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PORTABLE HEART RATE MONITOR FEASIBILITY STUDY

A PORTABLE HEART RATE MONITOR

by

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ABSTRACT

It is well known that heart rate is an important physiological parameter. In some cases, continuous monitoring of active subjects is desireable.

The report focuses on the feasibility of a miniaturized, portable heart rate monitor. Good artifact rejections, low power consumption small size and ease of use are of primary importance in such design. In order to keep size as low as possible a single cell (1.5 V) supply voltage is used.

A LED - phototransistor type of transducer was chosen for its good artifact rejection and simplicity of application. The transducer clips on the subject's earlobe. Light transmitted through the earlobe is amplitude modulated by the heart (blood) pulses and detected by the phototransistor thus providing electrical signal.

In order to keep current drain low, the LED was powered by $\sim$ 1% duty cycle pulses. The rest of the system was designed to comply with the requirements of the transducer. The detected train of pulses were preamplified and the original modulating waveform (heart pulse) reconstructed by a "sample and hold" circuit. The reconstructed signal was amplified by a narrow-band-pass amplifier filter.

An astable and two monostable multivibrators perform the necessary timing. Two integrated circuits were also employed: a voltage regulator, to provide stable reference voltages where needed and the output amplifier filter, providing the bulk of the gain.

A working prototype was built and suggests that a personalized heart rate monitor is quite feasible. The whole circuit can be integrated, with the exception of few capacitors and perhaps some trimming potentiometers.

ACKNOWLEDGMENTS

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CHAPTER 1

INTRODUCTION

1.1 The Heart Rate as a Physiological Parameter

The heart beat rate can be used in a number of instances to provide indication of the functioning of the heart and for the human body, or the level of physical fitness, for healthy subjects.

A heart rate monitor may be used to provide means of biofeedback; a subject with heart disease can be trained to voluntarily decrease his/her heart rate when the number of ectopics (irregular pulses) becomes dangerously large. Even thought the absolute heart rate in this case is not important in itself, irregular pulses of increased frequency occur, in general at higher heart rates. Therefore, it is desirable that the heart rate be decreased upon appearance of dangeroulsy irregular patterns.

Heart rate, as does body temperature, can be indicative of how a patient is reacting to an illness and helps diagnose instabilities. Its role is extemely important in cases of patients under intensive care.

During exercise the heart rate increases steeply in the beginning, to level off at a certain maximum value which maintains for the duration of the exercise. It returns to its rest value after the end of the exercise. The maximum obtained rate depends upon the age and the fitness index, and varies among individuals. The maximum rate decreases with age and imporvement of the physical condition. With repeated exercise the heart rate reached for a particular task decreases and it returns to the rest value (which also decreases with training) faster. This is an unambiguous indication of improved physical fitness.

It is well known that exercise is a critical factor in cardiovascular well being. However, in order for the exercise to be effective in training the heart, as well as other muscles, it must be vigorous. The heart rate must be in the upper 40% of its range.[1]

1.2 A Portable Heart Rate Monitor

A heart rate monitor used on working or exercising subjects must possess qualities not normally found in bedside monitors where the subject is still.

It must not be affected by the movements of the subject under any conditions. Even during the most vigorous exercise, the accuracy of the instrument must be reasonable if continuous and reliable observation of the heart rate is to be maintained. However, absolute accuracy is not possible if heart activity is monitored indirectly through blood pressure pulses during activity. The reason is simply that acceleration or deceleration of the part of the body to which the transducer is attached will cause blood displacements or transducer displacements and therefore outputs which may not correspond to the heart pulses.

Simplicity in positioning of the transducer as well as operation of the device are essential, and if it is to be used by persons not having special skills, it should require no maintenance other than, say, replacement of a battery, while calibration should be retained for long periods of time.

Used outside controlled environment and especially by athletes, it is bound to experience mechanical shocks which it should be able to withstand successfully.

Worn by people during their daily activities, it must comply with common aesthetic criteria. This basically means that it has to be as unobtrusive as possible.

Inteded for use by working and exercising persons, the device must cause minimal discomfort and not interfere with the subject's activities.

Finally, presented as a consumer product, it must be affordably priced.

In summary, the requirements a personalized heart rate monitor must satisfy are:

(a) Good artifact rejection, to permit use under any conditions of human activity,

(b) Mechanical ruggedness combined with compactness and light weight, for durability, convenience and aesthetic reasons.

The implications of a small-sized, lightweight device are that it must operate on, if possible, a single cell, hearing aid type of battery. Since the capacity of such power supplies is limited, the current drain of the device has to be kept minimal,to allow reasonably long battery life.

In view of a rather complex system with all the stated volume, weight and power consumption restrictions, the use of integrated circuit technology becomes necessary.

The object of this work is to study the feasibility of realizing a system meeting the above requirements.

CHAPTER 2

TRANSDUCERS

2.1 Possible Types of Transducers

Heart rate can be measured indirectly by either sensing electrical signals generated by heart activity (polarization - depolarization of the pacemaker cells) or by sensing blood pressure impuleses.

Possible types of transducers include:

(a) Electrodes

Electrodes are used exclusively in electrocardiography where besides the heart rate they provide additional diagnostic information.

They are almost exlusively made of pure silver or silver chloride on silver, to reduce offset voltages produced by electro-chemical action when brought in contact with the skin.[2]

Electrodes are attached to the subject through the use of adhesive rings or suction cups. The skin must be degreased before the application of the electrode, and a conductive paste is normally used to ensure good electrical contact and reduce artifacts generated by electrode-skin capacitance changes caused by subject movement.

At least three of them are needed, connected to a high input impedance differential amplifier (>1 MΩ differential, >10 MΩ common mode), with high common mode rejection ratio (better than 80 dB).

Typically, the peak to peak voltage output is of the order of 2 mV.

Changes of the body impedance have also been observed to accompany heart pulses. Two electrodes connected to the wrists are used to feed a small audio frequency current, and to sense impedance changes.

The major advantage of the electrodes, for use with a portable heart rate monitor, namely, no power consumption, is overshadowed by several obvious disadvantages, such as inconvenience of use, skin irritation when used for prolonged periods of time and the fact that it is difficult to interface, due to their signal amplification requirements.

(b) Acoustic Transducers

Acoustic transducers are simply contact microphones of either electromagnetic or piezoelectric type. When brought in contact with the skin they convert the minute displacements caused by the blood pressure pulses into electrical signals. The output characteristics are different, typical of the transducer type.

Electromagnetic transducers have low output impedance (2-3 kΩ) and a few millivolts of output signal, while the piezoelectric crystal or ceramic type, have high output voltage ($\sim$ 100 mV) into high impedance (1 MΩ type).[3]

Acoustic transducers offer significant advantages such as high output signal, no power consumption, and they are easy to use and interface. They probably make excellent transducers for monitoring non-active subjects. However, they are extremely susceptible to movement artifacts. Because of their mass, acceleration produces forces at the output of the transducer.

(c) Optoelectric Transducers

Optoelectric transducers are based on changes of the optical transmissivity of tissue with blood pulses. Light emerging from tissue containing blood vessels is amplitude modulated at the heart beat rate.

An optoelectric transducer basically consists of a light source and photodetector. There are two types of optoelectric transducers used for heart pulse detection. In one type, the photodetector and the light source are mounted side by side and the assembly is kept in contact with a digit, with the aid of a band, which also prevents ambient light from reaching the transducer.

Light shone by the source is back-scattered by the tissue and bone and detected by the photodetector.

The second kind of optoelectric transducer is usually in the form of a clip, attached to the earlobe. In this type, the light source and the photodetector are mounted across each other and the ear lobe is inserted between them. The photodetector senses light transmitted through the tissue. Optoelectric detectors can be made small and lightweight.

Artifacts are generated by changes in blood pressure of the location where the transducer is placed due to subject movements or transducer movements relative to the ear lobe.

The transmission type of optoelectric transducer is less artifact susceptible than the digital one, since large pressure pulses are generated when flexing the finger muscles or moving the hands.

The transmission type of optoelectric transducer is the most suitable kind for an active subject heart rate monitor.

The modulation of the light is $\sim$ 4-6% but the output signal amplitude depends upon the intensity of the light source and photo-detector response.

A comparison of the three types of transducers described as to how well they comply with a number of contraints is given in Table I. The ratings may not be absolutely correct in all cases.

TABLE I

Compliance of the Different Types of Transducers to a Number of Constraints

Constraint:	A (electrodes)	B (acoustic)	C (optoelect. transducer)
Electrical			
1. Low power consumption	1	1	2
2. Low voltage operation	1	1	2
3. Immunity to subject movement	3	3	2
4. Immunity to electrostatic noise	3	1	1
5. Immunity to electromagnetic noise (60 Hz)	3		1
6. Immunity to muscle electrical activity	3	1	1
7. Immunicty to light and I.R. radiation	1	1	2
8. Immunity to electrochemical changes of the skin	3	1	1
9. Ease of interfacing	3	1	3
Mechanical			
1. Lightweight	1	1	1
2. Small-sized	2	1	1
3. Ruggedness	1-2	1	1
4. Ease of installment	2-3	2	1
5. Position stability	2	2	1-2
Others			
1. Aesthetic acceptability	2		1-2
2. Chemical resistance	2	2	1

Key to Table:

A - electrodes 1 - Good

B - pressure transducer 2 - Fair

C - photoelectric transducer 3 - Poor

2.2 Light Source

The major disadvantage of the optoelectric transducer is that it requires power for the light source. The only way that it can be used successfuly in battery operated circuits is to be powered by current pulses of small duty cycle.

The obvious choice of light source would be a light emitting diode since it has distinct advantages over incandescent lamps, such as, better efficiency in converting electricity to light, and lack of thermal inertia. The latter feature makes it possible to power the light source with short duration pulses, thus significantly reducing the average power consumption.

Other advantages include longer life, higher reliability and better resistance to mechanical shocks. Furthermore, light emitting diodes radiate all the energy within a very narrow spectrum of wavelengths, which if matched with the region of the peak response of the photodetector, results an improved overall efficiency.

There are several factors that favour the use of GaAs infrared emitting diodes over visible L.E.D.'s:

The transmissivity of living tissue increases considerably with the wavelength which reduces the required light power.

The mechanism of modulation of light transmitted through blood vessels containing tissue, may be either expansion of the blood vessels with each heart pulse due to increased pressure and thus increase of the absorbing material area or due to the fact that the blood filling the vessesl at the beginning of each pulse contains more oxygen than that leaving, towards the end, and modulation of the light occurs due to the difference in absorptivity between HbO_2 and Hb.

It is most likely that both of the above mechanisms contribute to the modulation. However, the effects add for wavelengths above 805 nm where the absorptivity of HbO_2 is higher than that of Hb, while below 805 nm the HbO_2 is lower than that of Hb[3] and the two mechanisms compete against each other.

Another reason for using infrared light is that the most common type of photodetectors, namely silicon photodetectors, have peak response at wavelengths just over 800 nm and good matching with GaAs sources is possible.

Supply voltage considerations also favour infrared emitters, over common LED's. The forward voltage required for a significant current to flow through the diode increases with the photon energy of the emitted radiation. Conversely, longer wavelength emission requires lower forward voltage.

Typical values of such forward voltages are 1.1 - 1.3 Volts for infrared emitters (900 nm), 1.5 - 1.8 V for red light emitting diodes (660 nm) and 1.8 - 2.2 V for green (540 nm).

Thus infrared emitting diodes are the most appropriate type of light source for this application.

A Motorola type MLED 900 diode was used in the construction of the transducer since it was the most efficient of the readily available ones.

Some of the parameters provided by the manufacturer are:

	Min.	Typ.	Max.	Unit
Total radiated power @ 50 mA	200	550		μW
Forward voltage		1.2	1.5	Volts
Peak emission wavelength		900		nm
Spectral line half width		40		nm

Furthermore, the radiated power is a linearly increasing function of the current with a slope of ∿ 10 μW/mA.

2.3 Photodetectors

A suitable photodetector for this application must be as sensitive as possible and it must retain its sensitivity at very low irradiance levels (∿ 50 $\mu W/cm^2$) since the ear lobe absorbs most of the light emitted by the source.

At the same time it must have fast rise time in order for it to be able to follow the short duration light pulses emitted from the light source.

Finally it must have its peak response as closely as possible to 900 nm to match that of the GaAs diode.

The last requirement is best satisfied by silicon photodetectors: photodiodes, photovoltaic cells, phototransistors and darlington pair phototransistors.

Each of the above detectors has different sensitivity-rise time characteristics.

Photodiodes are extremely "fast" devices with rise times of less than 1 μsec. (down to 1 nsec.) but of limited sensitivity (∿ 1 $\mu A/mWcm^{-2}$). Photovoltaic cells are large area photodiodes operated at zero external bias. Their sensitivity is in the region of 0.1 mA/mWcm^{-2}), depending on the area of the cell, but they have very long rise times due to the extremely large capacitance of the large area zero bias junction.

Photodarlingtons are highly sensitive devices ($\sim$ 20 mA/mWcm^{-2}) but slow (rise time > 50 μ sec).

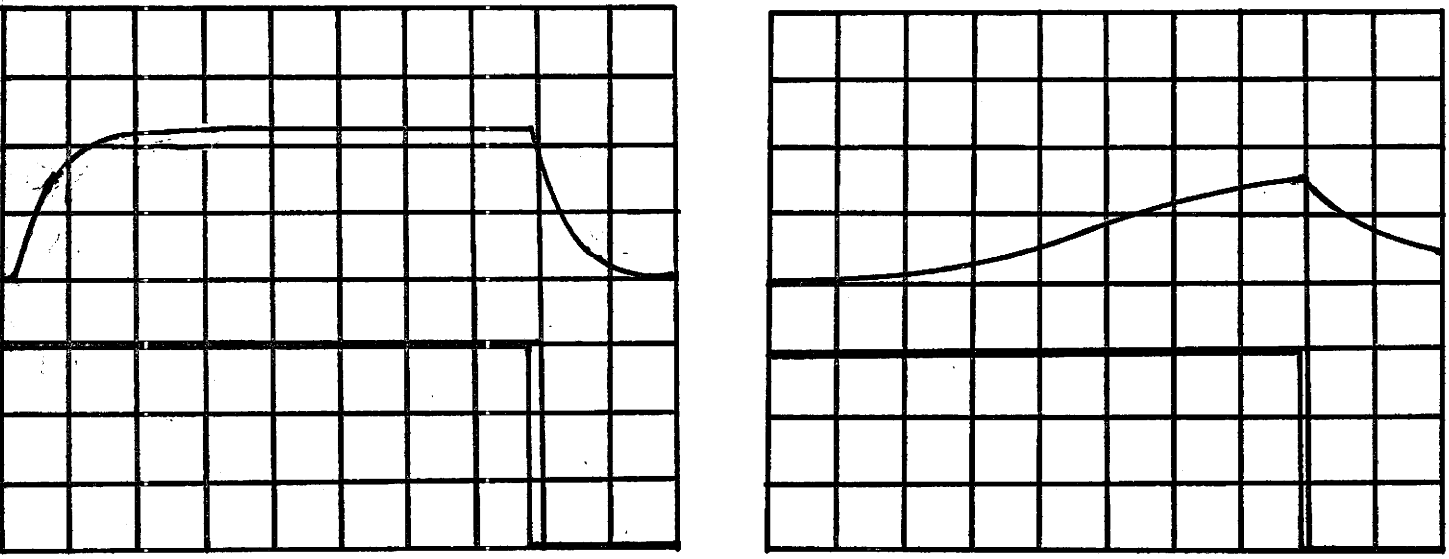
Phototransistors combine good sensitivity (2-3 mA/mW cm^{-2}) with relatively fast response ($\sim$ 4 μ sec. rise time).

A phototransistor type MRD450 was used. Some of the specifications given by the manufacturer are repeated below:

	Min.	Typ.	Max.	Unit
Sensitivity	0.2	0.8		mA/mWcm^{-2}
@ 5 mW/cm^2, from a tungsten source at 2870° K colour temperature				
Rise time	-	-	2.5	μ sec.
for 1 mA peak current				
Wavelength of maximum sensitivity:		800		nm

The sensitivity for the nearly monochromatic light of the GaAs diode is greater than the figure given above since the 900 nm wavelength is close to the wavelength of peak response. However the sensitivity decreases with decreasing irradiance and it is probably close to 10% of the given sensitivity of 0.8 mA/mWcm^{-2}, at the flux density levels involved in the transducer. The other undesirable effect of low light levels is a dramatic increase in the rise time.

The oscilloscope traces (Fig. 1) show the response of the transistor for different intensity light pulses.



(a) (b)

Fig. 1

(a) Response of the phototransistor when "looking" directly into the infrared emitting diode. (Horizontal scale: 50 μ sec/div., Vertical: Top 50 μA/div., Bottom 10 mA/div.)

(b) Response of the phototransistor with the transducer attached on the ear lobe. (Horizontal scale: 50 μ sec/div., Vertical scale: Top 1 μA/div., Bottom 10 mA/div.)

2.4 Transducer Design Considerations

The artifact problem is of major concern in the design of a transducer to be used on active persons.

Three sources of artifacts can be traced to the transducer:

(a) Artifacts due to the mass of the transducer, which as mentioned earlier causes inertial forces,

(b) Forces transmitted to the clip through the connecting leads, and

(c) Changes in the intensity of the ambient light appear as pulses at the output of the detector if it is not covered properly.

It is therefore necessary to keep the mass of the clip as low as possible, the leads very flexible and the area of the clip fairly large to protect the photodetector from ambient light (visible or infrared).

A good way to construct the clip is to avoid the use of springs. Instead, frictional forces at the hinge (Fig. 2b) could be used to retain the the distance between diode and photodetector, which can be rigidly mounted on the clip to retain alignment. (Fig. 2a).

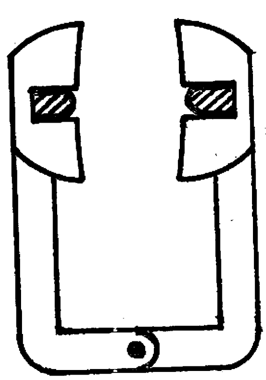


Fig. 2a: Proposed Transducer Construction

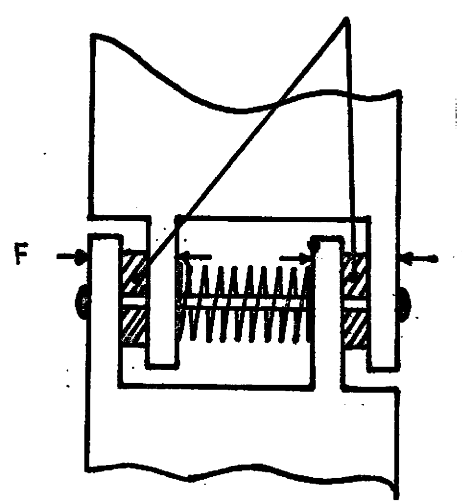


Fig. 2b: Detail of the Hinge

The clip could be opened inserted on the earlobe and then pressed to close to the extent that it is comfortable for the subject. This design minimizes changes in the pressure applied by the clip on the ear lobe.

Other approaches to transducer design are presented in references 5 and 6.

The transducer used for experimentation was built using 7/16" fibreglass washers, as shown in Fig. 3, for the holders (two for each holder) which were joined with a piece of stainless steel wire.

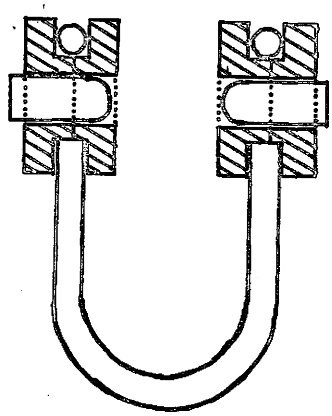


Fig. 3: Transducer used for Experimentation

The protection from the ambient light was not sufficient since pulses of twice the amplitude of the actual output were generated when shading the transducer from the room light. The best solution to the ambient light artifact problem is the use of optical filter passing wavelengths greater than ∿ 800 nm.

Optical techniques may also be employed to expand the size of the radiated beam. This would average out any nonuniformities in the distribution of capillaries in the earlobe, making the amplitude of the output signal independent of the position of the transducer.

A peak current of at least 30 mA was necessary to flow through the infrared emitting diode in order to produce sufficient light for a reasonable phototransistor rise time. The width of the pulse was 400 μ sec. to allow the phototransistor to reach a good percentage of the final value of the current.

CHAPTER 3

SYSTEM

3.1 Conceived System

The output of the transducer is heart beats in sampled form, one sample for each time the infrared emitting diode turns on. Thus to reconstruct the original waveform, a sample and hold circuit must be employed. The block diagram of a conceived system is shown in Fig. 4.

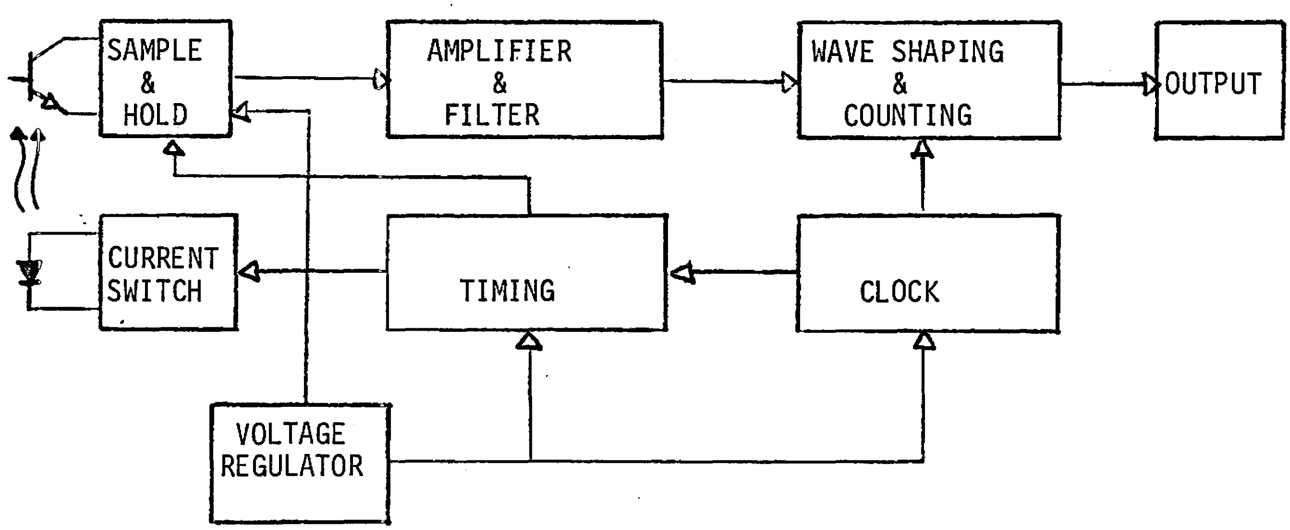


Fig. 4: Block Diagram of a Conceived System

The output of the Sample and Hold is then amplified and filetered to remove frequencies outside the heart beat spectrum ($\sim$ 10 Hz). However since the exact shape of the pulse is not needed, a narrower band pass filter may be used to attenuate noise and artifacts of higher frequencies.

The filtered pulses are subsequently fed into a wave shaping circuit, such as a Schmitt trigger, and then they are counted by a timed counter to determine the rate. The output could be either a digital display of the actual rate or a warning signal when the heart beat rate is not within a preset range.

The clock produces the sampling frequency and can be used to drive a time base to gate the counters.

The timing circuit provides pulses of proper width and phase to drive the current switch for the light source and the sample and hold gate.

A voltage regulator provides constant bias currents to ensure constant frequency of oscillators for the clock, timing pulses and proper operation of the sample and hold circuitry despite battery voltage decrease resulting from aging.

3.2 Sample and Hold

The circuit developed to sample the phototransistor signal and hold it, consists of a preamplifier giving 10 db of signal gain, a low voltage JFET which is gated to transfer the amplifier signal to a storage capacitor and a JFET voltage follower which outputs the changes on capacitor voltage without discharging the capacitor.

Due to the very slow rise time of the phototransistor the "SAMPLE" pulse gating the sample and hold circuit must not coincide with the pulse powering the infrared emitting diode, for if it did, the capacitor would have to discharge first and then rise to its final value - following the response of the phototransistor. This would appear as negative-going spikes at the output. The current pulse to the light source is 400 μ sec. long and the "SAMPLE" pulse 100 μ sec. delayed by 300 μ sec. so that both pulses end simultaneously but the sample pulse begins after the phototransitor transient is over. In the sampling period the time constant through which the capacitor charges/discharges to each new voltage value is 36.3 μ sec. so that at the end of each sampling interval the capacitor is within 6.4% from the final value.

The output voltage of the preamplifier must at all times lie between the battery voltage and the JFET pinch off voltage for the sample circuit to function. In order to accommodate widely varying input signals and low battery voltages, the pinch-off voltage of the JFET must be as low as possible. 2N3687 type JFET transitors were used. They have the following electrical characteristics:

Pinch-off Voltage: V_P = 1.0 Volts

Saturation Current: I_{DSS} = 0.5 mA

"On" resistance (V_{GD} or V_{GS} = 0): R_{ON} = 1.k Ω

The output of the sample and hold is a JFET source follower which operates in the triode region below pinch-off. This is so because with a 1.55 volt supply, the source voltage cannot rise far enough above the gate voltage (this is also a 2N3687) to exceed the pinch-off voltage. This results in a loss of gain of 4.5 db in the output stage but gives very low current draw while maintaining a low output inpedance of about 2 kΩ.

During the pulsing of an LED with optical transmission through the ear lobe this preamplifier, sample and hold circuit produces typically a 150 mV signal which is modulated about 4% or 6 mV by the heart beat rate (Fig. 5).

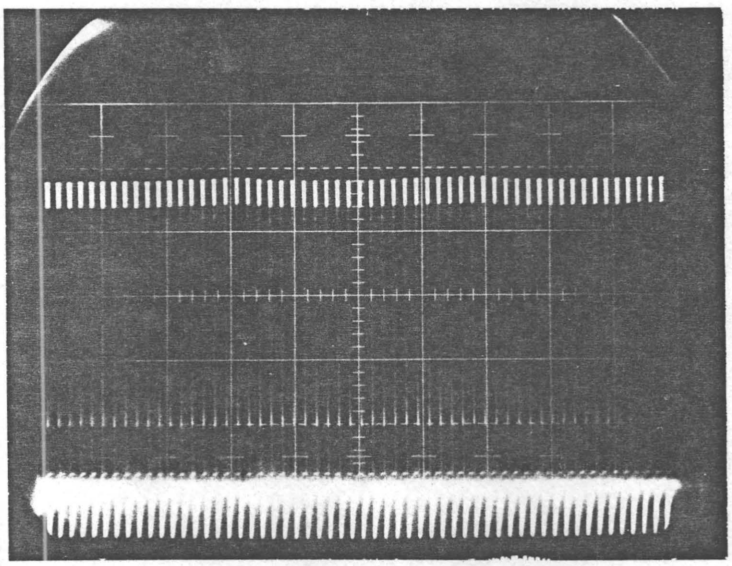


Fig. 5: Output of the preamplifier (Vertical 50 mv/div., Horizontal 0.2 sec/div.). The duty cycle has been increased to 10% for photography clarity. The modulation can be seen as a slight ripple on the top of the waveform.

In order to reconstruct the heart beat rate pulse accurately the pulse rate must be significantly higher than the heart rate and sufficiently high so that the discharge of the "Hold" capacitor does not mark the 6 mV signal. The discharge rate of this circuit was 7 mV/sec. and therefore a repetition rate of 24 Hz was chosen for the LED pulsing and sampling in order to maintain 5% accuracy in the sample and hold circuit.

The preamplifier and sample and hold circuit requires a total of 45 μA at 1.55 volts to operate.

3.3 Amplifier-Filter

The frequency spectrum of the heart pulse extends from about 0.6 to 10 Hz. If the exact shape of the pulse is not important, but only the rate as is the case, the frequency response of the output amplifier may be limited to reduce artifacts and noise.

An LTI type LD505 integrated circuit was used as amplifier-filter (Fig. 6). It consists of a high gain amplifier with differential inputs and single ended output connected to the base of an output transistor. The feedback loop is internally closed from the emitter of the output transistor to the inverting input.

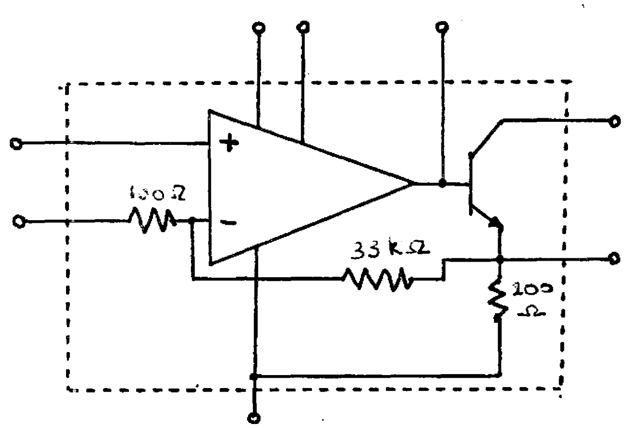


Fig. 6: Functional Diagram of the LD505

The 200 Ω emitter resistor and the 100 Ω resistor at the inverting input are also included on the chip. The input bias currents are set internally so that the unit must be capacitively coupled to any external circuitry.

The internal feedback resistor was utilized in conjunction with two capacitors and an external resistor to form a bridged T feedback network, as shown in Fig. 7. This feedback network makes the amplifier frequency selevtive (band pass) with center frequency fo = 1.6 Hz and

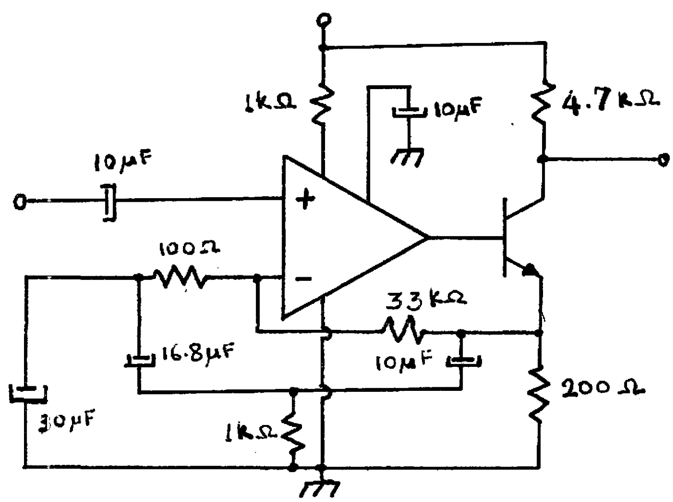


Fig. 7: Narrow band pass amplifier using the internal 33 kΩ feedback resistor of the LD505 with two external capacitors and a resistor to form a bridged T network.

a bandwidth of 0.6 Hz as can be seen from the diagram of gain vs frequency (Fig. 8).

The addition of 1 kΩ resistor in series with the power supply connection of the preamplifier results in significant reduction in current drain and a slight decrease in the overall gain.

As can be seen from the graph of gain and current drain (Fig. 9) vs. battery voltage, the preamplifier appears as a 1.8 kΩ resistor as far as dc current drain is concerned; therefore the addition of the 1 kΩ resistor results in a 21% reduction of the current. The voltage across the preamplifier is also reduced by the same amount. If, however the battery voltage is within the range 1.4-1.55 volts the voltage across the preamplifier will be in the range of 1.1-1.22 volts with an average drop in gain of only ∿0.5 dB while the current is reduced by an average of 75 μA. The output transistor current also drops by 50 μA to give a total current drain of 275 μA at 1.55 V. As shown in Figure 7, the gain from 0.6 to 10 Hz is 50 db or greater.

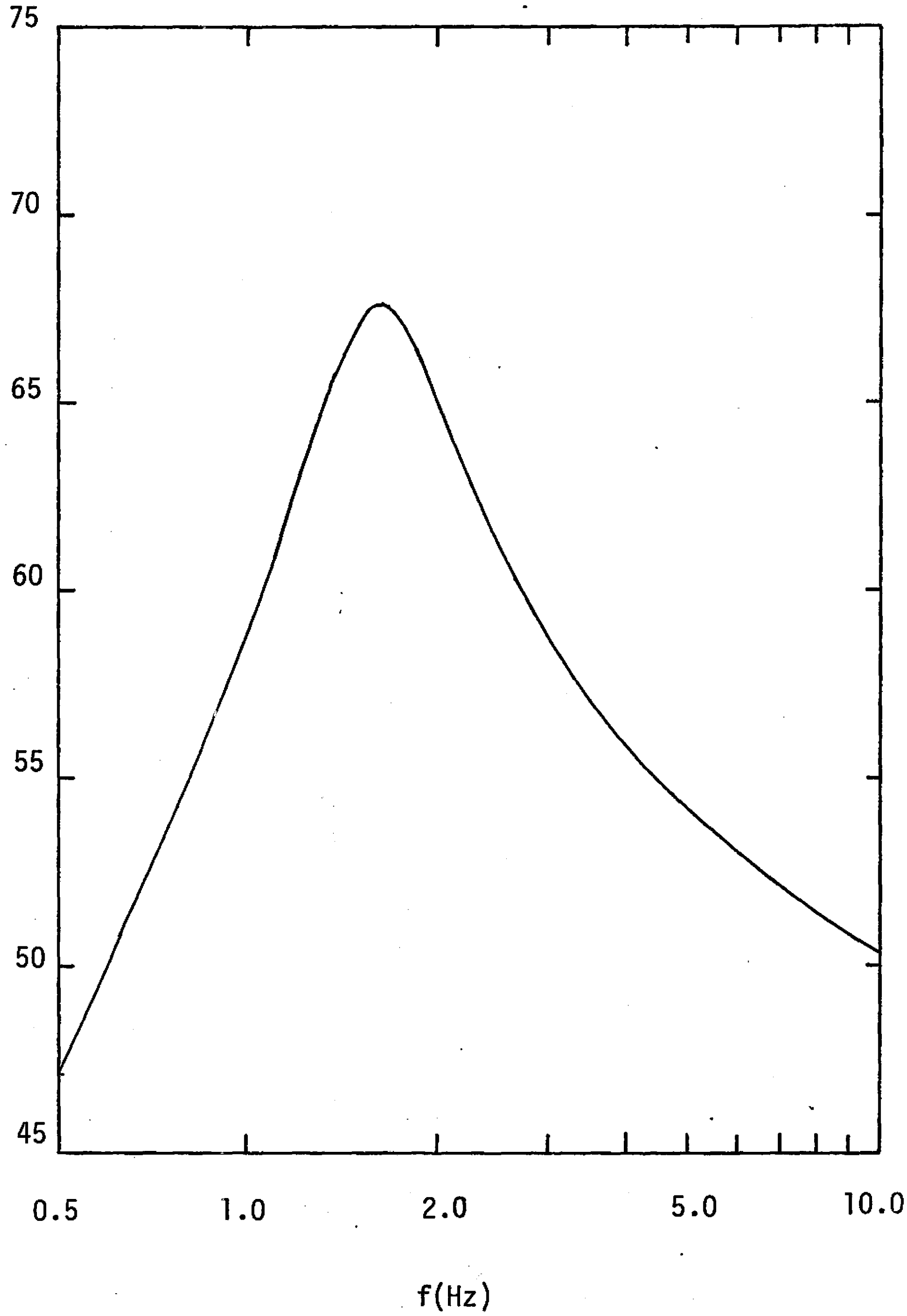


Fig. 8: Frequency response of the amplifier-filter.

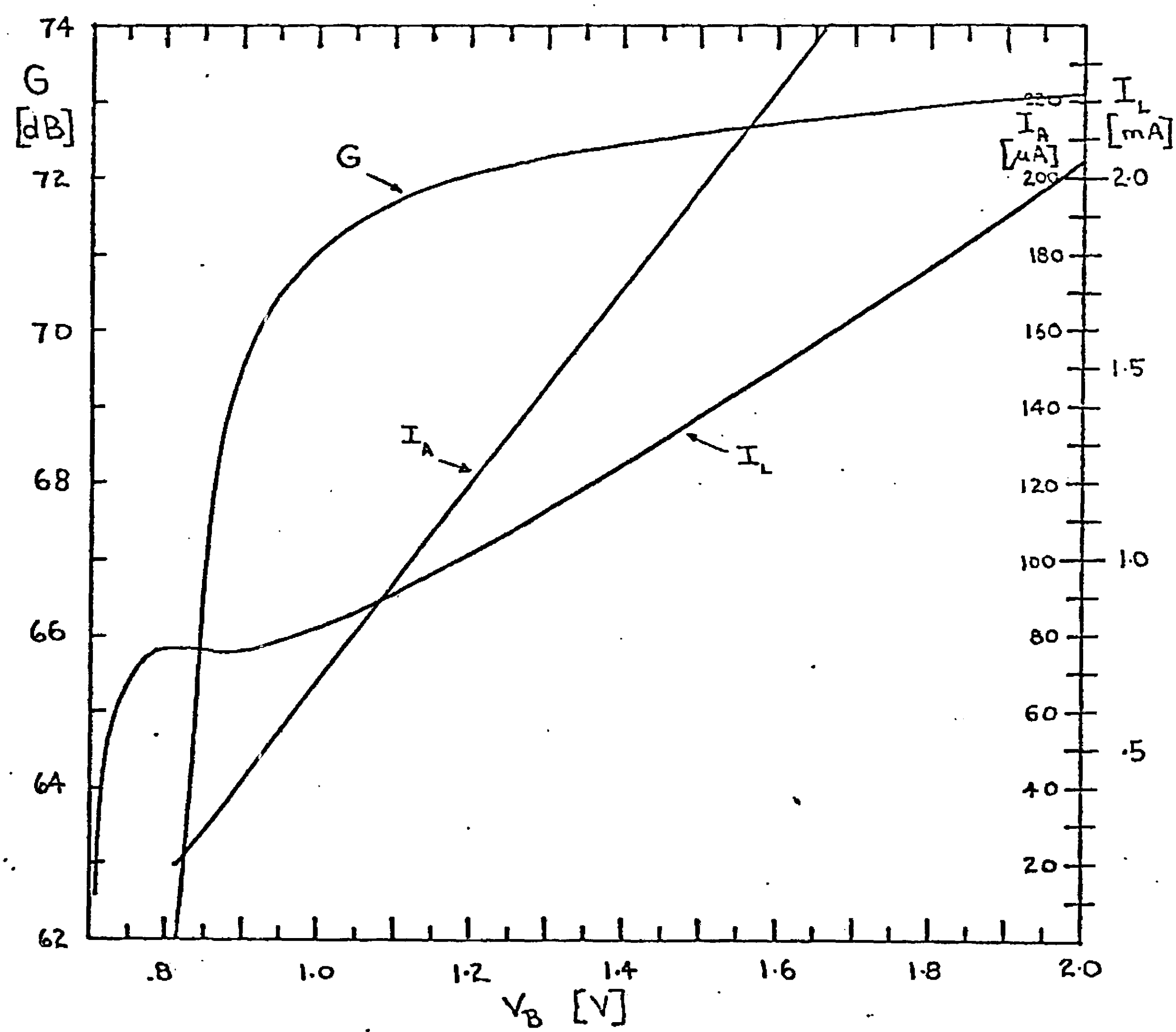


Fig. 9: Effect of supply voltage (V_B) on amplifier gain (G), output dc current (I_L) and preamplifier dc current (I_A).

3.4 Auxiliary Circuitry

3.4.1 Clock

The "clock" consists of an emitter coupled astable multi-vibrator and an amplifier-buffer. The operation of the multivibrator may be analyzed as follows:

Referring to the simplified diagram of the oscillator (Fig. 9) assume that Q_1 has just turned "OFF" and Q_2 "ON". The emitter current of Q_2 (I_e^{ON}) is determined by the two current sources (I):

$$I_e^{ON} = 2I \tag{1}$$

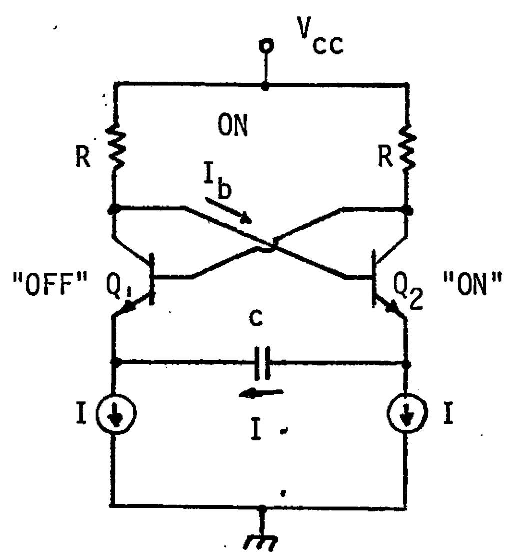


Fig. 10: Simplified diagram of the oscillator

The base collector voltage of Q_2 (V_{Bc}^{ON}) is:

$$V_{BC}^{ON} = V_B^{ON} - V_C^{ON} = (V_{cc} - I_b^{ON}R) - (V_{cc} - I_c^{ON} R)$$

where V_B^{ON} and V_C^{ON} are the base and collector voltages of Q_2 referred to ground. I_b is the base current and R is the load resistance.

or

$$V_{BC}^{ON} = (I_c^{ON} - I_b^{ON})\ R \qquad (2)$$

since

$$I_b^{ON} = I_e^{ON} - I_c^{ON}$$

(2) becomes:

$$V_{BC}^{ON} = (2I_c^{ON} = I_e)\ R$$

using (1) and solving the above for I_c^{ON} we get:

$$I_c^{ON} = I + \frac{V_{Bc}^{ON}}{2R} \qquad (3)$$

The Ebers-Moll equations, written for the direction of the currents shown, read:

$$I_e = \frac{I_{eo}}{1 - \alpha_N \alpha_I} (e^{V_{BE}/V_T} - 1) - \frac{\alpha_I\ I_{ce}}{1 - \alpha_I\ \alpha_N} (e^{V_{BC}/V_T} - 1) \qquad (4)$$

and

$$I_c = \frac{\alpha_N\ I_{eo}}{1 - \alpha_I\ \alpha_N} (e^{V_{BE}/V_T} - 1) - \frac{I_{co}}{I - \alpha_I \alpha_N} (e^{V_{BC}/V_T} - 1) \qquad (5)$$

where:

I_{eo}: the reverse saturation current of the emitter base junction with the collector open.

I_{co}: the reverse saturation current of the collector-base junction with the emitter open.

α_N: the current gain under normal conditions (i.e. the collector junction reverse biased and the emitter junction forward biased)

α_I: the current gain with collector and emitter interchanged

V_T: kT/q

Eliminating the term $(e^{V_{BE}/V_T} - 1)$ among (4) and (5) and using expressions (1) and (3) for the collector and emitter currents we get:

$$e^{V_{BC}^{ON}/V_T} + \frac{V_T}{2RI_{co}} \left[\frac{V_{BC}^{ON}}{V_T} \right] - \left[1 + (2\alpha_N - 1) \frac{I}{Ico} \right] = 0 \quad (6)$$

The collector-emitter voltage of Q_2 (V_{CE}^{ON}) is

$$V_{CE}^{ON} = V_{BE}^{ON} = V_{BC}^{ON}$$

and can be expressed in terms of I and V_{BC}^{ON} through (1), (4) and (6) as:

$$V_{CE}^{ON} = V_T \ln \left[\left(\frac{\alpha N}{\alpha I}\right) \frac{(2 - \alpha_I) I - \alpha_I \frac{V_{BC}^{ON}}{2R}}{(2\alpha_N - 1) I - \frac{V_{BC}^{ON}}{2R}} \right] \quad (7)$$

As the capacitor charges, the emitter voltage of Q_1 (OFF) will drop and as its emitter base junction becomes forward biased, Q_1 will start conducting. Since the total current flowing to ground is fixed (= 2I), any increase in the emitter current of Q_1 will cause an identical decrease of the collector current of Q_2 (neglecting changes in the base current of Q_2).

This means that the collector voltage of Q_2 will increase and the emitter-base junction of Q_2 will become more forward biased. The circuit now exhibits positive feedback and it will change state as soon as the loop gain reaches unity.

$$- \frac{\partial(-I_c^{ON} R)}{\partial V_{BE}^{OFF}} \cong \frac{\partial(I_e^{OFF} R)}{\partial V_{BC}^{OFF}} = 1$$

Q_1 is in the normal operating region (its collector junction is reverse biased) so that (4) reduces to:

$$I_e^{(OFF)} = \frac{I_{eo}}{1 - \alpha I \alpha N} (e^{V_{BE}^{OFF} / V_T} - 1) + \frac{\alpha I \, I_{co}}{1 - \alpha I \alpha N}$$

(The superscript "OFF" refers to Q_1)

therefore:

$$\frac{\partial(I_e^{OFF} R)}{\partial V_{BE}^{OFF}} = \frac{R}{V_T} \frac{I_{eo}}{1 - \alpha_I \alpha_N} e^{V_{BE}^{OFF} / V_T} = 1$$

from which one obtains the base-emitter voltage of Q_1 (V_δ), at which change of state takes place:

$$V_\delta = V_T \ \ln \left[\frac{V_T}{I_{eo} R} \ (1 - \alpha_I \ \alpha_N) \right] \tag{8}$$

Now the voltage across the capacitor, the moment the circuit switches, can be found:

$$V_{cap} = V_{BC}^{ON} + V_\delta - V_{BE}^{ON}$$

or

$$V_{cap} = V_\delta - V_{CE}^{ON} \tag{9}$$

Since the capacitor charges and discharges through constant current I, the voltage across the capacitor will have a triangular waveform of slope

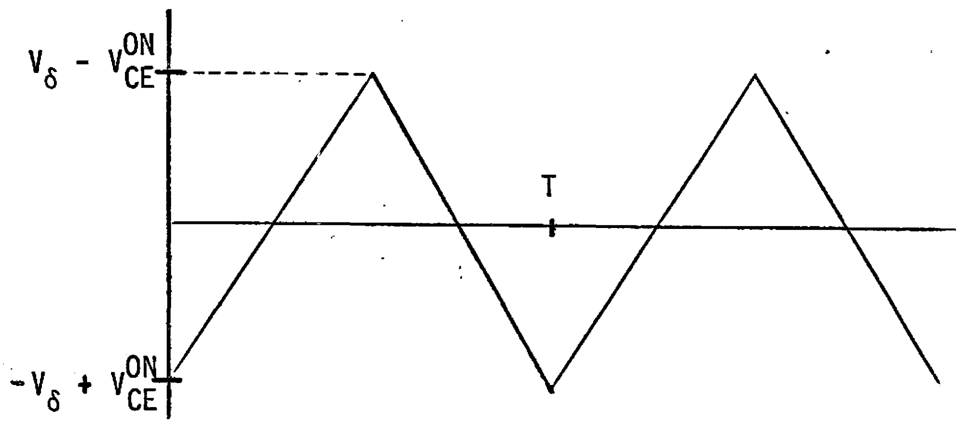


Fig. 11: Voltage waveform across the capacitor

f I/C (Fig. 11), and the period will be:

$$T = 4\frac{C}{I}\,(V_\delta - V_{CE}^{ON})$$

or

$$f = \frac{I}{4C}\;\frac{1}{V_\delta - V_{CE}^{ON}} \tag{10}$$

Equations (6), (7) (8) and (10) may be used to calculate f for any given set of parameters.

Calculated and measured values of the maximum capacitor voltage and frequency of oscillations graphs (Fig. 12 and 13) show fair agreement.

From the above analysis it is clear that there is no first order dependence of the frequency on the battery voltage.

Stability of frequency better than 1% was obtained in the range of 1.4-1.55 volts.

A second point to be noticed is that the frequency vs. current curve exhibits a minimum. If the oscillator is operated at that point, small variations of the current of the sources will not affect the frequency.

A square wave output may be taken from the collector of Q_1 or Q_2. The amplitude of this output is:

$$V = IR + \frac{V_{BC}^{ON}}{2}$$

which is of the order of 0.6 V($I = 10\ \mu A$, $R = 33\ k\Omega$, $V_{BC}^{ON} = 0.52$ V). The oscillator is followed by an unbalanced differential amplifier and a pnp transistor output stage. The output swings practically from 0 Volts to Vcc. The oscillator amplifier was not optimized for minimum current drain and even though the current drain of the multivibrator is approximately 30 μA, the total current drain of the circuit is about 110 μA at 1.55 Volts.

(b) Infrared Emitting Diode Pulser

The LED pulser circuit consists of a monostable multivibrator, triggered by the clock, which drives a current switch to turn the diode "ON". The pulse width of the monostable is about 400 μs. At a current of 30 mA the voltage drop across the infrared emitted diode (MLED 900)

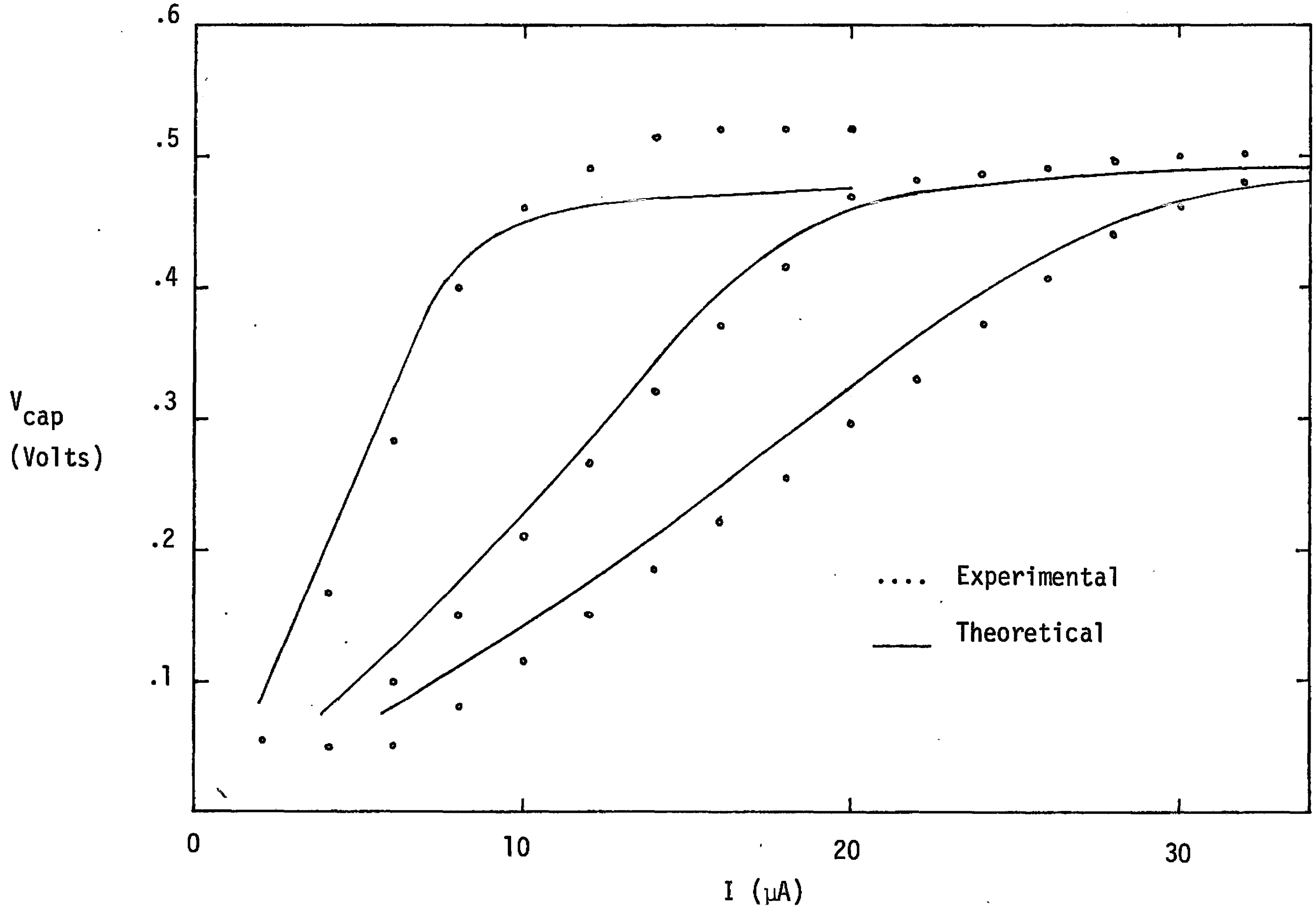


Fig. 12: Peak voltage across the capacitor vs. current of the current sources, for different load resistances.

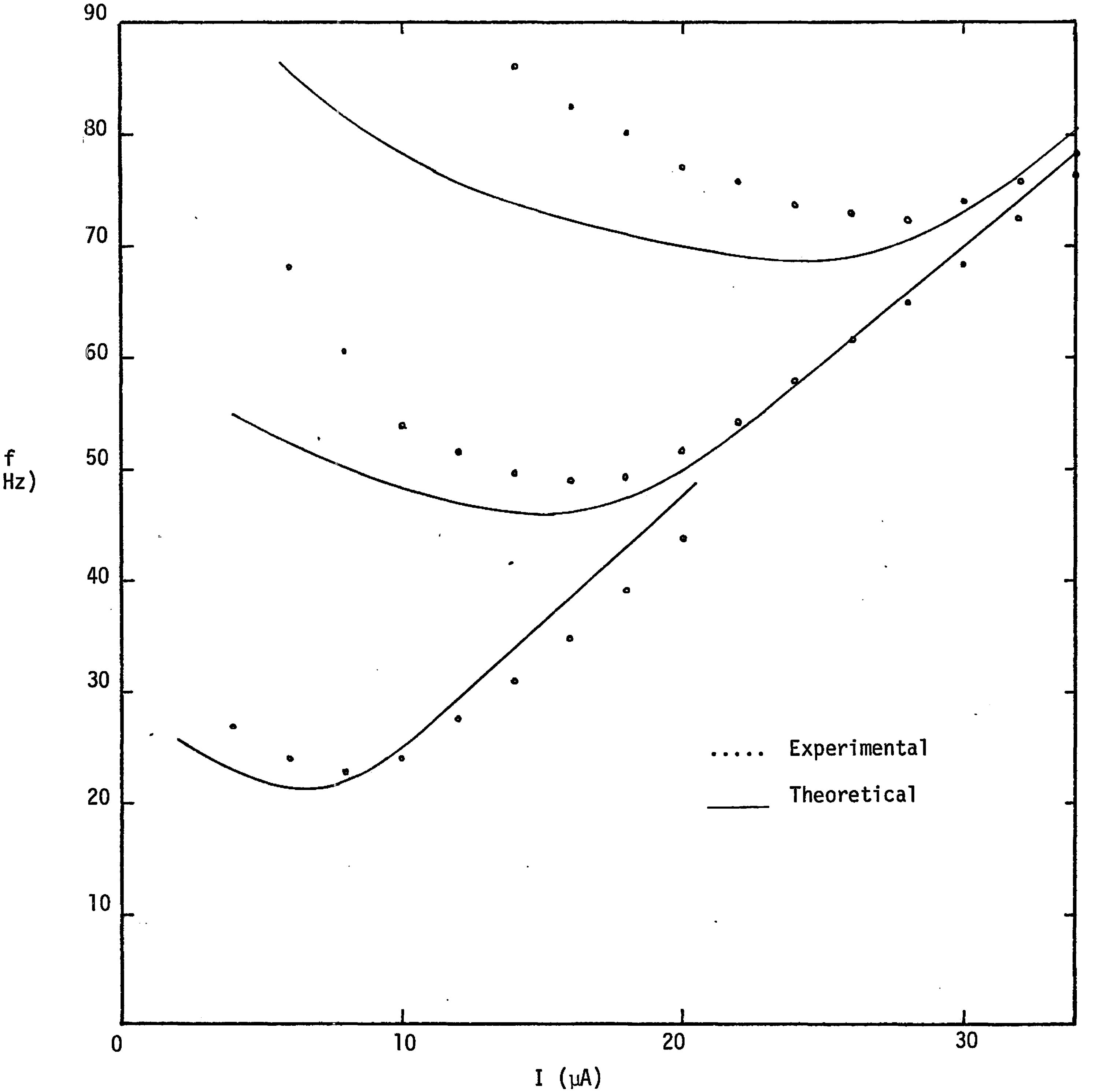
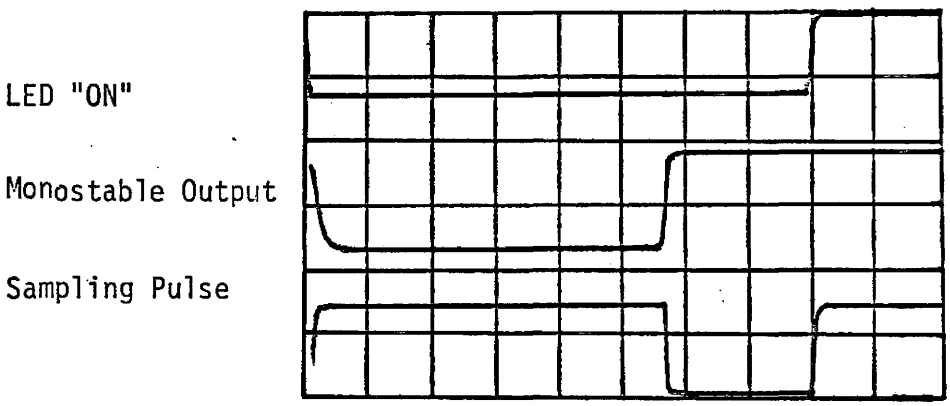


Fig. 13: Frequency of oscillattory vs. current of the current source for different values of load resistance. The value of the capacitor is 0.22 μF.

is about 1.1 Volts. If operation to 1.4 Volts battery voltage is desired, then, assuming battery resistance of 4.7 Ω, the "ON" resistance of the current switch must be no larger than 5.3 Ω. With such a low "ON" resistance it is difficult to control the current over the range of battery voltages experienced. However excess current only increased the battery drain, it will not harm the diode which in the pulsed mode with low duty cycle can take over 500 mA of current. The quiescent current of the circuit is about 270 μA and the LED average current is 380 μA at 1.55 V.

(c) Sampling Pulse Derivation

As already described, it is necessary for the "sample" pulse to the sample and hold circuit to be delayed by 300 μsec with respect to the LED pulse in order to reject the photo-transistor transients. This is accomplished by means of a 300 μsec monostable multivibrator triggered by the clock. This 300 μsec pulse is used to drive a gate which permits only the last 100 μsec of the LED driver pulse to reach the sample and hold circuit. This is illustrated in the timing diagram of Figure 13. The current required for this circuit is 180 μA at 1.55 V.



Vertical: 1 Volt/division

Horizontal: 50 μsec/division

Fig. 14: Relation of the various pulses produced by the circuits.

The voltage reulator used to provide constant voltage to various parts of the circuits, is an integral part of another integrated circuit, the WC501, made by Linear Technology Inc., with a nominal voltage output of 1.05 volts. It draws an unloaded current of 80 μA at 1.55 V.

(d) Battery Line Decoupling

Assuming internal battery resistance 4.7 Ω and a minimum infrared emitting diode "ON" current of 30 mA, the battery line voltage will drop by 140 mV every time the light source is turned "ON". These negative going pulses make battery line decoupling circuitry necessary. This problem was approached using the circuit shown in Fig. 14.

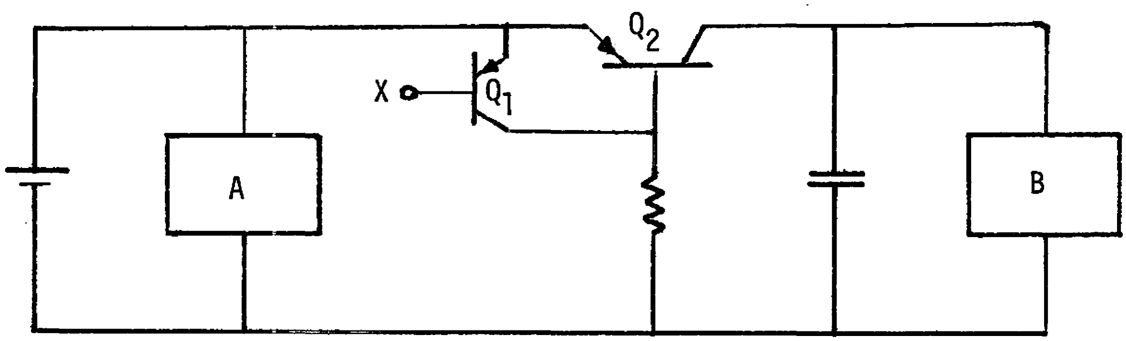


Fig. 15: Decoupling is achieved by switching Q_2 "OFF" when the infrared emitting diode is "ON"

The box labelled "A" represents the infrared emitting diode and the associated circuitry including the clock and all the timing circuitry which is not sensitive to the line ripple. The Box labelled "B" contains all the circuitry that is sensitive to battery line ripple such as the Sample and Hold circuit and the amplifer.

Point "X" is connected to the LED Pulser Circuit. The operation of the circuit may be explained as follows:

Q_1 is normally "OFF" so that Q_2 is saturated and current flows through to power and charge the capacitor C.

When the infrared emitting diode is turned "ON", Q_1 is turned "ON" which in turn cuts off Q_2. In this way circuitry B does not "see" the voltage pulse on the battery line. Current to power "B" is now drawn from the capacitor C which of course discharges. If I is the current drawn by "B" then at the end of the period (τ) for which Q_2

is "OFF" the voltage across C will have dropped by:

$$\Delta V \cong \tau \cdot \frac{I}{C}$$

The maximum current that the amplifier and the Sample and Hold draws at Vcc = 1.55 Volts is I = 380 μA and with τ = 400 μS.

$$\Delta V = \frac{0.15}{C} \text{ Volts/}\mu\text{F}$$

The decoupling circuitry draws about 10 μA average current at 1.55 volts.

CHAPTER 4

RESULTS AND DISCUSSION

An electro-optical heart rate monitor system has been designed and built. It consists of a 24 Hz clock which provides the timing for a 400 μsec LED switch and a gated preamplifier sample and hold circuit used to amplify and reconstruct the heart beat modulation of the phototransistor detection signal. This reconstructed signal is then amplified by a narrow band amplifier to levels acceptable for digital or analogue counting. The complete system operates from a 1.55 V battery and draws a total of 1.35 mA. Typical output waveforms for two different heart rates are shown in figure 16.

While feasibility of low power operation has been demonstrated, artifact immunity remains a question and several areas of the system could be greatly improved. Some of these are discussed below.

The peak to peak output signal voltage varies between 50 mV and 500 mV depending on the battery voltage even when the light source is powered from a separate constant voltage. The main reason for this variation is the poor biasing method used for the input transistor of the Sample and Hold circuit. When the battery voltage drops the transistor current is lowered with corresponding decrease in the gain of the stage.

A lower saturation resistance transistor must be used for the infrared emitting diode current switch since the one used fails to provide adequate current at 1.4 Volts battery voltage.

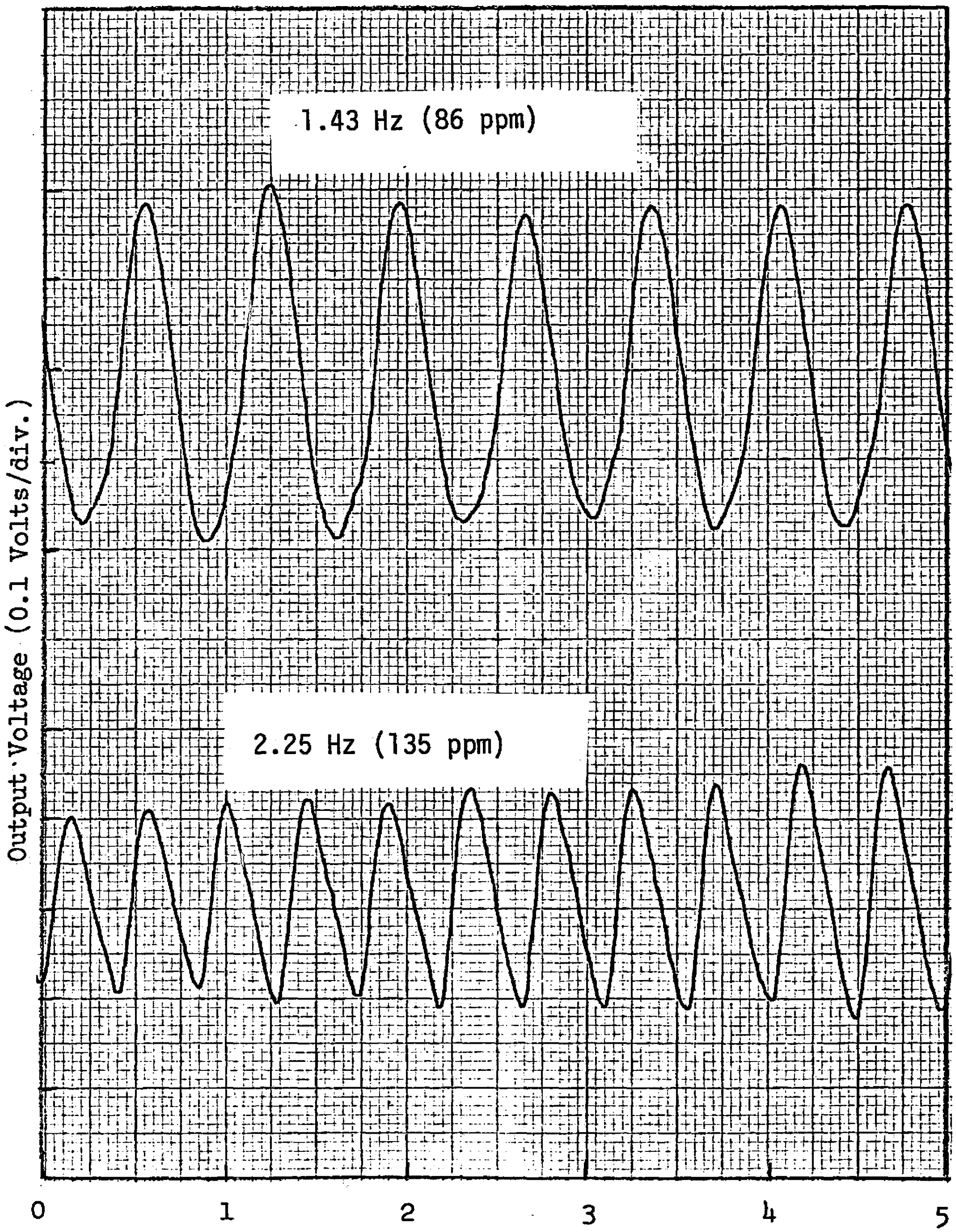


Fig. 16: Typical Output Waveforms.

The particular infrared emitting diode used is not the most efficient existing diode (it was the most efficient readily available). The Monsanto made type ME7161 infrared emitter has a light output of 3 mW at 50 mA (compared to 0.55 mW @ 50 mW of the MLED 900 used) but at slightly higher voltage (1.3 V). Unfortunately no further details were available but it is expected that a current of 10 mA would be sufficient. In such a case the diode driving transistor used would probably be sufficient. Lower current would reduce the battery line ripple and extend the battery life.

Battery line decoupling is a problem yet to be solved. Even though the circuit used is better than any R-C type filter, its efficiency strongly depends on the current load of the decoupled side. A possible improvement would be to use a second such decoupling circuit for the most sensitive part of the circuit which is the output FET of the Sample and Hold.

Transducer design has also to be improved as indicated in Section 2.4. Even though the transducer used had a rather poor ambient light and movement artifact rejection, it gave a useable output signal when located anywhere on the earlobe.

Lastly, the band width of the amplifer must be somewhat increased because it strongly distorts the pulses depending on how far their frequency is from the centre frequency of the amplifier - which might cause false triggering of the following stages.

Digital processing of the output signal is considered in references 8 and 9.

CHAPTER 5

CONCLUSIONS

It is possible to use the best available transducer, namely the transmission type optoelectric transducer in a circuit powered by a single cell battery. However the system complexity increases due to the fact that power conservation makes pulsed mode operation and thus timing "Sample and Hold" and decoupling circuitry necessary.

Battery line decoupling needs considerable improvement if the circuit is to operate properly.

The circuit, with the exception of the capacitors and, perhaps, the two FET's can be integrated. However, some of the stages have to be redesigned using npn transistors in place of the pnp's.

Provisions should be made to allow for external trimming of the clock frequency.

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