TECHNICAL SPECIFICATIONS

I. Introduction

This electromyograph (EMG) aims to determine reflex time of patients' muscle responses by measuring the electrical activity of their muscles. By utilizing a reflex hammer, we can provide controlled stimuli to the muscle. Once the stimulus has been administered to the patient, surface electrodes can record the muscle response. The time difference between these two signals will provide physicians with the reflex time.

Figure 1 shows the basic schematic that our EMG hardware follows. We begin with various hardware components that filter or amplify our signals in order for LabView to properly analyze them.

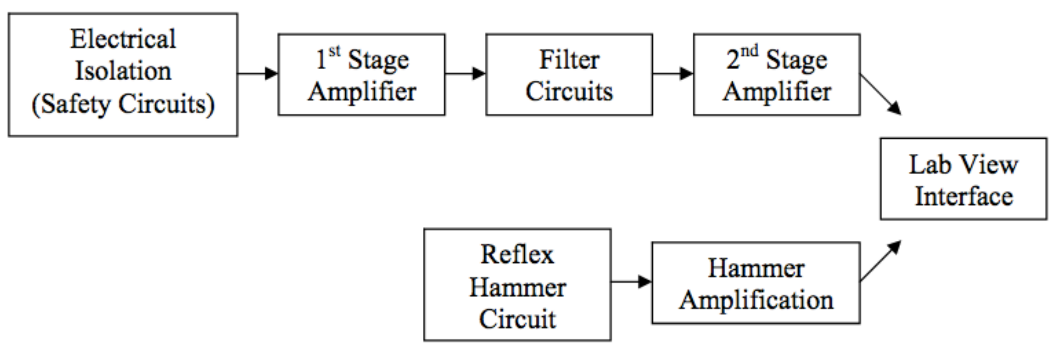

Figure 1: Electromyogram Schematic

The two main signals our hardware aimed to modify were that of the reflex hammer and the muscle under investigation. The signal of the muscle requires many hardware components to successfully obtain for there are three main issues to overcome – noise, signal size and electrical safety. Safety circuits must be implemented to ensure that the voltage will travel from the muscle to the EMG rather than from the EMG to the muscle. If there are faulty components or circuits on the EMG, some amount of voltage might want to travel backwards to the subject. As a preventative measure, electrical isolation circuits were placed in our electromyograph. In addition to safety circuits, the voltage difference measured by the two electrodes is very small and thus amplification is needed. In order to optimize our amplification, two rounds are carried out. The first round takes place before the signal is sent to the filter. A filter is necessary because the muscle signal is a noisy one. There are many electrical activities that occur in the body as well as intrinsic body movements. Thus, filters must be implemented to rid the signal of that noise. After filtering, one last round of amplification is administered to create a reasonably sized filtered signal. Unlike the muscle signal, the signal of the reflex hammer will simply need an amplifier for there is very little noise associated with the output signal and the hammer does not directly have any electrical pathways to the body.

II. Reflex Hammer Circuitry

The reflex hammer signal will produce a generally constant voltage with changes in voltage corresponding to any hammer strikes. This will allow us to determine the time at which the tendon is being struck. The hammer and the protoboard both have male DB9 connectors and thus, an alternative way to connect the hammer output to the protoboard was needed. We obtained an unconnected female DB9 connector and soldered wires into its connectors. The hammer's male DB9 connector could then be plugged into the female DB9 connector. Then, we connected the soldered wires via the protoboard to the appropriate power sources and circuitry, according to the hammer pin-out diagram displayed in Figure 2.

PIN	DESCRIPTION
1	Shield
2	V_{in} +
3	Ground
4	V_{in} -
5	Shield
6	+ 5 V (ref)
7	No Connection
8	No Connection
9	- 5 V (ref)

Figure 2: DB9 Male Connector Pin-out Diagram for Hammer Circuit

We placed all of the wires into the DB9 connector pins section of the protoboard. Each wire was placed into its corresponding numerical pin. Pins 1 and 5 were shield and were subsequently grounded. Pin 6 and pin 9 required input voltage that would power the hammer. The hammer requires connection to power sources of +5 v and -5 V. The hammer will not be damaged by slight variations in positive voltage so we were able to directly connect our +5 V DC voltage supply to pin 6. However, the hammer is very sensitive to excessive negative voltage and thus the input negative voltage could not be significantly less than -5 V without causing damage to the hammer. In order to prevent this, the following voltage divider was implemented (seen in Figure 3).

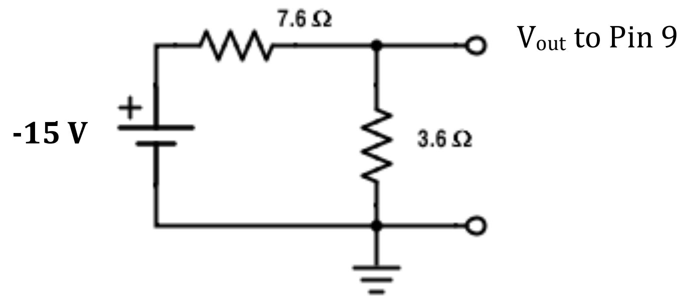

Figure 3: Voltage Divider for Pin 9

This voltage divider was designed to divert -4.86 V of the total -15 V DC power supply to pin 9 which is able to reduce the risk of damage to our hammer by remaining above -5 V at all times.

Once the wiring and powering were in place, we could then obtain an output voltage from the hammer by measuring $V_{in}+$ and $V_{in}-$. The output voltage of the reflex hammer is quite small and so amplification of the signal was necessary. In order to amplify the signal to a more analyzable size, our team designed a differential amplifier with an OP07 operational amplifier and two identical sets of two resistors, one for each input as seen in Figure 4.

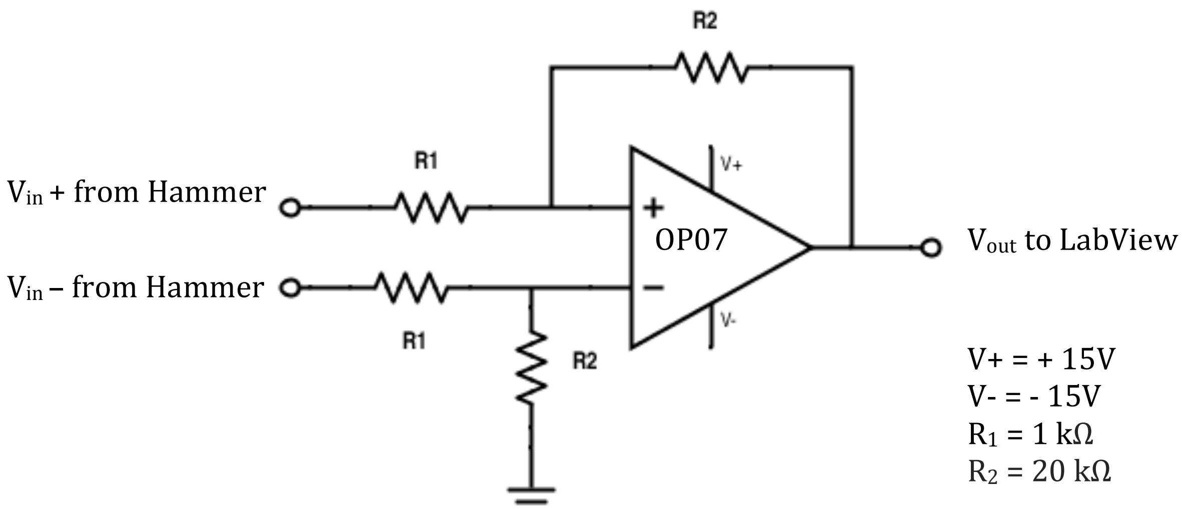

Figure 4: Differential Amplifier for Hammer Circuit

A differential amplifier was chosen to maximize amplification of the voltage difference between $V_{in}+$ and $V_{in}-$ while rejecting any common voltage. This amplifier is able to maintain a second input instead of resorting to a ground connection, the voltage difference is amplified instead of simply obtaining a single voltage input. The gain for our operational amplifier configuration followed the equation below:

$$\frac{V_{out}}{V_{in+} - V_{in-}} = \frac{R_2}{R_1}$$

The combination of resistors that allowed our amplifier to provide the recommended gain of 20 for our hammer output signal was R_1 = 1 kΩ and R_2 = 20 kΩ. After amplification, this signal will be sent to LabView for analysis and calculations in conjunction to measured muscle signal to obtain reflex time.

III. Muscle Electrical Activity Detection Circuitry

The purpose of the electrical activity detection circuitry is to obtain an easy to analyze signal of the muscle contraction. This begins at the patient electrode interface. As the stimulus is applied, the muscle will begin to contract. As the contraction travels through the muscle fibers in the form of action potentials, positive and negative electrodes placed along the length of the muscle fibers will detect the voltage change. Voltage from each electrode will then go through an electrical isolation circuit. This circuit (depicted in Fig. 5) ensures there is one-way voltage flow for any voltage greater than 0.6 V by incorporating diodes. 0.6V is the cutoff threshold for unidirectional voltage flow through the diodes. Anything greater than the threshold will instead be diverted through the diodes to ground. In addition, utilizing two diodes in the configuration shown in Fig. 5 will allow for both large positive and negative voltages to be directed to ground.

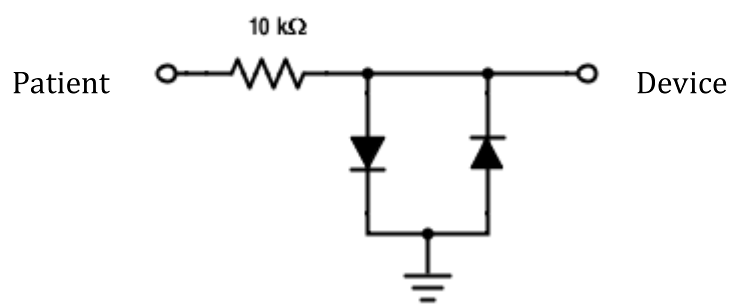

Figure 5: Electrical Isolation Circuit

By limiting the entrance of large voltages into the body, we can protect the patients and reduce shock risk.

The two voltages will now be fed into an AD620 instrumentation amplifier powered with 15 V and -15 V. For this kind of amplifier, only change of one resistor is required in order to modify gain. This amplifier is a type of differential amplifier, so while it is able to amplify voltage differences and reject signals common to both inputs, it also has some additional advantages. Instrumentation amplifiers are able to buffer the two inputs so that there is no need for impedance matching and the common mode rejection will not suffer as a consequence of different current amounts flowing through each input branch.

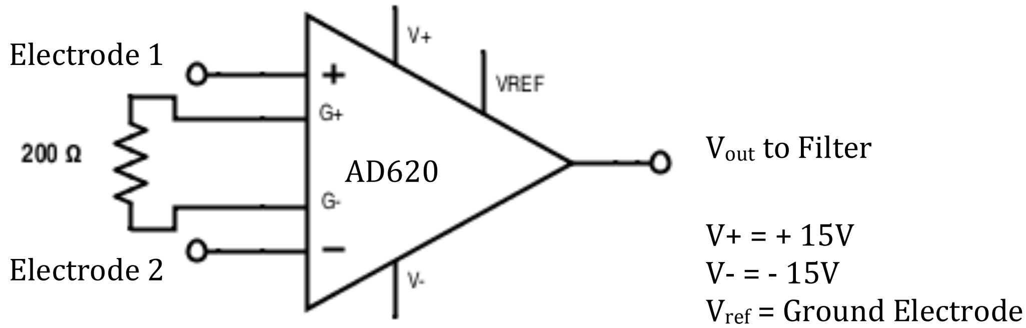

Figure 6: Instrumentation Amplifier

Previous research demonstrated that the anticipated output voltage would range from µV to mV[1]. Thus, our team decided that a gain of ~250 would provide a reasonable output signal. If the signal became too small, further amplification could be implemented after the signal has been filtered accordingly. In order to design our instrumentation amplifier configuration, we utilized the following formula to demonstrate the mathematical connection between our gain and the resistor value chosen (obtained from amplifier datasheet):

$$G = \frac{49.4\ k\Omega}{R_G} + 1$$

A resistor value of 200 Ω and a gain of 248 were calculated based on the above equation. Since the inner components of this amplifier is set up for gain changes to be dependent upon one external resistor, this is the only resistor necessary for this circuit.

In addition, the ground electrode output was inputted into V_{ref} and serves as a way for the amplifier to reduce the amplification of that signal. Different patients have different surface electrical signals and movements throughout the signal collection process. In order properly analyze the reflex signal, we want to reduce amplification of these portions as much as possible. By including a ground, this circuit is able to more effectively reduce noise and analyze the muscle signal.

Lastly, we placed our amplified signal through a bandpass filter to remove extraneous electrical activity collected because of movements of the body and thermal noise generated by the components.

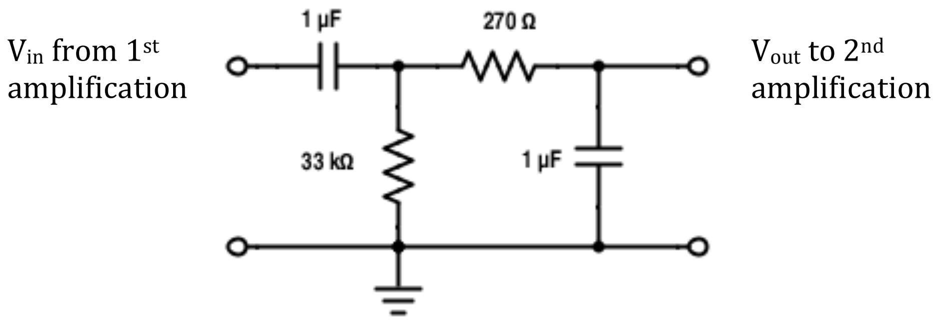

Figure 7: Passive Band-Pass Filter

Our team selected a passive filter because the components would be easier to fix if they broke and additionally, using active filters would have no real benefit for us, since we are amplifying our signal separately. The range of frequencies that are pertinent to our signal will lie in the 5 – 500 Hz range and thus, our team implemented a band pass filter with cutoff frequencies of 589 Hz and 4.82 Hz to rid our signal of extraneous frequencies[2]. These cutoff frequencies were chosen because they allow for us to filter away noise and have minimal attenuation of the desired signal. In the design of this circuit, the following equation was used to determine the appropriate capacitor and resistor values to give us the desired cutoff frequencies:

$$f_c = \frac{1}{2\pi RC}$$

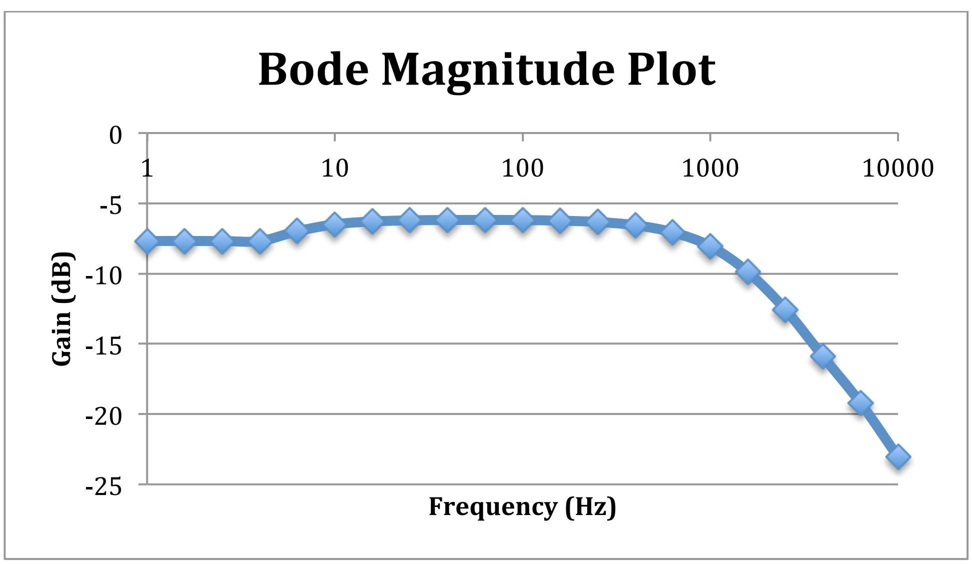

Figure 8: Bode Plot for Passive Band-Pass Filter

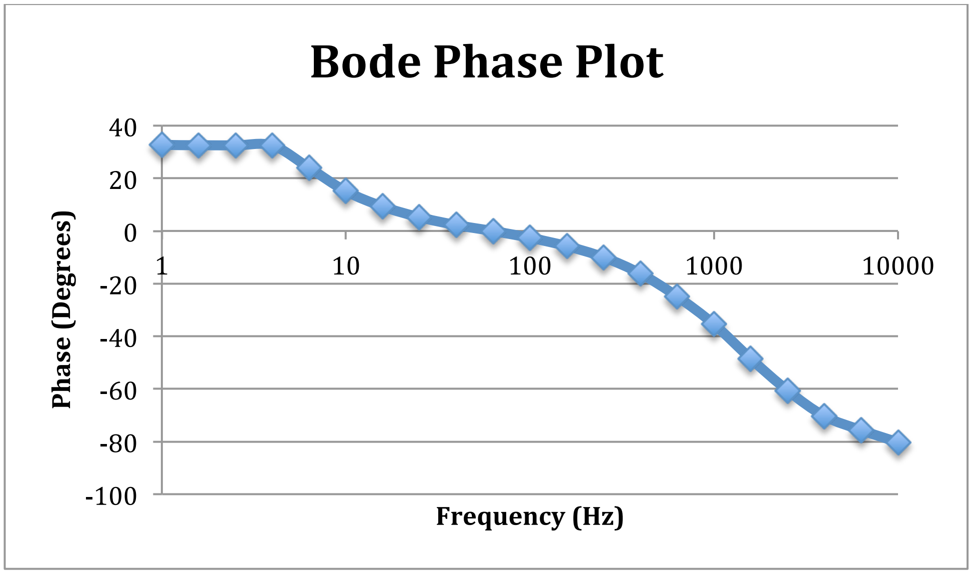

Figure 9: Phase v. Frequency Plot for Passive Band-Pass Filter

Shown above in Figures 8 and 9 are the Bode plot and phase v. frequency plot for the passive band-pass filter our team has designed. From the Bode plot we can observe a distinct lower and upper cutoff frequency as well as a fairly flat band response in the band-pass which is desired. The phase v. frequency plot demonstrates the conventionally observed phase shifts between 0 and 90 degrees. The high pass filter's phase is smaller than usual, due to the small cutoff frequency, the transition band is abnormally short.

Lastly once our signal has been filtered, it is amplified again to make it easier to analyze in LabView. Our program functions by incorporating thresholds and if the range of the signal is extremely small, it makes it much harder to detect changes within the signal via thresholds. We chose to amplify our signal with an OP07 operational amplifier set up in an inverting configuration. This simple amplification setup is ideal because it does not contribute a lot of expensive components and the gain of a single ended input is all that is necessary.

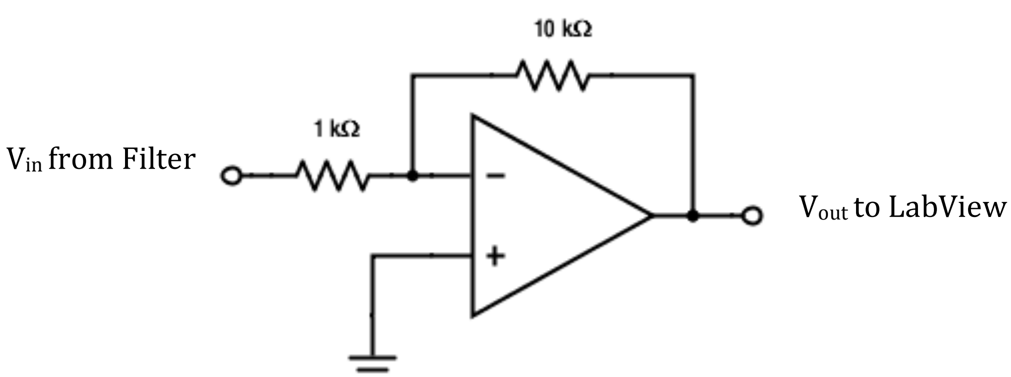

Figure 10. Inverting Operational Amplifier

After examining the collected muscle signals, we decided that a gain of 10 would give a reasonably sized signal to work with. The circuit was then set up with two different resistors whose values were determined by utilizing the following equation:

$$\frac{V_{out}}{V_{in}} = -\frac{R_F}{R_1}$$

From this equation, we derived resistor values of R_x = 1 kΩ and R_F = 10 kΩ for the inverting operational amplifier circuit.

This summarizes all of the steps that will lead to a reasonably sized and sufficiently noise free signal that can be sent into LabView for analysis.

a. Interactions Between Living and Non-living Systems

The interactions between electrical devices and humans, although mostly safe because of precautionary hardware and natural protective barriers of the body, can be dangerous. Although the skin normally has very high resistance, with strong enough voltages or a better conduction path facilitated by water, currents can end up traveling through the body. This is dangerous because currents into the body can cause many involuntary muscle movements, electric shocks and burns and electrocutions. If the patient is able to form a complete circuit with the hardware, electric shocks can occur. This allows for the current to flow through the patient and into a ground. This can have a variety of effects from tingling sensations to immediate cardiac arrest. In between there are loss of muscular control, the freezing zone (patient cannot let go of wires), respiratory arrest, and loss of rhythmic heart function. Most of these symptoms are a side effect of having large currents running through the body that disrupt normal muscle function and force muscles to contract. These involuntary muscle contractions are what prevent patients from letting go of objects, breathing properly or pumping blood properly when they are shocked.

Due to the inherent dangers of allowing interactions between living and non-living electrical systems, precautionary circuitry must be put in place to reduce the risk of electric shock. Frequently fuses or circuit breakers are implemented to prevent the passage of large currents. In our device, a big safety concern is that a malfunction in the hardware will cause large currents to travel from the device to the patient. Thus we utilized diodes to force unidirectional flow when the voltage gets too large as well as is going in the wrong direction. Safety circuits are essential to medical devices that interact with living systems because of the dangers electricity can pose on biological systems.

Signals from living systems is complicated by the many movements and other electrical signals coming from the system. The human body has many surface level movements, adjacent muscle contractions and bulk movements. These can all add noise to the signal we are trying to obtain. In addition, the usage of voltage sources in the device inevitably adds 60 Hz noise into our obtained signal. Thus when the desired signals are from a living system, the circuitry of the interacting non-living system must be modified to take into account the extra noise that will be incorporated by utilizing filters and amplifiers.

Lastly, one of the biggest challenges working with living systems is the inevitable patient variability. Each patient will have different surface noise as well as different muscle action potentials. This could pose problems for the device if not properly addressed. We tackled this problem by including a ground electrode which accounts for any noise that is not a result of muscle contraction. This is normally located over a bony area to ensure it does not pick up any muscle signals. By inputting both this ground signal and the detected muscle signal to a differential amplifier, we are able to selectively amplify for the difference between the two signals and essentially ignoring the variable surface signals, allowing for application for a wide audience. In addition, our system allows for calibration before each patient. Thresholds for action potential peak detection is then set based on these calibrations and thus can account for the different sized action potentials that might be detected from different patients.

IV. LabView VI

The LabView VI portion of our EMG is designed to acquire data from the analog signals on our protoboard, detect hammer strikes and muscle contractions, calculate reflex time, and display this data in an easy-to-use interface.

a. Front Panel

The Front Panel of the VI has three tabs: Main Display, Calibration, and Troubleshooting. These tabs can be seen in Figures 11, 12, and 13, respectively. The Main Display tab (Figure 11) prominently displays a graph of the Hammer Signal and Electrode (EMG) Signal, which records the data over time, and allows the user to scroll through past data. Additionally, a numerical indicator displays the calculated Reflex Time in milliseconds. The “Record Measurement in Table” button can be clicked in order to record whatever value is currently being displayed under Reflex Time into the Table below. The “Export Table to Excel” button can be clicked to export all the entries currently being displayed in the table into an excel spreadsheet with the column header “Reflex Time (ms).” The Calibration tab (Figure 12) simply contains two switches, which are used to perform calibration, and two numerical indicators, which display the calibrated threshold values after calibration is performed. This tab also displays instructions on how to perform the calibration. Finally, the Troubleshooting tab (Figure 13) displays three graphs; these graphs record data from the Hammer Signal, Partially Processed EMG Signal, and the Fully Processed EMG Signal. Visualizing these signals (which will be described in detail in later sections) can help users diagnose any problems they might encounter while using the device.

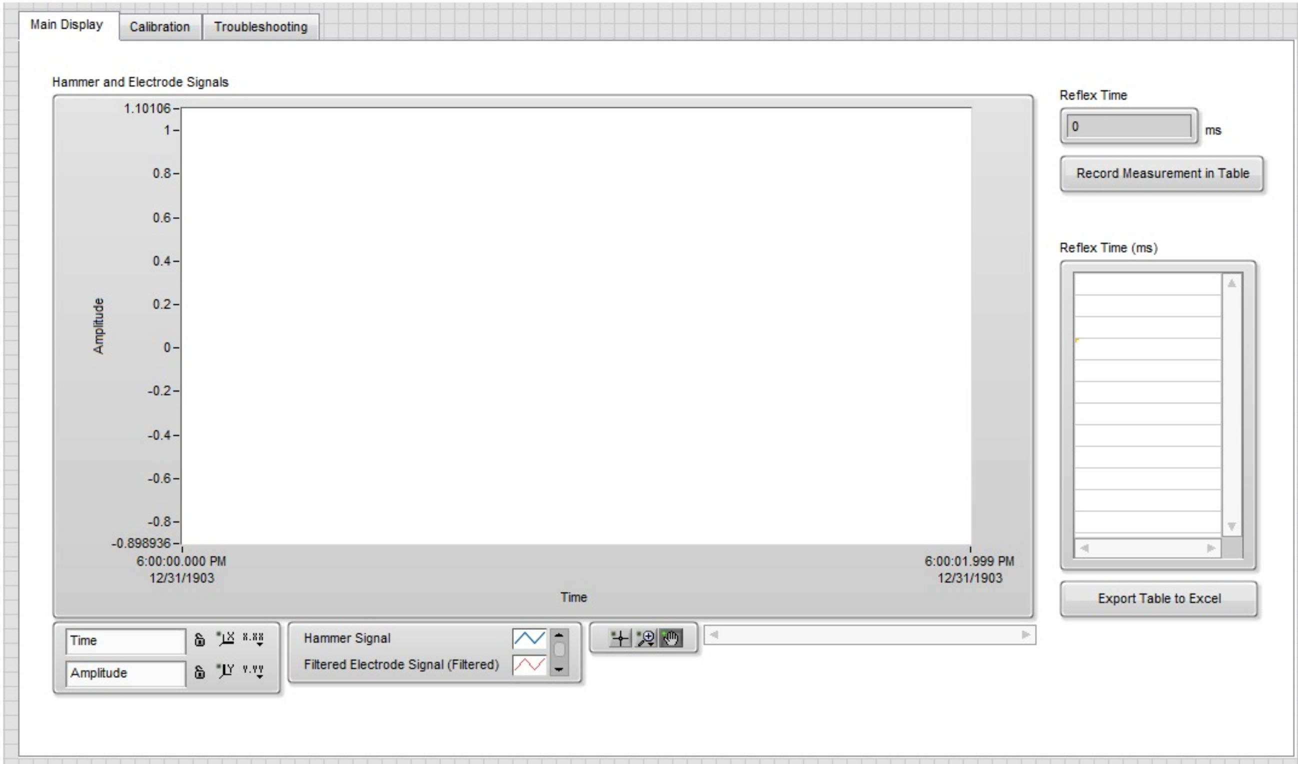

Figure 11. Front Panel: Main Display Tab

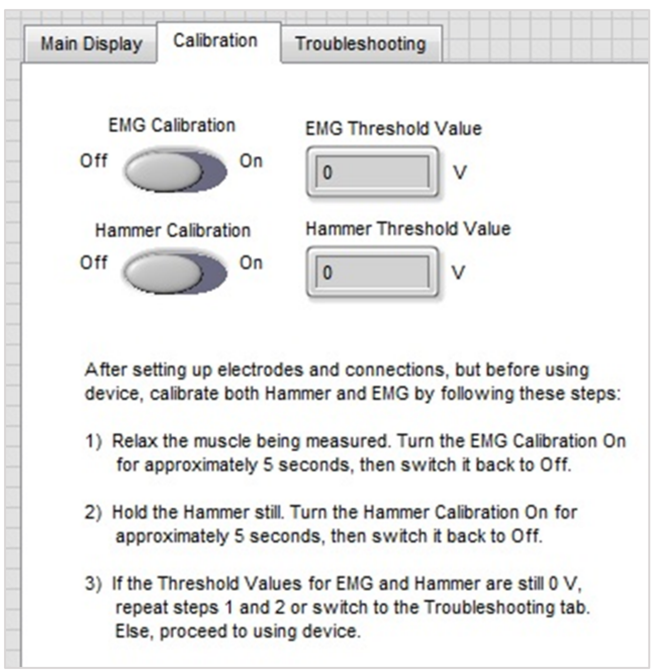

Figure 12. Front Panel: Calibration Tab

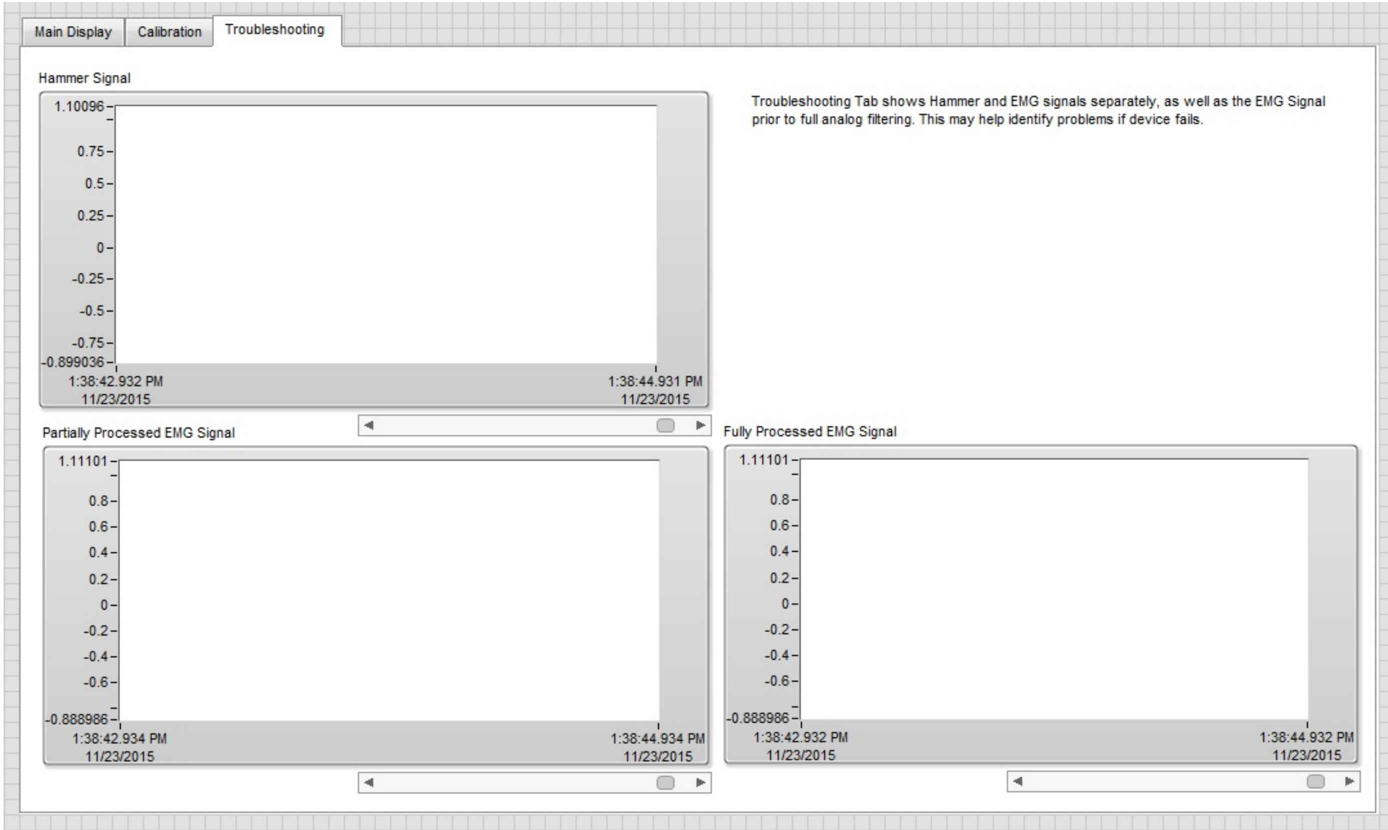

Figure 13. Front Panel: Troubleshooting Tab

b. Block Diagram

Our LabView block diagram can be functionally divided into eight main parts: Data Acquisition, Digital Filtering, Signal Display, Calibration of Threshold Values, Comparison to Threshold Values, Reflex Time Calculation, Recording Data in Table, and Exporting Table to Excel. All of these parts take place within a single while loop, which continuously runs the whole program while the On/Off switch on the Front Panel is switched to On. The eight parts are labeled on an image of the full VI Block Diagram in Figure 14. Although the details of the diagram are not discernible here, more explanation and larger, clearer images of each part will be presented in the following sections.

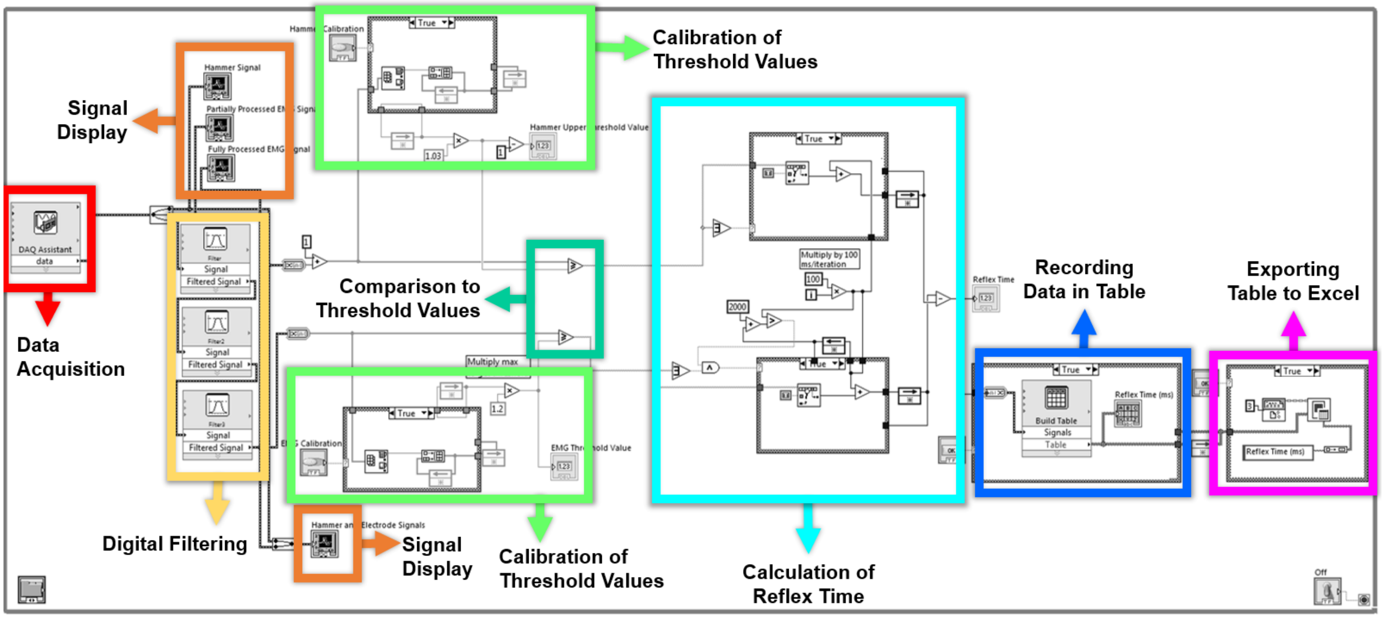

Figure 14. Block Diagram Overview

c. Data Acquisition

The first component of our Block Diagram is the DAQ Assistant sub-VI, which collects three channels of voltage data from Ach0, Ach1, and Ach2 on the protoboard. Ach0 transmits data from the hammer signal; Ach1 transmits data from an unfiltered and partially amplified EMG signal; Ach2 transmits data from the fully processed EMG signal. These signals are sampled at 1 kHz frequency, with 100 samples read at a time. This means that, for each channel, the samples are taken at 1 ms intervals, but are sent out from the DAQ as dynamic data arrays with 100 samples per array. In other words, the while loop runs once every 100 ms, and on each iteration, the DAQ assembles three arrays (one from each channel) of 100 samples each to be processed and interpreted by the program.

d. Digital Filtering

After the DAQ Assistant, the sampled data signal is split by a Split Signals (as can be seen in Figure 15), into the three different sampled channels. The third channel, which contains the data from the EMG signal after full analog processing, is sent through a series of digital filters. This is necessary due to the inherent noisiness of physiological signals and the presence of 60 Hz noise in the signal. These filters can be seen in figure 15.

The first filter is a band stop filter with cutoff frequencies of 55 Hz and 65 Hz, designed to eliminate 60 Hz mains noise. The second filter is another band stop filter, with cutoff frequencies of 115 Hz and 125 Hz. This was intended to attenuate any noise from the harmonic of the 60 Hz noise at 120 Hz. Finally, the third filter is a band pass filter, with cutoff frequencies of 5 Hz and 500 Hz. The EMG signal has already gone through an analog band pass filter with similar cutoff frequencies on the protoboard, before being acquired by LabView. This filter is simply intended to reinforce the analog filter.

The hammer signal was sufficiently clear from the analog processing, and therefore additional digital filtering was deemed unnecessary. The partially processed EMG signal from Ach1 is used only to display for troubleshooting purposes; since viewing noise in this signal could be useful for troubleshooting, no digital filters were used on this signal.

e. Signal Display

Our LabView Front Panel contains four charts, which are used to display various signals. One chart is on the Main Display tab and is used for viewing the hammer and EMG signals together, so that the reflex time can be visualized graphically. The other three charts are on the Troubleshooting tab; each chart shows one of the three channels of data (Hammer Signal, Partially Processed EMG Signal, and Fully Processed EMG Signal) and can be used to help identify the sources of any problems the device may have.

On the block diagram, the hammer signal, unfiltered and partially amplified EMG signal, and the digitally-filtered fully processed EMG signal are sent individually to separate charts for the Troubleshooting tab. The hammer signal and fully processed EMG signal are re-merged into one "wire" and sent to a single chart for simultaneous display on the Main Display tab. This can all be seen in figure 15.

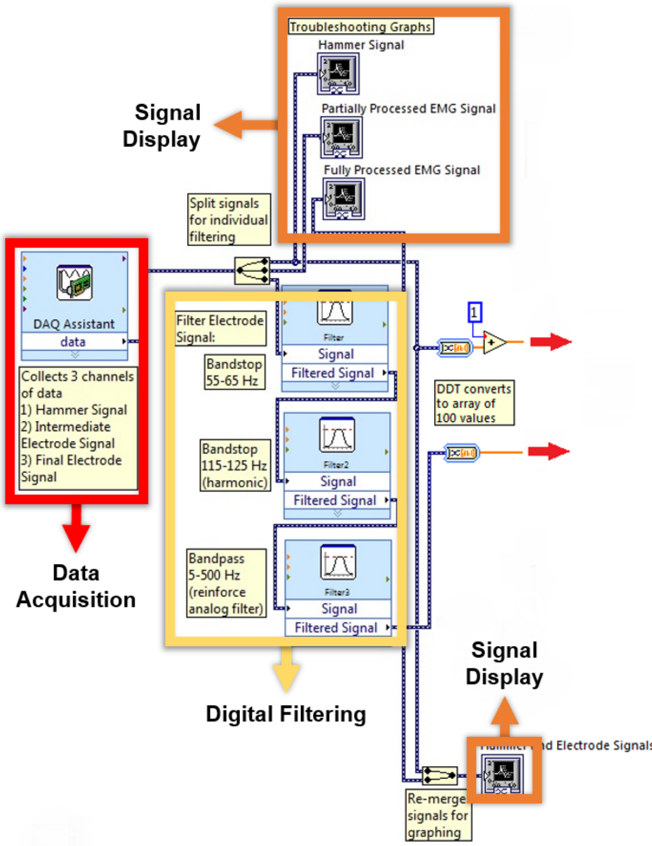

Figure 15. Block Diagram: Data Acquisition, Digital Filtering, and Signal Display

After the signal display, but before any other processes begin, the hammer signal and the EMG signal are converted from dynamic data to an array of values. These are simply two different data types recognized by LabView; dynamic data is produced by the DAQ Assistant, but arrays are useful for calculation purposes. Additionally, one is added to every sample of the hammer signal to serve as essentially an DC offset to make calculation of the threshold easier. This was implemented because a the hammer signal falls within the range of about -0.2 to -0.01 V. In order to properly set a max threshold, it was easiest to shift this data upward by 1 V to make all the data positive. These components can be seen on the right side of figure 15.

f. Calibration of Threshold Values

The calibration process is used to calculate threshold values for both the EMG and hammer signals; these threshold values are used to determine when a strike has occurred in the hammer signal and when a contraction has been detected by the EMG signal. These events are detected when the signal's voltage level increases above a certain level. However, different hammers may have different levels of electrical activity while at rest, and different muscles on different subjects will have different levels of activity at rest. The calibration process is designed to take these differences into account and calculate an appropriate threshold value for optimal device function for different hammers and subjects.

The calibration process begins when the user goes to the Calibration tab in the Front Panel and clicks one of the calibration switches to On. Directions on this tab instruct the user to leave the switch on for approximately five seconds and then turn it back to Off. There is one calibration switch for the hammer and one for the EMG. The calibration process works the same for each, but they are performed separately so that during the hammer calibration, the user can hold the hammer still, and during the EMG calibration, the user can hold the muscle still.

During the time that the calibration switch is turned on, a case structure on the block diagram (pictured in Figure 16, below) is switched to true. On any given iteration of the while loop, if this structure is true, it will take the array of 100 samples from the signal being calibrated and find the maximum value within it. That max value is then put into a new array. On the next iteration, if the structure is still true, it will again find the max value from the current iteration and put that value into the array with the max value from the previous iteration. This will continue for as many iterations as the switch is turned on, with the array of max values gaining a new element at each iteration. When the switch is turned off, and the case structure switches to "False," the maximum value from the array of max values is found and sent out of the case structure. This value is the maximum voltage sampled from the signal during the time the switch was on. Therefore, the baseline signal (while the hammer or muscle is at rest) should never get much higher than that value. The threshold value is calculated by multiplying the max value by a number (1.03 for hammer calibration; 1.2 for EMG calibration). This threshold value is then displayed on the Calibration tab of the Front Panel, and passed on to the next part of the VI for comparison. Note that the hammer threshold value that is displayed is shifted back down (equal to calculated threshold value minus one), for the user's convenience.

g. Comparison to Threshold Values

In order to determine the time of a hammer strike or muscle contraction, the VI must determine which sample voltage values are greater than the threshold. This is accomplished simply by wiring the array of sampled data and the threshold value into a greater than or equal to Boolean operator. The output from this operator is a Boolean array of 100 elements. Each element is either true or false: True if the corresponding element of the data array was greater than or equal to the threshold, and False otherwise. These two Boolean arrays are passed on to be used for Reflex Time Calculation. This can all be seen in Figure 16.

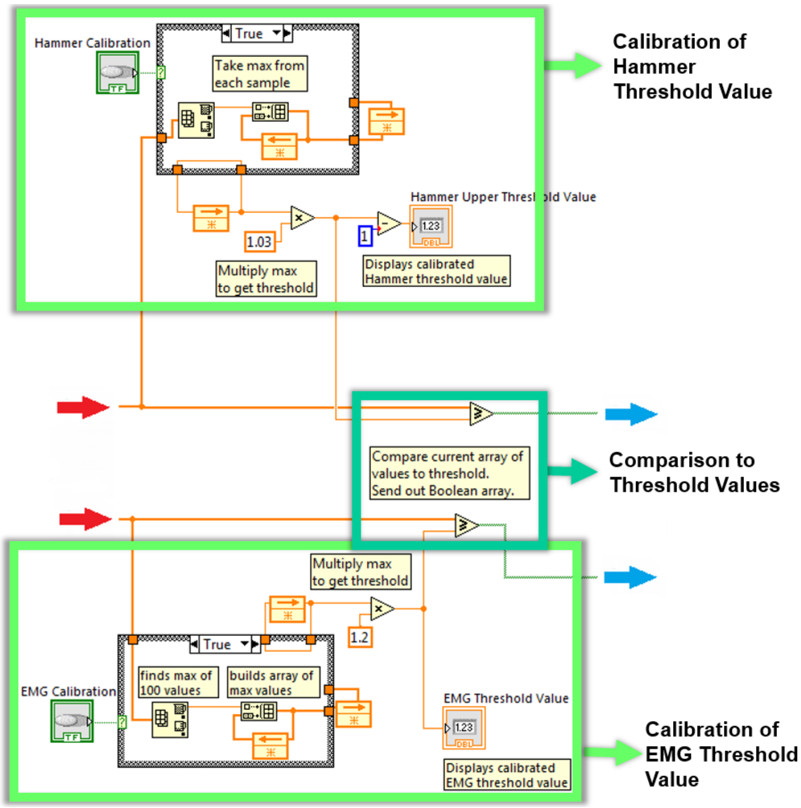

Figure 16. Block Diagram: Calibration of Threshold Values, and Comparison to Threshold Values

h. Reflex Time Calculation

The Reflex Time Calculation utilizes two case structures: one for determining the time of the hammer strike, and the other for determining the time of the muscle contraction. The hammer signal's case structure is controlled by a Boolean operator that returns true if the hammer signal's Boolean array has any trues in it, or false otherwise. Essentially, when the signal goes above its threshold, the case structure will be switched to true. When the case structure is true, it will take the Boolean array and search for the first element that is true. It will then take the index number of that element (which is equal to the number of milliseconds since the start of the iteration at which the sample was taken) and add to it the iteration number times 100 (since each iteration takes 100 ms, this value is the number of milliseconds since the start of the program). The result of this calculation is the time (in milliseconds) of the hammer strike. The resulting number is sent out of the case structure and looped back through the false case, so that even after the peak has passed, the same time value will still be sent out of the structure.

The calculation for the EMG signal is slightly more complicated because a muscle contraction results in several peaks in a row, while the hammer strike generally only causes a single peak. Therefore, it was necessary to make the program stop looking for peaks in the EMG signal for a short period after detecting the first peak. The EMG case structure was switched to true in response to two conditions and an "And" Boolean operator. The first condition was the same as for the Hammer Signal (any true values in the array returns true). The second condition checks if it has been over 2 seconds since the last time it was switched to true. We check for this condition by first looping the iteration time (iteration number times 100) obtained during the true case to the false case. Thus, when it switches to "False," the same iteration time will be sent out until the next time the case structure switches to true. With each iteration, the current iteration time is compared to this outputted iteration time plus 2000 milliseconds. If the current iteration time is greater, it returns true; if not, it returns false, and the case structure is forced to remain "False" for that iteration. This allows for us to delay the activation of the true case by 2 seconds because so long as current iteration time is less than outputted iteration time plus 2000 milliseconds, we would be able to maintain the case structure in the false state. The rest of the operations within the EMG (contraction-detecting) case structure are the same as that in the hammer (strike-detecting) case structure.

Once both of these case structures have been momentarily switched to "True" (detected a strike or contraction), they each send out a value for the time (in milliseconds) of their respective detected events. These times are subtracted to obtain the Reflex Time. This Reflex Time is sent to a numerical indicator to be displayed on the Front Panel's Main Display, as well as to the Recording Data in Table part of the block diagram. All of this can be seen in figure 17.

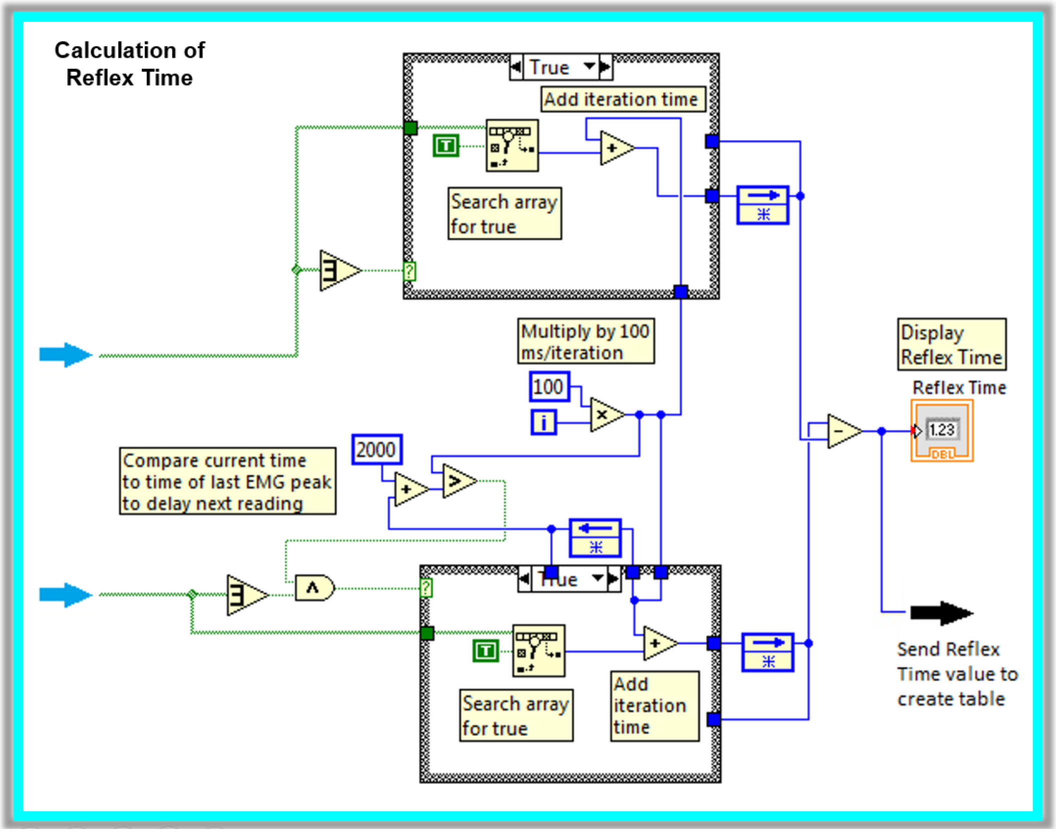

Figure 17. Block Diagram: Calculation of Reflex Time

i. Recording Data in Table

As an additional feature of this device, users can record their measured Reflex Time values into a table for viewing and saving. A case structure (activated by a button labeled "Record Measurement in Table" on the Main Display tab of the Front Panel) is used to grab Reflex Time values at the time that the button is pressed and build those values into a table to be displayed on the Main Display. When the case structure is "False", the same table is being looped around and sent out of it. This can be seen on the left side of figure 18.

j. Exporting Table to Excel

The table being sent out from the table-building case structure goes into another case structure, which contains 2 sub-VIs; one of the sub-VIs creates a blank report, and the other exports reports to Excel. When the button labeled "Export Table to Excel" is pressed, the case structure receives the table from the previous case structure and exports it to an excel document, complete with the column header "Reflex Time (ms)". This excel document can then be saved so the subject data is not lost.

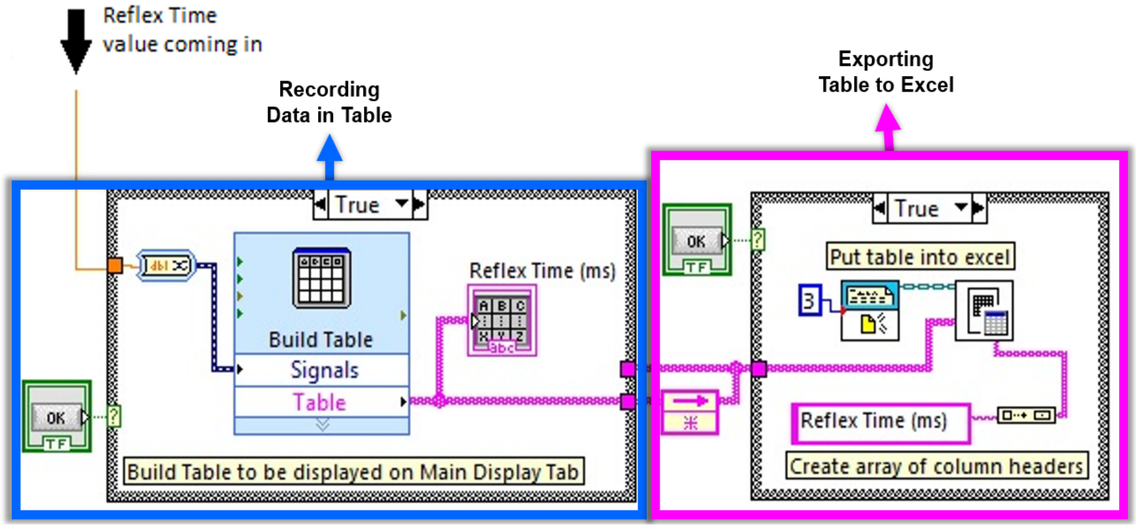

Figure 18. Block Diagram: Recording Data in Table and Exporting Table to Excel

REFERENCES

1. Day, Scott. "Important Factors in Surface EMG Measurement." *Important Factors in Surface EMG Measurement By Dr. Scott Day* (n.d.): n. pag. *Bortec Biomedical*. Web. 21 Nov. 2015. <http://www.bortec.ca/Images/pdf/EMG%20measurement%20and%20recording.pdf>.

2. "JAT Standards for Reporting." *Journal of Electromyography and Kinesiology* 4.2 (1994): 126. *NATA*. National Athletic Trainers Association, 1996. Web. 29 Nov. 2015.