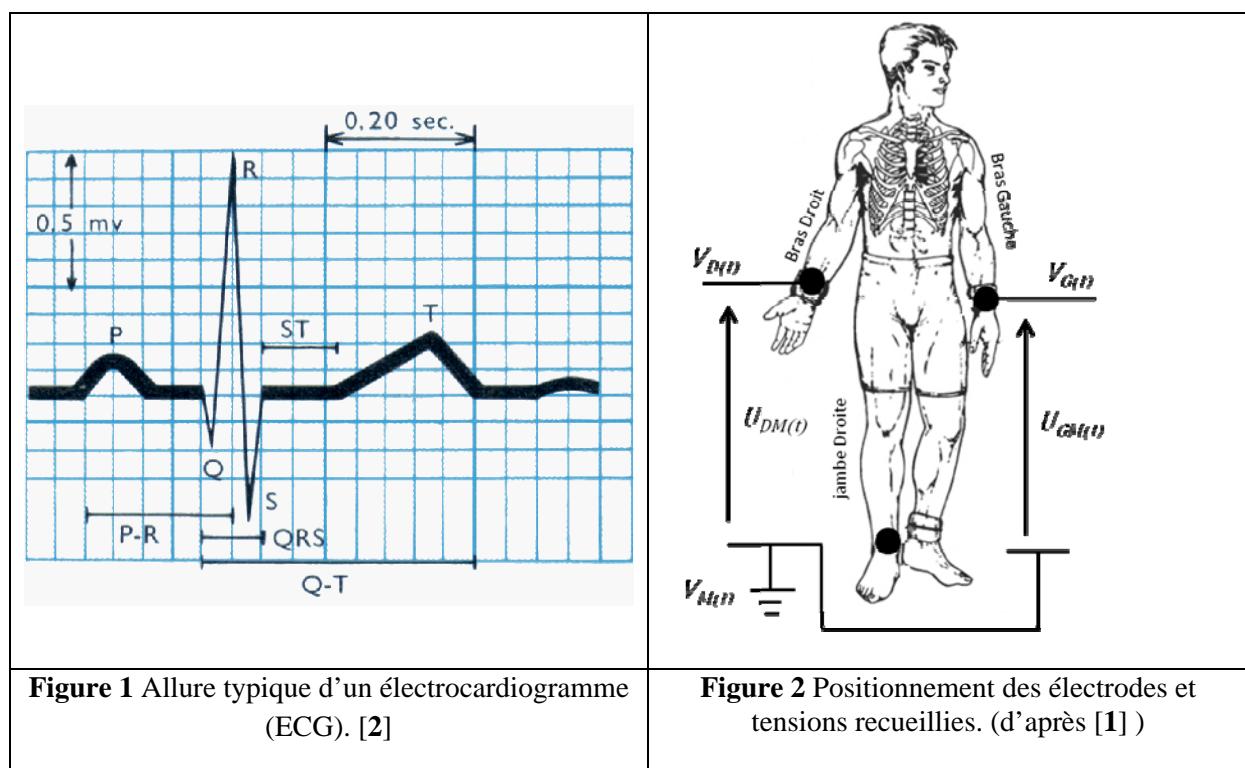


Le but de sujet est de préparer la réalisation d'une application qui doit relever et afficher l'electrocardiogramme d'un patient.

I. PRESENTATION

La propagation de l'activité électrique cardiaque provoque l'apparition de différences de potentiels à la surface du corps, qui peuvent être enregistrées en différents points de référence, c'est l'électrocardiogramme (ECG).

Ces potentiels seront prélevés grâce à des électrodes positionnées sur le corps humain. Pour assurer une prise d'information convenable il faut que la résistance de contact entre l'électrode et la peau soit la plus faible possible. Pour cela on interpose entre l'électrode métallique et la peau un matériau conducteur à l'état liquide ou à l'état de gel, à base de chlorure de potassium le plus souvent. Pour figer le positionnement, les électrodes sont solidaires d'un anneau autocollant [3]. L'allure « classique » d'un ECG (visualisation de la tension $U_{CARD(t)}$) est donnée sur la figure 1.



Dans la suite de ce sujet, il y aura systématiquement une électrode disposée sur le poignet droit, une autre sur le poignet gauche et généralement une troisième électrode fixée sur la cheville droite (sauf dans les questions spécifiées!) (cf figure 2).

Le potentiel $V_{M(t)}$ est choisi comme référence (masse) pour l'électronique de mesure ainsi que pour l'ensemble des potentiels.

La différence de potentiel entre le poignet droit et la masse est telle que :

$$U_{DM(t)} = V_{D(t)} - V_{M(t)}$$

et de même pour le poignet gauche:

$$U_{GM(t)} = V_{G(t)} - V_{M(t)}$$

La tension $U_{DG(t)}$ que l'on souhaite visualiser comporte :

- une **composante utile** que l'on souhaite visualiser $U_{CARD(t)}$, image de l'activité cardiaque . Son amplitude est de l'ordre de 0.5 mV à **5.0 mV** et son spectre fréquentiel varie selon le rythme cardiaque ; ainsi la largeur de bande utile d'un signal ECG, selon l'application, peut s'étendre de 0.5 Hz à 50 Hz (pour une application dans des services de soins intensifs) et jusqu'à 1 kHz (détection de stimulateur cardiaque) [4]. Pour ce sujet, nous choisirons une application ECG clinique standard avec une largeur de bande comprise entre **0.05 Hz et 100 Hz**.
- une **composante continue** lentement variable $U_{CONT(t)}$ comprise typiquement entre **0** et **$\pm 300 \text{ mV}$** (résultant du contact peau/électrode).
- d'autres composantes indésirables induites par exemple par le **50 Hz** du réseau et par des alimentations à découpage dont le spectre de bruits est au delà de la dizaine de kHz. On note $U_{BRUIT(t)}$ cette tension induite.

On a donc :

$$U_{DG(t)} = V_{D(t)} - V_{G(t)} = U_{CARD(t)} + U_{CONT(t)} + U_{BRUIT(t)}.$$

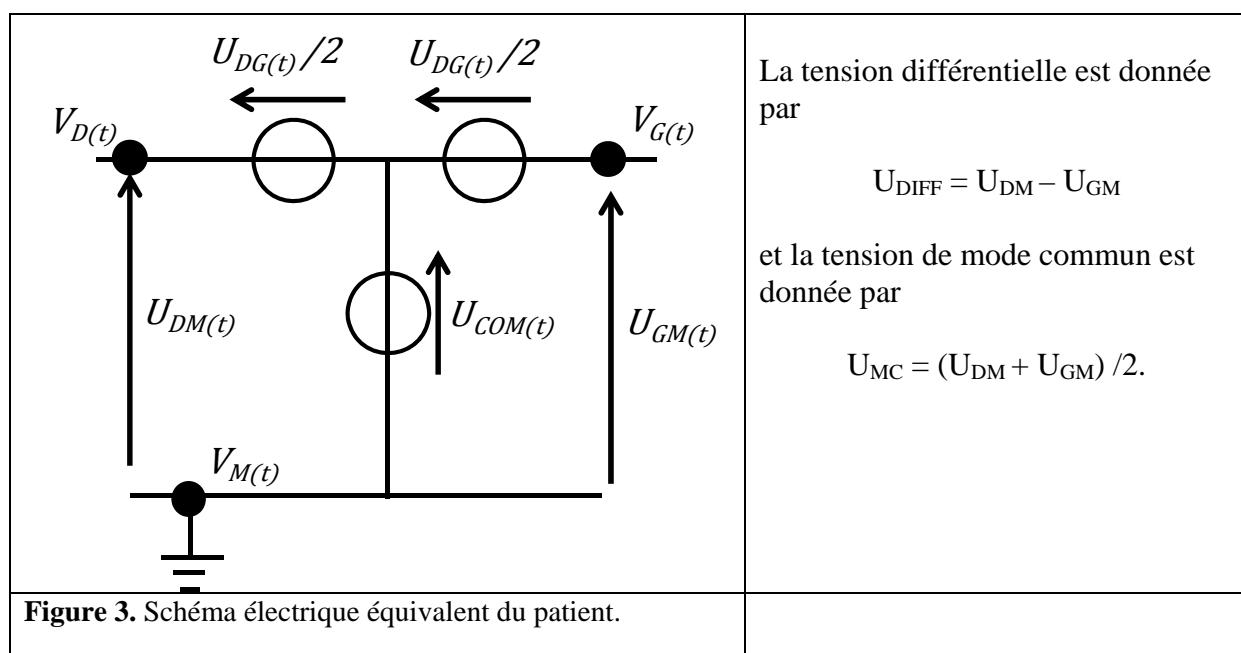
Il faut noter que les tensions $U_{DM(t)}$ et $U_{GM(t)}$ comportent en plus :

- une **composante de bruit de mode commun** $U_{COM(t)}$ (essentiellement dûe au **50Hz**)

II. POUR SE CHAUFFER ET SE METTRE DANS LE SUJET.

| **Q1** Exprimez $U_{DG(t)}$ en fonction de $U_{DM(t)}$ et $U_{GM(t)}$. (facile !)

Entre les points de contacts D, G et la masse, le comportement électrique du patient peut être modélisé par le schéma ci-dessous (figure 3).



| **Q2** Vérifier que $U_{DIFF} = U_{DG}$ et que $U_{MC} = U_{COM}$

III. UNE CHAINE D'ACQUISITION, AH OUI ...

La figure ci-dessous présente une chaîne d'acquisition typique pour relever un ECG.

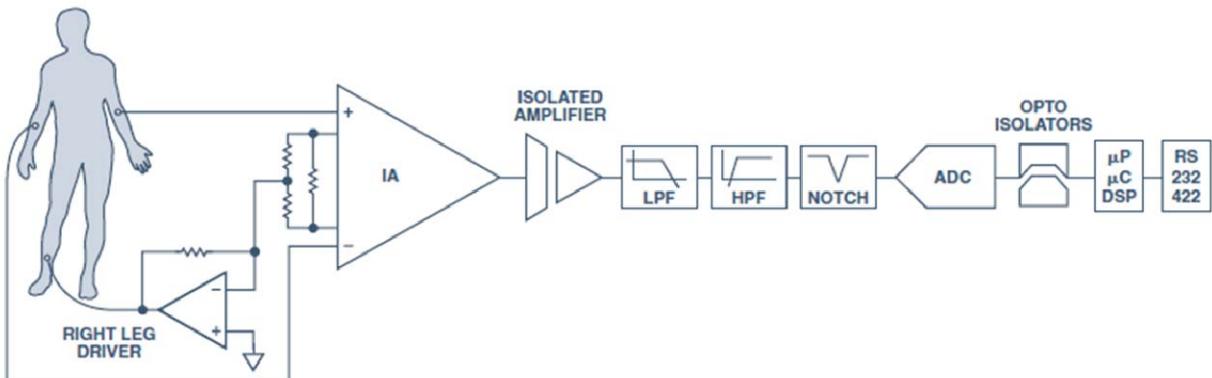
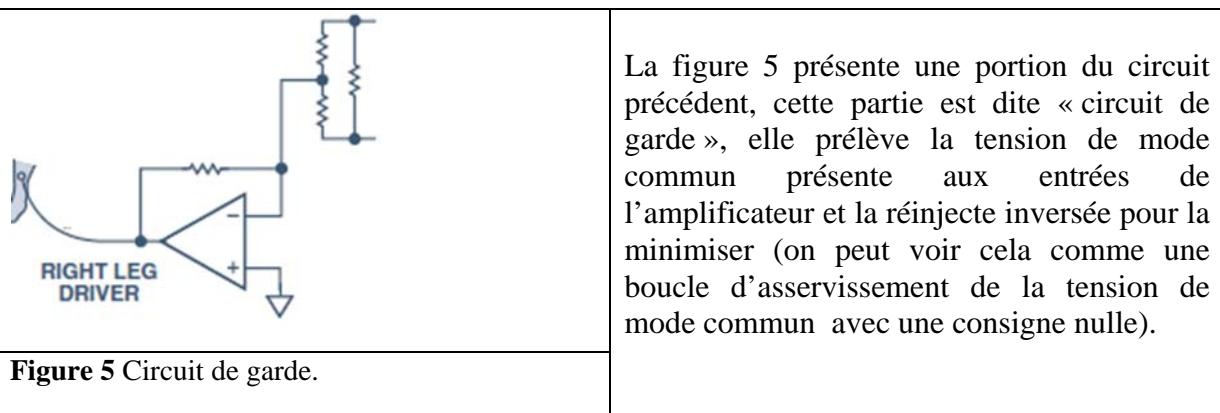


Figure 4 Chaîne typique d'acquisition pour un ECG (en sortie, l'information va dans un PC).



Q3 La figure 4 présente différents étages (9 pour être précis en excluant le circuit de garde), pour chacune de ces fonctions, précisez son nom, sa fonction (ou ses fonctions) et en quoi elle est nécessaire (si elle est nécessaire). La suite du sujet pourra vous aider à compléter ...

IV. CHEVILLE ET MODE COMMUN ...ET QUELQUES EXPERIMENTATIONS

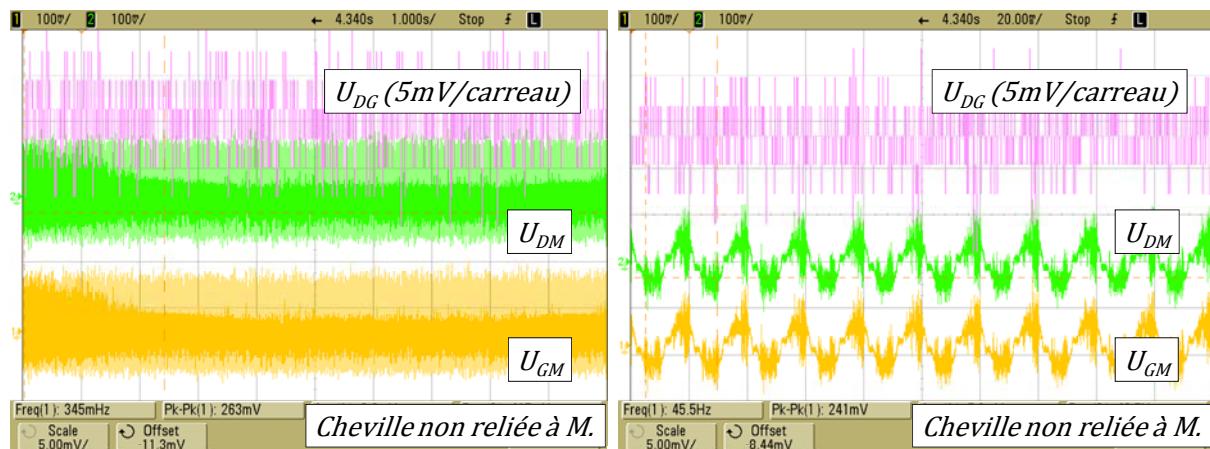


Figure 6 Mesures sans liaison entre la masse et la cheville.

Les relevés de U_{DM} et U_{GM} suivants ont été directement relevés sur un oscilloscope (masse de l'oscillo connecté à M), dans un premier cas sans référencer la cheville à la masse M (figure 6), puis en référençant la cheville à la masse M (figure 7).

L'oscilloscope calcule alors la tension U_{DG} .

Remarque : Pour la figure 7, l'oscilloscope réalise en plus un filtrage numérique (mode *High Resolution*) pour enlever le bruit haute fréquence.

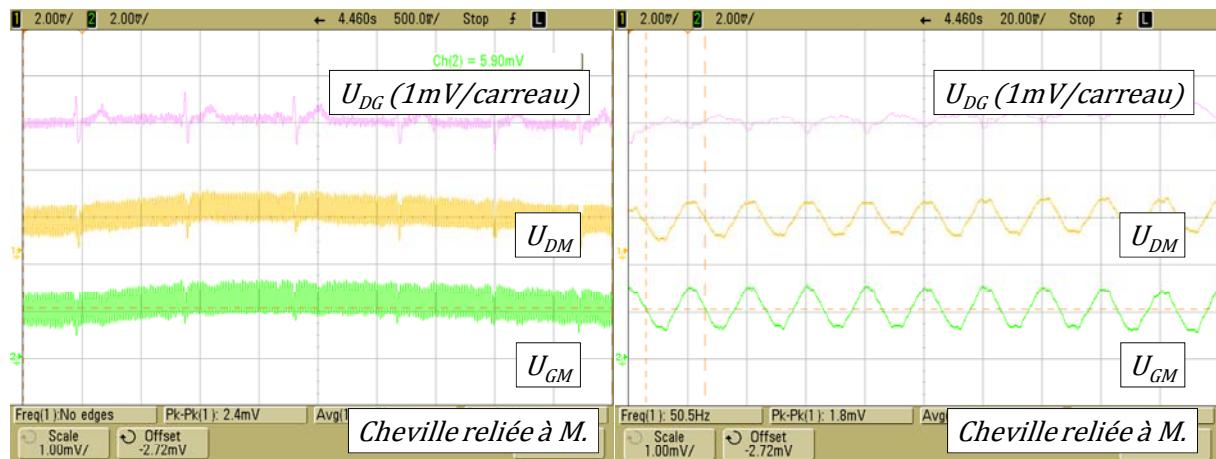


Figure 7 Mesures avec liaison entre la masse et la cheville

Q4 A partir de ces figures, que valent les tensions de mode commun (amplitude, fréquence) dans les deux cas ?

Q5 Pourquoi est-il préférable de relier D, G et la cheville du patient à M (ou d'utiliser un circuit de garde) plutôt que de ne relier que D et G ?

Q6 Figure 7, reste-t-il une composante 50Hz sur U_{DG} ? Quelle est son amplitude?

Q7 Figure 7, la composante utile est-elle visualisable ? Si oui quelle amplitude pic à pic ? Est-ce cohérent avec le début du sujet ?

Q8 Quel rythme cardiaque peut-on mesurer (en pulsation/min) ?

Dans toute la suite de l'étude, la cheville est reliée à un circuit de garde.

On supposera **pour toute la suite de l'étude**, que U_{COM} ne comporte que du **50 Hz** et on fixe : $U_{COM} = 1\text{mV}$ d'amplitude.

De même, **pour toute la suite de l'étude**, on suppose que U_{BRUIT} ne comporte que du **50Hz** avec une **amplitude de 0,2 mV** et du **20kHz** avec une **amplitude de 0,2 mV**.

Q9 A partir du paragraphe précédent, tracez le spectre d'amplitude des composantes de U_{BRUIT} (fréquence et niveau)

V. QUELS SIGNAUX ?

Q10 Pour une personne dont le cœur bat à 60 pulsations par minutes, quelle sera la période de $U_{CARD(t)}$?

Q11 En déduire la fréquence du fondamental du signal $U_{CARD(t)}$.

Le spectre du signal $U_{CARD(t)}$ est donné sur la figure 8 (pour 60 pulsation /minute).

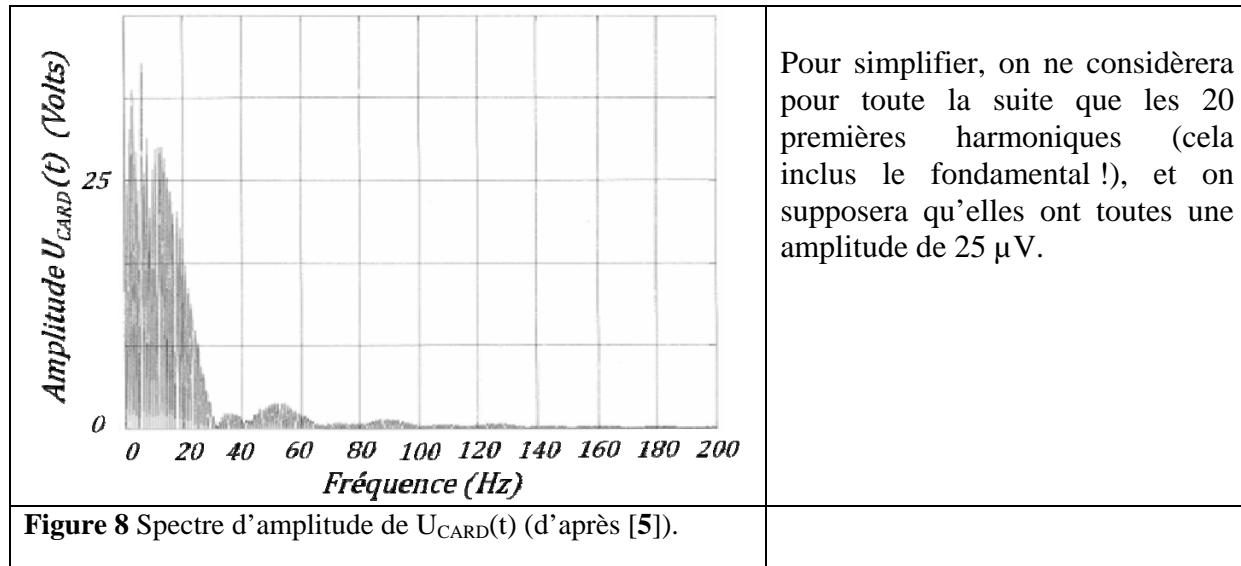


Figure 8 Spectre d'amplitude de $U_{CARD}(t)$ (d'après [5]).

Pour simplifier, on ne considèrera pour toute la suite que les 20 premières harmoniques (cela inclus le fondamental !), et on supposera qu'elles ont toutes une amplitude de $25 \mu\text{V}$.

Q12 Quelle serait alors le niveau maximal de $U_{CARD(t)}$ possible ?

Q13 Sur le même graphe, donner le spectre d'amplitude de toutes les composantes de \mathbf{U}_{PG} pour un cœur qui bat à 60 pulsation/minutes En déduire ce qu'il va falloir amplifier et ce qu'il va falloir atténuer par la suite.

Q14 donner le spectre d'amplitude de toutes les composantes de \mathbf{U}_{DG} pour un cœur qui bat à 180 pulsation/minutes ... cela pose-t-il un problème pour traiter le signal ?

VI. L'AMPLI D'INSTRUMENTATION

Pour la suite, nous utilisons le circuit présenté sur la figure suivante :

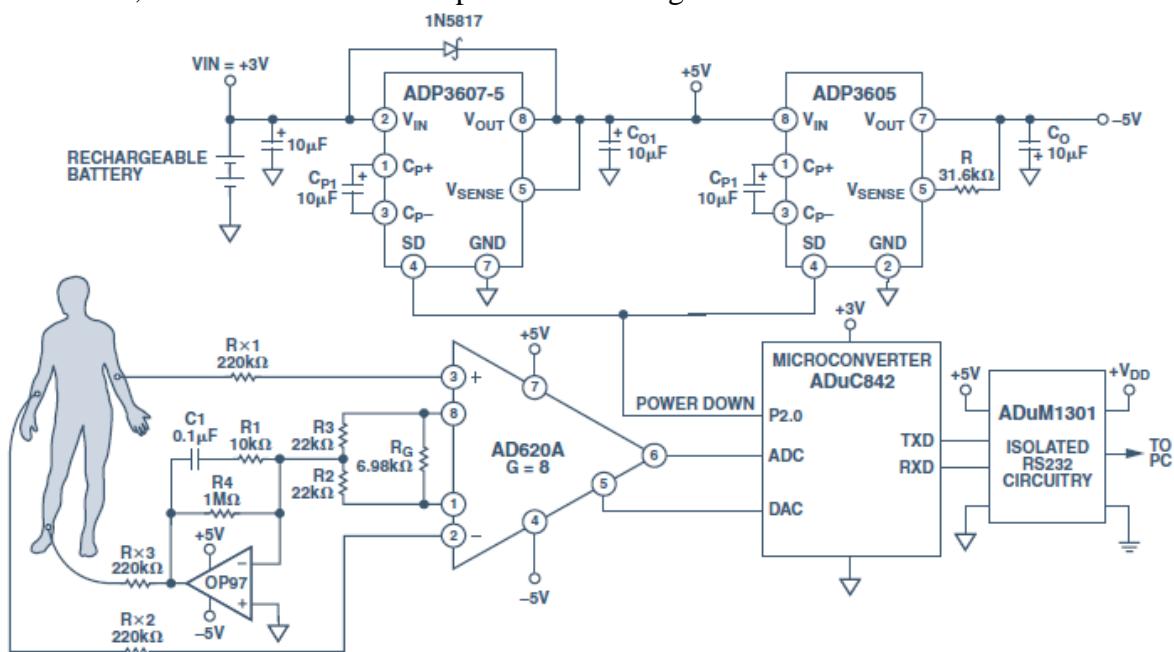


Figure 9. Proposed ECG configuration.

Figure 9 Structure utilisée pour la chaîne d'instrumentation [4].

L'amplificateur choisi est un **AD620A**, il est alimenté en $\pm 5 \text{ V}$

Q15 Quelle est la plage de tension admissible pour les tensions en entrée de l'ampli ?

Q16 Quelle est le niveau max du signal V_{DM} (ou V_{GM}) en entrée de l'ampli? Est-ce compatible avec l'ampli ?

Q17 Quelle est le niveau max du signal de sortie (output) de l'ampli?

L'ADC qui suivra l'ampli a une entrée entre 0 et 5V(12 bits). Pour l'ampli, nous avons la relation :

$$\text{OUTPUT(pin6)} - \text{REF(pin5)} = G \cdot (\text{IN}^+(pin3) - \text{IN}^-(pin2))$$

Q18 Quel est le gain maximal possible (tenir compte aussi de Q17)?

La sortie DAC du μ C est reliée à REF de l'AD620 ce qui permet au programme de décaler le niveau d'OUTPUT pour que la tension d'entrée de l'ADC reste le plus possible au milieu de gamme (mais cela est sans importance pour le calcul précédent).

Q19 De combien peut-on ainsi décaler le potentiel d'OUTPUT ? (cf doc AD 620A)

Pour la suite, on prendra un gain **G = 8.**

La doc de l'AD620 donne $R_G = 49.4k\Omega / (G - 1)$

Q20 Quelle est la valeur de R_G correspondante?

Q21 Quelle est le taux de réjection du mode commun?

Q22 Quelle est alors la valeur maximale de la tension issue du mode commun restante en sortie de l'ampli AD620 ?

Q23 Quelle est la valeur maximale de la composante utile en sortie de l'ampli (pour $U_{CARD} = 5mV$)? Conclure sur le mode commun.

VII. LA SECURITE

Q24 Compte tenu de l'application, quelle peut bien être le rôle des résistances RX1, RX2 et RX3 de $220k\Omega$?

Q25 L'alimentation principale est fournie par une pile rechargeable de 3.3 V, pourquoi une pile et pas une alimentation classique?

Q26 Le $\pm 5V$ est fournie par deux composants, lesquels ?

Q27 Avant d'être transmise à un PC sur le port série, le relevé des mesures passe dans un driver RS232 ADuM1301 . Pourquoi est-il « isolated »?

VIII. LE SOFT

Q28 Voyez-vous un filtre anti-repliement sur la figure 9 ? (oui/non)

Q29 La fréquence d'échantillonnage est fixée à 500 Hz. Qu'en pensez-vous?

Q30 Quels sont les fonctions que doit réaliser le programme dans le μ C à partir des acquisitions? Expliquer à partir de la doc fournie.

Q31 Quels types de filtres doit-on utiliser pour garder la forme de U_{CARD} ?

IX. LES ERREURS

Q32 (courte mais assez longue ...) : Analysez et quantifiez les sources d'erreurs du montage.

Fin.

Références :

[1] <http://www.open-ecg-project.org/tiki-index.php?page=Refik's%20ECG>

[2] ALPHA, tome 6, page 2157, éd. Grange Batelière.

[3] http://michel.hubin.pagesperso-orange.fr/capteurs/biomed/chap_b10.htm#debut

[4] *ECG Front-End Design is Simplified with MicroConverter*, Enrique Company-Bosch, Eckart Hartmann, Analog Dialogue 37-11, November (2003).

[5] épreuve U4.2 Physique appliquée -BTS systèmes électroniques 2009

FEATURES

Easy to use

Gain set with one external resistor

(Gain range 1 to 10,000)

Wide power supply range (± 2.3 V to ± 18 V)

Higher performance than 3 op amp IA designs

Available in 8-lead DIP and SOIC packaging

Low power, 1.3 mA max supply current

Excellent dc performance (B grade)

50 μ V max, input offset voltage

0.6 μ V/ $^{\circ}$ C max, input offset drift

1.0 nA max, input bias current

100 dB min common-mode rejection ratio (G = 10)

Low noise

9 nV/ $\sqrt{\text{Hz}}$ @ 1 kHz, input voltage noise

0.28 μ V p-p noise (0.1 Hz to 10 Hz)

Excellent ac specifications

120 kHz bandwidth (G = 100)

15 μ s settling time to 0.01%

APPLICATIONS

Weigh scales

ECG and medical instrumentation

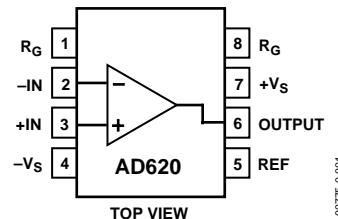
Transducer interface

Data acquisition systems

Industrial process controls

Battery-powered and portable equipment

CONNECTION DIAGRAM



00775-0-001

Figure 1. 8-Lead PDIP (N), CERDIP (Q), and SOIC (R) Packages

PRODUCT DESCRIPTION

The AD620 is a low cost, high accuracy instrumentation amplifier that requires only one external resistor to set gains of 1 to 10,000. Furthermore, the AD620 features 8-lead SOIC and DIP packaging that is smaller than discrete designs and offers lower power (only 1.3 mA max supply current), making it a good fit for battery-powered, portable (or remote) applications.

The AD620, with its high accuracy of 40 ppm maximum nonlinearity, low offset voltage of 50 μ V max, and offset drift of 0.6 μ V/ $^{\circ}$ C max, is ideal for use in precision data acquisition systems, such as weigh scales and transducer interfaces.

Furthermore, the low noise, low input bias current, and low power of the AD620 make it well suited for medical applications, such as ECG and noninvasive blood pressure monitors.

The low input bias current of 1.0 nA max is made possible with the use of Super β eta processing in the input stage. The AD620 works well as a preamplifier due to its low input voltage noise of 9 nV/ $\sqrt{\text{Hz}}$ at 1 kHz, 0.28 μ V p-p in the 0.1 Hz to 10 Hz band, and 0.1 pA/ $\sqrt{\text{Hz}}$ input current noise. Also, the AD620 is well suited for multiplexed applications with its settling time of 15 μ s to 0.01%, and its cost is low enough to enable designs with one in-amp per channel.

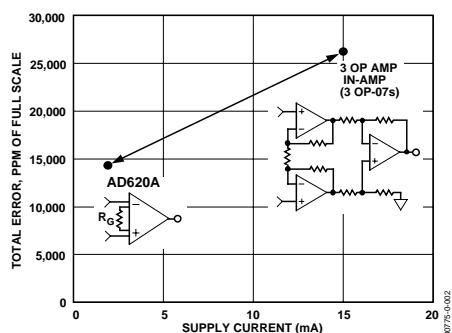


Figure 2. Three Op Amp IA Designs vs. AD620

Rev. G

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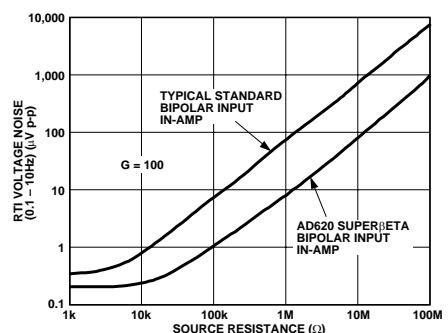


Figure 3. Total Voltage Noise vs. Source Resistance

SPECIFICATIONS

Typical @ 25°C, $V_S = \pm 15$ V, and $R_L = 2$ kΩ, unless otherwise noted.

Table 1.

Parameter	Conditions	AD620A			AD620B			AD620S ¹			Unit
		Min	Typ	Max	Min	Typ	Max	Min	Typ	Max	
GAIN											
Gain Range	$G = 1 + (49.4 \text{ k}\Omega/R_G)$										
Gain Error ²	$V_{\text{OUT}} = \pm 10$ V	1		10,000	1		10,000	1		10,000	
$G = 1$			0.03	0.10		0.01	0.02		0.03	0.10	%
$G = 10$			0.15	0.30		0.10	0.15		0.15	0.30	%
$G = 100$			0.15	0.30		0.10	0.15		0.15	0.30	%
$G = 1000$			0.40	0.70		0.35	0.50		0.40	0.70	%
Nonlinearity	$V_{\text{OUT}} = -10$ V to +10 V										
$G = 1\text{--}1000$	$R_L = 10$ kΩ	10	40		10	40		10	40		ppm
$G = 1\text{--}100$	$R_L = 2$ kΩ	10	95		10	95		10	95		ppm
Gain vs. Temperature											
$G = 1$			10			10			10		ppm/°C
$\text{Gain} > 1^2$			-50			-50			-50		ppm/°C
VOLTAGE OFFSET	(Total RTI Error = $V_{\text{OSI}} + V_{\text{OSO}/G}$)										
Input Offset, V_{OSI}	$V_S = \pm 5$ V to ± 15 V	30	125		15	50		30	125		μV
Overtemperature	$V_S = \pm 5$ V to ± 15 V		185			85			225		μV
Average TC	$V_S = \pm 5$ V to ± 15 V	0.3	1.0		0.1	0.6		0.3	1.0		μV/°C
Output Offset, V_{OSO}	$V_S = \pm 15$ V	400	1000		200	500		400	1000		μV
	$V_S = \pm 5$ V		1500			750			1500		μV
Overtemperature	$V_S = \pm 5$ V to ± 15 V		2000			1000			2000		μV
Average TC	$V_S = \pm 5$ V to ± 15 V	5.0	15		2.5	7.0		5.0	15		μV/°C
Offset Referred to the Input vs. Supply (PSR)											
$G = 1$	$V_S = \pm 2.3$ V to ± 18 V	80	100		80	100		80	100		dB
$G = 10$		95	120		100	120		95	120		dB
$G = 100$		110	140		120	140		110	140		dB
$G = 1000$		110	140		120	140		110	140		dB
INPUT CURRENT											
Input Bias Current		0.5	2.0		0.5	1.0		0.5	2		nA
Overtemperature			2.5			1.5			4		nA
Average TC		3.0			3.0			8.0			pA/°C
Input Offset Current		0.3	1.0		0.3	0.5		0.3	1.0		nA
Overtemperature			1.5			0.75			2.0		nA
Average TC		1.5			1.5			8.0			pA/°C
INPUT											
Input Impedance											
Differential Common-Mode		10 2			10 2			10 2			GΩ_pF
Input Voltage Range ³	$V_S = \pm 2.3$ V to ± 5 V	$-V_S + 1.9$	+ $V_S - 1.2$	$-V_S + 1.9$	+ $V_S - 1.2$	$-V_S + 1.9$	+ $V_S - 1.2$	$-V_S + 1.9$	+ $V_S - 1.2$		GΩ_pF
Overtemperature	$V_S = \pm 5$ V to ± 18 V	- $V_S + 2.1$	+ $V_S - 1.3$	- $V_S + 2.1$	+ $V_S - 1.3$	- $V_S + 2.1$	+ $V_S - 1.3$	- $V_S + 2.1$	+ $V_S - 1.3$		V
Overtemperature		- $V_S + 1.9$	+ $V_S - 1.4$	- $V_S + 1.9$	+ $V_S - 1.4$	- $V_S + 1.9$	+ $V_S - 1.4$	- $V_S + 1.9$	+ $V_S - 1.4$		V
Overtemperature		- $V_S + 2.1$	+ $V_S - 1.4$	- $V_S + 2.1$	+ $V_S + 2.1$	- $V_S + 2.3$	+ $V_S - 1.4$	- $V_S + 2.3$	+ $V_S - 1.4$		V

AD620

Parameter	Conditions	AD620A			AD620B			AD620S ¹			Unit
		Min	Typ	Max	Min	Typ	Max	Min	Typ	Max	
Common-Mode Rejection Ratio DC to 60 Hz with 1 kΩ Source Imbalance	$V_{CM} = 0 \text{ V}$ to $\pm 10 \text{ V}$										
G = 1		73	90		80	90		73	90		dB
G = 10		93	110		100	110		93	110		dB
G = 100		110	130		120	130		110	130		dB
G = 1000		110	130		120	130		110	130		dB
OUTPUT											
Output Swing	$R_L = 10 \text{ k}\Omega$ $V_S = \pm 2.3 \text{ V}$ to $\pm 5 \text{ V}$										
Overtemperature	$V_S = \pm 5 \text{ V}$ to $\pm 18 \text{ V}$										
Overtemperature Short Circuit Current											
DYNAMIC RESPONSE											
Small Signal –3 dB Bandwidth											
G = 1			1000			1000			1000		kHz
G = 10			800			800			800		kHz
G = 100			120			120			120		kHz
G = 1000			12			12			12		kHz
Slew Rate		0.75	1.2		0.75	1.2		0.75	1.2		V/μs
Settling Time to 0.01%	10 V Step										
G = 1–100			15			15			15		μs
G = 1000			150			150			150		μs
NOISE											
Voltage Noise, 1 kHz		$\text{Total RTI Noise} = \sqrt{(e_{ni}^2) + (e_{no}/G)^2}$									
Input, Voltage Noise, e_{ni}		9	13		9	13		9	13		nV/√Hz
Output, Voltage Noise, e_{no}		72	100		72	100		72	100		nV/√Hz
RTI, 0.1 Hz to 10 Hz											
G = 1			3.0			3.0	6.0		3.0	6.0	μV p-p
G = 10			0.55			0.55	0.8		0.55	0.8	μV p-p
G = 100–1000			0.28			0.28	0.4		0.28	0.4	μV p-p
Current Noise	f = 1 kHz		100			100			100		fA/√Hz
0.1 Hz to 10 Hz			10			10			10		pA p-p
REFERENCE INPUT											
R_{IN}			20			20			20		kΩ
I_{IN}	$V_{IN+}, V_{REF} = 0$		50	60		50	60		50	60	μA
Voltage Range Gain to Output			$-V_S + 1.6$	$+V_S - 1.6$		$-V_S + 1.6$	$+V_S - 1.6$		$-V_S + 1.6$	$+V_S - 1.6$	V
1 ± 0.0001			1 ± 0.0001			1 ± 0.0001			1 ± 0.0001		
POWER SUPPLY											
Operating Range ⁴		± 2.3	± 18		± 2.3	± 18		± 2.3	± 18		V
Quiescent Current	$V_S = \pm 2.3 \text{ V}$ to $\pm 18 \text{ V}$		0.9	1.3		0.9	1.3		0.9	1.3	mA
Overtemperature			1.1	1.6		1.1	1.6		1.1	1.6	mA
TEMPERATURE RANGE											
For Specified Performance		–40 to +85			–40 to +85			–55 to +125			°C

¹ See Analog Devices military data sheet for 883B tested specifications.

² Does not include effects of external resistor R_G .

³ One input grounded. G = 1.

⁴ This is defined as the same supply range that is used to specify PSR.

Patient Safety

In addition to the digital isolation and the safe power supply, the series resistors, Rx1, Rx2, and Rx3, provide protection for the patient—in order to comply with AAMI (Association for the Advancement of Medical Instrumentation) standards for safe current levels (see References). These standards require that rms ground currents or fault current from the electronics must be less than 50 μ A.

Signal Processing

The ADuC842 MicroConverter is well suited for the main signal processing tasks. It features a fast, 12-bit ADC and other high-performance analog peripherals, a fast 8052 microprocessor core, integrated 62KB flash memory for code, and several other useful peripherals, as shown in Figure 4.

The key components of the MicroConverter for this design are the ADC and the 8052 core. The ADC converts the analog output of the instrumentation amplifier to a digital signal. The software written for the 8052 core processes the digitized signal to produce the data for the ultimate ECG trace. As in many MicroConverter designs, the software includes both complex high level code written in C and time sensitive routines written in *assembly code*. In this case, the implementation of band-pass filters and notch filters is in C, while the ADC is controlled by assembly code. Assembly code, combined with converter speed, enables the accumulation of multiple samples, enhancing the effective resolution of the ADC well beyond its normal 12 bits.

Figure 5 gives a good indication of the effectiveness of the MicroConverter. The top trace is the signal from the instrumentation amplifier applied to the ADC. The middle trace shows the initial results achieved using the C-code filtering only, while the bottom trace shows the final result after the processing of multiple conversions, using assembly code.

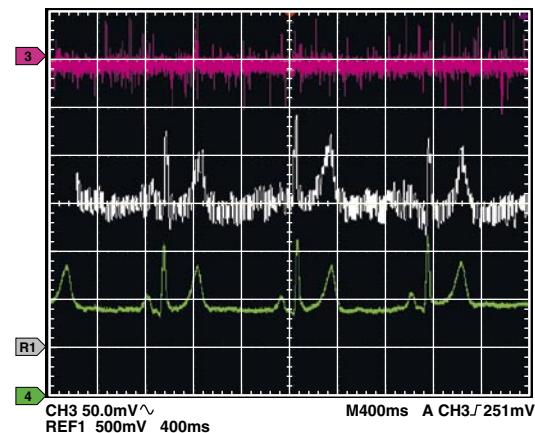


Figure 5. Oscilloscope traces.

Filters in C Code

The acquired signal is processed by digital filtering in the MicroConverter. For this purpose, we designed two second-order digital *infinite impulse-response* (IIR) filters, based on a sampling frequency of 500 Hz. A notch filter was designed to suppress the 50-Hz interference. The chosen design procedure was the *pole-zero placement method*, with a notch frequency of 50 Hz and notch width 10 Hz. To achieve this required the following transfer function:

$$H(z) = \frac{1 - 1.618z^{-1} + z^{-2}}{1 - 1.5164z^{-1} + 0.8783z^{-2}}$$

The transfer function can be converted into a programmable recursive algorithm:

$$NOut_k = NIn_k - 1.618NIn_{k-1} + NIn_{k-2} + 1.5164NOut_{k-1} - 0.8783NOut_{k-2}$$

In this equation the subindex, k , means the present value, $k-1$ means the value in the previous instant, and so on.

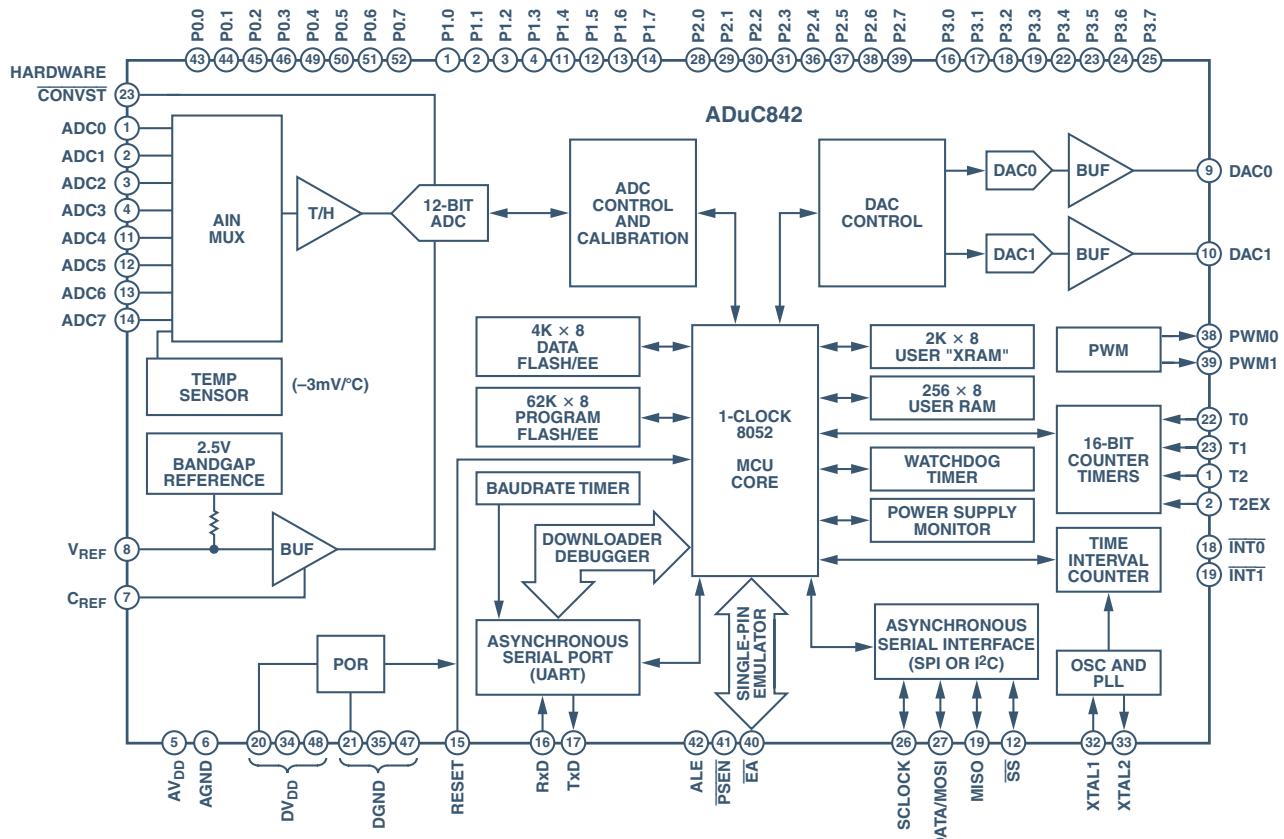


Figure 4. ADuC842 block diagram.

We now need to turn this equation into code. C coding was the automatic choice for this arithmetic-intensive processing, as programming it in assembly would have been too time consuming. Implementing the filter equations directly would be inefficient with the ADuC842, since it is not tailored for floating-point calculations. Fortunately we can just scale the coefficients (e.g. by 4096) and implement the notch code as:

```
iNOut = (4096L*iNIn-6627L*iNIn1+4096L*iNIn2+6211L*iNOut1-3598L*iNOut2)/4096;
```

This implements a second-order filter. Although we can calculate higher order filters, in practice it seems workable to simply cascade second-order filters.

The second filter was a Butterworth pass-band filter with a 0.05-Hz low cutoff frequency and a 100-Hz high cutoff frequency. The transfer function and recursive algorithm are:

$$H(z) = \frac{0.4206 - 0.4206z^{-2}}{1 - 1.1582z^{-1} + 0.1582z^{-2}}$$

```
BOut_k = 0.4206BIn_k - 0.4206BIn_{k-2} + 1.1582BOut_{k-1} - 0.1582BOut_{k-2}
```

This is implemented in C code by:

```
iBOut = (1723L*iBIn-1723L*iBIn2+4745L*iBOut1-650L*iBOut2) /4096;
```

Note that the outputs can be scaled simply by changing the coefficients of the inputs. Also note that, for efficiency (if the signals are all positive), the division by 4096 is accomplished at the end by shifting 12 right.

The implementation shown in Figure 6 is for a cascade of five band-pass filters and two notch filters. The signal is scaled up by a factor of 4 in each of the first and second band-pass filters. The 12-bit right shift accomplishes the divide-by-4096.

```
while(1)
{
    while(c25ms<64); //Wait for 64 measurements to be done.
    iBIn = iAdc0>>3; //Save accumulated measurement.
    iAdc0 = 0; //Zero for new measurement accumulation.
    c25ms = 0; //Reset synchronization timer.
    // 5 Band pass 0.05 - 100Hz Fs=500 first 2 with gain of 4 each.
    iBOut = (6891L*iBIn-6891L*iBIn2+4745L*iBOut1-650L*iBOut2)>>12L;
    iB010 = (6891L*iBOut-6891L*iBOut2+4745L*iB011-650L*iB012)>>12L;
    iB020 = (1723L*iB010-1723L*iB012+4745L*iB021-650L*iB022)>>12L;
    iB030 = (1723L*iB020-1723L*iB022+4745L*iB031-650L*iB032)>>12L;
    iNIn = (1723L*iB030-1723L*iB032+4745L*iNIn1-650L*iNIn2)>>12L;
    // 2 notch filters.
    iNOut = (4096L*iNIn-6627L*iNIn1+4096L*iNIn2+6211L*iNOut1-3598L*iNOut2)>>12L;
    iN30 = (4096L*iNOut-6627L*iNOut1+4096L*iNOut2+6211L*iN301-3598L*iN302)>>12L;

    iBIn2 = iBIn1; //Save delayed values for filters.
    iBIn1 = iBIn;
    iBOut2 = iBOut1;
    iBOut1 = iBOut;

    iNIn2 = iNIn1;
    iNIn1 = iNIn;
    iNOut2 = iNOut1;
    iNOut1 = iNOut;

    iB012 = iB011;
    iB011 = iB010;
    iB022 = iB021;
    iB021 = iB020;
    ...
    //Other delayed values not shown.

    if(iBIn>24000) iDac -= 1; //Control AD620 output level.
    if(iBIn<8000) iDac += 1;
    iout1 = (iDac)&0xffff;
    DACOH = (iout1>>8)&0xf;
    DACOL = iout1&0xff;

    if((iN30+iOfs)>3000) iOfs -= 1; //Control output level.
    if((iN30+iOfs)<1000) iOfs += 1;
    iout = ((iN30+iOfs))&0xffff;
    DAC1H = (iout>>8)&0xf; //Output to oscilloscope for evaluation.
    DAC1L = iout&0xff;
    if(!(c2++&3)) printf("%4d\r\n",iOut); //Output to PC.
}
```

Figure 6. Essential part of C code.

Note the lines, `if(iAdc0>24000)iDac -= 1,` and `if(iAdc0<8000)iDac += 1,` which adjust the DAC output of the ADuC842 to drive the level-shifting input of the AD620 to shift the AD620 output to a comfortable value for the MicroConverter's ADC input. This is desirable to reduce the effects of the variable dc offsets that result from slight differences in the way the electrodes are applied to the skin. A similar technique is used to ensure that the output voltage is centered within the output range.

Processing in Assembly Code

The assembly code's main functions are to measure the input signal at regular intervals and to ensure that the C code calculations are repeated at the required rate of 500 times per second. In the first instance, we programmed Timer0 to run continuously and generate its interrupts at 1-ms intervals. Each interrupt restarts Timer0, gets an ADC conversion result, and increments a variable, `c2ms`, which is used to synchronize the C code. At this stage of code development, the first few lines of C code were:

```
while(c2ms<2); //Used in first phase.
c2ms = 0;
iAdc0 = iAdc0;
```

Initially, `c2ms` is 0, and the C code will wait at the line `while(c2ms<2);`. After 1 ms, a Timer0 interrupt occurs, and `c2ms` is incremented to 1. After a further 1 ms, `c2ms` is incremented to 2. Now `while(c2ms<2);` is no longer satisfied, and the C code continues by resetting counter `c2ms` to 0 and doing the filter calculations. Thereafter, the C code shifts the results down the chain of variables representing the various delayed results ready for the next iteration of the loop. The final part of the loop is the `printf(...)`, which sends the result to the PC for display. The processing of the data on the PC, beyond the scope of this article, can be as simple as importing it to a spreadsheet for graphical display—or as sophisticated as the designer wishes to make it. This solution produced the middle trace of Figure 5.

To improve the result, the Timer0 interrupt rate was shortened to 1/32 ms, and the data was accumulated in `iAdc0`, to make use of multiple measurements instead of just a single measurement. At the same time, the `while` was changed to `while(c2ms<64)` so that the C code would wait for 64 measurements to be accumulated before doing each filter loop. The value in `iAdc0` is saved in `iAdc00` for further processing, and then `iAdc0` is cleared—ready to accumulate the next 64 measurements. Figure 7 shows the assembly code. This improved solution produced the lower trace of Figure 5.

```
IntT0: push ACC
       push PSW
       clr TR0
       mov TH0,#0fdh ;Stop T0.
       mov TL0,#0f6h ;Reload 1/32ms.
       setb TR0 ;Restart T0.
       mov a,ADCDATAL
       add a,iAdc0+3
       mov iAdc0+3,a
       mov a,ADCDATAH ;Add a conversion to iAdc0.
       anl a,#0fh
       addc a,iAdc0+2
       mov iAdc0+2,a
       clr a
       addc a,iAdc0+1
       mov iAdc0+1,a
       mov ADCCON2,#0 ;Start a conversion.
       mov ADCCON2,#10h
       inc c25ms ;Increment ms counter.
       IntT0R: pop ACC
               pop PSW
               reti
```

Figure 7. Assembly code.

Gain

Signal gain is always an important consideration in an ECG signal chain. In the above-described design, it depends on a number of factors. The analog gain is set to 8 \times , as discussed previously. Next, a gain of 64 \times results from accumulating 64 measurements of this signal. Next there is a signal loss of 8 \times from the code `iBIn = iAdc0>>3;`, and finally, a gain of 4 \times twice from the scaling of the first two band-pass-filter equations. This results in a total gain of $G = (8 \times 64/8) \times 4 \times 4 = 1024$, which is typical of analog ECG circuits.

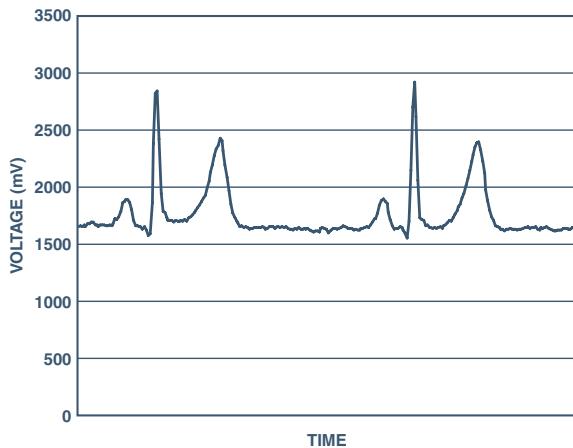


Figure 8. Graphs of practical measurements.

CONCLUSION

Figure 8 shows results for a subject connected in Einthoven lead I configuration. As can be seen, good results are achieved despite the simplicity of the electronic hardware used. The article demonstrates that significant improvements can be achieved

with simple hardware combined with attention to software. The improvement in this example is by no means at the optimum level; it should be possible for a dedicated designer to significantly improve the results. Additional improvements could be made if code with different filter frequencies or other special characteristics were to be implemented. The code memory of the ADuC842 is flash based, allowing such customizations to be made after a product using it is manufactured—or even as the patients' needs change. An ultimate result could be a compact, inexpensive ECG for a potentially large-volume market. □

REFERENCES

Webster John G., *Medical Instrumentation. Application and Design.* 3rd edition, Wiley, 1998.

Firth J. and Errico P., "Low-Power, Low-Voltage IC Choices for ECG System Requirements," *Analog Dialogue*, Volume 29, Number 3, 1995.

AAMI, *American National Standard, Safe Current Limits for Electromedical Apparatus (ANSI/AAMI ES1-1993)*. Association for the Advancement of Medical Instrumentation, 1993.

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