Network of Wireless Medical Devices to Assess the Gait of Rehabilitation in Patients for Walking and Running

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***Abstract*—In this paper, we present the design of two smart sensor systems to monitor the gait of patients. These sensor systems were developed for deployment within a body worn wireless network system of medical devices. Telemetry, ambulatory and remote monitoring systems composed of micro-mechanical systems have gained importance in the last decade as medical and rehabilitation institutions try to reduce costs by discharging patients earlier while still requiring various levels of monitoring. Most of the systems currently on the market are bulky, closed architecture, static in configuration and use wired medical devices, all of which limit their usage. Gait monitoring is mainly done in laboratories that are fixed and expensive. The aim of the research which encompasses both systems discussed in this paper is to develop an open architecture using Real-Time Object Oriented Modeling that will allow wireless, wearable medical devices to join a dynamically configurable monitoring environment. The intent of the system is to monitor patients recovery by measuring biometrics and biomedical signals as they go about their daily activities. The sensors that are being developed as part of this research are smart sensors that can provide pre-processed information, reducing the load on the wearable computer.**

I. Introduction

Our research project is to develop an ambulatory medical telemetry system that will be light, non-intrusive and flexible and allow many different kinds of sensors to be added or removed from the system with minimal effort. The system will allow health care professionals to use daily activities of the patients as part of the rehabilitation plan.

This part of our research originally was centered to assist physicians and health care workers in rehabilitation to monitor the gait of soldiers after orthopedic surgery. A career in the Army subjects the body to high impact and prolonged physical stress, especially on the lower body due to activities such as marches with full combat order, obstacle courses and combat fitness tests. Lower body injuries are the most common injury among soldiers and veterans, but these injuries are also prevalent in sports and other highly physical occupations such as construction or landscaping. While these injuries can vary in severity (some only requiring physical rehabilitation and therapy, others requiring surgery), extensive recovery and kinesiology is usually required.

In order to monitor the progress of recovery, empirical data should be used to monitor and quantify improvement or regression and designate if a patient has recovered enough to return to regular activities. While gait measuring systems currently exist, none are portable, and all involve specialized equipment. This specialized equipment limits access to hospitals in major centres. Those systems are very expensive and require the facilities housing these systems to maximize their use through scheduling and coordination with physicians. A portable gait monitoring system allows the technology to be taken to remote areas at a low cost, giving more patients access to the system.This type of disparity in the quality of health care away from major centres is one of the central themes of the Romanow report on the future of health care in Canada [1].

In a previous paper, our research group presented the preliminary analysis for the Health Activity Assessment System (HAAS) in the form of use cases that elaborate the requirements of the proposed HAAS. We have also presented two systems to monitor breathing and range of motion of a knee [2]. In this paper we present two systems to analyze two different components of gait analysis, that of identification of abnormal gait while walking and while running. Both present different diagnosis, challenges and although originally developed for different purposes, they combine to provide a system-of-systems. The two gait analysis systems presented in this paper were designed to be integrated into Health Activity Assessment System using the addressing scheme of up to sixteen sensor system developed in [2].

II. Background

A person's gait is described as the way in which a person moves while moving whether it be walking or running. Studying a person's gait will allow for a diagnosis as to whether they are using a proper gait or not. Understanding the biomechanical aspects and the overall process of running is vital to the correct identification of gait. Walking and running gait are different and present various kinds of anomalies.

There are currently several popular ways in which the gait of a person is analyzed. The first is using a pressure plate on the ground to measure the impact of the foot with the

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ground. Although this is an effective method to measure foot strike it also presents several issues. The first issue is the accessibility of such testing centres; due to the immobility of this system only certain research facilities are able to offer this technology. The second issue with this test is the occurrence of an expectation bias. This illustrates an obvious issue because as an individual approaches the plate on the ground they tend to either exaggerate or lessen their issues based on what they subconsciously want to achieve and subsequently skew the results.

A second technology that is present in gait analysis is the use of video analysis. The aforementioned pressure plates are sometime present in a hybrid lab that is also equipped with this video equipment that will produce a three dimensional model of the runner. However, the video analysis does become subjective as it is limited by the knowledge and experience of the health care professional that reviews the video [3].

A. Walking Gait Monitoring

Human walking gait can be divided in three phases: the loading phase, the stance phase and the swing phase [4]. The loading phase is defined as the acceptance of the weight from the trailing to the leading limb. The stance phase is when the leading limb bears the body's weight. The swing phase is when the leading limb, now trailing, is not touching the ground as it swings back to the leading position.

A lower limb injury is referred to as a peripheral musculoskeletal problem in the field of gait classification. There are four walking gait impairments that are caused by peripheral musculoskeletal problems (along with their characteristics) [4]:

- Arthritic gait
  - Excessive subtalar joint eversion motion that manifests itself in stance phase in patients with pes planus, arthritis and myelomeningocele. The gait motion is coupled where the internal rotation of the leg relative to the rear foot is visibly abnormal.
  - Caused by chronic synovial inflammation and progressive erosion of cartilage and bone have been described for the tibiotalar, subtalar and midtarsal joints in rheumatoid arthritis. the clinical manifestations is seen through as valgus deformation of the rearfoot, *usually accompanied by medial longitudinal arch collapse, and this is estimated to occur in 67% of patients with arthritis* [5].
- Antalgic gait
  - An abnormal gait that is the results from pain on the weight-bearing leg/foot in which the stance phase of gait is shortened on the affected side leading to a shortened stance phase relative to swing phase.
  - Caused by the patient wanting to avoid or counteract the pain in the affected leg. *Ground reaction force measures have been shown to be an effective means of quantifying the antalgic gait of hip arthroplasty patients.* [6]. Such pain can generate from back, hip or knee injuries.
- Myopathic gait (or wadding gait)
  - Pelvis on contralateral side of the pelvis drops during the swing phase. Can be observed by an unsteady gait, weakness and pain when climbing stairs [7]
  - Caused by weakness in proximal muscles of the pelvic girdle. In patients in need of Total Hip Arthroplasty the weakness stems from the gluteal abductor muscle [8]. This condition is normally associated with pregnancy [9], hip dysplasia, spinal atrophy and muscular dystrophies.
- Peripheral neuropathic gait
  - Attempt to lift leg high enough during walking so that the foot does not drag on the ground. *NP subjects also felt significantly less safe during standing and walking in unusual conditions, such as low light or unfamiliar terrain.* [10]
  - Caused by weakness of foot dorsiflexion. This can occur through traumas, such as from motor vehicle accidents, falls or sports injuries, can sever or damage peripheral nerves. Nerve pressure can result from having a cast or using crutches or repeating a motion such as typing many times. [11]

The aim of the walking gait monitoring system is to use existing electronic sensors to design, construct, test and evaluate a prototype for a system of sensors and software to record and analyze the gait of a patient. The device acquires data from the sensors, with the possibility of interpreting the data to compare it against a control limb from the same or similar patient. The data will be gathered from a mix of strategically placed biomedical sensors along the injured and uninjured limbs. The data will be transmitted to a microcontroller to be worn by the patient.

In this first version of the walking gait monitoring system, we characterize an irregular gait. The characterization of an irregular gate was tested and validated in our lab on multiple volunteer human subjects simulating the four gait disorders specified above. The system does not aim at fully documenting a subject's gait through tracking positions of the various limbs and joints. We found that position provides minimal data. As such, the system concentrates on rate of change and with the help of rehabilitation health care professional identify what is in the range of normal and what is abnormal.

B. Running Gait Monitoring

An unfavorable running gait that is used regularly can lead to a wide array of injuries in the long and/or short term. Injury incidents among runners is quite common, studies show that the amount of injured runners can be as high as 85% [3]. These injuries are caused by prolonged or over usage as well as slight biomechanical abnormalities [12]. Due to the prevalence of such issues there is an increase in analysis of the causes and effects of these biomechanical abnormalities.

Studying a person's gait will allow for a diagnosis as to whether they are using a proper gait or not. Understanding the biomechanical aspects and the overall process of running is vital to the correct identification of gait. Each injury that occurs from running has a different profile that is able to be identified using the aforementioned analysis methods. For the purpose of this project the profiles of four common running injuries will be examined Iliotibial Band Syndrome (ITBS), Patellofemoral Pain Syndrome (PFPS), Achilles Tendonitis (AT), and Medial Tibial Stress Syndrome (MTSS).

ITBS and PFPS are both characterized by pain that occurs at the knee both during and after running. The knee is a problem area for running injuries as it is the most common injury site due to the forces and moments that occur [13]. However, the two injuries differ in their gait and can be detected by monitoring and analyzing knee motion. The Iliotibial (IT) Band serves to stabilize the lateral motion of the hip and knee as well as resists adduction and internal rotation. However, when ITBS is present the IT Band tightens and the profile of one's gait will have an adverse effect as the motion about the hip and knee will contain more abduction [14]. Similar to that is the motion when a runner is experiencing PFPS which is caused by the weakness of the hip abductors as opposed to the IT Band. This profile of a weakened hip causes a lack of control and will result in a more sporadic gait with regards to lateral knee movement [15].

Both AT and MTSS can be attributed to the striking pattern that one produces while running. Where the two differ is in the length of stride produced throughout the running cycle. AT is often examined using rear foot and ankle kinematics; however, it has been shown that when analyzing AT a lengthened stride will result in a more extended knee. This over extension of the knee will place tension on the Gastrocnemius and Soleus complex which transfers its stress to the Achilles Tendonitis [16]. Conversely, a shortened stride will act as an interruption in momentum and increase the force that is observed throughout the body. MTSS is often caused by a increased loading rates and large tibial shocks; both of which can be attributed to the effects of a shortened stride [17].

C. *Summary*

Although both systems could be used to analyze walking and running gait, they would either provide partial results or would require additional sensors and software. The purpose of both systems are distinct and require different instrumentation, telemetry and analysis. Running is more demanding when tracking the motion of limbs, hence the need for a more precise system and data gathering process.

In the following two sections we introduce the architecture and design for each of the smart sensor systems.

III. System Designs

A. *Walking Wireless Gait Analysis Monitor (WWGAM)*

The system is integrated into a knee brace and inside running shoes and will use a self-contained power system in order to enable maximum portability. The patient being monitored must be able to live a regular life without having his/her movement unduly impaired. The microcontroller, power supply and user interface will all be self-contained in

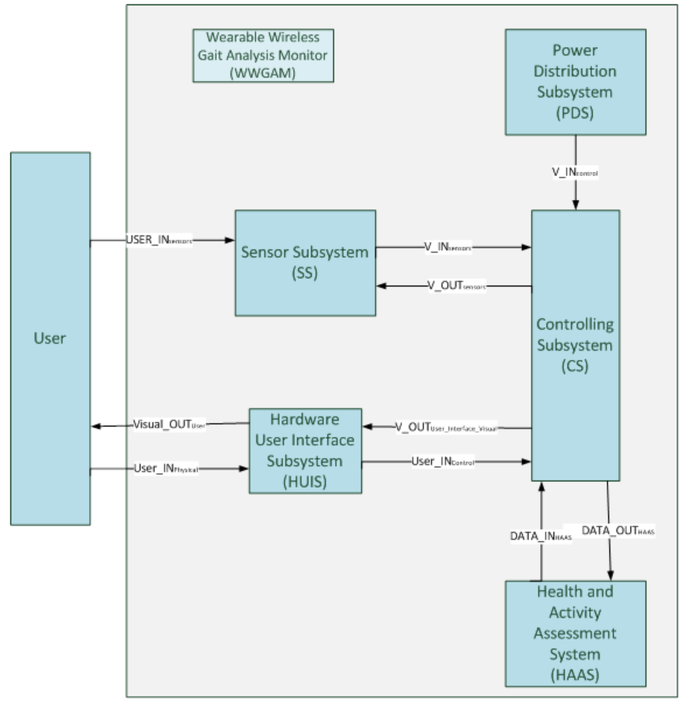


Fig. 1. Walking Wireless Gait Analysis Monitor Top Level Block Diagram

a carrying case in order to be easily worn. A top level block diagram of the system can be seen in Fig 1.

There are four buttons on the Hardware User Interface Subsystem (HUIS) seen in figure 1 to select each mode of operation. In calibration mode, the WWGAM iterates through all seven sensors 100 times and finds the maximum, minimum and average values of each of the sensors without any weight on them and at complete rest. A hysteresis is then applied to the unloaded calibrated value which was found for each sensor type through experimental calibration trials. The range of values between the maximum and minimum value are all considered to be 0 for the following iteration of data gathering. The base, maximum and minimum values are printed out at the end of the calibration routine as well.

The data processing mode takes place in two phases. In the first phase, once the patient or clinician has pressed the button, the system gathers data from the sensors on the uninjured limb. When printing out the values on the display screen, if they are not between these maximum and minimum values from the calibration routine, the average value is subtracted from the value returned by the sensor as this average value is considered to be the 0 value for that sensor.

Data is gathered ever 10 mSec. After the system has iterated through gathering data for each of the seven sensors, the analyzeSensors() function is called, where the data goes through processing.

After the analyzeSensors() function is called the loadRange, XMax, YMax and avgStepTime values are all saved for results processing. Instructions to put the system on the injured limb are printed. Once the user is ready, he presses a button to start data collection again. The values of highLoad and loadRange, firstTime, step, XMax and YMax are initialized to 0 in order to allow data to be gathered for the injured leg.

During the processing mode, the number of steps taken, the number of times the patient shuffled, the loadRange, X and Y max values of both the injured and uninjured limbs as well as the average stance and swing phase of the injured limb are printed. Then, the system runs the results through four tests. If the results fail any of these tests, a message with the anomaly detected is printed. If not, a message saying that the gait passed all tests is printed.

If the value returned from the load sensor is higher than the previous highest value, the highLoad value is updated, as well as the loadRange value. As deternined through experimentations with the sensors, the loadRange value is 2/3 of the value of the highRange value. This value was determined through experimentation with the load sensors to identify when a foot strike or the start of a loading phase occurs. The load phase of gait is characterized by a rapid heel strike for longer than 20 mSec. If the loadRange value is surpassed for two iterations in a row, this is considered to be a step and a message is printed. The time between steps is also recorded by comparing the number of iterations it has been since the last step.

Shuffling, or dragging the feet against the ground is characterized in a similar way. If a value 20 more than the maximum resting value of the sensor is recorded 6 times in a row, shuffling is detected, and a message is printed. The number of times the patient has shuffled is also recorded.

Finally, the maximum acceleration on both the x and the y planes are recorded. However, motion artefact is a considerable risk when using accelerometers close to the skin. In order to eliminate the possibility of false data interfering with results, the average of both sensors is taken. These values have the potential to be updated every iteration.

After the analyzeSensors() function is called, the values of highLoad and loadRange, firstTime, step, XMax and YMax are put back to 0 in order to allow data to be gathered for the injured leg. Before this, the loadRange, XMax, YMax and avgStepTime values are all saved for results processing. Instructions to put the system on the injured limb are printed. Once the user is ready, they press button 2 to start data collection again.

During the processing mode, the number of steps taken, the number of times the patient shuffled, the loadRange, X and Y max values of both the injured and uninjured limbs as well as the average stance and swing phase of the injured limb are printed. Then, the system runs the results through 4 tests. If the results fail any of these tests, an message with the anomaly detected is printed. If not, a message saying that the gait passed all tests is printed.

While we could have calculated and calibrated for bad gait and good gait, we measured empirically the values that would qualify a bad gait from a good one. If the difference between the loadRange of the uninjured and injured limbs is found to be more than 20, uneven loading was detected. If the swing phase was average of 100mSec longer than the stance phase, a shortened stance phase was detected. If the difference between either the XMax of YMax values between the legs was more than 10, then a lateral or medial anomaly was detected. Figure 1 shows the top level block diagram for the WWGAM system. The following quickly defines each of the blocks:

- The Sensor Subsystem (SS) is comprised of the sensors used for measuring a patient's gait. The SS includes a Motion Monitoring Sensor Module (MMSM). The MMSM is composed of a load sensor (SFE Load Sensor) that records a heel's contact with the ground and two high accuracy accelerometers (LilyPad accelerometer ADXL335) that are place above and below the knee and can accurately senses and measure a patient's stride length, walk speed and lower limb acceleration in three axis.
- The Power Distribution System (PDS) provides two level of powers 5 V and 12 V DC to power the Controlling Subsystem (CS) and drive the sensors and hardware interfaces.
- The HUIS acts as a hardware interface between the user and the CS. It allows the passage of information from the CS to the user visually, through LEDs and allows the user to send control inputs to the CS, through hardwired button inputs. The HUIS consists of:
  - Visual Output Module (VOM)
  - User Input Module (UIM). The UIM consist of four buttons:
    - Button 1 Calibrates the system and zeros all the readings;
    - Button 2 Starts and stops testing mode;
    - Button 3 Starts processing mode;
    - Button 4 Shuts off the system.

  When data needs to be displayed to the user, the acVOM-receives a signal from the CS. Based on this signal, the VOM visually displays information to the user, through LEDs. For example, when the button for testing mode is pressed the CS simultaneously turns on the corresponding LED. When the user needs to send control inputs to the CS he presses the inputs button, the UIM interprets the meaning of a button pressed by the user, and sends a control signal to the CS. Users can obtain their information on measurements by pressing another button on the HUIS which in turns changes to mode of the system to data display.
- The Controlling Subsystem (CS) is the brains of the entire Walking Wireless Gait Analysis Monitor operation. It controls what is happening within the system, and when it happens. To perform its duties, the CS is composed of:
  - System Control Module (SysCM)
  - Sensor Control Module (SensCM)
  - User Communication Module (UCM)
  - Data Transfer Module (DTM)

  The SysCM is software that exists within the micro-controller. It sends and receives a variety of signals

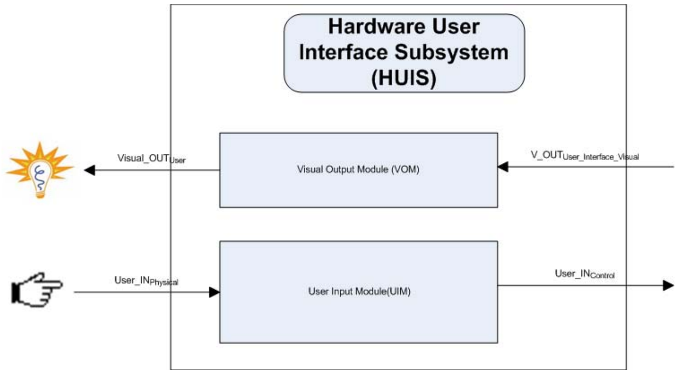


Fig. 2. Hardware User Interface Subsystem Top Level Block Diagram

from different modules in order to coordinate the smooth operation of the WWGAM.

The SensCM acts as a hybrid hardware/software interface between the SysCM and theSS. When polling the sensors for data, the SensCM sends signals out to the SS and then receives an input signal back. The SensCM also identifies to the SysCM which sensor it is communicating with.

The UCM acts as a hybrid hardware/software interface between the SysCM and the Hardware User Interface Subsystem HUIS. The UCM sends signals to the HUIS when certain data needs to be outputted visually to the user, and receives control inputs from the HUIS.

The DTM acts as a hybrid hardware/software interface between the SysCM and the Health Activity Assessment System.

The system has four modes: calibration, data gathering, processing and shutdown. User control of the system is achieved through the use of buttons and LEDs. These buttons are located within the HUIS module show in figure 2. Each button has a corresponding LED to show the user what mode they are in. The buttons allow the user to switch between modes and the respective LED lights up in order to display which mode the system is in.

Data processing is done through the use of the MC9S12DP256C microcontroller. In order to record data, the microcontroller polls the sensors and record gait data from the array of sensors. It then runs a data analysis function to process the data and stores data for the later more detailed processing. It relies on a port timer and the saturation of the load sensor and accelerometers to effectively record a patient's gait. The RS232 serial port is used to send desired gait data to the Health Activity Assessment System (HAAS) base-station.

B. Runners Gait Analysis System (RGAS)

Throughout the conceptualization of the Runners Gait Analysis System many assumptions had to be made in order to determine the final design of the system. The choice to go with four sensors (Sparkfun Accelerometers 3D- ADXL335) was made based on the assumption that this configuration would be sufficient to collect the necessary data needed to determine the running gait of the patient/runner. Studies suggest that sensor placement is key to providing accurate output and should be placed on the area of interest and should be able to observe motion on all three planes [3]. The placement of these sensors determined the number that was going to be used; on each leg there would be one sensor just above the knee that will measure the articulation of the hip joint and the movement of the thigh. The second sensor on each leg will be placed on the shin just above the ankle and will measure the range of motion of the knee. These two measurements will give sufficient data that will be analyzed.

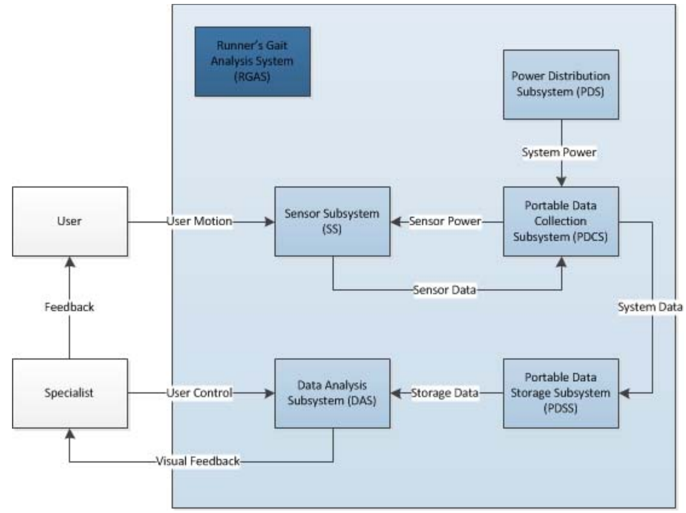


Fig. 3. Runners Gait Analysis System Top Level Block Diagram

Figure 3 shows a breakdown of the system and its different components. As well, Figure 3 illustrates the interfaces between the different components. The system-based diagram begin to illustrate the overall flow of the system and build upon the conceptualization that has been described thus far.

The Data Analysis Module (DAM) is composed of multiple java classes and methods. The architecture of this module is a pipe and filter architecture, we found that this architecture was the best for this kind of analysis based on comparing architectures as discussed in [18]. This means that the data that passes through the DAM goes through multiple filters which are described next. This architecture of the DAM is displayed on Figure 4. Each filter in the pipe and filter is described briefly below:

- *AxisSplit* filter takes the raw output from the 3D accelerometers readings and splits the data in their respective axes. This is a parsing operation that creates an array of strings of data for each axis.
- *CleanArray* filter removes null strings if the data collection did not fill the entire arrays of strings so that what is left is data from the gait of the subject.
- *ToFloat* converts the raw data to arrays of floats.
- *Fast-Fourier Transform Low-Pass Filter* is used on the resulting array of floats in order to eliminate all the noise on our signal to obtain a smooth curve the algorithm

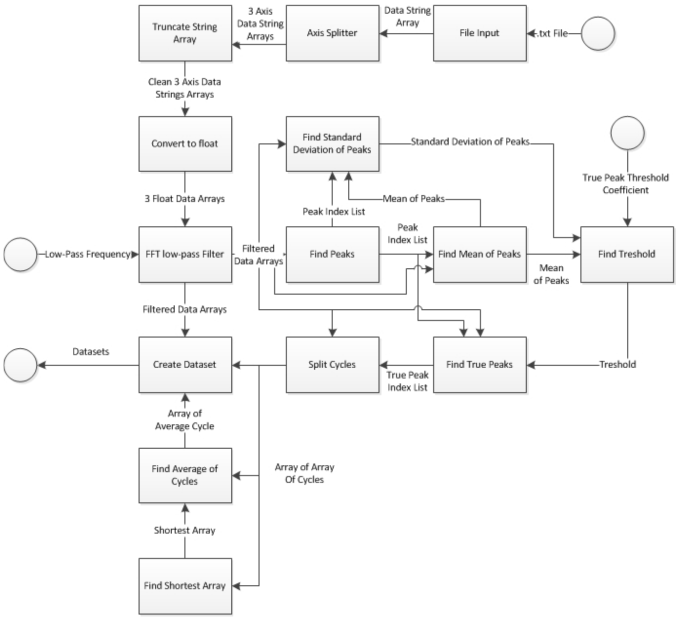


Fig. 4. Data Analysis Module Pipe-and-Filter Architecture

was chosen because our signal has a high frequency, the FFT filter removes the high frequency noise, leaving only the true signal. Other algorithms like Savitzky-Golay and adjacent averaging work well when the noise is normally distributed, but in our case, the noise is greater on localized spikes. We create a dataset from this filtered signal that is be sent to the Visual Output Module.

- *FindPeaks* finds the true peaks so we can split the signal into its different cycles. The algorithm in the filter needs a float array as input and returns a list which contains the indexes of all the peaks in the array.
- *FindMean* helps us we can find the mean of the peaks using the index and the amplitude of those peaks and the filtered array. -item*FindStdDev*: Using the mean, the Peak Index List as well as the float array the system can find the standard deviation of the peaks with the FindStdDev filter.
- *FindTruePeaks*: With all this data that is transformed by each stage of the pipe and filter the health care professional can now set a threshold coefficient. With this threshold set by a domain expert, we can create a True Peak Index List.
- *SplitCycles*: The True Peak List allows us to separate the cycles that form the gait of the person under measurements. In order to do that, the filter first finds the shortest cycle of all the cycles because it will limit the size of the average cycle to the size of the shortest cycle (*FindShortest*) in order to avoid null pointer exceptions.
- *FindAvg* The system then calculates the average value of cycles at a given time and outputs the resulting array.

Each of the filters in the arhitecture is described by an algoritm. One of the most important filter is that of finding true peek, for this we use a method described by [19].

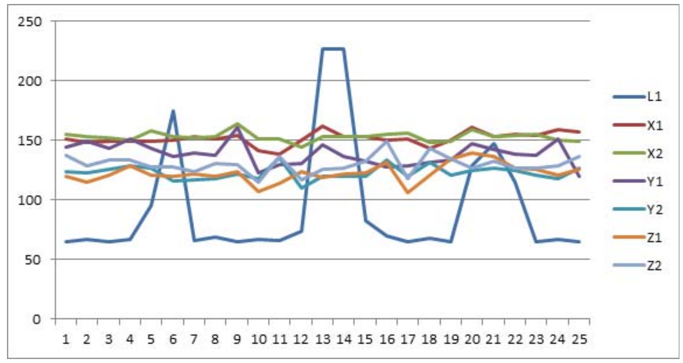


Fig. 5. System Output Normal Gait

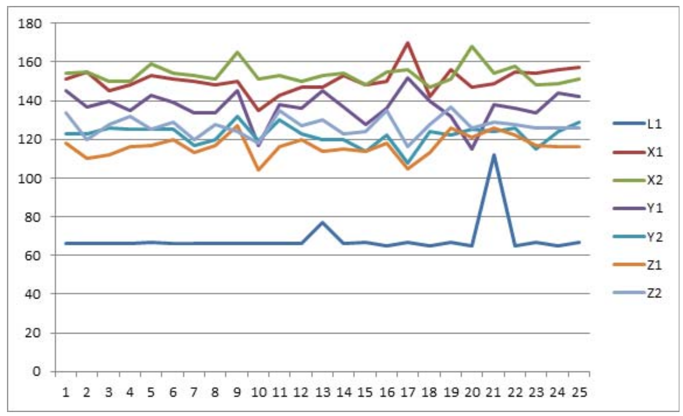


Fig. 6. System Output Antalgic Gait

IV. RESULTS

A. WWGAM Results

Figure 5 displays the output of our A/D ports during the testing mode of the WWGAM for normal gait. Note the three peaks in the blue line represent steps taken, as blue is the saturation of the load sensor, and the load sensor approaches maximum saturation during the loading phase of gait. There is very little movement in the x and y directions, while the z direction shows small variations.

Figure 6 displays the output for an antalgic gait. Note that the blue load sensor value does not even approach full saturation, especially when compared to normal gait. There is very little movement in the x and y directions, while the z direction shows small variations.

Figure 7 displays an arthritic gait. Note that the blue load sensor value shows frequent but low saturation, especially when compared to normal gait. There is very little acceleration in any of the three planes.

Figure 8 displays a myogenic gait. Note that the blue load sensor value shows a saturation that is much more similar to normal gait compared to the other two gait abnormalities. The y direction is the plane that shows the greatest variation in acceleration.

Figure 9 displays a peripheral neuropathic gait. Note that the blue load sensor value shows a saturation that is much

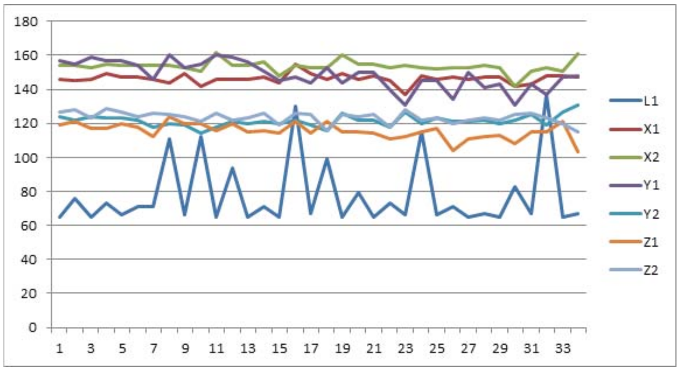


Fig. 7. System Output Arthritic Gait

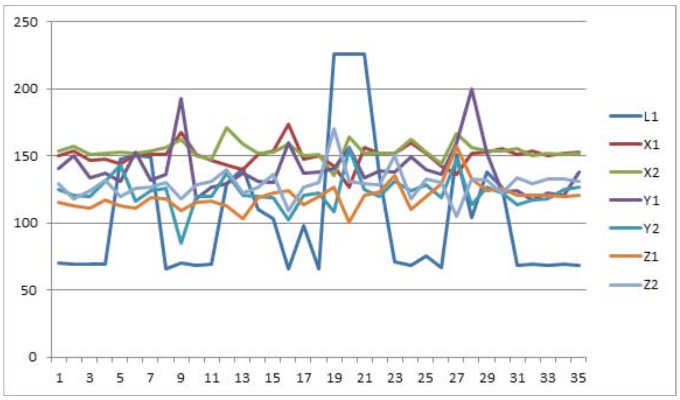


Fig. 8. System Output Myogenic Gait

more similar to normal gait compared to the two first gait abnormalities. The x direction is the plane that shows the greatest variation in acceleration.

B. RGAS Results

As of this writing, automated analysis of the running subsystem has not been completed, but the results obtained thus far are promising. Figure 10 shows one of several cycles performed by a test subject displaying a normal running gait.

From this data, we retain the accelerations in X,Y and Z

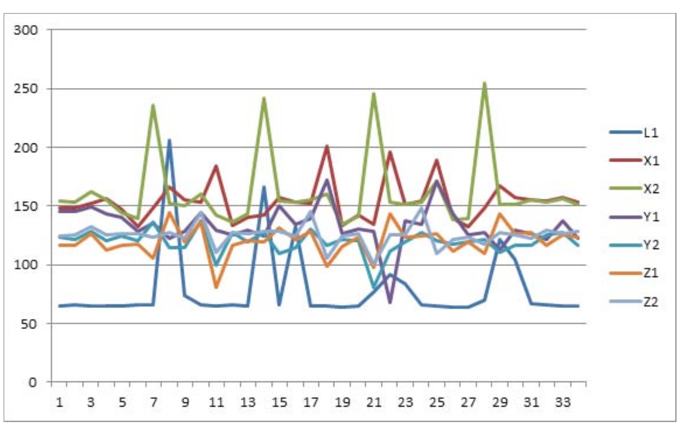
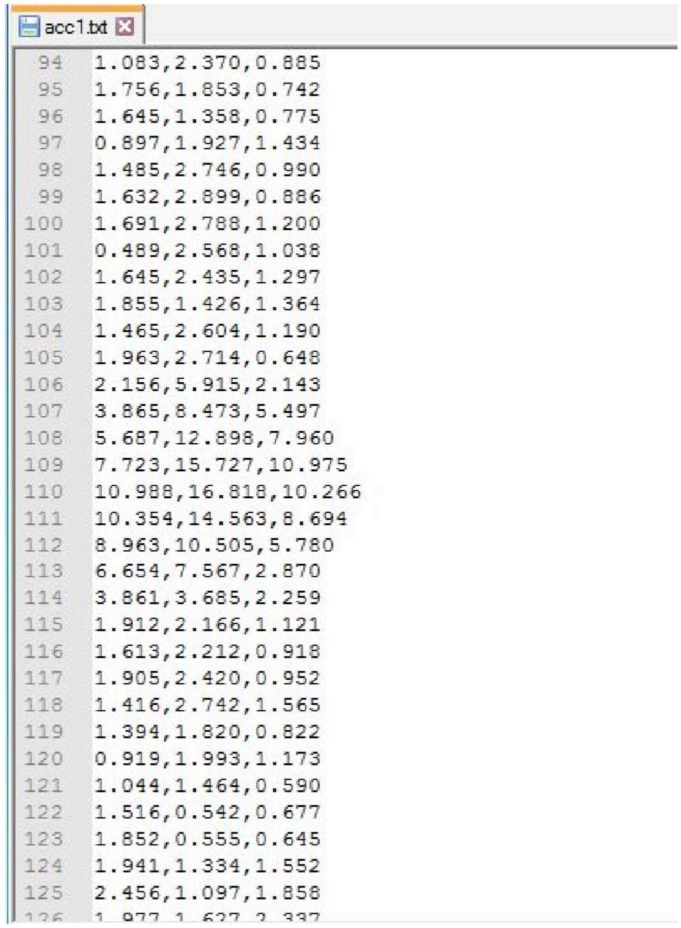


Fig. 9. System Output Peripheral Neuropathic Gait



acc1.txt

```
94  1.083,2.370,0.885
95  1.756,1.853,0.742
96  1.645,1.358,0.775
97  0.897,1.927,1.434
98  1.485,2.746,0.990
99  1.632,2.899,0.886
100 1.691,2.788,1.200
101 0.489,2.568,1.038
102 1.645,2.435,1.297
103 1.855,1.426,1.364
104 1.465,2.604,1.190
105 1.963,2.714,0.648
106 2.156,5.915,2.143
107 3.865,8.473,5.497
108 5.687,12.898,7.960
109 7.723,15.727,10.975
110 10.988,16.818,10.266
111 10.354,14.563,8.694
112 8.963,10.505,5.780
113 6.654,7.567,2.870
114 3.861,3.685,2.259
115 1.912,2.166,1.121
116 1.613,2.212,0.918
117 1.905,2.420,0.952
118 1.416,2.742,1.565
119 1.394,1.820,0.822
120 0.919,1.993,1.173
121 1.044,1.464,0.590
122 1.516,0.542,0.677
123 1.852,0.555,0.645
124 1.941,1.334,1.552
125 2.456,1.097,1.858
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Fig. 10. System Output Raw Data Normal Running Gait

axes. Here we retained the Y axis in Figure 11 after the pipe and filter. The red line is the unfiltered data, blue is filtered with a hysteresis clipping from calibration. From this data we extract the True Peaks which can be used to classify faulty gait.

V. Future work

So far the smart sensors and the dynamic software architecture have evolved in parallel and have not been integrated together. All the communication tests between the smart sensors and the Health Activity Assessment System have been performed using stubs and drivers.We will continue this research further by completing the integration of the smart sensors within the Health Activity Assessment System architecture. Several new smart sensors have been considered for development, the next monitoring system that we are considering is a set of sensors to Monitor the strain on the spine.

VI. Conclusion

In this paper, we present the design of two smart sensors systems that were developed for deployment within a body worn wireless network of medical devices. Both monitoring systems were developed using technology that is becoming

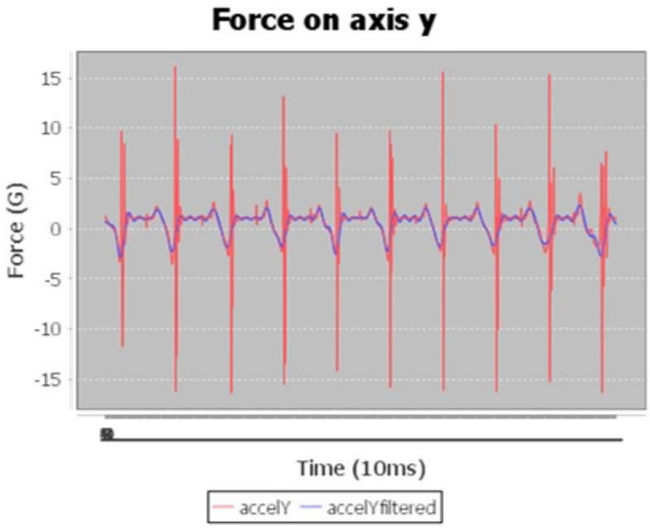

Fig. 11. System Output acceleration on Y Axis Running Gait

more and more the standard to replace static facilities. Both systems surpassed our expectations. Both systems are unintrusive and light weight; together weigh less than 1 kilogram including the power source. Finally the cost of producing both prototype systems was less than $500 CDN. The production cost can be reduced significantly by merging several sensors per microcontroller using our addressing scheme and having the wireless transmitter on-board the microcontroller.

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