DIPLOMA THESIS

DEVELOPMENT OF AN MRI-COMPATIBLE ECG SYSTEM

CSABA HAJDU

SUPERVISORS

Dr Dávid Légrády   Béla Mihalik
BME   Mediso

MŰEGYETEM 1782

Budapest University of Technology and Economics
Faculty of Natural Sciences
Institute of Nuclear Techniques
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ÖNÁLLÓSÁGI NYILATKOZAT (DECLARATION OF INDEPENDENT WORK)

Kijelentem, hogy jelen szakdolgozat saját munkám eredménye, azt önállóan, nem megengedett segédeszköz igénybevétele nélkül készítettem el. A szakdolgozatban csak a megadott forrásokat használtam fel. Minden olyan részt, amelyet szó szerint, vagy azonos értelemben, de átfogalmazva más forrásból vettem, egyértelműen, a forrás megadásával jelöltem.

Budapest, 2012. május

Hajdu Csaba

ABSTRACT

Cardiac gating is an important technique used in many different modalities of medical imaging to eliminate artifacts resulting from the motion of the heart. It is often implemented with the use of an electrocardiograph (ECG).

This thesis project was executed at Mediso Medical Imaging Systems. The main objective was the development of an ECG system aimed at cardiac gating, to be used in the pre-clinical (small animal) and clinical (human) imaging devices of the company.

The ECG system is based on a microcontroller and an ECG-specific analog frontend. Prototype hardware was created using development kits and the appropriate firmware was developed. Since the tests carried out with the prototype system confirmed the suitability of the concept, the final hardware was designed. The process is explained in detail.

Cardiac gating also finds use in Magnetic Resonance Imaging (MRI), but this causes challenges in ECG acquisition. Measurements were carried out to assess the usability of the prototype system in MRI. These results are also reported.

KIVONAT

A szívkapuzás fontos technika, amelyet számos különböző orvosi képalkotó modalitás esetében használnak a szív mozgásából eredő műtermékek kiküszöbölésére. A szívkapuzást gyakran elektrokardiográf (EKG) használatával valósítják meg.

Ez a diplomamunka a Mediso Orvosi Berendezés Fejlesztő és Szerviz Kft.-nél készült. A fő célja egy szívkapuzásra szolgáló rendszer kifejlesztése volt, a vállalat preklinikai (kisállat) és klinikai (humán) képalkotó készülékeivel való használatra.

Az EKG rendszer egy mikrokontrolleren és egy EKG-specifikus analóg frontend áramkörön alapul. Fejlesztőkészletek felhasználásával prototípus hardvert alkottam, és kifejlesztettem az azon futó szoftvert. A prototípus rendszerrel lefolytatott tesztek alátámasztották, hogy az elképzelés megfelel a célnak, így megterveztük a végleges hardvert. Ezt a folyamatot a dolgozat részletesen leírja.

Szívkapuzást mágneses rezonancia képalkotó berendezésben (MRI-ben) is alkalmaznak, de ebben az esetben kihívást jelent az EKG mérése. A prototípus rendszer MRI-ben való használhatóságának vizsgálatára méréseket hajtottam végre, amelyek leírása szintén szerepel a dolgozatban.

ACKNOWLEDGMENTS

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ACRONYMS

AD   Analog-to-Digital

ADC   Analog-to-Digital Converter

AFE   Analog Frontend

CT   Computed Tomography

CAN   Controller Area Network

CMRR   Common-Mode Rejection Ratio

DSP   Digital Signal Processor

ECG   electrocardiograph

EVM   Evaluation Module

FIR   Finite Impulse Response

GPIO   General Purpose Input/Output

IC   Integrated Circuit

IIR   Infinite Impulse Response

LCD   Liquid Crystal Display

MCU   Microcontroller Unit

MDK   Medical Development Kit

MRI   Magnetic Resonance Imaging

PDK   Performance Demonstration Kit

PET   Positron Emission Tomography

PGA   Programmable Gain Amplifier

PSRR   Power Supply Rejection Ratio

RLD   Right Leg Drive

SAR   Successive Approximation Register

SPI   Serial Peripheral Interface

TI   Texas Instruments

UART   Universal Asynchronous Receiver/Transmitter

USB   Universal Serial Bus

WCT   Wilson Central Terminal

INTRODUCTION

This thesis describes the development of an electrocardiograph (ECG) system designed for cardiac gating. This chapter is aimed at highlighting the significance of cardiac gating in pre-clinical (animal) and clinical (human) medical imaging.

Section 1.1 discusses electrocardiography based on [1]. Assuming a basic knowledge of the anatomy of the heart, the physiological background of human ECG is explored, without delving into the clinical significance of ECG traces. Next, the practical aspects of ECG recording relevant to this project are reviewed.

Section 1.2 gives an overview of cardiac gating, and Section 1.3 deals with the challenges of ECG measurement in Magnetic Resonance Imaging (MRI) machines.

1.1 ELECTROCARDIOGRAPHY (ECG)

1.1.1 *Physiological background*

As the heart undergoes depolarization and repolarization during the cardiac cycle, the electrical currents that are generated spread not only within the heart, but also throughout the body. This electrical activity generated by the heart can be measured by an array of electrodes placed on the body surface. The recorded tracing is called an electrocardiogram (ECG, or EKG). Figure 1.1 shows the schematic representation of a "typical" human ECG tracing. The different waves that comprise the ECG represent the sequence of depolarization and repolarization of the atria and ventricles.

The P wave represents the wave of depolarization that spreads from the sinoatrial (SA) node throughout the atria, and is usually 80-100 ms in duration. The brief isoelectric period after the P wave represents the time in which the impulse is traveling within the atrioventricular (AV) node and the bundle of His.

The period of time from the onset of the P wave to the beginning of the QRS complex is termed the P-R interval, which normally ranges from 0.12 to 0.20 seconds in duration. This interval represents the time between the onset of atrial depolarization and the onset of ventricular depolarization.

The QRS complex represents ventricular depolarization. Heart rate can be calculated by determining the time interval between QRS complexes. The duration of the QRS complex is normally 0.06 to 0.1 sec-
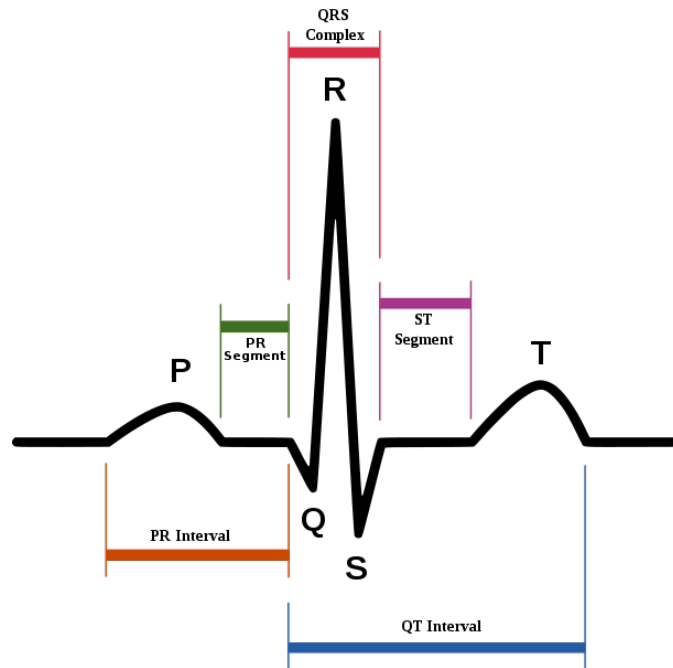
Figure 1.1: Schematic representation of normal ECG. Courtesy of *Wikipedia*.

onds. This relatively short duration indicates that ventricular depolarization normally occurs very rapidly.

The isoelectric period (ST segment) following the QRS is the time at which the entire ventricle is depolarized.

The T wave represents ventricular repolarization and is longer in duration than depolarization, since conduction of the repolarization wave is slower than the wave of depolarization.

The Q-T interval represents the time for both ventricular depolarization and repolarization to occur. This interval can range from 0.2 to 0.4 seconds depending upon heart rate.

There is no distinctly visible wave representing atrial repolarization in the ECG because it occurs during ventricular depolarization. Because the wave of atrial repolarization is relatively small in amplitude, it is masked by the much larger ventricular-generated QRS complex.

1.1.2 *Recording an ECG*

By convention, in a 5-electrode ECG configuration, electrodes are positioned on each arm and leg, and one electrode is placed on the chest. The limb electrodes are referred to as Left Arm (LA), Right Arm (RA), Left Leg (LL) and Right Leg (RL), while the chest electrode is referred to as V_1. LA, RA, LL and V_1 are used as input electrodes in monitoring the electrical activity of the heart, whereas RL is electrically driven by the Right Leg Drive (RLD) circuitry to reduce the common-mode

voltage on the other electrodes. In general, the RL signal is obtained as the average of the other limb electrodes.

The conventional ECG *leads* are formed from the *electrode* signals. The ECG leads are subdivided into the following categories:

- Standard limb leads: I, II, III

- Augmented limb leads: aVL, aVR, aVF

- Chest lead: V_1

The leads are formed according to the following rules:

$$
\begin{aligned}
I &= LA - RA \\
II &= LL - RA \\
III &= LL - LA \\
aVL &= LA - \frac{1}{2} \cdot (RA + LL) \\
aVR &= RA - \frac{1}{2} \cdot (LA + LL) \\
aVF &= LL - \frac{1}{2} \cdot (LA + RA) \\
V_1 &= V_1 - \frac{1}{3} \cdot (LA + RA + LL)
\end{aligned}
$$

The standard limb leads are *bipolar* because the lead voltage is measured between two electrodes, while the other leads are *unipolar* because they are measured with respect to the Wilson Central Terminal (WCT) voltage, which is defined as $V_W = \frac{1}{3} \cdot (LA + RA + LL)$.

In practice, ECG signal acquisition may be implemented with differential amplifiers. In this case, the standard limb leads may be obtained directly by connecting the appropriate electrodes to the inputs of the differential amplifiers. The chest lead may be obtained by generating the WCT voltage and measuring the chest electrode with respect to it. The augmented limb leads may be calculated digitally from the other leads.

In the case of small animal measurements, three electrodes are generally used. The electrode configuration may be LA, RA and LL, or alternatively, LA, RA and RL to provide feedback for common-mode signal reduction.

1.2 CARDIAC GATING

Cardiac gating is a technique used to reduce artifacts due to heart motion in medical imaging.

The heart is in motion during the cardiac cycle. It completes the contraction required for its pump function during the systolic phase,

then relaxes in the diastolic phase. Regardless of the imaging modality in use, it is desirable to create images of the heart while it is stationary in order to minimize motion artifacts. The diastolic periods of lesser cardiac motion are ideal candidates for artifact-free imaging, however, the temporal resolution of current imaging techniques is not sufficient for the creation of images in a single stationary period. This problem can be addressed with the use of ECG-synchronized techniques to record and combine data from multiple stationary periods for motion-free imaging.

The occurrence of the QRS complex in the ECG corresponds to the period of greatest heart motion. Therefore, by recording QRS triggers along with the image data, data from the periods of lesser motion can subsequently be extracted for the reconstruction of motion-free images (retrospective ECG gating). Moreover, the occurrence of QRS complexes can be predicted on the fly from the previous ECG data, which can be used in Computed Tomography (CT) imaging for reducing the output power of the X-ray tube during systole in order to reduce patient dose (prospective ECG triggering) [2].

Cardiac gating has been proven to be efficient in cardiac CT and coronary angiography [2], cardiac Positron Emission Tomography (PET) [3] and even in MRI diffusion tensor imaging of the brain [4].

1.3 ECG IN MRI

As discussed above, ECG acquisition in MRI is clinically relevant. However, it is a challenging task.

In addition to the inherent distortion of the ECG waveform in MRI systems operating at high field strengths, the static, gradient and RF electromagnetic fields of the MRI system can cause additional artifacts in the ECG signal [5]. Moreover, the ECG electrode leads may act as antennas, guiding radio-frequency interference into the MRI bore, thereby generating imaging artifacts or even causing image acquisition to fail.

An existing MRI-compatible ECG system for human applications employs long MRI-compatible cables and filtering in order to leave the electronics outside of the MRI chamber [6], but solutions allowing the monitor to be placed near the patient are also available [7, 8].

Solutions also exist for small animal applications [9, 10]. The documentation available on-line reveals very limited detail about these systems, in particular with respect to the MRI architectures and field strengths they are compatible with. However, both systems appear to rely on MRI-compatible ECG electrodes and lead wires, and the elimination of artifacts in the ECG signal by digital filtering.

SPECIFICATIONS

2

The thesis project was executed at Mediso Medical Imaging Systems, Hungary. The company's specifications for the ECG system project require the development of a modular ECG system. The main module should implement the acquisition of the ECG signals with QRS complex detection, trigger generation and transmission for cardiac gating in pre-clinical and clinical imaging machines, along with pre-clinical respiratory measurement and triggering. The slave module should implement a temperature control loop for small animal systems, measuring and regulating the temperature of the animal bed. Each device should implement two channels, i.e. it should be able to operate with two patients (animals) simultaneously, with a shared microprocessor (see Figure 2.1). Although the possibility to visualize signal waveforms is required, the project is not aimed at the development of a diagnostic quality ECG system.

The thesis deals with the development of the main module, and only describes details pertaining to the ECG functionality.
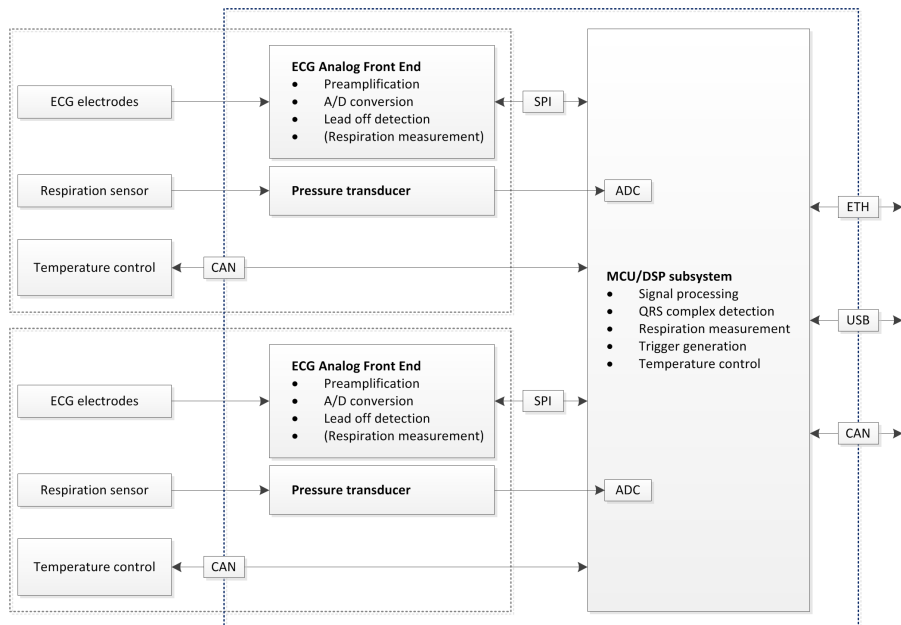
Figure 2.1: Block diagram of the ECG system.

2.1 SPECIFICATIONS OF THE ECG SYSTEM

An excerpt of the original specifications by Mediso follows.

1. Input/output

   - Digitization of signals from 4 ECG electrodes in parallel for 7-lead ECG

   - RLD bias electrode

   - ECG cables with standard connectors should be usable

   - Communication interfaces: IEEE 1588 Ethernet, Universal Serial Bus (USB), Controller Area Network (CAN)

2. Electronics, signal processing

   - Digitization of human and rat/mouse ECG signals

   - Automatic QRS complex detection, trigger generation and transmission

   - Configurable ECG preamplifier gain

   - Power line interference filter

   - Switchable high- and low-pass filters

   - Automatic detection and reporting of ECG electrode faults

   - Protection against overvoltages from defibrillators

3. Safety

   - Patient isolation: at least 4 kV from communication interfaces and power network

   - Defibrillator protection: at least 5 kV at inputs

   - Compliance with the IEC 60601-2-25 standard for medical electrical equipment

4. Design parameters (see Table 2.1)

Table 2.1: Design parameters of the ECG system.

	Min	Typ	Max
Supply voltage (DC)	9 V	24 V	36 V
Number of ECG electrodes		4 + 1	
Analog preamplification	1x	6x	12x
AD converter resolution	12 bits		
Input bandwidth	0.05 Hz		250 Hz
Low pass filter frequency (human)		35 Hz	
Low pass filter frequency (animal)		75 Hz	
High pass filter frequency		0.05 Hz	
Power line filter notch frequency		50 Hz / 60 Hz	
Input sensitivity	10 μV		
CMRR	100 dB		
Input voltage range		$\pm$300 mV	
Input impedance		20 MΩ	
Input-referred noise			15 μV
Heart rate (human)			400 bpm
Heart rate (animal)			800 bpm
Trigger latency			100 ms
Trigger jitter			10 ms

PROTOTYPING

3.1 CHOICE OF ARCHITECTURE

Many digital ECG designs are based on discrete electronic components. Depending on the level of complexity used, the entire analog signal processing stage is implemented using numerous separate Integrated Circuits (ICs) up to the point of Analog-to-Digital (AD) conversion. Publicly available examples include simple, 2-electrode ECG monitor circuit designs by Analog Devices [11] and by Texas Instruments (TI) [12], and the ADS1258 Medical Development Kit (MDK) by TI [13].

However, at present several Analog Frontend (AFE) ICs are available that implement a significant portion of the analog functionality required for ECG in one piece of silicon [14]. This might include lead signal preamplification, an RLD DC bias circuit, generation of the WCT voltage used as reference voltage, ECG lead-off detection, respiration measurement over the ECG electrodes and AD conversion.

In order to facilitate hardware development and accelerate the design, I decided to execute the project based on an AFE IC. Two options were commercially available for consideration:

- the ADS119x/129x[1] family by TI,

- the ADAS1000[2] by Analog Devices.

The features put forward by both candidates are very similar. In contrast to TI's high-end ICs, the ADAS1000 only has 5 input channels, but this would be sufficient to implement the 4+1 electrode ECG required by the specifications (see Chapter 2). However, the ADAS1000 is still in the pre-release phase at the time of writing this document, therefore TI's offer was the reasonable choice.

The TI ADS129x family offers better performance than the ADS119x, notably in terms of resolution, noise and Common-Mode Rejection Ratio (CMRR). Optionally, it also integrates the possibility to implement respiration measurement over ECG electrodes [15]. Moreover, it is relatively inexpensive, thus it was chosen for development.

The prototyping phase was carried out using an ADS1298R IC, but the final product does not require all the input channels it offers and, depending on the configuration, the respiration capability might also

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1 http://www.ti.com/ww/en/analog/ads1298/index.shtml?DCMP=analog_
signalchain_mr&HQS=ads1291-pr

2 http://www.analog.com/en/analog-to-digital-converters/ad-converters/
adas1000/products/product.html
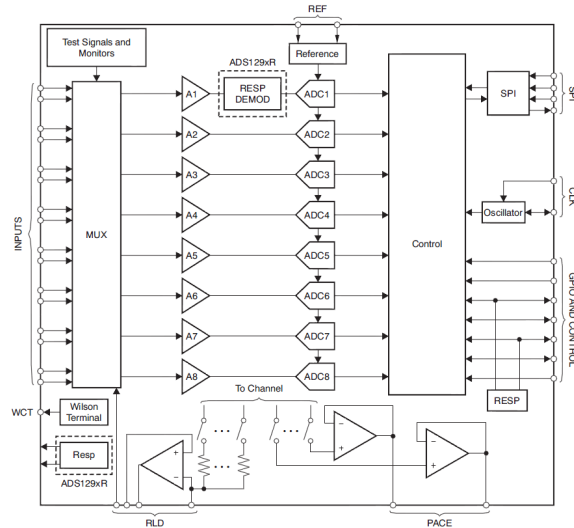
Figure 3.1: Block diagram of the ADS1298R [15].

be unnecessary, so one of the less expensive pin-compatible variants may be used in the end-product.

3.1.1 The ADS1298R IC

The ADS1298R[3] is an 8-channel AFE meant for biopotential measurements (see Figure 3.1). It features eight low-noise Programmable Gain Amplifiers (PGAs) and eight high-resolution Analog-to-Digital Converters (ADCs) for fully parallel data acquisition at a wide range of configurable data rates. It offers low power consumption and small input-referred noise alongside a high CMRR. It provides built-in RLD circuitry, lead-off detection and WCT circuitry for ECG. It also integrates support for respiration impedance measurement over ECG electrodes.

The ADS1298R provides a convenient Serial Peripheral Interface (SPI) for configuration and measurement data readout, thus it requires only a limited number of additional digital control signals [15].

3.2 THE ECG PROTOTYPE

3.2.1 The ADS1258 MDK

At the beginning of development, an ECG MDK based on TI's ADS1258[4] IC was already available at Mediso. This development kit consists of an Evaluation Module (EVM)[5] by Spectrum Digital based on TI's

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3 http://www.ti.com/product/ads1298r
4 http://www.ti.com/product/ads1258
5 http://support.spectrumdigital.com/boards/evm5505/revd/
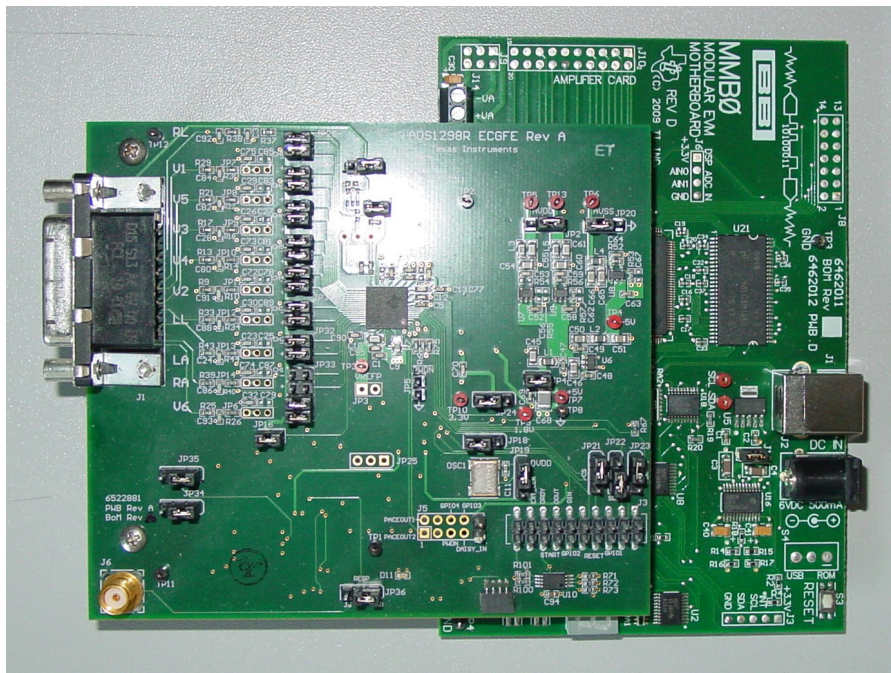
Figure 3.2: Photo of the ADS1298R ECG FE-PDK.

TMS320VC5505[6] Digital Signal Processor (DSP), and a piggy-back module hosting the ADS1258 and accompanying circuitry. The Performance Demonstration Kit (PDK) also includes the firmware running on the DSP implementing ECG signal acquisition and processing, and a PC application which receives the ECG waveforms over a serial link and plots them.

Since the ADS1258 is merely a 16-channel ADC and all ECG-related functionality is implemented using external components, this IC was not considered for design. However, the TMS320VC5505 EVM has proved to be essential in later development (see Section 3.2.3), and I also reused the PC application supplied with the MDK (see Section 3.3).

3.2.2 *The ADS1298R ECG FE-PDK*

To start experimenting with the AFE I've chosen, I ordered an ADS1298R PDK. The kit contains the MMB0 Modular EVM motherboard that hosts a TI TMS320VC5509[7] DSP, and the ADS1298R ECG AFE on a piggy-back board. The assembled system is shown in Figure 3.2.

Once out of the box and wired up to a PC, the demo kit acquired ECG signals perfectly, yet it wasn't usable for further development because of the lack of documentation and development tools for the

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6 http://www.ti.com/product/tms320vc5505
7 http://www.ti.com/product/tms320vc5509
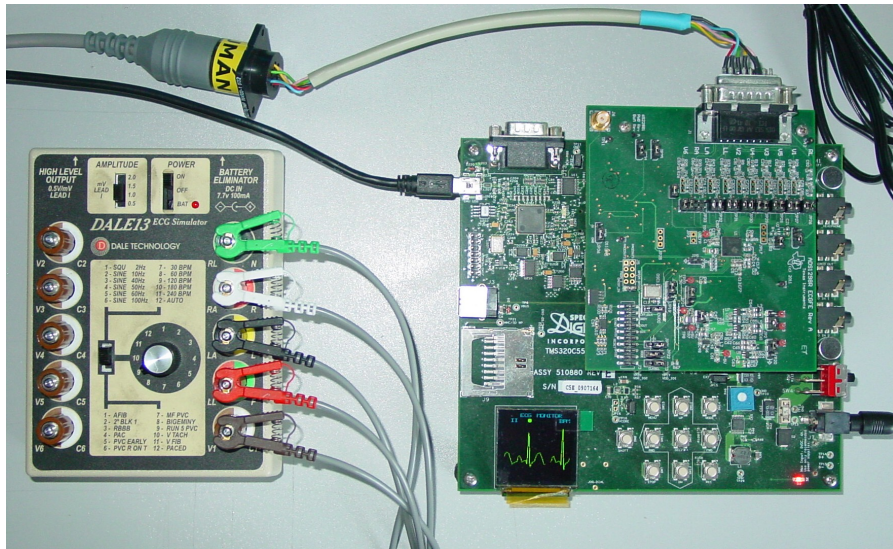
Figure 3.3: Photo of the ECG prototype system connected to an ECG simulator. The acquired ECG waveforms are visible on the LCD display.

MMB0 board. According to TI, "The MMB0 is not intended to be a full embedded development tool, it is an evaluation platform."[8]

3.2.3 *The ECG prototype system*

Fortunately, the two PDKs share the same piggy-back interface, hence mounting the ADS1298R piggy-back module on the TMS320VC5505 EVM is possible. The piggy-back interface provides all the necessary signals for the SPI interface used in the communication with the ADS1298R IC and the digital control signals. Since the EVM offers excellent development and debug support through TI's Eclipse-based Code Composer Studio suite, it is an ideal platform for a detailed exploration of the features of the ADS1298R AFE. The EVM also features a color AMOLED[9] Liquid Crystal Display (LCD) and a Universal Asynchronous Receiver/Transmitter (UART) port which were useful for visualizing the ECG waveforms both locally and through data transmitted to a PC (see Section 3.3).

Figure 3.3 shows the ECG prototype system connected to an ECG simulator (see Section 5.1.1). The digitized output of the simulator is displayed on the LCD screen.

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8 Official comment on http://e2e.ti.com/support/data_converters/precision_data_converters/f/73/t/151713.aspx, retrieved on 09/05/2012.

9 Active-matrix organic light-emitting diode.

3.3 FIRMWARE DEVELOPMENT

Since the ECG prototype system is implemented using the DSP module of the ADS1258 MDK, several features of its firmware are based on the firmware supplied with the MDK by TI.

3.3.1 *The firmware of the ADS1258 MDK*

Following are the features of the firmware supplied with the ADS1258 MDK, as listed in TI's application note [13]. I also appended brief explanations, mainly focusing on how they relate to the features required in the firmware of the ECG prototype system (see Section 3.3.2).

Data acquisition through ADC

The firmware establishes communication with the ADS1258 IC over the SPI bus for configuration and data retrieval. The ADS1298R also has an SPI interface, however, the protocol is different. In addition to adapting the protocol, I rewrote the entire SPI implementation in the ECG prototype system firmware for clarity.

Lead-off detection

ECG lead-off detection is implemented using external circuitry on the ADS1258 MDK board. The firmware takes care of the communication with these. Since the ADS1298R implements lead-off detection internally and the status information can be read out over SPI, this feature is not required in the ECG prototype system.

DC signal removal

The firmware implements a first-order digital Infinite Impulse Response (IIR) filter for baseline restoration of the ECG signal. The frequency response of the filter is shown in Figure 3.4. The filter is implemented in platform-independent C code and I reused it in the prototype firmware almost without modifications.

Multi band-pass filtering

The firmware implements a digital Finite Impulse Response (FIR) filter. The coefficients of the filter realize a low-pass filter for the elimination of high-frequency noise with a notch for the suppression of power line interference. Figure 3.5 shows the frequency response of the filter with coefficients realizing a notch at 50 Hz.

The FIR filter is implemented in architecture-specific assembly code. I reused the implementation in the ECG prototype system firmware with different filter coefficients yielding different filter characteristics.
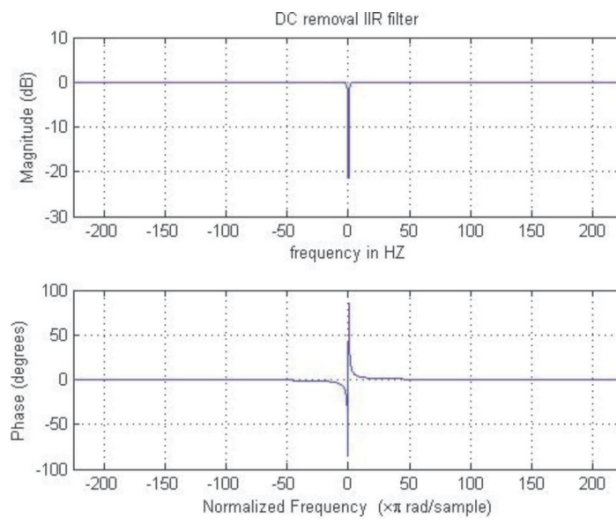
Figure 3.4: Frequency response of the IIR filter of the ADS1258 MDK [13]. The filter suppresses the DC component of the signal for baseline restoration.
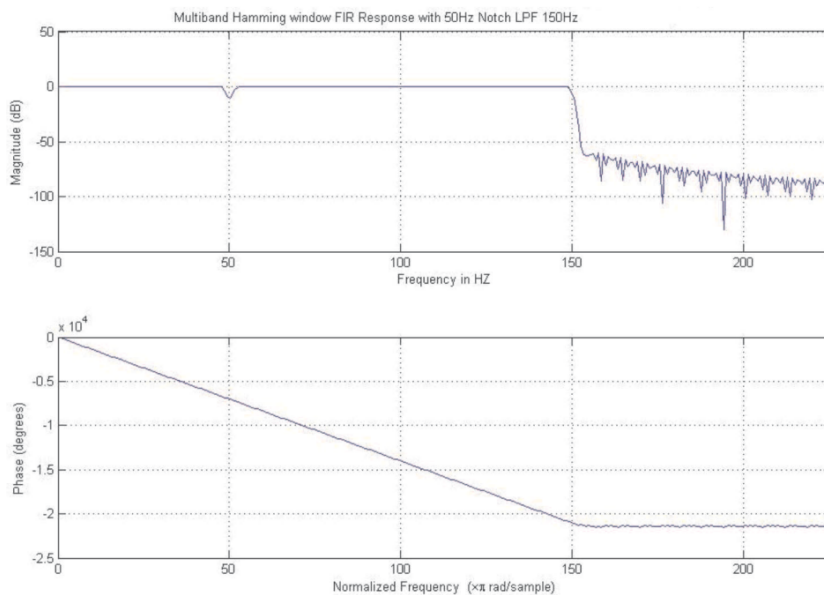
Figure 3.5: Frequency response of the FIR filter of the ADS1258 MDK [13]. The filter has a notch at 50 Hz to suppress power line interference and has a stopband with a very high attenuation above 150 Hz to filter out high frequency noise.

ECG leads formation

The firmware carries out ECG lead formation from the single ended input signals with simple arithmetics according to the rules described in Section 1.1.2. Since the ADS1298R has differential inputs, the arithmetic operations to perform are different.

QRS (heart rate) detection

The firmware implements the detection of the QRS complex (see Section 1.1.1) in the ECG signal for heart rate measurement. QRS detection is a crucial feature required for cardiac gating, but this requires real-time detection with limited latency. However, this algorithm is only aimed at heart rate measurement, it has unacceptably high latency and is unsuitable for the purposes of the ECG system.

Display of ECG data

The acquired ECG waveform is plotted on the AMOLED LCD screen with a possibility to select the ECG lead to plot using the keys of the EVM5505. This feature was useful for demonstration purposes and has been reused in the ECG prototype system with negligible modifications.

UART communication

The acquired ECG data is transmitted over UART to the PC application for display. This feature was also useful for demonstration and has been reused in the ECG prototype system firmware.

3.3.2 *The firmware of the ECG prototype system*

The main purpose of the ECG system is to issue triggers for gated imaging at the occurrence of QRS complexes with deterministic delay, i.e. limited and unvarying latency, and small jitter (see Chapter 2). The firmware has been designed with this requirement in mind. Most of the project is implemented in platform-independent C code, with the exception of the low-level CPU peripheral drivers and the FIR filter.

The flowchart of the ECG prototype system firmware is shown in Figure 3.6. A detailed description follows.

The main function

The main function performs the initialization of the entire ECG system.

It initializes the peripherals of the DSP: SPI for communication with the ADS1298R, UART for data transfer to the PC, a Successive Approximation Register (SAR) ADC that reads out keypresses for LCD configuration, the LCD for ECG waveform display, and General Purpose
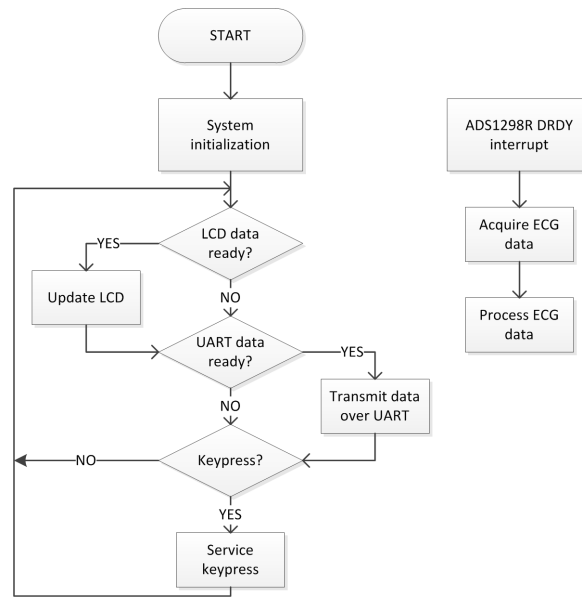
Figure 3.6: Flowchart of the ECG prototype system firmware.

Input/Outputs (GPIOs) for controlling the ADS1298R and debugging features.

It then initializes and configures the ADS1298R AFE. The register settings can be found in Listing A.1 (for detailed explanations, see [15]). The following set of main configuration parameters has proved to be usable in all cases tested so far (see Section 5.1):

- Sample rate set to 1000 SPS by default.

- The gain of all PGAs is set to 6.

- Lead-off detection is enabled for all ECG electrodes.

- The RLD and WCT signals are generated as the average of the LA, RA and LL signals (see Section 1.1.2).

Finally, it enters an infinite loop where, whenever data is available, it plots ECG waveforms to the LCD and transmits ECG data over UART to the PC for display. It also monitors keypresses and updates the LCD display configuration accordingly.

Figure 3.7 shows a screenshot of the PC application. The ECG waveforms and actual measured heart rate of a rat are displayed as acquired by the ECG prototype system. The measurement configuration only included the electrodes necessary for the calculation of Lead I (see Section 5.1.3), thus the waveform displayed for Lead II is calculated and displayed incorrectly and should be considered an arithmetic artifact.
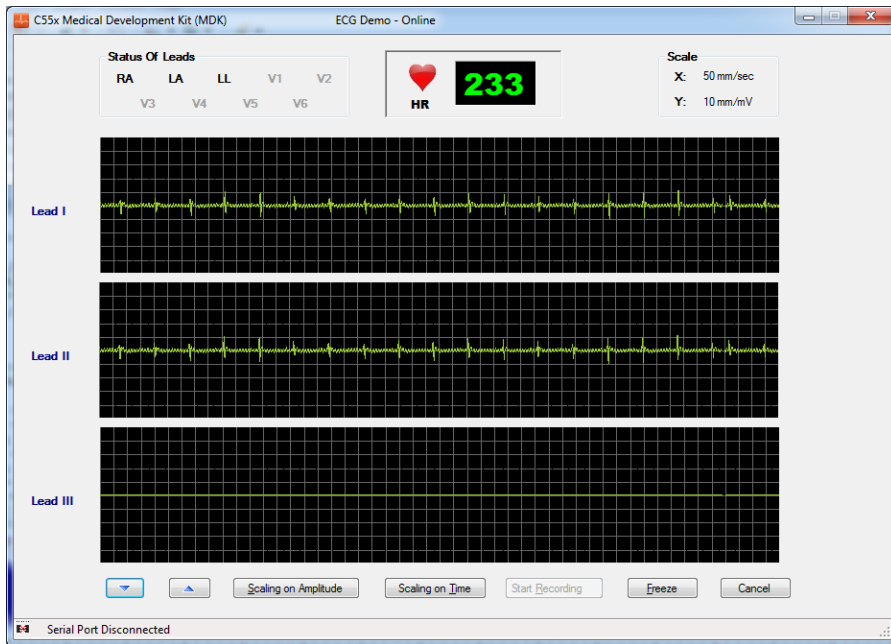
Figure 3.7: Screenshot of the PC application displaying the ECG waveforms and actual measured heart rate of a rat, acquired by the ECG prototype system.

The DRDY interrupt subroutine

The ADS1298R is configured for continuous data acquisition, thus it provides new data at the sampling rate specified in the configuration, signaling its availability through its Data Ready (DRDY) output signal. The prompt response of the ECG system to the availability of new data is ensured by an interrupt mapped to the DRDY signal. Whenever new data is available, the firmware reads it out over the SPI bus, then executes the following steps of processing:

- *Data preprocessing.* Lead-off information is extracted from the status word received from the ADS1298R.
  The 24-bit output of the ADC is converted to 16-bit resolution by truncation of a configurable number of bits. To allow for a flexible input configuration, the ADC channel values are sorted into an internal buffer according to a changeable map. The input values that contain signal from any disconnected electrode are zeroed out. Based on the lead-off status, one healthy lead is selected for QRS detection.

- *Filtering.*

  - *Baseline restoration.* A first-order IIR filter adapted from the ADS1258 MDK (see Section 3.3.1) is used for suppressing the DC component of the ECG signal. The frequency response of the filter is shown in Figure 3.8.
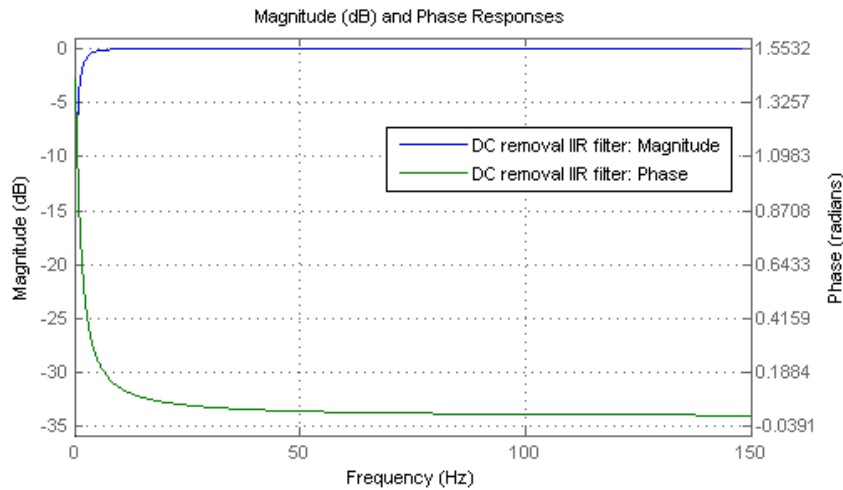
Figure 3.8: Frequency response of the DC removal IIR filter. The filter restores the baseline of the ECG signal.

– *Band-pass filtering.* I reused the FIR filter implementation in the ADS1258 MDK with a set of custom coefficients I designed in Matlab for a 128-tap FIR filter. The lower stopband of the band-pass filter reduces undesirable signal contributions from muscle movement, weakens the P and T waves and softens the QRS complex, but since the project is not aimed at the creation of a diagnostic quality ECG, this is acceptable. The higher stopband is designed to filter out high frequency noise. Figure 3.9 and Figure 3.10 show the frequency response of the filters used in human and small animal applications, respectively (see Chapter 2). Both filters have linear phase in the passband, which is a desirable characteristic since it implies equal delay at all frequencies. FIR filters have a characteristic delay which is determined by the location of the peak of the filter impulse response. Since both filters have this peak at half the number of taps, 64 samples (see Figure 3.11), the delay introduced is 64 ms at 1000 SPS.

Due to the integer arithmetics used in the FIR filter implementation, the filter coefficients were converted from double-precision floating point to integers. As a result, preserving the relatively flat response in the passband, the filters have significant amplification. However, with sufficient care taken to avoid arithmetic overflows, this was very easily compensated.

– *Notch filtering.* Since the stopband of the band-pass filter for preclinical applications lies above the frequency of power line interference, in these applications, a separate filter is required for the suppression of these undesirable contribu-
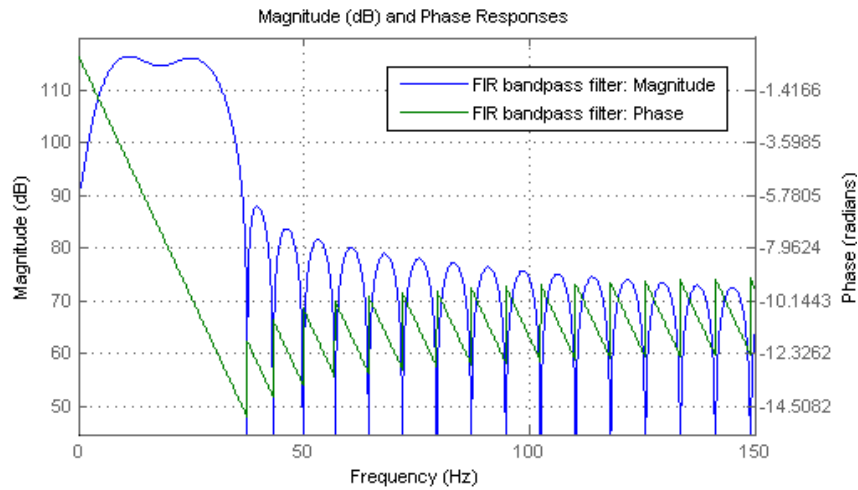
Figure 3.9: Frequency response of the human band-pass FIR filter. Smaller heart rates allow for a lower cutoff frequency.
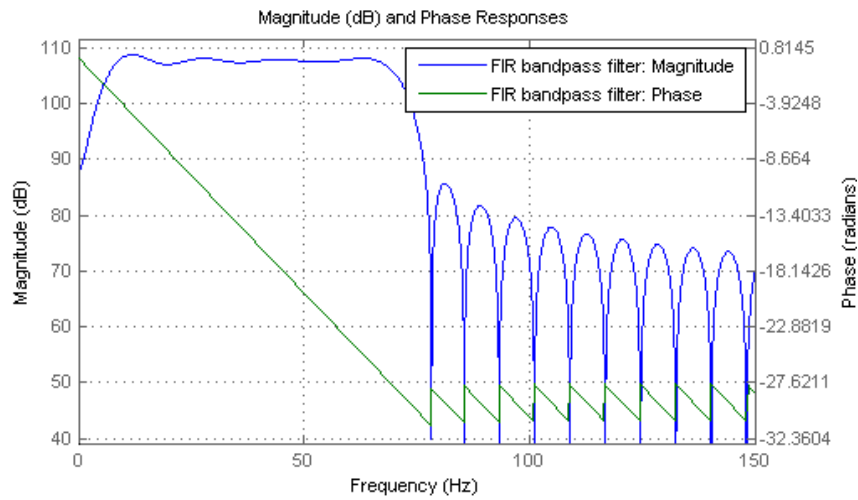
Figure 3.10: Frequency response of the small animal band-pass FIR filter. Higher heart rates require a wider passband.
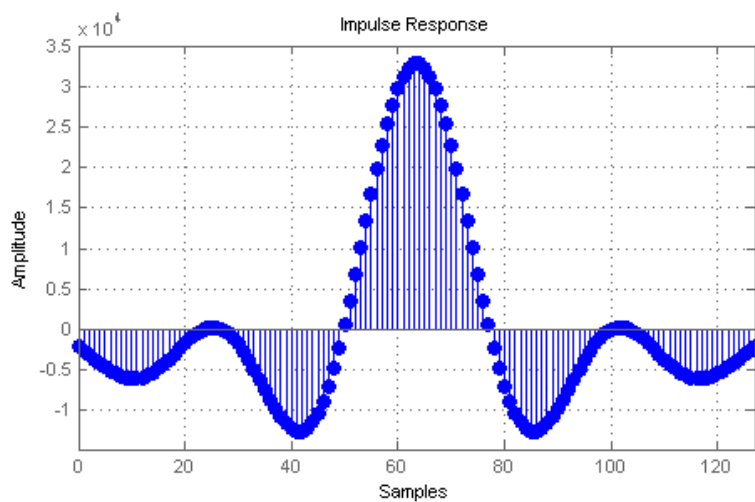
Figure 3.11: Impulse response of the human band-pass FIR filter.
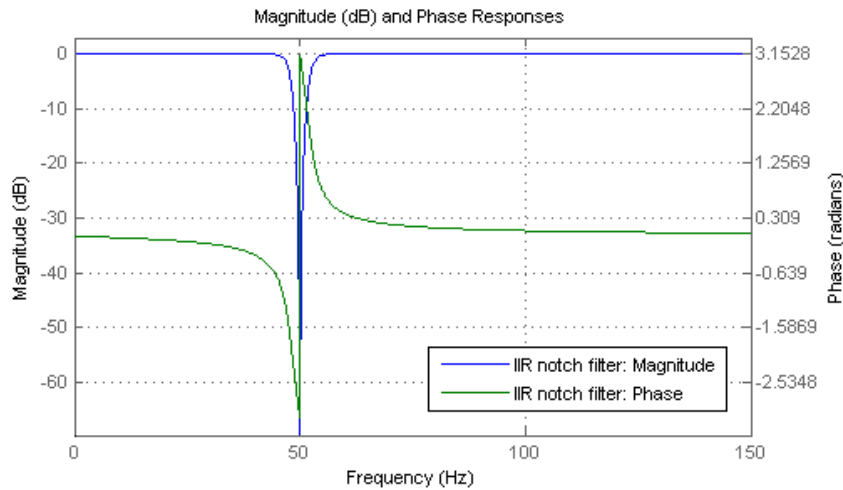
Figure 3.12: Frequency response of the IIR notch filter for power line interference suppression. Notch frequency is 50 Hz. The filter has a relatively blunt notch to reduce transients.

tions. For this purpose, I designed an IIR filter in Matlab. Based on the coefficients obtained, I implemented this filter as a cascade of two second-order IIR stages in platform-independent C code. Figure 3.12 depicts the frequency response of the filter with a 50 Hz notch.

- QRS *detection.* As discussed in Section 3.3.1, the QRS detection algorithm implemented in the ADS1258 MDK is unsuitable for the purposes of the ECG system. Therefore, I implemented a detection algorithm based on the Pan-Tompkins algorithm [16].
  The algorithm relies on the previously filtered signal. As mentioned above, the band-pass filter distorts the P and T waves in the ECG signal. It also softens the peak of the QRS complex, however, by subsequent application of a five-point digital derivative filter and squaring its output, the output has proved to be an excellent metric for QRS detection in all cases tested (see Section 5.1). Pan and Tompkins applied a moving window average filter to this signal, which yields a series of pulses and allows for QRS detection through rising edge detection. I simply perform threshold crossing detection on the squared derivative with an adaptive threshold established based on the last five detection amplitudes. Each detection inhibits further detections for a configurable amount of time, thereby avoiding multiple detections of the same QRS complex.

- *UART and LCD flag generation.* Not all samples are sent to the UART and LCD. The ratio of decimation is configurable and the corresponding data ready flags, processed by the main loop, are generated accordingly.

HARDWARE DESIGN

I carried out the hardware design in cooperation with my colleague Sándor Török. Many design choices originate from his professional experience.

4.1 CHOICE OF MICROPROCESSOR

The specification of the project requires the implementation of a variety of communication interfaces (see Chapter 2). Among those specified, the TI TMS320C5000 Ultra Low-Power DSP family supports USB, but not IEEE 1588 Ethernet or CAN. These interfaces could have been realized using external ICs, but in order to keep the number of electronic components to a minimum, I decided to search for a microprocessor that has integrated support for all required interfaces. This means a change of architecture, but since a significant part of the ECG system firmware is platform-independent high-level C code, porting the code is possible with reasonable effort.

In order to provide ample processing power for filtering the ECG signal, I started looking for another DSP. Among the manufacturers I considered - TI, Analog Devices and Freescale -, TI appeared to have the most detailed documentation, and they offer excellent on-line tools for microprocessor selection[1]. In addition, I had already familiarized myself with their design tools, therefore I decided to choose a microprocessor by TI.

In TI's high performance DSP & ARM CPU line, I was left with two candidate families. The high-end DaVinci DSP family[2] has several members that offer all required communication interfaces but the capabilities of these processors far exceed the requirements of the project and they are also rather expensive. The selection tool also recommended the Sitara family[3]. Even though they are not DSPs, these ARM-based CPUs are aimed at performance applications and seemed to offer sufficient processing power for the ECG application at an affordable price. However, before I could order a development kit, a colleague brought TI's Concerto Microcontroller Unit (MCU) family[4] to my attention. These microcontrollers feature all communication interfaces required by the specification, and development boards were readily available at the office.

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[1] http://focus.ti.com/en/multimedia/flash/selection_tools/dsp/dsp.html

[2] http://www.ti.com/lsds/ti/dsp/platform/davinci/device.page

[3] http://www.ti.com/lsds/ti/dsp/platform/sitara/device.page

[4] http://www.ti.com/mcu/docs/mcuproductcontentnp.tsp?sectionId=95&familyId=2049&tabId=2743
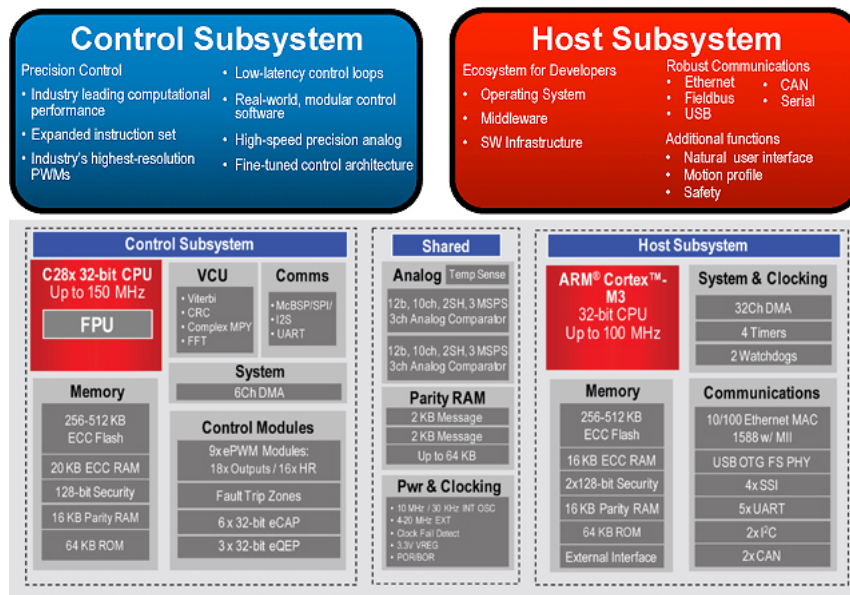
Figure 4.1: The architecture of the Concerto MCU family [17].

The Concerto family is based on a very interesting architecture, which is shown in Figure 4.1. These MCUs combine an ARM Cortex-M3 core aimed at communication with a C2000 DSP core that offers similar performance to the C5000 DSP I had used.

I carried out preliminary tests with the development board. I ported the most computationally intensive part of the firmware, the FIR filter (see Section 3.3) to the C2000 DSP core, and I also experimented with the communication between the two cores. These tests suggested that the high-end Concerto F28M35H52C1[5] would be very suitable for the ECG system project. Since it is quite reasonably priced and also offers an integrated recyclic ADC useful for small animal respiratory measurement (see Section 4.2.4), it was chosen for hardware development.

4.2 HARDWARE DESIGN CONSIDERATIONS

Since the specification of the ECG system requires the implementation of two channels on a single board, the entire ECG hardware (see Section 4.2.3) and small animal respiratory measurement stage (see Section 4.2.4) is doubled.

Nearly all digital signal lines in the design feature series resistors for limiting transient currents.

The design features several LEDs driven by the MCU to provide a direct means of conveying information to the user.

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5 http://www.ti.com/product/f28m35h52c

4.2.1 *Patient protection*

The IEC 60601-2-25 standard for ECGs requires that in human applications, the patient be electrically isolated from the power network and communication interfaces of the device. The breakdown voltage must be at least 4 kV.

The alternatives considered for the implementation of the isolation are the following:

- Isolation on the analog side, before the inputs of the ADS1298R.

- Isolation of the signal paths between the ECG AFE and the MCU, keeping the AFE in a separate power domain.

- Isolation of the external communication interfaces and power, keeping the MCU and the AFE in the same isolated power domain.

Creating separate, isolated power domains for each ECG AFE and isolating the digital signal paths to the MCU offers several advantages over the two other methods. Excellent methods are readily available for the isolation of on-board digital signal paths, making this alternative less cumbersome than the medical quality isolation of the communication interfaces and more robust than the isolation of analog signals. Moreover, this solution intrinsically fulfills the requirements for isolation between the two ECG stages for two patients, thus it appeared to be the best choice.

In order to provide the required isolation strengths, the portions of the board containing the ECG AFEs have been protected against leakage currents by incisions in the board layout (see Figure 4.3).

Additionally, the main power supply (see Section 4.2.2) and the CAN interfaces (see Section 4.2.6) are also electrically isolated, even though no medical quality isolation is provided in these cases.

4.2.2 *Power supplies*

The external components required for each regulator have been selected following the guidelines set forth in the respective datasheets. A number of ferrite bead chokes have been used throughout the design for the suppression of high frequency interference on power supply lines.

Main power supply

Since the highest supply voltage required in the ECG system is 5 V, this has been chosen as the output voltage of the main power supply. In order to comply with the requirement for a wide input voltage range and provide electrical isolation between the power network and

the board, the TEN5-2411WI DC-DC converter by TracoPower has been selected. This power supply provides a regulated output voltage of 5 V with a maximal output current of 1 A, and offers an efficiency of typically 78% [18]. Its output current is sufficient for the application, and since the sensitive components require 3.3 V power, its ripple will be suppressed by additional linear regulators.

ECG power supplies

The ECG circuitry resides in a separate power domain, which must be isolated from the rest of the board to comply with the requirements of the standard (see Section 4.2.1). In order to achieve this, the RV-0505S DC-DC converter by Recom was used. This device possesses the medical approval required by the application. With an input voltage of 5 V, it supplies a maximum output current of 400 mA at an unregulated output voltage of 5 V, with standard efficiencies of 70-75% [19].

Since the ADS1298R requires 3.3 V power and the output of the isolated power supply is unregulated, additional regulation is necessary. In order to minimize the ripple of the output voltage, linear regulation is desirable. Moreover, to suppress interference from the digital side of the IC to the analog side and thereby reduce the noise of the ADC, the best solution is having two separate regulators provide power to the analog and digital domains of the AFE. The TPS79201 by TI was chosen for its ultralow noise and high Power Supply Rejection Ratio (PSRR) [20]. Its current sourcing capabilities are largely sufficient for the requirements of the ECG stage.

CAN power supplies

The two CAN interfaces used (see Section 4.2.6) are also isolated. Their isolated power domains are supplied by TME 0505S isolated DC-DC converters from TracoPower, which provide a maximal output current of 200 mA at 5 V with an input voltage of 5 V, with a typical efficiency of 70% [21].

MCU power supply

The F28M35H52C1 MCU can operate with a single 3.3 V power supply since it has internal regulators supplying all other voltages it requires. TI's TPS79601 was selected for its excellent performance similar to the regulators used for the ADS1298R and its higher current sourcing capabilities up to 1 A [22]. Since in this case, the suppression of noise to the analog subsystem is not a critical issue, the same regulator supplies both digital and analog power, with a ferrite bead choke separating the two power domains to ensure high frequency noise reduction.

4.2.3 *ECG hardware*

The external components required by the ADS1298R AFE have been
selected following the guidelines of the datasheet [15].

Patient protection

As discussed in Section 4.2.1, medical quality electrical isolation is
required in ECG systems. In addition to placing the ECG electronics
in an isolated power domain, this also requires the isolation of the
digital lines to and from the AFE. Traditionally, optocouplers were
used for this purpose, but this technology suffers from aging effects
in the LED and difficulties at higher signaling speeds. The *i*Coupler
technology by Analog Devices overcomes these limitations, it offers
far superior performance and significantly lower power consumption
[23]. The ADuM2401 device chosen for isolation offers a suitable num-
ber of input and output channels, can operate at the foreseen SPI bus
frequency of 10 MHz and complies to the requirements of the IEC
60601 standard.

   Since the two ECG AFEs share the same SPI bus, additional buffers
gated with the respective SPI Enable signals were required on the
Master In Slave Out (MISO) lines to avoid two ADuM2401 outputs con-
currently driving the MISO input of the MCU.

   The clock signal is routed to both AFEs from the same 2.048 MHz
clock generator residing in the main power domain through the elec-
trical isolators.

The analog input stage

The ADS1298R features a Delta-Sigma ($\Delta\Sigma$) AD converter. The $\Delta\Sigma$ mod-
ulator and the on-chip digital decimation filters can be used to filter
out the noise at higher frequencies, thus the complexity of analog
antialiasing filters required can be dramatically reduced with respect
to Nyquist ADCs. Effectively, the input antialiasing filters are only re-
quired to filter out any interference at frequencies around multiples
of f_{MOD}, the sampling frequency of the modulator [15]. In High Res-
olution mode, $f_{MOD} = f_{CLK}/4 = 512\,kHz$.

   Therefore, only a simple RC filter is required for antialiasing at the
input of the ADS1298R ICs, which was adapted from the circuit design
put forward by TI in [24]. Figure 4.2 shows the magnitude response
of the antialiasing filter. Complemented by the input EMI filter within
the ADS1298R IC [15], this ensures the required antialiasing.

   A neon lamp was added in parallel to the filter in order to provide
the defibrillator protection required by the specifications on the in-
puts of the AFE IC. The neon lamp has a maximum strike voltage of
90 V [25]. In conjunction with the resistor in series, it limits transient
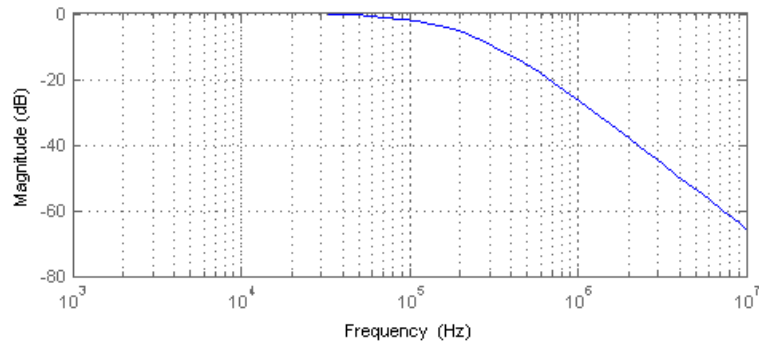
Figure 4.2: Frequency response of the input antialiasing filter. The graph confirms that, together with the EMI filter within the ADS1298R, the filter ensures the antialiasing required.

currents reaching the inputs of the ADS1298R to approximately 9 mA, which the inputs can withstand [15].

4.2.4 *Small animal respiratory measurement*

Based on previous experience, the SDP1108-R differential pressure sensor by Sensirion has been chosen for the purpose of respiratory measurement. It requires a 5 V DC power supply and provides a fully calibrated analog voltage output. It offers a full scale pressure difference of 500 Pa with superior accuracy, outstanding resolution and low temperature dependence through internal compensation [26].

The proposed measurement configuration consists of a tiny respiration sensor cushion[6] the small animal lies on for measurement. The output of the cushion is connected to the Hi terminal of the SDP1108-R, while the Lo terminal is left open, thus atmospheric pressure is used as reference. The performance of the SDP1108-R makes it highly suitable for measuring the small volume flows generated by the cushion through the breathing of the animal.

The analog output voltage of the SDP1108-R is connected to an ADC input of the MCU for digitization. Since the output voltages range from 0.25 V to 4 V and the ADC operates with a reference voltage of 3 V, a voltage divider is required between the output and the input. In addition, an optional RC filter has been added to the design to ensure the possibility of limiting high-frequency noise.

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6 E.g.   http://www.medicare.ie/mother-baby/infant-respiration-monitors-rental-1/graseby-respiration-sensor.html.

4.2.5   *Microcontroller unit*

The quartz oscillator, decoupling capacitors, pullup/pulldown resistors and other required external components have been laid out according to the guidelines of the datasheet [27].

A JTAG interface has been implemented for programming and debugging.

The F28M35H52C1 offers a total of five SPI interfaces. Both ECG AFEs are connected to the only one of those that is controlled by the DSP core to allow for readout and data processing by the same core.

In addition to the output of the pressure sensors (see Section 4.2.4), two supply voltages have been connected to ADC inputs of the Concerto through resistive dividers to allow monitoring the state of the power supplies.

4.2.6   *Communication interfaces*

Ethernet

In addition to the Ethernet MAC in the Concerto, a transceiver IC and an RJ-45 connector is required for Ethernet communication. This circuitry has been adapted from the circuit design of the TI H52C1 Concerto ControlCard [28].

USB

The F28M35H52C1 includes the implementation of the physical layer, therefore only a USB connector is required for communication. Since the ECG system is only meant to be used as a USB device, a simple USB-B connector has been implemented in the design, along with protection diodes on the data lines.

CAN

The ECG system design makes use of both CAN interfaces of the Concerto. CAN is used as an external communication interface and also for connecting to the slave modules (see Chapter 2). 9-pin D-SUB male connectors are used.

To complement the isolated power supplies, TI's ISO1050 isolated CAN transceivers [29] have been used in the design for CAN interface isolation.

In addition to protection diodes on the CAN data lines, the slave module ("bed") CAN circuitry features a solution that disables the local CAN termination resistor if the slave modules for both channels are attached. In contrast, the external CAN interface complies to the protocol in use at Mediso, thus it features a termination resistor that can be enabled externally through the D-SUB connector.
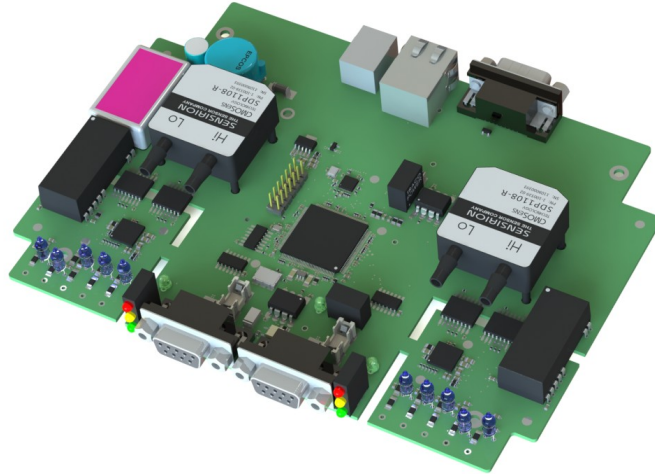
Figure 4.3: Rendered visualization of the populated ECG printed circuit board. Note the incisions protecting the ECG circuitry from leakage currents.

4.3 THE HARDWARE

Based on the aforementioned considerations, the schematic (see Appendix A) and layout design of the final printed circuit board have been completed. At the time of writing this thesis, the circuit board is being manufactured. Figure 4.3 shows a rendered visualization of the populated printed circuit board.

MEASUREMENTS

Since no final ECG system hardware has been produced yet at the time of writing this thesis, all measurements described in this chapter were carried out using the ECG prototype system. However, the expected performance of the final hardware is very similar.

The output of the ECG system has not yet been calibrated for converting output values to voltages, thus all amplitude axes show values in arbitrary units.

5.1 ECG SIGNAL ACQUISITION

5.1.1 *ECG simulator*

Initial ECG measurements were made with an ECG simulator, a device featuring connectors for ECG lead wires and capable of generating ECG waveforms. The ECG system was connected to the simulator with a standard human ECG cable in a 5-electrode configuration, as shown in Figure 3.3. The following examples illustrate the performance of the system with input signals from the simulator.

Figure 5.1 shows a recording of the ECG simulator output with amplitude set to 1 mV and heart rate set to 60 bpm. The unfiltered ECG data features very characteristic P and T waves and QRS complex, along with a negligible contribution from power line interference. In this case, the unfiltered data would be directly usable for QRS detection, however, filtering was carried out for comparison. Since the bandpass filter for human applications was used, use of the notch filter for power line interference suppression was not required.

Suppression of the DC component by the baseline restoration filter is well visible in the filtered data. The limited delay and waveform distortion resulting from the FIR filter are also identifiable, however, these effects are acceptable (see Section 3.3.2). As a result, the discrete derivative provides a perfectly usable QRS metric.

Figure 5.2 depicts a recording of the output of the simulator with the same settings (note the different scale on the time axis), deliberately corrupted by 50 Hz power line noise by placing a soldering iron close to the ECG system. The filtering does a remarkable job in eliminating power line interference and a usable QRS metric is produced.
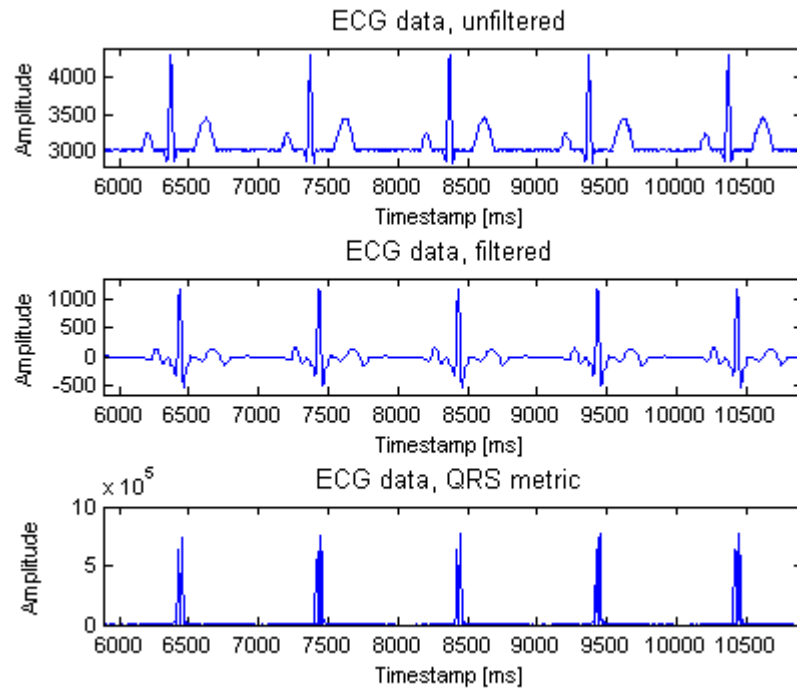
Figure 5.1: Recording of the ECG simulator output.
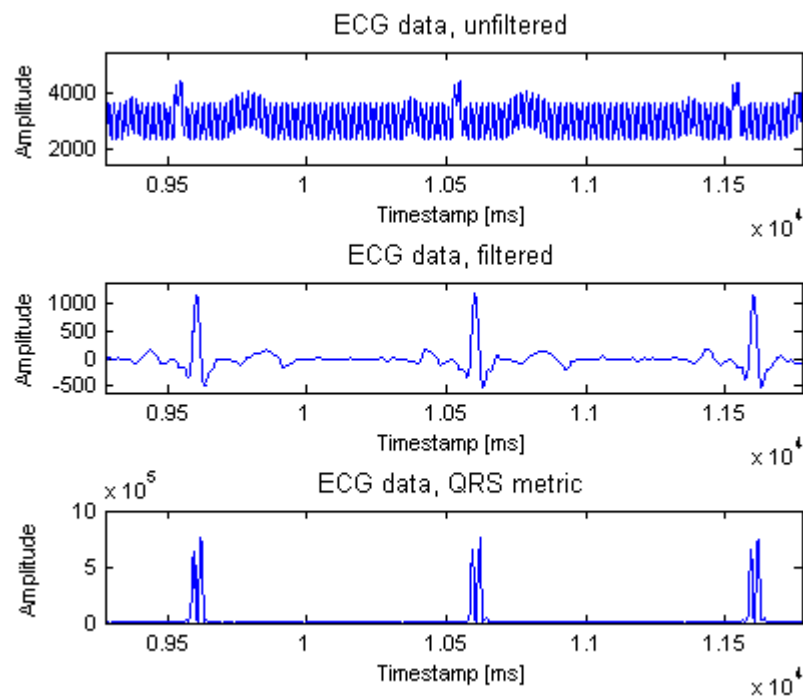
Figure 5.2: Recording of the ECG simulator output, corrupted by 50Hz noise. This figure demonstrates the efficiency of the notch filter in eliminating power line interference.
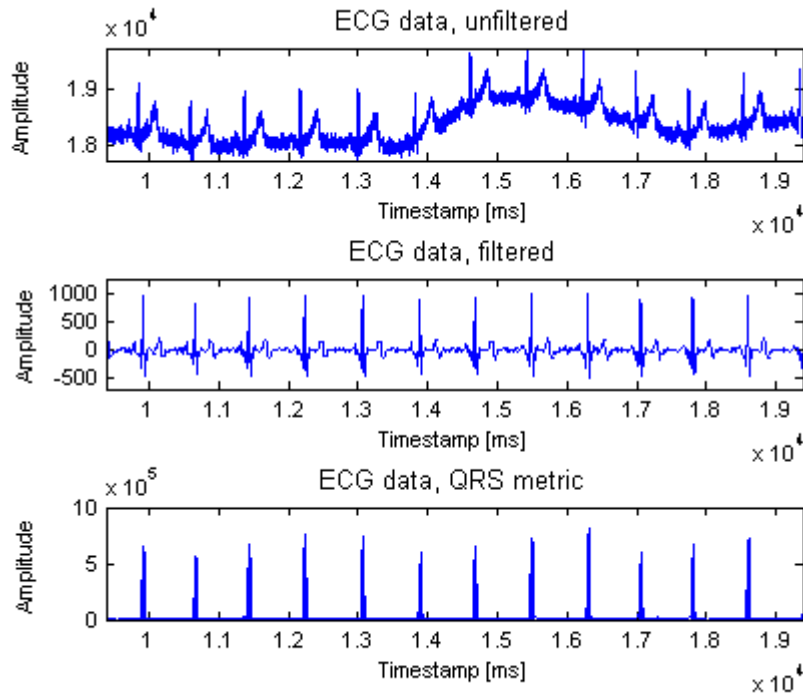
Figure 5.3: Human ECG recording. This figure illustrates the power of the filters in suppressing power line interference and baseline wander.

5.1.2 *Human ECG*

Figure 5.3 shows the author's ECG recording at rest, acquired in a 3-electrode configuration with LA and RA input electrodes, and the RL feedback electrode connected to the appropriate limbs. The recording exhibits noticeable power line interference and breathing causes significant baseline wander in the ECG trace. However, these distortions are eliminated by the DC suppression and bandpass filters. This yields a very high quality metric for QRS detection. The average heart rate during the measurement can be identified as approximately 75-80 bpm.

Figure 5.4 shows the author's stress ECG recording while performing squats. The baseline restoration filter performs remarkably once more, however, due to muscle movement, the unfiltered ECG data exhibits sharp transitions the currently used filters cannot eliminate satisfactorily. This leads to the presence of spurious pulses in the QRS metric, thus unwanted QRS triggers.

5.1.3 *Small animal ECG*

To verify its capability to acquire the ECG signals of rodents, the ECG system was tested with anesthetized small animals. These recordings
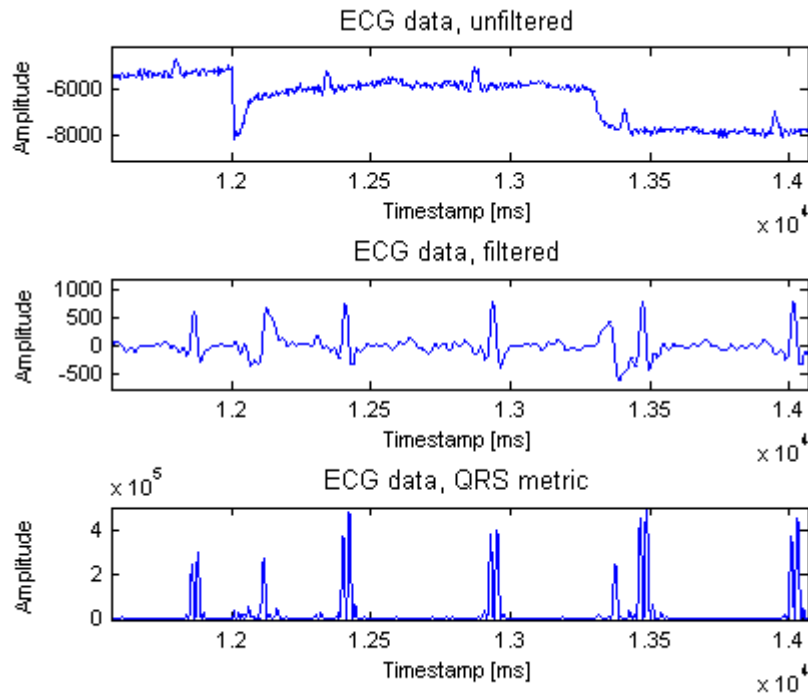
Figure 5.4: Human stress ECG recording. This figure demonstrates the challenges faced in the processing of stress ECG signals.

were carried out in a 3-electrode ECG configuration with LA and RA input electrodes and a feedback electrode, RL, connected to the limbs of the animal. 3M Red Dot neonatal, pre-wired, radiolucent monitoring electrodes were employed with a standard, unshielded ECG cable. The bandpass filter for small animal measurements was used, therefore the notch filter for power line interference was required (see Section 3.3.2).

The photo of an anesthetized rat fitted with ECG electrodes can be seen in Figure 5.5. Figure 5.6 shows the ECG signal recording. The unfiltered signal suffers from significant power line interference. Due to the transients of the notch filter, this interference is only attenuated, not completely eliminated. Filtering also reduces the amplitudes of the QRS complexes. However, both the unfiltered and the filtered signal yield a usable QRS metric through the discrete derivative filter. The heart rate of the rat was measured to be around 235 bpm (see Figure 3.7), which is in accord with the values found in the literature [30].

Figure 5.7 shows the ECG of a hamster. Surprisingly, the signal features inverted QRS complexes due to biological reasons. The signal exhibits limited baseline wander and negligible power line interference. In this case, filtering yields no obvious improvement with respect to the unfiltered signal, and the amplitudes of the QRS complexes are re-
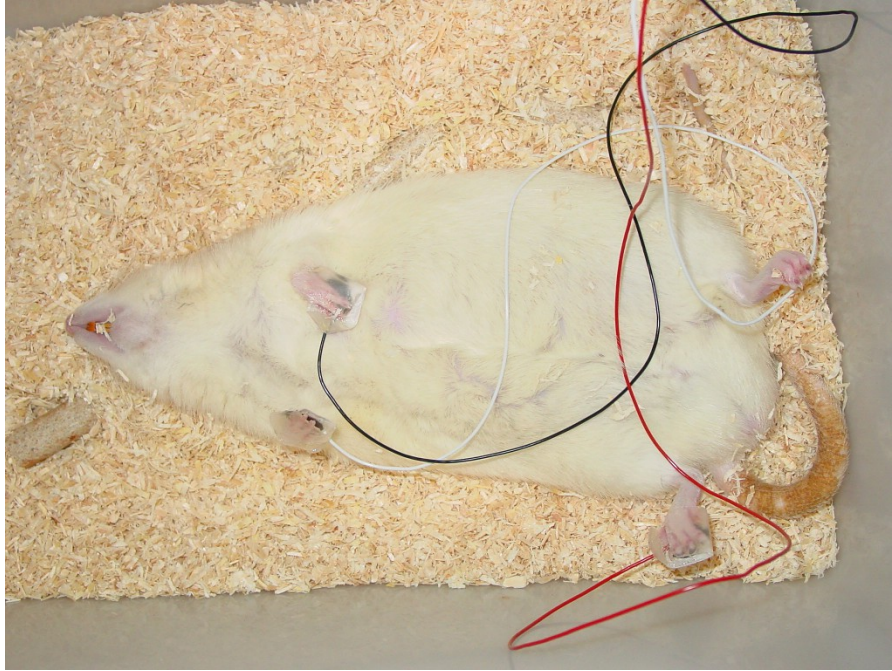
Figure 5.5: The anesthetized rat fitted with ECG electrodes.
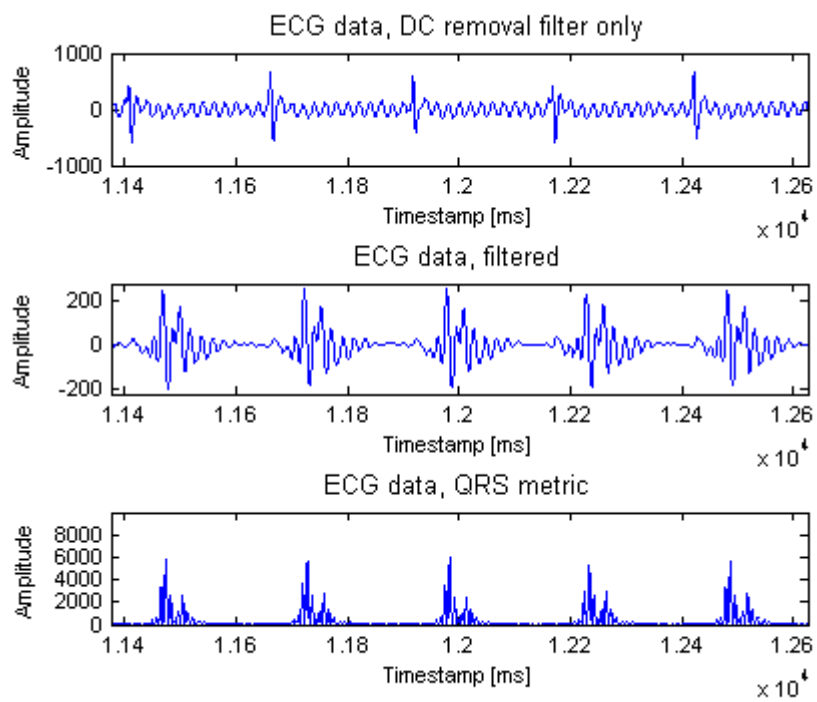
Figure 5.6: Rat ECG recording. This figure illustrates the capability of the processing to create a usable QRS metric.
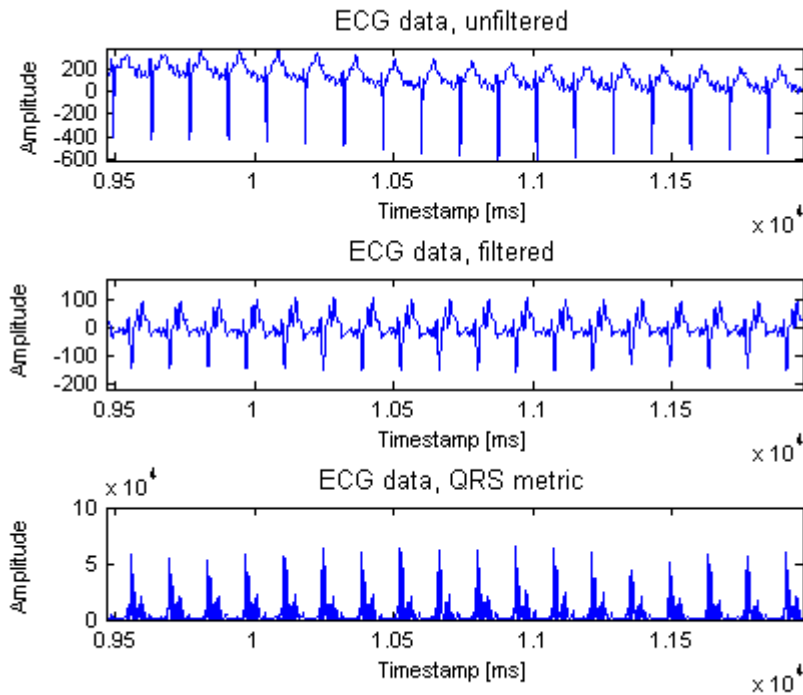
Figure 5.7: Hamster ECG recording. This figure demonstrates the ability of the processing to create a usable QRS metric.

duced. However, as was the case with the ECG of the rat, both filtered and unfiltered signals are usable for QRS detection. The heart rate of the hamster was measured to be around 430 bpm, which corresponds to the values found in the literature [31].

5.2 PERFORMANCE IN MRI

Mediso assembles permanent magnet MRI machines for small animal applications. Compared to electromagnet-based devices, these machines offer the advantage of a limited fringe field [32], thus metallic objects may be brought into the vicinity and ordinary electronic devices may be operated near the machine. This means that ECG measurement in MRI might be possible by setting up the ECG system next to the MRI machine and using MRI-compatible ECG electrodes within the bore of the MRI magnet.

The manufacturer claims that the 3M electrodes I used for small animal ECG measurements (see Section 5.1.3) are not MRI-compatible. Nevertheless, the electrodes are non-ferrous and the pre-attached lead wires are made of carbon. Moreover, Small Animal Instruments, Inc. use the same electrodes in their MRI-compatible ECG system [10], thus these electrodes were good candidates for measurements in MRI.

5.2.1 *Initial measurements*

In order to assess the viability of measuring ECG as outlined above, I carried out initial measurements with an MRI water phantom and the 3M electrodes within the magnet bore. The two ECG electrodes that form the ECG lead under measurement were connected together with approximately 100 kΩ contact impedance, which reasonably approximates usual ECG contact impedances. This setup was placed in the MRI animal bed underneath the phantom and inserted into the MRI bore. No actual signal source was connected to the electrodes, thus the measurement was expected to register background noise and eventual MRI contributions.

Figure 5.8 shows the result of the acquisitions. The upper plot depicts the spectrum of the background acquired within the MRI bore, with no MRI sequence in progress. The spectrum is clearly dominated by 50 Hz power line interference with significant harmonic content. The lower plot shows the spectrum of the data acquired during the execution of a scout MRI sequence. I deemed the results promising, since MRI contributions are only visible around 260 Hz and 480 Hz, outside of the frequency band of interest, and should therefore be easy to attenuate. It is noteworthy that 260 Hz corresponds to the repeat period of the radio frequency pulses in the scout sequence, 3.8 ms. It should also be noted that although the spectra of the 50 Hz harmonics are visually similar, MRI interference appears to amplify several harmonics.

The MRI images acquired of the phantom with the electrodes in place were only slightly deteriorated, showing limited signs of interference generated by the electrodes. The electrodes were not visible on the MRI image.

5.2.2 *Small animal measurements*

ECG measurements

Based on the positive experience gained with the initial measurements, I attempted measuring the ECG of a hamster within MRI. I used the same electrode configuration as in Section 5.1.3. Measurements with the hamster inside the MRI bore, but with the gradient and radio frequency systems of the MRI machine turned off, yielded results in excellent agreement with those obtained outside of the machine. Thus, the field of the permanent magnet appears to have no noticeable influence on the measurement.

Figure 5.9 shows the result of the measurements. The upper plot depicts the spectrum of the hamster ECG acquired within the MRI bore, with no MRI sequence in progress. It is interesting to note that the spectrum appears to be discrete, which suggests that the heart rate of the hamster is very steady. The lower plot shows the spectrum of
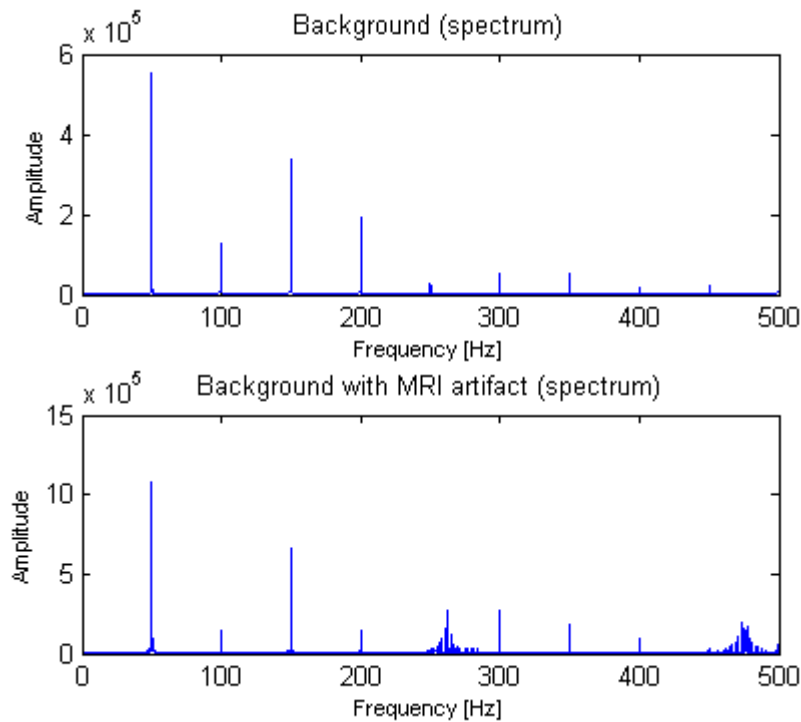
Figure 5.8: Spectrum of the background signal acquired within the MRI bore, without MRI sequence in progress and during a scout sequence.

the data acquired during the execution of a scout MRI sequence. MRI interference appears at the same frequencies that were identified earlier, however, in this case, it dominates the signal. This demonstrates a weakness of the filtering in use: Figure 5.10 shows that the low frequency spectrum of the hamster ECG is partially preserved, with some frequency components diminished. However, the filtering cannot sufficiently attenuate the high-frequency interference, thus QRS detection on the filtered signal fails.

Figure 5.11 shows an ECG spectrum acquired during execution of a spin echo MRI sequence. The high frequency components of the spectrum were attenuated by the small animal band-pass filter. This MRI sequence appears to cause a wide-band interference within the frequency band relevant for ECG acquisition, along with some very high peaks. QRS detection on this signal yields no usable results.

MRI measurements

In contrast to the phantom measurements described in Section 5.2.1, no usable MRI images of the hamster fitted with ECG electrodes could be acquired with the sequences tested.
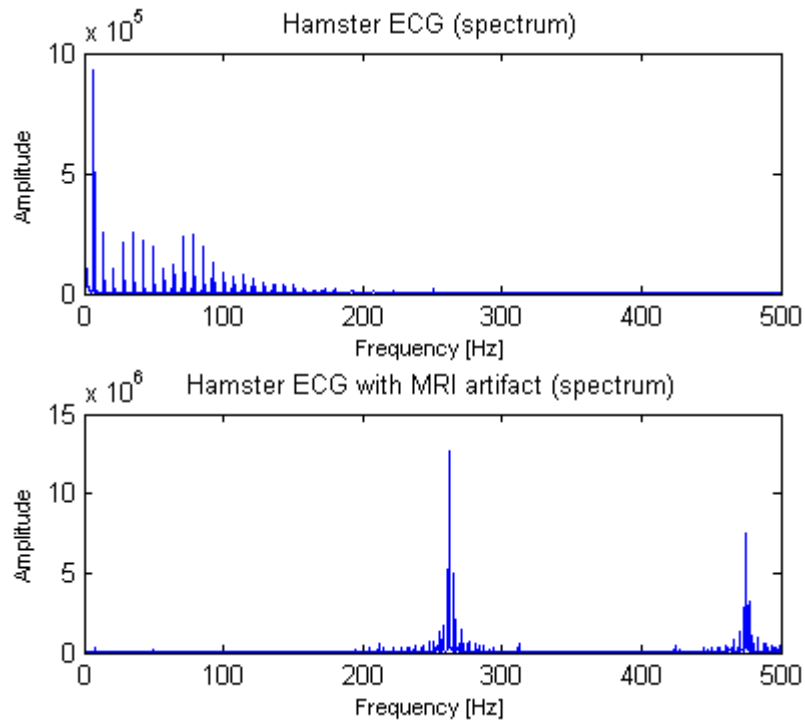
Figure 5.9: Spectrum of a hamster ECG acquired within the MRI bore. During the scout sequence, MRI artifact dominates the spectrum.
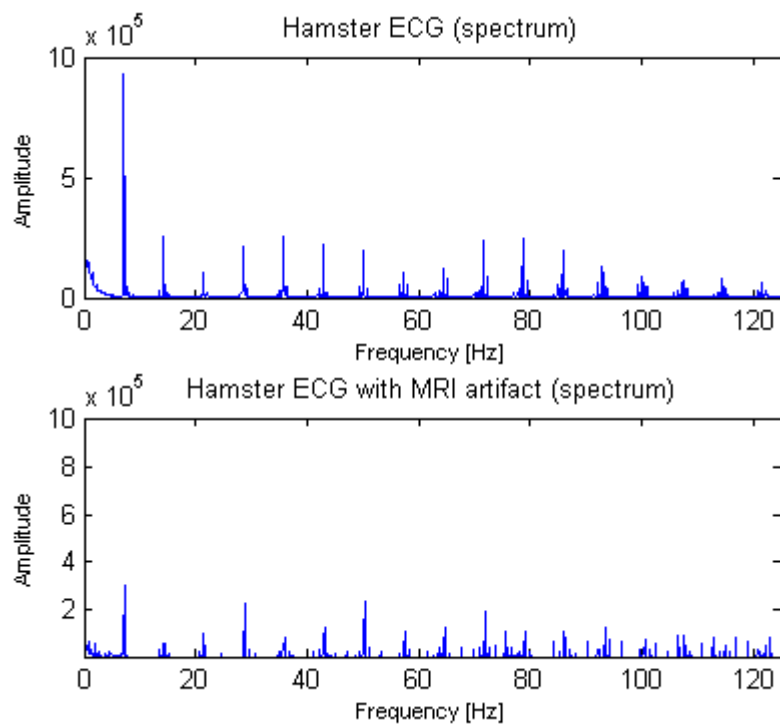
Figure 5.10: Spectrum of a hamster ECG acquired within the MRI bore, zoomed in. The spectrum has been partially preserved at low frequencies, even despite MRI artifact from the scout sequence.
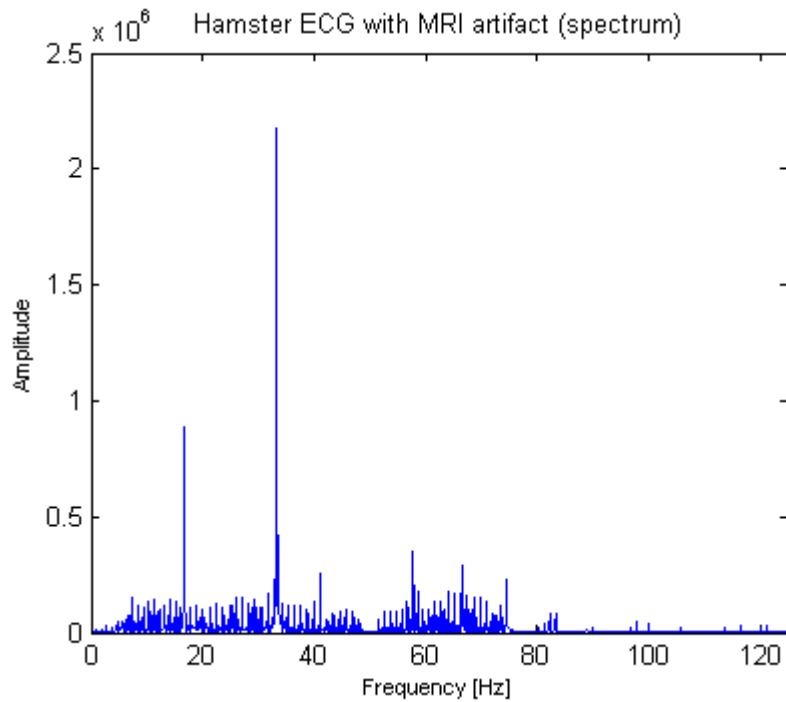
Figure 5.11: Spectrum of a hamster ECG acquired within the MRI bore during a spin echo sequence. The high frequency components were attenuated by the small animal band-pass filter.

5.2.3 *Discussion*

At present, the interactions between the ECG system and the MRI machine impair both ECG signal acquisition and MRI image creation.

MRI image creation

The comparison of MRI images acquired of the phantom and of the hamster, both in presence and absence of ECG electrodes, shows a degradation in the quality of the MRI images when the electrodes are present. The images suggest that this effect is due to radio-frequency interference caused by the presence of the ECG electrodes in the MRI bore, which probably act as antennas in guiding interference into the MRI acquisition volume. However, the degradation is much more significant in the case of hamster measurements, where it yields unusable images, whereas for phantom measurements, it merely causes grainy artifacts but image quality remains acceptable. This implies that the body of the animal amplifies the radio-frequency interference guided into the MRI volume by the electrodes.

Nonetheless, in the case of phantom measurements, the ECG electrodes were connected together with a contact impedance (see Section 5.2.1), while in the case of hamster measurements, they were connected to the limbs of the animal (see Section 5.2.2). In order to

evaluate the role of the animal body in the amplification of the interference, phantom measurements should be carried out in a similar configuration, ensuring that the electrodes are physically separated. Since the phantom is made of plastic and is thus probably an insulator, appropriate contact impedances must be ensured between the different electrodes. Due to the limited amount of time available after the first animal MRI experiments, this measurement has not been carried out yet.

Colleagues with MRI-related experience suggested adding 1 nF filtering capacitors between all ECG electrode lines and the common ground to reduce radio-frequency interference, but our measurements carried out with the phantom showed no quantifiable improvement nevertheless.

ECG acquisition

A tentative simulation showed that by applying a band-pass filter with sufficient attenuation in the higher stopband to the ECG signal corrupted by interference from the scout sequence, the remaining low-frequency components (see Figure 5.10) can be used to produce an almost usable QRS metric. This implies that in the case of this sequence, further tweaking of the already implemented digital filtering could probably be used to satisfactorily eliminate MRI interference from the ECG signal.

However, several other sequences that were tested, e.g. spin echo, produce wide-band interference within the frequency band of interest for ECG measurement (see Figure 5.11). In these cases, the filtering used in the ECG prototype system gives little hope of eliminating MRI interference, thus new filtering techniques need to be considered. Since the stray magnetic field of the solenoidal radio-frequency excitation coil of the MRI machine can be expected to be small, and as discussed in Section 4.2.3, the inputs of the ECG AFE are equipped with sufficient analog filtering against out-of-band interference, digital filtering will probably be appropriate.

Measurements also revealed that the position of the ECG cable has a great influence on the sensitivity of the system to power line interference. Further investigations could help establishing good practices of cable layout.

SUMMARY

6.1 ACCOMPLISHMENTS

I spent the past year working at Mediso Medical Imaging Systems. During this time, based on two different development kits, I produced an ECG prototype system, which is fully functional with the firmware I designed. I demonstrated the ability of the system to acquire human and small animal ECG waveforms and detect QRS complexes.

Based on the experience gained with the prototype system, we designed the final ECG system board. The circuit board is being manufactured at the time of writing this thesis. Utilizing my work done for the prototype and a demonstration board, I have completed the development of a great part of the firmware of the future ECG system.

6.2 POSSIBILITIES OF IMPROVEMENT

The measurements carried out so far show that the system is not yet able to perform satisfactorily in the MRI environment. The interactions of the ECG system with MRI require further investigation in order to achieve MRI-compatibility.

The firmware developed for the final ECG system needs to be completed and it should be tested with the final board as soon as the latter is available.

The filters in use should be reviewed in order to eliminate the glitches encountered during the acquisition of human stress ECG.

The ECG output amplitudes need to be calibrated in order to allow voltage readout, taking into consideration the amplification of the band-pass filters.

By keeping the currently used filters for QRS detection, and implementing another set of filters which satisfactorily preserve all key features of the ECG waveform for display, a diagnostic quality ECG could probably be produced.

APPENDIX

Listing A.1: Configuration of the ADS1298R.

```c
// ADS1298R register values
Uint8     ADS1298RegVal[25] = {
#if   ECG_SAMPLERATE == 500
  0x86,   // Register CONFIG1: Hi-Res EN, data rate = 500SPS
#elif ECG_SAMPLERATE == 1000
  0x85,   // Register CONFIG1: Hi-Res EN, data rate = 1000SPS
#elif ECG_SAMPLERATE == 2000
  0x84,   // Register CONFIG1: Hi-Res EN, data rate = 2000SPS
#elif ECG_SAMPLERATE == 4000
  0x83,   // Register CONFIG1: Hi-Res EN, data rate = 4000SPS
#else
  #error Sample rate not supported!
#endif
  0x10,   // Register CONFIG2: variable WCT chopping frequency
  0xDD,   // Register CONFIG3: RLD buffer EN; Vref = 2.4V
  0x03,   // Register LOFF: DC current source Loff detection
  0x00,   // Register CH1SET: normal operation, PGA gain = 6
  0x00,   // Register CH2SET: id.
  0x00,   // Register CH3SET: id.
  0x00,   // Register CH4SET: id.
  0x00,   // Register CH5SET: id.
  0x00,   // Register CH6SET: id.
  0x00,   // Register CH7SET: id.
  0x00,   // Register CH8SET: id.
  0x06,   // Register RLD_SENSP: channels 2 and 3 (LA, LL) in RLD
  0x02,   // Register RLD_SENSN: channel 2 (RA) in RLD
  0xFE,   // Register LOFF_SENSP: Loff EN on all chans but IN1P
  0x02,   // Register LOFF_SENSN: Loff EN on IN2N (RA)
  0x00,   // Register LOFF_FLIP: Loff curr direction: default
  0x00,   // Register LOFF_STATP (read only)
  0x00,   // Register LOFF_STATN (read only)
  0x0F,   // Register GPIO: all GPIOs set to input (unused)
  0x00,   // Register PACE: PACE detect buffer off
  0xF0,   // Register RESP: respiration switched off
  0x22,   // Register CONFIG4: Loff comparators powered up
  0x0A,   // Register WCT1: IN2P (LA) -> WCTA
  0xE3    // Register WCT2: IN3P (LL) -> WCTB, IN2N (RA) -> WCTC
};
```

ECG

Root Sheet — Fő Lap
EKG (Humán és Kisállat)

ECG for Human & Small-Animal

ECG_Inputs1

ECG Interface 1

ELEC_LA1	ELEC_LA
ELEC_RA1	ELEC_RA
ECG_LA1	LA
ECG_RA1	RA
ECG_LL1	LL
ECG_RL1	RL
ECG_C1	C

DIN — MOSI
SCLK — SCLK
CS — CS
DOUT_O — MISO
DRDY_O — DRDY
CLK — CLK
RESET — RESET
START — START

ECG_Inputs2

ECG Interface 2

ELEC_LA2	ELEC_LA
ELEC_RA2	ELEC_RA
ECG_LA2	LA
ECG_RA2	RA
ECG_LL2	LL
ECG_RL2	RL
ECG_C2	C

DIN — MOSI2
SCLK — SCLK2
CS — CS2
DOUT_O — MISO2
DRDY_O — DRDY2
CLK — CLK2
RESET — RESET2
START — START2

Isolators

MOSI — ECG1_MOSI — ECG1_MOSI
SCLK — ECG1_SCLK — ECG1_SCLK
CS — ECG1_CS — ECG1_CS
MISO — ECG1_MISO — ECG1_MISO
DRDY — ECG1_DRDY — ECG1_DRDY
RESET — ECG1_RESET — ECG1_RESET
START — ECG1_START — ECG1_START

MOSI2 — ECG2_MOSI — ECG2_MOSI
SCLK2 — ECG2_SCLK — ECG2_SCLK
CS2 — ECG2_CS — ECG2_CS
MISO2 — ECG2_MISO — ECG2_MISO
DRDY2 — ECG2_DRDY — ECG2_DRDY
RESET2 — ECG2_RESET — ECG2_RESET
START2 — ECG2_START — ECG2_START

Pressure Sensors

UP1 — UP1 — UP1
UP2 — UP2 — UP2

VCC_MEAS — VCC_MEAS
3V3_MEAS — 3V3_MEAS

Power_Supplies

VCC_MEAS — VCC_MEAS
3V3_MEAS — 3V3_MEAS

DSP

ECG1_MOSI
ECG1_SCLK
ECG1_CS
ECG1_MISO
ECG1_DRDY
ECG1_RESET
ECG1_START

ECG2_MOSI
ECG2_SCLK
ECG2_CS
ECG2_MISO
ECG2_DRDY
ECG2_RESET
ECG2_START

UP1
UP2
VCC_MEAS
3V3_MEAS

Ethernet Interface

ETH_MDC — ETH_MDC
ETH_MDIO — ETH_MDIO
ETH_TXEN — ETH_TXEN
ETH_TXD[3:0] — ETH_TXD[3:0]
ETH_TXCLK — ETH_TXCLK
ETH_COL — ETH_COL
ETH_CRS — ETH_CRS
ETH_RXD[3:0] — ETH_RXD[3:0]
ETH_RXER — ETH_RXER
ETH_RXCLK — ETH_RXCLK
ETH_RXDV — ETH_RXDV
ETH_INTR — ETH_INTR
ETH_RST — ETH_RST

USB interface

USB_DP — USB_DP
USB_DM — USB_DM
USB_VBUS — USB_VBUS

SA Bed CAN Interfaces

CANTX — CANTX
CANRX — CANRX

Main CAN Interface

MCANTX — MCANTX
MCANRX — MCANRX

Feedback_LEDs

Bed_LED_R1 — Bed_LED_R1
Bed_LED_Y1 — Bed_LED_Y1
Bed_LED_G1 — Bed_LED_G1
ECG_LED1 — ECG_LED1
Bed_LED_R2 — Bed_LED_R2
Bed_LED_Y2 — Bed_LED_Y2
Bed_LED_G2 — Bed_LED_G2
ECG_LED2 — ECG_LED2

ECG

Schematic component labels:

ELEC_LA1, ELEC_RA1, ECG_LA1, ECG_RA1, ECG_LL1, ECG_RL1, ECG_C1

J1 — ELEC_LA1 — R45 22K — C56 47pF — DS1 MC08010000 — R55 10K — C55 47pF — ECG_LA1 — GNDISO1

J2 — ELEC_RA1 — R46 22K — C57 47pF — DS2 MC08010000 — R56 10K — C58 47pF — ECG_RA1 — GNDISO1

J3 — ELEC_LL1 — R48 22K — C61 47pF — DS4 MC08010000 — R58 10K — C62 47pF — ECG_LL1 — GNDISO1

J4 — ELEC_RL1 — R47 22K — C60 47pF — DS3 MC08010000 — R57 10K — C59 47pF — ECG_RL1 — GNDISO1

J5 — ELEC_C1 — R49 22K — C63 47pF — DS5 MC08010000 — R59 10K — C64 47pF — ECG_C1 — GNDISO1

ECG

Power supply (TPS79601DCQR / TEN5-2411WI)

U20 TPS79601DCQR
EN IN OUT PG

VCC
C183 100nF
GND

R154 30K
GND

R139 51K
3V3

C224 15pF

C182 100nF
C158 10uF
C157 10uF
GND

L1 BLM21PG221SN1D
A3V3

C159 10uF
C160 10uF
GND

VCC_MEAS / 3V3_MEAS

VCC
R162 13K
VCC_MEAS
R160 7.5K
VCC_MEAS
C187 100nF
GND
GND

3V3
R161 13K
3V3_MEAS
R159 7.5K
3V3_MEAS
C188 100nF
GND
GND

TEN5-2411WI

U23 TEN5-2411WI
+VIN 22
+VIN2 23
R_ON/OFF 1
-VIN 2
-VIN2 3
+VOUT 14
-VOUT 16

C228 10nF
C227 10nF

VCC
R145 100
C120 10uF
C185 100nF
GND

24V
C226 470nF
C237 22uF
GND24V

L19 2x1mH

C225 470nF

D1 1N4007

VIN
J26

GND_VIN
J36

CAN Powers

U22 TME0505S
VOUT 4
GND2 3
VS 2
GND1 1

VCCISO
C236 22uF
GNDISO
C186 100nF
GNDISO
GNDISO

L7 BLM21AG102SN1D
VCC
C232 1uF
C231 1uF
GND
GND

U21 TME0505S
VOUT 4
GND2 3
VS 2
GND1 1

VCCISOM
C235 22uF
GNDISOM
C184 100nF
GNDISOM
GNDISOM

L6 BLM21AG102SN1D
VCC
C229 1uF
C230 1uF
GND
GND

GNDISO1 R146 100 C190 100nF 3V3ISO1
C122 10uF
GNDISO1
L8
C214 15pF
R155 30K R140 51K C151 BLM21AG102SN1D 1uF
GNDISO1
GND=GNDISO1
PG 5
IN OUT 6
BYP
U14 EN
TPS79201
C220 10nF
GNDISO1

GNDISO1 R148 100 C192 100nF A3V3ISO1
C125 10uF
GNDISO1
L10
C215 15pF
R156 30K R141 51K C152 BLM21AG102SN1D 1uF
GNDISO1
GND=GNDISO1
PG 5
IN OUT 6
BYP
U15 EN
TPS79201
C221 10nF
GNDISO1

GNDISO1 R147 100 C191 100nF
C123 10uF
GNDISO1
C255 100nF GNDISO1 C193 100nF GNDISO1
L9
BLM21AG102SN1D

R149 100 C194 100nF
C124 10uF
GNDISO1
L11
BLM21AG102SN1D

VCCISO1
12
13
10
15
+Vout1
+Vout2
-Vout1
-Vout2
GNDISO1

U18
+Vin -Vin
RV-0505S
C121 10uF
GND
L2
BLM21PG221SN1D
VCC
C189 100nF
GND

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M
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ECG for Human & Small-Animal
EKG (Humán és Kisállat)
Power Supplies
Tápegységek
ECG
Hajdu Cs., Török S.
Czeller Miklós
2012-04-11

GNDISO2

R150 100
C196 100nF
3V3ISO2
C127 10uF
GNDISO2

R157 30K
R142 51K
C216 15pF
L12
C153 1uF
BLM21AG102SN1D
GNDISO2

GNDISO2

GND=GNDISO2
U16
IN   PG
      OUT
EN   BYP
TPS79201
C222 10nF
GNDISO2

GNDISO2
R151 100
C197 100nF
C128 10uF
GNDISO2
L15
BLM21AG102SN1D
C256 100nF
GNDISO2

GNDISO2
R152 100
C198 100nF
A3V3ISO2
C129 10uF
GNDISO2

GNDISO2
R158 30K
R143 51K
C217 15pF
L13
C154 1uF
BLM21AG102SN1D
GNDISO2

GND=GNDISO2
U17
IN   PG
      OUT
EN   BYP
TPS79201
C223 10nF
GNDISO2

GNDISO2
R153 100
C200 100nF
C199 100nF
GNDISO2
C130 10uF
GNDISO2
L14
BLM21AG102SN1D

VCCISO2
GNDISO2

+Vout1 +Vout2
-Vout1 -Vout2

U19
+Vin   -Vin
RV-0505S
C126 10uF
GND

L3
BLM21PG221SN1D
VCC
C195 100nF
GND

ECG

ECG for Human & Small-Animal
EKG (Humán és Kisállat)

Power Supplies
Tápegységek

Designer «Tervező»: Hajdu Cs., Török S.
Controller «Ellenőr»: Czeller Miklós
Date «Dátum»: 2012-04-11
Date «Dátum»: <DATE>
NME «Név»:
Title «Cím»:

Drawing number «Rajzszám»: EL-23-50-00
Doc. Rev.: A1

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DSP_Core

ETHMDC — ETH_MDC
ETH_MDIO — ETH_MDIO
ETHTXEN — ETH_TXEN
ETHTXD3 — R166 49.9 — ETH_TXD[3:0]
ETHTXD2 — R167 49.9 — ETH_TXD3
ETHTXD1 — R168 49.9 — ETH_TXD2
ETHTXD0 — R169 49.9 — ETH_TXD1
ETH_TXD0
ETH_TXCLK — ETH_TXCLK
ETH_COL — ETH_COL
ETH_CRS — ETH_CRS
ETH_RXD[3:0] — ETH_RXD[3:0]
ETH_RXER — ETH_RXER
ETH_RXCLK — ETH_RXCLK
ETH_RXDV — ETH_RXDV
ETH_INTR — ETH_INTR
ETHRST — ETH_RST
USBVBUS — R165 49.9 — USBVBUS
USB_DP — USB_DP
USB_DM — USB_DM
CAN_TX — R97 100 — CANTX
CANRX — CANRX
MCAN_TX — R98 100 — MCANTX
MCANRX — MCANRX

R163 49.9 ETH_MDC
R164 49.9 ETH_TXEN

R225 100K USB_VBUS

Bed_LED_R1 — Bed_LED_R1
Bed_LED_Y1 — Bed_LED_Y1
Bed_LED_G1 — Bed_LED_G1
ECG_LED1 — ECG_LED1
Bed_LED_R2 — Bed_LED_R2
Bed_LED_Y2 — Bed_LED_Y2
Bed_LED_G2 — Bed_LED_G2
ECG_LED2 — ECG_LED2

ECGMOSI — R87 100 — ECG1_MOSI
ECGSCLK — R88 100 — ECG1_SCLK
ECG1CS — R89 100 — ECG1_CS
ECG_MISO — ECG_MISO
ECG1_DRDY — ECG1_DRDY
ECG1RESET — R91 100 — ECG1_RESET
ECG1START — R90 100 — ECG1_START
R96 100 — ECG2_MOSI
R95 100 — ECG2_SCLK
ECG2CS — R94 100 — ECG2_CS
ECG2_DRDY — ECG2_DRDY
ECG2RESET — R93 100 — ECG2_RESET
ECG2START — R92 100 — ECG2_START
UP1 — UP1
UP2 — UP2
VCC_MEAS — VCC_MEAS
3V3_MEAS — 3V3_MEAS

U24 SN74LVC125AD
VCC=3V3
ECG1_CS
R224 10K 3V3
C94 100nF
GND
ECG1 MISO

U24 SN74LVC125AD
VCC=3V3
ECG2_CS
R223 10K 3V3
ECG2 MISO

U24 SN74LVC125AD
VCC=3V3
GND

U24 SN74LVC125AD
VCC=3V3
GND

T3 T4
GND GND

ECG

ECG for Human & Small-Animal
EKG (Humán és Kisállat)

DSP
DSP

Designer <Tervező>
Hajdu Cs., Török S.
Controler <Ellenőr>: Czeller Miklós
Date <Dátum>: 2012-04-11

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Doc. Rev:

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ECG

ECG for Human & Small-Animal

EKG (Humán és Kisállat)
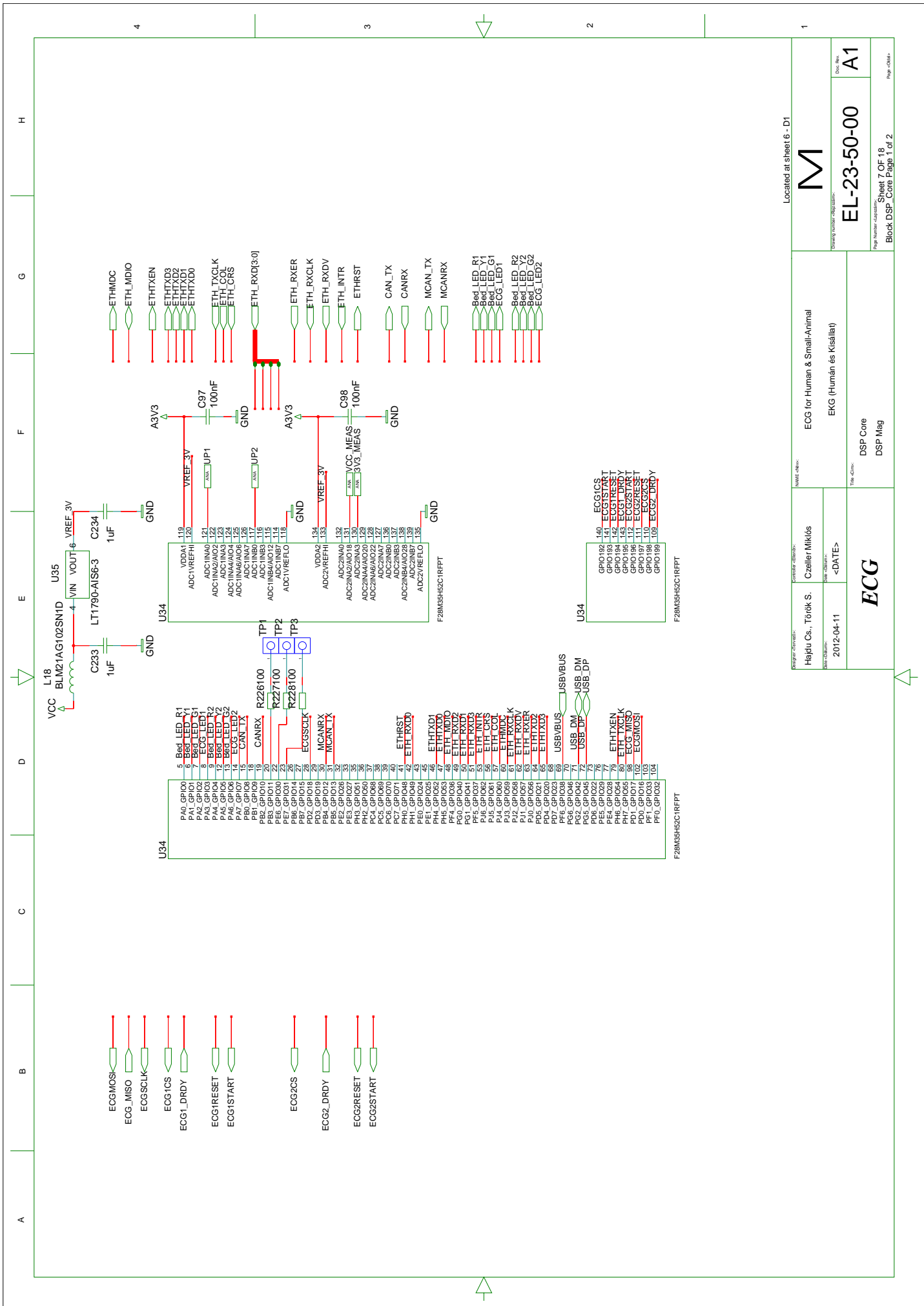
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DSP Mag

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EL-23-50-00

A1

Drawing number <Rajzszám>

Doc. Rev.

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Block DSP_Core Page 1 of 2

Page <Oldal>

NAME <Név>

Contoller <Ellenőr> Czeiler Miklós

Date <Dátum> <DATE>

Designer <Tervező>

Date <Dátum> 2012-04-11 Hajdu Cs., Török S.

Title <Cím>

M

U35 LT1790-AIS6-3

U34 F28M35H52C1RFPT

U34 F28M35H52C1RFPT

U34 F28M35H52C1RFPT

U34 F28M35H52C1RFPT

L18 BLM21AG102SN1D

VCC

VREF_3V

C233 1uF
C234 1uF
GND

ECGMOSI
ECG_MISO
ECGSCLK
ECG1CS
ECG1_DRDY
ECG1RESET
ECG1START
ECG2CS
ECG2_DRDY
ECG2RESET
ECG2START

ETHMDC
ETH_MDIO
ETHTXEN
ETHTXD3
ETHTXD2
ETHTXD1
ETHTXD0
ETH_TXCLK
ETH_COL
ETH_CRS
ETH_RXD[3:0]
ETH_RXER
ETH_RXCLK
ETH_RXDV
ETH_INTR
ETHRST
CAN_TX
CANRX
MCAN_TX
MCANRX
Bed_LED_R1
Bed_LED_Y1
Bed_LED_G1
ECG_LED1
Bed_LED_R2
Bed_LED_Y2
Bed_LED_G2
ECG_LED2

VREF_3V
A3V3
C97 100nF
GND
VUP1
VUP2
A3V3
VREF_3V
C98 100nF
VCC_MEAS
3V3_MEAS
GND
GND

TP1 TP2 TP3
R226 100
R227 100
R228 100

VDDA1
ADC1VREFHI
ADC1INA0
ADC1INA2/AIO2
ADC1INA3
ADC1INA4/AIO4
ADC1INA6/AIO6
ADC1INA7
ADC1INB0
ADC1INB3
ADC1INB4/AIO12
ADC1INB7
ADC1VREFLO
VDDA2
ADC2VREFHI
ADC2INA0
ADC2INA2/AIO18
ADC2INA3
ADC2INA4/AIO20
ADC2INA6/AIO22
ADC2INA7
ADC2INB0
ADC2INB3
ADC2INB4/AIO28
ADC2INB7
ADC2VREFLO

119 120 121 122 123 124 125 126 117 116 115 114 118 134 133 132 131 130 129 128 127 136 137 138 139 135

PA0_GPIO0
PA1_GPIO1
PA2_GPIO2
PA3_GPIO3
PA4_GPIO4
PA5_GPIO5
PA6_GPIO6
PA7_GPIO7
PB0_GPIO8
PB1_GPIO9
PB2_GPIO10
PB3_GPIO11
PB6_GPIO30
PB7_GPIO31
PB6_GPIO14
PB7_GPIO15
PD3_GPIO18
PD2_GPIO19
PB4_GPIO12
PB5_GPIO13
PE2_GPIO26
PE3_GPIO27
PH2_GPIO50
PH3_GPIO51
PC4_GPIO68
PC5_GPIO69
PC6_GPIO70
PC7_GPIO71
PH0_GPIO48
PH1_GPIO49
PE0_GPIO24
PE1_GPIO25
PH4_GPIO52
PH5_GPIO53
PF4_GPIO36
PG0_GPIO40
PG1_GPIO41
PI6_GPIO37
PJ6_GPIO62
PJ5_GPIO61
PJ4_GPIO60
PJ3_GPIO59
PJ2_GPIO58
PJ1_GPIO57
PJ0_GPIO56
PF5_GPIO37
PF6_GPIO38
PG2_GPIO42
PG5_GPIO45
PD6_GPIO22
PE5_GPIO25
PE4_GPIO28
PH6_GPIO54
PH7_GPIO55
PD1_GPIO17
PD0_GPIO16
PF1_GPIO33
PF0_GPIO32

5 6 7 8 9 12 13 14 15 18 19 20 21 22 23 26 27 28 29 30 31 32 33 35 36 37 38 39 40 41 42 43 45 46 47 48 49 50 51 53 56 57 60 61 62 63 64 65 68 69 70 71 72 73 76 77 79 80 98 102 103 104

Bed_LED_R1
Bed_LED_Y1
Bed_LED_G1
ECG_LED1
Bed_LED_R2
Bed_LED_Y2
Bed_LED_G2
ECG_LED2
CAN_TX
CANRX
ECGSCLK
MCANRX
MCAN_TX
ETHRST
ETH_RXD0
ETHTXD1
ETHTXD0
ETH_MDIO
ETH_RXD2
ETH_RXD3
ETH_INTR
ETH_CRS
ETH_COL
ETH_RXCLK
ETH_RXDV
ETH_RXER
ETHTXD2
ETHTXD3
USBVBUS
USB_DM
USB_DP
ETHTXEN
ECG_MISO
ECGMOSI

GPIO192
GPIO193
GPIO194
GPIO195
GPIO196
GPIO197
GPIO198
GPIO199

140 141 142 143 112 111 110 109

ECG1CS
ECG1START
ECG1RESET
ECG1_DRDY
ECG2START
ECG2RESET
ECG2CS
ECG2_DRDY

USBVBUS
USB_DM
USB_DP

VIN VOUT

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Doc. Rev.

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R200
2.2K
GND

3V3
R201
2.2K
C254
100nF
GND

T2
GND

T1
GND

R222
4.7K
R221
4.7K
3V3   3V3

3V3
R221
4.7K

EMU0 83
EMU1 86
TRST 85
TMS   87
TDI   88
TDO   84
TCK   89
XRSn 4
ARSn 144
VSSOSC 94
X1   93
X2   95
XCLKIN 97
XCLKOUT 82
BOOT3 81
BOOT2 52
BOOT1 78
BOOT0
NC   91
FLT1 16
FLT2 21

U34
F28M35H52C1RFPT

C219
15pF
GND

X2
20Mhz
X1
C218
15pF
GND

3V3 3V3 3V3 3V3 3V3
C109 100nF C111 100nF
GND GND
3V3 3V3 3V3
C110 100nF
GND GND GND GND GND
3V3 3V3
C112 100nF
C113 100nF
GND GND GND GND
3V3 3V3
C115 100nF
C114 100nF
GND GND GND GND
3V3 3V3
C105 100nF
GND GND GND GND
3V3 3V3
C106 100nF
C108 100nF
GND GND GND GND
3V3 3V3
C107 100nF
GND GND GND GND
3V3 3V3
C104 100nF
C103 100nF
GND GND GND GND
3V3 3V3
C101 100nF
GND GND GND GND
3V3 3V3
C102 100nF
GND GND GND GND
3V3 3V3
C100 100nF
C99 100nF
GND GND GND GND

TRST
EMU1
GND

J33
JTAG
HEADER 7X2

TMS
TDI
TDO
TCK
EMU0
3V3

C250 470nF
GND
C251 470nF
C249 470nF
C248 470nF
GND
C247 470nF
C246 470nF
C245 GND
C244
GND
C240
C241
2.2uF
2.2uF
GND
3V3

VREG12EN 101
VREG18EN 113
VDD12_11 11
VDD12_24 24
VDD12_55 55
VDD12_58 58
VDD12_66 66
VDD12_75 75
VDD12_90 90
VDD12_99 99
VDD18_1 1
VDD18_108 108
VDDIO_2 2
VDDIO_3 3
VDDIO_10 10
VDDIO_17 17
VDDIO_25 25
VDDIO_34 34
VDDIO_44 44
VDDIO_54 54
VDDIO_59 59
VDDIO_67 67
VDDIO_74 74
VDDIO_92 92
VDDIO_96 96
VDDIO_100 100
VDDIO_105 105
VDDIO_106 106
VDDIO_107 107
VSS 145
GND

U34
F28M35H52C1RFPT

VCC

L16
BLM21AG102SN1D

C238
2.2uF

C201
100nF
GND

GND

R99 100

UP1

C204
100nF
GND

R186
20K
GND

R188
10K

OUT 3

U25
SDP1108-R

dP
VCC 1
GND 2
GND

UP1

VCC

L17
BLM21AG102SN1D

C239
2.2uF

C202
100nF
GND

GND

R100 100

UP2

C203
100nF
GND

R185
20K
GND

R187
10K

OUT 3

U26
SDP1108-R

dP
VCC 1
GND 2
GND

UP2

ECG

ECG for Human & Small-Animal
EKG (Humán és Kisállat)

NAME <Név>:

Controller <Ellenőr>:
Czeller Miklós
Date <Dátum>:
<DATE>

Designer <Tervező>:
Hajdu Cs., Török S.
Date <Dátum>:
2012-04-11

Title <Cím>:
Pressure Sensors
Nyomás érzékelők

Drawing number <Rajzszám>:
EL-23-50-00

Doc. Rev.:
A1

Page Number <Lapszám>: Sheet 9 OF 18
Block Pressure_sensor Page 1 of 1

Page <Oldal>:

ELEC_LA2 → ELEC_LA2
ELEC_RA2 → ELEC_RA2
ECG_LA2 → ECG_LA2
ECG_RA2 → ECG_RA2
ECG_LL2 → ECG_LL2
ECG_RL2 → ECG_RL2
ECG_C2 → ECG_C2

J6 ELEC_LA2 R54 22K R64 10K DS10 MC08010000 C74 47pF ECG_LA2 GNDISO2 C73 47pF GNDISO2 GNDISO2
J7 ELEC_RA2 R53 22K R63 10K DS9 MC08010000 C71 47pF ECG_RA2 GNDISO2 C72 47pF GNDISO2 GNDISO2
J8 ELEC_LL2 R51 22K R61 10K DS7 MC08010000 C67 47pF ECG_LL2 GNDISO2 C68 47pF GNDISO2 GNDISO2
J9 ELEC_RL2 R52 22K R62 10K DS8 MC08010000 C70 47pF ECG_RL2 GNDISO2 C69 47pF GNDISO2 GNDISO2
J10 ELEC_C2   R50 22K R60 10K DS6 MC08010000 C65 47pF ECG_C2   GNDISO2 C66 47pF GNDISO2 GNDISO2

SA Bed 1 connector

SA Bed 2 connector

SA Bed 1 connector

DSUB9 R FEMALE

J34

DSUB9 R FEMALE

J35

24VPWR1
CANL
CANH
GNDISO
GND GNDISO
GND_VIN

24VPWR2
CANL
CANH
GNDISO
GND GNDISO
GND_VIN

T5
GNDISO

RT1
MINISMDC020F

RT2
MINISMDC020F

VCC

VCC

VCCISO

R208
5.1K

CANTSW1

R207
5.1K

VCCISO

CANTSW2

SD2
NUP2105L

GNDISO

U29
PS7241

VCCISO

R210
100

R209
360

Q1
2N3904

GNDISO

SD1
BAT754C

R144
51K

GNDISO

CANH
CANL

VCCISO

C95
100nF

GNDISO

3V3

C252
100nF

GND

U30
ISO1050

TXD   CANH
RXD   CANL

CANTX

CANRX

R101
100

VCC1=3V3 GND1=GND VCC2=VCCISO GND2=GNDISO

F1
FUSEHOLDER

24VPWR1

F2
FUSEHOLDER

24VPWR2

VIN

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Page <Oldal>

NAME <Név>

ECG for Human & Small-Animal

EKG (Humán és Kisállat)

Title <Cím>

SA Bed CAN Interfaces

Kisállat ágy CAN interfészek

Designer <Tervező>

Hajdu Cs., Török S.

Date <Dátum>

Controller <Ellenőr>

Czeller Miklós

Date <Dátum>

<DATE>

2012-04-11

ECG

J31

USB_VBUS
USB_DM
USB_DP

USB_VBUS   VBUS   1
USB_DM   DATA-M 2
USB_DP   DATA-P 3

USB-B 90R
SHLD=GND

VCC

D2   PRTR5V0U2AX

GND

M

Doc. Rev.

EL-23-50-00   A1

NAME <Név>

ECG for Human & Small-Animal

EKG (Humán és Kisállat)

Title <Cím>

USB Interface

USB Interfész

Controlled <Ellenőr.>

Czeller Miklós

Date <Dátum>

<DATE>

ECG

Designer <Tervező>

Hajdu Cs., Török S.

Date <Dátum>

2012-04-11

Drawing number <Rajzszám>

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Block USB_Intf Page 1 of 1

Page <Oldal>

VDD1=3V3 GND1=GND VDD2=3V3ISO1 GND2=GNDISO1

3V3ISO1

3V3

C77 100nF
3V3ISO1
GNDISO1

U5 ADuM2401BRWZ

R69 100 CLK CLK
R65 100 SCLK SCLK
R67 100 CS CS
MISO MISO

ECG1_CLK
ECG1_SCLK
ECG1_CS
R68 49.9 ECG1_MISO

3V3
C75 100nF
GND

VDD1=3V3 GND1=GND VDD2=3V3ISO1 GND2=GNDISO1

3V3ISO1

3V3

C76 100nF
3V3ISO1
GNDISO1

U6 ADuM2401BRWZ

R71 100 RESET RESET
R66 100 MOSI MOSI
R70 100 START START
DRDY DRDY

ECG1_RESET
ECG1_MOSI
ECG1_START
R72 49.9 ECG1_DRDY

3V3
C78 100nF
GND

VDD1=3V3 GND1=GND VDD2=3V3ISO2 GND2=GNDISO2

3V3ISO2

3V3

C82 100nF
3V3ISO2
GNDISO2

U7 ADuM2401BRWZ

R80 100 CLK2 CLK2
R73 100 SCLK2 SCLK2
R75 100 CS2 CS2
MISO2 MISO2

ECG2_CLK
ECG2_SCLK
ECG2_CS
R76 49.9 ECG2_MISO

3V3
C79 100nF
GND

VDD1=3V3 GND1=GND VDD2=3V3ISO2 GND2=GNDISO2

3V3ISO2

3V3

C81 100nF
3V3ISO2
GNDISO2

U8 ADuM2401BRWZ

R78 100 RESET2 RESET2
R74 100 MOSI2 MOSI2
R79 100 START2 START2
DRDY2 DRDY2

ECG2_RESET
ECG2_MOSI
ECG2_START
R77 49.9 ECG2_DRDY

3V3
C80 100nF
GND

VE1 VIA VE2 VOA VIB VOB VIC VOC VOD VID

R81 75
R82 75

U9
OE 1
OUT 4
GND
2.048MHz 3.3V
V33=3V3
3V3
C83 100nF
GND

Located at sheet 1 - C3

ECG for Human & Small-Animal
EKG (Humán és Kisállat)

Isolators
Galvanikus leválasztás

Czeller Miklós
Hajdu Cs., Török S.
2012-04-11
<DATE>

ECG

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Block Isolators Page 1 of 1

VCC1=3V3 GND1=GND VCC2=VCCISOM GND2=GNDISOM

CANTX

CANRX

U31

TXD CANH

RXD CANL

ISO1050

R102 100

3V3

C253 100nF

GND

VCCISOM

C96 100nF

GNDISOM

CANH

CANL

SD3 NUP2105L

GNDISOM

T6

GNDISOM

120

J32

DSUB9 R MALE

GNDISOM

M

ECG for Human & Small-Animal

EKG (Humán és Kisállat)

NAME <Név>

Controle <Elenőr.> Czeller Miklós

Designer <Tervező.>
Hajdu Cs., Török S.

Date <Dátum.>
2012-04-11

Date <Dátum.> <DATE>

Title <Címe.>

Main CAN Interface

Fő CAN interfész

ECG

Drawing number <Rajzszám.>

EL-23-50-00

Doc. Rev.
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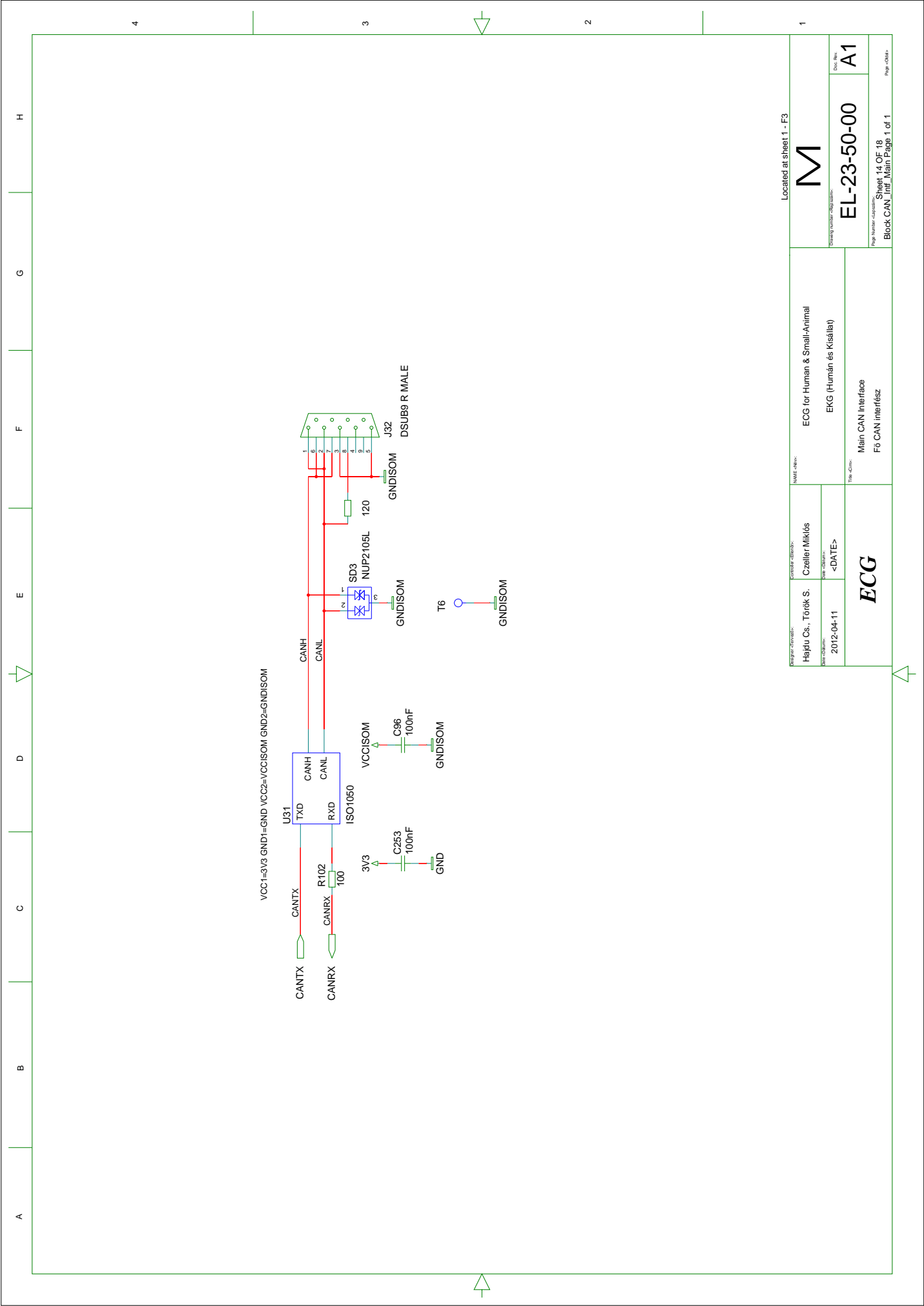
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Page <Oldal>

ECG

U28
J0011D01BNL

TXP
TXCT
TXN
RXP
RXCT
RXN
C_YLW
A_YLW
C_GRN
A_GRN
N.C.
CHS_GND3
CHS_GND2
CHS_GND1

1
4
2
3
5
6
11
10
9
7
8
13
14
12

R205 330
R206 330
GND
GND

3V3 R204 10
3V3 R203 10
C206 100nF
C205 100nF
GND
GND

R182 49.9
R181 49.9
R180 49.9
R179 49.9
ENET_VDDA

ENET_TXP
ENET_TXN
ENET_RXP
ENET_RXN

R190 10K
GND

C242 33pF
C243 33pF
GND
X1 25MHz X32
R202 12K
GND

U27
LAN8710A

VDDA2
VDD1A
VDDIO
VDDCR
MDC
MDIO
TXEN
TXD3
TXD2
TXD1
TXD0
TXCLK
COL/CRS_DV/MODE2
CRS
RXD3/PHYAD2
RXD2/RMIISEL
RXD1/MODE1
RXD0/MODE0
RXER/RXD4/PHYAD0
RXCLK/PHYAD1
RXDV
nINT/TXER/TXD4
nRST
XTAL1/CLKIN
XTAL2
RBIAS
VSS
LED2/nINTSEL
LED1/REGOFF

27
1
12
6
17
16
21
25
24
23
22
20
15
14
8
9
10
11
13
7
26
18
19
5
4
32
33
2
3

GND

R183 75
R178 75
R177 75
R173 75
R172 75
R171 75
R170 75
R174 75
R175 75
R176 75
R184 75
R191 2.2K
3V3

ENET_VDDA
ENET_VDDIO
ENET_VDDCR

L4 BLM21PG221SN1D
L5 BLM21PG221SN1D
3V3
3V3

C162 10uF
C208 100nF
C209 100nF
C210 100nF
C207 100nF
C161 10uF
C211 100nF
C163 10uF
GND

POWER
GROUND

ETH_MDC
ETH_MDIO
ETH_TXEN
ETH_TXD[3:0]
ETH_TXCLK
ETH_COL
ETH_CRS
ETH_RXD[3:0]
ETH_RXER
ETH_RXCLK
ETH_RXDV
ETH_INTR
ETH_RST

3V3 R189 10K
3V3 R192 2.2K

3V3 R199 2.2K ETH_RXER
R198 2.2K
GND
R197 2.2K ETH_RXCLK
GND
R196 2.2K ETH_RXD3
GND
ETH_RXD2
GND

3V3 R195 2.2K ETH_RXD0
3V3 R194 2.2K ETH_RXD1
3V3 R193 2.2K ETH_COL
GND

DIN
SCLK
CS
START
CLK
RESET

DRDY_O
DOUT_O
RL

R83 100
R84 100
DRDY DRDY_O
DOUT DOUT_O
RLDOUT

U10 ADS1298RIZXG

B8 DRDY
D6 DOUT
E8 DIN
H8 DIN
F8 SCLK
F7 CS
G7 START
G6 RESET
G8 CLK

CLKSEL
DRDY
DOUT
DIN
SCLK
CS
START
GPIO1
GPIO2
GPIO3
GPIO4
RESET
CLK

F5 GPIO1
E7 GPIO2
E6 GPIO3
E5 GPIO4

PWDN
VREFN
VREFP
VREF
DAISY_IN
DGND3
DGND2
DGND1
AVSS_5
AVSS_4
AVSS_3
AVSS_2
AVSS_1
VCAP2
VCAP1
TESTN PACE_OUT2
TESTP PACE_OUT1

VREFN
VREFP

C86 100nF
C155 10uF
C133 1uF
C131 1uF
C132 1uF
C116 10uF
C117 10uF
GNDISO1

VCAP1
VCAP4
VCAP3

WCT
IN1N
IN1P
IN2N
IN2P
IN3N
IN3P
IN4N
IN4P
IN5N
IN5P
IN6N
IN6P
IN7N
IN7P
IN8N
IN8P
RESP_MODN
RESP_MODP
RESV1

RA
LA
LL
C

DVDD2 D8
DVDD1 C8
AVDD5 A7
AVDD4 C6
AVDD3 B6
AVDD2 A6
AVDD1 B4
VREF A4
RLDREF C4
RLDINV A3
RLDIN B3
RLDOUT B2

C134 1uF
C84 100nF
C135 1uF
C85 100nF

GNDISO1 GNDISO1 GNDISO1
A3V3ISO1
3V3ISO1

C164 1nF
WCT
IN1N
IN1P

R103 10K
PWDN
3V3ISO1

R106 10K
R107 10K
R108 10K
R109 10K
R110 10K
R111 10K
R112 10K
R113 10K
IN5N IN5P IN6N IN6P IN7N IN7P IN8N IN8P
A3V3ISO1

RLDINV RLDOUT RLDIN
R137 1MEG
C180 1.5nF

RESP_MODP
RESP_MODN

IN1N
R129 10MEG
C169
R130 10MEG C125 40.2K
C168 2.2nF GNDISO1
C176 100nF
ELEC_LA
A3V3ISO1

IN1P
R132 10MEG
C171
R131 10MEG C126 40.2K
C170 2.2nF GNDISO1
C177 100nF
ELEC_RA
A3V3ISO1

GNDISO1

ECG

Feedback LEDs
Visszajelző LED-ek

ECG for Human & Small-Animal
EKG (Humán és Kisállat)

Located at sheet 1 - F2

Drawing number <Rajzszám>
EL-23-50-00

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Block Feedback_LEDs Page 1 of 1

Designer <Tervező>
Hajdu Cs., Török S.

Controller <Ellenőr>
Czeller Miklós

Date <Dátum>
2012-04-11

Date <Dátum>
<DATE>

NÉVE <Név>

Title <Cím>

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