because only two spectral lines are of interest, the zero spectral line corresponding to the zero frequency background intensity and the N spectral line corresponding to the number of heartbeats for the data set, the Fourier Transform need only be made at the two frequency components and not of the entire spectrum, and the computation efforts required by the microprocessor are significantly diminished.

The apparatus of the present invention can be used for either time domain or frequency domain analyses, and includes inputs for the plethysmographic detected optical signals and ECG signals of a patient, an analog to digital converter for converting the analog plethysmographic signal to the digital optical signals and for converting the analog ECG signals into digital ECG signals (unless the plethysmographic or ECG signals are provided in digital form), and a digital signal processing section for receiving the digital signals and processing the digital detected optical signal in accordance with one of the foregoing analysis techniques of the present invention, including a microprocessor, memory devices, buffers, software for controlling the microprocessor, and display devices.

In its context, the apparatus of the present invention is a part of an oximeter device which has the capability to detect the red and infrared light absorption, and receive at ECG signal from the patient. In the preferred embodiment, the apparatus of this invention is a part of the Nellcor N-200 Pulse Oximeter (herein the "N-200 oximeter"), a commercially available noninvasive pulse oximeter device manufactured and sold by Nellcor, Incorporated, Hayward, Calif. U.S.A.

The N-200 oximeter is an improved version of the enhanced N-100 oximeter described above and in the prior application Ser. No. 742,720. The N-200 includes circuits that perform many of the same functions as in the N-100 device, but includes some changes, for example, to expand the dynamic range of the device over the N-100 device and to include a 16 bit microprocessor manufactured by Intel Corporation, Model No. 8088. The N-100 oximeter uses an 8 bit microprocessor manufactured by Intel Corporation, Model 8085. The N-200 oximeter includes software for controlling the microprocessor to perform the operations of the preferred embodiment of the time domain analysis techniques of present invention in addition to the conventional oximeter functions, and has some structure and processing methods that are unrelated to the present invention, and therefore are not discussed herein. The software could be modified to perform any of the other time domain or frequency domain analysis techniques of the present invention.

BRIEF DESCRIPTION OF THE DRAWINGS

FIGS. **1A** and **1B** are a block diagram of the apparatus of this invention and the apparatus associated with the present invention.

FIG. **2A** is a detailed circuit schematic of the saturation preamplifier in the patient module of FIG. **1**.

FIG. **2B** is a detailed circuit schematic of the ECG preamplifier and input protection circuit in the patient module of FIG. **1**.

FIGS. **3A** and **3B** are a detailed circuit schematic of the saturation analog front end circuit of FIG. **1**.

FIG. **4** is a detailed circuit schematic of the LED drive circuit of FIG. **1**.

FIG. **5** is a detailed circuit schematic of the ECG analog front end circuit of FIG. **1**.

FIGS. **6A** and **6B** are a detailed circuit schematic of the analog to digital converter section of FIG. **1**.

FIGS. **7A**, **7B**, and **7C** are a detailed circuit schematic of the digital signal processing section of FIG. **1**.

FIG. **8** is a detailed circuit schematic of the external ECG circuitry of FIG. **1**.

FIGS. **9A**, **9B**, **9C**, **9D**, **9E** and **9F** are flow charts for the time domain ECG and optical signal processing of this invention.

FIG. **10** is a flow chart for the frequency domain optical pulse processing of this invention.

FIGS. **10A**, **10B**, **10C**, **10D** and **10E** are a series of waveforms corresponding to the flow chart of FIG. **10**.

DETAILED DESCRIPTION OF THE PREFERRED EMBODIMENT

Referring to FIGS. **1A** and **1B**, the preferred embodiment of the present invention relates to the apparatus for processing the detected analog optical signal and the analog ECG signal and comprises portions of analog to digital conversion section ("ADC converter") **1000** and digital signal processing section ("DSP") **2000**, including the software for driving microprocessor **2040**, which processes the digitized optical signals and ECG signals to determine the oxygen saturation of hemoglobin in arterial blood. Associated with the invention, but not forming a part of the invention, is the apparatus for obtaining the detected analog optical signals and the analog ECG signals from the patient that is part of or is associated with the commercially available Nellcor N-200 Pulse Oximeter. Such apparatus include plethysmograph sensor **100** for detecting optical signals including periodic optical pulses, patient module **200** for interfacing plethysmograph sensor **100** and the conventional ECG electrodes with saturation analog front end circuit **300** and ECG analog front end circuit **400** respectively, saturation analog circuit **300** for processing the detected optical signals into separate red and infrared channels that can be digitized, and ECG analog front end circuit **400** for processing the ECG signal so that it can be digitized. The N-200 oximeter also includes external ECG input circuit **500** for receiving an external ECG signal and processing the signal so that it is compatible with the N-200 processing techniques (as explained below), LED drive circuit **600** for strobing the red and infrared LEDs in plethysmograph sensor **100** at the proper intensity to obtain a detected optical signal that is acceptable for processing, and various regulated power supplies (not shown) for driving or biasing the associated circuits, as well as ADC **1000** and DSP **2000**, from line current or storage batteries.

The associated elements are straightforward circuits providing specified functions which are within the skill of the ordinary engineer to design and build. The associated elements are briefly described here, and reference is made to the corresponding detailed schematics in the Figures and circuit element tables set forth below, to place the apparatus for using the present invention in its operating context in the preferred embodiment.

In the preferred embodiment, the invention requires three input signals, the two plethysmograph or detected optical signals (e.g., red and infrared) and the ECG signal of the patient. If analog signals are provided, they must be within or be adjusted by, for example, offset amplifiers, to be within the voltage input range for the ADC. In circumstances where the signals have been digitized already, they must be bit compatible with the digital signal processing devices, DSP.

The plethysmograph signal is obtained in a conventional manner for a non-invasive oximeter, typically by illuminating the patients tissue with red and infrared light in an alternating fashion, in the manner described above for the N-100 oximeter. Referring to FIGS. **1A** and **1B**, sensor circuit **100** has red LED **110** and infrared LED **120** connected in parallel, anode to cathode, so that the LED drive current alternately illuminates one LED and then the other LED. Circuit **100** also includes photodetector **130**, preferably a photodiode, which detects the level of light transmitted through the patient's tissue, e.g., finger **140**, as a single, analog optical signal containing both the red and infrared light plethysmographic, detected optical signal waveforms.

Referring to FIGS. **1A**, **1B**, **2A**, and **2B**, patient module **200** includes preamplifier **210** for preamplifying the analog detected optical signal of photodetector **130**, ECG preamplifer **220** for preamplifying the analog ECG signal detected from the ECG electrodes that would be attached to the patient in a conventional manner, and protection circuitry **250** interposed between instrumentation amplifier **220** and inverter **230** and the three ECG signal leads, to prevent high voltage transients from damaging the ECG preamplifier electronics.

Preamplifier **210** may be an operational amplifier configured as a current to voltage converter, biased by a positive voltage to extend the dynamic range of the system, thereby converting the photocurrent of photo-diode **130** into a usable voltage signal. ECG preamplifer **220** is preferably a high quality instrumentation amplifier which amplifies the differential signal present on the two ECG signal electrodes. The

common-mode signal present on the two signal electrodes is inverted by inverter **230** and returned to the patient by the third ECG lead, effectively nulling the common-mode signals. A biasing network on tee two ECG signal leads is provided to aid in the detection of when an ECG electrode lead becomes disconnected from patient module **200** or the patient. Patient module **200** also includes leads for passing the LED drive voltages to LEDs **110** and **120**.

Referring to FIGS. **1**A, **1**B, **3**A and **3**B, saturation analog front end circuit **300** receives the analog optical signal from patient module **200** and filters and processes the detected signal to provide separate red and infrared analog voltage signals corresponding to the detected red add infrared optical pulses. The voltage signal is passed through low pass filter **310** to remove unwanted high frequency components above, for example, 100 khz, AC coupled through capacitor **325** to remove the DC component, passed through high pass filter **320** to remove any unwanted low frequencies below, for example, 20 hertz, and passed through programmable gain stage **330** to amplify and optimize the signal level presented to synchronous detector **340**.

Synchronous detector **340** removes any common mode signals present and splits the time multiplexed optical signal into two channels, one representing the red voltage signals and the other representing the infrared voltage signals. Each signal is then passed through respective filter chains having two 2-pole 20 hertz low pass filters **350** and **360**, and offset amplifier **370** and **380**. The filtered voltage signals now contain the signal information corresponding to the red and infrared detected optical signals. Additionally, circuits for use in preventing overdriving the amplifiers in saturation circuit **300** may be applied, for example, level-sensing circuits **312** and **314** (located after low pass filter **310**) for indicating unacceptable LED drive current, and level sensing circuit **315** (located after programmable gain amplifier **330**) for indicating unacceptable input amplifier gain setting.

Referring to FIGS. **1**A, **1**B, and **8**, ECG analog front end circuit **4000** receives the preamplified ECG signal from patient module **200** and processes it for use with the present invention. The analog ECG signal is passed through 2-pole 40 hertz low pass filter **410** for removing unwanted frequencies above 40 hertz, and programmable notch filter **420** for removing unwanted line frequency components. Optionally, circuitry may be provided to measure the line frequency and to select an appropriate clock frequency for the notch filter. The ECG signal is then passed through low pass filter **430**, preferably configured to remove further unwanted components above about 40 hertz, and in particular any frequency components that may have been generated by notch filter **420**. Thereafter, the ECG signal is passed through 2-pole 0.5 hertz high pass filter **440** to remove any low-frequency baseline shifts present in tee original ECG signal, and then passed through offset amplifier **450** to add an offset voltage that the voltage is within the input signal specifications of the analog to digital converter device and the complete waveform will be properly digitized.

It also is desirable to pass the signal output from low pass filter **410** into a circuit that detects whether or not the ECG signal is being detected to identify a "leads-off" condition. The signal voltage is passed through absolute value circuit **480** to take the absolute value of the low pass filter output voltage and sends the value to comparator **490**. Comparator **490** compares the absolute value voltage to a reference threshold or range and, when the absolute value voltage is not within the acceptable range, comparator **490** changes state which change is input to latch **495**, to indicate this condition to, for example, the microprocessor.

Referring to FIGS. **1**A, **1**B and **8**, the Nellcor N-200 device also is equipped with external ECG circuit **500** for receiving the ECG signal of a stand alone ECG detector device and processing the ECG signal so that it can be used with the N-200 oximeter and the present invention. Circuit **500** receives the external analog ECG signal, passes it across capacitor **510** to remove any DC offset voltage and then passes the signal through peak detection circuit **530**. A portion of the AC coupled signal also is passed through buffer amplifier **520** and input to comparator **570**. The held peak voltage is used as the reference threshold voltage that is fed to the other input of comparator **570** so that subsequent QRS complexes in the ECG signal that rise above the threshold generate a trigger signal that is transferred to DPS 2000 by an electrically isolated optical serial communication link comprising serial driving opto-isolator **580**, electrically isolated optical link **590**, and corresponding serial driving opto-isolator **2590** in DSP 2000.

Referring to FIGS. **1**A, **1**B, **6**A and **6**B, ADC 1000 provides the analog to digital conversions required by the N-200 oximeter. The aforementioned three voltage signals, the red detected optical signal, the infrared detected optical signal, and the ECG signal (preferably the ECG signal from patient module **200**), are input to ADC **1000**. These three signals are conventionally multiplexed and digitized by an expanded range 12-bit analog to digital conversion technique, yielding 16-bit resolution. The input signals are passed through multiplexor **1010** and buffer amplifier **1020**. The converter stage includes offset amplifier **1030**, programmable gain circuitry **1040** which allows a portion of the signal to be removed and the remainder to be further amplified for greater resolution, sample and hold circuit **1050**, comparator **1060**, and 12-bit digital to analog converter **1080**. The buffered signal is passed through offset amplifier **1030** to add a DC bias to the signal wherein a portion of the signal is removed and the balance is amplified by being passed through programmable gain circuitry **1040** to improve the resolution. The amplified signal is then passed through sample and hold circuit **1050**, the output of which is fed to one input of comparator **1060**. The other input of comparator **1060** is the output of digital to analog ("DAC") converter **1080** so that when the inputs to comparator **1060** are the same, the analog voltage at the sample and hold circuit is given the corresponding digital word in DAC converter **1080** which is then stored in an appropriate memory device as the digitized data for the sample, and the next sample is sent to sample and hold circuit **1050** to be digitized.

Referring to FIGS. **1**A, **1**B, **4**, **6**A, **6**B, **7**A, **7**B, and **7**C, DAC **1080** also generates the sensor LED drive voltages, under the control of microprocessor **2040**, using analog multiplexor **610**, which separates the incoming analog signal into one of two channels for respectively driving the red and infrared LEDs, having respective sample and hold circuits **620** and **630**, and LED driver circuit **640** for converting the respective analog voltage signal into the respective positive and negative bipolar current signals for driving LEDs **110** and **120**.

Alternate techniques of converting the analog signals to digital signals could be used, for example, a 16-bit analog to digital converter.

Referring to FIGS. **1**, **7**A, **7**B and **7**C, DSP **2000** controls all aspects of the signal processing operation including the signal input and output and intermediate processing. The apparatus includes **16**-bit microprocessor **2040** and its associated support circuitry including data bus **10**, random access memory (RAM) **2020**, read only memory (ROM)

2030, a conventional LED display- device **2010** (not shown in detail), system timing circuit **2050** for providing the necessary clock synchronizing and notch filter frequency signals. In the preferred embodiment, microprocessor **2040** is a model 8088 microprocessor, manufactured by Intel Corporation, Santa Clara, Calif. Alternate microprocessors may be used, such as any of model nos. **8086**, **80186**, and **80286**, also made by Intel Corporation.

Referring to FIGS. **9**A, **9**B, **9**C, **9**D, **9**E, and **9**F and software Appendix A, the flowcharts for the software operation of the preferred embodiment are shown and described. Software appendix A is written in the standard programming language for Intel Model **8088** microprocessor devices.

Similar to the enhanced N-100 oximeter described in U.S. application Ser. No. 742,720, the N-200 oximeter incorporating the present invention is designed to determine the oxygen saturation in one of two modes, an unintegrated mode wherein the oxygen saturation determination is made on the basis of pulses detected in the optical pulse signal that are determined to be optical pulses in accordance with conventional pulse detection techniques, and in an ECG synchronization mode wherein the determination is based on the synchronized additive, composite optical signal information is accordance with the preferred embodiment of the present invention. In an alternate embodiment of the present invention, the determination of saturation in the unintegrated mode may be based on the frequency domain analysis techniques in accordance with this invention with or without the ECG synchronization feature of the time domain analysis techniques.

Referring to FIG. **9**F, interrupt programs control the collection and digitization of incoming optical and ECG data. As particular events occur, various software flags are raised which transfer operation to various routines that are called from the Main Loop Processing routine. For example, Main Loop Processing calls the ECG routine at **3600**, calls a routine that checks the LED levels at **3610** to make sure that there is enough and not too much light being transmitted, looks for the presence of new data at **3615**, and if there is new data, calls the MUNCH routine at **3620**, looks for processed pulse data at **3635** and passes such data to the Leve**13** routine that calculates saturation at **3640**, and also runs various maintenance routines related to the oximeter functions which are not pertinant to the present invention, e.g., at **3625**, **3630**, **3645**, **3650**, **3655**, and **3660** and are not discussed herein. The routines pertinent to the present invention are discussed here. Examples of similar and peripheral other routines may be found in the software appendix to application Ser. No. 742,720.

The detected optical signal waveform is sampled at a rate of 57 samples per second. When the digitized red and infrared signals for a given portion of detected optical signals are obtained, they are stored in a buffer called DATBUF and a software flag indicating the presence of data is set at **3615**. This set flag calls a routine referred to as MUNCH as **3620**, which processes each new digitized optical signal waveform sample. The MUNCH routine is called once per data point and determines pairs of maximum and minimum amplitudes in the detected signal data and presents the pairs to the Leve**13** routine. The Leve**13** routine evaluates the pair of maximum and minimum amplitudes determined by MUNCH, preferably utilizing conventional techniques for evaluating whether a detected pulse is acceptable for processing as an arterial pulse and performs the saturation calculation based on accepted pulses. The MUNCH routine first queries whether or not there is ECG synchronization. If there is ECG synchronization, then the MUNCH routine obtains from the SLIDER routine the enhanced pulse data on which the ECG synchronized saturation calculation will be mad. If there is not synchronization, MUNCH obtains the sample stored in DATBUF on which the unintegrated saturation calculation will be made.

Referring to FIG. **9**A, the SLIDER routine processes each new digitized sample portion of detected optical signal containing the optical pulse to create and maintain the enhanced composite red and infrared optical pulse waveforms, synchronized with the occurrence of successive ECG R-wave events.

The SLIDER routine first inquires whether there is an ECG signal at **3100**. If there is not, then the routine aborts to exit at **3160** to main line operation. If there is an ECG signal, then the SLIDER routine continues and checks the validity of the optical signal at **3110**. If the digitized sample in the buffer DATBUF for either of the red or infrared channels contains an invalid datapoint, the full content of the slider buffer (SLIDEBUF) is erased and the routine exited. The validity of the data is checked by looking for zeros placed in the buffer. Zeros are placed in the buffer when the signal level of the LEDs changes to permit the 20Hz filters to settle, or if the signal exceeds the voltage limits of the processing electronics. This prevents processing of data known to be bad.

If the data are determined to be valid, then the SLIDER routine queries whether or not the data should be skipped at **3120**. The optical signal sampling and data collection and processing of the sampled data are asynchronous processes. On occasion, the data buffer will have several unprocessed samples of data by the time the ECG R-wave event trigger occurs (described below). The R-wave event resets the slider buffer pointer to the beginning of the slider buffer and marks the R-wave data sample in DATBUF SLIDER will not process a data point if the slider buffer pointer is reset already to the beginning of the slider buffer and if the incoming data point was digitized in DATBUF before the data point marked by the R-wave event. Data in the DATBUF buffer prior to the R-wave event are to be skipped. If the data are to be skipped, SLIDER is exited.

If the data are to be processed, SLIDER calculates the updated value for the composite portion waveform sample as "slider data" using the following formula:

$$\text{slider\_data} = \frac{(\text{WEIGHT} - 1)}{\text{WEIGHT}} \times \text{slider\_data} + \frac{1}{\text{WEIGHT}} \times \text{new-data}$$

wherein "WEIGHT" is the aforementioned fractional weighting fraction; "new data" is the data point taken from the incoming sample in DATBUF, and "slider data" is the pre-existing data point in the composite waveform in the slider buffer (SLIDEBUF) before the new data point is added and becomes the updated data point after the computation.

The computation is performed for each data point in DATBUF and any corresponding pre-existing data in the slider buffer. The occurrence of an R-wave event indicates the beginning of the heartbeat sample.

Before making the computation, SLIDER checks the slider buffer to see if there is any existing data at **3130**. If there are data, then at **3150** SLIDER calculates the new value for the composite optical signal. If, however, the slider buffer is empty, then the WEIGHT value is assigned a numerial value of 1 at **3140**, and subsequent new data points

will be weighted 100% and the routine continues to calculate a new value for the composite optical signal at **3150** until the occurrence of the next R-wave event corresponding to the following heartbeat and portion of detected optical signal.

The SLIDER routine also performs other housekeeping chores for the processing associated with the slider buffer. First, in the preferred embodiment, the slider buffer is given a specific length and is able to store about three seconds worth of data. If, for whatever reason, the microprocessor does not receive or recognize an R-wave for more than three seconds, the pointer of the slider buffer is set to point to the last location and does not get incremented beyond that location. Subsequently processed samples are each placed in the last location of the buffer until the next accepted R-wave occurs or a time-out condition occurs. Time-out occurs when no further R-wave events are accepted for a predetermined time of, e.g., five seconds. After time out has occurred, MUNCH is notified that ECG synchronization is lost so that saturation calculations will be based only on the optical signals in DATBUF in the unintegrated mode.

Second, SLIDER continuously compares the updated composite waveform in the slider buffer to the previous composite waveform. If there is a large discrepancy, for example, during electromechanical disassociation, SLIDER takes immediate action to disregard the slider buffer data.

Third, to avoid corrupting the integrity of the waveform data in the slider buffer whenever the apparatus hardware or software triggers a change that influences the signal level of the detected optical signal or the optical pulse waveform, the content of the slider buffer is erased.

Referring to FIG. **9**B, the ECG BOX routine processes the ECG signal obtained through patient module **200** and analog ECG front end circuit to detect ECG R-wave events. The ECG signal is digitized every 5 msec and the digitized values are maintained in a circular buffer. The R-wave event is detected by observing the derivative of the ECG signal. The derivative is obtained at **3200** by application of the following algorithm:

$$\text{derivative\_ecg\_data}[n] = \text{abs}[-0.5*\text{ecg\_data}[n-3] + 0.5*\text{ecg\_data}[n-2] + 0.5*\text{ecg\_data}[n-1] - 0.5*\text{ecg\_data}[n]]$$

where "ecg data[n]" is the digitized value for the ECG signal at sample location n and "ab []" is the absolute value of the bracketed quantity.

The largest magnitude spike in the derivative buffer marks the R-wave. Because the algorithm generates the absolute value of the derivative, the derivative buffer contains two spikes very close to each other, one for the positive-going portion and the other for the negative-going portion of the R-wave. After the first spike is recognized, a timer ecg block is set at **3250** to case ECG BOX to ignore the second spike.

Once the derivative value is obtained, and if the ecg block timer is not active, then the derivative value is compared to the ECG threshold at **3240**. The ECG threshold is a value that is set at about 75% of the previous maximum derivative value. If the derivative is greater than the threshold, ECG BOX starts the ecg block timer by setting ecg block equal to true at **3250**, and it replaces the maximum derivative value with the current derivative value at **3260**, and calls the R-WAVE CHECKING routine at **3270**. After the R-WAVE CHECKING routine is completed (as discussed below), ECG BOX is exited at **3280**.

If the derivative is not greater than the threshold, then ECG BOX is exited at **3280**. Once the ecg bock timer is activated, ECG BOX will continue to calculate the derivative and compare the derivative to the prior maximum derivative value ay **3220**. If the calculated derivative is greater, then the maximum derivative value is set equal to the current derivative ecg data[n]at **3230** and the routine is exited. Otherwise the routine is exited.

Referring to FIG. **9**E, the R-WAVE CHECKING routine receives the detected R-wave event at **3500** and checks the elapsed time since the last R-wave at **3510**. If the elapsed time is less than the minimum internal time limit, preferably set at about **200** msec, the R-wave event is marked as a false R-wave event at **3520**. If the elapsed time is greater than the minimum limit, then the routine starts a phasedelay timer/counter at **3530**. The purpose of the phase-delay counter is to ensure that the optical data is placed into the beginning of the slider buffer after the optical signal minimum from the preceding pulse, but before the signal maximum of the next pulse. The preferred phase-delay period is **40** msec, based on the results of experimentation, and corresponds to the opening of the time window. It may be desirable to have a phase delay period that can be adjusted to accommodate varying optical signal detection conditions for different patients.

The Nellcor N-200 device is equipped with an external ECG input circuit as described above. The main line operating system controlling the operation of the N-200 device receives an interrupt when the external circuit **500** detects an R-wave. On receipt of the interrupt, a message is sent across isolated optical data transmission path **580**-**590**-**2590** (FIGS. **1**A and **1**B) to microprocessor **2040**. The microprocessor then indicates to the ECG processing routines that an externally detected R-wave event has occurred, and the R-wave event is passed to the R-WAVE CHECKING routine. The external ECG analog circuit **500** thus performs the same function as the ECG BOX routine, i.e., determination of an R-wave event followed by the R-WAVE CHECKING routine. The ECG BOX routine is given priority over external ECG circuit **500** in passing signals to R-WAVE CHECKING.

Referring to FIG. **9**C, the ECG routine provides for ECG synchronization, the initialization for slider buffer use, and various other tasks associated with ECG enhancement of the detected optical signal. The ECG routine is entered from the Main Loop Processing system (FIG. **9**F, at **3600**). Its first task is to maintain the ECG related counters/timers, such as ecg block and phase-delay, at **3300**. Next, at **3310**, it checks whether or not the ECG leads from patient module **200** are present, and if not, it checks at **3320** for the presence of an external R-wave event trigger from external ECG circuit **500**. If no R-wave event is detected, then the ECG routine is exited at **3370**.

At this point in the processing, the main line processing system is receiving R-wave events, either from external circuits **500** or from patient module **200** and ECG BOX. Regardless of the source of the R-wave event, the subsequent processing of the R-wave event is the same.

When an external R-wave event is detected or the ECG leads are present, the ECG routine calls the ECG LV3 routine, shown in FIG. **9**D. ECG LV3 runs through a similar patient module **200** lead checking at **3410** or external circuit **500** trigger at **3420** at the ECG routine and if no R-wave event has occurred the routine is exited at **3480**. If an

R-wave event is detected, it is first checked at **3425** to determine whether or not it is a new R-wave event, and if it is not, the ECG LV**3** routine is exited. If it is a new R-wave event, **3430** uses the false R-wave flag (set by the R-WAVE CHECKING routine) to determine whether or not it was a true or false R-wave event. False R-waves will cause the routine to be exited at this point.

If the R-wave event is determined not to be a false R-wave, then the ECG-LV**3** routine builds up a history of R-wave events based on the R-wave to R-wave interval at **3435**. The criteria for accepting an R-wave includes the R-R period and the amplitude of the R-wave. For external ECG circuit **500** triggers, the R-wave even is a uniform pulse resulting from a comparison of the R-wave amplitude to a determined threshold signal.

After computing the R-R interval (or R-R delta) and history, the ECG LV**3** routine checks to see if the ECG is synchronized at **3440**. The ECG is synchronized after receiving the predetermined number, preferably five, acceptable R-wave triggers. For example, the ECG synch counter is initialized at five. The routine tests the ECG synch counter **3440** so that if it is greater than zero, the ECG is determined to be not synched, and then the ECG synch counter is decreased by one at **3455**. Thus, when the ECG synch counter is at zero at **3440**, indicating that the required prior five acceptable R-wave event have successively occurred, then it is determined that there is ECG synchronization and the device will proceed through MUNCH to calculate oxygen saturation based on the enhanced composite slider buffer calculations. Whether or not there is EC synchronization, any R-wave event is checked again at **3450** against the history and R-R interval, if any, to determine whether there is an error in synchronization. If there is an error, the ECG LV**3** routine is exited. If there is no synchronization error, a routine is called at **3460** to compute the maximum length of time after which data in the slider buffer (SLIDEBUF) is disregarded. For example, if there is no prior R-R interval or history, then there will be no error for the first R-wave event. Subsequent true R-wave events will be compared to the prior R-R interval and history and if it appears to be a valid true pulse, then a routine is called to reset slider buffer pointers. However, the saturation calculation will be based upon the slider buffer data only after five R-waves have passed in synch and the synchronization flag is raised. Loss of synchronization resets the ECG synch counter to five.

The ECG LV**3** routine also calculates the maximum length of the slider buffer based on the heart rate, which length is preferably **3** seconds or **2.5** times the determined R-R interval, whichever is the smaller. The ECG LV**3** routine also maintains the slider pointers and counters, resetting them or clearing them as necessary, resets the ecg timeout and bad R-wave counter, computes and displays heart rate based on the R-R interval at **3465**, updates the history buffers and sets the trigger for the MUNCH routine to calculate pulse data for determining oxygen saturation based on the updated slider buffer data at **3470**, sets and computes windows for selecting the portion of detected optical signal to be processed for each heartbeat, based on the history and the most recent data at **3475**. In the preferred embodiment, the windows are set to open by the R-WAVE CHECKING routine phase-delay counter/timer **40** ms after the R-wave occurs and before the maximum optical pulse wave has occurred at the detection site, and set to close by the MUNCH routine after a maximum and minimum pair has been found.

Upon exiting ECG LV**3**, the program returns to the ECG routine and checks the threshold of the derivative buffer of the ECG BOX. If the maximum derivative value is changes substantially, which indicates that the R-wave slope is changing, then the threshold is adjusted.

Referring to FIGS. **10**, **10**A, **10**B, **10**C, **10**D, **10**E and the software appendix B, the flow chart for the software operation of the frequency domain embodiment of the present invention are shown. Software appendix B is written is the Asyst computer language which is a commercially available language.

The routine begins at **4000** with the acquisition of **512** data points for each of the digitized red and infrared optical signals, which are shown graphically at FIG. **10**A. At **4010**, the complex data set, f(t)=Red(t)+jIR(t), is formed. At **4020**, the "D.C." component is formed by summing all of the data points, and the "D.C." component is then removed from the complex data set by subtraction at **4030**, which is graphically shown at FIG. **10**B. The resulting data is then decimated in time to **64** samples at **4040**, which is illustrated in FIG. **10**C, and the time decimated data is then processed by the Hamming Window function at **4050**, which result is illustrated in FIG. **10**D. Thereafter, the Fourier Transform is taken at **4060**. The spectral components of the transform are shown in FIG. **10**E. The Fourier Transforms of the red and infrared components are then calculated at **4070** in accordance with the aforementioned equations, and at **4080** the maximum value at the fundamental heart rate and the minimum value at the zero frequency are determined for each of the red and infrared transforms. The saturation ratio R is calculated as:

$$R = \frac{\dfrac{\text{peak at heartrate for red}}{\text{Re("D.C.")}}}{\text{peak at heartrate } \dfrac{\text{for infrared}}{\text{Im("D.C.")}}}$$

The minimum values for the red and infrared waveforms are taken from the respective real and imaginary components of the "D.C." component. Thereafter, the pulse data is declared ready and saturation is calculated in accordance with the foregoing saturation formula. With each occurrence of the heartbeat, new data is acquired, the **512** data point set is updated and the routine operates to determine the saturation ratio R.

In the preferred embodiment, the blood constituent measured is the oxygen saturaton of the blood of a patient. The calculation of the oxygen saturation is made based on the ratio of the pulse seen by the red light compared to the pulse seen by the infrared light in accordance with the following equation:

$$\text{Saturation} = 100\% \times \frac{BR2 - R(BR1)}{R(BO1 - BR1) + BR2 - BO2}$$

wherein

BO**1** is the extinction coefficient for oxygenated hemoglobin at light wavelength **1** (Infrared)

BO2 is the extinction efficient for oxygenated hemoglobin at light wavelengths 2 (red)

BR1 is the extinction coefficient for reduced hemoglobin at light wavelength 1

BR2 is the extinction coefficient for reduced hemoglobin at light wavelength 2

light wavelength 1 is infrared light

light wavelength 2 is red light

and R is the ratio of the optical density of wavelength 2 to wavelength 1 and is calculated as:

$$R = \frac{\ln [I_{max2}/I_{min2}]}{\ln [I_{max1}/I_{min1}]}$$

wherein

I_{max2} is the maximum light transmitted at light wavelength 2

I_{min2} is the minimum light transmitted at light wavelength 2

I_{max1} is the maximum light transmitted at light wavelength 1

I_{min1} is the minimum light transmitted at light waveguide 1

The various extinction coefficients are determinable by empirical study as is well known to those of skill in the art. For convenience of calculation, the natural log of the ratios may be calculated by use of the Taylor expansion series for the natural log.

Circuit Tables

REF #	CHIP	MFR PART #	Manufacturer	DESCRIPTION OF CHIP
FIG. 2				
210	U2	LF442	NATIONAL SEMICONDUCTOR	DUAL LOW POWER OP AMP
220	U1	INA101HP	BURR BROWN	INSTRUMENTATION AMP
230	U2	LF442	NATIONAL SEMICONDUCTOR	DUAL LOW POWER OP AMP
FIG. 3.				
312	U27	LF444	NATIONAL SEMICONDUCTOR	QUAD JFET OP AMP
312	U28	LP365N	NATIONAL SEMICONDUCTOR	QUAD VOLTAGE COMPARATOR
310	U27	LF444	NATIONAL SEMICONDUCTOR	QUAD JFET OP AMP

320	U27	LF444	NATIONAL SEMICONDUCTOR	QUAD JFET OP AMP
330	U44	MP7524LN	MICROPOWER	8-BIT DAC
330	U32	LF444	NATIONAL SEMICONDUCTOR	QUAD JFET OP AMP
330	U32	LF444	NATIONAL SEMICONDUCTOR	QUAD JFET OP AMP
315	U20	LP365N	NATIONAL SEMICONDUCTOR	QUAD VOLTAGE COMPARATOR
340	U32	LF444	NATIONAL SEMICONDUCTOR	QUAD JFET OP AMP
340	U14	DG243CJ	SILICONIX INCORPORATED	ANALOG SWITCH
340	U7	LF444	NATIONAL SEMICONDUCTOR	QUAD JFET OP AMP
340	U13	LF444	NATIONAL SEMICONDUCTOR	QUAD JFET OP AMP
350	U7	LF444	NATIONAL SEMICONDUCTOR	QUAD JFET OP AMP
360	U13	LF444	NATIONAL SEMICONDUCTOR	QUAD JFET OP AMP
370	U7	LF444	NATIONAL SEMICONDUCTOR	QUAD JFET OP AMP
380	U13	LF444	NATIONAL SEMICONDUCTOR	QUAD JFET OP AMP
340	U19	DG211CJ	SILICONIX INCORPORATED	CMOS ANALOG SWITCH
FIG. 4				
640	U19	DG211CJ	SILICONIX INCORPORATED	CMOS ANALOG SWITCH
640	U32	LF444	NATIONAL SEMICONDUCTOR	QUAD JFET OP AMP
FIG. 5				
410	U12	LF444	NATIONAL SEMICONDUCTOR	QUAD JFET OP AMP
420	U6	LTC1059CN	LINEAR TECHNOLOGY	SWITCHED CAPACITOR FILTER
430	U12	LF444	NATIONAL SEMICONDUCTOR	QUAD JFET OP AMP
440	U12	LF444	NATIONAL SEMICONDUCTOR	QUAD JFET OP AMP
440	U19	DG211CJ	SILICONIX INCORPORATED	CMOS ANALOG SWITCH
450	U12	LF444	NATIONAL SEMICONDUCTOR	QUAD JFET OP AMP
480	U5	LF444	NATIONAL SEMICONDUCTOR	QUAD JFET OP AMP
490	U4	LM393N	NATIONAL SEMICONDUCTOR	VOLTAGE COMPARATOR
495	U10	74HC00	TEXAS INSTRUMENTS	HIGH SPEED CMOS
495	U3	74HC74	TEXAS INSTRUMENTS	HIGH SPEED CMOS

FIG. 6

1010	U24	DG528CK	SILICONIX INCORPORATED	OCTAL ANALOG SWITCH
1020	U25	LF444	NATIONAL SEMICONDUCTOR	QUAD JFET OP AMP
1030	U25	LF444	NATIONAL SEMICONDUCTOR	QUAD JFET OP AMP
1040	U38	AD7524LN	ANALOG DEVICES	DAC
1040	U42	74HC374	TEXAS INSTRUMENTS	HIGH SPEED CMOS
1040	U37	LF442N	NATIONAL SEMICONDUCTOR	LOW POWER OP AMP
1050	U36	LF398N	NATIONAL SEMICONDUCTOR	SAMPLE & HOLD OP AMP
1060	U29	LM211P	TEXAS INSTRUMENTS	LOW OFFSET VOLTAGE COMPARATOR
1080	U43	AD7548KN	ANALOG DEVICES	CMOS 12-BIT DAC
1080	U31	LF411ACN	NATIONAL SEMICONDUCTOR	LOW OFFSET OP AMP
1080	U25	LF444	NATIONAL SEMICONDUCTOR	QUAD JFET OP AMP
610	U18	DG528CK	SILICONIX INCORPORATED	OCTAL ANALOG SWITCH
620	U11	LF444	NATIONAL SEMICONDUCTOR	QUAD JFET OP AMP
630	U11	LF444	NATIONAL SEMICONDUCTOR	QUAD JFET OP AMP

FIG. 7

	U2	82C84A-2	NEC	CMOS 8 MHZ CLOCK GENERATOR
	U1	74HC74	TEXAS INSTRUMENTS	HIGH SPEED CMOS
	U1	74HC74	TEXAS INSTRUMENTS	HIGH SPEED CMOS
2040	U8	MSM80C88RS-2	OKI ELECTRIC	CPU 8MHZ, 125ns
	U3	74HC74	TEXAS INSTRUMENTS	HIGH SPEED CMOS
	U33	74HC374	TEXAS INSTRUMENTS	HIGH SPEED CMOS
	U9	74HC04	TEXAS INSTRUMENTS	HIGH SPEED CMOS
	U3	74HC74	TEXAS INSTRUMENTS	HIGH SPEED CMOS
	U9	74HC04	TEXAS INSTRUMENTS	HIGH SPEED CMOS
	U19	74HC00	TEXAS INSTRUMENTS	HIGH SPEED CMOS
	U9	74HC04	TEXAS INSTRUMENTS	HIGH SPEED CMOS
2030	U21	MBM27C512-25	FUJITSU LIMITED	CMOS 64K X 8 ROM
2020	U15	DS1242	DALLAS SEMICONDUCTOR	CMOS 32K X 8 RAM
	U23	74HC138	TEXAS	HIGH SPEED CMOS

U17	74HC138	TEXAS INSTRUMENTS	HIGH SPEED CMOS
U19	74HC00	TEXAS INSTRUMENTS	HIGH SPEED CMOS
U19	74HC00	TEXAS INSTRUMENTS	HIGH SPEED CMOS
U16	82C51A	OKI ELECTRIC	CMOS UART
U22	MSM82C59A-2RS	OKI ELECTRIC	CMOS INTERRUPT CONTROLLER
2050 U34	MSM82C53-2	OKI ELECTRIC	CMOS TRIPLE TIMER
2050 U38	MSM82C53-2	OKI ELECTRIC	CMOS TRIPLE TIMER
2050 U9	74HC04	TEXAS INSTRUMENTS	HIGH SPEED CMOS
2050 U39	74HC393	TEXAS INSTRUMENTS	HIGH SPEED CMOS
2050 U35	D2732A	INTEL CORPORATION	4096 X 8 ROM
2050 U40	74HC374	TEXAS INSTRUMENTS	HIGH SPEED CMOS
2050 U28	74HC374	TEXAS INSTRUMENTS	HIGH SPEED CMOS
FIG. 8			
520 U3	LF444	NATIONAL SEMICONDUCTOR	QUAD JFET OP AMP
530 U2	LF444	NATIONAL SEMICONDUCTOR	QUAD JFET OP AMP
530 U3	LF444	NATIONAL SEMICONDUCTOR	QUAD JFET OP AMP
570 U7	LM311N	NATIONAL SEMICONDUCTOR	VOLTAGE COMPARATOR W/STROBE

```
ECG:
;ECG MAIN "LOOP"
ecg_00:
        mov     al,ecg_sw       ;check if to do any ECG
        or      al,al
        jn.     short ecg4

        call    ecg_clock
        CALL    LEADS_CHK       ;can't do anything if leads off
        JNC     SHORT ECG1
        TEST    ECG_MODE,EXT_TRIG
        JNZ     SHORT ECG1      ;USING EXTERNAL TRIGGER IF SET
        cmp     ecgtmoflg,1
        je      short ecg0
        cmp     ecgsdly,0
        jne     short ecg0
        CALL    ECG_TMO
        mov     ecgtmoflg,1
ecg0:
        RET

ECG1:
        CALL    ecg_lv3
        call    set_thresh      ;initial threshold adjustment
        call    chk_thresh      ;check and adjust threshold
ECG3:
        MOV     AX,ecgoidx
        ADD     AX,2
        AND     AX,ECG_B_MSK
        MOV     ecgoidx,AX
        RET
ecg4:
        cmp     ecg_sync,1
        je      short ecg5
        call    ecg_init1
ecg5:
        ret

        ; SUBTTL  ECG LEVEL 3

ECG_LV3:
;COMPUTE HEART RATE FROM ECG R WAVE THIS ROUTINE FIGURES OUT WHETHER EKG
;AND/OR OXIMETER ARE SYNCED. IF EKG ONLY, THEN DO CONFIDENCE CHECKING ON
;PERIOD AND DISPLAY AND ALARM CHECK FOR THE HEART RATE!IF OXIMETER IS SYNCED,
;THEN DETERMINE Heart Rate BY TIE R-WAVE PERIOD. IF THE CONFIDENCE IS GOOD,
;
```

```
;BEGIN CHECKING R-WAVE
          TEST      ECG_MODE.LEADS_OFF      ;if leads are on
          JZ        SHORT ECGLV31           ;   yes
          TEST      ECG_MODE.EXT_TRIG       ;   or external trigger in use
          JNZ       SHORT ECGLV3T           ;   yes, continue
          JMP       LV3_RET                 ;else get out
ECGLV31:
          TEST      ECG_MODE.ECG_FLAG       ;if it's ok to do the rest of ecg_lv3
          JNZ       SHORT ECGLV32           ;   continue
          JMP       LV3_RET                 ;else get out
ECGLV32:
          CMP       fals_rwv,0              ;if false rwave
          JE        ecglv32z                ;   no, continue
          MOV       al,ecgsdly              ;else
          CMP       al,0                    ;   reset false r wave counter
          JE        ecgex                   ;   if 0<ecgsdly<4 then ecgsdly++
          CMP       al,4
          JAE       ecgex
          INC       ecgsdly
ecgex:    JMP       lv3_ret                 ;   get out

ecglv32z:
          CALL      HST_ECG                 ;COMPUTE HISTORY PARAMETERS -
                                            ;jump to level3's hstcmp routine
;dab ///
          MOV       ax, dlyavg1             ;copy average r-max, r-slope
          MOV       dlyavg1cpy, ax          ;to visible location
          MOV       ax, dlyavg2
          MOV       dlyavg2cpy, ax
;dab ///
          MOV       ax, ecgmax              ;note interrupt issue: don't raise  ///
          SUB       ax, ecgmin              ;flag until you've calculated delta ///
          MOV       init_max_min,1          ;request next max_min pair          ///
          MOV       ecg_delta, ax           ;                                   ///
ecglv32a:
          CMP       ecgsdly, 0              ;are we synced?
          JE        ecg_is_sync             ; yes, continue
          JMP       ecg_not_sync            ; no, get out
ecg_is_sync:
          CMP       ecg_syndly, 0           ;
          JE        ecgsyn0                 ;
          DEC       ecg_syndly              ;
ecgsyn0:
          XOR       AL,AL                   ;WE'RE SYNC'D
          MOV       ecg_SYNC,AL             ;SHOW ECG'S BEEN USED
          MOV       ecgtmoflg,al            ;
          MOV       CL,ECGLOST              ;show on the display
          CALL      CLRLIT2                 ; that ecg is on
          MOV       CL,ECGON                ; again
          CALL      SETLIT2                 ;
          call      ecg_sync_err_ck         ;did sync error occur
          jnc       ecgsyn8                 ; no, continue
          jmp       ecg_sync_err            ; yes, get out
ecgsyn8:
          xor       ax,ax                   ;get ecg count
          xchg      ecgperiod,ax            ; and clear it at the same time
          mov       rrper,ax                ;update r-r period
          call      calc_slide_tmo          ;calculate what slide_tmo should be ///
          call      do_slider_stuff         ;deal with slider and metermode junk ///
          MOV       ECG_TMR,ECGTMO          ;PERIOD IS GOOD...RESET TIMEOUT
          MOV       BADECG,4                ;RESET BAD ECG TICKET BOOK
          MOV       CX,rravg                ;CALC HEART RATE dab - use rravg not rrper ///
          MOV       AX,ECGRATEFACTOR
          CALL      COMRAT0
          CALL      DSPSR
          CALL      SEND_SAT
          XOR       AL,AC
          CMP       SYNFLG,AL               ;CHECK OXIMETER SYNC FOR ALARM CHECKING
          JNE       SHORT ecgsyn10          ;IF EQUAL, SYNC'D...DO WHOLE THING
          INC       ECGALMCHK               ;set semaphore to tell almchk
          CALL      ALMCHK                  ;   ecg is calling it
          DEC       ECGALMCHK               ;   then clear the semaphore
          JMP       ecgsyn11                ;
ecgsyn10:
                                            ;ecg not synced
          INC       ECGALMCHK               ;set semaphore again for almchkb
          AND       ALMFLG,1 OR ECGTMOF OR FLSTMOF  ;MASK OFF heart rate ALARM
          CALL      ALMCHKB                 ;   check if alarm should be on
          DEC       ECGALMCHK               ;   then clear the semaphore
ecgsyn11:
          CALL      ECG_INP                 ;HISTORY UPDATE FOR ECG
          CALL      setTRIG                 ;SET WINDOW & FLAG TO TRIGGER MUNCH
          call      clear_wi                ;
          call      do_windows              ;
                                            ; accepted r-wave while synced.
LV3_RET:
lv3_ret2:
          mov       mch_flg, 1              ;reenable munch
          AND       ECG_MODE,NOT(ECG_FLAG)  ;disallow ecg_lv3 to be entered
          RET

ECG_not_sync:
ecgnsyn0:
          xor       ax,ax                   ;read ecg counter and reset
          xchg      ecgperiod,ax            ; it at the same time
          cmp       first_rwv_seen, 1       ;is this the first rwave seen
                                            ;since power up/ timeout?
          je        ecgnsyn0a               ;yes, ecgperiod is not yet valid
          mov       rrper, ax               ;no, save ecgperiod in r-r period
```

```
ecgnsync0a:
        mov     first_rwv_seen, 0       ;
        call    ecg_nsync_err_ck        ;
        jc      ecg_not_sync_ex         ;an error occurred
        dec     ecgsdly                 ;ecgsdly will now be )= 0
        cmp     synflg, 0               ;synced on sat =) already displaying a pulse //
        ja      ecg_not_sync_ex         ;so skip the next part
        MOV     FRATN,255               ;SET HR FILTER LOW & CALC BASELINE HR
        MOV     CX,RRPER
        MOV     AX,ECGRATEFACTOR
        CALL    COMRAT0                 ;CALC BUT DON'T DISPLAY YET

ecg_not_sync_ex:
        test    ecgnotsyncerr, ecg_amp_err  ;don't update history if failed
        jnz     ecg_not_sync_ex1        ;peak-peak voltage test
        call    ecg_hup                 ;update ecg history
ecg_not_sync_ex1:
        JMP     LV3_RET

ecg_sync_err:
        MOV     AL,BADECG
        DEC     AL
        jns     ecgserr1                ;not yet out of coupons
        xor     ax,ax                   ;get ecg count
        xchg    ecgperiod,ax            ; and clear it at the same time
        mov     rrper,ax                ;update r-r period
        call    ecg_hup                 ;out of coupons, store it and get out
        jmp     short ecgserr2          ;
ecgserr1:
        MOV     BADECG,AL
        mov     ax, rravg               ;clear ecgperiod if it's too large
        add     ax, diflim + 10         ;last word in diflim is the
                                        ;difference limit for r-r period
        cmp     ax, ecgperiod           ;
        jae     ecgserr2                ;rravg + rr diff lim ) ecgperiod - ok
        mov     ecgperiod, 0            ;
ecgserr2:
        jmp     lv3_ret         ;

ecg_sync_err_ck:
        mov     bl, 0                   ;bl contains error codes
        test    ecg_mode,ext_trig       ;if external trigger do not check
        jnz     ecgserrk1               ;min amplitude
        mov     ax, ecg_delta           ;
        cmp     ax,min_pp               ;check min amplitude requirement
        ja      ecgserrk1               ;fall through if passed
        or      bl, ecg_amp_err         ;
ecgserrk1:
        push    bx                      ;
        CALL    ECG_D_CHK               ;CHECK DIF CODES
        pop     bx
        TEST    CURDIF,bit5             ;ECGBIT...CHECK IF FAILED PERIOD
                                        ;IF NZ, FAILED...DON'T CALC HR
        JZ      ECGserrk2               ;OUT OF BOUNDARY JUMP SOLVED
        or      bl, rr_per_err          ;
ecgserrk2:
        mov     ecgsyncerr, bl          ;
        cmp     ecgsyncerr, 0
        ja      ecgserrk3               ;jump if ecgsyncerr = 0
        stc                             ;return carry -> bl != 0
ecgserrk3:
        ret

ecg_nsync_err_ck:
        mov     bl, 0                   ;bl contains error codes
        test    ecg_mode,ext_trig       ;if external trigger do not check
        jnz     ecgserrk1               ;min amplitude
        mov     ax, ecg_delta           ;
        cmp     ax,min_pp               ;check min amplitude requirement
        ja      ecgnserrk1              ;fall through if passed
        or      bl, ecg_amp_err         ;
ecgnserrk1:
        cmp     ecgsdly, 1              ;only call ecg_v_chk just before
                                        ;about to sync
        jne     ecgnserrk2              ;
        push    bx                      ;
        CALL    ECG_v_CHK               ;CHECK DIF CODES
        pop     bx
        TEST    CURvar,bit5             ;ECGBIT...CHECK IF FAILED PERIOD
                                        ;IF NZ, FAILED...DON'T CALC HR
        JZ      ECGnserrk2                 ;OUT OF BOUNDARY JUMP SOLVED
        or      bl, rr_per_err          ;
ecgnserrk2:
        mov     ecgnotsyncerr, bl       ;
        cmp     ecgnotsyncerr, 0
        ja      ecgnserrk3              ;jump if ecgnotsyncerr = 0
        stc                             ;return carry -> bl != 0
ecgnserrk3:
        ret

do_slider_stuff:
;** note = between here and the next set of stars, al contains slide_mode
;
        mov     al, slide_mode          ;for slider - if passed all checks. ///
        mov     elapse_hb, 0            ;                                   ///
        mov     elapse_sec, 0           ;                                   ///
        mov     bl, daftrig             ;skip the next few samples if
        cmp     dopidx, bl              ;dopidx is lagging where diplox
        je      doslj0                  ;was when the rwave occurred
        or      al, shld_skip           ;set shld_skip and use_ecg to true. ///
doslj0:
        test    al, skip_next_rwv       ;skip this r wave?                  //
        jz      short doslj1            ;no                                 ///
        and     al, not (skip_next_rwv) ;yes                                ///
        mov     slide_mode, al
        jmp     short doslj4
```

```
dosljl:
        or      al, use_ecg             ;
        test    al, first_time          ;                                       ///
        jz      short dosljl2           ;                                       ///
        and     al, not (first_time)    ;                                       ///
        mov     slide_mode, al          ;first time, have clr_newbuf clear
        mov     ax, 0                   ;entire newbuf
        jmp     short dosljl3           ;
dosljl2:
        mov     slide_mode, al          ;not the first time, have clr newbf
        mov     ax, newlidx             ;reset 1st & 2nd newbuf ptrs                    ///
        mov     newlidx2,ax             ;copy newbuf position of ptr1 to ptr2
        mov     newlidx,0               ;reset ptr1 tp start
        mov     newlidx2_on,1           ;turn on ptr2
        jmp     short dosljl4           ;and done
dosljl3:
        call    clr_newbf               ;           ///
        mov     newTidx, 0              ;                                       ///
        mov     newlidx2_on,0           ;turn off newbuf ptr2   axp///
dosljl4:
        ret                             ;

RWAVE_CHK:
;R-WAVE "INTERRUPT" -- CALLED BY THE ACTUAL I.S.R. (ABOVE) OR BY 2.5 mS ROUTINE
        PUSH    AX
        mov     al,ecg_sw               ;check if ok to continue
        or      al,al
        jnz     short intr_ret
RWV1:
        test    first_rwv_flg,0ffh
        jz      rwv12                   ;jump if it is not the first r-wave
        mov     first_rwv_flg,0         ;this is the first one so reset flag
        inc     find_max_req            ;go and search for max deriv value
        mov     find_cnt,TIME200        ;set 2 sec timer
        jmp     intr_ret
rwv12:
        inc     block_flg               ;r-wave block is active
        XOR     AX,AX
        CMP     ECG200MS,AL             ;CHECK IF 200 mS TIMER IS ZERO
        JE      rwv14                   ;IF Y, >200 MS SINCE LAST R-WAVE, CONTINUE
                                        ;ELSE, RATE IS TOO FAST AND/OR TRIGGERED ON
                                        ;SOMETHING OTHER THAN AN R-WAVE
        inc     fals_rwv        ;false R-wave counter
        jmp     short intr_ret
rwv14:
        MOV     ECG200MS,TIME200        ;SET 200 MS COUNTER
        mov     phase_dly,TIME40        ;start 40 msec delay having enough
                                        ;time to put Min in slider buffer
INTR_RET:
        mov     rwv_time_out,TIME5000   ;start 5 sec r-wave watch dog
        POP     AX
        RET

CHK_TRIG:
;CALLED FROM 2.5 mS INTERRUPT TO CHECK STATUS OF EXTERNAL TRIGGER FLAG
;
        cmp     front_ecg_tmout,0
        JNE     SHORT CHKTRIG_XIT       ;IF LEADS ARE ON
        MOV     AL,EXT_ECG_TRIG         ;GET STATUS OF TRIGGER
        OR      AL,AL
        JZ      SHORT CHKTRIG_XIT       ;NOT SET
        XOR     AL,AL
        MOV     EXT_ECG_TRIG,AL
        cmp     ext_ecg_blk,0           ;check ext. rwave-block counter
        jne     chktrig_xit             ;probably a false trigger so jump
        mov     ext_ecg_blk,time50*2    ;init for 100 msec
        OR      ECG_MODE,EXT_TRIG
        mov     ext_trig_tmr, ext_trig_tmo      ;reset back panel timeout
        CALL    RWAVE_CHK
CHKTRIG_XIT:
        RET

;-----------------------------------------------------------------------
;TITLE:  ECGATOD
;FUNCTION: store the ecg signal value in the ecg ring buffer. If front
; panel ecg input flag is set, send ecg data to the powerbase.
; Increment the ecgsperiod and five_ms counters. Also sets the ignore
; rwave flag in ecg_mode if the ecg value isn't within allowable limits.
;
;REGS CHANGED.  ax,bx,dx
;CALLS:
;   adcvt, send_ecg, ecgbox, chktmrs
;-----------------------------------------------------------------------
ECGATOD:
        inc     ecgperiod               ;
        MOV     BX,OFFSET DGROUP: ECGTBL        ;
        CALL    ADCVT                   ;get 12 bit dc voltage in dx
        MOV     BX,ECGIIDX              ;save index into ecg_buf
        push    bx                      ;for later
        CMP     DX,0FFFH                ;railed voltage ?
        jne     atod0                   ;  no, continue to adjust voltage offset
        xor     dx,dx                   ;  yes, no offset to adjust
        jmp     short atod_1            ;    so go on
atod0:
        OR      DX,DX                   ;is dc voltage = 0 ?
        JE      SHORT ATOD_1            ;yes, do nothing
        ADD     DX,GNDV                 ;no, deal with p-gain offset
ATOD_1:
        MOV     OFFSET DGROUP: ECG_BUF [BX],DX  ;load dc voltage
                                                ;  into the ecg buffer
        cmp     init_max_min, 1 ;time-to initialize ecg max and min ?
        je      atod_1a                 ;yes, go do it
                                        ;no, see if we have new ecg extremes
```

```
        cmp     dx, ecgmax      ;if a new dc max is found
        jbe     atod_1b         ;
        mov     ecgmax, dx      ;  update ecgmax
        jmp     atod_1c         ;
atod_1b:
        cmp     dx, ecgmin      ;else if a new dc min is found
        jae     atod_1c         ;
        mov     ecgmin, dx      ;  update ecgmin
        jmp     atod_1c         ;
atod_1a:                        ;if we are in a initializing mode
        mov     ecgmax, dx      ;  load the new max
        mov     ecgmin, dx      ;  and min
        mov     init_max_min,0  ;  & clear the request to initialize
atod_1c:
        INC     BX              ;point ecdiidx
        INC     BX              ;  to the next element
        AND     BX,ECG_B_MSK    ;  in the ecg_buf
        MOV     ECDIIDX,BX      ;  ring buffer
        INC     FIVE_MS         ;count one more time running this routine
        mov     ax,ecgzv        ;get the absolute ecg signal
        add     ax,gndv         ; i.e. Vecg = ecg noise gate + ground voltage
        sub     ax,dx           ;      delta is Vecg - Vecgzero
        jns     atod11          ;
        neg     ax              ;
atod11:
        cmp     ax,41                   ;is the ecg signal less than 100 mV
        jb      atod12                  ;yes, then it's inside noise range
        mov     front_ecg_tmout,TM1@5EC ;no, re-trig front panel ecg timeout
        jmp     short atod13            ;  and go send out the data
atod12:                                 ;ecg signal inside noise
        cmp     front_ecg_tmout,0       ;any front panel ecg input ?
        je      atod_15                 ;no, then nothing to send out
atod13: MOV     AX,DX           ;send ecg data
        CALL    SEND_ECG        ;TO POWER BASE
atod_15:
        pop     bx              ;retrieve index of value just converted
                                ;that pointed to ecg_buf
        MOV     dx, OFFSET DGROUP: ECG_BUF [BX] ;if ecg value
        cmp     dx, max_ecg_lev                 ;  is not within
        jae     short atod_15a                  ;  the max and min
        cmp     dx, min_ecg_lev                 ;  of the allowable
        jbe     short atod_15a                  ;  ecg levels
        jmp     short atod_15b                  ;
atod_15a:
        or      ecg_mode, ignore_rwv            ;  next rwave
        mov     ignore_rwv_tmr, ignore_rwv_time ;  for a limited time
atod_15b:
        call    ecgbox          ;make entry into parallel buffer ///
        call    chktmrs         ;
        RET

ECG_INIT:
        mov     ext_ecg_blk,0           ;init extern. rwave block counter
        MOV     ECGSDLY,4
ecgin0a:
        MOV     ECGPGN,1
        MOV     ECGOFV,0
        MOV     ECGCTR,2        ;sampling intrv=5 ms.
        MOV     ECGFQ,2
        AND     SH_PAT_SAV,ECGDSABL
        MOV     DELTAV,DELTA_MIN
        mov     front_ecg_tmout,TM15SEC         ;init front panel timeout
                                                ;counter with 15 sec
ecg_init1:
        MOV     ecg_SYNC,1
;*---------------------------------------------------------------------
;TITLE:  ECG_TMO
;FUNCTION: ecg time out
ECG_TMO:
ECG_TMO_2:
        CMP     ecg_SYNC,1
        JE      SHORT TMO_3
        TEST    ECG_MODE,TMO_FLG
        JZ      SHORT TMO_4
        JMP     TMO_1
TMO_4:
        OR      ECG_MODE,TMO_FLG
TMO_3:
        mov     bl,ecgsdly
        cmp     ecgsdly,0
        jne     short tmo_3a
        mov     ecgisecalarm,1
tmo_3a:
        MOV     ECGSDLY,4
tmo_3a0:
        MOV     WINDOFAC,3              ;START OUT AT 12.5%
        MOV     MCH_FLG,1               ;ENABLE MUNCH ON TIMEOUT
        mov     ecg_syndly, 4           ;dab ///
;since chktmrs will increment wintmr1 and 2 as long as winflg1 and 2 are
;zero, and since you don't want wintmr1 or 2 to begin counting
;until you're synced on ecg, initialize these variables to something
;non zero. dab ///
        mov     winflg1, 2              ;
        mov     winflg2, 2              ;
        mov     wintmr1, 0              ;
        mov     wintmr2, 0              ;
        call    clrpdbfrs               ;
        call    clr_rr_hist             ;
        mov     first_rwv_seen, 1       ;for ecg lv3 dab ///
        cmp     plstmr,40               ;check pulse timeout counter
        jbe     tmo_33
        mov     plstmr,40
```

```
tmo_33:
        MOV     AL,ECG_MODE
        AND     AL,NOT (EXT_TRIG)
        OR      AL,MODE_1                       ;SET IDLE MODE
        MOV     ECG_MODE,AL
        XOR     AX,AX
        MOV     ECGPERIOD,AX
        MOV     RRPER,AX
        mov     ecg_tmr,al
        MOV     AL,ecg_SYNC
        OR      AL,AL
        JNZ     SHORT TMO_1
        CMP     SYNFLG,0
        JE      SHORT TMO_0
        XOR     AX,AX
        MOV     FRATE,AL
tmo_33a:
        MOV     RATE,AL
        MOV     VRATE,AX
        call    send_sat                ;update analog output    //anb
TMO_0:
        CALL    DSPSR
        OR      ALMFLG,ECGTMOF
        AND     ALMFLG,1 OR ECGTMOF OR PLSTMOF
        INC     ECGALMCHK
        CALL    ALMCHKB
        DEC     ECGALMCHK
        MOV     CL,LOWRATE OR HIRATE
        CALL    CLRLIT2
        CMP     ALICTR,0
        JE      SHORT TMO_1
        MOV     ALMDLY,1
TMO_1:
        mov     al, slide_mode          ;for slider - if use_ecg, ///
        test    al, use_ecg             ;use_ecg (= false, and    ///
        jz      short tmo_1a            ;clear munch's parameters ///
        and     al, not(use_ecg)        ;                         ///
        mov     slide_mode, al          ;                         ///
        mov     perctr, 0               ;just for fun, reset perctr ///
        mov     al, 0                   ;
        MOV     BX,OFFSET MCHMOD        ;RESET MUNCH'S PARAMETERS ///
        MOV     CX,11                   ;                         ///
tmo_1b:
        MOV     BYTE PTR [BX],AL        ;             ///
        INC     BX                      ;             ///
        LOOP    tmo_1b                  ;             ///
tmo_1a:
        call    rwv_lost
        ret

rwv_lost:
        mov     runsum, 0               ;for ecgbox ///
        mov     ecgboxcnt, 4            ; ditto     ///
        mov     ax,thresh_min
        mov     max_thresh,ax           ;set threshold for min. temp.
        mov     max_deriv,ax            ;
        mov     first_rwv_flg,1         ;wait for first r-wave
        mov     init_max_min, 1         ;next call to ecgatod should
                                        ; initialize ecgmax, ecgmin
        mov     least_rwv_lv3, 255      ;the next call to ecg_lv3 should
                                        ; update this value
        RET


LEADS_CHK:
;CHECK FOR ECG LEADS OFF...SET CARRY IF LEADS ARE OFF
        XOR     AX,AX
        TEST    STATUS,LEADS_BIT
        MOV     CLR_LEADS,AL            ;CLEAR LEADS BIT FOR NEXT TIME
        JZ      SHORT LCL_1             ;IF Z,LEADS ARE OK
        OR      ECG_MODE,LEADS_OFF      ;SET LEADS OFF FLAG
        OR      COSTA,ECGOFF
        AND     SH_PAT_SAV,ECGDSABL     ;DISABLE ECG INPUT TO COMPARATOR
        STC
        RET
LCL_1:
LCL_2:
        AND     ECG_mode, NOT (LEADS_off); CLEAR BIT
        AND     COSTA,NOT ECGOFF
        OR      SH_PAT_SAV,NOT ECGDSABL
        CLC
        RET

; Checking and adjusting threshold for ecgboxcar modul
chk_thresh:
        test    ecg_block,0ffffh        ;wait til 30 ms is over
        jnz     ct_ret
        test    block_flg,1             ;adjust just one time after every r-wave
        jz      ct_ret
        dec     block_flg               ;clear block active flag
        mov     ax,old_deriv            ;take 75% of old max
        mov     bx,ax
        mov     cl,2
        shr     bx,cl
        sub     ax,bx
        mov     bx,max_deriv            ;and 25% of new max
        shr     bx,cl
        add     ax,bx                   ;new average max value
        mov     old_deriv,ax
        mov     bx,ax                   ;and take 75% of this value
        shr     bx,cl
        sub     ax,bx                   ;new threshold
        jmp     ct21                    ;jump! ignore the rest
```

```
ct21:
        cmp     ax,thresh_min   ;if threshold < min
        ja       ct3
        mov      ax,thresh_min   ;set min value
ct3:
        mov      max_thresh,ax   ;set threshold
ct_ret:
        ret

set_thresh:
        test    find_max_req,0ffh
        jz       st1             ;return if no max deriv search mode
        test    find_cnt,0ffffh
        jnz      st_ret          ;return if inside 2 sec time interval
        mov     find_max_req,0  ;reset flag
        mov     ax,max_deriv    ;calculate 75% of max deriv value
        mov     old_deriv,ax     ;save average max_deriv
        mov     bx,ax
        mov     cl,2
        shr     bx,cl
        sub     ax,bx
        mov     max_thresh,ax   ;new threshold value
        inc     set_flg         ;check if any r-wave in the next 2 sec
        mov     set_flag_cnt,TIME2000*5
        jmp     min_chk
st1:
        test    set_flg,0ffh    ;inside the next 2 sec interval
        jz       st_ret          ;jump if no
        test    set_flag_cnt,0ffffh
        jnz      st_ret          ;jump if < 2 sec
        mov     set_flg,0       ;reset flag
        cmp     ecgsdly,0
        je       st_ret          ;jump ecg is synced

        mov     ax,max_thresh
        mov     bx,ax
        shr     bx,1
        shr     bx,1
        shr     bx,1
        sub     ax,bx
        mov     max_thresh,ax   ;load min value
        inc     set_flg  ;look for max again
        mov     find_cnt,TIME2000*5
min_chk:
        mov     ax,thresh_min   ;check threshold value against min_thresh
        cmp     ax,max_thresh
        jbe      st_ret
        mov     max_thresh,ax   ;load minimum threshold value
st_ret:
        ret

;calc_slide_two calculates how long an interval may elapse since
;the last good R wave before the slider buffer should be cleared.
;It uses 2 and 1/2 times the average R-R period. ///
calc_slide_two:
        mov     ax, rravg                       ;
        mov     bx, ax                          ;
        shl     ax, 1                           ;2 * rravg
        jc      calc1                           ;overflow
        shr     bx, 1                           ;1/2 * rravg
        add     ax, bx                          ;
        jc      calc1                           ;overflow
        jmp     calc2                           ;
calc1:
        mov     ax, 0ffffh                      ;set to maximum
calc2:
        mov     slide_two, ax                   ;
        ret                                     ;

;clr_rr_hist clears the r-r history buffer  dab ///
clr_rr_hist:
        mov     cx, 8                   ;clear 8 words from es:di
        mov     di, offset ds:rrper
        push    ds
        pop     es
        xor     ax, ax
        cld
        rep     stosw
        ret

;Ecg boxcar filter.
ecgbox:
        mov     dx, bx                          ;save ecgdlidx
        mov     di, offset ds:ecg_buf           ;0 data?
        cmp     word ptr [bx+di],0              ;
        jne     ecgb0                           ;no
        mov     runsum, 0                       ;yes. initialize the sum. set countdown
        mov     ecgboxcnt, 4                    ;
        mov     cx, 0                           ;
        jmp     ecgbl                           ;initialization complete. leave
ecgb0:
        sub     bx, 8                           ;have bx index ecg_buf[ecgidx-4]
        and     bx, ecgmask                     ;
        MOV     AX,runsum                       ;
        cmp     ecgboxcnt, 0                    ;safe to subtract oldest value?
        jne     ecgb0b                          ;no
        SUB     AX,WORD PTR [BX + DI]           ;yes
ecgb0b:
        mov     bx, dx                          ;retrieve ecgdlidx
        add     ax,WORD PTR [BX + DI]           ;add in newest value
        MOV     runsum,AX                       ;AX contains runsum
        cmp     ecgboxcnt, 0                    ;sums valid yet?
        je      ecgb2                           ;yes
        dec     ecgboxcnt                       ;no
```

```
        cmp     ecgboxcnt, 0                    ;sums valid yet to compute entry?
        ja      ecgb2                           ;yes
        mov     cx, 0                           ;
        jmp     short ecgb1                     ;
ecgb2:
        SHR     AX,1                            ;get 1/2 running sum
        sub     bx, 2                           ;
        and     bx, ecgmask                     ;bx indexs ecgbuf[ecgidx-1]
        mov     cx, word ptr [bx + di]          ;cx contains ecgbuf[ecgdidx-1]
        sub     ax, 2                           ;
        and     bx, ecgmask                     ;bx indexes ecgbuf[ecgdidx-2]
        add     cx,WORD PTR [BX + DI]           ;cx contains sum of middle two points
;following code determines whether midsum ) 1/2 running sum
;cx to point to larger value, ax to smaller.
;Purpose is to compute (midsum - 1/2 running sum).
;On entry ax contains 1/2 running sum.
        cmp     cx, ax                          ;
        jae     ecgb1a                          ;
        xchg    cx, ax                          ;
ecgb1a:
        sub     cx, ax                          ;
ecgb1:
        mov     bx, dx                          ;retrieve ecgdidx
        mov     di, offset ds:ecgboxbuf         ;
        mov     word ptr [bx + di], cx          ;enter answer
        cmp     ecg_block, 0                    ;should be finding maximum?
        je      ecgb1b                          ;no
        cmp     cx, max_deriv                   ;data ) max_deriv?
        jbe     ecgb12                          ;no, exit
        mov     max_deriv, cx                   ;yes
        jmp     ecgb12                          ;now exit
ecgb1a:
        cmp     cx, max_thresh                  ;data ) max_thresh?
        jbe     short ecgb12                    ;no
        cmp     ecg200ms, 0                     ;safe to block? ///
        jne     short ecgb1c                    ;no             ///
        mov     ecg_block,TIME100               ;yes, set block_out timer
ecgb1c:
        test    ecg_mode, ignore_rwv            ;should ignore rwave because signal rolled?
        jz      short ecgb1d                    ;no
        jmp     short ecgb12                    ;yes
ecgb1d:
        cmp     first_rwv_flg, 1                ;first rwave seen?
        je      short ecgb1ca                   ;yes, don't update max_deriv
        mov     max_deriv, cx                   ;
ecgb1ca:
        call    rwave_chk                       ;
ecgb12:
        ret                                     ;

slider:
        push    bx                              ;
        mov     al, slide_mode                  ;
        test    al, use_ecg                     ;are we using ecg?
        jnz     slide0                          ;yes
        jmp     no_ecg                          ;no
slide0:                                 ;using ecg
        xor     al,al                           ;anb
        cmp     fmode,2                         ;if mode2 (beat-beat)
        je      slide1                          ;   this_weight = 0
        mov     al, weight                      ;else this_weight = weight
slide1:
        mov     this_weight, al                 ;
        lea     bx, datbuf                      ;check if ir = red = 0
        mov     al, dopidx                      ;
        xor     ah, ah                          ;
        add     bx, ax                          ;
        mov     cx, [bx]                        ;       cx (- datbuf[dopidx]  (IR)
        mov     ax, [bx+2]                      ;       ax (- datbuf[dopidx+2]  (Red)
        or      cx, ax                          ;ir = red = 0?
        jnz     short slide4                    ;no
        mov     al, slide_mode          ;ir = red = 0
        and     al, not (use_ecg)               ;   reset using ecg flag to no
        or      al, skip_next_rwv               ;   & skip next r wave interrupt
        mov     slide_mode,al                   ;
        cmp     meternode, 4                    ;change meternode 4
        jne     short slide2                    ;
        mov     meternode, 3                    ;  to 3
        jmp     short slide3                    ;
slide2:                                         ;
        cmp     meternode, 6                    ;change meternode 6
        jne     short slide3                    ;
        mov     meternode, 5                    ;  to 5
slide3:                                         ;
        mov     ax, 0                           ;
        call    clr_newbf                       ;clear entire newbuf
        mov     newIdx2_on,0                    ;& turn off 2nd newbuf ptr
        jmp     slidx                           ;& we're done
slide4:                                 ;ir and red have data not equal to 0
        mov     al, dopidx                      ;has dopidx caught up to dattrig?
        cmp     al, dattrig                     ;  i.e. to where the ecg occurred
        mov     al, slide_mode                  ;
        jnz     short slide5                    ;no
        and     al, not (shld_skip)             ;yes, then don't skip putting data
        mov     slide_mode, al                  ;  into the slider
        mov     mchead,0                        ;and per dab insistence !!!
        jmp     short slide7                    ;
slide5: test    al, shld_skip                   ;should we skip putting data into
                                                ;  slider buffer?
        jz      short slide7                    ;not skipping
        mov     al,newIdx2_on                   ;yes skipping
        or      al,al                           ;  is slider's ptr2 active?
        jnz     slide6                          ;  yes,
        jmp     slidx                           ;  not on, leave routine
```

```
slide6:
        lea     bx,datbuf               ;skip mode - we advance slider's ptr2 alone
        mov     al,dopidx               ;
        xor     ah,ah                   ;
        add     bx,ax                   ;
        mov     ax,[bx]                 ;get ir data from datbuf
        push    bx                      ;save datbuf pointer
        lea     bx,newbuf               ;
        add     bx,newidx2              ;
        mov     cx,[bx]                 ;get ir data at slider's 2nd pointer
        push    bx                      ; and save ptr2
        call    new_avg                 ;average the two ir values and
                                        ;  return new weighted ave in ax
        pop     bx                      ;restore slider's 2nd ptr
        mov     [bx],ax                 ;  and store new ir average
                                        ;now get weighted average for the red
        inc     bx                      ;retrieve the red
        inc     bx                      ;  value from
        mov     cx,[bx]                 ;  slider's 2nd pointer
        pop     dx                      ;
        push    bx                      ;save slider's 2nd pointer
        mov     bx,dx                   ;put datbuf pointer in bx
        inc     bx                      ;retrieve the red
        inc     bx                      ;  value from
        mov     ax,[bx]                 ;  datbuf
        call    new_avg                 ;average the two red values
        pop     bx                      ;and store the weighted ave
        mov     [bx],ax                 ;  at slider's 2nd pointer
        jmp     slidx                   ;and we're done
slide7:                         ;not in skip mode - both slider ptrs advanced
        lea     bx, newbuf              ;find slider's 1st pointer
        mov     ax, newidx              ;
        add     bx, ax                  ;
        push    bx                      ;save it for later
        mov     cx, [bx]                ;get data it points to
        or      cx, cx                  ;is it zero?
        jnz     short slide8            ;no, leave weight alone
        xor     al, al                  ;yes.
        mov     this_wieght,al          ;clear weight
slide8:
        lea     bx, datbuf              ;point to data in datbuf
        mov     al, dopidx              ;
        xor     ah, ah                  ;
        add     bx, ax                  ;
        pop     cx                      ;restore slider's 1st ptr
        lea     dx, newbuf              ;get slider's 2nd pointer
        mov     ax,newidx2              ;
        add     dx,ax                   ;
        call    put_newbf               ;add new data to slider
                                        ;bx points to datbuf[dopidx],
                                        ;cx points to newbuf[newidx]
                                        ;dx points to newbuf[newidx2]
        jmp     short slidx             ;we're done
no_ecg:                                 ;
        mov     al, slide_mode          ;
        test    al, first_time          ;
        jnz     short slidx             ;first_time flag already set
        mov     ax, elapse_sec          ;get number of seconds since time out
        cmp     ax, slide_tmo           ;
        jae     short set_flg           ;eight seconds or more have elapsed, set flag
        mov     al, elapse_hb           ;get number of heart beats since time out
        cmp     al, 5                   ;
        jae     short set_flg           ;5 or more heart beats since time out
        jmp     short slidx             ;everything's ok
set_flg:                                ;
        mov     al, slide_mode          ;
        or      al, first_time          ;set the first_time
        mov     slide_mode, al          ;
slidx:                                  ;
        pop     bx                      ;
        ret                             ;

;put_newbf moves entries from datbuf to newbuf using the following
;formula:
;newbuf[newidx] (- ((this_weight-1)/this_weight)newbuf[newidx] + (
;                       1/this_weight)datbuf[dopidx]
;
;This procedure assumes bx points to datbuf[dopidx], cx points to
;newbuf[newidx].  Remember, both IR and red values must be placed.
;dx points to newbuf[newidx2]
put_newbf:
        push    dx                      ;save newbuf[newidx2]   axp///
        push    bx                      ;
        push    cx                      ;
        mov     ax, [bx]                ;ax = datbuf[dopidx] IR
        pop     bx                      ;
        push    bx                      ;
        mov     cx, [bx]                ;cx = newbuf[newidx] IR
        call    new_avg                 ;
        pop     bx                      ;
        mov     [bx], ax                ;
        inc     bx                      ;
        inc     bx                      ;
        push    bx                      ;
        mov     cx, [bx]                ;cx = newbuf[newidx] red
        pop     dx                      ;
        pop     bx                      ;
        push    dx                      ;
        inc     bx                      ;
        inc     bx                      ;
        mov     ax, [bx]                ;ax = datbuf[dopidx] red
        call    new_avg                 ;
        pop     bx                      ;
        mov     [bx], ax                ;
;axp///>                                ;bx = newidx for the red data
        mov     dl,newidx2_on           ;is 2nd newbuf pointer active?
```

```
        or      dl,dl           ;
        pop     dx              ;(meanwhile,get 2nd newbuf pointer to ir)
        je      short putnew0   ;no, newidx2 inactive
        dec     bx              ;yes,copy ir,red from 1st to 2nd newbuf ptr
        dec     bx              ;  back up to 1st ir
        mov     cx,[bx]         ;  & save contents
        mov     bx,dx           ;  point to 2nd ir
        mov     [bx],cx         ;  copy into 2nd ir
        inc     bx              ;  advance to 2nd red
        inc     bx              ;
        mov     [bx],ax         ;  and copy into 2nd red
utnew0:
        ret                     ;
```

```
new_avg - if weight is 0, return datbuf[dopidx] in ax.
          else, return (weight-1)/weight newbuf[newidx] + 1/weight datbuf[dopidx] in ax.
Routine assumes ax:= datbuf[dopidx], cx = newbuf[either newidx or newidx2]
Improvement - round division returns rather than truncating
ew_avg:
        cmp     cx,0            ;zero in newbuf?
        je      short newx      ;yes, then return 100% boxcar
        mov     bh, 0           ;
        mov     bl, this_weight ;
        or      bx, bx          ;
        jz      short newx      ;
        mov     dx, 0           ;
        div     bx              ;dxax/bx = ax rem dx =  datbuf[dopidx]/weight
        push    ax              ;save result
        mov     ax, cx          ;
        mov     dx, 0           ;
        div     bx              ;dxax/bx = ax rem dx = newbuf[newidx]/weight
        dec     bx              ;get weight - 1
        mul     bx              ;ax * bx = dxax = (weight-1)(newbuf[newidx]/weight)
        or      dx, dx          ;result too big?
        jz      newl            ;no
        pop     ax              ;yes, clear stack
        mov     ax, 0ffffh      ;and return
        jmp     newx            ;
ewl:
        mov     bx, ax          ;save result
        pop     ax              ;
        add     ax, bx          ;result too big?
        jnc     newx            ;no
        mov     ax, 0ffffh      ;yes, return
ewx:
        ret                     ;

clr_newbf - routine to clear newbuf. Assume ax contains the index
into newbuf from which to clear it! i.e., clear newbuf from where
ax points to til the end.
lr_newbf:
        lea     bx, newbuf      ;
        add     bx, ax          ;bx points to address from which to clear
        mov     cx, newlen      ;
        sub     cx, ax          ;cx contains number of bytes to clear
        or      cx, cx          ;
        jz      short clr2      ;zero is a problem
        js      short clr2      ;so is a negative number
        cmp     cx, 600         ;
        ja      clr2            ;so is too large a number
        shr     cx, 1           ;need to know how many words to clear
        mov     di, bx          ;
        push    ds              ;
        pop     es              ;
        xor     ax, ax          ;loop counter is fine
        cld                     ;
        rep     stosw           ;
lr2:
        ret                     ;
```

Software
ppendix B

```
 0 D/A.TEMPLATE CHNL0

 2 A/D.TEMPLATE CHNLS.23
 2 A/D.TEMPLATE CHNLS.CAL

 4 STRING FILE.NAME

NTEGER DIM[ 512 , 2 ] ARRAY RAW.DAT.1
       DIM[ 512 , 2 ] ARRAY RAW.DAT.2
       DIM[ 1 ] ARRAY ISAT
       DIM[ 2048 , 8 ] ARRAY DAT.BUF

   SCALAR PTR  SCALAR CTR  SCALAR HOW.LONG  SCALAR PTR.MAX
   SCALAR PTR.MIN  SCALAR RCAL SCALAR MIN1 SCALAR MIN2
```

```
COMPLEX
    DIM[ 128 ] ARRAY SHORT.DAT
    DIM[ 128 ] ARRAY SCONJ.DAT
    DIM[ 512 ] ARRAY FOR.DAT
    SCALAR FOR.DC

REAL
    DIM[ 512 ] ARRAY HAMMING.DAT
    DIM[ 1024 ] ARRAY SAT.DAT
    SCALAR RED.MAX SCALAR IR.MAX  SCALAR RNEW
    SCALAR SAT SCALAR BO1 SCALAR BO2 SCALAR BR1 SCALAR BR2 SCALAR SLOPE
    SCALAR INTERCEPT  SCALAR QUAL

: set.up
    HAMMING.DAT [IRAMP
    HAMMING.DAT 2. * PI * 512. / COS -.46 * .54 + HAMMING.DAT :=
    0 MIN1 := 0 MIN2 := 4095 ISAT :=
    1 set.$.optima
    5 set.$.points
;

: SET.TEMPLATE
    CHNLS.23
    RAW.DAT.1 RAW.DAT.2 CYCLIC DOUBLE.TEMPLATE.BUFFERS
    .6 CONVERSION.DELAY
    CHNL0 ISAT CYCLIC TEMPLATE.BUFFER
;

: WAIT.FOR.USER
    SCREEN.CLEAR
    10 10 GOTO.XY
    INTEN.ON
    ." SET LOCATION 24 TO 0 ... THEN: "
    ." STRIKE ANY KEY TO BEGIN. "  CR
    INTEN.OFF
    PCKEY ?DROP DROP
: OPEN.DATA.FILE
    NORMAL.DISPLAY
    ." HOW LONG A DATA SET IN MINUTES ? " #INPUT 7 * HOW.LONG :=
    CR
    ." ENTER THE FILE NAME XXXXXXXX.DAT " "INPUT FILE.NAME ":=
    CR ." PLEASE WAIT WHILE THE DATA FILE IS CREATED... "
    FILE [illegible]
    COMPLEX DIM[ 512 ] SUBFILE
    HOW.LONG TIMES
    END
    FILE.NAME DEFER> FILE.CREATE

: START.ACQ
    DAS.INIT
    CLEAR.TASKS
    CHNL0 1 TASK ARRAY>D/A.OUT
    CHNLS.23 2 TASK A/D.IN>ARRAY
    17 TASK.PERIOD
    PRIME.TASKS
    TRIGGER.TASKS
;
```

```
: STOP.ACQ
    STOP.TASKS
    CLEAR.TASKS
    NORMAL.DISPLAY
;

: FIND.BETAS                    \ GET BETAS FOR SAT CALC
    RCAL                        \ PLACE RCAL ON THE STACK
    CASE
\ RCAL      B01       BR1       B02       BR2       SLOPE        INTERCEPT
   OF       10563     27960     5069      51763     -15743       9237     ENDOF
    ENDCASE

 INTERCEPT :=
 SLOPE :=
 BR2 :=
 B02 :=
 BR1 :=
 B01 :=
 ;



 : SHORT.FLIP
    128 2 DO
    130 I - PTR :=
    SHORT.DAT [ I ] SCONJ.DAT [ PTR ] :=
    LOOP
    SHORT.DAT [ 1 ]  SCONJ.DAT [ 1 ] :=
    SCONJ.DAT CONJ SCONJ.DAT :=
\ ***** COMPUTE SATURATI..
   QUAL .2 > IF
   BR2 RNEW BR1 * - B01 BR1 - RNEW * B02 - BR2 + / 100. *
   SAT :=
   SAT 70. < if
   RNEW 4. / SLOPE * INTERCEPT + 4. * 256. /
   SAT :=
   else
   THEN
   SAT 102. > IF 102. SAT := ELSE THEN
   SAT 0. < IF 0. SAT := ELSE THEN
   SAT SAT.DAT [ CTR ] :=
   SAT 100. > IF 100. SAT := ELSE THEN
   SAT 40.95 * FIX ISAT :=
   CR  ." SAT = " CR SAT . CR  CTR .
   1 CTR + CTR :=
   .25 .51 VUPORT.ORIG
   .75 .50 VUPORT.SIZE
   SHORT.DAT ZMAG SUB[ 1 , 40 ] Y.AUTO.PLOT PTR.MIN 5 + CR .
   STACK.CLEAR
   ELSE
   THEN
```

```
: SHORT.PROCESS
home screen.clear
    FOR.DAT []SUM SI2. / DUP FOR.DC := FOR.DAT SWAP - FOR.DAT :=
    FOR.DC ZREAL 200.. > IF
    FOR.DAT ZREAL []MAX FOR.DAT ZREAL []min - for.dc zreal  /
    dup .2 <. IF
    0. >. if
    FOR.DAT HAMMING.DAT * FOR.DAT :=
    FOR.DAT SUB[ 1 , 128 , 4 ] FOR.DAT SUB[ 2 , 128 , 4 ]
    FOR.DAT SUB[ 3 , 128 , 4 ] FOR.DAT SUB[ 4 , 128 , 4 ] + + + SHORT.DAT :=
    SHORT.DAT FFT SHORT.DAT :=
    SHORT.FLIP
    SHORT.DAT SCONJ.DAT * ZMAG
    SUB[ 4 , 30 ] DUP LOCAL.MAXIMA DROP 1 - PTR.MIN :=
    SUB[ PTR.MIN , 3 ] []SUM IR.MAX :=
    SHORT.DAT SCONJ.DAT - ZMAG
    SUB[ 4 , 30 ] SUB[ PTR.MIN , 3 ] []SUM RED.MAX :=
    RED.MAX FOR.DC ZIMAG / dup 256. / . cr
    IR.MAX FOR.DC ZREAL / dup 256. / . cr DUP QUAL := /
    .' R= ' ? cr RNEW :=
    sat.comp
    THEN
    THEN
    ELSE .' TOO SMALL '
    THEN
    STACK.CLEAR
: GET.IT
HOW.LONG 1 + 1 DO
BEGIN
?BUFFER.SWITCH
UNTIL
?BUFFER.A/B
IF
RAW.DAT.1 XSECT[ 1 , 1 ] MIN1 -
.25 .01 VUPORT.ORIG
.75 .50 VUPORT.SIZE
DUP Y.AUTO.PLOT
RAW.DAT.1 XSECT[ 1 , 2 ] MIN2 -
ELSE
RAW.DAT.2 XSECT[ 1 , 1 ] MIN1 -
.25 .01 VUPORT.ORIG
.75 .50 VUPORT.SIZE
NO.LABELS
DUP Y.AUTO.PLOT
RAW.DAT.2 XSECT[ 1 , 2 ] MIN2 -
THEN
Z=X+IY FOR.DAT :=
FOR.DAT CTR SUBFILE ARRAY>FILE
    SHORT.PROCESS
LOOP
1

: IDLE.IT
    1 CTR :=
    BEGIN
    BEGIN
    ?BUFFER.SWITCH
```

```
UNTIL
?BUFFER.A/B
IF
RAW.DAT.1 XSECT[ 1 , 1 ] MIN1 -
.25 .01 VUPORT.ORIG
.75 .50 VUPORT.SIZE
DUP Y.AUTO.PLOT
RAW.DAT.1 XSECT[ 1 , 2 ] MIN2 -
ELSE
    RAW.DAT.2 XSECT[ 1 , 1 ] MIN1 -
    .25 .01 VUPORT.ORIG
    .75 .50 VUPORT.SIZE
    DUP Y.AUTO.PLOT
    RAW.DAT.2 XSECT[ 1 , 2 ] MIN2 -
    THEN
    Z=X+IY FOR.DAT :=
    SHORT.PROCESS
    ?KEY
    UNTIL
    PCKEY DROP
;

: SHOW.IT
STOP.ACQ
NORMAL.DISPLAY
SCREEN.CLEAR
home ." 1 STARTS A DISPLAY ONLY OXIMETER."
CR CR ." 2 DOES OXIMETRY AND STORES DATA FILES."
CR CR ." 3 CHECKS OFFSET CALIBRATION."
CR CR ." 4 READS RED AND IR VOLTAGES. "
CR CR ." 5 REPLAYS RECORDED DATA FILES "
CR CR ." 6 RETURNS TO DOS. "
CR CR ." 7 CLOSES DATA FILES ON ERRORS. "
cr cr ." 8 PRINT INSTRUCTIONS ON SCREEN. "
cr cr ." ENTER A SELECTION."
    SET.TEMPLATE
    1 CTR :=
    OPEN.DATA.FILE CR
    FILE.NAME DEFER> FILE.OPEN
    ." ENTER RCAL VALUE CODE ( 63 - 84 ) " #INPUT RCAL :=
    FIND.BETAS
    START.ACQ
    GRAPHICS.DISPLAY
    GRID.OFF
    GET.IT
    STOP.ACQ
    FILE.CLOSE
    NORMAL.DISPLAY
    SHOW.IT
    CR


: IDLE
    SET.TEMPLATE
    1 CTR :=
    ." ENTER RCAL VALUE CODE ( 63 - 84 ) " #INPUT RCAL :=
```

```
FIND.BETAS
START.ACQ
GRAPHICS.DISPLAY
GRID.OFF
IOLE.IT
STOP.ACQ
NORMAL.DISPLAY
SHOW.IT
CR
;


: CAL.CHECK
    SET.TEMPLATE
    NORMAL.DISPLAY
    SCREEN.CLEAR
    WAIT.FOR.USER
    START.ACQ
    BEGIN
    ?BUFFER.SWITCH
    UNTIL
    STOP.ACQ
    RAW.DAT.I XSECTI [ , 1 ] [IMIN DUP MIN1 := FLOAT 405.5 /
    ." IR zero is "  . cr
    RAW.DAT.I XSECTI [ , 2 ] [IMIN DUP MIN2 := FLOAT 409.5 /
    ." Red zero is "  .
    3000 MSEC.DELAY
    SHOW.IT
;

: READ.VOLTS
    cr
    CHNLS.CAL
    A/D.INIT
    A/D.IN 409.5 / ." red volts = " . 409.5 / ." IR volts = " .
    CHNL0 D/A.INIT
    1 CTR :=
    NORMAL.DISPLAY DIR *.DAT
    CR ." ENTER THE FILE TO REPLAY " "INPUT FILE.NAME ":=
    CR FILE.NAME DEFER> ?FILE
    CR ." ENTER RCAL VALUE " #INPUT RCAL :=
    FIND.BETAS
    CR ." WHICH FILE TO BEGIN REPLAY ? " #INPUT
    DUP 1 + CR ." HOW MANY FILES TO REPLAY " #INPUT + SWAP
    graphics.display DO
    I .SUBFILE FOR.DAT FILE>ARRAY
    .25 .01 VUPORT.ORIG
    .75 .50 VUPORT.SIZE
    for.dat zreal y.auto.plot
    SHORT.PROCESS
    ISAT [ I ] D/A.OUT
    2000 msec.delay
    LOOP
    FILE.CLOSE
    SHOW.IT
;
```

```
: main.loop
set.up
show.it
begin
    BEGIN
    input
    ?dup not if
    cr ." Invalid number, re-enter: "
    then
    UNTIL
        case
        1 of idle endof
        2 of doit endof
        3 of cal.check endof
        4 of read.volts endof
        5 of replay.dat endof
        6 of bye endof
        7 of file.close  endof
        8 of print.inst endof
        endcase
    again
    ONERR:

      key.drop
      MYSELF
;

banner: my.banner
CR
."                                    NELLCOR PULSE OXIMETER"
CR CR CR
."                                  FOR EXPERIMENTAL USE ONLY"
CR CR CR
."                                      COPYRIGHT 1988"
CR CR
."                                  PROPRIETARY INFORMATION"
CR CR
."                                  ALL RIGHTS RESERVED"
CR CR
."                                  Written by: R. T. Stone, Ph.D."
;BANNER
```

We claim:

1. A method for calculating [the] *an* amount of a blood constituent from the blood flow characteristics of a patient comprising:

detecting an absorption signal corresponding to [the] absorption of light measured at two or more wavelengths in the patient's tissue including periodic changes caused by periodic arterial pulses in the blood flow characteristics and aperiodic changes unrelated to the patient's heartbeat;

detecting an ECG signal corresponding to the patient's ECG waveform corresponding to the periodic electrical heart activity of the patient;

selecting a first time-measure of the absorption signal including a maximum and minimum amplitude change in absorption;

selecting a second time-measure of the ECG signal including components corresponding to the occurrence of the heartbeat;

correlating the ECG signal and the absorption signal by multiplying the first time-measure of the absorption signal and the second time-measure of the ECG signal together to form a waveform product and thereafter shifting the time-measure of the absorption signal backwards in time relative to the time-measure of the ECG signal and multiplying the shifted and unshifted waveforms together to identify a maximum waveform product corresponding to the product of the ECG signal corresponding to the occurrence of the heartbeat and the maximum amplitude of the absorption signal;

processing the absorption signal and the determined correlation to identify the periodic changes in the absorption signal associated with the determined maximum waveform product likely to correspond to arterial pulses in the patient's blood flow characteristics; and

calculating the amount of the blood constituent from the identified periodic changes in the absorption signal.

2. The method of claim 1 wherein the ECG signal further comprises the R-wave component of the ECG signal so that the maximum waveform product corresponds to the product of the R-wave component and the maximum of the absorption signal.

3. The method of claim 2 wherein the first time-measure is shorter in length than the second time-measure so that as the absorption signal time-measure is shifted backward relative to the ECG time-measure the latest maximum waveform product corresponds to the most recent heartbeat activity in the first and second time-measures.

4. The method of claim 3 wherein the first time-measure includes a first plurality of maximum amplitudes and the second time-measure includes a second plurality of R-wave signals so that the maximum waveform product further comprises an array of maximum waveform product.

5. Apparatus for calculating [the] *an* amount of a blood constituent from the blood flow characteristics of a patient comprising:

means for photoelectrically detecting an absorption signal corresponding to [the] absorption of light measured at two or more wavelengths in the patient's tissue including periodic changes caused by periodic arterial pulses in the blood flow characteristics and aperiodic changes unrelated to the patient's heartbeat;

means for electrically detecting an ECG signal corresponding to the patient's ECG waveform corresponding to the periodic electrical heart activity of the patient;

means for selecting a first time-measure of the absorption signal including a maximum and minimum amplitude change in absorption;

means for selecting a second time-measuring of the ECG signal including components corresponding to the occurrence of the heartbeat;

means for correlating the ECG signal and the absorption signal by multiplying the first time-measure of the absorption and the second time-measure of the ECG signal together to form a waveform product and thereafter shifting the time-measure of the absorption signal backwards in time relative to the time-measure of the ECG signal and multiplying the shifted and unshifted waveforms together to identify a maximum waveform product corresponding to the product of the ECG signal corresponding to the occurrence of the heartbeat and the maximum amplitude of the absorption signal;

means for processing the absorption signal and the determined correlation to identify the periodic changes in the absorption signal associated with the determined maximum waveform product likely to correspond to arterial pulses in the patient's blood flow characteristics; and

means for calculating the amount of the blood constituent from the identified periodic changes in the absorption signal.

6. The apparatus of claim 5 wherein the means for electrically detecting an ECG signal further comprises means for detecting the R-wave component of the ECG signal so that the maximum waveform product corresponds to the product of the R-wave component and the maximum of the absorption signal.

7. The apparatus of claim 6 wherein the first time-measure is shorter in length and later in time than the second time-measure so that as the absorption signal time-measure is shifted backwards relative to the ECG time-measure, the latest maximum waveform product corresponds to the most recent heartbeat activity in the first and second time-measures.

8. The apparatus of claim 7 wherein the first time-measure includes a first plurality of maximum amplitudes and the second time-measure includes a second plurality of R-wave signals so that the maximum waveform product further comprises an array of maximum waveform products.

9. A method for calculating [the] *an* amount of a blood constituent from a patient's blood flow characteristics including aperiodic information corresponding to artifacts and periodic information corresponding to arterial pulses, comprising:

directing light of two or more wavelengths toward the patient's tissue;

detecting a set of first signals representing [the] absorption of light measured at *said* two or more wavelengths [of light] in the patient's tissue including detected information corresponding to changes in absorption as a result of changes in the blood flow characteristics;

detecting a second signal representing the occurrences of a selected portion of [the] *an* ECG waveform of the patient;

processing the first and second signals to enhance [the] periodic information contained in each individual signal whereby the first signals are processed in a repetitive manner, one discrete portion at a time, and the occurrence of a selected portion of the patient's ECG waveform in the second signal initiates processing of a discrete portion of the first signals; and

calculating the amount of *said* blood constituent using the enhanced periodic information in the processed portions of the first signals.

10. The method of claim **9** wherein processing the first and second signals further comprises correlating the discrete portions of the first signals and the occurrences of **[**a**]** *said* selected portion of the patient's ECG waveform in the second signals.

11. The method of claim **10** wherein detecting the second signal further comprises detecting a signal representing the occurrences of the R wave portion of the ECG waveform of the patient.

12. The method of claim **11** wherein processing the first and second signals further comprises determining whether or not the detected information in the *discrete* portions of the first signals are likely to be arterial pulses by using the determined correlation and preferentially processing detected information determined likely to be arterial pulses for use in calculating *the amount of said* blood **[**constituents**]** *constituent*.

13. The method of claim **12** wherein preferentially processing detected information further comprises rejecting detected information determined not likely to be arterial pulses so that rejected information is not used in calculating the blood constituent.

14. The method of claim **12** wherein processing the first and second signals further comprises determining whether or not detected information in a discrete portion of the first signals is likely to be an arterial pulse by using the determined correlation after each occurrence of an R wave portion in the second signal.

15. The method of claim **11** wherein correlating the first and second signals further comprises:

determining a period of time after the occurrence of **[**an**]** *said* R wave in which detected information in the *discrete* portions of the first signals corresponding to blood flow characteristic changes caused by arterial pulses are likely to be detected;

determining that detected signal information in the discrete portions of the first signals are likely to be arterial pulses when detected in the determined period of time after the occurrence of **[**an**]** *said* R wave; and

calculating the *amount of said* blood **[**constituents**]** *constituent* further comprises preferentially processing any determined detected information for use in calculating *the amount of said* blood **[**constituents**]** *constituent*.

16. The method of claim **15** wherein preferentially processing detected information further comprises rejecting detected information detected other than in the determined period of time after the R wave occurs so that rejected information is not used in the calculation of *the amount of said* blood **[**constituents**]** *constituent*.

17. The method of claim **15** wherein determining whether or not detected information in a discrete portion of the first signals is acceptable for preferential processing as corresponding to an arterial pulse by using the determined correlation after each occurrence of an R wave portion in the second signal.

18. The method of claim **11** wherein calculating the amount of *said* blood constituent further comprises calculating oxygen saturation of hemoglobin in arterial blood.

19. The method of claim **10** wherein correlating the first and second signals further comprises:

synchronizing the occurrence of a plurality of discrete portions of the first signals;

synchronizing the occurrence of a plurality of occurrences of the R wave portion in the second signal; and

correlating the synchronized discrete portions of the **[**firs**]** *first* signals and the R wave portions.

20. Apparatus for calculating **[**the**]** *an* amount of a blood constituent from a patient's blood flow characteristics including arterial pulses and artifacts comprising:

means for photoelectrically detecting a set of first signals representing the absorption of light measured at two or more wavelengths in the patient's tissue including detected information corresponding to changes in absorption as a result of changes in the blood flow characteristics;

means for detecting a second signal representing the occurrences of a selected portion of **[**the patient's**]** *an* ECG waveform *of the patient*;

means for processing the first and second signals to enhance the periodic information contained in each individual signal whereby each occurrence of **[**a**]** *said* selected portion in the second signal initiates processing of **[**a**]** discrete portions of the first signals so that the information in the first signals are processed in a repetitive manner; and

means for calculating the amount of *said* blood constituent using the enhanced information in the processed portions of the first signals.

21. The apparatus of claim **20** wherein the means for processing the first and second signals further comprises means for correlating the discrete portions of the first signals with the occurrences of **[**a**]** *said* selected portion of the patient's ECG waveform in the second signal.

22. The apparatus of claim **21** wherein the means for detecting a second signal further comprises means for detecting a signal corresponding to the occurrences of the R wave portion of the patient's ECG waveform.

23. The apparatus of claim **21** wherein the processing means further comprises:

means for determining whether or not the detected information in the discrete portions of the first signals are likely to be arterial pulses using the determined correlation; and

means for preferentially processing detected information determined likely to be arterial pulses for use in calculating *the amount of said* blood **[**constituents**]** *constituent*.

24. The apparatus of claim **23** wherein the means for preferentially processing detected information further comprises means for rejecting detected information determined not likely to be arterial pulses so that rejected information is not used in calculating *the amount of said* blood **[**constituents**]** *constituent*.

25. The apparatus of claim **23** wherein the processing means further determines whether or not detected information in a discrete portion of the first signals is likely to be an arterial pulse using the determined correlation after each occurrence of an R wave portion in the second signal.

26. The apparatus of claim **21** wherein the means for correlating the first and second signals further comprises:

first determining means for determining a time period after the occurrence of an R wave in which detected information in the first signals corresponding to blood flow characteristic changes caused by arterial pulses are likely to be detected;

second determining means for determining that detected information in the discrete portions of the first signals are likely to be arterial pulses when detected in the determined period of time after the occurrence of **[**an**]** *said* R wave; and

wherein the means for calculating *the amount of said* blood constituent further comprises means for preferentially processing detected information for use in calculating *the amount of said* blood **[**constituents**]** *constituent.*

27. The apparatus of claim **26** wherein the second determining means further comprises means for rejecting detected information detected other than in the determined period of time after the **R** wave occurs so that rejected information is not used in calculating *the amount of said* blood **[**constituents**]** *constituent.*

28. The apparatus of claim **26** wherein the second determining means further comprises means for determining whether or not detected information in a discrete portion of the first signals is likely to be an arterial pulse using the determined correlation after each occurrence of an R wave.

29. The apparatus of claim **21** wherein the means for calculating further comprises calculating the amount of oxygen saturation of hemoglobin in arterial blood.

30. A method of calculating an amount of a blood constituent from the blood flow characteristics of a patient comprising:

subjecting the patient's tissue to electromagnetic energy, thereby facilitating the transformation of arterial blood flow in the patient's tissue into a blood flow signal, the blood flow signal corresponding to the arterial blood flow including periodic changes caused by periodic arterial pulses in the blood flow characteristics and changes caused by artifact;

detecting the blood flow signal;

detecting the occurrence of a heartbeat of the patient;

correlating the detected blood flow signal and the occurrence of a heartbeat;

processing the blood flow signal and the determined correlation to identify the periodic changes in the blood flow signal likely to correspond to arterial pulses in the patient's blood flow characteristics; and

calculating the amount of the blood constituent from the identified periodic changes in the blood flow signal.

31. The method of claim 30 wherein said blood constituent is the oxygen saturation of hemoglobin in arterial blood.

32. The method of claim 31 wherein said step of detecting the blood flow signal comprises detecting an absorption signal corresponding to the absorption of the electromagnetic energy measured at two or more wavelengths in the patient's tissue.

33. The method of claim 31 wherein said correlating step comprises the steps of:

synchronizing the occurrence of a plurality of changes in the blood flow signal;

synchronizing the occurrences of a plurality of selected portions of the heartbeat; and

correlating the synchronized changes in the blood flow signal with the synchronized portions of the heartbeat.

34. The method of claim 30 wherein said blood constituent is the oxygen saturation of hemoglobin in arterial blood;

said step of detecting the blood flow signal comprises detecting an absorption signal corresponding to the absorption of the electromagnetic energy measured at two or more wavelengths in the patient's tissue; and

said correlating step comprises the steps of

synchronizing the occurrence of a plurality of changes in the blood flow signal,

synchronizing the occurrences of a plurality of selected portions of the heartbeat, and

correlating the synchronized changes in the blood flow signal with the synchronized portions of the heartbeat.

* * * * *