Polygraph implementation using Arduino microcontroller board

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Contents

1 Introduction

This paper describes the design and implementation of a polygraph, more commonly referred as a *lie detector*. The purpose of the device is to determine whether a person, the *subject*, is telling the truth when asked simple yes or no questions, e.g. "Have you seen this man before?". The polygraph measures the symphatetic nerve activity of the subject through the electrodermal response, EDR, combined with pulse and a video recording of the subjects facial movements during the test. It is believed, but not implied by this paper, that telling a deliberate lie induces a corresponding, spontaneous response from the nervous system, which, when measured, can be used to discriminate the questions where the subject is not telling the truth. The basis for measuring the spontaneous response is the EDR, which measures the sweat gland activity of the subject that is known to correlate well with symphatetic nerve activity [1]. The measurements are done using a micro-

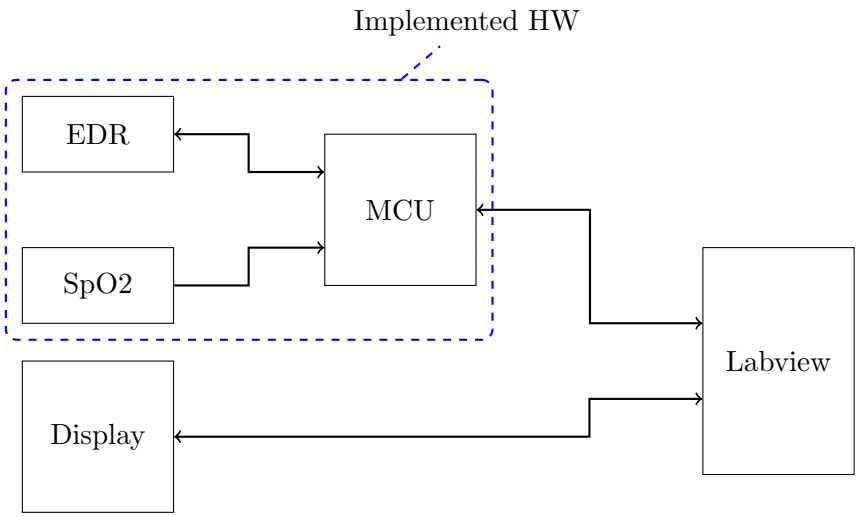

Figure 1: Block diagram of the device.

controller that converts the analog signals to digital form and, additionally, communicates with a host pc running a Labview software. The Labview softaware acts as a graphical user interface, which displays and records the gathered data. Furthermore, from the example question presented above it is obvious that a method for displaying images for the subject is required. The images must be synchronised with the questions in order to reliably combine the measured response with a question. The images are displayd to the subject using the same Labview software and an additional touch screen display placed in front of the subject. A block diagram of the device is presented in figure 1.

2 The Electrodermal Response

The skin consist of layers of alive and dead cells, called *epidermis*, *dermis*, and *hypodermis* with the epidermis being the outer, and hypodermis being the inner layer of skin. The layers protect the body from bacteria, mechanical assaults, and dehydration and, in general, have a high electrical resistance; the epidermis in particular is a strong electrical barrier. However, since the body uses water to transfer excess heat from the body the skin contains sweat glands and ducts that transfer sweat from the hypodermis to the epidermis. The sweat glands are controlled by the sympathetic nervous system, which implies two things. First, the nervous system ensures that dehydration does not happen in excess and water is excreted through the skin only during periods of elevated body temperature or stress, such as heavy excercise or during a humiliating situation. Second, sweating cannot be controlled by will; a consequence of vast importance in the theory of polygraphs. [1]

Sweat is a liquid with properties similar to a weak electrolyte. The skin therefore consists of a number of parallel tubes filled with conducting substance. When the sympathetic nervous system activates the sweat glands, sweat is pumped through the outer layers of skin, which result in a significant reduction in the skin resistance and, consequently, the EDR measurement is an impedance measurement for which several well known and tested circuit topologies exist. Like the standard impedance measurement, the EDR has a high dynamic range; a problem routinely encountered in regular multimeters. To counter this problem, multimeters use a variable dynamic range to ensure sufficient resolution even at extremely high impedances in the megaohm range. Since the described method for solving the range problem is routinely used by the industry, it is also the method chosen to measure the electrodermal response.

Since the subject is a human being care must be taken not to cause any pain or discomfort during the measurement. The impedance can be measured either through the *exosomatic* or *endosomatic* method. The first uses an external measurement current to cause a voltage drop between two skin electrodes on the subject while the other operates using the internal potentials of the body [1]. The exosomatic method, while simpler, has the potential to induce large enough currents in the body to induce sensory stimulus, i.e. pain. In contrast, the endosomatic method requires an amplifier with substantially higher input impedance than the skin resistance, but does not drive a current through the body. [1]

The skin measurement can be *tonic* measuring the background level of the impedance, or *phasic* measuring the changes in the signal filtering out any DC components. The suggested frequency range for these measurements are typical for biosignal measurements: 0 to 5Hz for the tonic, and 0.03 to 5Hz for the phasic measurements. [1] The simplified equivalent circuit of figure 2 assumes lumped parameters, which may or may not be appropriate

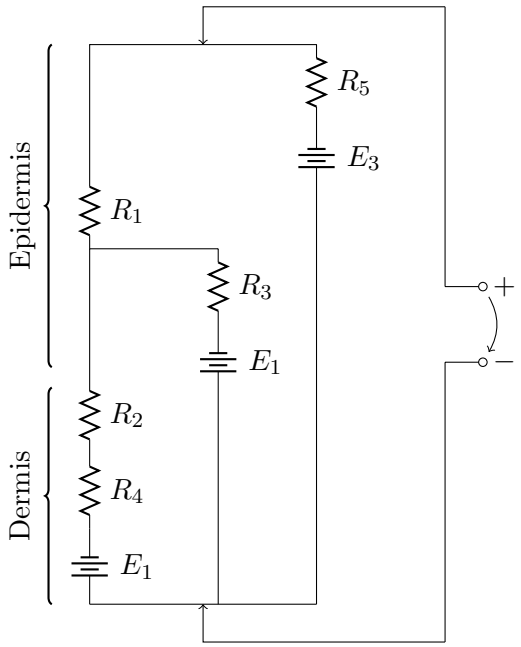

Figure 2: A simplified qualitative equivalent circuit of the skin impedance between two electrodes.

for the distributed parameters of the skin. Additionally, the model assumes that electrodes are used that provide an efficient mechanism to convert between electron current carriers inside the copper wires and the ions in the body. The electrodes are placed firmly on the skin using a NaCl solution that mimics the characteristics of sweat [1].

Skin impedances decreases nearly exponentially with increasing frequency, decreasing from 1 to 3.5MΩ at frequencies below 10 Hz to just 220 Ω at 1 MHz. However, the impedance is greatly lowered, becoming nearly independent of frequency by removing the stratum corneum, or the layers of dead cells in the epidermis, through abrasion. [2] Other studies have found that skin hydration increasingly affects the resistance of the measurement at frequencies above 100 kHz [3].

The recommended method for measuring electrodermal responses is to measure the skin conductance, SC, using a approximately 0.5 V constant voltage source and a small series resistor. The value of the series resistor should be chosen such that the applied voltage between the two skin electrodes stays constant and nearly 0.5 V. In essence, the series resistor has to be considerably smaller than the skin resistance, and a value of 500 Ω has been recommended in literature. Similar recommendations have been made as per the electrodes used in the measurement. The electrodes should be silver based, with Ag/AgCl being a common choice, and they should be located on active sites with the palmar side of hand being a particularly suitable

place. In case of skin potential measurements, the other electrode has to be placed in a indifferent location near the elbow. The electrode surface area should be approximately $1\,\mathrm{cm}^2$. [1–4]

3 Measurement Hardware

The MCU used by the device is the Arduino Uno, which is an Atmega328 based prototyping board largely used by enthusiastist and academics alike. The board contains some IO pins, 6 analog inputs with a 10-bit ADC, and a usb-serial connection to the PC. The Arduino is programmed, using an open source Arduino GUI, through the serial interface with the help of a custom bootloader precompiled into the Atmega328. The process is fairly straightforward and the board is quickly operational. The software comes with a multitude of libraries for device control and data acquisition and masks many of the tedious tasks, such as register handling and interrupts traditionally necessary for microcontroller programming. In addition to the simple IO ports, some IO pins have an alternative function, namely the hardware supported Universal Serial Interface (USI) capable of communicating with peripheral devices using SPI. In the PC end, the data is easily read into Labview, since a community maintained Arduino library for communicating with the microcontroller exists and is freely available through the VI Package Manager.

EDR

As mentioned, the EDR is an impedance measurement with relatively high dynamic range. Since the operator of the polygraph is interested in the *changes* in the signal, two possible implementations exist that solve the dynamic range problem. First, the phasic measurement can be used to disconnect any DC signals from the inputs of the amplification circuit. Second, a variable gain amplifier can be used to adjust the signal gain so that the signal is never clipped and meaningful information can be read from the waveform. The latter option can be implemented using, e.g. a digitally programmable amplifier with an optional settable biasing voltage, generated for example using a single channel DAC. The former option is easier to implement, but exceedingly large capacitor values are needed to allow the very low frequencies of the EDR to pass through the decoupling capacitor. The second option is arguably more complicated and has many more design parameters to be optimised, but the constraints on each parameter are much more relaxed.

The EDR signal is a fairly weak signal in the presense of a strong common mode voltage. Since the signal amplitude is small, the overall system gain has to be large, which requires that the common mode signal is somehow attenuated before amplification. The DC common mode signal is removed by

AC coupling the amplifier inputs, effectively removing the DC-component of the signal. To ensure that even slow, gradual changes are measured, the decoupling capacitors must be relatively large to position the -3dB frequency sufficiently low. An appropriate value for the corner frequency should be ¡ 1 Hz, and 0.3Hz was was chosen. Since the op amp inputs have negligible, but non-zero, input bias currents a DC path must be provided from the decoupled op amp inputs to ground. The easiest way to accomplish this is to place a resistor between the input and ground.

Since EDR is concerned with the changes in the electrical properties of skin, the measurement should be reasonably linear. Consequently, it is convenient to measure the inverse of skin resistance, the skin conductance g_s, which varies linearly when the sweat glands begin to conduct. To measure the skin conductance, current through the skin resistance has to be measured and to measure the current, a small sense resistor has to be placed in series with the measurement voltage and the skin conductance. This series resistance, R_{sense}, causes a small nonlinearity in the measurement, but, provided that $R_{sense} << R_{skin}$, the effect is negligible. The basic measurement setup is presented in figure 3.

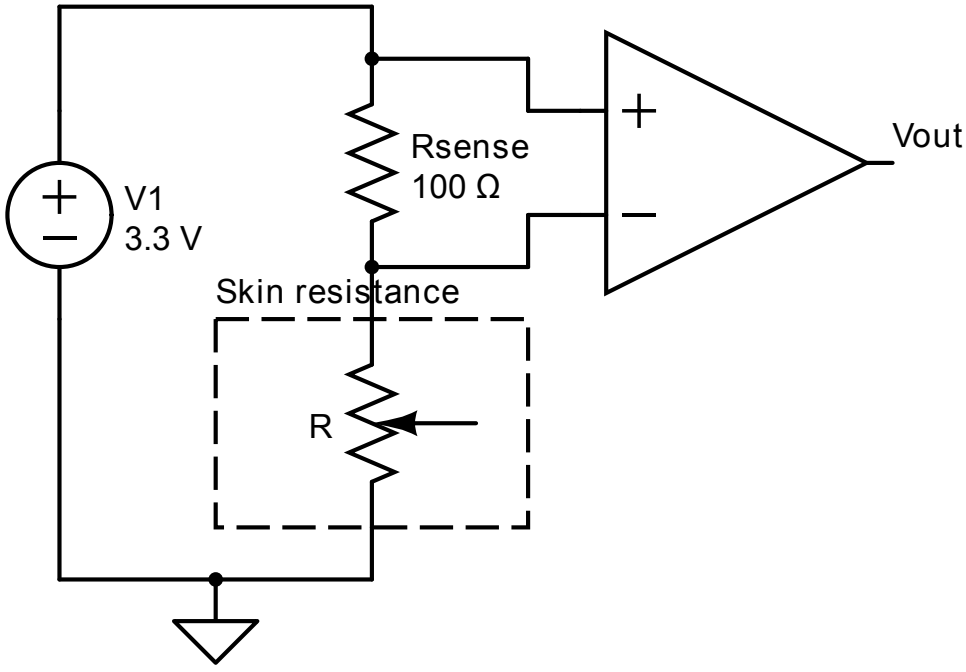

Figure 3: Conductance measurement

The output voltage of the setup presented in figure 3 is iGR_{sense}, where i is the current through sense resistance and GR_{sense} is the overall gain of

the amplifier. The current i is calculated from the excitation voltage as

$$i = \frac{V_{meas}}{R_{skin} + R_{sense}} \approx \frac{V_{meas}}{R_{skin}} = g_s V_{meas}, \tag{1}$$

which is approximately linear in g_s. The overall gain of the system determines the sensitivity of the measurement. The Atmega328 powering the Arduino contains a 10-bit ADC with 5V reference voltage, setting the minimum detectable voltage difference as $5/2^{10} \approx 49\,\mathrm{mV}$. Therefore, the necessary gain can be calculated from

$$G = \frac{49\,\mathrm{mV}}{R_{sense} V_{meas} \Delta g_s}, \tag{2}$$

where Δg_s is the smallest measrurable change in conductance. However, if the smallest measurable conductance is fixed to an absolute value, the *dynamic range* of the amplifier has to be fairly large since a $1\,\mathrm{k}\Omega$ change is roughly 10% in the 10k range but only 0.1% in the mega Ohm range. Therefore, the required sensitivity is defined as the smallest measurable relative change, and given as a percentage.

Since the total required gain is quite high, the gain block has to be divided into several blocks with reasonable gains to prevent the amplifier stage from becoming unstable. Amplifiers with very high gains might start to oscillate if, at some frequency, the gain has reduced to unity and the phase of the signal has turned 180°, i.e. inverted, resulting in positive feedback which quickly saturates the output. To prevent this, stages with lower gain are used in series to achieve the same required total gain. Since the desired signal, changes in the skin conductivity, is measured across the sensing resistor R_{sense} in the precence of a large common mode DC-voltage. As instrumentation amplifiers have, in general, much larger common mode rejection ratios than ordinary operational amplifiers, the first amplifier stage was built using the AD8236 instrumentation amplifier, which has adjustable gain from 5 to 200 using only one external resistor. The AD8236 has a reference terminal to which the output voltage is referenced. By capacitively coupling the input of the amplifier back to the reference terminal, the gain of the amplifier is reduced to unity at low frequencies while remaining high at higher frequencies, effectively AC coupling the amplifier. The AC coupled instrumentation amplifier circuit topology is presented in figure 4. The voltage Vref shifts the output of the amplifier to the middle of the available voltage range, providin Vs/2 volts of swing room for the varying EDR signal. The op amp selected as the buffer for the REF pin was chosen to be the same as recommended by the datasheet.

The second amplifier stage is also AC coupled to prevent the virtual ground at Vs/2 from being amplified thus saturating the output. Since the reference pin is buffered by a AD806, it is logical to choose the same amplifier as the second amplifier stage since the op amp comes in single, dual, and

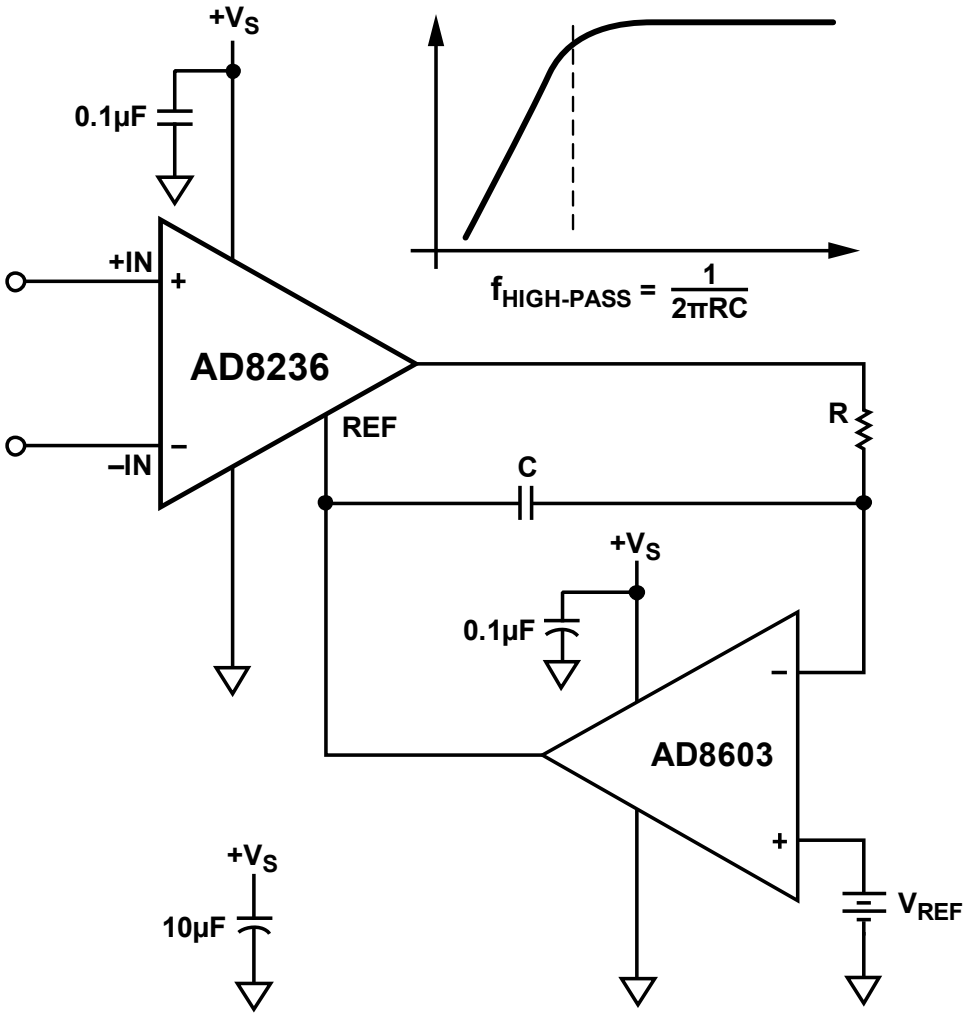

Figure 4: AC coupled AD8236. Schematic taken from the datasheet.

quad configurations, which is convenient and saves board space. The second AD806 is connected as a AC-coupled, non-inverting amplifier that is biased at Vs/2 to provide the same head room for the AC signal as the first stage. The biasing is done by simply forming a voltage divider at the positive input terminal. The voltage divider also serves as a DC return path to ground, preventing the input bias currents of the amplifier from generating a DC voltage to between the inputs. The gain of the amplifier is initially chosen as 36, but should this be inadequate, the gain can be safely increased by adjusting the feedback resistors.

The final amplifier stage is a programmable amplifier with user selectable gains from 1 to 16, and is intended for fine tuning the signal amplitude before the ADC inputs. The gain is initially set to the middle of the gain

range, i.e. 8, and can then be altered via the microcontrollers SPI interface. This configuration can then be used to either double or halve the gain if the signal amplitude is too weak or too large respectively. The amplifier initially chosen for this task was the TI LMP8100, as the supply of SPI controllable amplifiers seemed somewhat scarce, which was a fallacy since there actually exists a wide range of different programmable amplifier devices. One alterna-

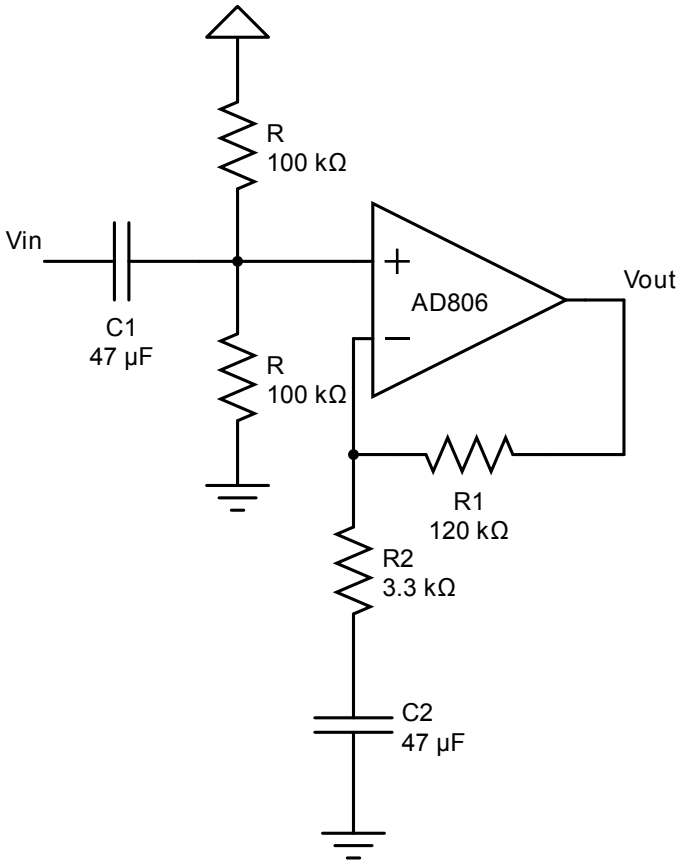

Figure 5: AC coupled non-inverting amplifier

tive would have been a regular op amp with a digitally controllable feedback resistor network. This would have had the advantage that the gains could be easily selected and adjusted, but on the other hand, the configuration would have taken much more board space than a single programmable device. The DIY option could have been implemented with the help of a analog switch and the output pins of the Arduino mini, optionally using a parallel out shift register, such as the 74HC595, to reduce the output pin usage.

Pulse

Pulse is measured using a Medlab pulse oximetry module EG00352, which connects to a SPO2 led sensor, measures the amount of light absorbed by oxygen in the blood, and calculates the oxygen saturation and pulse of the patient. The signals are continuously transmitted over a serial transmit wire operating at CMOS logic levels, i.e. from 0 to 3.3V, with a baudrate of 9600. The module requires only a well regulated 3.3V supply and ground signals, as well as connections to the SPO2 sensor leads; it is otherwise self contained.

The required 3.3V is generated from the 5V usb lead with the help of a LM317 voltage regulator. The LM317 is a three terminal positive adjustable voltage regulater, with output voltage selectable using two external resistors. The device operates by maintaining a fixed 1.25 V voltage between the OUT and GND, also commonly referred to as ADJ, terminals. The regulator circuit is presented in figure 6. The output voltage is given by

$$V_o = 1.25\left(1 + \frac{R_2}{R_1}\right) + I_{gnd}R_2, \tag{3}$$

where I_{gnd} is typically less than 100 µV and can be neglected for most applications. The equation follows from the fact that current $I_1 = \frac{1.25\,\mathrm{V}}{R_1}$ flows into the GND-terminal node, and since I_{gnd} is small, the voltage from the output terminal is $1.25V + I_1R2 = 1.25(1 + \frac{R_2}{R_1})$.

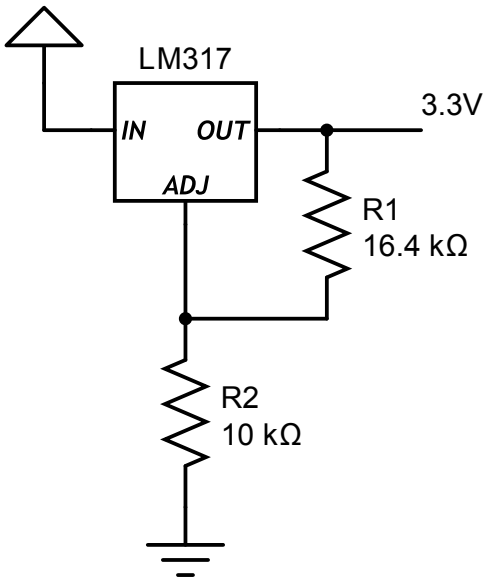

Figure 6: 3.3 V voltage regulator.

The serial data from the SPO2 module has to be translated to TTL logic, i.e. from 0 to 5V, before it can be connected to the Arduino. This is achieved

by driving the gate of an NPN MOSFET circuit presented in figure 7. The MOSFETs buffer the serial signal and cause the voltage at the serial input to swing from GND to 5V, sufficiently for the Arduino to register each bit.

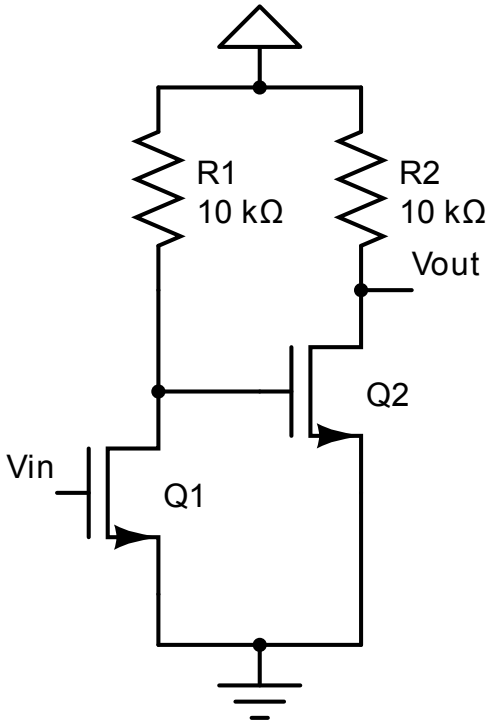

Figure 7: Logic level translation. When Vin is HIGH Q1 is conducting and the gate of Q2 is pulled close to ground, causing Vout to be pulled HIGH. When Vin is pulled low, Q1 closes and the gate of Q2 is set to Vcc thus opening the DS-channel and pulling Vout to ground.

Since the main RX/TX pins of the Arduino are required for communicating with the host computer, the SPO2 data is read using a software serial library available for Arduino. The library assigns external interrupts for each registered IO-pin and when the pin state changes the software reads the current pin state directly from the pin, essentially transforming any IO-port to a serial interface, provided that the baud rate for communications is sufficiently low. The baud rate of the Medlab module is, as mentioned, 9600, which a relatively slow communication speed and should pose no problems to the software serial implementation.

4 Host Software

The polygraph device sends data, and is controlled by the host computer running a Labview program, which displays the questions to the subjects

and records the response. The Arduino is loaded with the Labview Arduino interface firmware, which can be acquired through the VI Package Manager. The library also provides many usefull subVIs for control and data acquisition from the ADC. The program consists of an user interface and a data acquisition loop that continuously collects data from the Arduino. The UI and main loop are implemented using the Labview event structure, which can be registered to wait for an user interface event and has an inbuilt timeout event, which is fired if no UI events occur before the preset timeout value, given in milliseconds. The timeout event is practical for acquiring data, since the code is fired at roughly equal intervals, and the event structure is superior in monitoring the UI for actions, since no polling is required for any of the front panel controls.

The questions are displayd to the subject through a subVI that is launched on a separate window. The subVI takes two Labview queue references as parameters and uses these to communicate to the main application. Similarly, when the main application wishes to send data to the remote application it simply puts the data in the send queue and Labview core handles the rest. The subVI showing the questions is operated using a touch screen, and consists of a large area for displaying pictures and text and two boolean buttons for *yes* and *no* answers.

5 PCB Assembly

The PCB was designed using the freeware version of CadSoft Eagle PCB layout software. Some of the parts were directly available in the various libraries, but the majority of the footprints had to be made by hande, or acquired from external libraries. Particularly good 3rd party libraries were found from www.sparkfun.com and www.lement14.com. Since majority of components manufactured today come in surface mount packages, and some of the components were only available in SMD packages, no effort was made to search for alternative devices that do come in through hole packages. The Medlab SPO2 module and Arduino are attached to the PCB using pin headers, which allow the modules to be removed if the device ever becomes obsolete, or if another improved version is made. The SPO2 module connects to a low profile 2x4 and 2x3 (rox x col) male pin headers using 2mm pitch. The only suitable manufacturer was found to be Samtec, but the somewhat expensive headers were sold in batches of 10, making the headers one of the most expensive parts in the assembled device. Luckily, two 2x5 Samtec low profile headers were found lying around in a junk box and, although larger than required, have much better price/functionality ratio than the expensive headers of the right size. The price of each component was assessed critically since the intended use of the device is not precision resistance measurement but to give qualitative information about the subconscious reactions of a

subject being asked questions.

The freeware version of Eagle is capable of producing two layer PCBs with size restrictions, but since much of the electronics were separately packaged in the SPO2 and Arduino modules, all the components were easily placed on the top layer, while the bottom layer holds mainly the ground plane and a few signals. No particular high speed design considerations were made in the design as the signals being measured are relatively slow biosignals. EMC was also considered as a non-issue since the intended use site of the device should not contain possibilities for capacitively coupled fields in the measurement frequency range since the pass band of the device terminates before the 50 Hz power line frequency.

If this were a medical device, the patient would have to be isolated from the main power supply. Murata, for example, makes handy power isolators, but their price in small batches is unreasonably high, in tens of euros, for this application. Another possibility would have been the use of a transformer coil and a switching circuit generating the required AC signal that would induce voltage in the secondary coil. This approach however requires, in addition to the transformer coils, a switching IC, a rectifier bridge on the isolated side as well as a voltage regulator. Provided the component sizes were kept reasonable to enable easy hand soldering, this would require board space, but would improve the safety of the device. However, since the probe leads connected to the subject are placed in adjacent fingers, and since the 100 Ω sensing resistor limits the maximum *theoretical* current to 33 mA, which is not sufficient to cause skin burns, the isolation was considered, but ultimately left out of the apparatus.

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