

Design and Development of an MRI Compatible Circuit suitable for Pre-Clinical Physiological Monitoring in a 4.7T Magnetic Resonance Imaging.

ABSTRACT

As monitoring preclinical MRI is essential, it is important that the equipment used must be MRI-compatible. This is because interaction with an MRI radiofrequency (RF) transmitting field can cause RF burns due to inappropriate equipment. And non-compatible MRI monitoring equipment can give inaccurate readings in the MRI system if the equipment has ferromagnetic components. Preclinical MRI is designed for animals, the model used to study diseases related to humans. Animal models are imaged using a smaller scanner with high magnetic fields like the 4.7T. The main parameters of interest are the respiratory rate, the temperature, the pressure transducer and the electrocardiogram (ECG), this is because the animal model is used as a laboratory experiment for human disease. An MRI-compatible lead II ECG simulator, with approximately 0.5 mV output amplitude was attached to the input of the ECG preamplifier to represent a small animal's signal as shown in Figure 1. The transmitted signal is the ECG, the pressure sensor, the temperature, and the battery (voltage), these are all linked to different channels as seen in Figure 5-8. Figure 5-7 (outside, near and inside magnet) were acquired with the pressure sensor pad taped on a male adult volunteer's thumb to record the human pulse of 70 bpm, which is similar to the respiration rate of an anaesthetized mouse or rat. Then Fig.8 was acquired with the pressure sensor taped to the surface of a phantom to pick up vibration from the gradient system. The use of a signal gating generator should be explored in future work, this could make the monitoring parameters to be independent of the RF pulse influence as seen in Fig 5-8. Finally, fibre optics should be included in the design process.

Keywords: MRI, ECG, Compatible, Physiological monitor, preclinical, Temperature, pressure sensor and respiratory sensor.

INTRODUCTION

Physiological monitoring for preclinical magnetic resonant imaging (MRI), is designed to mainly meet the physiological monitoring and gating requirement for anaesthetizing small rodents such as rats and mice, and even larger animals in an MRI setting. The major benefits of the MRI is excellent in soft tissue contrast, and also has the ability to image 3-D structure and function in

high inter-session replicate [1]. Biomedical research has increased significantly through the use of non-invasive imaging technologies in the study of biological and biochemical processes within living animals and humans.

Certainly, imaging small animals are used widely as a preclinical experimental instrument in modelling both animal and human diseases as well as cancer studies, musculoskeletal disorders, cardiovascular, and neurodegenerative. Their application signified, the most important influence on the improvement in studying animal models in vivo. Using longitudinal monitoring from the beginning and the development of disease around similar animals, and studying the effectiveness of biological drug therapy. This provides a fast and efficient method for classifying and modelling.

This research work is mainly Centre on pre-clinical MRI scanners, because, the animal model used widely is based on pre-clinical scanners owing to the substantial benefits in the quality of image acquisition, sensitivity, and spatial resolution, that can be acquired using a smaller scanner with a high magnetic field like the 4.7T. As rat and mouse brains are about 3,000 times much smaller compare to the brain of humans, the pre-clinical scanner is necessary to achieve high resolution [3].

In physiological monitoring of basic parameters such as temperature, respiration, heartbeat and electrocardiogram (ECG) for small animals such as rats and mice to remain still throughout the period of the MRI examination, a general anaesthetic is required to acquire decent high-quality MRI images. As monitoring preclinical MRI is essential, it is important that the equipment used must be MRI-compatible. This is because interaction with an MRI radiofrequency (RF) transmitting field can cause RF burns due to inappropriate equipment. And non-compatible MRI monitoring equipment can give inaccurate readings in the MRI system if the equipment has ferromagnetic components. It can cause missile hazard, and can also give rise to image artefacts such as signal loss due to ferromagnetic distortion of static field or RF interference lines [4].

THEORETICAL BACKGROUND

It is widely known that physiological monitoring for preclinical MRI is achieved using a high field magnets B_0 with a very small bore specifically designed for rodents (rats and mice). The B_0 is a static magnetic field, and the significance of using the high magnetic field (B_0) is the high

resolution that is achieved due to the increase in signal, as a result of increased sample magnetization caused by a greater number of aligned spins from the Boltzmann distribution and the increased Faraday induction from the higher frequency given as;

$$fL = \left(\frac{\gamma}{2\pi}\right)B_0 \quad (1)$$

where γ is gyromagnetic, fL is the Larmor frequency and B_0 is the magnetic field.

Usually, preclinical uses a higher magnetic field B_0 range of 4.7T to 7T. For the purpose of this project magnetic field of 4.7T (200 MHz) is used [8].

Preclinical MRI is designed for animals, the model used to study diseases related to humans. Animal models are imaged using a smaller scanner with high magnetic fields like the 4.7T. Neurological situations, for diseases of the joints and muscles, for assessing tumours, and also used for showing abnormalities in the heart and blood cells. Modern clinical whole-body MRI scanner has a magnetic field strength range from 0.5T-3.0T [13].

The magnetic dipole moment μ in the classical moment experiences a torque L , that will align with the magnetic field B_0 was given as

$$L = \mu \times B_0 \quad (2)$$

As the nucleus also has angular momentum p , the nucleus precesses because of the torque around the magnetic field B_0 , relatively, similar to a gyroscope in a gravitational field. The angular momentum changing rate is given as:

$$\frac{dp}{dt} = L = \mu \times B_0 \quad (3)$$

and from the equation (2), $\mu = \gamma p$, then

$$\frac{dp}{dt} = \gamma p \times B_0 \quad (4)$$

where γ is the gyromagnetic ratio, and the Larmor frequency is given as:

$$\omega L = -\gamma B_0 \quad (5)$$

Eqn. (4) and (5), show the direction of the applied field at which the nucleus precesses, at ω rad/s, and the negative sign indicates the direction of rotation. The resonant frequency is known as the Larmor frequency.

MRI rely on the process called nuclear magnetic resonant (NMR) which is the way atomic nuclei interacted with an externally applied magnetic field. It can be achieved through any type of nucleus, as long as it is a non-zero spin indicated by $I \neq 0$, and has angular momentum aligned with the spin axis as

$$p = \hbar I \quad (6).$$

where I is the spin, \hbar is the Plank constant given as $\hbar = h/2\pi$

METHODOLOGY

The reason why main parameters of interest in this research are the respiratory rate, the temperature, the pressure transducer and the electrocardiogram (ECG), this is because the animal model is used as a laboratory experiment for human disease.

The temperature sensor used was a chip thermistor with dimensions x mm, y mm, z mm (NTCS0603E3103JHT, Vishay, US) [18], with b value of 3960K, tolerance of $B_{25/85} = \pm 1$, and R_{25} value of 10k Ω from equation (1) below. The thermistor obeys the exponential law.

$$R = Ae^{(b/T)} \quad (7)$$

where b is the constant temperature in Kelvin (K), and depends on the composition; 3000 – 4000K, the value within the datasheet and A is in (Ω) depending on the sensor shape. Supplied with calibration value of R_0 at T_0 which allows the elimination of A from equation (7) to give

$$R = R_0 e^{b(1/T - 1/T_0)} \quad (8)$$

where R is the unknown resistance, R_0 is the calibrated resistance, $1/T$ is the unknown temperature and $1/T_0$ is the calibrated temperature.

The pressure transducer's datasheet is, Freescale; MPX5100AP/GP pressure transducer with Graseby sensor pad, used for respiration. It is an on-chip condition integrated silicon pressure sensor, suitable for microcontrollers or microprocessors. The MPX5100 is a piezoresistive transducer series.

The abdominal respiratory sensor used is a small pressure pad designed for infant apnoea alarms (MRI0, Graseby, UK). A small volume of air passes through the sensor tube to the monitor, caused by the expansion of the abdominal wall during breathing. Other components datasheet used are; LM285Z – 2.5RAG 2.5V reference, and TC 7660S Charge pump (negative voltage generator).

The ECG preamplifier uses an instrumentation amplifier as the input stage. (INA118, Texas Instrument, USA). It is made up of three operational amplifiers, seven resistors arranged to give a differential gain

stage with a good common mode rejection ratio and a high input impedance. Six of the resistors and the three op-amps are connected to an 8-pin package of the INA118 instrumentation amplifier chip.

The gain of the INA118 instrumentation amplifier is set at the input stage by using an external resistor R_G to a value given as;

$$G = 1 + 50k\Omega/R_G \quad (9)$$

The choice for the gain is to boost the ECG output signal level to about 1 V, and the gain for an R_G value of 50 ohms was calculated to be 1001.

First among the measurement was the thermistor designed to be inserted inside the body of the animal to determine the body temperature. The negative temperature coefficient (NTC) thermistor was soldered to a twisted pair of 0.2 mm enameled copper wires to allow resistance measurements. The thermistor was temporarily glued to a copper block of approximate dimensions of 5 mm x 20 mm x 40 mm to which a silicon temperature sensor with a 10 mV/ $^{\circ}$ C output (LM35, Texas Instrument, USA) [25], was also attached. The copper block was heated up to approximately 40 $^{\circ}$ C by applying a lot of soldering iron for a few seconds, and then was placed on a polystyrene block and covered so that it cooled to room temperature over a period of several minutes. Copper has high thermal conductivity and a high heat capacity, so this helps the thermistor and reference sensor at the same temperature during cooling. The measurement of resistance versus temperature shows that as temperature increases it led to a decrease in resistance.

A signal generator (Gwinstek, SFG-1013, China) was used to generate a sign wave signal of amplitude approximately 1 V at a frequency between 1 Hz and 120 Hz and this was applied to the input of the ECG preamplifier via a 1000:1 potential divider to reduce the amplitude to the level of an ECG R-wave. The amplitude of the output and input voltages were recorded using an oscilloscope (Tenma colour signal, Tenma Testing Equipment, USA) and the transfer function $|T(f)|$ was calculated from the ratio and plotted against frequency as seen if Figure 1.

RESULTS AND DISCUSSION

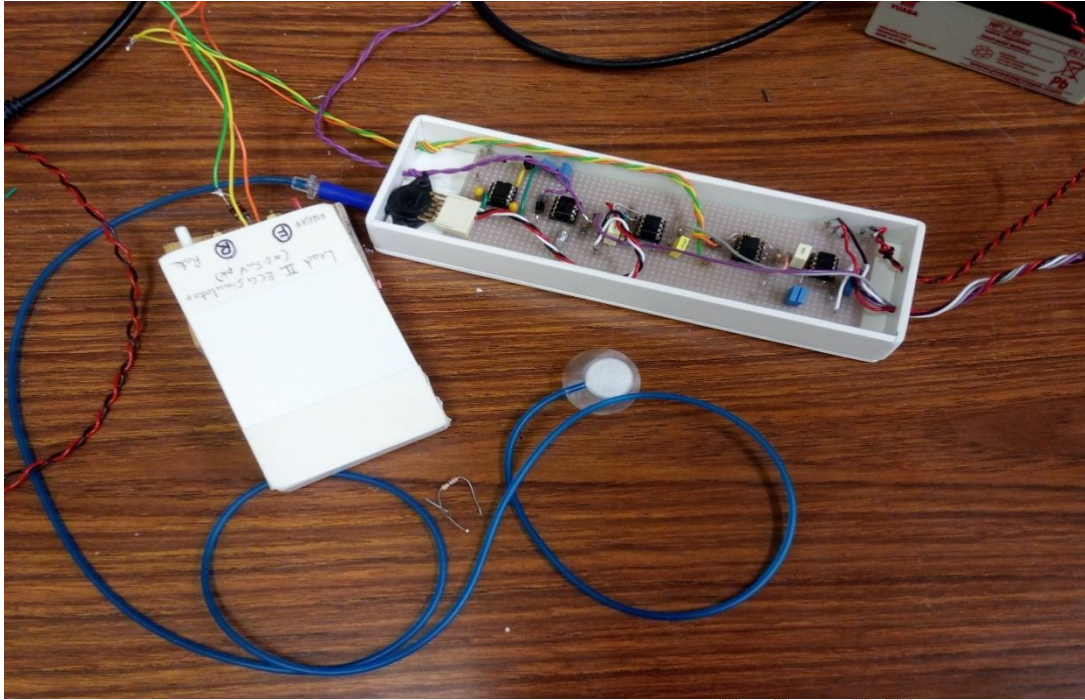


Figure 1: ECG simulator connected to the circuit board, with the respiration pad attached to the silicon pressure sensor

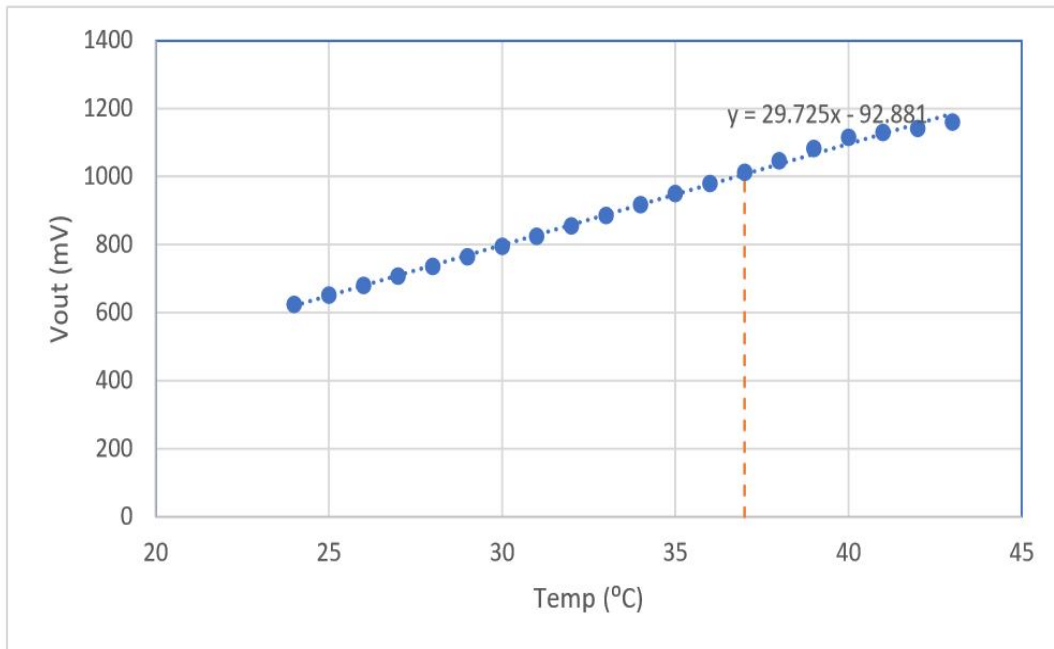


Fig 2: Thermistor response versus temperature when operated with voltage bias and linearizing

The fitted line shows that when operated in the voltage-biased circuit, the thermistor response to temperature is more linear and has a sensitivity of approximately $30 \text{ mV}/^{\circ}\text{C}$. The vertical dashed line indicates a normal body temperature of 37°C .

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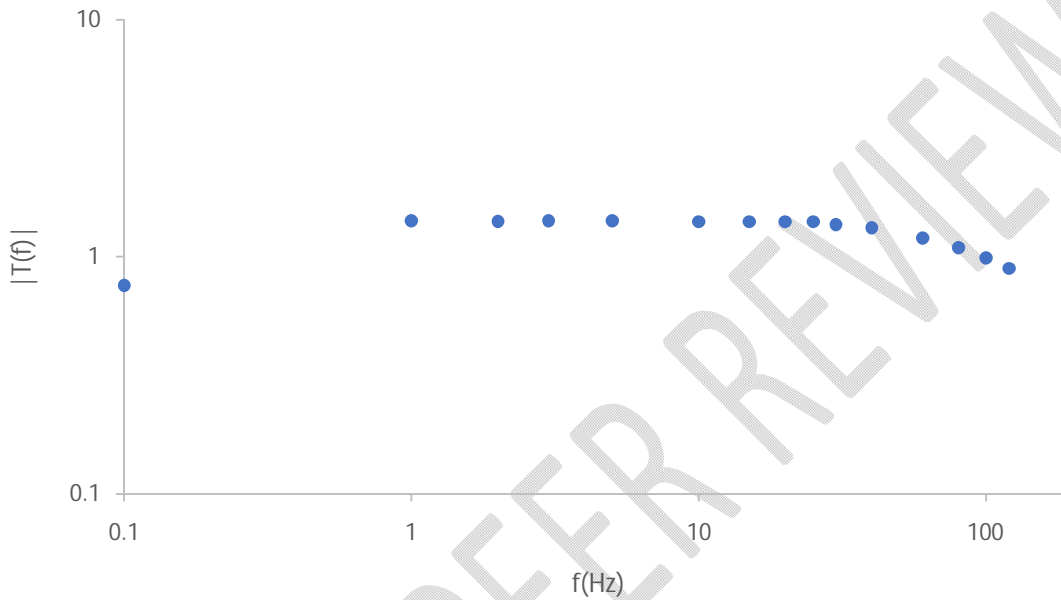


Figure 3: The frequency response of the pressure sensor's preamplifier.

The calculated high pass cut-off frequency is 0.16Hz, while the cut-off frequency from the oscilloscope is 0.2Hz. The specific low pass cut-off frequency is 100Hz, and the measured frequency from the oscilloscope is 99Hz. In this case, the output signal leads while the input signal lags behind, and there was a distortion in the waveform, as the amplifier tends to deliver higher output voltage, the waveform started clipping. Clipping is a situation where the output tends to go beyond the power supply rails which is not symmetrical. The gain and low characteristics of the pressure sensor as shown in Fig3 above.

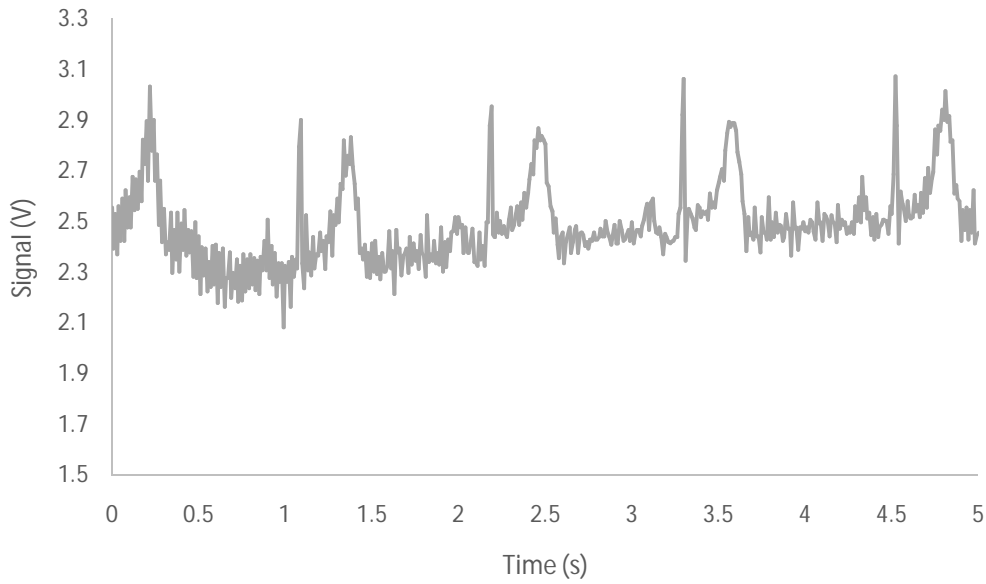


Figure4: The author's ECG.

The author's ECG trace show noticeable 50Hz interference.

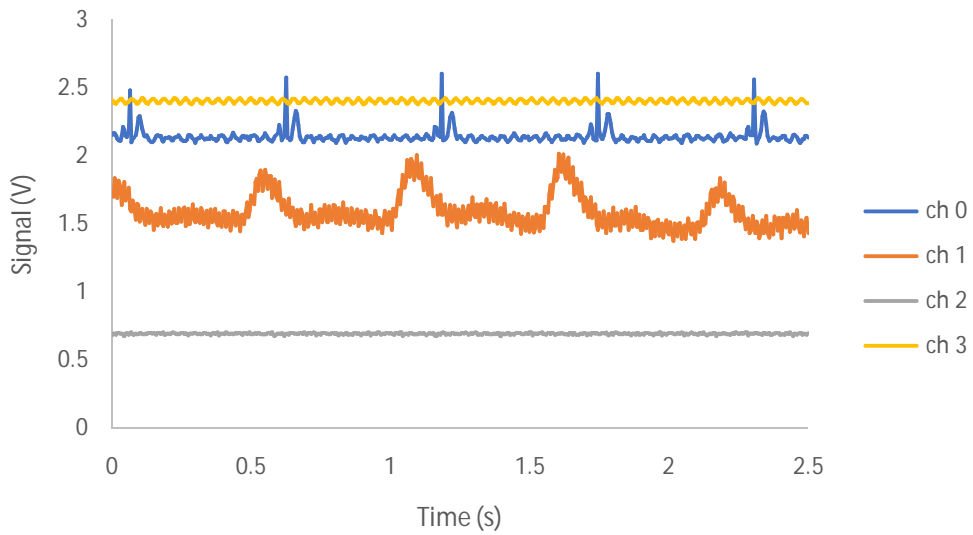


Figure5: Image inside the magnet

Channel 0: ECG, channel 1: pulse, channel 2: temperature and channel 3: battery voltage.

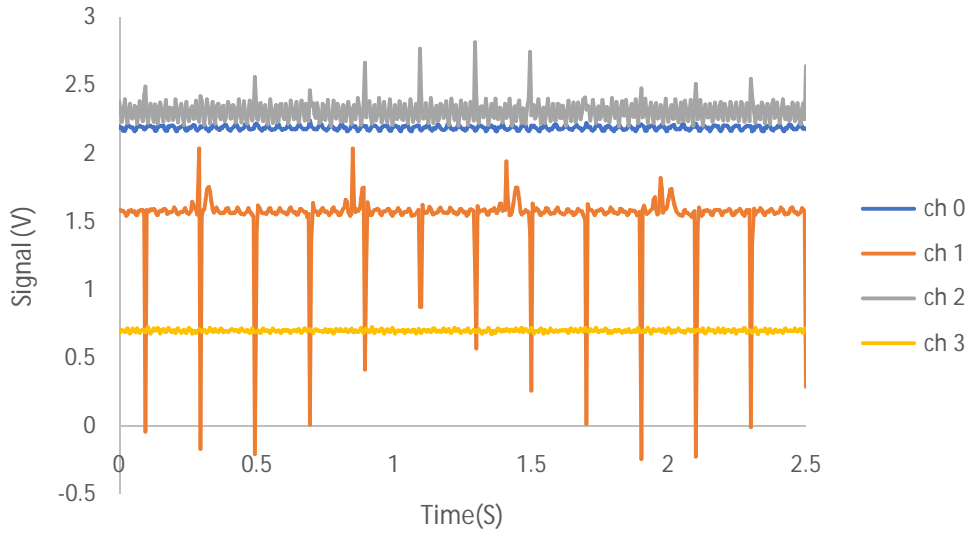


Figure6: Image outside magnet

Channel 0: battery voltage, channel 1: ECG, channel 2: vibration and channel 3: temperature.

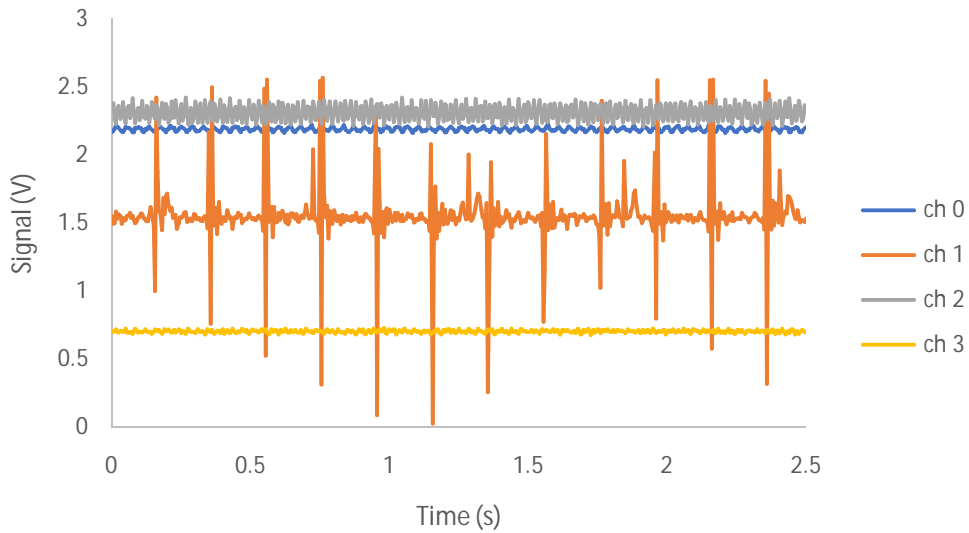


Figure7: Image near the magnet

Ch. 0: battery, ch. 1: ECG, ch. 2: Vibration, and ch. 3: temperature.

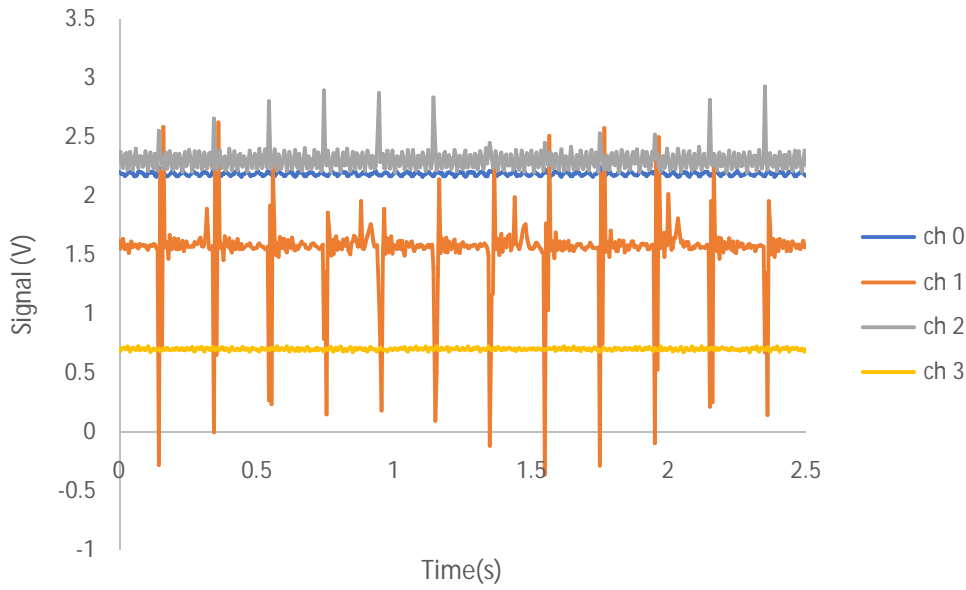


Figure8: Image with RF and gradient on

Ch.0: battery, ch. 1: ECG, ch. 2: vibration, ch.3: temperature.

DISCUSSION AND RECOMMENDATION

The physiological sensors and signal processing circuits were tested in the MRI system. An MRI-compatible lead II ECG simulator, with approximately 0.5 mV output amplitude was attached to the input of the ECG preamplifier to represent a small animal's signal as shown in Figure 1. The circuit board is then connected to a fibre optic link with a transmitter and a receiver, and the receiver is linked to a 12-bit ADC (1208FS, Measurement Computing, USA) and waveforms were captured using oscilloscope software (TracerDAQ, Measurement Computing, USA). The transmitted signal is the ECG, the pressure sensor, the temperature, and the battery (voltage), these are all linked to different channels as seen in Figure 5-8. Figure 5-7 (outside, near and inside magnet) were acquired with the pressure sensor pad taped on a male adult volunteer's thumb to record the human pulse of 70 bpm, which is similar to the respiration rate of an anaesthetized mouse or rat. Then Fig.8 was acquired with the pressure sensor taped to the surface of a phantom to pick up vibration from the gradient system.

The gradient echo measurements of the parameters were carried out to test interactions between the imager's RF and gradient pulses and physiological sensors. The tests used a gradient echo pulse sequence with a bandwidth of 50kHz, 256 x 256 pixels, slice thickness 1mm with a field of view (FOV) of 50mm. The gradient ramp time is 1000 μ s, the repetition time (TR) of 200ms, the echo time (TE) of 11ms and the flip angle of 45 $^{\circ}$.

In this research, our major consideration is “does the scanner influence the components and does the components influence the final image?”. Most of the components used are slightly ferromagnetic (i.e. the resistors and the op-amps). The testing and measurement carried out show that all the monitoring parameters do not perform as expected in the MRI system. The component’s interaction with the MRI needs more research for MRI compatibility to be achieved. And nonferromagnetic components should be used.

The use of a signal gating generator should be explored in future work, this could make the monitoring parameters to be independent of the RF pulse influence as seen in Fig 5-8. Finally, fibre optics should be included in the design process.

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