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Hardware Development of Wearable ECG Devices*

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3.1 Introduction

Personal healthcare devices find many applications where noninvasive monitoring of biopotential signals is required. Among physiological parameters ECG (Electrocardiogram) is one of the most important vital signals because it directly reflects the heart condition. Wearable devices used for recording ECG related information may continuously record heart rate and/or ECG for several hours or days and store it on the system memory. The stored ECG can then be used by cardiologists for subsequent analysis and diagnosis. Holter monitors, developed for the first time by N. J. Holter [44], are commonly used for this purpose. Event monitors record ECG for short periods which are set by the user control [87]. More functionality may be incorporated into the monitoring device by providing interface to PC (Personal Computer), mobile phone and PDA (Personal Data Assistant) through USB (serial Bus Interface) or other standards. In advanced monitors short range transceivers send stored ECG to the control center in the hospital.

Meeting medical standards, size, weight, power dissipation and cost are important factors in battery-operated personal or portable ECG monitoring devices. These specifications imply special care in the design and implementation of the internal hardware. In this chapter common functional modules of the signal conditioning Electronics in these devices are introduced. In general, system may be implemented using commercially available chips or special purpose integrated circuits. In either case the device may be a single lead or multi-lead depending on the requirements. Detailed circuits will be given for each module based on the prototypes developed by a design team in IIT (Indian Institute of Technology) Bombay.

*An invited chapter.

S. Chaudhuri et al., *Ambulation Analysis in Wearable ECG*, DOI: 10.1007/978-1-4419-0724-0_3, 27

3.2 Basics of Personal ECG Instruments

3.2.1 System Modules and Operation

A personal ECG monitoring system, called **Silicon Locket**, has been developed in IIT Bombay. Block diagram of **Silicon Locket** is shown in Fig. 3.1.

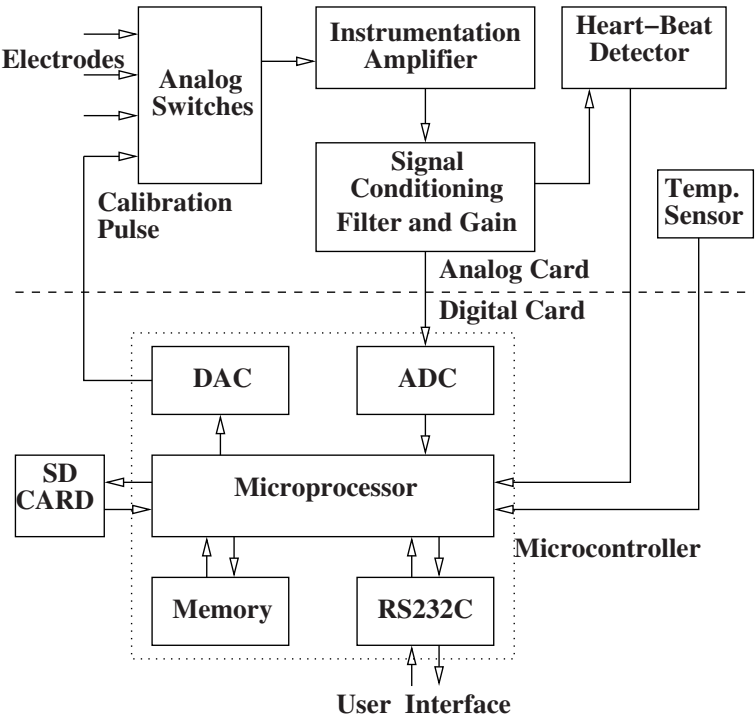


Fig. 3.1. Block diagram of **Silicon Locket**, a personal ECG monitoring device. Interfacing modules may be different among different products.

To be able to understand the requirements of the hardware we should start from ECG itself. ECG signals from electrodes are inherently low voltage analog potentials, ranging from 0.1mV to 4mV [142]. These signals are more often mixed with common mode noise which naturally exist on the human body. An example of such common mode noise is mains supply interference which is induced on the body due to coupling capacitances between the body and AC power boards. Amplitude of the common mode noise is normally large compared to the amplitude of ECG signal. Therefore an INA (instrumentation amplifier) is required at the input stage of the ECG acquisition system to attenuate the common mode noise and amplify the ECG signal which is a differential mode signal, without adding additional noise. Number of ECG

channels, varying from one to twelve channels, determines the number of required instrumentation amplifiers if simultaneous acquisition of ECG through all channels is required. Otherwise a multiplexing scheme at the input may be used.

Typical bandwidth of ECG signal is from 0.05Hz to 150Hz and maximum up to 200Hz [142, 91]. Therefore it should pass through a bandpass filter. However it is possible to highpass filter the signal before or inside INA itself [116, 117]. INA is then followed by an active bandpass/lowpass filter, as shown in Fig. 3.1. In case of heart beat detection cut-off frequency of the lowpass filter can be as low as 40Hz [42].

To store the ECG signal for further processing it is digitized. High resolution ADCs (analog to digital converter) with sampling frequencies not more than few KHz are used for this purpose. Usually a low power microcontroller provides all control signals for multiplexing, sampling and digitization, interfacing with memory and heart beat detection. Nowadays ADCs are also integrated on the microcontroller chips. Limited ECG processing for detection of abnormalities may be also implemented by the microcontroller. Recorded ECG can be sent through a wireless link to a base station or through a data transmission link to PC (personal computer) and/or mobile phone or it can be easily stored on a memory card.

3.2.2 System Requirements

Summary of main performance requirements for electrocardiographs is given in [139]. However all of them are not applicable to wearable ECG recorders because these devices are battery operated. For these devices some important performance requirements are given in table 3.1.

Table 3.1. Some of the main performance requirements for wearable ECG recorders

Parameter	**Value**
Minimum range of input signal	±5mV
Minimum tolerable input DC offset voltage	±300mV
Maximum Gain error	5%
Minimum upper cut-off frequency (-3dB)	150Hz
Maximum DC current flowing through each patient electrode	1μA
Minimum input impedance at 10Hz (each lead)	2.5MΩ
Minimum CMRR (common mode rejection ratio) at 60Hz	86dB

3.3 Electrodes

As it was explained in the introduction chapter, Ag/AgCl electrodes are common type of electrodes used for sensing ECG signals. A picture of a single

disposable foam-pad adhesive ECG electrode has been shown in Fig. 1.1. Generally electrolyte gel is used to maintain a good contact between the electrode and the skin. A biopotential electrode in contact with the gel and skin can be represented by an equivalent circuit. The equivalent circuit, briefly is a parallel RC impedance in series with a dc voltage source, called half-cell potential [139]. To reduce the magnitude of the impedance, electrodes should have proper contact with the gel and hence with the skin. Since ECG signals are recorded in differential-mode half-cell potential appears as a dc input offset voltage. Value of this offset voltage can be as high as ±300mV [139]. Therefore ECG signal conditioning circuit should tolerate such a high value of input DC offset voltage.

3.4 Signal Conditioning

Performance of any personal biomedical system depends on the performance of analog signal conditioning. Signal conditioning includes extraction and amplification of differential signals from sensors, with maximum amplitude of a few mV, from a noisy environment. These tasks are mainly achieved by INAs (instrumentation amplifiers) followed by gain and filter stages. Often conditioned signals will be converted into digital form for further digital processing. In the context of ECG main analog signal conditioning modules for a multi-lead ECG acquisition system are shown in Fig. 3.2 [89]. In Fig. 3.2 each

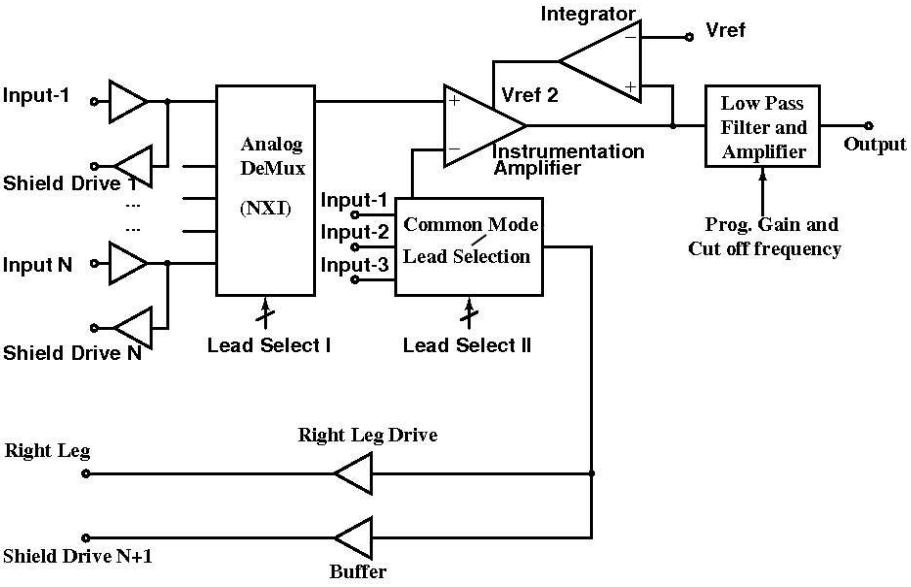


Fig. 3.2. Basic ECG signal conditioning modules.

input **i** ($1 \leq i \leq N$) comes from a single ECG electrode on the body. An analog demultiplxer (**DeMux** module in Fig. 3.2) selects the electrode of which ECG signal should be applied to the input of INA. Another input of INA either receives ECG signal from another electrode (in the case of limb leads) or receives average of two/three ECG signals (for example Wilson central terminal).

3.4.1 Implementation Using General-Purpose ICs

It is always possible to implement a signal conditioning circuit using available general-purpose ICs. INA as the main module of signal conditioning may be implemented using opamps (operational amplifiers) and resistors. Fig. 3.3 shows a commonly used three-opamp configuration for implementing an INA [139].

The INA, shown in Fig. 3.3, constitutes three operational amplifiers and seven resistors comprising three matched pairs and one single resistor R1. Direct connection of input differential voltage to the opamp terminals provides an effective high input impedance for INA. Therefore loading of INA on the sensor will be almost negligible. However sometimes bias resistors at the input of INA are provided to establish a DC bias voltage at the input of INA. In this case value of resistors should be high enough to provide a high input resistance, e.g. at least 2.2MΩ.

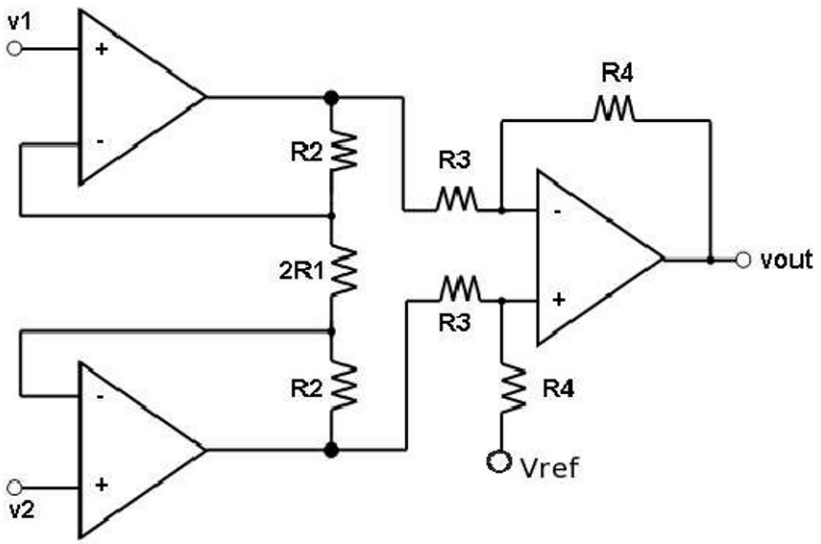


Fig. 3.3. Three-opamp configuration for INA implementation.

It is straightforward to derive the input-output characteristics of the INA, shown in Fig. 3.3. Considering ideal opamps the input-output voltage relationship is given by Eq. (3.1).

$$\frac{V_{out} - V_{ref}}{V_{in+} - V_{in-}} = \frac{2R_2 + R_1}{R_1}\frac{R_4}{R_3} \tag{3.1}$$

Eq. (3.1) is in fact input-output equation for a differential amplifier. As shown in table 3.1 high CMRR (common mode rejection ratio) is one of the main specifications of INAs. The value of CMRR often should be more than 90dB for wearable ECG recorders. Opamps used in three-opamp should have high CMRR. Still three-opamp INA needs accurate matching of the resistor pairs that implement the feedback networks (Fig. 3.3). CMRR degrades by the amount of 6dB for every 2% mismatch between resistor values. For a discrete implementation resistors with accuracy matching better than 1% and opamps with CMRR better than 90dB are required. Since equivalent input noise voltage of two input opamps appears directly at the input of instrumentation amplifier these opamps must be low noise. For a proper design RMS (root mean square) value of the input noise voltage of each opamp over ECG frequency band should not exceed 5μV.

General-purpose INA ICs overcome the matching requirements of three-opamp configuration by integrating the entire circuit in one IC. Therefore system developers may use these available INA ICs directly on the ECG signal conditioning board.

Mains power supplies cause interference currents flow through the human body [85]. The interference current is induced due to coupling capacitances between the human body and power lines as well as between human body and ground. There are different paths for this interference current to flow. Some part of the current flows through the body, which generates a common mode signal, at frequency of 50Hz (or 60Hz), on the body. Some part of the current may flow through the input impedance of INA if INA is AC-coupled. AC-coupled INA is desired because INA should tolerate at least ±300mV input DC offset voltage.

If all components between the body and input terminals of the instrumentation amplifier are matched still CMRR of the signal conditioner module is a finite value and hence a small portion of the induced common mode signal appears as 50Hz (or 60Hz) noise in the recorded ECG. This noise disturbs the ECG, specifically P-Wave and T-Wave part of it. Using a sharp notch filter with center frequency of 50Hz (or 60Hz) seems to be a solution for reducing the interference. However care should be taken to avoid ECG distortion. Therefore interference reduction is an important practical implementation issue even for portable ECG monitoring devices which are operated by batteries. One of the common ways of attenuating the common mode noise is to drive the right leg using a common mode negative feedback circuit as shown in Fig. 3.4. Right leg drive circuit is also used in bedside electrocardiograph systems [139].

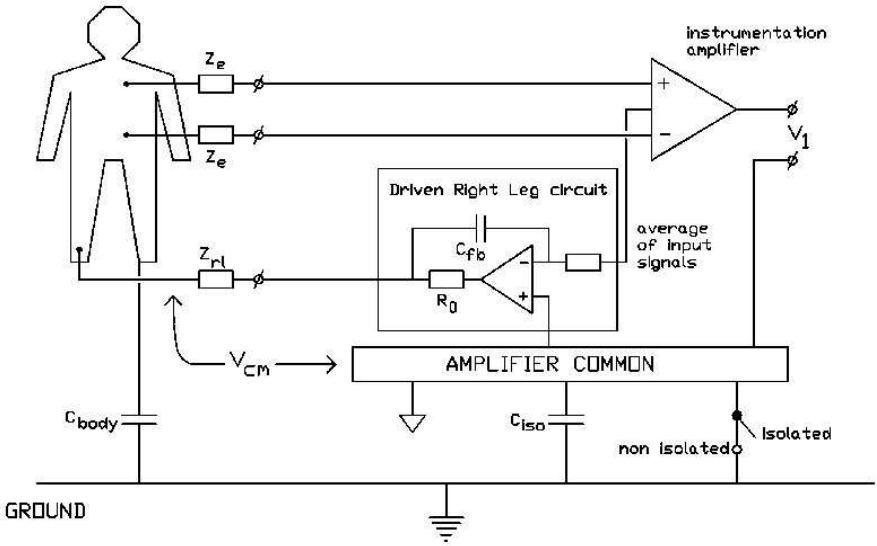


Fig. 3.4. Right leg drive circuit(Used with permission from Biosemi Inc. [85])

For further amplification and filtering of ECG signals more often INA is followed by a LPF (lowpass filter). This lowpass filter is an active filter made by on-chip opamps or OTAs (operational transconductance amplifiers) and discrete resistors and capacitors. The cut-off frequency of the filter is set around 150Hz as specified in table 3.1. However it can be programmable. The overall gain of the analog ECG signal conditioning module is also programmable, usually in the range of 200 to 1000.

Wearable ECG recorders are often interfaced with personal computers, mobile phones or removable memory cards through USB (universal serial bus), IrDA (Infrared interface) or RS232 ports [136]. Therefore digitization of captured ECG signal is required. A low power microcontroller with on-chip ADC (analog to digital converter) is a good choice for digitization of conditioned ECG signal.

3.4.2 ASIC (Application-Specific Integrated Circuit) Design for Signal Conditioning

To reduce the cost, weight and power compact design of personal medical instruments is desired. These requirements motivate development of custom signal conditioning chips. The ultimate goal is the ASIC design as per the recommended specifications for personal health care devices, integration of more functionalities into the chip and compact design of the final monitoring device.

ASIC development for medical instruments is not recent. In fact in 80s research had already started for the development of micropower personal monitoring devices. For example a comprehensive test chip was reported in 1988 for the acquisition of physiological signals [84]. An integrated micropower heart rate indicator with power supply voltage of 10V, developed in 3μ CMOS process, was reported in 1989 [109]. This heart rate monitor was designed for the continuous monitoring of the heart rate. In heartbeat detectors exact retrieval of ECG waveform is not a concern. The detector must only precisely detect QRS peaks to provide a base for counting heart beats. Therefore bandpass filter of analog front end is designed in such a way to attenuate frequencies below 10Hz or above 60Hz. In 1996 a CMOS nine channel ECG measurement IC with complete data acquisition was published [23]. The ASIC was developed in 2μm CMOS process for general measurement purpose, operating at 10V power supply with power consumption of 270mW. Hayes-Gill et al. reported a generic ASIC again comprising analog modules and ADC in [42]. As CMOS technology advanced more functionalities were incorporated in the integrated circuits aimed for biomedical instruments. An example of such test chips was published by Chih-jen, et al. in 1999 [143]. The test chip included analog signal processing unit, transmitter, receiver and digital processing unit. Analog processing unit constituted instrumentation amplifiers implemented using opamps and analog filters. Transmitter included amplitude modulators, analog adder, frequency modulator and RF transmitter. In the continuation of that work an analog processor IC for wireless bio-signal monitor was reported in 2003 [144].

The research has continued in different directions, for example reducing power dissipation or covering more number of physiological signals or improving noise performance and finally making the chip specifications quite robust using programmable gain, filtering and base line drift compensation [68, 91, 117, 142, 144]. High-performance CMOS biomedical signal conditioning ASICs are not usually developed in nano regime technologies due to low frequency operation, stringent noise and offset requirements, and cost. In this regard 0.35μm and 0.5μm CMOS technologies are attractive technologies.

Instrumentation Amplifiers

Instrumentation amplifier (INA) is a key module among all signal conditioning modules. In low power personal health monitoring devices, biopotential INAs with high common mode rejection ratio (CMRR), low input-referred noise voltage and low offset voltage as well as very low power consumption are required. As it was discussed in Section 3.4.1 monolithic implementation of INA using traditional three opamp configuration needs accurate matching of the resistors used in its feedback network to achieve high CMRR. Also this structure is not a proper solution for very low power design. Another approach for the design of integrated ECG INAs is to use current balancing technique [80, 108]. Fig. 3.5 shows the basics of current-balancing technique.

At the transconductance stage, the input difference voltage is converted into a differential current i_g, flowing through resistor R_g. Current i_g is mirrored to the transimpedance section with the unity gain. The mirror current, called i_s, is converted into the voltage by flowing through a resistor R_s. Referring to Fig. 3.5, expression (3.2) applies.

$$i_g = i_s \Rightarrow \frac{V_{out} - V_{ref_{internal}}}{V_{in+} - V_{in-}} = \frac{R_s}{R_g} \tag{3.2}$$

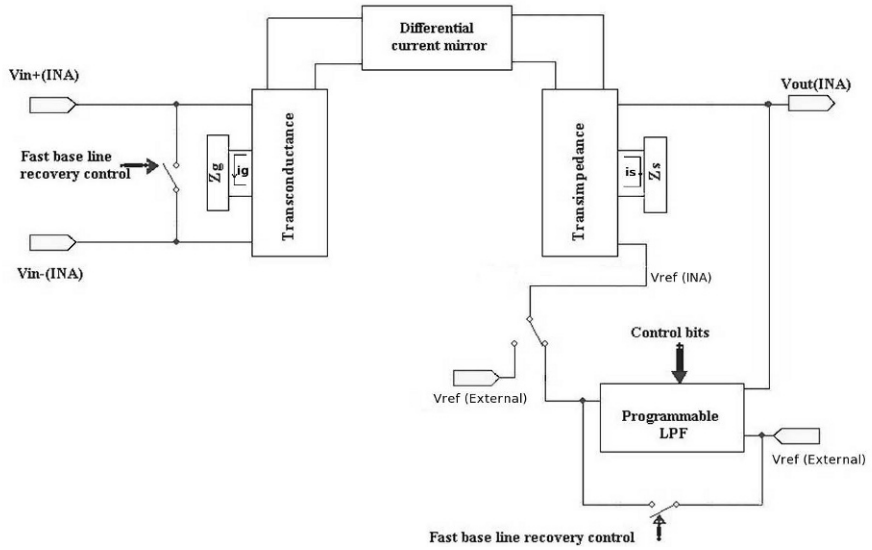


Fig. 3.5. Basics of current balancing technique. (Reprinted with permission from M. Shojaei Baghini, S. Nag, R. K. Lal, D. K. Sharma, "An Ultra Low-Power Current-Mode Integrated CMOS Instrumentation Amplifier for Personal ECG Recorders", Journal of Circuits, Systems, and Computers (JCSC) Vol: 17, Issue: 6, ©2008 World Scientific Publishing Co. Pte. Ltd, Singapore.)

Although current balancing technique is a known method, suitability of this technique for achieving ultra low power ECG signal conditioning chips was reported for the first time in [116]. Integrated INA reported in [116, 118] consumes only 9μA. The circuit has been implemented in a test chip and fabricated in 0.35μm CMOS technology with supply voltage range of 2.8V to 4V. The circuit schematic of the implemented INA is shown in Fig. 3.6. In Fig. 3.6 transistors ML1 and ML2 act as active loads for input transistors M1 and M2. Transistors MGm1 and MGm2 make an internal transconductance amplifier with feedback path to the input stage through cascode current mirrors. This feedback path keeps drain currents of M1 and M2 almost constant. When a differential voltage is applied, output currents of the transconductance ampli-

fier become unbalanced in order to maintain the drain currents of M1 and M2 equal. Therefore input voltage will drop across R_g. At the transimpedance stage transistors M7 and M8, linearized by the internal opamp, convert the input current into a voltage across resistor R_s. Thus input voltage is amplified at the output by the amount of R_s/R_g. Cascode current mirror is used to obtain high CMRR [29].

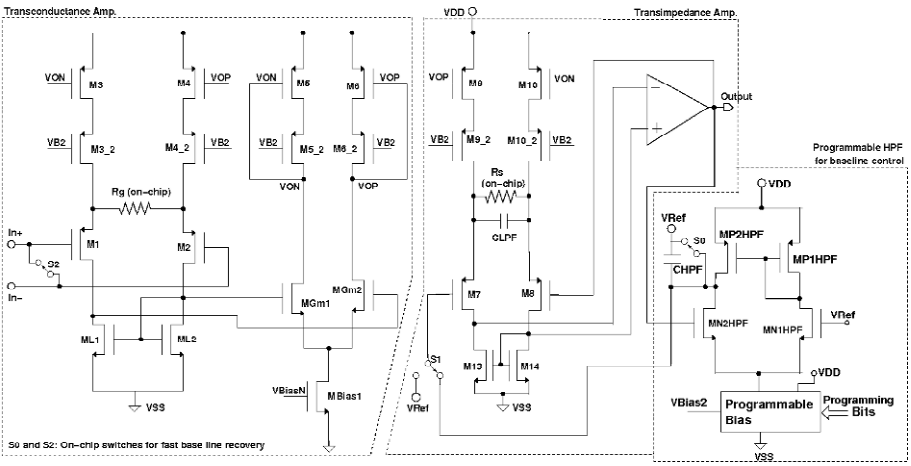


Fig. 3.6. The circuit schematic of ultra low power CMOS INA reported in [116, 117, 118]. (©2004, 2005 IEEE)

Amplifiers and Filters

With current balancing technique it is possible to implement main analog signal conditioning functions by the instrumentation amplifier (INA) itself and so complexity, power and cost of the whole analog circuit reduces. For example, in Fig. 3.6 two external capacitors CHPF and CLPF determine the lower and higher cutoff frequency of the amplifier, respectively [117]. Typical range for corner frequency of lowpass filtering is 150Hz to 200Hz. Typical range for corner frequency of highpass filtering is from 0.05Hz to 0.5Hz. For heart beat detection applications corner frequency of lowpass filtering can be reduced to 40Hz. Programmability of both lowpass and highpass filtering is desired. For example in [118] two control bits program frequency response of HPF. Also by setting both control bits to zero HPF will be disabled and reference voltage of the transimpedance stage will be connected to Vref. It should be noted that highpass filtering also remove out-of-band slow motion artifacts. However for very fast motion artifacts it will take a long time for the base line to recover. In this case to restore the base line to the normal level, on-chip internal switches across input transistors M1 and M2 (switch

S2 in Fig. 3.6) and across CHPF (switch S0 in Fig. 3.6) are used to rapidly discharge external capacitors of INA.

Ultra low power INAs with bandpass frequency response are the main modules for compact integrated analog signal conditioning [116, 117]. In addition to that since amplitude of ECG QRS complex changes from person to person adjustable gain for each channel of ECG signal conditioning is always desired. One simple programmable gain stage after instrumentation amplifiers provides the desired range of channel gains. Typical values of channel gain is from 200 to 1000. The signal conditioning circuit reported in [117] draws DC current of 22μA from 3.3V battery supply voltage for each ECG channel. The front-end differential stage is AC coupled to the body through 1μF capacitors.

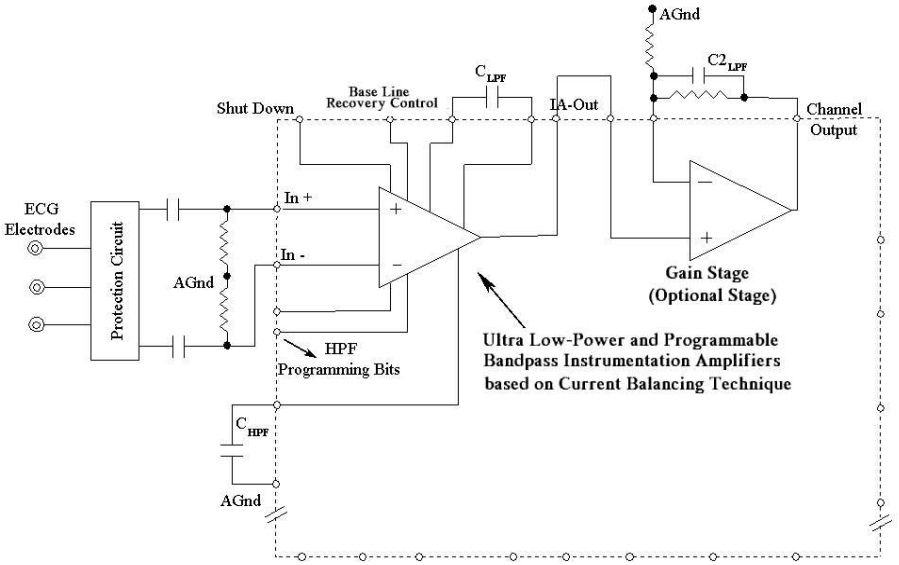
Block diagram of one channel of the compact and low power CMOS signal conditioning chip, called **SLAC1.1**, which is designed in IIT Bombay, is shown in Fig. 3.7(a) [117]. Off-chip components are also shown in the figure. In Fig. 3.7(a) INA, bias generator module and each operational amplifier of the chip draw 9μA, 10.5μA and 8μA dc current from 3.3V battery supply voltage, respectively. Operational amplifiers have phase margin of 70 degree while driving 40pF load capacitor. Unity gain frequency of the opamps was measured 130kHz suited for low frequency biopotential signals. Fig. 3.7(b) shows the chip photo. Table 3.2 shows measured specifications of each channel on the chip [117].

Table 3.2. Performance specifications of each ECG channel achieved by custom ECG signal conditioning chip reported in [117].

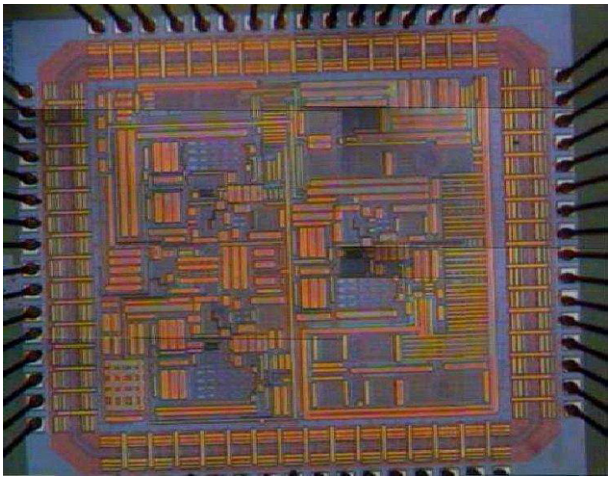
Parameter	**Value**
Vdd	3V$\leq$Vdd$\leq$ 4V
Voltage gain of the channel	200
Input voltage range for high linearity	$\pm$6mV
Tolerable input DC offset voltage	Any value due to AC-coupling
Input referred noise voltage (RMS) (INA noise plus thermal noise of bias resistors)	6 μV (BW=200Hz)
CMRR	100dB (at 60Hz)
Input impedance of each channel at 10Hz	Very high due to AC-coupling
HPF cut off frequency	0 to 0.07Hz (programmable)
LPF cut off frequency	170Hz (adjustable by external capacitors)

Interference Issues

Recording of bioelectric signals is always liable to electrically induced and magnetically induced interference. Mains power supply is a common cause of



(a)



(b)

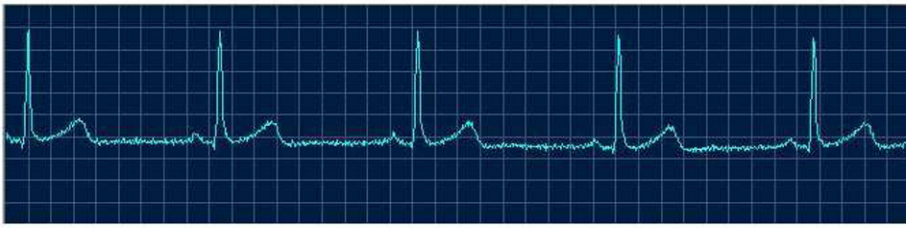
Fig. 3.7. One channel of SLAC1.1 reported in [117], (a) ultra low power CMOS ECG signal conditioning chip and (b) chip photo of SLAC1.1. (©2005 IEEE)

interference for both bedside and personal monitoring devices. The capacitances between the patient and mains power supply cables (or boards) cause an interference current, typically in the range of few μA, to flow through the body [85]. This interference current exists even in personal battery operated ECG monitoring instruments. A part of the interference current flows through right leg drive electrode and enters into the ECG recording device. In return common mode voltage of the ECG instrument changes with frequency of 50Hz (or 60Hz), which appears as common mode noise on the reference level of ECG signal. Therefore INAs with high CMRR (more than 90dB) are required. In addition to common mode noise there is possibility of differential mode noise as well. This is due to the inherent mismatch between impedance of ECG electrode skin interface for every two electrodes. 50/60Hz interference signal on the body is converted into a small differential voltage due to this mismatch. This differential voltage appears at the input of ECG lead, which in turn is amplified and appears as noise on the ECG signal at the output of ECG channel. Fig. 3.8(a) shows recorded Limb lead I ECG using **SLAC1.1** under strong 50Hz interference but with proper shielding. Similar ECG was then recorded without shielding considerations to demonstrate the effect of mains supply interference on the ECG signal as shown in Fig. 3.8(b).

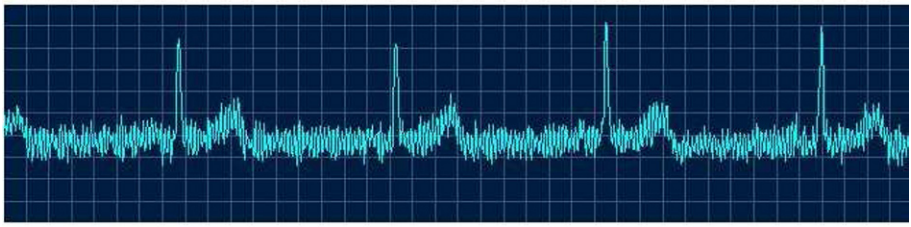
Mains supply interference cannot be easily filtered from ECG because its frequency is in the band of ECG signal. Therefore cables and the entire Electronics of ECG device should be shielded to reduce this noise. Shields should be driven by appropriate signals. The best way is to drive a shield with the signal at the inner wire [85]. However for each input signal an extra shield drive amplifier is required. For example for a simultaneous three lead ECG recording device, three shield drives are required. For higher number of leads, analog demultiplexing reduces the required number of shield drives. For example for a 12 lead arrangement, shown in Fig. 3.2, eight additional op-amps are required for buffering and shield drive purposes. Fig. 3.9 shows the schematic of Wilson Central Terminal generation, guarding and right leg drive circuits for a 64 channel ECG amplifier [85].

3.5 Analog to Digital Converter

ECG is a common non-invasive diagnostic tool for monitoring and detecting cardiac diseases. Main motivation for continuous monitoring of the patients' ECG is to detect transient arrhythmias or transient distortion in ECG waveform, which may not be present during short-time ECG tests in the hospital. ECG is not only analyzed by the cardiologist but also in automatic post-processing ECG analysis systems which act as an assistant to the cardiologist. For example, heart rate variability and ST segment deviations can be automatically detected by post processing of ECG data [18, 61, 110]. This is specially crucial in ambulatory situations. Signal conditioned ECG should be digitized for the purpose of post processing.



(a)



(b)

Fig. 3.8. ECG signal traces from lead-I, using custom ASIC **SLAC1.1**, designed in IIT Bombay, (a) clean ECG signal (b) effect of strong 50Hz interference on ECG signal

Post processing of ECG data is not only used for detection of abnormalities but also sometimes for further noise and artifact removal, specifically in portable and personal health care devices. For example, mains supply interference or motion artifacts can be savior when the patient prefers more freedom in mobility and movements.

Resolution of ADC (Analog to digital converter) in personal ECG monitoring devices depends on the amount of information which will be extracted from recorded ECG or accuracy the cardiologist requires. Resolutions up to 16 bits and sampling rates up to 2k samples/s are used when sophisticated post processing like ST-T micro-variabilities need to be detected [61]. It has been shown that beat-to-beat micro-variations of the T wave, are related to arrhythmia [61]. However these micro-variations are not visually apparent from ECG. On the other hand if sophisticated analysis is not required and noise removals like removal of motion artifacts are concerned resolutions maximum to 12 bits with sampling frequencies as low as 256 samples/s are usually enough [40, 100, 103].

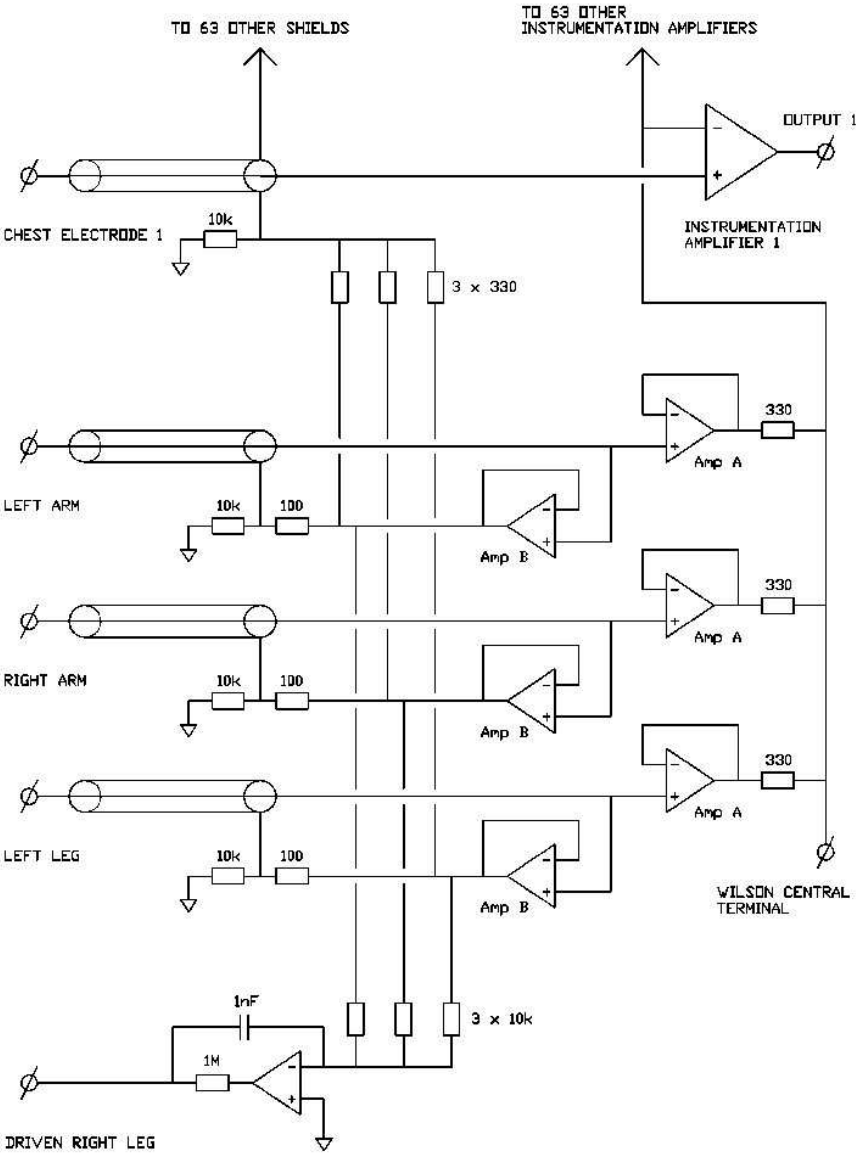


Fig. 3.9. A typical Wilson Central Terminal generation, guarding and driven right leg circuits in front end of a multi-channel ECG signal conditioning circuit (Used with permission from Biosemi Inc. [85])

3.6 Digital Modules

In standard monitoring sophisticated signal processing is not necessary. Therefore the device may display the heart rate or provide crucial but limited functionalities like continuous recording and storage of ECG with or without wireless link and detecting vital abnormalities like arrhythmia. As far as standard monitoring is concerned the entire control and processing tasks can be handled by a low power microcontroller. However diagnostic monitoring needs advanced and highly-accurate DSP algorithms to be embedded into the device. This will drastically increase the power consumption of the medical gadget much beyond the rate at which a battery-operated instrument can work continuously at least for 24 hours. As a result diagnostic monitoring is handled by the base station or central processing unit in the medical control center.

Main digital modules in a personal ECG monitoring device are as follows.

1. Microcontroller (more often with built in data converters)
2. Clock generator
3. Memory modules for storing application software (usually compiled in assembly code), patient data and digitized ECG
4. Display driver (some ECG monitoring devices do not have display)
5. Data transfer interface like USB interface or serial link
6. Data transmission interface like interface with short range ECG wireless transmitter like infrared and Bluetooth.

3.6.1 Microcontroller

Main digital modules of **Silicon Locket** are shown in Fig. 3.1. Low power microcontroller MSP430F149 from TI (Texas Instrument) was used in the first prototype of **Silicon Locket** [90]. This microcontroller has an internal 12 bit ADC with reference voltage of 2.5V. MSP430F149 consumes 280μA current at 1 MHz operational frequency with standby current of 1.6μA [2]. On-chip 60KB flash memory is used for temporary storage of digital data as ECG is acquired and digitized in **Silicon Locket** [90].

3.6.2 Data Storage

In many cases real time transmission of ECG data is not required. Instead cardiologist needs latest ECG data from the patient recorded, say for 24 hours. One of the best ways of storing ECG data is SD (secure digital) card [3]. SD card is a non-volatile memory card which can be easily plugged into portable electronic devices such as cameras, mobile phones, etc. Capacity of the card may change from few Mega bytes to tens of Giga bytes. As shown in Fig. 3.1 **Silicon Locket** uses a removable SD card interface. Patient can easily carry the stored data any time she/he visits the doctor.

3.6.3 Data Retrieval

Data storage, transmission and retrieval are crucial in wearable healthcare instruments. In the context of ECG, data retrieval methods depend on the storage media. For example ECG retrieval from SD card can be addressed in two ways. One way is to provide a removable SD card interface to the instrument and then use a media-card reader. Another way is to provide a reader based on one of the commonly used connectivity standards, such as USB or Bluetooth [60]. In latter case additional conversion chips are required to interface the microcontroller to the peripheral devices. In case monitoring should be real time or acquired ECG should be sent to a base station with minimum delay low power wireless transceivers supporting short range connectivity standards such as IEEE 802.15.4 may be used.

3.7 Discussion

Personal health monitoring technologies provide medical facilities in small devices. The device can be like a mobile phone so that a person can easily carry it or will be inserted in proper textiles so that the person can wear it. Low cost wearable or mobile personal healthcare devices record vital physiological signals or provide standard measures of health. Many of these devices are already available in the market. However to make these devices reliable and affordable for every body research is going on to provide solutions with more functionalities, better performance, less energy consumption, less size and lower price. In this chapter an example of a low cost power efficient custom IC for signal conditioning of simultaneously three channel ECG was presented. From hardware side integration of sensor, signal conditioning, data conversion, control and processing modules all in one chip will significantly improve the performance, cost, size, price and energy consumption of the device. Development of special microcontrollers and processors for wearable healthcare devices plays a vital role in embedding more algorithms for annotation and post processing of recorded or measured signals.