2009-01-01

Application of Sensor Fusion to Human Locomotor System

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Dedicated to

my parents Alexander and Victoria Avor

and my uncles Mr. Robert and Edmund Nartey

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by

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THESIS

Presented to the Graduate School of

The University of Texas at El Paso

in Partial Fulfillment

of the Requirements

for the Degree of

MASTER OF SCIENCE

Department of Electrical and Computer Engineering

THE UNIVERSITY OF TEXAS AT EL PASO

August 2009

Acknowledgements

I would like to thank God Almighty for giving me strength each day, without which I would not be able to accomplish anything. Also I would like to thank my adviser Dr. Thompson Sarkodie-Gyan for his guidance and support in supervising this thesis work. I would like to especially thank Dr. Jody Riskowski for her guidance, advise and time dedicated to this research and also express my sincere gratitude to Dr. Patricia Nava for her continual financial support through the ECE Department throughout my peroid of study. I would also like to thank Dr. Bill Tseng for serving on my committee and the contributions and advise he gave to make this work a success.

I am very grateful to members of Laboratory of Industrial Metrology and Automation LIMA and my friends. Lastly but certainly not the least, I am indebted to my family for everything and my thankfulness to them is unique and is incomparable to anyone.

Abstract

This research study proposes a multi-sensor data fusion technique to determine the complex interactions between the sensory, muscular and mechanical components of the human locomotor systems (neuromechanics). The object in this work is to demonstrate the viability in using inertial sensors for accurate gait phase determination and to present the acceleration effect specific body segment contributes in order to establish a functional gait. The current method used in determining gait phases consist of combining ground reaction force and angular data, this method is time consuming and cannot be totally reliable. We found out that gait phases can be determined distinctively by measuring vertical and horizontal accelerations (X-Y) in the sagittal plane on the foot. The foot acceleration graph show 98% unique consistency in determining the seven gait phases, during the experiment when the subjects were walking at normal; fast; even running. Note that this foot acceleration pattern was accurate for three consecutive experiments carried out on three different days under the same experimental conditions. Although the shank, thigh and hip acceleration showed uniqueness in determining the gait phases when the subjects were walking at normal speed, we discovered irregularities in the graph pattern when the subjects were running. For 132lbs, 5.6ft male subject walking at 0.9m/s we noticed that the highest negative acceleration (-0.460g, foot y-direction in the sagittal plane) occurred at heel strike and this is due to the slowing down of the muscles as the leg stabilizes from the

swing phase of the sagittal plane. The corresponding vertical acceleration was 1.226g with a vertical ground reaction force of 24N. As the right leg transitions from heel strike to toe off, a maximum ground reaction force of 619N was detected. At toe-off maximum positive acceleration in the foot y-direction was recorded (0.327g) corresponding to the second highest peek of the vertical acceleration (x-direction 0.311g). We also found out that the only lowest negative acceleration in the vertical plane occurred at the cross over (initial swing) as the foot leaves the ground and the hip flexor muscles are activated to accelerate the leg forward.

In total six able bodied subjects were involved in this research (four male and two female). The experiment was performed using rate gyroscopes and linear accelerometers attached to the right hip, right thigh, right knee, right shank, right ankle and right foot. The assumption in this work is that walking pattern in able-bodied people is symmetrical, thus we assume the same acceleration conditions and gait phases detections as well is symmetrical. The subjects were allowed to walked at their normal walking speeds, however, for comparison in some cases all subjects walked at 0.9m/s and 1.08m/s. The experiment lasted for 120s when the subjects walked on an instrumented treadmill and the setup was synchronized with ground reaction force sensor, and Simi motion capture system. The sampling frequency was 280Hz for all four sensing technologies. When the subjects walked over ground, the experiments lasted for about 5sec. Outputs from the sensors were fed into fuzzy inference system, where the concept of fuzzy similarity was applied to determined the coordinated walking pattern for each subject (CWP).

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Chapter 1

Introduction

1.1 Background and Significance of the Study

1.1.1 Summary

In order to determine the forces transmitted in the muscle, ligaments and joint articular surfaces during human locomotion, a complete knowledge of muscle recruitment has to be understood [1]. However, current understanding of this concept is inadequate, thus computer simulations designed to depict human locomotor system fall short in real time applications. It is evidential that the draw backs in proposed simulation models is due to the complexity of sensory coordination of inputs from the motor cortex, cerebellum and basal ganglia coupled with feedback from the vestibular, visual and proprioceptive (position and velocity of the limb) sensors of a healthy person during locomotion [3]. Due to the afore mentioned constraints, couple with computational power limitations, most gait analysis models are planar and comprise only few degrees of freedom and most models do not even consider the dynamic behavior of muscles [4]. Hence the simulations yields ground reaction forces and joint torques, but not muscle forces or neural excitations. Based on these inadequacies we

can say we are still far from quantitative description of the human locomotor system.

It is important to note that, the complexity of the human locomotor system, makes it difficult to achieve accurate mathematical models to represent some parameters especially when simulating effects. Thus these simulations results contain some assumptions which necessarily do not represent the true physiology of the musculoskeletal interactions locomotion. As part of the goal of this study, we seek to demonstrate the sensory coordination that takes place for a functional gait to be attained. We also demonstrated accurate determination of seven gait phases from accelerometric data.

1.1.2 The concept of Sensor Fusion

Sensor fusion is increasingly viewed as a perceptual activity; i.e., a collection of observations from multiple sensors and combining these observations into a single, coherent precept [6]. However, this perception is characterized by the range of information that the sensor can provide and by the complexity of their operation [7]. For example, for a mobile robot to be able to plan and guide the execution of task it has to coordinate the observations (sensor information) into a consistent consensus view of the environment and some form of intelligent has to be applied to this information for the robot to precisely and robustly make a confident decision [22]. Unlike robots, human's use biological senses (ear, skin, eye, nose and tongue) to provide appropriate information such as visual and proprioceptive signals which are fused together in the brain (just as the intelligence in robots) to make reasonable locomotor decision. In humans, locomotion consists of involvement from the sensory receptors in bones, skeletal muscles, cartilage, tendons, ligaments, joints and other connecting tissues [23]. The combinations of these individual organs form the musculoskeletal system and its function is to

provide stability and movement to the human body. In [23], it was stated that for any bipedal walking to occur, there should be periodic movement of each foot from one position of support to the next and sufficient ground reaction forces must be applied through the feet to support the body. When this step is executed properly, then a functional gait is registered.

1.1.3 Existing System

Recent researches have been concentrating on understanding the complex interaction (forces generated, muscle activation, etc) of the human locomotor system. The fact however is that, in order to determine the forces transmitted in the muscle, ligaments and joint articular surfaces during human locomotion, a complete knowledge of muscle recruitment has to be understood, but this knowledge is currently inadequate [1]. This is a result of several factors including, lack of computational power (calculating higher degrees of freedom), assumed parameters in simulation models, inadequate of neuronal firing sequence and the experimental approaches used to investigate biomechanics.

As an effort to understand the human locomotion process (human gait analysis), several approaches including:

- Observation
- Footswitches
- Gait mats
- Force Plates

- Optical System
- Questionnaires.

The most common method employed is the optical system (high speed camera). Despite the high cost of this investigation method, the system has set-backs such as: occlusion, interference from shining surfaces, computational limitations and lastly this system only provide trajectory information from reflective maker displacement after which vector analysis is used to derive rotational and linear kinematic data. Evidential this system are limited to laboratory investigations only due to their setup.

1.1.4 Way Forward

In view of these, researchers are proposing the gradually drift from the use of optical system in the investigation of human gait to the use of micro-mechanical devices such as linear accelerometers, inclinometers, goniometers, force sensitive resistors and rate gyroscope [8], [9]. These sensors are inexpensive thus sparking extra-ordinary interested in researchers. Many researchers have already started using single or multi-accelerometer combinations to analysis gait phase changes or assessments and these studies so far are plausible and find application in clinical and robotics [10], [11]. Accelerometer data was implemented to detect static and dynamic activities, in domestic real-time experiments [12]. It was determined that the accelerometers have high inclination sensitivities when its sensitive axis is horizontal.

In [12] gait phases were determined by using accelerometer to distinguish between stance and swing phases, but high sensitivity of the sensors sometimes introduces undesirable signals which affect real-time precision. Determination of knee unlock during stance was investigated in [10]. Although accelerometer signals do not contain

information about rotation around the vertical axis, their work was possible with the use of goinoimeters and quantitative vector analysis. In [], Auvinet and et al, used two accelerometric devices attached to the center of mass of 282 healthy subjects ranging from age 20-98 and monitored their gait dynamics as they walked on a corridor of 40m. It is worth mentioning that the vertical acceleration recorded from this study cannot be distinctively separated for each body segment of the lower extremity, instead it is a combination of the effect from all the body segments involved in producing the said acceleration. This is good, however, in situations where individual components need to be examined it would not be of much help.

Unlike the optical system, accelerometers and rate gyroscopes measure acceleration and angular velocities directly for biomechanics analysis. However, the problem with this system is that they suffer from fluctuating offset, which results from temperature change or small changes in orientation. In order to maintained high precision H. J. Luinge proposed a Kalman filter that fuses tri-axial accelerometer and rate gyroscope signals for ambulatory recording for human body segment orientation [12]. According to Luinge, fusion technique such as Kalman filter continuously correct's offset errors from accelerometer outputs.

1.2 Motivation

Commercial gait data collection systems currently in use utilize optical systems for data collection by monitoring trajectories of reflective markers attached to the subject. This system has defects such as occlusion, in-accurate tracking of reflective markers during data analysis, interference from shining surfaces and specially designed laboratories for the data collection in order to eliminate brightness and increase contrast. With regards to this, the systems can only be used in confined locations which is not

helpful in recording real life gait experiences such as climbing stair cases or walking through complex environments.

Notwithstanding the above, since the system measures only trajectory data, computation methods such as integration and differentiation are employed to find velocity and acceleration. As well know, these numerical methods introduce noise to the measure quantity thus reducing the signal to noise ratio. Since gyroscopes and accelerometers are ideal devices used to measure acceleration, velocity and angular displacement and are cheaper we desired to explore their viability in human motion analysis.

1.3 Reseach Goal

This research work has two main goals - the first is to measure gait parameters and the second part consists of developing an intelligent sensor fusion system that can validate dynamic and kinematic gait parameters (biometric) of people. Triaxial linear accelerometers, rate gyroscopes, Bertec force plates, instrumented treadmill and electromyography were used to measure the dynamic and kinematic gait features from ten subjects. The claim is that this system will provide a more accurate measurement of accelerations, angular velocity and joint angles, since this system is free from mathematical manipulations to arrive at the above mentioned parameters.

The first step in this work consists of quantitative human motion analysis of the lower extremity and validation of the measured parameters from the developed sensing devices used to measure the gait features. Changes in the hip, shank, thigh, arm, foot, ankle were determined from the rate gyroscopes and accelerometers. The EMG and force plates provided information of muscle activity and distribution during

walking. Three experimental scenarios were investigated, walking/running on an instrumented treadmill, walking in a room freely and navigating through office setting, and sit-stand.

In the second step, an intelligent sensor fusion model is developed using fuzzy inference system to validate gait parameters measured in the previous step. This fusion algorithm is developed based on gait analysis results including, lower extremity movement obtained with the sensor system described above and compared with the motion analysis results obtained using laboratory optical motion analysis system.

1.4 Overview

Chapter 2 will discuss the general concept of locomotion both biological and robotic system approach. Chapter 3 will introduce the sensor fusion concept and validation, design of the experiment and the approach used.. Chapter 4 presents kinesthetic and dynamic experiments on group of subjects and discussion of results. Chapter 5 presents the conclusions and future work.

Chapter 2

The Human Locomotor System

2.1 Summary

There are seven fundamental components that form the functional basis of the way in which we walk: central nervous system, peripheral nervous system, muscle, synovial joint, rigid links segment, movement and external forces. This sequential behavior of the various processes from top-down constitutes a cause-and-effect model, figure 2.2, [19]. It is important to note that this cause-and-effect model occurs by conversion of an idea in the supraspinal centers into a pattern of muscle activity that is necessary for walking. Thus activation of the lower neural centers establishes muscle activation patterns and the sensory feedback from muscles, joints and other receptors modifiers the movement.

2.2 Sensor Fusion

For any bipedal walking to occur, there should be periodic movement of each foot from one position of support to the next and a sufficient ground reaction forces must be applied through the feet to support the body. These cyclic performances can be simplified as stance or swing phases, figure 3.3. In [19],they outlined the sequence of events that must occur for a reasonable locomotion. Firstly, there must be registration and activation of the gait command in the central nervous system, this should be followed by the transmission of the gait signals to the peripheral nervous system. The signals intend contracts the muscles to develop tension. Consequentially there should be generation of forces at the moments across, synovial joints which should be regulated by the rigid skeletal segments based on their anthropometry. Then displacement of the segments in a manner that is recognized as functional gait; and lastly generation of ground reaction forces.

To be able to understand the above mentioned sequence of determining human gait, two approaches can be implemented, direct dynamic or inverse dynamic. Direct dynamic problem involves the analysis of forces applied to a mechanical system (force sensors) and the objective is to determine the motion that results. The second approach which is inverse dynamic, is when the motion of the mechanical system is defined in precise detail and the objective is to determine the forces causing that motion. With the pre-established control process that occurs in walking, the human body for that matter movement can as well represented in three fundamental planes; sagittal, coronal (or frontal), and transverse.

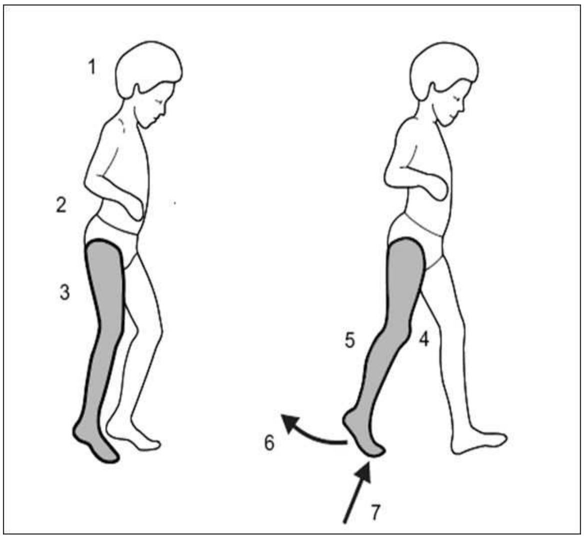


Figure 2.1: Stages involved for a meaningful functional gait to be established

[19]

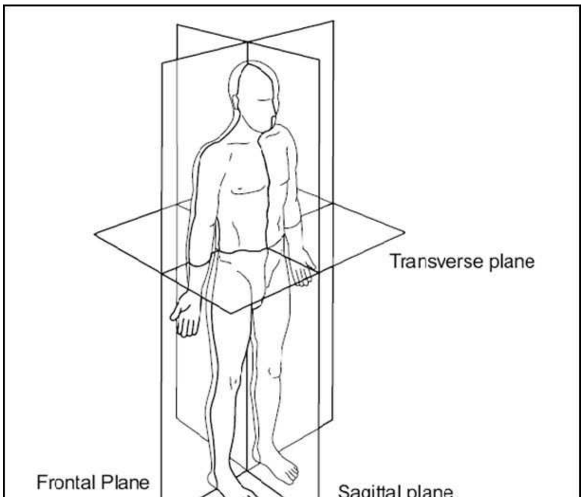


Figure 2.2: The three main planes of locomotion [19]

2.2.1 Biological Sensory Coordination

In humans, regulating movement consists of involvement from the sensory system, bones, skeletal muscles, cartilage, tendons, ligaments, joints and other connecting tissues [23]. The combinations of these individual organs form the musculoskeletal system and its function is to provide stability and movement to the human body.

These cyclic performances can be simplified as stance or swing phases. Also, there should be sequential occurrence of some events for a reasonable locomotion to occur. The first is there should be registration and activation of the gait command in the central nervous system; this should be followed by the transmission of the gait signals to the peripheral nervous system. The signals intend contracts the muscles to develop tension. Consequentially there should be generation of forces at the moments across, synovial joints which should be regulated by the rigid skeletal segments based on their anthropometry.Then displacement of the segments in a manner that is recognized as functional gait; and lastly generation of ground

Each of these steps clearly shows, there must be a well orchestrated information fusion archetiture to accurately combine all these information and trigger required actions. The brain however, does the fusion in normal human beings, for prior to muscle contraction inputs from the sensory system (visual, vestibular and kinesthetic) are integrated in the motor cortex and these sends electrical signal through the spinal cord and peripheral nerves to the muscles to initiate the desired contraction [24]. The sensors in the muscles and joints as well relay proprioceptive sensory information through peripheral nerves to the cerebellum and other parts of the brain [5]. Thus for any biologically meaningful movement to be made, a complex and complete sensory and muscle interaction must occur.

It is also important to note that the motor pattern activity of humans during locomotion changes with respect to the environment and the information fed to the feedback sensory system. Missing information from any of these sensors will alter gait phases synchronization and will result in clumsy movement. It is now clear that sensor information has to be available when needed for confident decision making, however, in mobility disabilities like cerebral vascular accidents, which occurs as a consequence of cell death; the neurons does not function correctly, consequently fail to provide the information needed for fusion [6]. This can change the loading and unloading patterns in humans during gait cycles. This has been a major huddle in neuroscience and developers of medial robots (assistive rehabilitation devices for learning new motor commands).

2.2.2 Robotic Sensory coordination

The concept of sensor fusion demonstrates that, the ability of mammals and multi-purpose robots to move through complex environment in a flexible manner depends on integration of their sensory information (sensor fusion) into flexible fusion system architecture [7], [25]. We can say the human brain perhaps is the best paradigm to illustrate the fusion concepts conceived, for information processing on a multisensory environment. The brain generally combines five kinds of signals (hearing, touch, sight, smell and taste) received from the basic sensors (ear, skin, eye, nose and tongue) and comes out with a consensus and confident decision in real time [26].

Fusion occurs at different levels for different functions, for instant; perceptions from the visual, vestibular coupled with kinesthetic are obtained at prcised and distinct time frames, however, these responses are properly integrated into the central

neuronal network in the brain to produce a meaningful intentional movement. Therefore, motor impairments often result whenever there is missing information from a particular sensor/sensory group during the fusion process. This means that biologically useful behavior does not comprise of an action of single afferents, neurons, muscles or limbs, because the information obtained from single sensory sources are limited in terms of triggering motor control mechanisms [24].

Models have been developed for sensor fusion for artificial intelligence purposes. These models however, are characterized by their application domain, fusion objective and the sensor types. The most significant classification criteria is the sensor type mainly because the information used for fusion depends on the physical sensor's measured quantities [21]. For instance in engineering process monitoring, the sensor types are specifically designed to measure a desired parameter and correlated it to the process of interest.

However, in situations where during measurement the principal signal component has distortion higher than the desired measurement; the signal yields no meaningful correlation of the desired quantity. In order to eliminate this problem, more than one sensor is used to measure the same parameter and the outputs integrated to obtain the resultant [22]. Thus sensor fusion techniques such as evidential reasoning, fuzzy logic, neural networks, etc, are applied to combine information from theses different sensors and provide appropriate feedback signals for confident decision making.

Comparatively, in the musculoskeletal system the neuronal sensory network provides feedback to the central neural system about proprioceptive activity for accurate decision making, which is similar to the feedback concept mentioned earlier in robotic systems. In this work, the authors will be investigating the sensory coordination based

on the measured kinesthetic and dynamic behavior of the lower extremity body segments.

Chapter 3

Gait Analysis Using Inertia Sensors

3.1 Summary

In this study we developed motion sensing device comprising of one triaxial accelerometer (ADXL330), and three single axis rate gyroscopes on a single prototyping board see Figure 3. The rate gyroscopes (ENC-05EA, ENC-05EB and ADXLS150) were calibrated to measure X-Y-Z components of angular velocity of the body segment respectively and the triaxial accelerometer was used to measure linear acceleration in X-Y-Z sensitive axis in the sagittal plane. Both sensor systems were designed with a second order band pass filter with (0.33-35Hz) with a 20-dB gain in the band pass. Based on the outputs from the sensors, a fuzzy inference system (FIS) is developed for the determination of gait phases. A digital filter is designed to remove noises from the output of the fuzzy inference system. Lastly, a sensor fusion algorithm is developed to validate optical motion analysis approach.

Table 3.1: Gryoscope operating characteristic at 5V supply- datasheet

Parameter	**Min**	**Typical**	**Max**	**Unit**
sensitivity	-150		+150	^0S
Initial Null		2.50		V
Rate Noise Density		0.05		$^0S,\sqrt{Hz}$
Angular rate		0,±1		^0S,G
Bandwidth		80		$\sqrt{Hz}$

3.2 Hardware Description

3.2.1 Gyroscopes

Table 3.1, shows summary of the initial operating conditions of the three gyroscope used (ENC 05EB, ENC 05EA, ADXLS150, size 1.58.04.3mm weight 0.2g). The three sensors each measure a single axis rotational motion. Figure 3.1 shows the arrangement scheme implemented in this work in order to have the sensors detect motion in X-Y-Z coordinates. We arranged the sensors in such a way that each detects rotational motion in a specific axis (X-Y-Z) in the sagittal plane. The murata ENC-05EB gyroscope measures the rotational velocity by sensing the mechanical deformation caused by the Coriolis force on an internal vibrating prism. The gyroscope signal was filtered by a second-order band-pass filter (0.25-25 Hz) with 20dB gain in the pass band.

The signals were sampled at 280Hz with a resolution of 16bits through A/D card (NI-USB6218). The filtered gyroscope signal was used to directly estimate the angular velocity of the foot, knee, ankle, shank, thigh and hip. I integrated the signals to

Figure 3.1: Arrangement of three single axis rate gyroscopes to detect three dimensional motion in the sagittal plane.

estimate the inclination of the body segments relative to the ground.

3.2.2 Accelerometers

The accelerometer-chip (ADXL330, size 4 mm 4 mm 1.45 mm) used is a small, thin, low power, complete 3-axis accelerometer with signal conditioned voltage outputs, all on a single monolithic IC. It measures acceleration with a minimum full-scale range of 3 g. The output signals are analog voltages that are proportional to acceleration and it can measure the static acceleration of gravity in tilt-sensing applications, as well as dynamic acceleration resulting from motion or vibration.
The bandwidth selected for this research is (0.25-50Hz) each for each axis (X-Y-Z),

and the signals were sampled at 280Hz. The accelerometer uses a single structure for sensing the X, Y, and Z axes. As a result, the three axes sense directions are highly orthogonal with little cross axis sensitivity. In order to obtained accurate results, the mechanical misalignment was calibrated out during the experiment.

3.3 Methodology

3.3.1 Proposed Approach

Figure 3.2 is the processing flow of the proposed method. The first block consists of registration and activation of the gait command in the central nervous system and transmission of the gait signals to the peripheral nervous system. This signal coupled with the proprioceptive feedback signals produces tension in the muscles (muscle dynamic) which results in a specific body dynamic. EMG measures the muscles dynamic whiles the body dynamic are the inputs to the accelerometer and rate gyroscopes. The accelerometer and gyroscopes recorded voltage changes what corresponds to a specific acceleration and angular velocity respectively. Outputs from the sensors are feed into a data acquisition system and the final output signal was fed into an intelligent fusion system. Fuzzy Inference System is used for intelligent decision making and the output from the FIS shows a distinctive coordinated walking pattern. The output from the FIS as well should correspond and be recognized as a functional gait as recorded by the force plate data.
Table 3.2, illustrated the meaning of the gait phases detected in this work, and Table 4.4 shows the data on each subject

Figure 3.3 shows the graphical illustration of the phases description in Table 3.2.

Table 3.2: Description of gait phases as proposed in [19]

Phase	Description
Heel Strike (HS)	Initial contact
Loading Response (LD)	Heel Strike to flat foot
Mid-Stance (MS)	Flat foot to mid-stance
Terminal Stance (HF)	Mid-Stance to heel off
Pre-Swing (PS)	Toe off
Initial Swing(IS)	Toe off to acceleration
Mid-Swing (MiS)	Acceleration to mid swing
Terminal Swing (TiS)	Mid Swing to deceleration

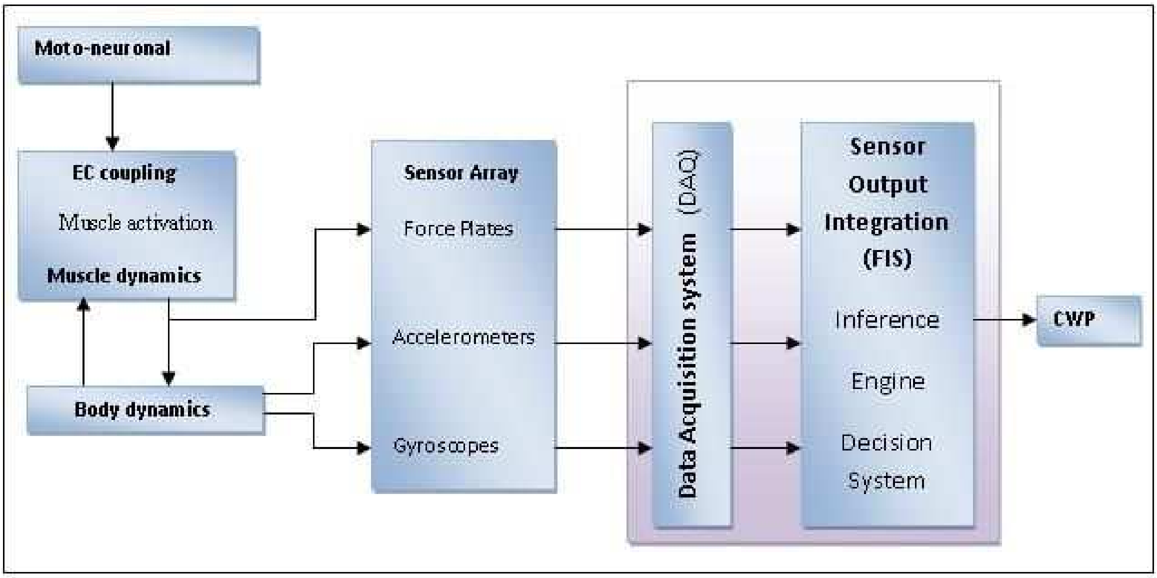


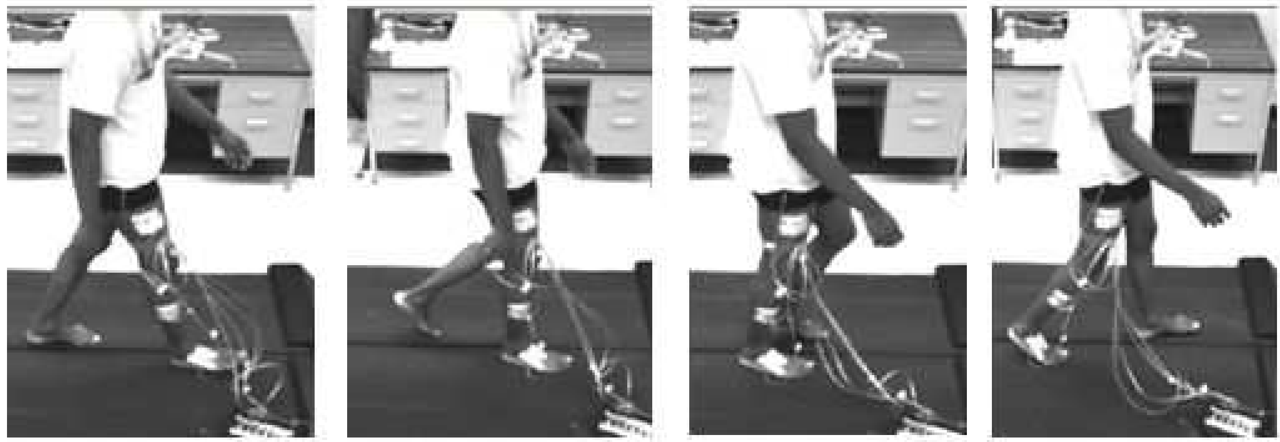
Figure 3.2: The proposed signal flow and processing scheme. The output from the fusion systems is a specific coordinated walking pattern (CWP)

The first four pictures shows the stance phase whiles the last four pictures illustrate the swing phase.

3.3.2 Experimental Procedure

3.3.3 Description of the setup

Four able-bodied male subjects and two females were involved, Table 4.4 provides details about the subjects. The sensors were attached to the right leg of each subject (Figure 3.3) with the assumption that walking in able-bodied persons is symmetrical. Measurements were taken on the right hip, right thigh, right shank, right knee, right angle, and right foot. The subjects were involved in three different experiments; 1. walking at different speeds on an instrumented treadmill, 2. Sit and Stand, and 3. Walking over ground. During each experiment, the sensors were aligned in order to measure X-Y coordinated movements in the sagittal plane. A second order band pass filter was designed to filter the unwanted frequency components with a limiting frequency range of (0.25-35Hz). In order to synchronize the sensor signals with the force plate data and the optical system (Simi motion capture system), a sampling frequency of 280Hz was chosen. At the start of each experiment, the sensors were placed on the test subjects when standing on the treadmill at a static position. The initial readings from the sensors were recorded as static voltages (Vs).These static voltages reading were used to calibrate the accelerometers to a zero (g) position for the subject before the experiment starts. The acceleration detection sensors can measure up to 3g; this corresponds to a specific analog output voltage recorded and it is determined from the transfer function shown in equation 1.



(a) Initial Contact (Right heel strike) (b) Flat foot (c) Mid stance (d) Heel – Off

(e) Toe-off (f) Acceleration (Swing Phase) (g) Mid Swing (h) Deceleration

Figure 3.3: Experimental description of the gait cycle

Table 3.3: Details about the subjects involved in this work

Subject	**Gender**	**Height (ft)**	**Weight (lbs)**
Subject A	Male	5.6	130.7
Subject B	Male	5.8	170
Subject C	Male	5.5	126.8
Subject D	Male	5.76	178
Subject E	Female	5.4	130
Subject F	Female	5.6	126

3.3.4 Data Acquisition System

NI6218 data acquisition card was used to collect the data from the sensors. The acquisition card has a resolution of 16bit A/D NI-USB6218 with a range of +/- 10V. In order to synchronize the measured acceleration data with the force plate data and the optical system,a sampling frequency of 280Hz was used. The force plate data was used as reference to confirm our sensors measured at least the four accurate gait phases the force plate data can provide.The sensors were calibrated at zero g levels before the test. Each cluster was set in position for 120s and sampled at 280Hz. The sensor system used has an excellent temperature stability thus there were no variation in DC baseline due to change in room temperature. The sensitivity was 330mV/g with an accuracy of +/-10n/a. The signals were band pass filtered between 0.25Hz and 50Hz. The output voltage of the accelerometer varies between 1.5 - 3V corresponding to an acceleration of +/- 3g.

3.3.5 The Calibration Process

In order to measure accurately, we need to calibrate the sensors to an initial reference point. This determines the precision of the measured quantity. At specific sensor orientation, a 0g/1g/-1g is produced and this corresponds to a particular V. In this work at zero (g) the output voltage of the accelerometers was 1.5V when supplied at 3V. An initial voltage was recorded when the subject is at a stationary position. The average of these signals was used to determine the calibration constant used to determine the zero (g) position.

3.3.6 The Sensor Fusion and Validation

Locomotion consists of periodic movement of each foot from one position to the next and two there must be sufficient ground reaction forces applied through the feet to support the body [19]. As part of the objective of this study we aim to determine the impact factors in terms of acceleration that affect walking in able bodied subjects. To validate our sensing system, the subjects performed the same set of experiment thrice and three sensing technologies were used to validated what was recorded in previous and subsequent experiments simultaneously. It is worth noting that each experiment was carried out under the same conditions.

Figure 3.4 shows the accuracy of our sensor scheme in the detection of changes in gait phase as compared to what was recorded by the force plate data. Traditionally, force plate data can distinctively determine four gait phase and angular data has to incorporated to determine the other three phase. Our proposed approach was able to determine all seven gait phases from one acceleration graph. This validation was possible because we carried out experiment in synchronism with eight high speed cameras and embedded force sensors in an instrumented treadmill all sampling at

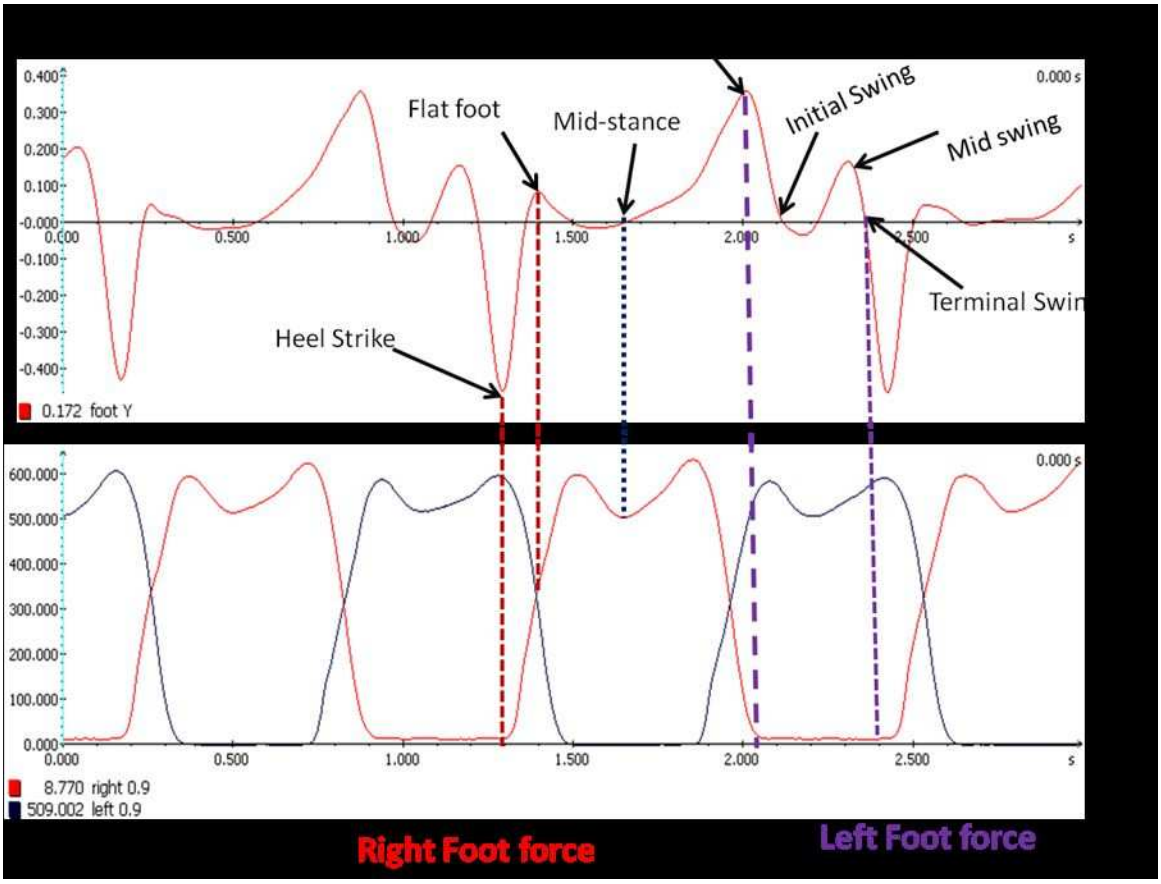


Figure 3.4: Comparison of acceleration data on subject A walking at 0.9m/s with the simultaneous force data recorded

280Hz/s. Fuzzy similarity concept was implemented on the data obtained during each experiment phase. This was done to determined the degree of accuracy in the detection of our sensing system. As show in the results obtained, the accuracy range between 89 to 100 percent. This factor is examined to determine the trend use for the sensor fusion.

As indicated in figure 3.2, the motor neuronal activities results in muscle activation, (muscle dynamics) there is a two way communication process between the muscle dynamics and the body dynamics. The previous is measured using EMGs and the corresponding ground reaction forces are measured with force plates embedded in the instrumented treadmill. The body dynamics (kinetics) are measured using accelerometers and rate gyroscopes and the output is a Coordinated Walking Pattern (CWP)

Chapter 4

Results and Discussion

The current practice in determining gait phases consist of analysis on force plate data and angular data obtained form mathematical manipulations. Validating the results obtained from this experiment shows that, it is possible to completely determine the seven gait phases accurately and distinctively from acceleration graph 4.1. It is important to note the experiment was carried out on the right leg of the subjects with the assumption that walking pattern in able-bodied persons is symmetrical. The graph in figure 4.4, and figure 4.5 it can be seen clearly that when the sensor was used to measure acceleration in the foot for subject walking at different speed, it was able to accurately reproduce the gait phases (refer figure 4.1 to see the definition of the gait phases as recorded by the accelerometers).

4.0.7 Determination and Validation of Gait phases

[h] From figure 4.1, heel-strike initiates the gait cycle and represents the point at which the body's center of gravity is at its lowest. This corresponds to the highest negative acceleration value. Flat foot is the time when the plantar surface of the foot touches

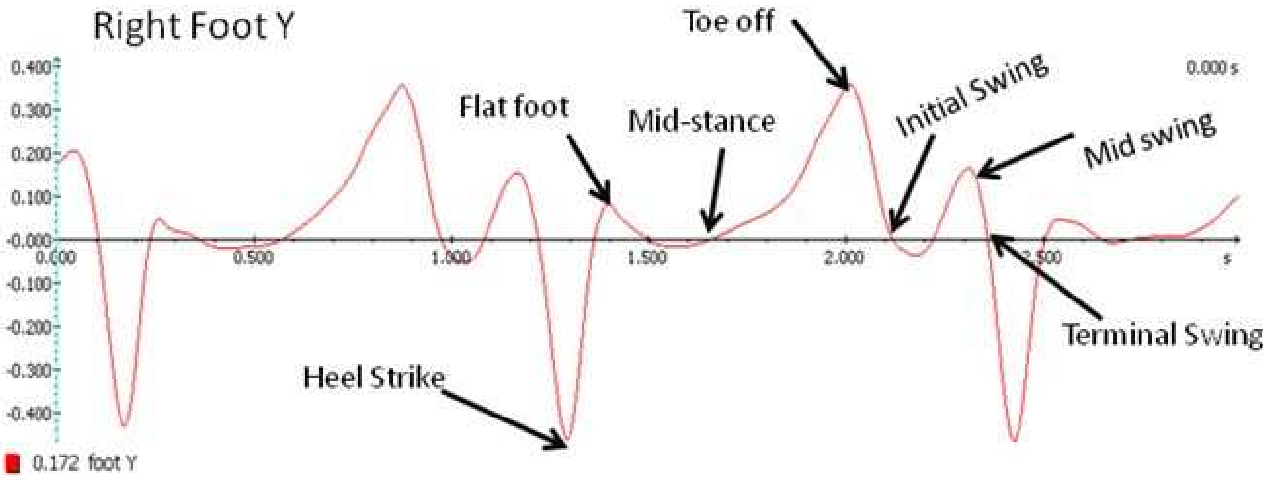


Figure 4.1: The seven Gait phases as clearly detected from the accelerometer, from the foot Y direction acceleration in the sagittal plane.

the ground. Mid stance on the other hand occurs when the left foot passes the stance foot and the body's center of gravity is at its highest position. Heel-off occurs as the heel loses contact with the ground and pushoff is initiated and toe off terminates the stance phase as the foot heaves the ground, at this point a the highest acceleration in the y-direction occurs, and this corresponds to the second largest vertical acceleration in the sagittal plane. Note that first largest vertical acceleration (x-direction) occurs at the initial contact.

Acceleration begins as soon as the foot leaves the ground and the subject activates the hip flexor muscles to accelerate the leg forward. Mid swing occurs when the foot passes directly beneath the body, coincidental with midstance of the left foot.

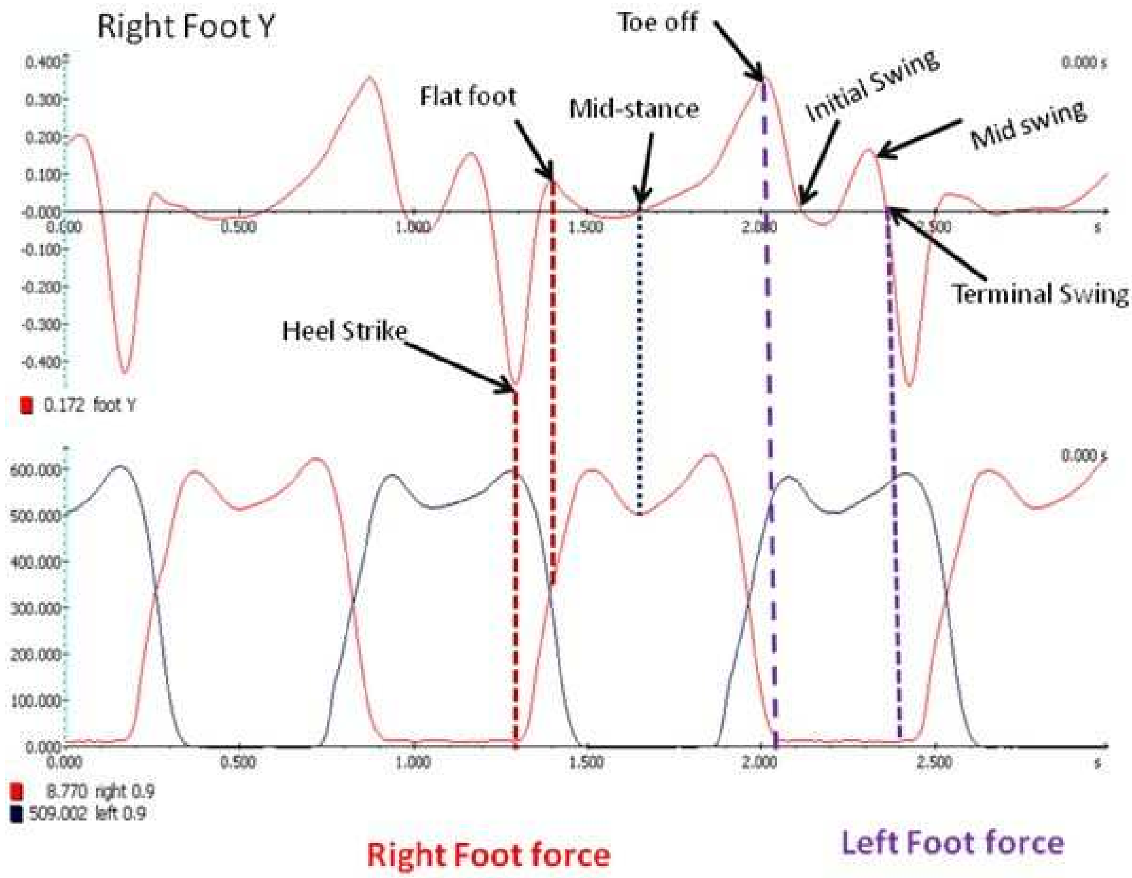


Figure 4.2: Shows the accuracy of the accelerometers detecting changes in gait phases. The Right and left force plate data have been shown to confirm this detection.

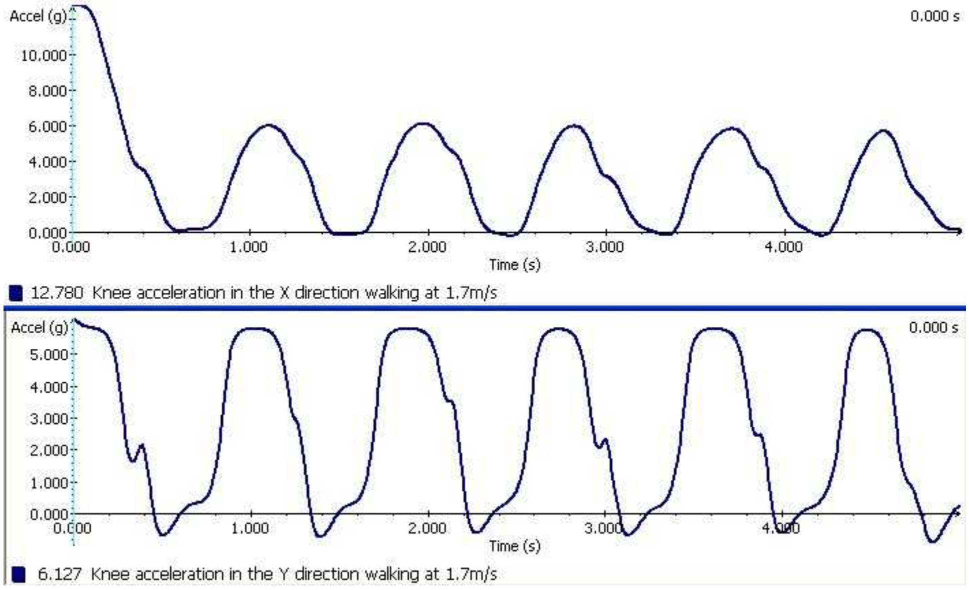


Figure 4.3: The Knee acceleration measured in the sagittal plane when the subject was walking at 1.7m/s on a treadmill

Deceleration describes the action of the muscle as they slow the leg and stabilize the foot in preparation for the next heel strike. This corresponds to a negative acceleration (deceleration) as recorded by the accelerometers. Detailed acceleration graphs on subject A are presented in appendix B, Appendix C has the acceleration graphs for subject B.

4.0.8 Subject A - Knee Acceleration Graphs

Figure 4.3 is the acceleration of the knee when the subject was walking at 1.7m/s note the distinction in the harmonic motion.

4.0.9 Acceleration Graphs

4.0.10 subject A - walking 0.9m/s

Analyzing the coordinated accelerations from the body segments examined for subject A walking at 0.9m/s shows that, heel strikes occur consistently with an acceleration of approximately 0.44g. The loading respond occurs 0.102s after the initial contact. During the mid-stance approximately 0.004g occurred at a periodic rate of 0.301s after the loading response. The terminal stance occurred 0.491s after the loading which is about 0.19s after the mid-stance and with an acceleration of 0.145g. For the initial swing, 0.005g was recorded and time preceded the terminal stance and loading response 0.22s and 0.712s respectively. The last two stages of the gait sequence showed acceleration of 0.15g and 0.001g for mid-swing and terminal swing respectively. The corresponding durations were 0.89s and 0.96s accordingly.

Figure 4.6 and 4.7 shows one complete gait cycle of foot acceleration in the Y-direction and X-directions for subject A respectively (please not figures 4.4 and 4.5 shows four complete gait cycles for a period of 5secs). From the graph it is clear that heel strike occurred at all most 0% with a higher acceleration of 2.5g. It drops down from sharply from 10% of the gait cycle to 0g at about 30% of the gait cycle. The reason been during that phase, the foot is in contact with the ground and there was no velocity to trigger a corresponding acceleration (see eqn. 2). During the flat foot phase, the acceleration remains sturdily at 0g and after 40% there was a gradual increase in the acceleration which is a preparation of the foot to enter a swing mode. The pre-swing phase occurs at about 50% when there is gradual increase in acceleration. The initial swing occurs from 60% of the gait phase and that is when the toe leaves the ground. It is worth noting that the swing lasted between 60-85% when the foot prepares for another heel strike. For the Y-axis at heel-strike, the velocity is

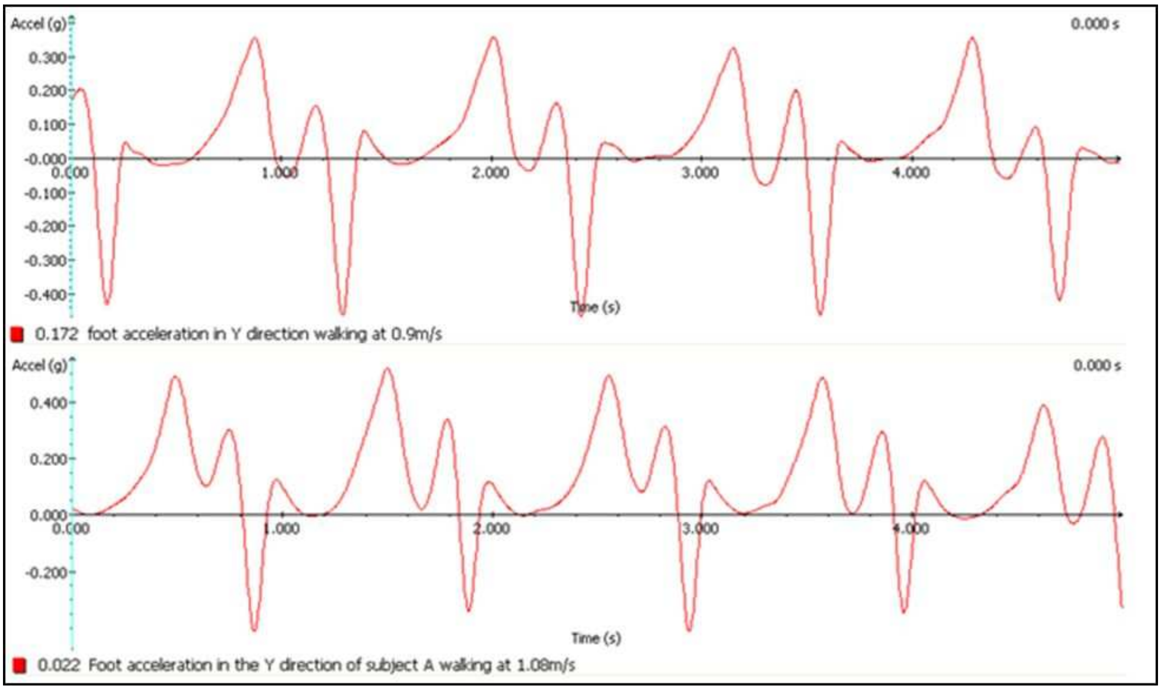


Figure 4.4: The above figure shows the accuracy of the accelerometers detecting all the gait phase for two different walking speeds. 0.9m/s and 1.08m/s in the for foot acceleration in the Y-direction of the sagittal plane. Notice that at during mid stance for the 0.9m/s speed, the acceleration graph slights drops below zero. This is due to the effect from the treadmill

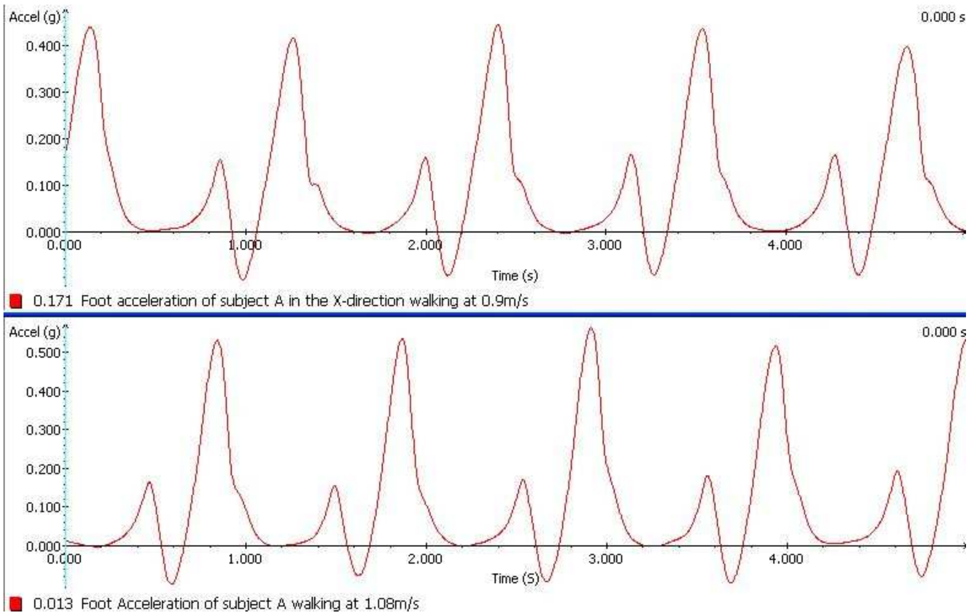


Figure 4.5: The above figure shows the accuracy of the accelerometers detecting all the gait phase for two different walking speeds. 0.9m/s and 1.08m/s in the for foot acceleration in the X-direction of the sagittal plane

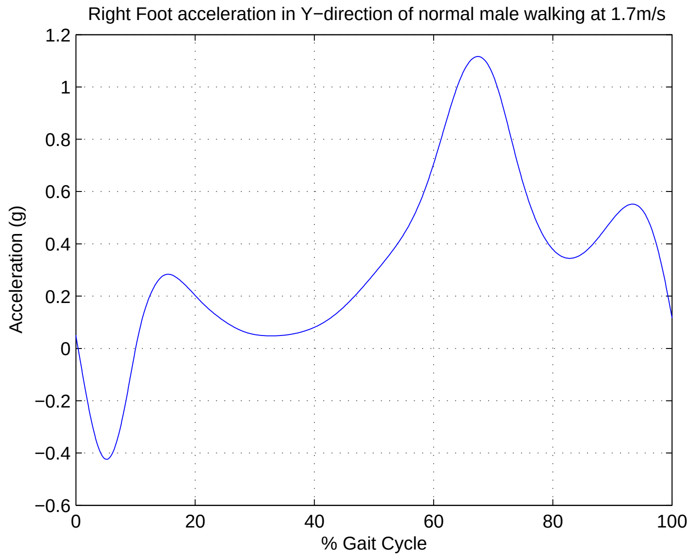


Figure 4.6: One complete gait cycle of foot Acceleration for Subject A in the Y axis

negative because the center of gravity is still moving downward.

Analyzing figures 6-1, 6-2, 6-3 it shows the gait variability in the right shank x-axis acceleration in the sagittal plane of an able bodied male subject walking at a speed of 0.9m/s, 1.08m/s and 1.70m/s respectively. The vertical axis shows the acceleration in (g) as a function of time. The 0.195g occurred at toe-off and the initial swing lasted for almost 0.231s until the acceleration returned to almost zero (0.006g) at mid-stance. It increases gradually to about 0.033g at terminal stance and drops to 0.009g at heel-strike again, due to the center of gravity.

Analyzing figures 4.9, 4.10, 4.11 and 4.12, it is clear the see the consistence in gait cycle. Note that though this work was carried out on the right side of the body, the same conditions apply to the left side for able -bodied persons. The first double

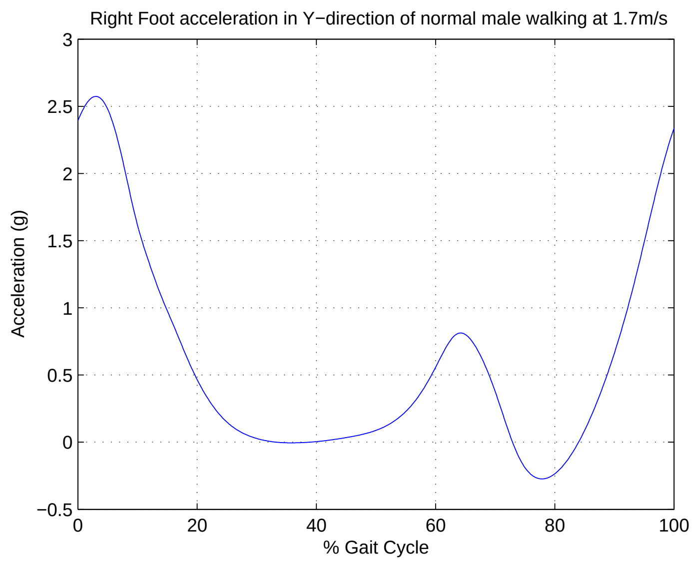


Figure 4.7: One complete gait cycle of foot Acceleration for Subject A in the X axis

support for the right side is the second double support for the left side and vice versa. This can be seen clearly in the force plate data shown in figure 3.4. Figure 4.8, shows the vertical ground reaction force data for subject A when walking at 0.9m/s as experience by the right plate and left plate. It is important that we emphasis the shape of each with respect to the walking speed. As can be seen for the subject when walking at 0.9m/s, it is visible that the mid-stance shape for the vertical ground reaction force is more flat as compared to the when walking at 1.08m/s. For the acceleration graphs can see the changes in the amplitude of the acceleration as the walking speed increases.

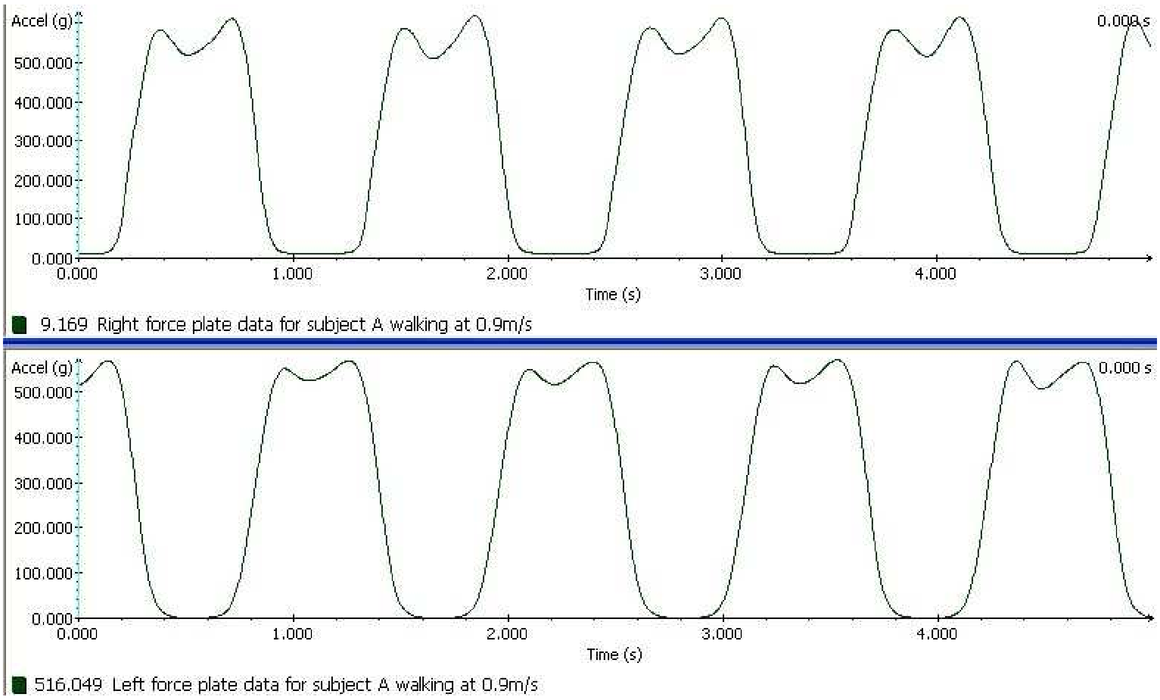


Figure 4.8: Vertical ground reaction force (N) on the left and right plates for subject A when walking at 0.9m/s. The force is plotted as a function of time (s)

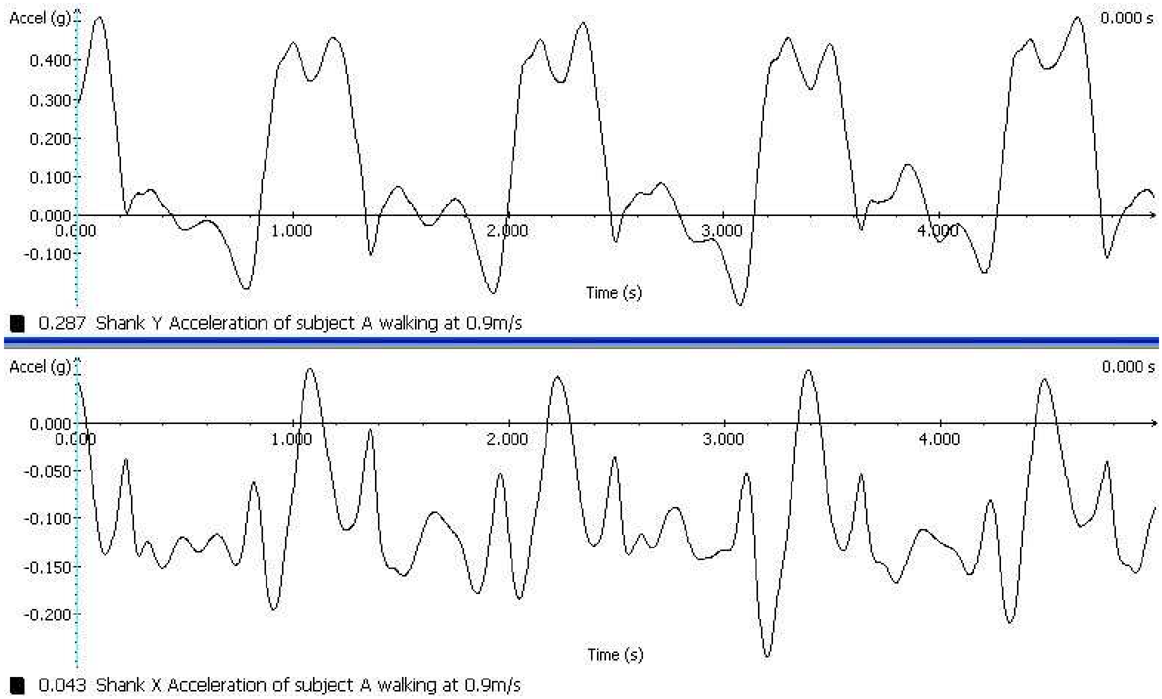


Figure 4.9: This shows 5sec shank acceleration data for subject A walking at 0.9m/s, note the labeling on the axis and compare the similarities for the subject when walking at 1.08m/s

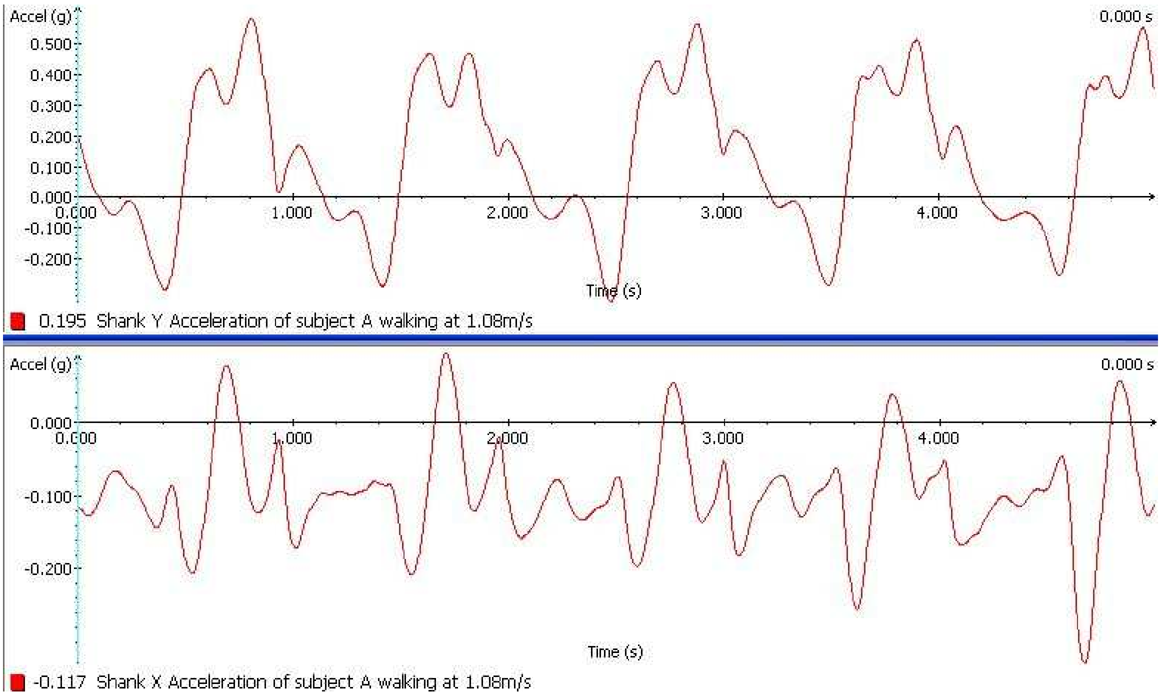


Figure 4.10: This shows 5sec shank acceleration data for subject A walking at 1.08m/s

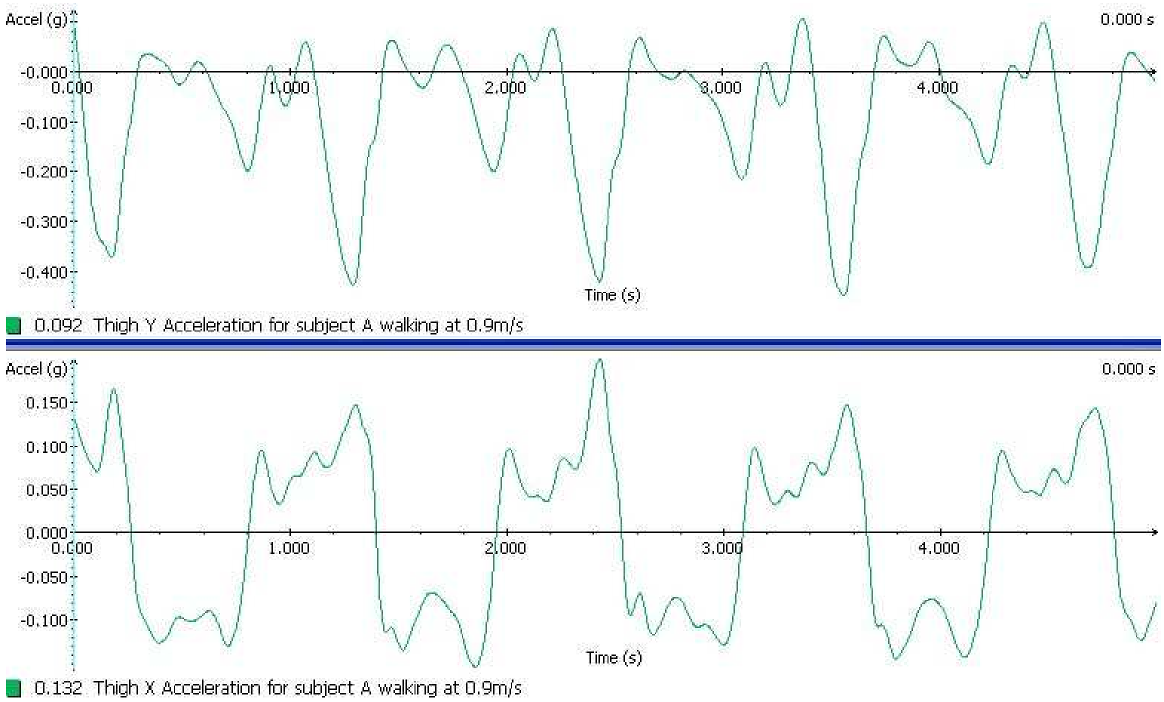


Figure 4.11: This shows 5sec thigh acceleration data for subject A walking at 0.9m/s, note the labeling on the axis and compare the similarities for the subject when walking at 1.08m/s

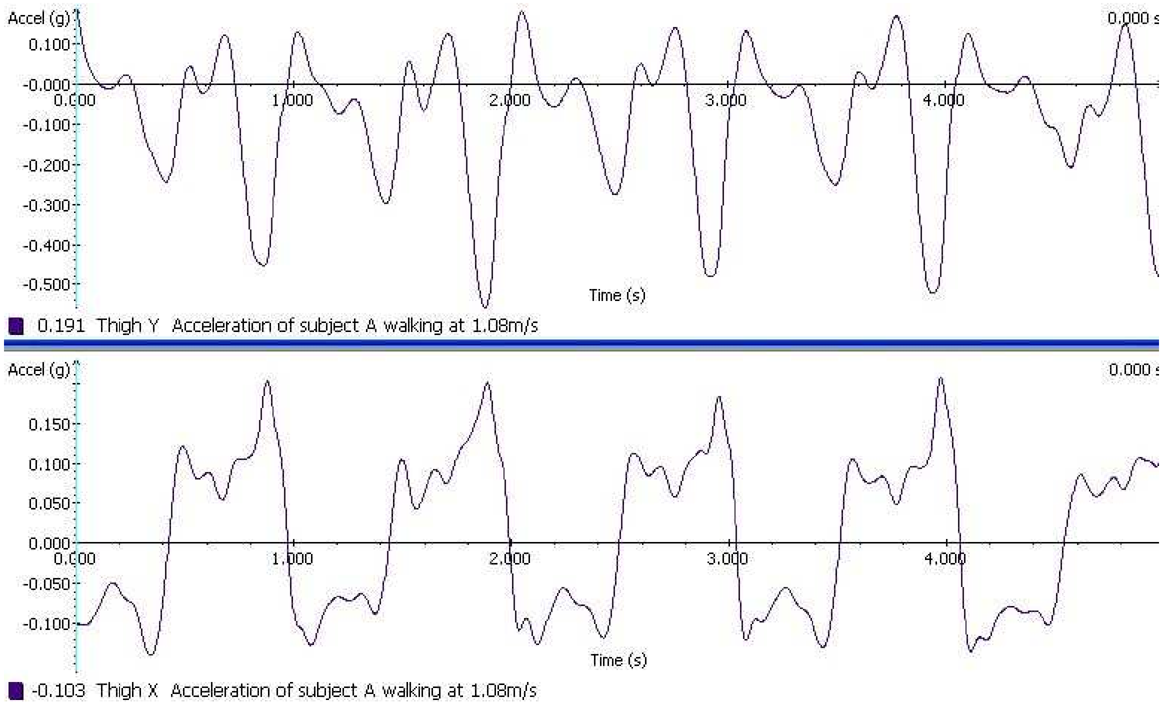


Figure 4.12: This shows 5sec shank acceleration data for subject A walking at 1.08m/s

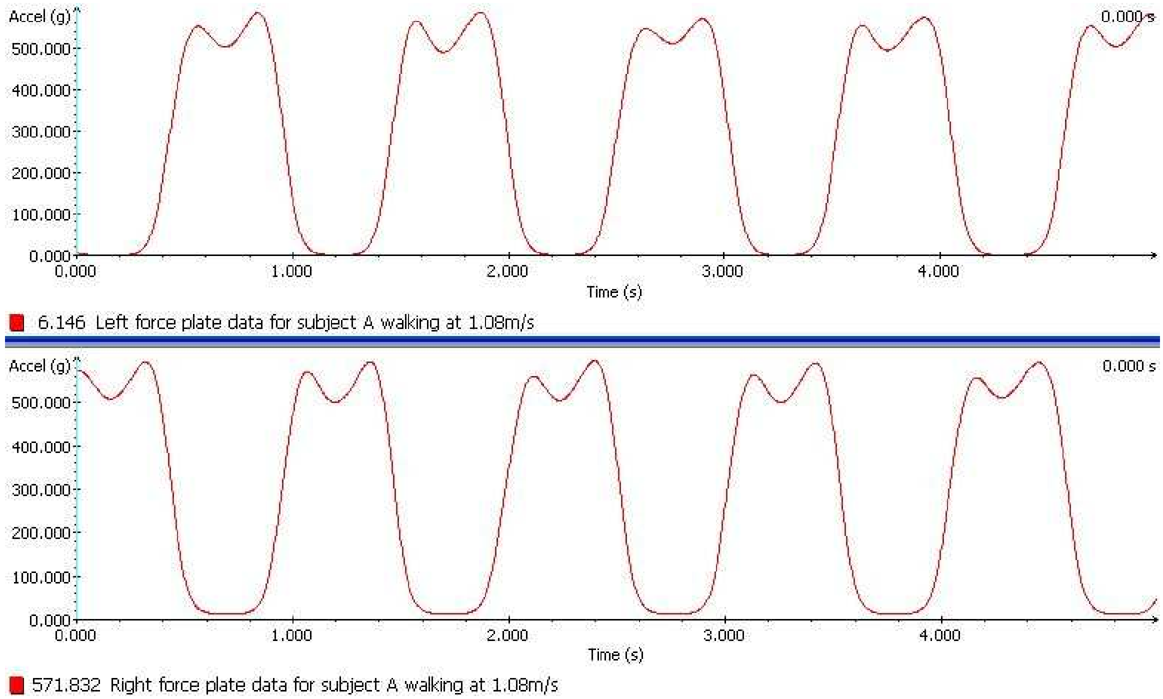


Figure 4.13: Vertical ground reaction force (N) on the left and right plates for subject A when walking at 1.08m/s. The force is plotted as a function of time (s)

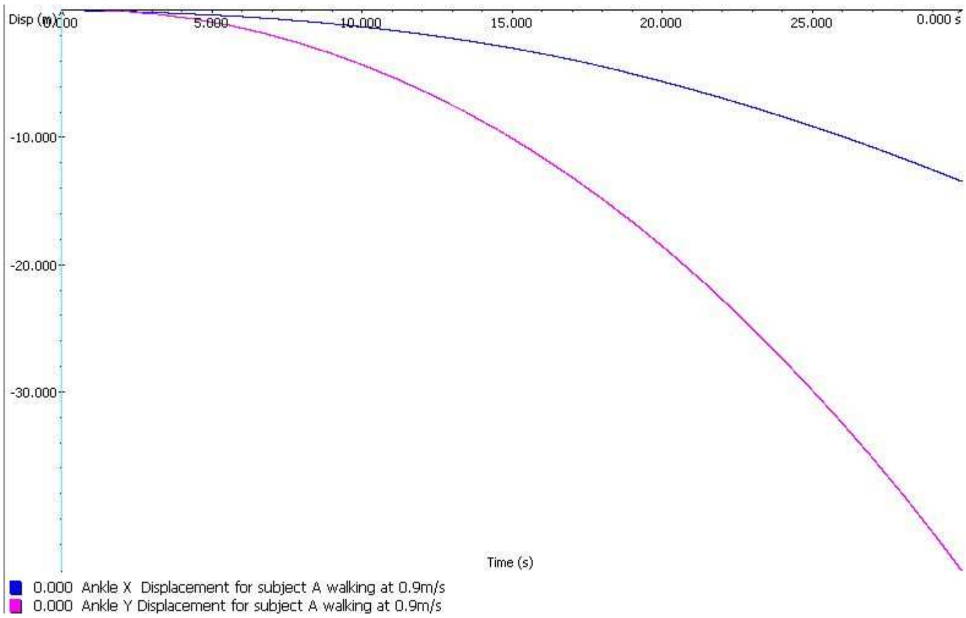


Figure 4.14: The ankle displacement measured in the sagittal plane when the subject was walking at 0.9m/s on a treadmill

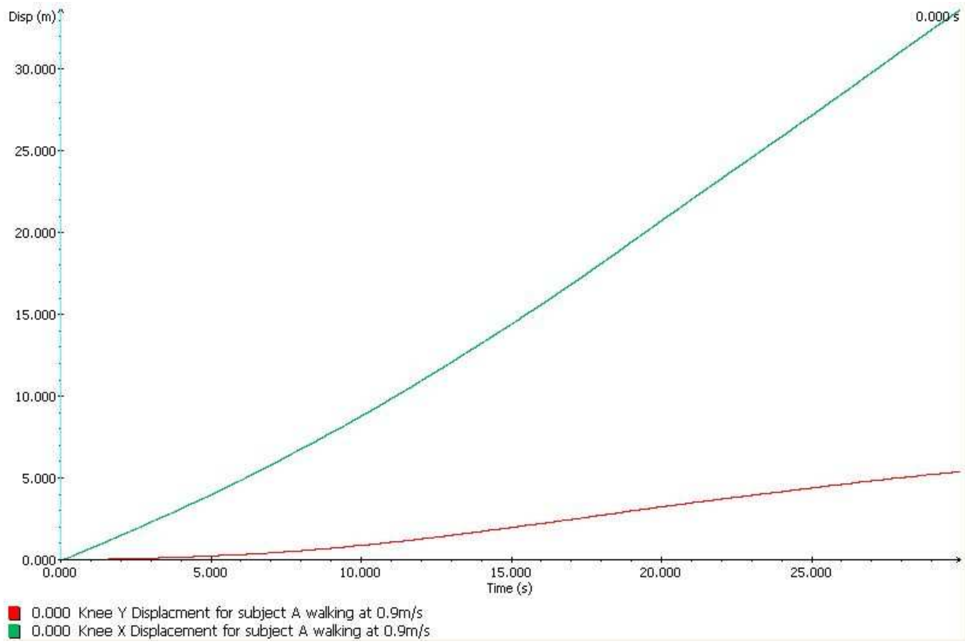


Figure 4.15: The knee displacment measured in the sagittal plane when the subject was walking at 0.9m/s on a treadmill

4.0.11 Acceleration Data

4.0.12 subject A - walking on treadmill

The results presented here are the acceleration data recorded during each gait phase for two subjects A &B. The acceleration values are in g (where 1g=9.8m/s/s). This values correspond to the peeks are which the phases are identified. The acronyms used stand as described in 3.2. The result presented in Table **??** shows the raw acceleration data for subject A during the experiment at a walking speed of 0.9m/s. Table **??** also shows the acceleration data for the same subject walking at the same speed during the second experiment. The idea here is to verify where the sensor will be able to produce the same acceleration values recorded in the previous experiment(note the first and second experiments took place on separate days). Table **??** is the output obtained using fuzzy similarity concept(max - min) to validate the results form both experiment on subject A.

From Table 4.3, it can be seen clearly that when the same subject perform the experiment at the same walking speed, the sensor produce an accuracy of 100% in recognizing terminal stance, Mid-Stance and Heel-off for the foot Y-X axis, and the shank x-direction movement. The overall efficiency of the sensor proved about 95% in recognizing the same when when walking at the same speed. It important to not the recognition accuracy of the sensors were all not 100% and this could be due to slight changes in the walking patten of the subject A. However, it is worth noting that the sensors were able to detect small changes in Hip and thigh movements accurately to about 98% figures.

Table 4.1: Acceleration data obtained from subject A walking at 0.9m/s on a treadmill during the first experiment

Segment	**HS**	**LD**	**MS**	**HF**	**PS**	**IS**	**MiS**	**TS**
Hip X	-0.161	-0.046	0.099	0.109	-0.095	0.041	-0.075	-0.142
Hip Y	-0.161	-0.046	0.099	0.109	-0.095	0.041	-0.075	-0.142
Thigh X	-0.164	-0.065	0.101	0.119	-0.094	0.042	-0.092	-0.144
Thigh Y	0.370	-0.151	0.016	-0.113	-0.083	0.005	-0.172	-0.425
Shank X	0.106	0.064	0.136	0.149	0.130	0.191	-0.037	0.102
Shank Y	0.241	0.026	-0.024	-0.145	0.084	0.360	0.361	0.217
Foot X	1.256	0.523	-0.008	0.121	0.512	-0.006	0.653	1.285
Foot Y	-0.421	0.030	-0.004	0.141	0.347	0.207	0.170	-0.463

Table 4.2: Acceleration data obtained from subject A walking at 0.9m/s on a treadmill during the second experiment

Segment	**HS**	**LD**	**MS**	**HF**	**PS**	**IS**	**MiS**	**TS**
Hip X	-0.161	-0.046	0.099	0.109	-0.095	0.041	-0.075	-0.142
Hip Y	-0.161	-0.046	0.099	0.109	-0.095	0.041	-0.075	-0.142
Thigh X	-0.161	-0.046	0.099	0.109	-0.095	0.041	-0.075	-0.142
Thigh Y	0.371	-0.134	0.020	-0.116	-0.072	0.007	-0.161	-0.422
Shank X	0.110	-0.084	0.122	0.149	0.139	0.187	-0.038	0.102
Shank Y	0.261	0.027	-0.031	-0.148	0.102	0.380	0.447	0.211
Foot X	1.295	0.492	-0.005	0.121	0.507	-0.005	0.756	1.290
Foot Y	-0.429	0.043	-0.004	0.145	0.351	0.202	0.154	-0.463

Table 4.3: Fuzzy similarity results - Subject A walking at 0.9m/s. This results shows sensor validation for subject A, when walking during three different experimental under the same condition.

Segment	**HS**	**LD**	**MS**	**HF**	**PS**	**IS**	**MiS**	**TS**
Hip X	98.2%	92.3%	86.5%	98.0%	97.8%	88.5%	81.6%	92.2%
Hip Y	85.4%	99.7%	84.5%	84.5%	95.6%	98.4%	99.4%	97.2%
Thigh X	98.1%	70.7%	98.0%	91.5%	98.9%	97.6%	81.5%	98.6%
Thigh Y	99.7%	88.7%	80.0%	97.4%	86.7%	71.4%	93.6%	99.2%
Shank X	96.3%	76.7%	89.7%	100%	93.5%	97.9%	97.3%	100%
Shank Y	92.3%	96.2%	77.4%	97.9%	82.3%	94.7%	80.7%	97.2%
Foot X	96.9%	94.0%	62.5%	100%	99.0%	83.3%	86.3%	99.6%
Foot Y	98.1%	69.7%	100%	97.2%	98.8%	97.5%	90.5%	100%

Table 4.4: Time lapse between gait phases as detected from the acceleration data recorded for subject A. Note, loading response is the time between initial contact and flat foot of the right foot. Midstance is the period between toe-off to mid-swing of the leg foot, and finally swing phase is the time between pre-swing and terminal swing of the right foot.

Subject	**Loading Response**	**Mid-Stance**	**Swing Phase**
Subject A	0.089	0.277	0.266
Subject B	0.08	0.334	0.409
Subject C	0.101	0.296	0.2 31
Subject D	0.133	0.378	0.210
Subject E	0.082	0.362	0.380
Subject F	0.106	0.401	0.432

4.0.13 subject A - Time stamps

In analyzing human gait, it is important to examine the period movement of each leg, Table **??** shows the time the right leg spent either in swing of stance phase with the corresponding ground reaction forces.

The times spent by each walking at their respective natural walking speeds can be represent in percent form. This shown in table 4.5,

As shown in Table 4.5, the male subjects spent approximately 0.701s to complete one stride when walking at a velocity of 1.08m/s, whiles the female subjects spent about 0.883s to complete the said task. According to [19], about 60% of the stride time is spent in the stance phase whiles 40% percent of is sent during the swing phase. This of course varies from subject to subject. Our results show an average of 43% of the stride time in the swing phase about about about 57% during stance phase.

Table 4.5: This shows the total time each subject to complete one stride represented in percentages.

Subject	**walking speed**	**Ttime (S)**	**LD**	**MS**	**S-Phase**
Subject A	1.08	0.632	%14	%44	%42
Subject B	1.08	0.823	%9.7	%40.5	%49.6
Subject C	1.08	0.628	%16	%47	%39
Subject D	1.08	0.721	%18	%52	%43
Subject E	1.08	0.827	%9.9	%43	%45
Subject F	1.00	0.939	%11	%45	%44

The stance phase can further be classified as mid stance or loading. From the results the maximum time delay occurred during the mid-stance with about 45% of the total stride time spent in this activity. The loading response was a little over 10%. With the assumption that walking pattern in able -bodied subjects is symmetrical, we can say that each subject spent approximately 45% of the time stride time swinging the left foot, and about 43% during mid-stance and a little over 10% for loading response.

Conclusively, stride time differ from each walking speed. As can be seen the total time for an constant velocity of 1.08m/s corresponds to about 0.701s for males and at a constant speed of 0.9m/s, this corresponds to about 1.120s whiles for the female subjects, this is 0.883s and 1.52s respectively.

The cyclic nature of human gait is very useful for reporting different gait variables. As you will later discover, detailed description of these gait parameters are presented and you will appreciate how minute changes can results in significant changes in gait parameters.

Table 4.6: X-Y dimensional acceleration of the hip, thigh, shank and foot of subject A walking over ground, recorded during each gait phase, the acroymns apply as discussed earlier.

Segment	HS	LD	MS	HF	PS	IS	MiS	TS
Hip X	0.135	0.017	0.019	-0.319	-0.389	-0.204	-0.284	-0.393
Hip Y	-0.105	0.151	-0.097	0.150	-0.018	-0.070	-0.121	-0.110
Thigh X	0.284	0.009	-0.017	0.095	0.061	0.056	0.141	0.163
Thigh Y	-0.655	-0.214	-0.088	-0.090	-0.030	-0.079	-0.317	-0.460
Shank X	-0.097	-0.004	-0.050	-0.135	-0.218	-0.160	-0.057	-0.081
Shank Y	0.164	0.093	-0.100	-0.008	0.448	0.472	0.402	0.330
Foot X	1.453	0.346	0.010	0.584	0.012	-0.255	1.135	1.383
Foot Y	-0.504	0.188	0.000	0.483	0.236	0.011	0.164	0.031

4.0.14 Subject A - walking over ground

Table 4.6, shows the acceleration data recorded on subject A when walking over ground. Note the duration for this experiment was 5sec due to some constraints as will be discussed later. One of the purposes for repeating the experiments for over ground walking was the investigate differences in walking on a treadmill and natural walking. As can be seen from the right foot X and right foot y shown in figure B.5,during mid-stance for the right foot, the acceleration was completely zero (g), this is correct because at this point the foot was in contact with the ground and there was no velocity to cause acceleration. Observing figure 4.4, you can see that there was a slightly negative acceleration during mid-stance of the right foot. This is because the treadmill was moving this caused negative velocity reaction.

As shown in Table 4.6, we have some exciting results - acceleration in the X-Y axis

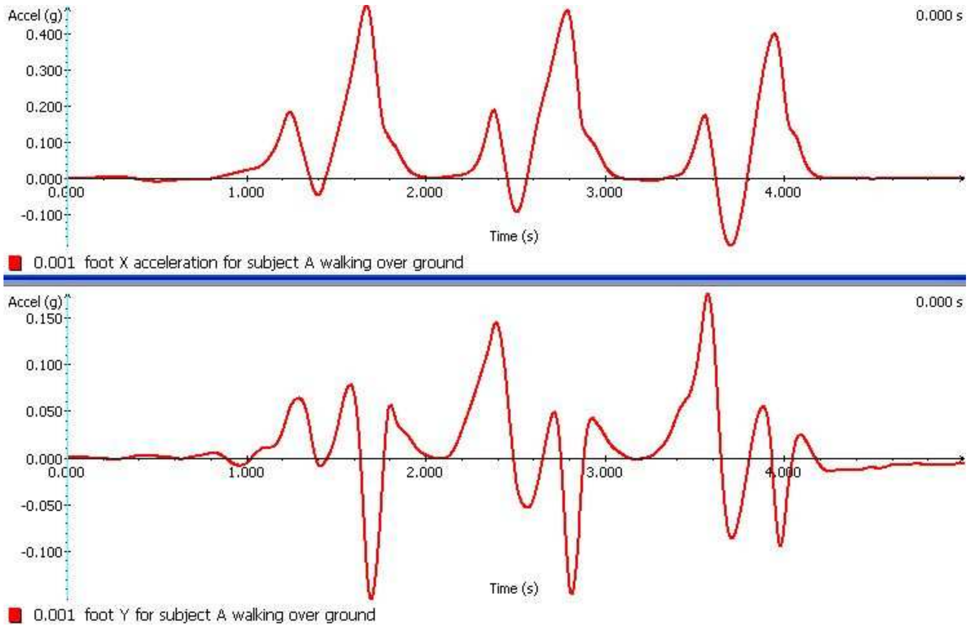


Figure 4.16: Acceleration in the X-Y direction of the sagittal of the right foot of subject A walking over ground ploted as a function of time. Note Table 4.6 shows the acceleration data recorded during this phase of the experiment.

of the sagittal plane for subject A walking at an unknown speed (more like to be the his natural walking speed). All thought during the earlier experiments, the subjects normally select a specific walking speed as their natural walking speed, this can not be said to be totally true since the subjects adapt to the speed of the instrumented treadmill and assume that to be their walking speed. From the table, we can see that the subject has his maximum positive acceleration of 1.453g at heel-strike in the X-direction with a corresponding -0.504g in the y-direction. Comparing this to when the subject was walking on treadmill, the highest acceleration occurred at heel strike with 1.256g in the x-direction and -0.421 in the y-direction. Although it is still preliminary to say walking on ground produces more accurate gait acceleration parameters, due to the time frame of the experiment, we can confidently determine the seven fundamental gait phases from the foot acceleration graph.

Chapter 5

Conclusion

5.1 Overview

We have proposed a new measurement method for distinguishing gait variables in able bodied people using an array of accelerometers. Results discussed demonstrated that gait phase can be determined from foot acceleration graph observed in the sagittal plane. The precision and reproducibility of the proposed method has been verified by experiments involving 6 subjects and a total of 36 experiments. The effectiveness of this method in application of human gait analysis has been verified by fuzzy similarity algorithm in validating the sensors This method enables us to measure and recognize gait variables in laboratory and real-life situations and eliminates the implementation of numerical analysis used in existing gait analysis techniques. As mentioned, current approach used in determining these gait phases consist of the use of force plate data and joint angles. The later when done with goniometers can be very unreliable due to changes in offset values and computational methods involved introduces noise and dimmishes the amplitude of the original signal. In this research, an improvement has been made to the signal acquisition and data processing technique thus eliminating the

use of integration or differentiation. With this, the slightest changes in gait variables during locomotion were recorded and taken into account for dynamic analysis. This is useful in clinical gait studies.

The architecture of the proposed system was described. And the results of its application in walking on treadmill, walking overground, sit-stand trails, and running on treadmill were presented. The results were compared and validated with vertical ground reaction force sensors and optical motion capture system with the proposed method showing high accuracies.

5.1.1 Future work

Preliminary results from this research shows inertial sensors have great prospect as far as gait analysis is concern. Current applications of these sensors however have been limited. With the prospect shown in this initial work, I do recommend that this work be carried out, in well defined phases. The duration of the experiment for the over ground walking was short, due to the length of the wire connecting the sensor system to the data acquisition card. It is recommended a wireless system should be implemented to transmit the inertial sensor data to the data acquisition system for processing. Better still, I portable data storage device to recorded the data and later download it for processing. This can help take more realistic gait data such as climbing stairs, walking through office environment etc.

The experiment was not carried out on both legs simultaneously, this was with the assumption that walking pattern in able bodied people is symmetrical. It is recommended that in future work, both legs should be examine simultaneously, in that the symmetrical concept can be investigated. Finally, it is recommended to carry out this experiment in a wide range subjects and have their acceleration and correspond force

data recorded.

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Appendix A

Rate Gryoscope

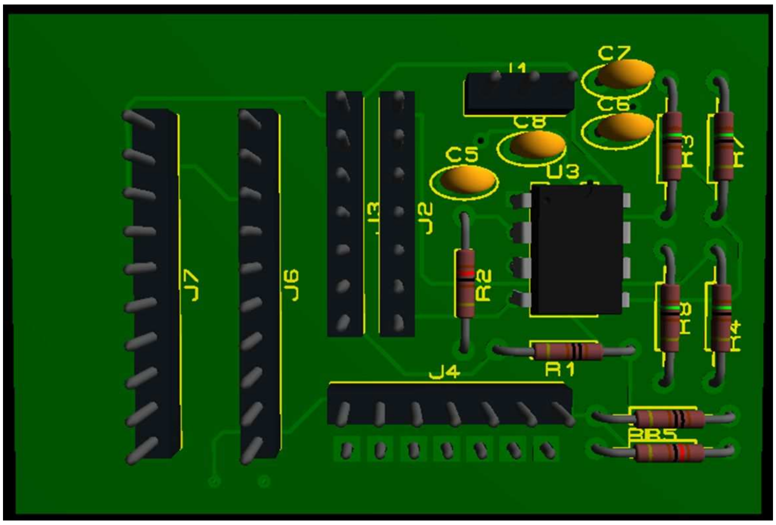


Figure A.1: Three dimensional view of the simulated sensor system.

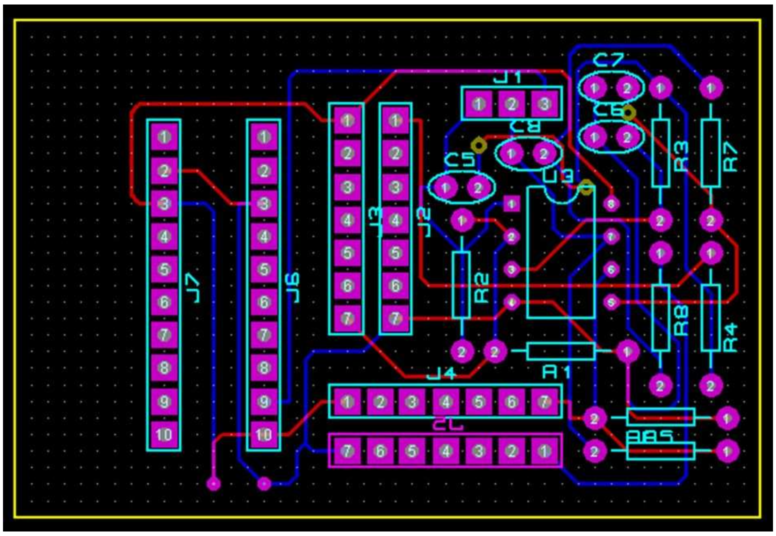


Figure A.2: PCB design of the proposed sensing system.

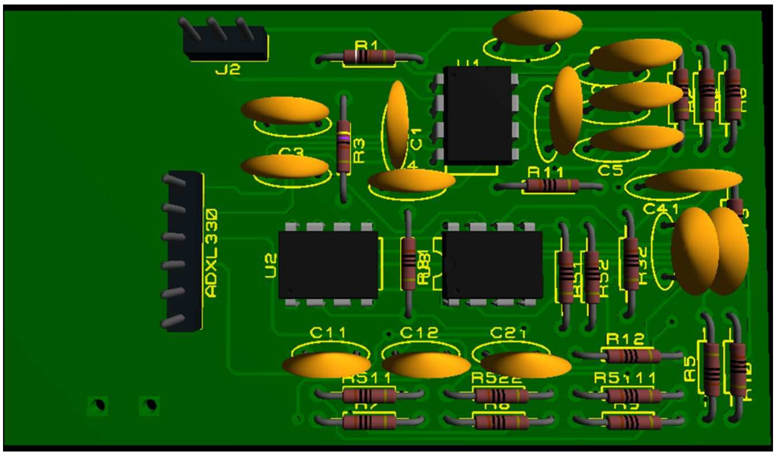


Figure A.3: PCB design of the proposed sensing system.

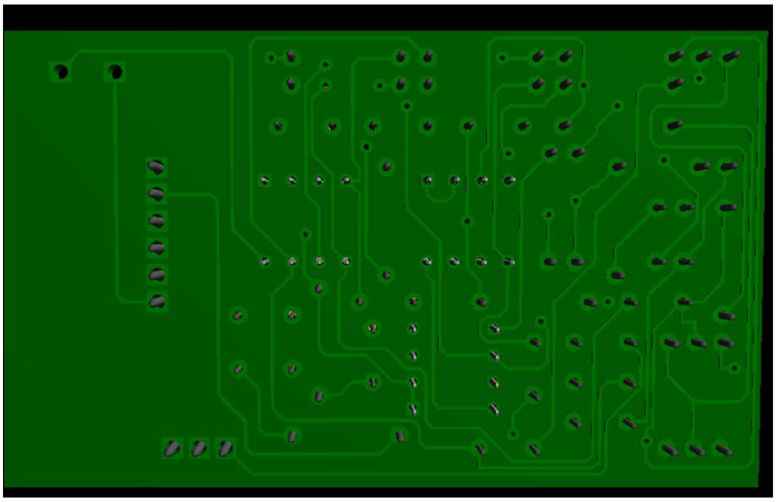


Figure A.4: PCB design of the proposed sensing system.

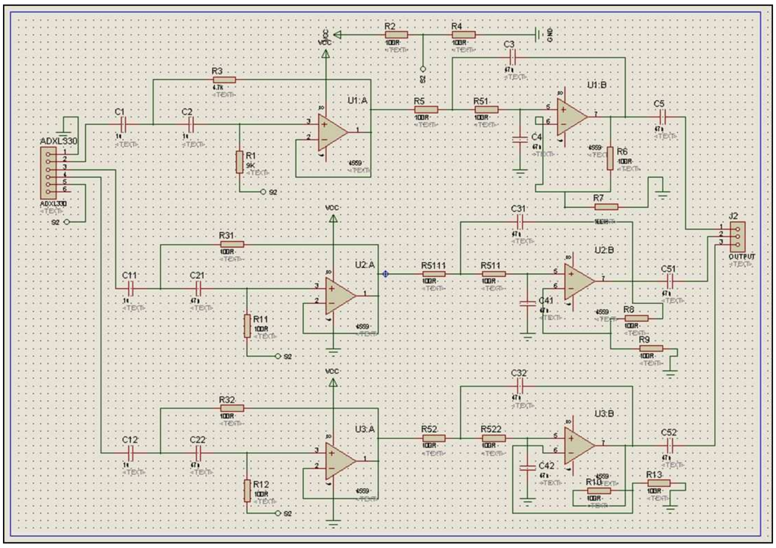


Figure A.5: PCB design of the proposed sensing system.

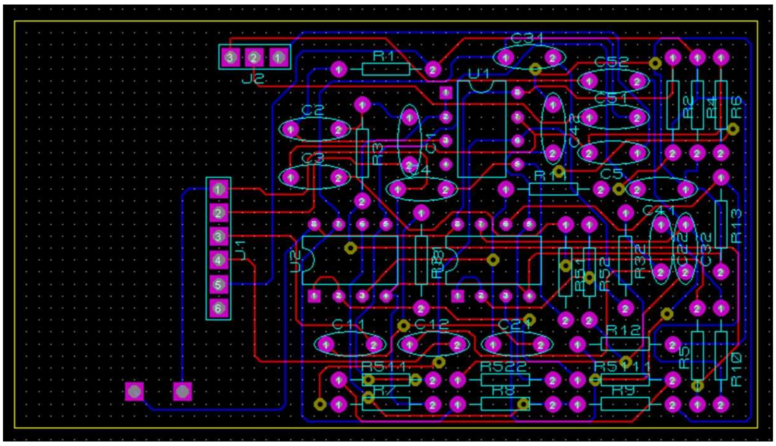


Figure A.6: PCB design of the proposed sensing system.

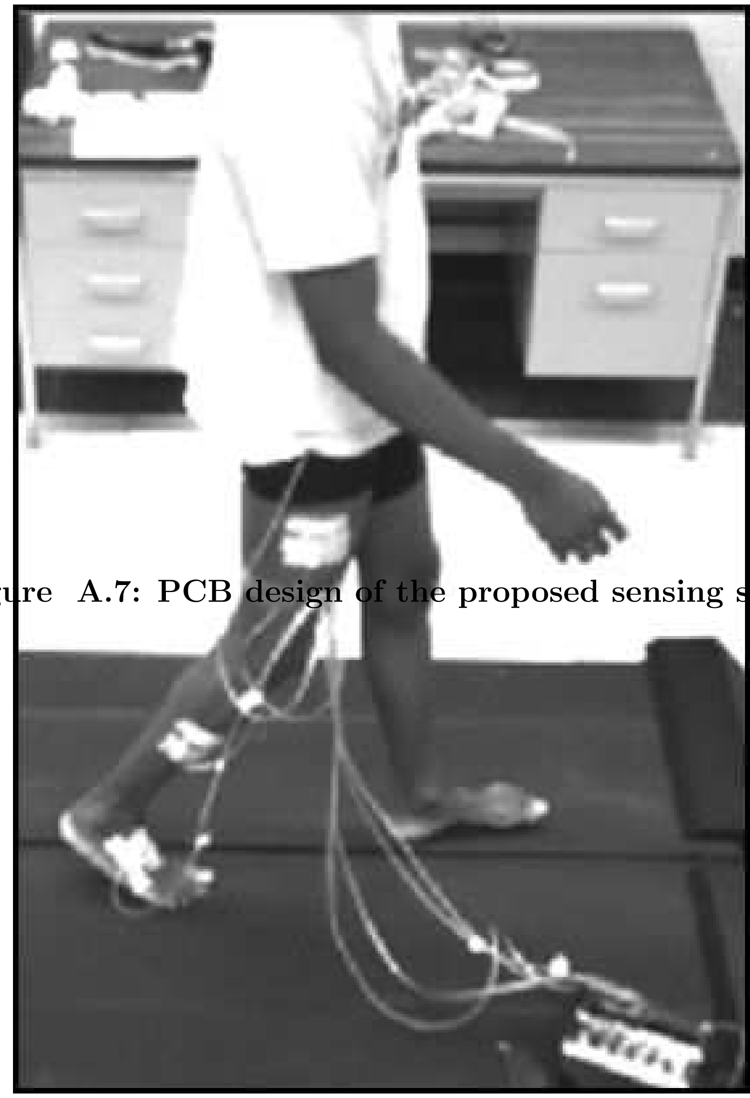


Figure A.7: PCB design of the proposed sensing system.

Appendix B

Subject A

B.0.2 Subject A - Ankle Acceleration Graphs

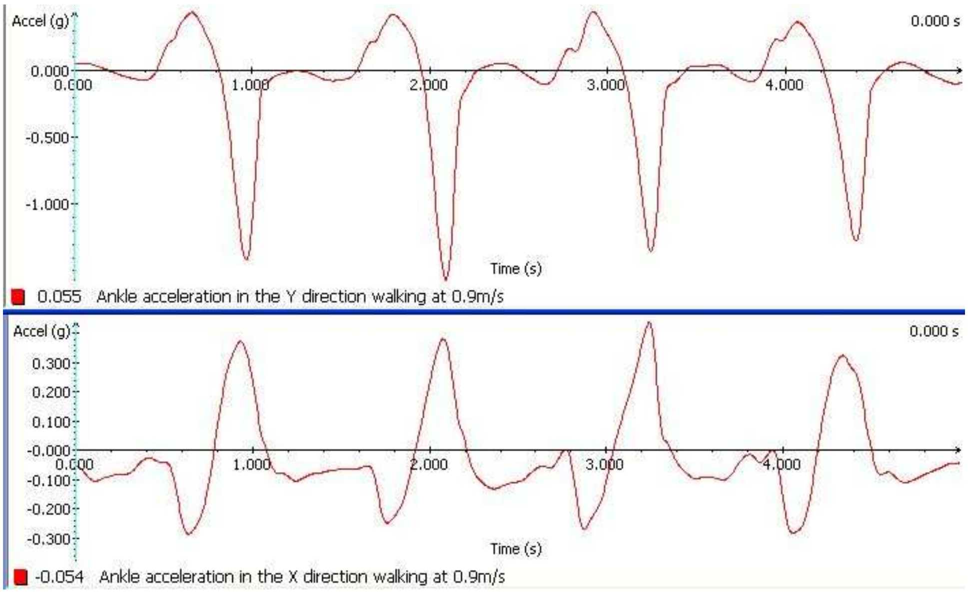


Figure B.1: The ankle acceleration measured in the sagittal plane when the subject was walking at 0.9m/s on a treadmill

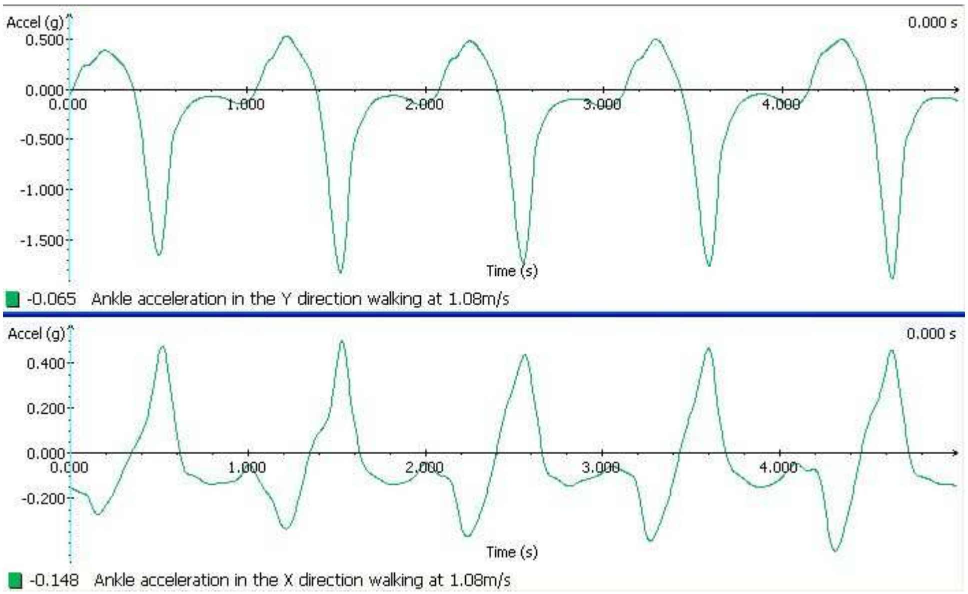


Figure B.2: The ankle acceleration measured in the sagittal plane when the subject was walking at 1.08m/s on a treadmill

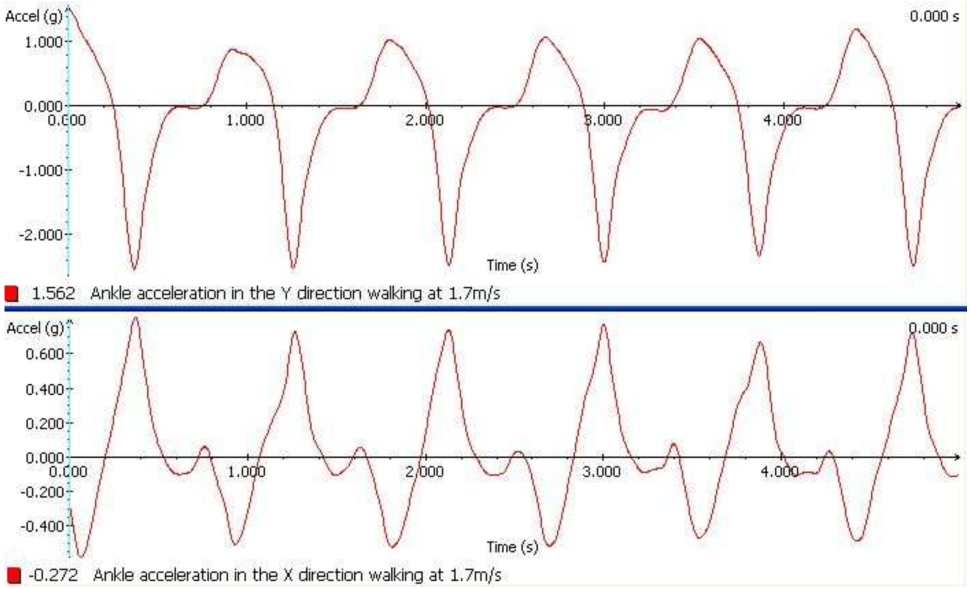


Figure B.3: The ankle acceleration measured in the sagittal plane when the subject was walking at 1.7m/s on a treadmill

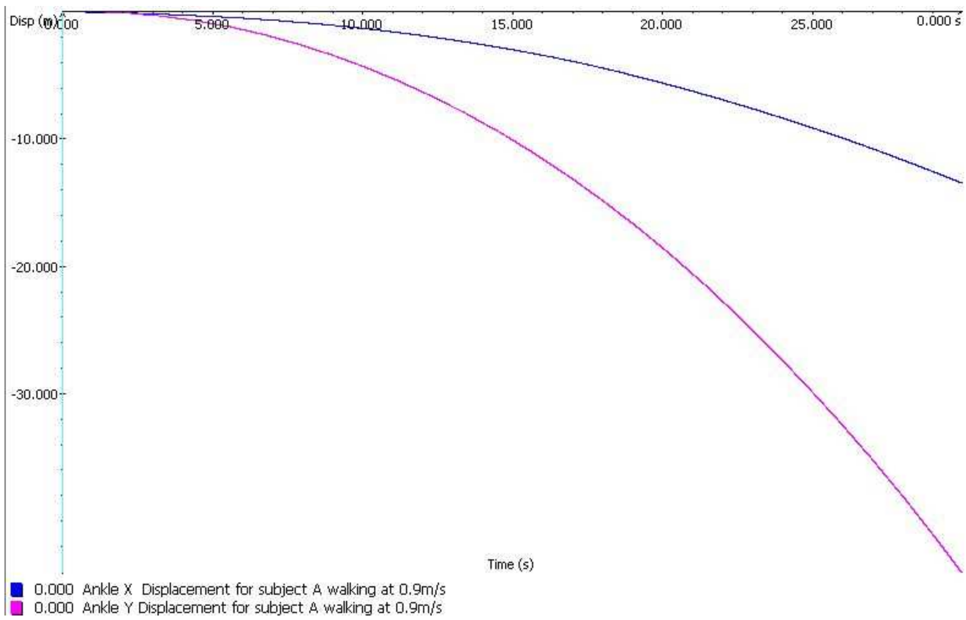


Figure B.4: The ankle acceleration measured in the sagittal plane when the subject was walking at 1.7m/s on a treadmill

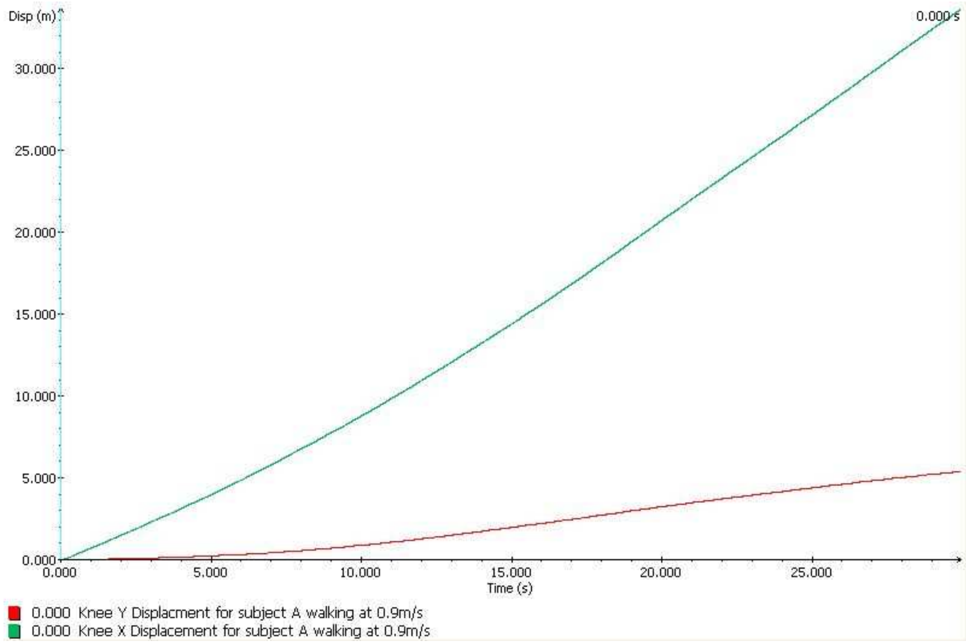
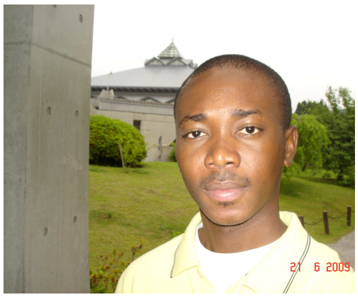


Figure B.5: The ankle acceleration measured in the sagittal plane when the subject was walking at 1.7m/s on a treadmill

Curriculum Vitae



John K Avor was born on November 23, 1983 in Adidome, Volta Region, Ghana, the eldest son of Rev. Alexander and Victoria Avor. He graduated from Kwame Nkrumah University of Science and Technology, Kumasi, Ghana with a bachelors degree in Electrical and Electronic Engineering in Spring of 2006. During his undergraduate studies, John had the opportunity to participate in valuable internships where he worked with Unified Communication (Engineering Systems and Services) and KRON Finance and Investment Ltd both in Accra. At Unified Communication, he assisted senior telecommunication engineers to provide internet access, establish local and wide area networks and troubleshoot failures in communication systems. He participated in several conferences and training seminars including, the annual African Telecom Summit "Bridging the Digital Divide".

Prior to the start of his master's program, John worked with Volta River Authority

(Utility Company) at Akosombo as Protection and Control Engineer under Transmission Systems Department. He interfaced with Senior Engineers to undertake several projects on power generation; transmission and distribution. He was also responsible for a Demand Side Management(DSM) project where he audited the efficiency of a Dynamic Motor Optimizer and served as the field supervisor for the DSM project to conscientizer raise the awareness of power optimization among consumers and also to reduce reactive power in the transmission system.

In Fall 2007, John entered the Graduate School to pursue his Masters in Electrical and Computer Engineering at The University of Texas at El Paso - Texas. During his masters program at UTEP he attended several national and international conferences ranging from professional to leadership and development. The recent international conference he attended was the IEEE International Conference on Rehabilitation Robotics - ICORR2009, Kyoto-Japan where he presented his published paper on "An approach of Sensor Fusion to Medical Robots". He also submitted a journal paper pending review at the IEEE-ASME Transaction on Mechatronics with the title "Application of inertial sensors to patient monitoring and rehabilitation devices". John hopes to start his PhD studies soon.

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