Brick Computing

Final Project Report

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1. Introduction

This project aims to explore the possible mechanisms and designs of a floor tile that can actively monitor a person's physiological measurements. The motivation behind this project is to have the ability to collect physiological measures over an extended period of time non-intrusively, i.e.; without the active participation of the user by incorporating the sensing and networking technology into the building material of the user's physical environment, such as the flooring or walls.

The report first outlines the project objectives and requirements in Section 2 and 3. Section 4 gives a detailed overview of the proposed mechanical design and the sensing methodologies to be used. Section 5 outlines the progress to-date with regards to the prototype development and data acquisition system. The report concludes by discussing some of the current issues with the prototype design and suggested solutions.

2. Project Objectives

- Determine the types of sensors and measurements that can be collected from the feet
- Investigate practical ways of embedding sensors into floor tiles
- Build and test an initial prototype of the proposed system

3. Project Requirements

In order to meet the objectives the system must be non-intrusive and must have the capability to automatically log and measure the physiological data.

4. Proposed Design

4.1 Sensing Technology

The following table (Table 1) outlines the physiological measures that can be obtained non-invasively form a person's feet with high accuracy, the sensing technology that is most applicable for measuring these values and their method of procurement. (Please refer to the earlier submitted Proposal document for detailed discussion on reasoning behind the methodology and sensing technology choices)

Physiological Measure	Sensors	Methodology	Drawbacks
Heart rate	Strain gages (load cell)	Measure changes in forces produced by BCG	Motion artefacts - can be removed by simultaneous ECG/EMG monitoring[1]
Blood Pressure	Dry electrodes[2]	Based on correlation between ECG R-peak and BCG J-peak	High correlation for SBP but moderate correlation for DBP[3]
Respiration Rate	Load cells	Respiration causes cyclic variation in BP wave	Movement artefacts
Body Weight	Strain gages (load cell)	Measure the total change in voltage produced by load cells	-
Body Temperature	Thermistor[4]	Change in temperature causes a measurable change in resistance	Heat dissipation in tile material; Highly influenced by room temp
Motion	Piezoelectric sensor or switch	Weight activated	Can be falsely activated

Table 1. Physiological measure and relevant sensors

Below is a detailed description of the various embodiments of the design that follow in connection with Figure 1, which shows the system overview and process flow of a general prototypic design for determining the above mentioned physiological measures.

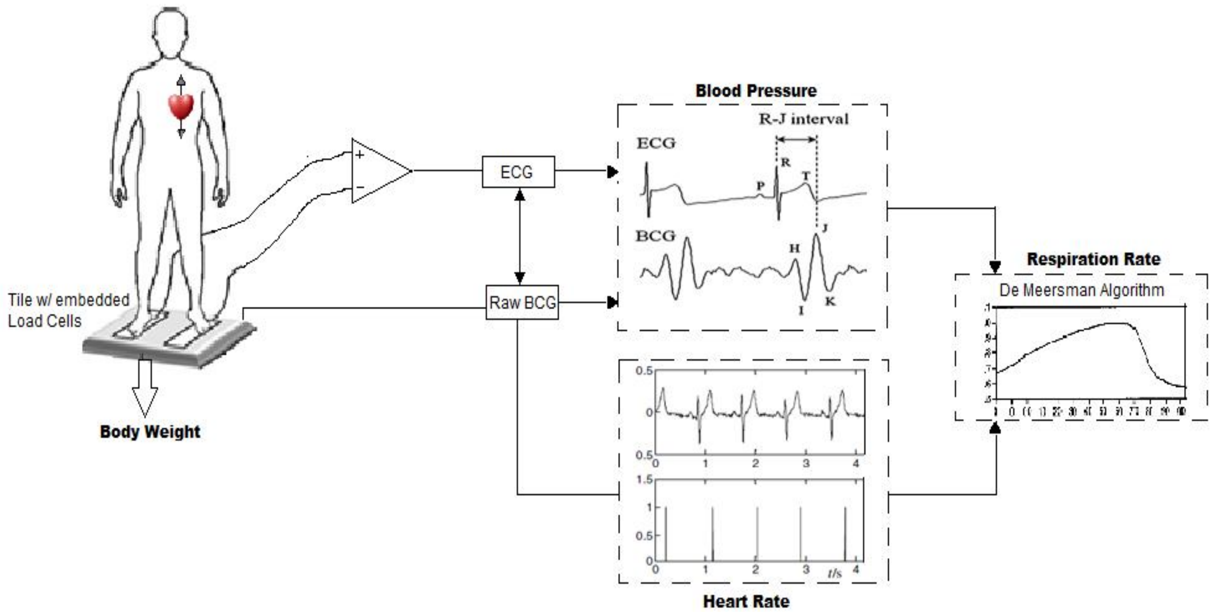

Figure 1. System overview of proposed design

The heart rate, one of the principal vital signs, acts as an important source of data itself; indicating a person's level of fitness or the presence of disease. In the proposed design, the subject's heart beat will be measured using the ballistocardiogram (BCG), a non-invasive diagnostic technique that measures the mechanical recoil of the body in reaction to the ejection of blood from the heart through the vasculature. During early systole, the left ventricle accelerates a volume of blood through the ascending aorta and according to Newton's 3rd law; the body moves footward and exerts a force on the ground in reaction. Similarly, during late systole the blood volume ejected accelerates through the descending aorta due to its compliance and gravity, making the body move headward, thus reducing the stress exerted on the ground. These changes in force exerted by the body can be detected using the strain gages or load cells attached to a platform[5].

Blood pressure (BP), the pressure exerted by circulating blood upon the walls of the blood vessels, will be measured using the heart rate signal in correlation with ECG data obtained from the subject[3]. A low-level current of 1 mA (rms) at 10 kHz will be injected into the subject through the dry electrodes to acquire an ECG signal. An automatic peak detection algorithm[6] will be run on the raw ECG signal to detect the R-peaks. The time interval between the ECG R-peak thus obtained and the BCG J-peak (obtained through post processing techniques explained below) will determine a BP estimation equation for each subject. BCG J-peaks will be identified as the maximum peak of the BCG signal between two sequential R-peaks of the ECG signal[3].

Since respiration modulates heart and blood pressure variability, knowledge of respiration can be extracted from the pulse wave and BP wave. De Meersman *et al* demonstrated that given a pulse wave, the respiratory rate could be accurately determined with high correlation (Table 2)[7]. The technique is based on the relationship between oscillations in the area under the dichrotic notch of the pulse wave and respiration.

Protocol	Correlation Coefficient	P Value
Controlled breathing	0.98	0.001
Exercise	0.79	0.05
Isometric contraction	0.98	0.001

Table 2. Correlation coefficients and probabilities between actual breathing Spectral peaks and blood pressure-derived spectral peaks11

The proposed design will include a thermistor embedded in the surface of the tile, to allow for measurement of subject's body temperature.

To monitor patient motion within the environment, the floor tiles will consist of piezoelectric sensors embedded under the load cells. Piezoelectric sensors consist of thin polymeric piezoelectric films that generate a voltage on undergoing mechanical deformation. The use of a rigid casing for the piezo film would ensure activation only by vertically applied forces[8]. To ensure power saving, the other sensors on the tiles could be weight activated using these piezoelectric sensors to sense a person's presence on the tile.

4.2 Physical Design Overview

The floor tile will contain four load cells, one at each corner of the tile to ensure equal distribution of the weight between them (Refer to Figure 3). Each load cell must be relatively compact with a flat disc structure and be sensitive enough to measure the minute force variations produced by the BCG. Experiments performed by Gonzales-Landaeta et al determined these heart-beat-related forces to vary form from 0.24 N minimal to 6N maximal in a single individual[5].

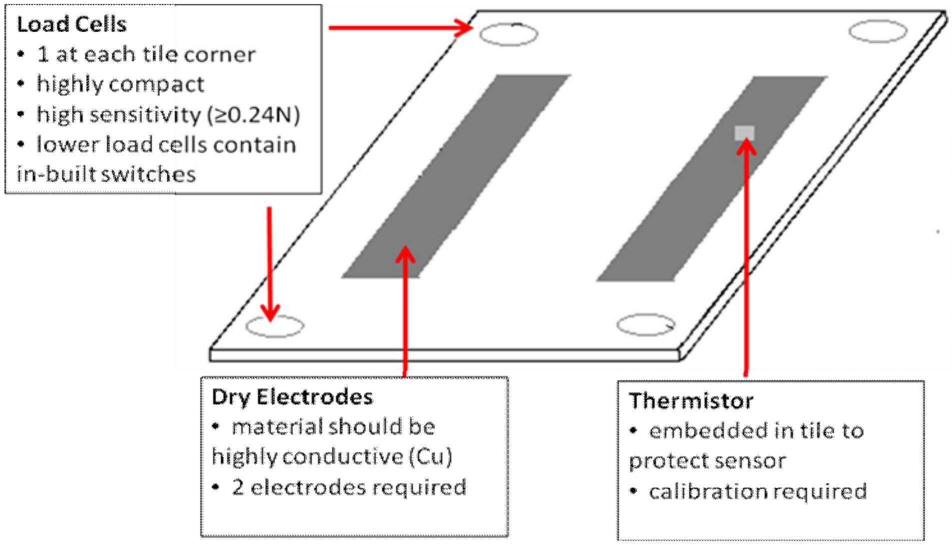

Figure 3. Proposed design layout

The tile will have two dry electrodes in-built onto the surface. The electrodes must be of a highly conductive and non-rusting material such as Copper and must be large enough to easily cover a subject's foot.

To protect the thermistor from the surrounding environment and possible damage, it is recommended that a 1-2m copper plate be epoxied on top of the sensor surface or the sensor be embedded below the dry electrodes itself. This would also ensure good thermal conductivity between the sensor surface and the tile surface. However, the sensor must be calibrated to ensure that the accurate tile surface temperature (and not the sensor surface temperature) is being logged.

The main pre-processing circuit board for the design will be housed in a compartment on the base of the tile.

4.3 Data Acquisition

To obtain the required pulse signal, only the AC components of the load cell signal must be extracted and a large signal-to-noise ratio (SNR) is required in order to detect the heart rate with signal processing methods. A block diagram of the measurement system used is given in Figure 4.

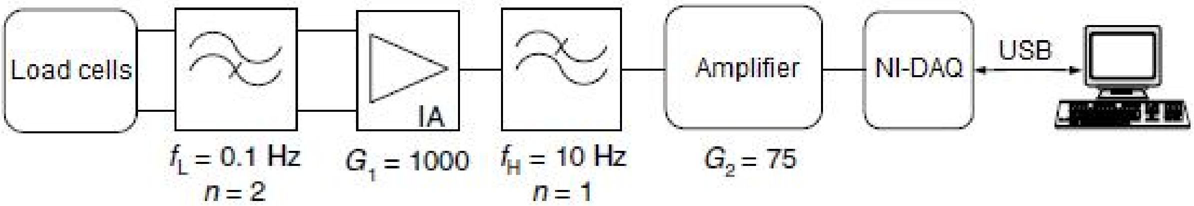

Figure 4. Block diagram of the measurement system to obtain a pulse signal from heart-beat force variations

The strain gauges on the tile are connected in a Wheatstone stone configuration and the bridge is excited by a DC voltage of +/-5V. Figure 5 shows the detailed circuit design.

To achieve a large ac dynamic range, the proposed circuit has two ac-coupled stages: a front differential ac-coupling network (passive input stage) and a high-pass difference amplifier (active dc suppression stage). The first stage consists of a high-pass differential filter in front of a dc amplifier. To remove any op-amp input offset voltage and reduce noise, a second stage is added that consists of an integrator in a feedback loop[9].

Because the circuit concentrates its gain in the first stage, its equivalent input noise is determined by the op-amp composing the input stage. The design uses the low noise, high gain (1000V/mV) and high common mode rejection ratio (CMRR) AD8221 op-amp.

OP1177 in the feedback loop ensures that only frequencies above a certain threshold (0.1Hz) are output from the differential amplifier. This output signal is then filtered by a first-order low-pass filter (24Hz) to reduce noise bandwidth and amplified (75V/mV) by a second low-noise amplifier (AD743). The signal is thus band limited between 0.1 and 24Hz (the average frequency range of a human heart beat)[8] and amplified with an overall circuit gain of 98dB.

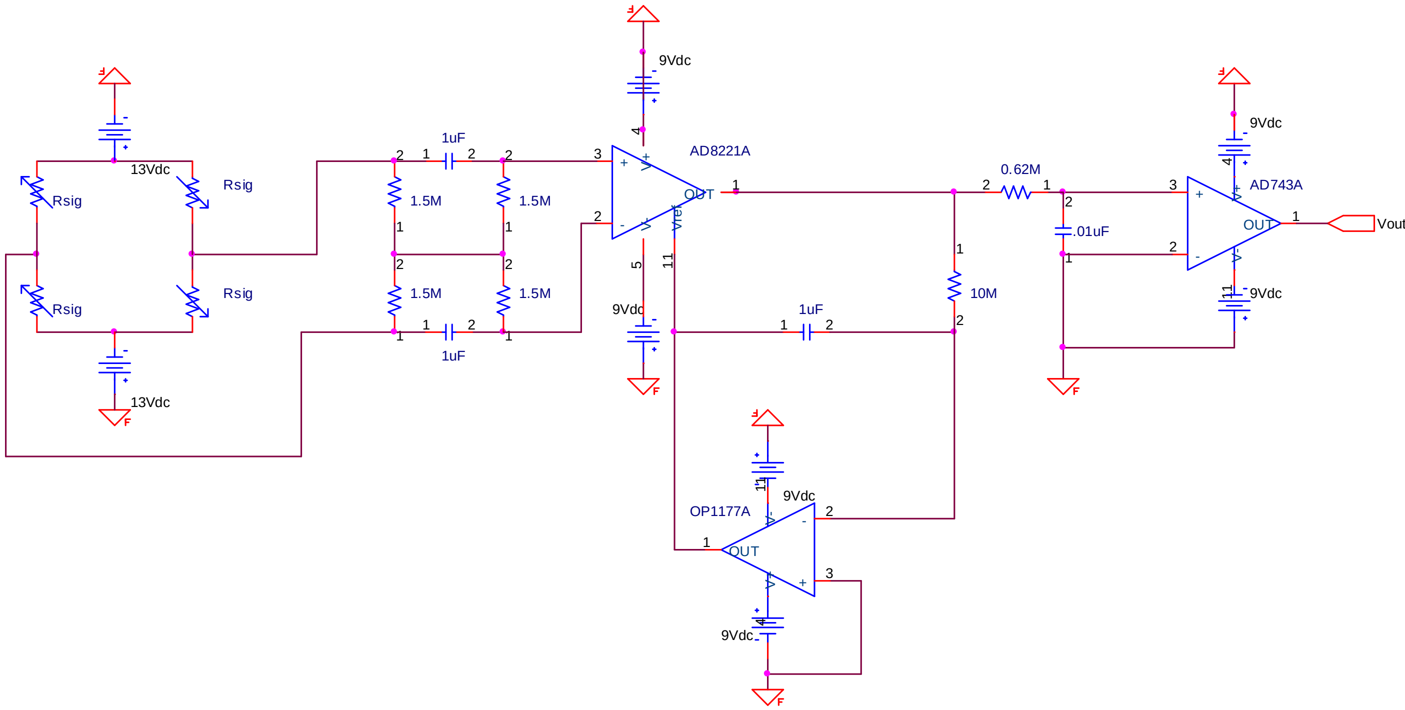

Figure 5. Pre-processing circuit design

A NI data acquisition system (NI cDAQ-9174 chassis and NI 9215 analog input module) controlled by a LabView® algorithm, will be used to record the BCG at a sampling rate of 1 kHz with a 16bit resolution.

A similar circuit can be used for recording the ECG signal. The ECG signal will be band-pass filtered from 0.1 to 40Hz and amplified by 60dB. The same data acquisition card will be used to record the data simultaneously with the BCG, using the same sampling rate.

4.4 Digital Signal Processing

To extract QRS complexes from the raw ECG data, a constant threshold; half the maximum value of the first n (n=5) readings of the ECG; will be set for the filtered ECG waveform, and the local maxima above this threshold will be automatically located and annotated as R-waves (using MATLAB®). BCG and ECG beats will be segmented by windowing the signals around these R-waves (Figure 6.). The windows to be used for segmenting BCG beats will start at the preceding ECG R-wave with a window width equal to the minimum R-R interval for that recording. BCG J-peaks will be identified as the maximum peak of the BCG signal in a windowed segment[3].

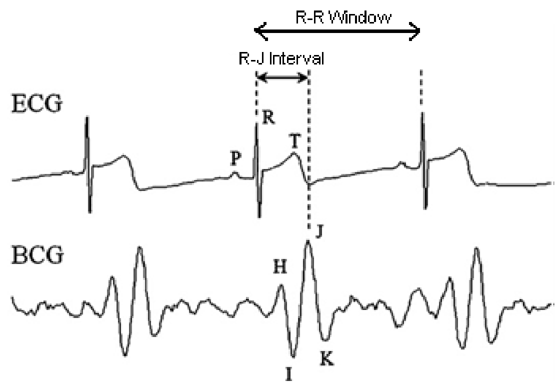

Figure 6. Measured ECG and BCG signals with lettered waves and time interval parameters

Previous work carried out by Shin et al[3] has shown that the ECG signal measured from the feet is often corrupted by EMG generated by the gastrocnemius and plantar muscles of the feet, especially when the subject is standing. An ensemble averaging technique developed by Lander and Berbari et al.[10] can be used to reduce noise components in the desired ECG signal.

Once the ECG R-peaks and BCG J-peaks have been accurately identified, a linear regression method can be used to estimate beat-by-beat systolic blood pressure form the R-J intervals

$SBP = a * RJ_{int} + b$; where a and b are calibration constants.

5. Prototype Overview

5.1 Mechanical Interface

A Home Collection 63-8709-8 digital bathroom scale was used as the sensor platform. The scale consists of four strain gauges located at each corner of the platform connected to form a Wheatstone bridge. The bridge was balanced at 0.7 V and the unloaded resistance of each strain gauge was 875Ω. The cross sectional area of the weighing scale was 13”x12” with an operating range of 2 – 150 kg and measurement accuracy of 0.1 kg (0.90 N).

5.2 Sensitivity estimation

Scale sensitivity depends on the strain gages and the excitation voltage supplied to the Wheatstone bridge. When applying a symmetrical excitation voltage (±*V*CC), the bridge output is a differential voltage proportional to 2*V*CC and to the force, with zero common-mode voltage. To estimate the static sensitivity, each load cell was supplied at ±2 V and its output voltage was measured by a digital multimeter (Fluke® 77-III). The output voltage was first measured for an unloaded scale and then for standard weights (2.5 lbs, 5 lbs and 15lbs). The output voltage increased linearly with the weight, as expected and the scale sensitivity was measured to be 1818 nV V^{-1} N^{-1}. A detailed schematic of the scale's load cell layout is given in Figure 7.

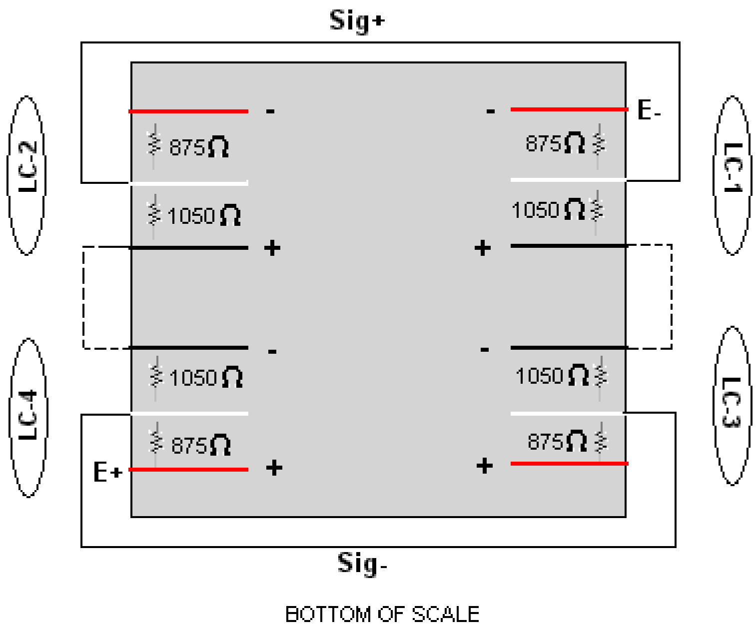

Figure 7. Home Collection digital bathroom scale - electronic layout.
LC – load cell, E – excitation voltage, Sig – output signal

5.3 Force measurement

To check if the proposed setup could in-fact measure cardiac-related forces (attained from the feet) that are superimposed on a much larger DC force (the subject's weight), a force plate (ATMI® AccuSway dual force plate) was used to detect BCG readings. Preliminary tests were run on two healthy subjects standing still on a force plate and the vertical and horizontal force variations were recorded. The subject's heart rate was simultaneously monitored using a pulse oximeter placed on the finger. Distinct 1.1Hz spikes were seen in the force signal which corresponded to the heart rate measured by the oximeter (Figure 8).

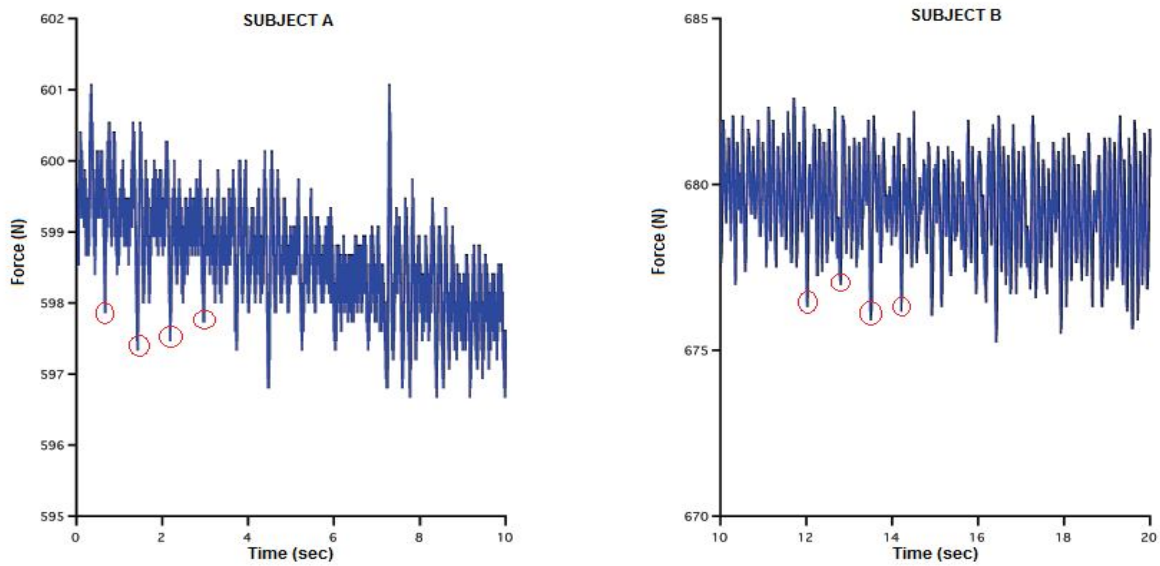

Figure 8. Force variation data recorded from quiet standing of subjects on a force plate.

6. Discussion

The main purpose of the prototype was to amplify force variations produced on a platform during quiet standing (BCG) and store the recorded data in an excel file. From research the expected BGG signal obtained from the subject's feet should mimic Figure 9.

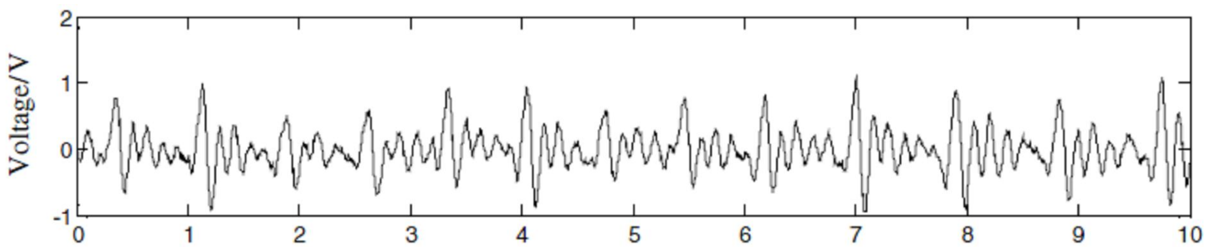

Figure 9. BCG force signal obtained from a subject with footwear.
(man, age = 23, weight = 92kg, height = 1.94m)[8]

The current prototype although displaying primitive functionality is detecting variations in force exerted on the platform, cannot effectively detect BCG due to the high noise level corrupting the signal. Figure 10 graphs the output obtained from a subject standing on the scale and periodically jumping. Figure 11 graphs the output obtained from lightly tapping on the surface of the scale.

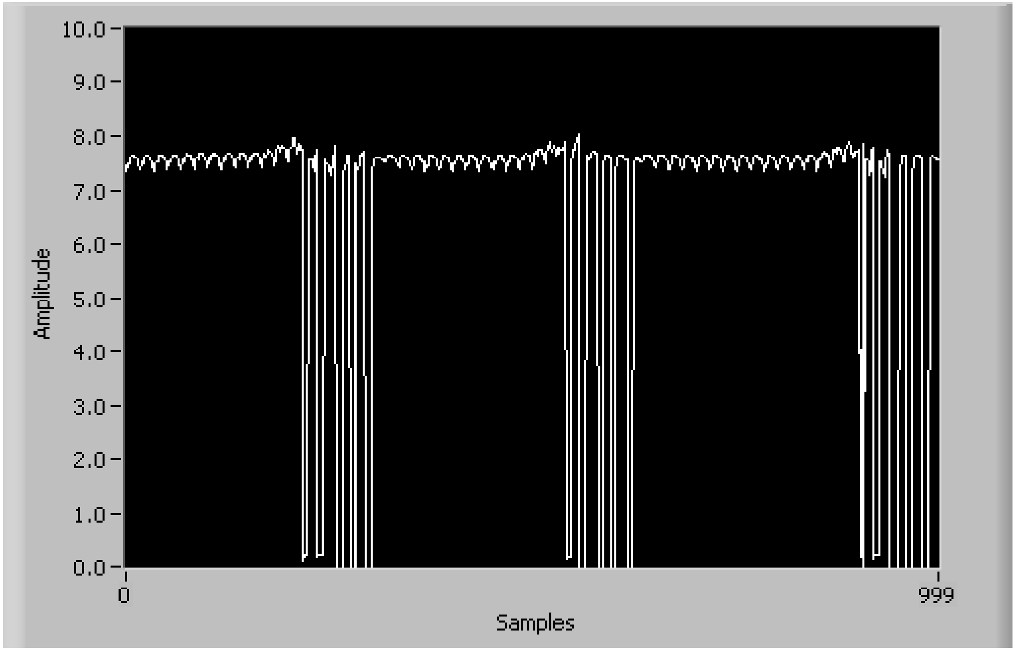

Figure 10. Time trace signal obtained from subject (female, age = 23, weight = 53kg) standing on scale and periodically applying extra pressure on platform

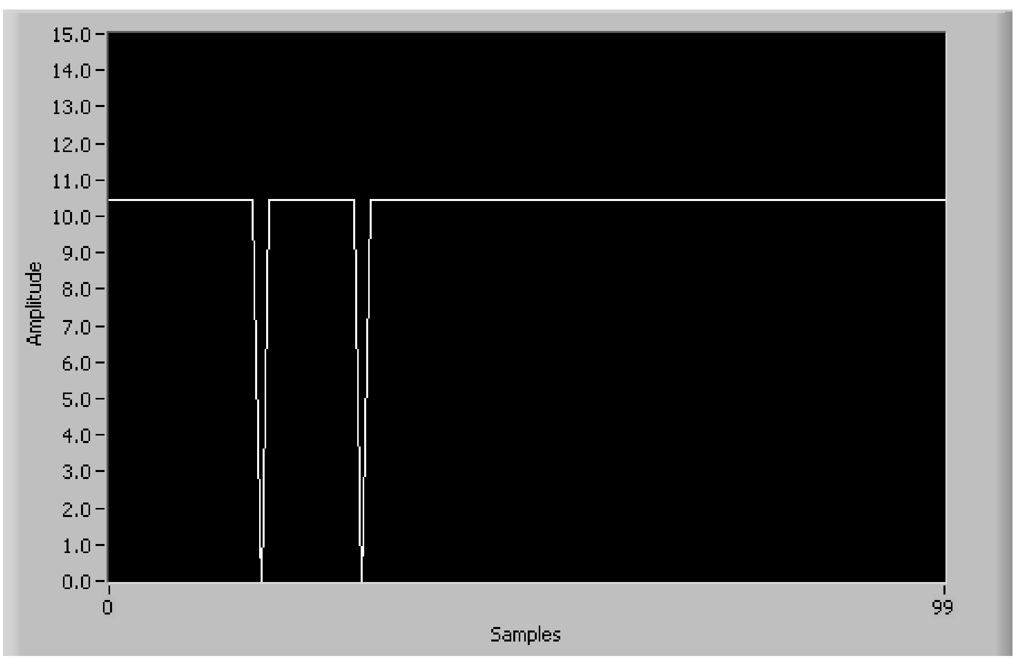

Figure 11. Time trace signal obtained from lightly tapping a finger on the scale platform

Outlined below are some of the possible sources of error/ causes of noise in the circuit and the steps that I have already taken to reduce these errors.

1. Poor or loose connections due to improper soldering

   The circuit was transferred to a soldered bread board and all connections were checked with a digital multimeter for any loose/wrongful connection.

2. Lack of a proper grounding plate for the DAQ system and for the circuit

   In order to keep the DAQ mobile a proper grounding system was not developed and grounding loops may exist in the circuit, making the circuit prone to noise from vibrations and movement. The circuit and all power lines are connected to a common ground which is connected to the ground of the DAQ.

3. Magnification of offset voltage by high gain operational amplifiers

   To eliminate offset voltage magnification the differential amplifier was filtered to reject low frequencies using an integrator in a feedback loop. However since the output signal does contain a dc component of approximately 10V, offset voltage amplification may not have been totally eliminated.

4. Noise produced on the power supply pins

   Bypass capacitors were added to each of the power lines to decouple the amplifiers. Additionally a stable DC power supply was used as the voltage source to minimize noise in the power supply lines.

5. Flicker noise or 1/f noise produced at lower frequencies

   To eliminate flicker noise the differential amplifier was filtered to reject low frequencies using an integrator in a feedback

6. Although the circuit is based on literature, the choice of capacitors or resistor may potentially be incorrect for the measurement of the BCG signal.

7. Interference noise due to current or voltage spikes, vibration, electrostatic or electromagnetic sources.

   Table 3 outlines common sources of interference noise, their typical magnitudes and suggested methods of addressing them.[11]

External Source	Typical Magnitude	Typical Cure
60Hz power	100pA	Shielding, attention to ground loops
120Hz supply ripple	3μV	Supply filtering
180Hz magnetic pick-up	0.5μV	Reorientation of components
Vibration	10pA (10 to 100Hz)	Elimination of leads carrying high voltages near input terminals
Cable vibration	100pA	Use low noise cables (suggestion: carbon coated dielectric)
Circuit Board	0.01 to 10pA	Clean board and use Teflon insulation as needed

Table 3. Interference noise

References

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