Project Number: ME-BJS-GA06

GAIT ANALYSIS OF RACE WALKING

A Major Qualifying Project Report:

Submitted to the Faculty

of the

WORCESTER POLYTECHNIC INSTITUTE

In partial fulfillment of the requirements for the

Degree of Bachelor of Science

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1. race walk
2. gait
3. efficiency

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Abstract

Vertical movement of the center of mass is wasted energy when walking. To evaluate this hypothesis a force plate and mathematical model were initially constructed. However due to manufacturing complications they were eliminated from the experimentation. Therefore, using only video capture and accelerometers we compared the subject's kinetic energy to their potential energy during both a normal and race walking gait to determine a subject's efficiency while walking. Upon comparing the two walking styles it has been determined that normal walking is more efficient then race walking.

ACKNOWLEDGEMENTS

We would like to thank Professor Savilonis, Professor Sullivan, Professor Furlong, Professor Hall and Dave Fogaren for all the help they gave us in completing this project. Without their help we would not have been able to complete this project.

TABLE OF CONTENTS

TABLE OF FIGURES

Table of Tables

Executive Summary

This report is looking at comparisons between normal walking and race walking. We tried to compare both the mechanical work throughout the leg and the efficiency of the two gaits while walking. This included the use of mathematical models and the design and fabrication of a dual axial force plate.

Recent changes in race walking rules require that walkers maintain a straight leg through the gait until that leg crosses under the center of gravity. We believed that this could cause stress on the knee and hip not experienced in a normal walk. We had all subjects walk in both fashions in order to compare the differences between the walks in all subjects. In addition to testing mechanical work we looked at the efficiencies between the two gaits. We wanted to see if one walk was more efficient then the other and also if race walkers had a naturally more efficient normal walk then a control group.

Gait efficiency was defined as the correlation between potential and kinetic energy. The subjects' centers of gravity were tracked for the duration of one stride to calculate their horizontal velocity and vertical position. In this model, any vertical motion that does not translate into horizontal motion is assumed to be unnecessary energy consumption.

The difference in joint moments and forces between race walking and normal walking could have been analyzed if horizontal and vertical forces and coordinates of segment center of masses were recorded. This model assumes the foot, shank, and thigh to be three separate segments. Thus, the moments could be calculated using moment equations and Winter's model for center of masses of various body segments.

The mathematical models mentioned required the accelerations at the ankle, knee, hip, center of mass, video capture of the vertical movement, and the horizontal and vertical force of the subject. We calibrated four MEMS ADXL 276 accelerometers in both the x and y direction and collected readings with LabView. For the video capture we used a digital camera to collect the movement and the software package ImageJ to analyze it.

We constructed a dual axial force plate light enough for transport. Strain was measured from two I-beams in each corner of the force plate perpendicular from the other to obtain vertical and horizontal measurements. Strain gages were applied, VISHAY SR-4, in a full Wheatstone bridge. The strain was then input into LabView and recorded.

Complications arose during the calibration of the force plate. The I-beams which measured the strain were machined at half the thickness which was specified in the drawings. This nearly quartered the amount of weight the force plate could hold. Attempts were made to continue with the original experiment but due to time and budget constraints the mechanical work section had to be dropped.

Four subjects were tested with both a normal and race walking gait. Five trials of both walks were analyzed using the ImageJ program and data taken was then put in excel. Using our model we found both the potential and kinetic energy of the subjects during their trials and compared the two throughout the walks.

A statistical t-test was used to analyze the difference in gait efficiency between normal walking and race walking. The race walking gait among subjects was less efficient than normal walking among all subjects using a significance of $\alpha=.05$. This proved that there was a significant difference in the two walks in our subjects.

More data about the mechanical differences of the leg's movement during the gait could possibly be seen with the use of mathematical models and equipment to capture ground forces and joint accelerations. We recommend that the experiment be repeated in an attempt to collect more information about the internal mechanical work throughout the leg.

I. Introduction

A. History of Race walking

The sport of race walking has a rich heritage in the history of man. People are competitive by nature and discovered this competitive spirit even in walking. The English started betting on race walking 400 years ago. Lords, dukes and other affluent men would take sport in racing each other's servants. In most cases the races would either go on for long periods of time or cover a vast amount of distance to allow ample time for betting [2, 7].

This continued until the late 1800's. By then race walking was becoming even more popular throughout the country. It was now a common man's sport and it was the second most bet on sporting event to horse racing. Because there were no formal leagues or rules to regulate the sport, judges simply had to insure the competitors were walking and not running in an attempt to gain advantage. One judge's views of a walking gait could differ vastly from another's so it was difficult to obtain a fair ruling from one race to another.

In 1880 the Amateur Athletics Association was formed and two rules were applied to race walking. The rules stated that all disqualifications were to be determined by the judge and all of their rulings were final. This did not help the sport develop because it simply stated what was already accepted as fact. Rules to determine the definition of a walking gait would need to be determined if the sport was to gain any credibility.

The addition of race walking to the Olympics in 1908 helped bolster the efforts to develop more rules defining walking. In 1912 the International Association of Athletics Federation (IAAF) founded a walking commission to recognize race walking as an official sport. However, it wasn't until 1928 that the IAAF made any efforts to add regulations. Finally in 1928 they stated that while walking one could not break contact with the ground. While not encompassing all situations this was an important step to get judging criteria. Judges could now disqualify people and be able to back up their claims with a specific infraction.

This was still not enough for the race walking community and efforts were continued to define the rules and make the sport fair for all participants. It took over twenty years for the rules to be adapted and in 1949 the rules were basically restated, but in a way that clouded judge's eyes even more. After a few controversial decisions in major matches, the rules were clearly stated in 1956. The rules now clearly defined two separate areas. Contact must be made with the ground at all times, and at one point in the stride a straight leg must be made. This cleared up many controversies for judges and brought clarity to a sport where craziness prevailed for so long.

This rule has stayed the same except for two minor modifications in 1972 and then again in 1996. The 1972 rule changed the need for a straight leg at any time during the stride. Now the heel would have to land with a straight leg. This was due to the increased speed at which the race walkers gait was kept. Judges could now always look for the straight leg at the same time during the gait to increase their effectiveness. The 1996 rule change was along the same lines and had the walkers continue to keep the leg straight until their body passed over the leg. This allowed the judges even more time to analyze straight legs during a walker's gait.

B. Clinical Significance

Race walking tends to yield fewer injuries than running and other strenuous activities, but there is a significant amount of stress put on certain joints that can produce complications over time. It has been discovered that the amount of injuries race walkers suffer are less than other strenuous sports, however they are not absent and must be addressed (the average race walker suffers 0.156 injuries per year according to Francis et al. [1].) The change in the rules of race walking in 1996 regarding the knee movement may further affect the rate of such injuries. Until 1996, one was allowed to have their knee slightly bent upon impact of the leading foot as long as it was straightened out by time it reaches the person's center of gravity.

In the beginning of 1996, the rules changed stating that the knee must be straight when the leading foot impacts the ground. Ever since these rule changes, there has been controversy whether or not this method of walking produces higher forces in the joints involved in race walking, specifically the knee and hip joints. Martin Rudow, the U.S. national men's race walking coach states, "Upon impact, the supporting leg should be relaxed instead of locked … Slightly tensing the quadriceps will help prevent knee joint fatigue and lessen the chances or injury" [2]. His opinion appears to support the fact that the new rules may prevent a higher rate of joint fatigue and injury since a race walker cannot "tense the quadriceps" or relax the supporting leg.

Table 1: Frequency of injuries in specific areas resulting from race walking [1]

Location	Frequency	Percentage of All Injuries at This Location With No Prior Orthopedic History
Knee	107 (21.3%)	66.0%
Foot	104 (20.7%)	75.0%
Shin	64 (12.7%)	94.2%
Hip	58 (11.6%)	74.5%
Back	46 (9.2%)	41.5%
Hamstring	41 (8.2%)	81.1%
Ankle	37 (7.4%)	84.6%
Groin	13 (2.6%)	
Thigh	9 (1.8%)	
Shoulder	6 (1.2%)	
Neck	4 (0.8%)	
Abdomen	2 (0.4%)	
Iliotibial band	1 (0.2%)	
Pelvis	1 (0.2%)	
Other	9 (1.8%)	
Total	502 (100.1%)	

Some common injuries in runners and race walker alike relating to the knee are as follows. Chondromalacia is one of the most common causes of pain in the knee [3]. Usually caused in younger, more active athletes, it is usually referred to as "runner's knee." This is caused when the patella, the underside of the kneecap becomes irritated due to constant strain on the joint. This problem occurs when the kneecap rubs unequally on one side of the knee joint, which directly results in inflammation. Therapy and the use of non-steroidal inflammatory medication (NSAIDS) have also been proven to be beneficial. There is a severe limitation when one is diagnosed with Chrondromalacia. The individual who experiences the condition must refrain from any activities that may irritate the patella for several weeks. Athletes with Chrondromalacia often must find alternate means of training to remain healthy, which quickly becomes a limitation.

According to the paper by Francis et al., the most common race walking injury is one that involves a hamstring strain. This occurs because of a tear in the hamstring muscles which include the semitendinosus, semimembranosus, and biceps femoris. In race walking, there is an extreme amount of force that the hamstrings must move, sometimes more than running requires. The increased loading of the muscles result in increased fatigue and injury. The severity of the hamstring strain varies from minor or "micro" tears that usually go undetected, or the complete rupture of the muscle [4]. Depending on the severity, the treatment varies. When the muscle ruptures, the athlete may not be able to even walk without the aid of crutches, but this type of injury usually does not occur in race walking.

Shin splints refer do symptoms more than the diagnosis. Sharp pains are noticed in the shin area and can be the result of muscle problems, bone problems, or the attachments of the muscles to the bone. The pain is commonly sensed in the front of the lower leg [5]. The most common cause of shin splint is medial tibia stress syndrome, followed by stress fractures and exercised induced compartment syndrome. This usually is the result of a rise in training intensity, which occurs commonly in race walkers. This type of injury commonly plagues runners, and race walkers alike, and is very troublesome because it can take months to go away. Often athletes must change their training regimen to lessen the affects of shin splints. Low impact exercises are recommended as treatment, along with special stretches and the use of prescription anti-inflammatories.

Table 2: Specific Diagnosis of Injuries Reported by 247 Race walkers [1]

Injury	Frequency
Hamstring strain	24
Shin splints	19
General ligament sprains	17
Other muscle strains	17
Tendinitis, foot	16
Spinal injuries	10
Tendinitis, knee	9
Iliotibial band syndrome	8
Sciatica	8
Plantar fasciitis	8
Chondromalacia patella	7
Groin pull/strain	6
Anterior tibial tendinitis	6
Arthritis, knee	6
Ligament strain, knee	6
Stress fracture, foot	6
Blisters, foot	6
Arthritis, other locations	6
Stress fractures, other locations	5
General tendinitis	5
Sesamoiditis	5
Nonspecific pain	5
Bursitis	5
Anterior compartment syndrome	4
Muscle spasm	2
Other conditions	31
Total	247

Foot injuries account for about 20.7% of race walking injuries right behind the knee injuries. The most common injury involves the inflammation of the band of tissue that runs from the heel along the arch of the foot [6]. The resulting pain is very bothersome. One notices it whenever they first use there feet such as in the morning or after resting for a prolonged amount of time. The pain may go away quickly, but returns after excessive standing or walking. This symptom alone can prevent race walkers from participating in their daily workout. The reason that this condition occurs is most likely the nature of the sport. Running and race walking can damage the fibrous tissue in the foot simply from repetition. Much like other injuries, rest is the most effective way of treating this condition. This is very undesirable, especially for professional race walkers who must train daily. Also, stretching is effective along with anti-inflammatory drugs reduce the pain felt by the athlete.

Hip injuries also appear to be common among about 11% of race walkers. A frequent hip injury that affects race walkers is hip bursitis. Bursa is the fluid filled sacs between uneven surfaces that helps produce smooth movement. When bursa becomes inflamed the individual experiences pain. Every movement results in pain, from moderate to severe, and can be very uncomfortable for the individual. The key to treating hip bursitis is to control the inflammation in the bursa. The way to control the inflammation is rest, the use of anti-inflammatory medications and ice. Sometimes, the inflammation causes an excessive amount of fluid and needs to be collected by a needle. Often when this is done, cortisone is administered. The common amount of time required before the athlete can continue his or her activities affectively is about six weeks. Surgery is rarely an option except in extreme cases.

It is evident that race walking injuries need to be minimized in order for athletes to maintain a constant training regimen. It is unclear whether the rule in 1996 requiring the knee to be straight as the foot strikes the ground will impose more forces against the knee or other lower extremities. We will be able to compare the internal forces of the knee and hip joints and compare them to everyday walking. This will be a good indication of whether or not race walking places higher loads on the joints that can be harmful after a certain amount of time.

C. Previous Studies

Cairns et al. showed that race walkers exhibited increased maximal ankle dorsiflexion, knee extension, angular displacement of the pelvis, reaction forces between the foot and the ground, plantar flexion moment, and external knee hyperextension moment. This study entailed the use of video and force plate data to study internal joint forces and moments acting on the ankle, knee, and hip. The foot, shank, and thigh were assumed to be three separate body segments which allowed the use of moment equations [7]. Winter's model for body segment center of mass was used to calculate the center of mass of each segment [17]. Franklin et al. have shown that the physiological and psychological profiles between marathon runners and race walkers are similar [8]. Based on these results it is clear that the biomechanical parameters vary between normal and race walking gait. Menier and Pugh and Marchetti et al. determined race walking to be less efficient than running at velocities greater than 8 km/hr [9, 10]. It has also been reported by Cavagna and Franzetti that race walking at higher speeds may reduce energy expenditure compared to normal walking [11].

Recent studies have presented a variety of models for gait analysis. Meichtry et al. have developed a technique for assessing external work and power in normal gait using trunk acceleration. This model placed accelerometers at the trunk and center of mass of the subject and calculated power as the sum of the product of the acceleration and velocity vectors and the product of gravity and vertical velocity. The power was then integrated to calculate total external work [12]. Gider at al. proposed and analyzed an assessment method for gait efficiency. This study validated the efficacy of a tracking an optical marker at the trunk to determine the cross correlation between normalized time courses of kinetic and potential energies where the correlation, for optimal gait efficiency, should be 1. The results were compared to other established gait energy consumption methods by measuring oxygen consumption and heart rate. It was determined that this technique is a practical method for determining gait efficiency in normal gait [13].

D. Problem Statement

There is little biomechanical research in the area of race walking. Specifically, this study will determine gait efficiency, and internal forces acting on the knee and hip using the aforementioned models. These results will help evaluate whether the recent rule changes to race walking can cause adverse effects on the body that could lead to injury.

A race walker would be interested in determining how efficient their walk is. It would be beneficial to know whether regular walking or race walking is more efficient. This can lead to race walkers improving their walk and retain more of their energy. Upon comparing a regular walk to a race walk, it will be determined which is more efficient.

The rule change in 1996 most likely affects the forces throughout the leg and hip joints. It is our goal to determine if these forces are significant enough to lead to injury over time. If this is the case, recommendations to the race walking governing body will be considered. One possible solution could have the knee slightly bent on impact which will result in a lower amount of energy being transferred throughout the leg.

E. Expected Outcomes

A race walker must be able to maintain a faster speed for longer amounts of time, using the least amount of energy over the period of a race. Seeing how professional race walkers can complete a mile in five minutes, it would take a great deal of efficiency to maintain this pace. Running, on the other hand, would be very inefficient. The high rate of up and down motion would equate to wasted energy because it takes energy to lift one's self up and down and this is not used directly in the forward motion. Ideally we are going to want no vertical motion as this would relate to all of one's energy being applied in a forward motion.

For these reasons and the simple fact that a race walker is trying to walk as fast as possible without consuming much energy, we expect to find that a race walk will be more efficient than traditional styles. When comparing the relationship between potential and kinetic energy, the desired ratio would be close to one. This means that all of the potential energy is converted to kinetic energy, and in our case, towards the forward motion. The numbers gathered should show that the correlation for regular walking is statistically significantly less than that of race walking. Therefore the race walk can convert more of its potential energy into forward momentum.

Furthermore, the stresses in the joints are going to be measured. Since the rule change in 1996, race walkers are forced to land with straight knees. This translates to more of the energy being applied to the joints, rather than muscles if the legs were slightly bent. When we gather the force readings of the race walk, initially we will notice that the heel strike is much larger than that of regular walking. It has been hypothesized that it also would be larger than if the knee was bent and the muscles can dissipate some of the energy. We should notice that based on our anatomical models, that more force is being applied towards the knee, hip, and ankle joints.

The fact that there may be more force being applied to the joints may be interesting for race walkers to know, because a higher rate of injury may occur. Being that the new rule is relatively new, not all of the effects of the rule change are observed. It would be very beneficial to determine if the new rule change led to increased injury. Necessary recommendations can be made to governing bodies to inform of any impending health risks as a result of the rule change in 1996.

II. MATERIALS AND METHODS

A. Force Plate Design

The force plate designed in this experiment had to satisfy the need for measuring forces on two axes. Additionally we wanted the force plate to be portable. Also it needed to be capable of holding an average sized human subject while walking. After researching possible designs, we designed a force plate using two main I-beams, incurring strain while a subject walked over the plate. Two I-beams at each corner would measure the strain at each corner of the force plate in both the horizontal and vertical direction. In order to measure strain in each of the directions, strain gages were attached to the I-beams in a full Wheatstone bridge, configuration I.

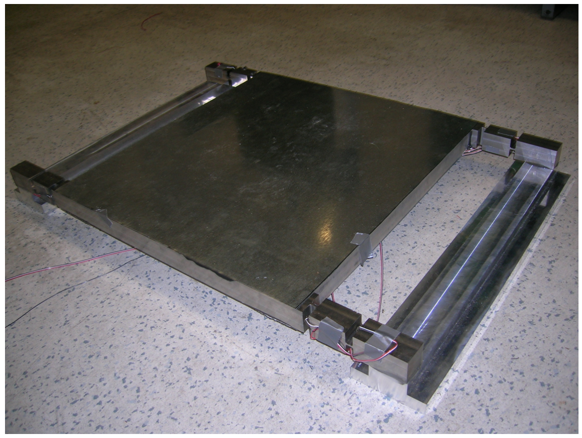

Figure 1: Force Plate

We wanted the walking subjects to feel like they did not have to focus on taking an accurate step on the force plate so the area for the force plate chosen was roughly .5 meters squared. This would give the walker an adequate area to place their foot inside the force plate. By having the force plate portable we can also take the subject's average stride length and position the force plate so that their average stride length would match the placement of the force plate.

In order to create appropriate dimensions, the materials for the force plate had to be selected accordingly. Our choices were mainly between steel and aluminum for materials. Aluminum offered a lighter design while steel is significantly stronger. We finally selected a combination of the two materials using steel for the beams experiencing strain and aluminum for the supporting base and the internal supports of the force plate. In order to ascertain the length and width of the I-beam we first needed to determine the maximum force that could be applied to the force plate.

The equation we used to gain the maximum force was force of the subject times the number of g's (gravities) on impact plus the weight of the force plate:

$$P_{\max} = P_s(g_{\max}) + P_p \tag{1}$$

For Eq. 1 we used generous assumptions to assure safety of both the force plate and the subjects walking on it. We selected the subject force (Ps) as 250 pounds or 1112 Newtons. We used the number of G's on impact (g_{max}) as 8 and rounded that number to 9000. The weight of the force plate on top of the two supporting beams would be negligible in relation to the force of the subject. Pmax was then multiplied by a safety factor of 2 to arrive at a Pmax of 18000 Newtons.

Once the maximum force that would be applied to the force plate was known, the dimensions of the I-beam needed to be determined. In order to know the dimensions of the I-beam we used the standard formula for stress in a beam is:

$$S = \frac{Mc}{I} \qquad (2)$$

S is the stress in the beam, M is the bending moment, c is the distance to the neutral axis, and I is the moment of inertia in the strain beam. Each component of this equation can then be isolated in order to arrive at a version relating the dimensions of the I-beam to the stress in the beam.

The three equations for M, c, and I are as follow:

$$I = \frac{bh^3}{12} \qquad (3)$$

$$c = \frac{h}{2} \qquad (4)$$

$$M = P_{\max} l \qquad (5)$$

In these terms P_{max} is the maximum force applied to the force plate, *l* is half the length of the I-beam, h is the thickness of the I-beam, and b is the height and width of the overall beam. The ratio of width of the beam to thickness of the beam will be referred to as n. In order to keep the crosstalk between the vertical and horizontal beams, it was recommended to keep the ratio at least 7. We can insert these variables into one equation and solve for h arriving at:

$$h = \left(6 \frac{P_{\max} l}{n \varepsilon} \right)^{\frac{1}{3}} \qquad (6)$$

For this equation n will be 7, ε is the strain, and we used the yield strength of 1080 carbon steel as 585 MPa, Pmax will be 18000 N, and we arbitrarily picked l as .01 m. We wanted a short thin beam because the longer the beam the thicker its dimensions. After inserting the numbers into the equation we arrived at a beam thickness of .005 m and a beam height and width of .035 m. The distance between the two beams was determined to be d and was recommended at 10 times the thickness of the I-beam. This made us select a dimension of .05 m for d. The following is a drawing of the beam and all the variables.

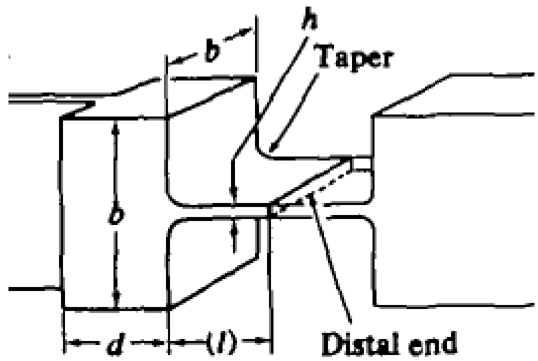

Figure 2: Force plate I-beam diagram [15]

With the dimensions of the strain beam and the walking area of the force plate known, we proceeded to construct a base for the force plate on which to rest and internal support beams on which to walk. We constructed these out of aluminum alloy in order to minimize the weight of the force plate. The bases were placed at two ends of the force plate with the ends of the strain beam resting on them. We did not make them perfectly rectangular in order to give us more room to place the strain gages at a later date. A model of the base constructed in SolidWorks can be seen below as well as the dimensions given in millimeters.

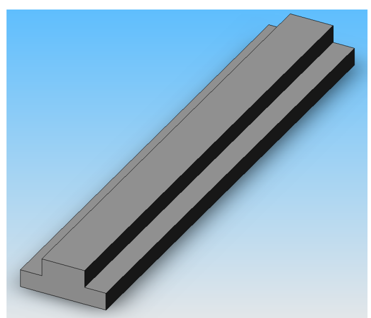

Figure 3: Solid model of force plate base

Something had to be placed internally in order to support the weight of the walkers as they moved across the force plate. Therefore, we used rectangular internal beams running in the direction of the walkers. These had a height of .035 m, a width of .010 m and a length of .480 m. There were nine beams spaced equally under the force plate .005125 m apart. This would ensure that any step on the force plate would land on an internal beam and would be distributed among the four corners of the plate. A picture of the internal model of the force plate is seen below.

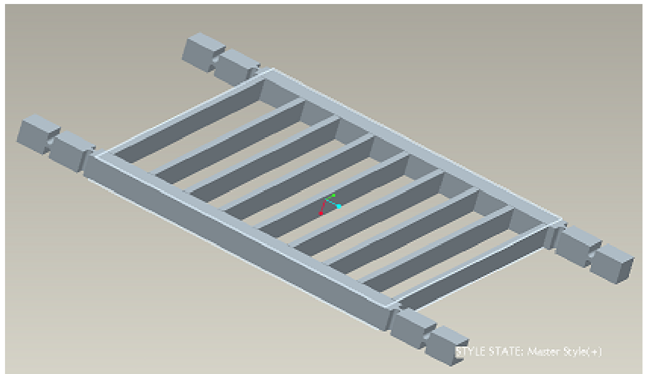

Figure 4: Solid model of internal section of force plate

All of the parts needed for the force plate were designed and taken to the WPI machine shop for manufacturing. Once the parts were collected from the machine shop they were assembled using epoxy rated at strength of 1,600 psi which was sufficient for the weight applied to the force plate. A thin aluminum sheet was also fastened to the top of the force plate for aesthetic purposes. Once constructed, the strain gages were applied to the force plate in the appropriate places as seen below.

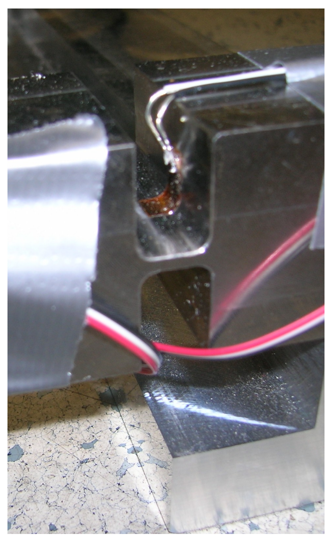

Figure 5: I-beam of force plate

Prior to placing the strain gages on the force plate each corner was cleaned and then the strain gage was applied using superglue. The strain gages used were VISHAY SR-4 with a grid resistance of 120 ohms and a gage factor of 2.1. The gages were placed at the center of the I-beam, running perpendicularly up the wall of the I-beam. They were then soldered together in a Wheatstone full bridge I construction.

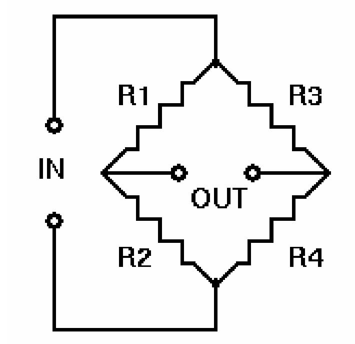

Figure 6: Wheatstone bridge design [18]

The Wheatstone full bridge I, as seen above, was constructed with the resistors, which are our strain gages, in both tension and compression. Resistors one and four are in tension, while resistors two and three are in compression. As pressure applied to the top of the force plate will cause the I-beams to deflect in an S shaped curve. This means that for the top section of a vertical I-beam the outside would be in compression while the inside in tension. The strain gages were placed accordingly for both the horizontal and vertical measurements and had an excitation voltage of 3.33 volts running through them.

B. Force Plate Ramp

The force plate design made the final horizontal plane of the force plate .075 m above the ground. Although this only translates to three inches it will still cause the subject to step up in order to place their foot on the plate. Due to this, we constructed a ramp to elevate the walker to roughly the height of the force plate in order to simulate a normal walking surface.

The force plate ramp we designed was in two sections. The first ramp segment was for the approach to the force plate and the second after stepping off the force plate. The second section was needed so the subject didn't alter their steps in coming down from the force plate. Both were constructed in the same fashion for simplicity.

Instead of making a sloping ramp to ease the approach we decided that by simply starting the subject on the raised platform, it would make it easier for them to walk on the level surface and not have a gradient affect their approach. We wanted the subject to be able to have at least two strides at the elevated level in order to get an accurate measurement on the force plate. The average stride length for men is 2.5 ft and 2.2 feet for women. [19] It can change depending on height and gait of the subject so we made the ramps eight feet to cover a wide range of gait lengths. An eight foot ramp would be too long for easy transport so we looked for a way to make it easier to transport by making it in smaller pieces.

To get the needed three inches of height we used plywood at a thickness of ½ inch and two by threes for support braces (for which the actual dimensions are 1 ½ x 2 ½). We started with an eight foot by four foot sheet of plywood. We then quartered it in order to get four two-foot by four-foot sections. The support braces were cut into four foot lengths. Five were applied to each section roughly five inches apart from the other. Lastly the two pieces were placed next to each other and a hinge was applied to the top of the ramp. This would allow for easy transport and easy setup when the experiment was ready to perform.

C. LabView Design

In order to measure results we constructed two different programs in LabView. LabView is a software package that allows the user to construct a program to measure a given instrument. The two transducers from which we needed to take measurement were accelerometers and strain gages. Screenshots from completed programs can be seen in Appendix A.

The acceleration program was created using a LabView assistant which helped in the making of programs. By first choosing what was needed to measure, the assistant would set up a program and then additional outputs could be chosen. We had the LabView assistant create a program which measured volts because that would be the output of the accelerometers. Later we could calibrate the program and change voltage to accelerations.

Once LabView created the base program, we needed to see the output in a waveform graph and then also change the format of the information from an array of data into one voltage value and time. After converting the data into usable information we constructed a formula box in order to later compute the measured value. We then constructed a second waveform graph in order to see the values after calibration. The data acquired in the program was then sent to an excel folder which could be chosen on the front panel. Data is inserted into excel horizontally through the cells. In order for the data to read down vertically “transpose” had to be chosen on the write to spreadsheet box on the block diagram. This allowed us to analyze the data easier. The program recorded absolute time, time from start of program, voltage, and accelerations in m/s^2.

After constructing a program for the accelerometers we created one for the strain gages. A program for strain gages is slightly more complicated then accelerometers so the data assistant could not be used. This was started with a pre-made program which we slightly changed in order to have the program meet our needs.

As in the accelerometer program we needed to convert the data from an array into a single piece of data. After that we constructed a similar formula box so that the data acquired could be translated into Newtons and then be used in our mathematical model. The data exiting the box was then placed in another waveform chart so the operator could clearly see the forces as they were applied to the plate. Lastly the data was written to a spreadsheet which was transposed so the data was read vertically. The program recorded the absolute time, time from start, strain, and Newtons.

D. Data Acquisition Calibration

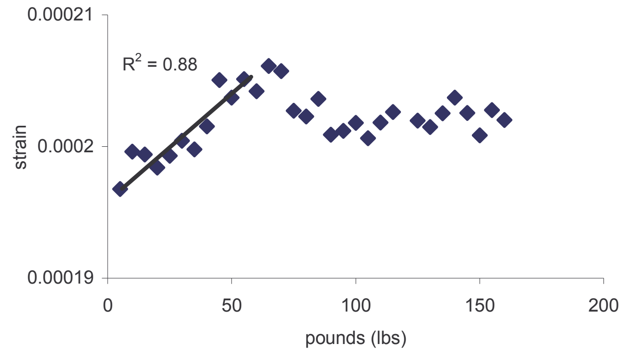

Figure 7: Stress-strain curve during force plate calibration

The force plate needed to be calibrated in both directions in order to interpret the data recorded during each step. The strain gauges on the force plate read in micro strain, and to interpret the force, it needed to be in pounds. We recorded the data for five pound intervals from 0-160 pounds. After plotting the results it was determined that the strain on the force plate was linear up to 70 pounds.

In the horizontal direction we recorded the data for five pound intervals from 0-45 pounds. After plotting the data the correlation between pounds and micro strain was determined. All force plate calibrations can be seen in Appendix B.

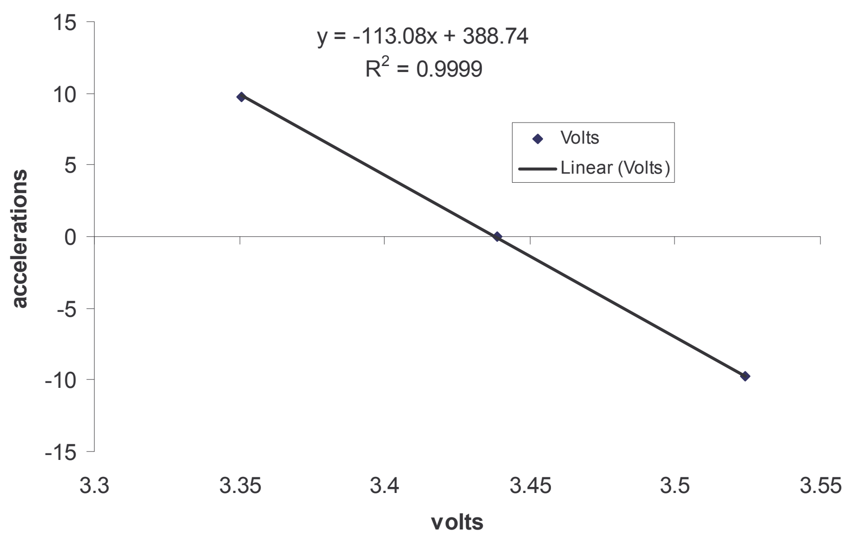

Figure 8: Hip accelerometer calibration in x-direction

The accelerometers needed to be calibrated to determine what each one read when they were in certain positions. They were held against a flat surface in order to ensure the repeatability and eliminate as much noise as possible. Three positions were used to calibrate the accelerometers in the X and Y directions. To calibrate the X direction the accelerometer was held horizontally, vertically on the back end, and vertically on the front end, signifying zero, a negative G, and a positive G respectively. To calibrate the Y direction the accelerometer was held vertically on the back end, vertically with the top to the right, and vertically with the top to the left, signifying zero, a negative G, and a positive G respectively. These positions were used because the output would be known, and constant. After obtaining the average of each of these trials they were plotted, and slope and intercept of the line they created was input into the LabView program. All accelerometer calibrations can be seen in Appendix C.

E. Subject Selection

Test subjects consisted of five randomly selected individuals, three male and two female, and were untrained in race walking disciplines. Five separate trials were performed on each subject for the two different types of walks, the race walk and normal style walking for a total of ten trials. Each trial consisted of one stride, from the toe off of on foot to the heel strike of another. The data was collected and subjects remained anonymous throughout the experimental analysis.

Subjects used in our study were not experienced race walkers. Subjects needed to be taught the basic concepts of race walking in order to complete the experiment using both walking formats. Before starting the experiment both walking formats were explained to each participant. The normal walk was how the subject normally walked on a daily basis. The race walk was unfamiliar to some and had to be explained further.

The three main rules of race walking were told to the participants: subjects must keep one foot on the ground at all times, subjects must land their feet with a straight leg, and participants must keep that leg straight until it passes under the walkers center of mass. The rules were explained and then demonstrated by one of the experimenters. Then the participants were shown footage of race walkers during competition and also demonstrating the gait. This gave the participants two examples of race walking. When subjects felt comfortable with the information, they were given the chance to practice a race walk under our supervision. They then underwent the experiment when they felt comfortable with their own race walk and we felt that they abided by the rules of race walking.

F. Accelerometer Placement

Four MEMS ADXL 276 accelerometers were attached to the subjects by adjustable straps on the ankle, knee, hip, and trunk. The accelerometer is placed on the trunk, or the center of gravity, of the subject in order to determine their vertical displacement while walking. The objective of walking is to move horizontally, any vertical movement that is not converted to horizontal movement is extra expended energy. The specific points on the leg were chosen because they are the joints at which the movement occurs, and the accelerations between each point can be determined by knowing the measurements of the subjects' legs. Based on the assumption that each step in a subjects gait is the same, it is only necessary to measure the accelerations on one leg. Wires are soldered to the accelerometers in order to connect to a LabView program which displays the voltage output from each accelerometer as a graph, and saves the data points in a spreadsheet. The data collected from each accelerometer can then be input into the equations with the acquired force plate data in order to determine the forces on the joints and the overall efficiency of the subjects walking stride.

G. Motion Capture

In order to measure a subjects' potential energy change while walking, the vertical displacement of their center of mass must be known. In order to find this we used video capture and a sacral marker to determine the vertical movement while walking. This data found would then be analyzed and compared to the subjects' kinetic energy.

Before the subject makes a walk the subjects' center of mass must be found. Instead of finding the exact center of mass for each subject we used ergonomic models for the center of mass of the body. The models showed the center of mass slightly above the base of the spine. The center of mass is difficult to precisely view while a subject is walking even when looking frame by frame. In order to assist, a sacral marker is used. This is usually a bright, circular object that can be easily seen by the film. We used a neon pink ping pong ball attached to a cylinder which could be easily taped to one subject then another. The subject could now walk with the camera recording the movement of the marker.

Any video that records digital video can be used in video capture. We used the Sony Mini-DV Network Handycam, (DCR-TRV70) which captures 30 images a second. This rate of capture would give us enough data points to look at the step as it lands on the force plate. The camera was assembled on a tripod and screwed in to ensure that each subject would be walking the same distance from the camera and all angles of all subjects would be the same.

It is extremely difficult to capture the exact moment when one step occurs. In order to do this we captured video of the subjects approach to the step as well as a few seconds after the step. A few extra seconds of video footage can add up to hundreds of extra frames that would need to be analyzed later. To decrease the extra footage, we spliced together the film from each subject in Adobe Premier®. This would allow us to look at all of the normal steps of the subject and the race walking steps of the subject in sequential order with no breaks.

After recording all of the video, the data needed to be analyzed. Most image analyzing software is extremely expensive ranging between 1,000 and 10,000 dollars [20]. Due to budget constraints, we used a free java image processing program called ImageJ [16]. This program could analyze nearly any format of pictures including .gif, .jpeg, and .bmp. It could also be spatially calibrated so that a known measurement on the image could be a dimension calibrated to pixels on the screen. The last thing we needed from the program was the ability to know the time and displacement of the sacral marker. ImageJ allows the user to zoom in on the object it is analyzing and place a marker from frame to frame. These data points of time and displacement could then be taken into excel and analyzed in order to find both the displacement of the subject vertically, (potential energy) and the velocity of the subject horizontally (kinetic energy). These values would allow us to find the efficiency of the different gaits.

H. Data Analysis

Gait efficiency was calculated according to the model developed by Gider et al., which correlates potential and kinetic energy. This model assumes that any potential energy which does not translate into kinetic energy is extra energy consumed, and therefore in fully optimal gait this efficiency index should be equal to 1.

$$PE_k^{(NORM)} = \frac{PE_k}{\sqrt{\sum_{k=1}^{N} PE_k^2}} \tag{7}$$

$$KE_k^{(NORM)} = \frac{KE_k}{\sqrt{\sum_{k=1}^{N} KE_k^2}} \tag{8}$$

The potential (Eq. 7) and kinetic (Eq. 8) energies were normalized for each stride, where k is the number of the sample in a given stride and N is the number of samples in one stride. One stride was defined as toe-off to heel-strike of the same foot.

The cross-correlation between potential and kinetic energy for each stride was calculated using the correlation function in Microsoft Excel 2003®. An unpaired t-test was performed between all the race walking samples and normal samples to determine whether there is a statistical difference in efficiency between normal gait and race walking.

The analysis of joint moments and forces using the method described by Meichtry et al. would have made it possible to determine whether there are any excessive strains on the hip, knee, or ankle during race walking, specifically due to the recent rule changes. This model uses force data, accelerometry, and kinematic analysis to determine joint moments and internal forces on the aforementioned segments.

Table 3: Definitions of joint strain model variables

Variable	Definition	How To Find
M_A	Joint moment at ankle where positive indicates plantar flexion	Eq (15)
M_K	Joint moment at knee where positive indicates knee extension	Eq (16)
M_H	Joint moment at hip where positive indicates hip extension	Eq (17)
R_Z, R_Y	Forces in Z- and Y-directions at foot	Ground Reaction Forces
A_Z, A_Y	Reaction forces at ankle	Eq (9), (10)
K_Z, K_Y	Reaction forces at knee	Eq (11), (12)
H_Z, H_Y	Reaction forces at hip	Eq (13), (14)
Y_R, Z_R	Coordinates of center of pressure on force plate	Y_R=0, Z_R=Estimated by sight
Y_A, Z_A	Coordinates of lateral malleolus	Calculated through Video Capture
Y_S, Z_S	Coordinates of center of mass of shank	Calculated through Video Capture
Y_K, Z_K	Coordinates of knee marker	Calculated through Video Capture
Y_T, Z_T	Coordinates of center of mass of thigh	Calculated through Video Capture
Y_H, Z_H	Coordinates of greater trochanter	Calculated through Video Capture
W_S	Weight of shank	Ergonomic model [17]
W_T	Weight of thigh	Ergonomic model [17]
α_S	Vector acceleration of shank	Vector of accelerometer placed on COM of calf
α_T	Vector acceleration of thigh	Vector of accelerometer placed on COM of thigh
I_S	Moment of inertia of shank	Eq (15)
I_T	Moment of inertia of thigh	Eq (16)
m_s,m_t	Mass of shank and thigh	Ergonomic model [17]
K_s,K_t	Radius of gyration proportional to length of shank and thigh	Ergonomic model [17]
L_s,L_t	Length of shank and thigh	Measured on subject

Table 2 defines the variables used in the method for calculating joint moments in the ankle, knee, and hip.

$$A_Y = R_Y \tag{9}$$

$$A_Z = R_Z \tag{10}$$

$$K_Z = A_Z - W_S - m_S \cdot \ddot{Z}_S \tag{11}$$

$$K_Y = A_Y - m_S \cdot \ddot{Y}_S \tag{12}$$

$$H_Z = K_Z - W_T - m_T \cdot \ddot{Z}_T \tag{13}$$

$$H_Y = K_Y - m_T \cdot \ddot{Y}_T \tag{14}$$

$$I_S = m_S (K_S L_S)^2 \quad (15)$$

$$I_T = m_T (K_T L_T)^2 \quad (16)$$

The assumptions made in this method are described in Eq.'s 9-16.

$$M_A = R_Z (Y_R - Y_A) + R_Y (Z_A - Z_R) \quad (17)$$

$$M_K = A_Z (Y_S - Y_A) - A_Y (Z_S - Z_A) - K_Y (Z_K - Z_S) + K_Z (Y_K - Y_S) - M_A + I_S \alpha_S \quad (18)$$

$$M_H = K_Z (Y_K - Y_T) + K_Y (Z_T - Z_K) + H_Z (Y_T - Y_H) + H_Y (Z_H - Z_T) - M_K + I_T \alpha_T \quad (19)$$

Joint moment calculations can be done according to Eq. 17-19 where Eq. 17 defines ankle moment, Eq. 18 defines knee moment, and Eq. 19 describes hip moment. In this method, the coordinates for optical marker placement would be determined according to the body segment model by Winter.

III. RESULTS

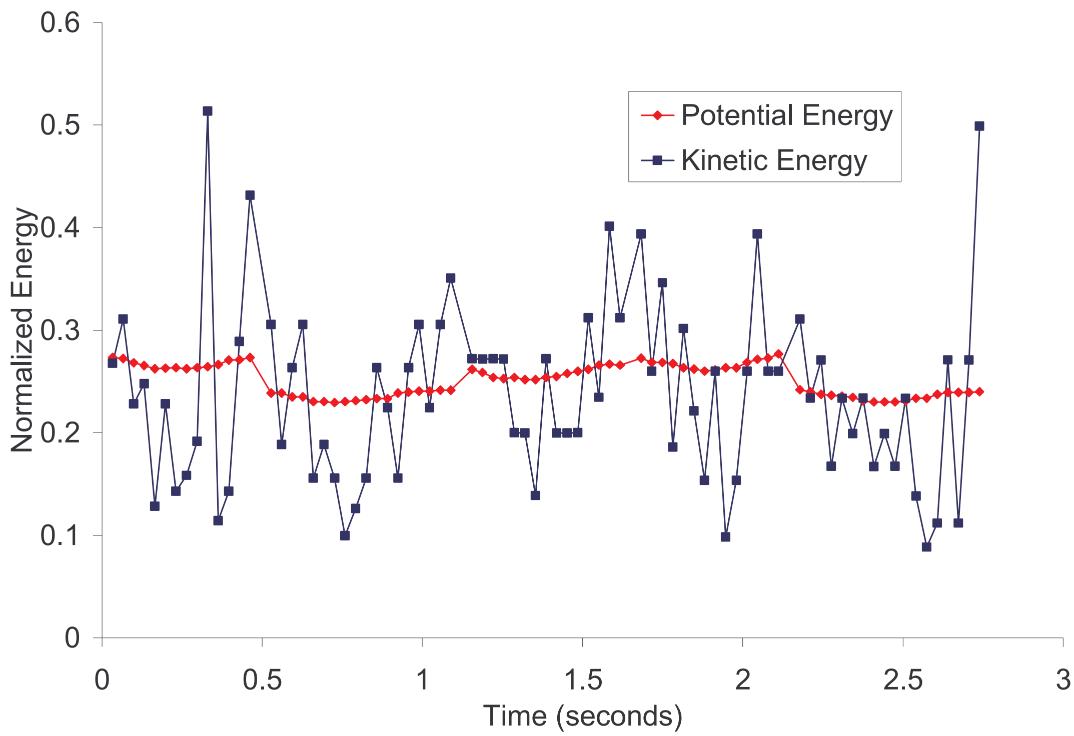

Figure 9: Normalized energies of normal walking

Figure 9 shows the normalized potential and kinetic energy of normal walking of one subject.

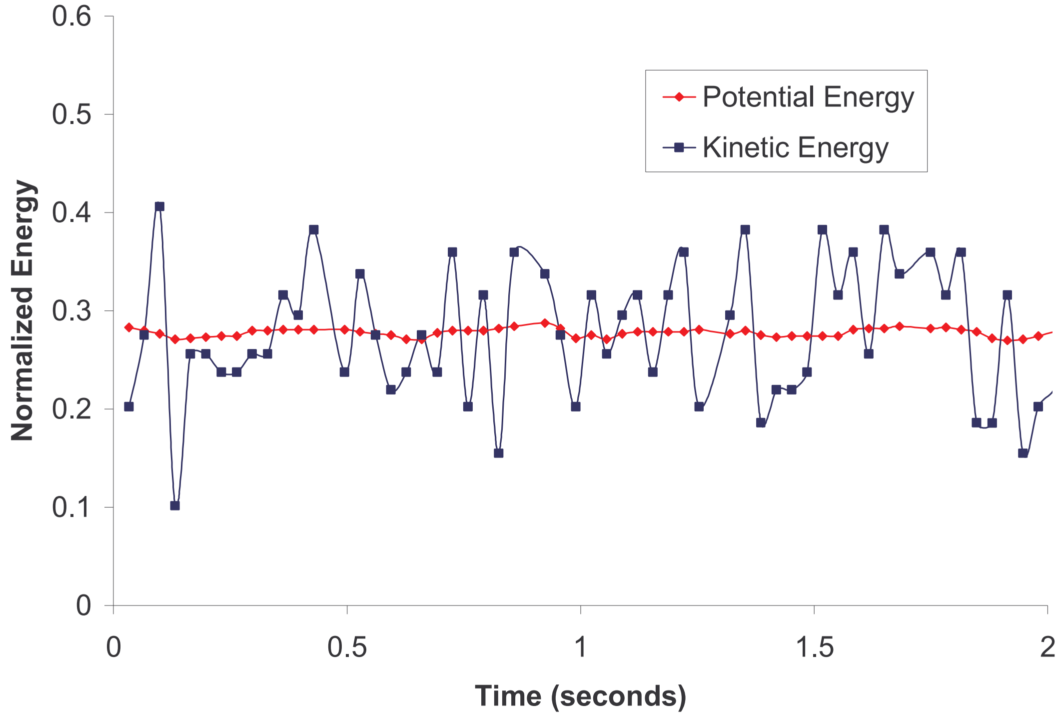

Figure 10: Normalized energies of race walking

Figure 10 shows the normalized potential and kinetic energy of race walking of one subject.

Table 4: Statistical t-test of correlation coefficients

	Normal	Race
Mean	0.541	0.395
Standard Dev.	0.185	0.191
Number of sampled strides	20	
% difference	73%	
α	0.05	
One-tailed t-test	0.009	
Two-tailed t-test	0.019	

Table 1 shows the results of a t-test between the efficiency coefficients of normal walking and race walking of all the sampled strides. A total of 4 subjects were used in this study, with 5 samples recorded for race walking and normal gait each. Using the model of correlating potential and kinetic energy to determine gait efficiency, race walking is 27 ± 3% less efficient than race walking. This is significantly less efficient than normal gait according to both a one-tailed and two-tailed t-test between all the calculated correlation coefficients for normalized potential and kinetic energy. However, two of the subjects did not demonstrate decreased efficiency in race walking, as seen in Appendix D.

IV. Discussion

The calculated correlation coefficients, which indicate gait efficiency according to Gider et al., show that race walking is less efficient than normal walking. This means that the subjects' vertical motions during race walking did not necessarily translate into horizontal motion as much as it did during normal gait. It is expected that race walking would increase energy consumption since it is a more arduous task than normal walking, but in this instance there was some extra energy consumed which did not increase velocity. This is inconsistent with previous studies and the established hypothesis that race walking is in fact more efficient. This variation could be for a number of reasons, such as the subjects are not trained race walkers, the model applied may only be viable for normal gait, and difficulty and human error in accurately locating the sacral marker in each frame during kinematic analysis. Also, while the data was normalized to minimize variations in velocities and time courses, it is unknown whether the efficacy of this method of determining gait is viable for all types of gaits.

There were also issues that were encountered during the design process which had to be dealt with. After receiving the parts from the machine shop and constructing the force plate a problem was seen in the manufacturing process. The I-beams which were originally designed at a thickness of .005 m were now at a thickness of .0025 m. This caused an enormous change in the hypothesized maximum load that the force plate could hold. Using Eq. 6 we changed the formula in order to calculate P_{max}. The new equation is seen here:

$$P\max = \frac{h^3 ns}{6l} \tag{20}$$

Using .0025 m for h, 14, for n 585 MPa for s and .005 for l, we can find the new P_{max} which is roughly 4265 Newtons. When that is placed into equation 1 using our safety factor of 2 and the hypothesized g's of 8 our Ps came out as 267 Newtons which equates to roughly 60 pounds. As seen in the calibration section this holds true to the results. A stress strain curve was created with roughly 70 pounds being the amount of stress that could be applied before the force plate reached a stage of plastic deformation.

Due to the state of our force plate, we could not have subjects walk on it without putting the force plate into plastic deformation. With each application past 70 pounds the force plate would need to be recalibrated to the correct amount. Ways to strengthen the force plate were considered, such as applying a brace, but in the end it was decided that there would be no way to accurately strengthen each I-beam uniformly.

Without the use of the force plate the mathematical model used to find mechanical work could not be used. This changed the layout of the experiment greatly. Without being able to find the ground forces vertically there would be no need to find the horizontal force or the accelerations at the ankle, knee, hip, and center of mass. Collecting the data would be useless without something to analyze it. With no time to manufacture the strain beams again to have the correct thicknesses, we had to move the focus of our experiment elsewhere.

Half of the experiment dealt with the mechanical work throughout the leg. The other half dealt with the efficiency differences between the two gaits. We would still be able to accomplish this using video capture and analyzing it through the use of ImageJ software.

Many recommendations can be made in order to carry out the ideas of the experiment and find clearer results. Ideally we wanted the experiment to go further and delve deeper into the differences between normal walking and race walking. With more experiments hopefully data can be found that will prove whether the rule change requiring straight leg throughout the gait is detrimental to the athletes of race walking.

In a subsequent experiment we recommend that the force plate design stay the same as the one designed previously in the experiment. Through calibration it was proven that the I-beam thickness of .0025 m would hold the predicted weight of 70 pounds. We believe that this will translate to a thicker I-beam being able to support a human while race walking.

Although we recommend the design of the force plate stay the same we believe that it would be beneficial to the experiment to use smaller, more sensitive strain gages. The strain gages were seen to record major differences in ten pound changes. With a more sensitive input we believe that the accuracy of the measurements seen would be increased. We recommend that the strain gages be smaller because the strain gage cannot cross the halfway point of the I-beam. An I-beam is set up so during deflection half is in compression and half is in tension. Once the strain gage crosses halfway it starts to negate strain within the beam. We recommend smaller strain gages so the maximum area of the strain gages experience the strain.

With consequent experimentation we think that more effort needs to go into understanding LabView for data acquisition. LabView has recently upgraded their software to LabView 8.0 and new features are available to the user. We were unfamiliar with the software and believe a better program can be made to acquire the data and decrease noise in the system. Less noise would allow the data to be more predictable while waking with fewer jumps from data point to data point. With smoother curves the data analyzed would have less variance between points.

In addition to changes in LabView, a more powerful data acquisition board was needed to properly do the experiment. A single data board cannot handle it eight channels from the accelerometers, and two channels from the force plate. Multiple computers would have to be time synched to do the experiment or a data acquisition board that could acquire data at several thousand hertz while switching channels would be needed.

We also recommend that a wide array of race walkers be used in the experiment to collect data. We recommend that the race walkers be the variable group and normal walkers as the control. Comparisons can then be made within the walks of the two groups and also between them. This could prove that due to extended walking in the sport, race walkers have a more efficient normal walk then the control. Also, the subjects should walk at different speeds in order to analyze whether walking at various speeds changes gait parameters such as efficiency and joint moments. We would do this to try and lessen the chance of lurking variables in the experiment.

If race walkers cannot be used due to a lack of resources we think that proper training would increase the validity of the experiment. Participants should complete several practices by an experienced race walker in order to ensure that subjects are indeed following the rules and walking in a competition style. The trial during the experiment should also be witnessed by a judge, or experienced race walker to ensure proper technique. Without this it is hard for the experimenters to truly know if subjects are walking correctly.

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APPENDIX A: LABVIEW DRAWINGS

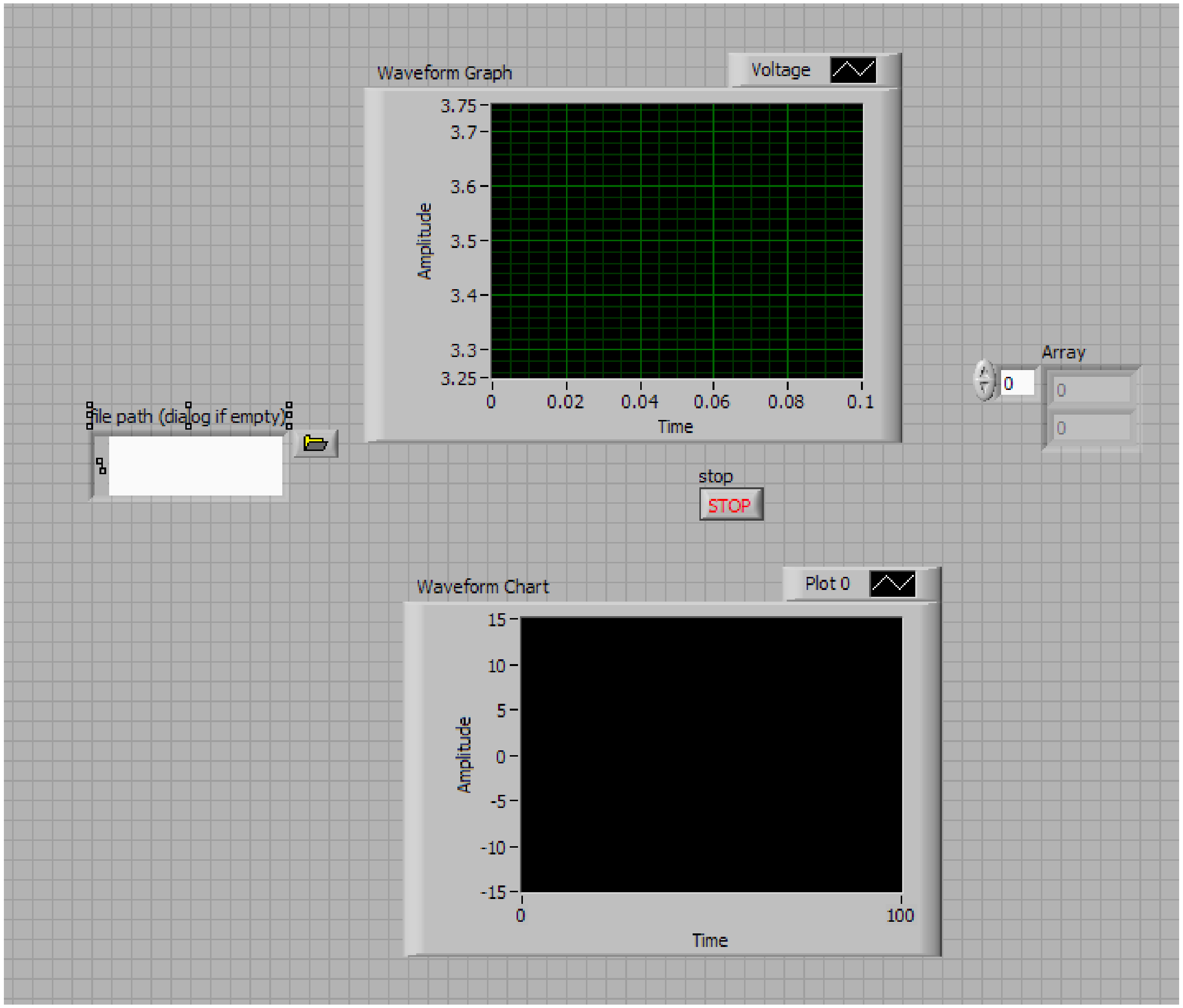

Figure 11: LabView accelerometer front diagram

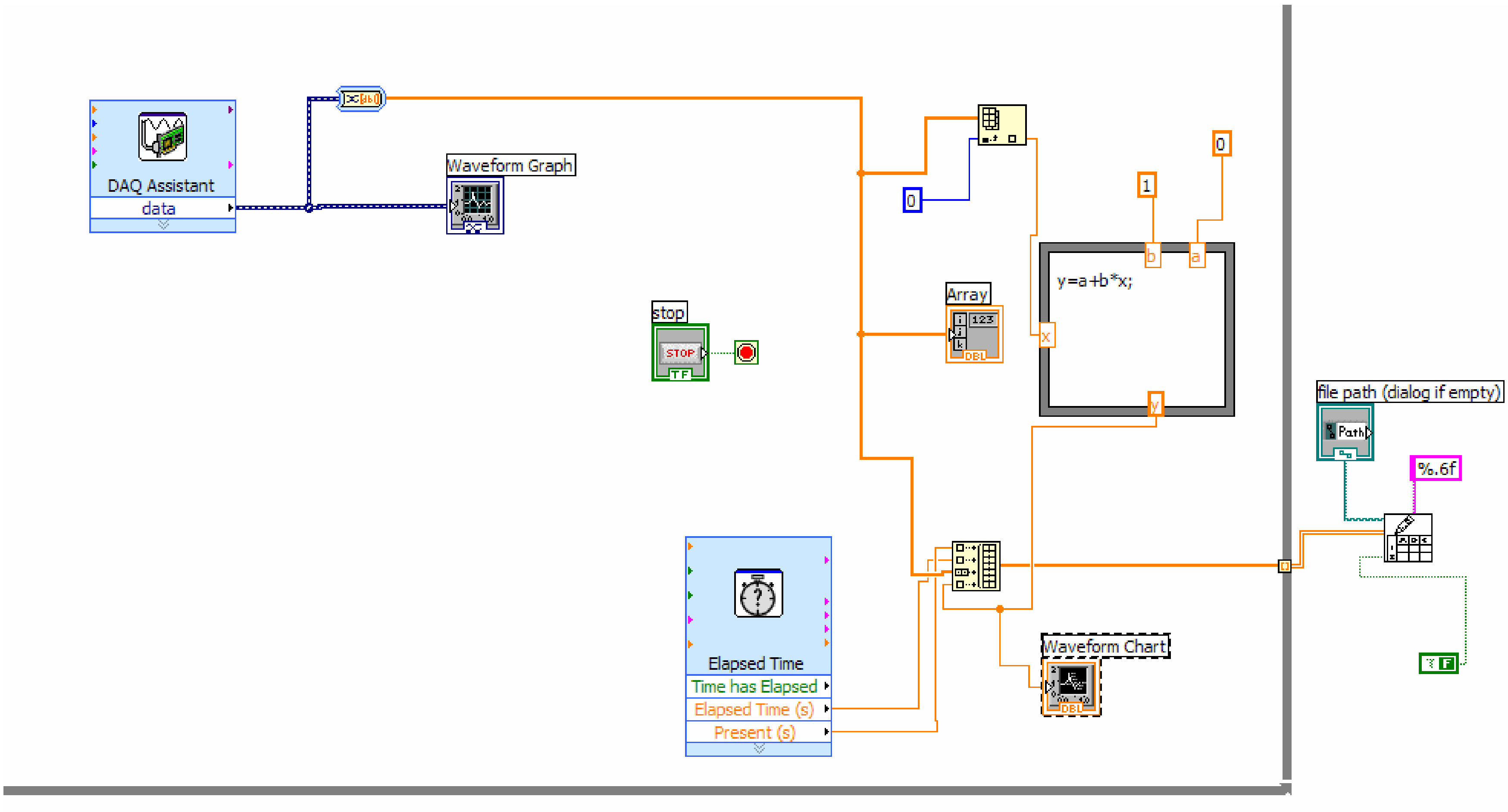

Figure 12: LabView accelerometer block diagram

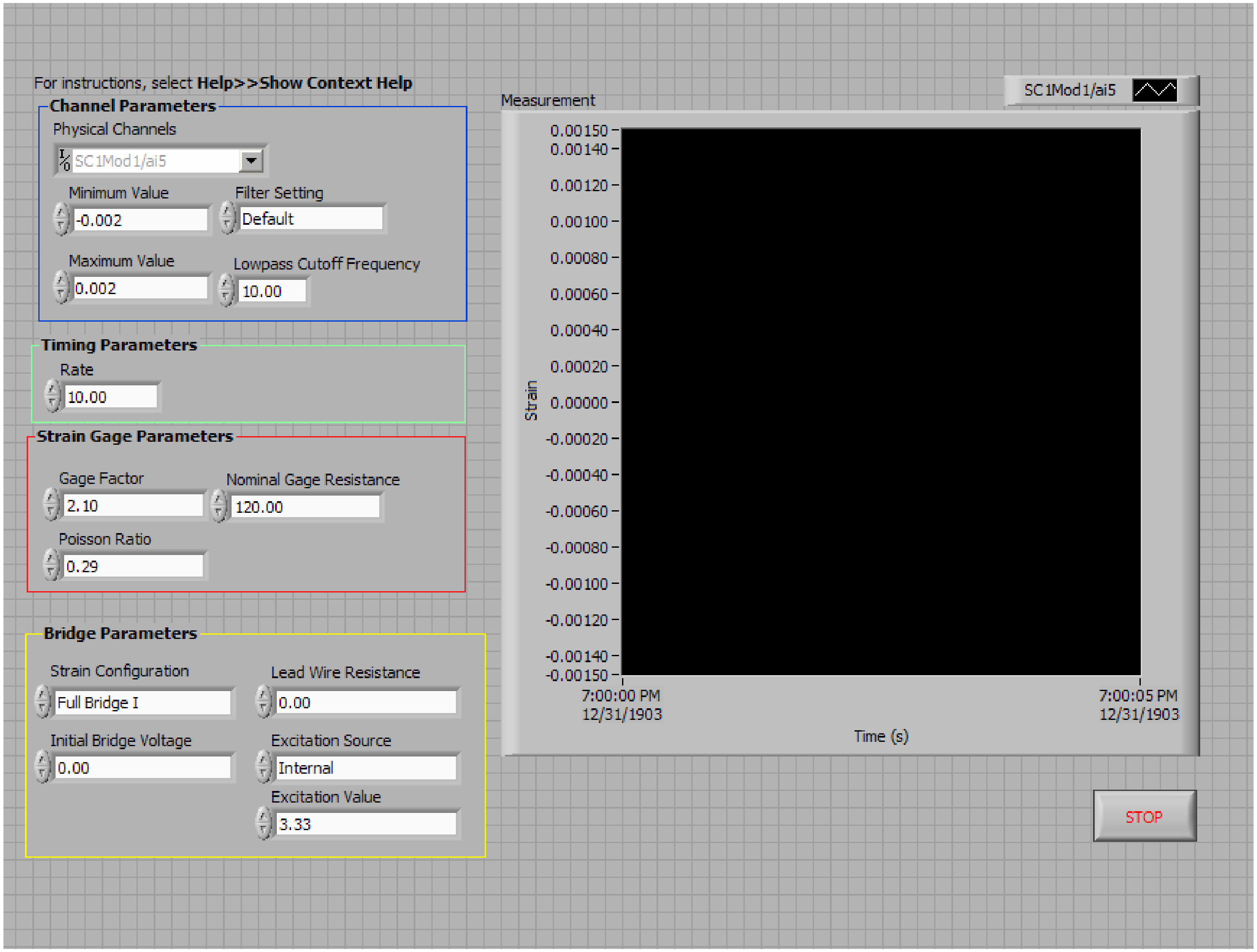

Figure 13: Original LabView strain gauge front diagram

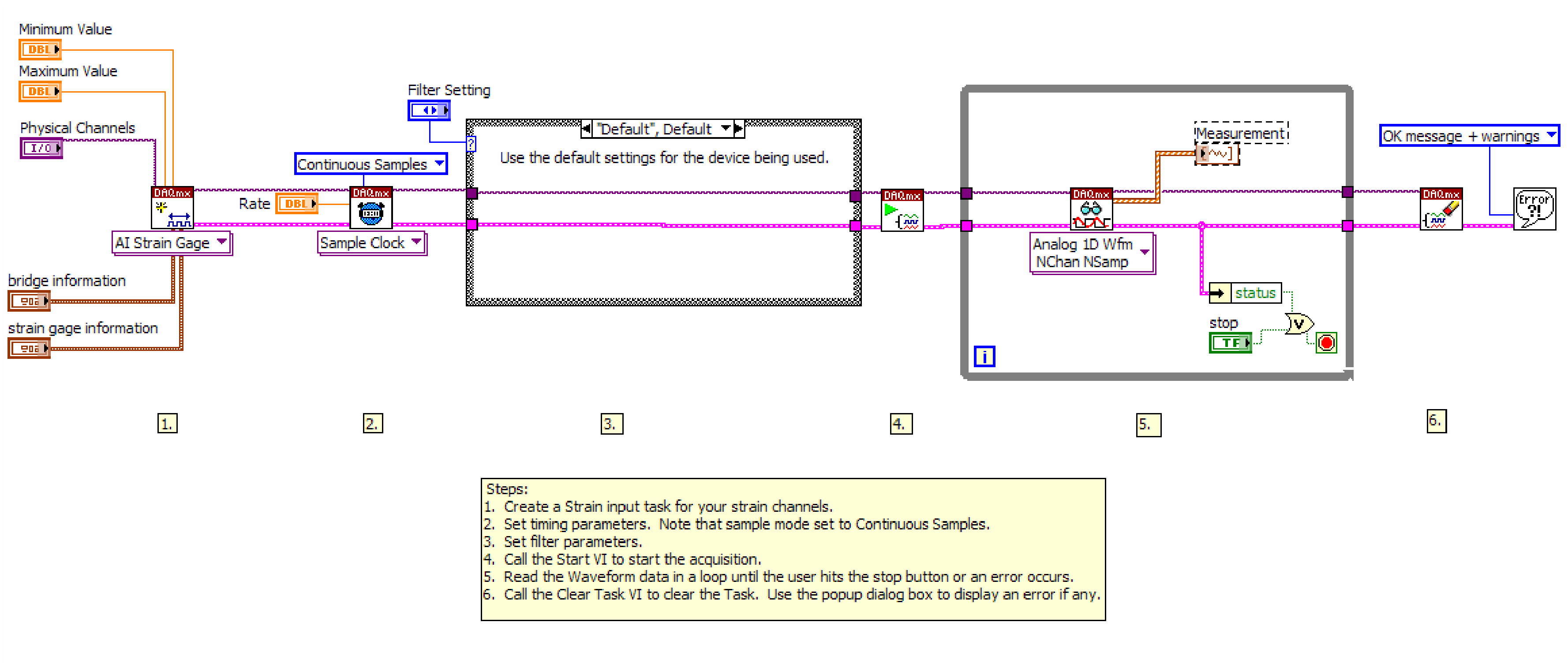

Figure 14: Original LabView strain gauge block diagram

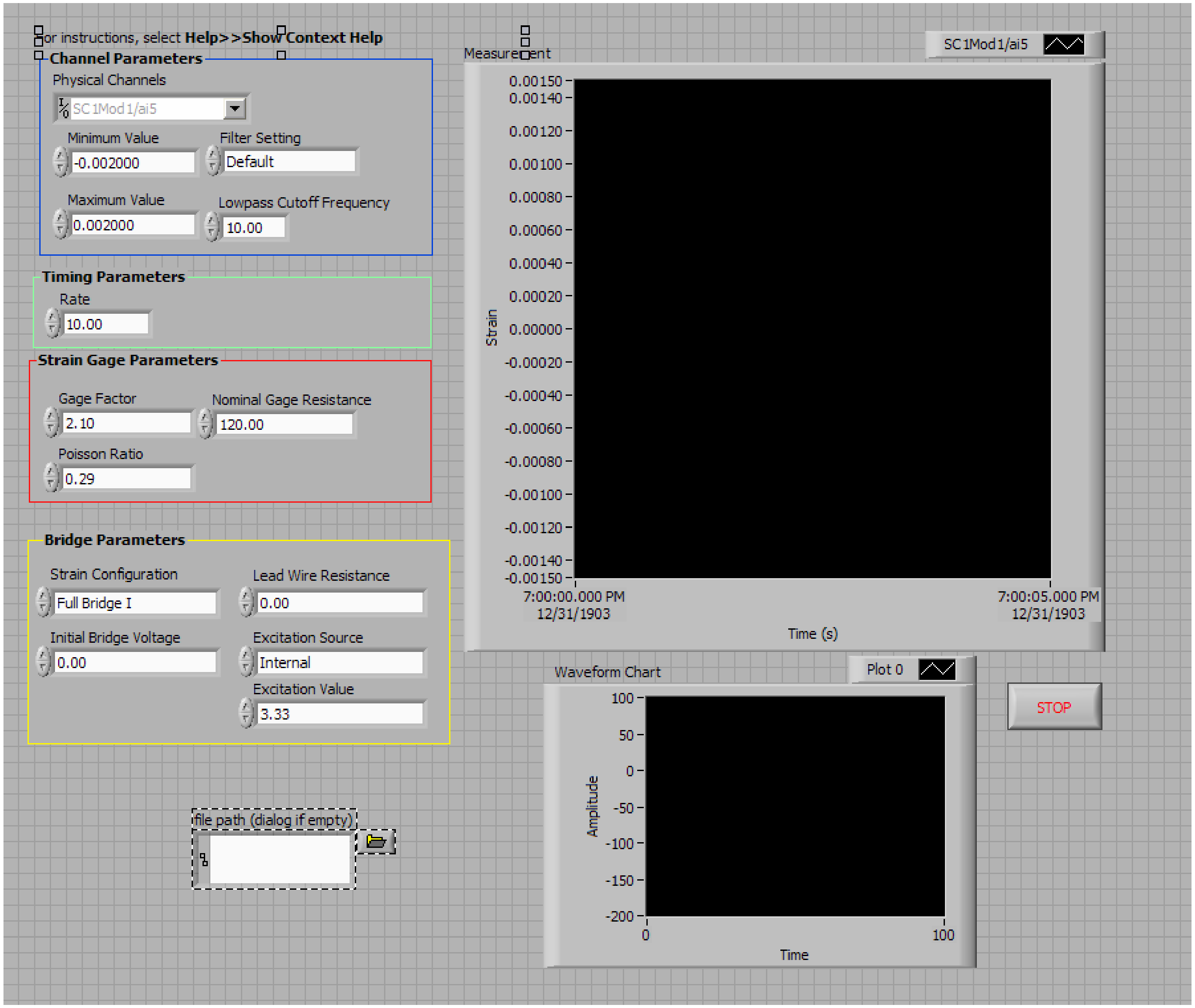

Figure 15: Modified LabView strain gauge front diagram

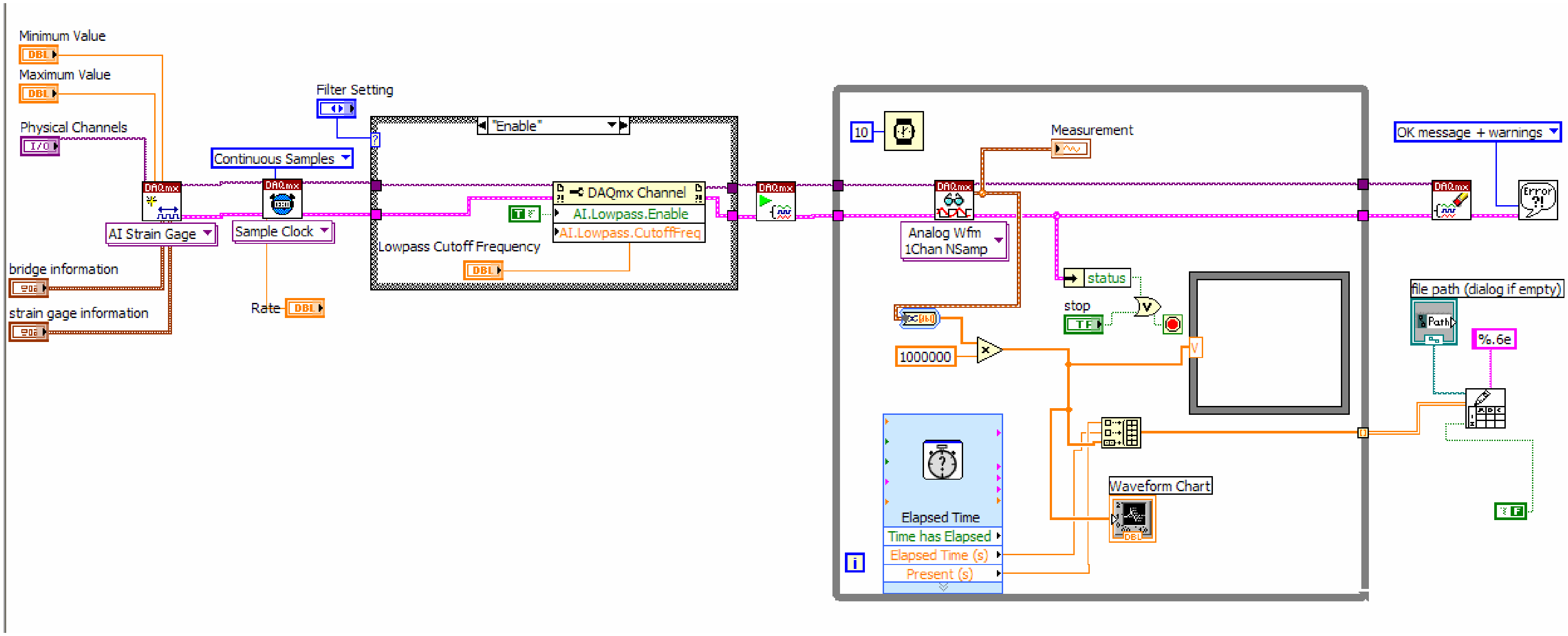

Figure 16: Modified LabView strain gauge block diagram

APPENDIX B: FORCE PLATE CALIBRATION

Table 5: Initial Vertical Calibration Points

Weight(lbs)	Strain
0	0.00019577
5	0.00019676
10	0.000199606
15	0.000199388
20	0.000198394
25	0.000199297
30	0.000200438
35	0.000199774
40	0.000201541
45	0.000205037
50	0.000203703
55	0.000205118
60	0.00020421
65	0.000206111
70	0.000205731
75	0.000202723
80	0.000202285
85	0.000203623
90	0.000200896
95	0.000201183
100	0.000201806
105	0.000200629
110	0.00020182
115	0.000202612
120	0.000202431
125	0.000201971
130	0.000201486
135	0.000202531
140	0.000203724
145	0.00020256
150	0.00020084
155	0.000202756
160	0.000202019

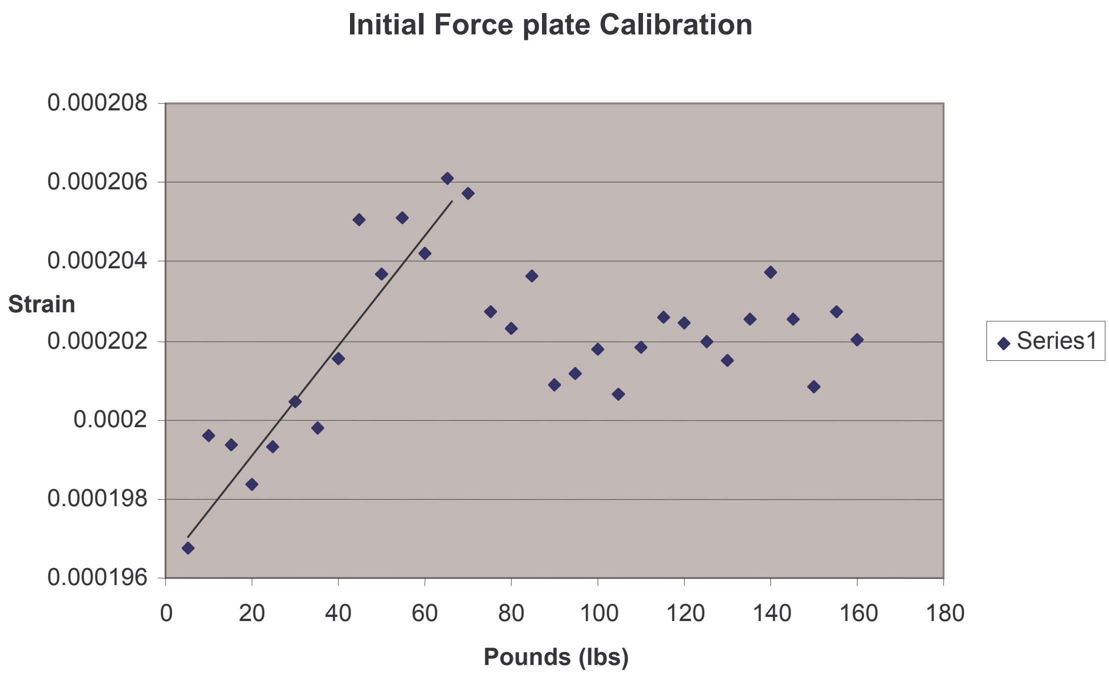

Figure 17: Initial Forceplate Calibration Data

Table 6: Final Vertical Calibration Points

Pounds (lbs)	Strain
0	0.000192624
5	0.000195553
10	0.000195563
15	0.000198953
20	0.000198872
25	0.000201625
30	0.00020238
35	0.000201773
40	0.000204134
45	0.000203767

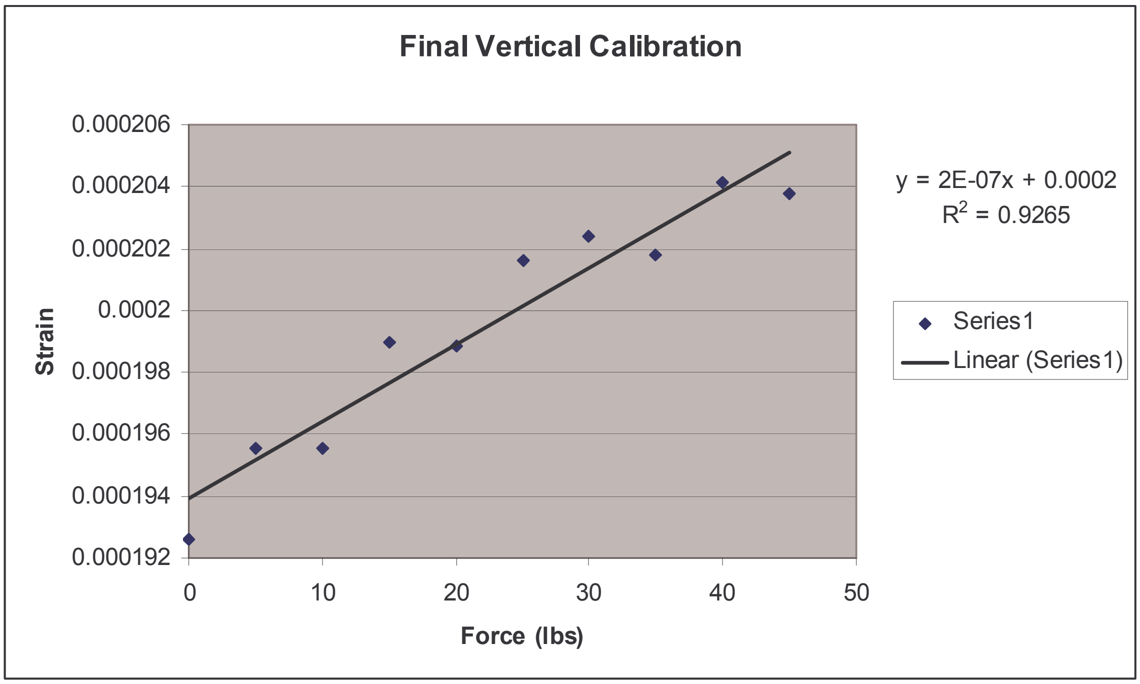

Figure 18: Final Vertical Calibration

Table 7: Horizontal Calibration Points

Weight (lbs)	Strain
0	6.04448E-05
5	6.10483E-05
10	6.11611E-05
15	6.12524E-05
20	6.18678E-05
25	6.33837E-05
30	6.36655E-05
35	6.47034E-05
40	6.48327E-05

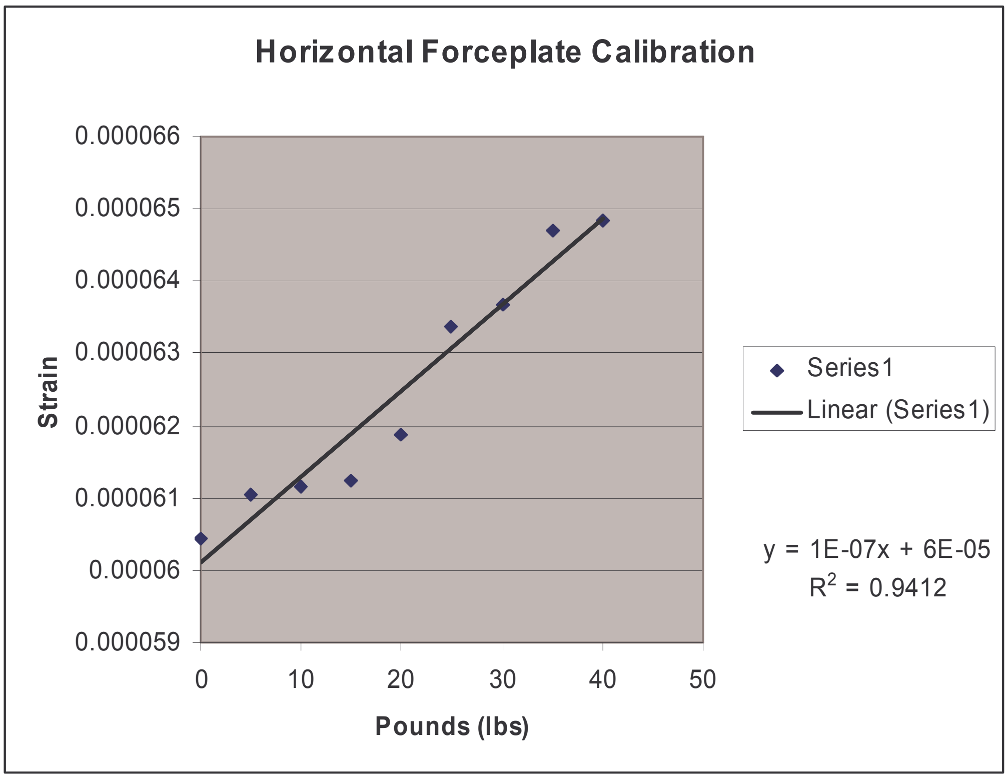

Figure 19: Horizontal Forceplate Calibration

APPENDIX C: ACCELEROMETER CALIBRATIONS

Table 8: Ankle Accelerometer X-direction

x-direction

Volts	Accelerations (m/s^2)
3.355899	9.8
3.446619	0
3.540226	-9.8

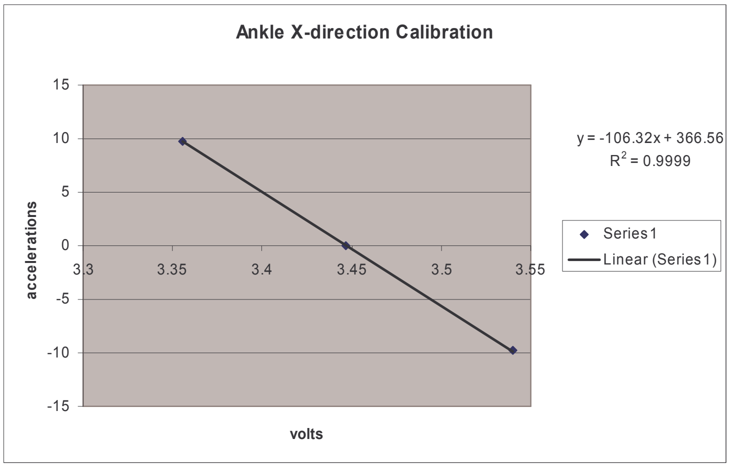

Figure 20: Ankle X-direction Calibration Equation

Table 9: Ankle Accelerometer Y-direction

y-direction

Volts	Accelerations (m/s^2)
3.259258	9.8
3.352192	0
3.436561	-9.8

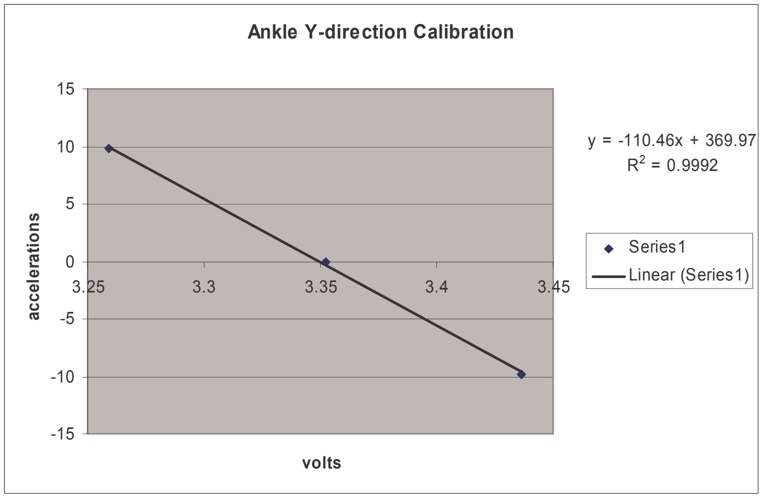

Figure 21: Ankle Y-direction Calibration Equation

Table 10: Knee Accelerometer X-direction

x-direction

Volts	Accelerations (m/s^2)
3.320389	9.8
3.405057	0
3.499045	-9.8

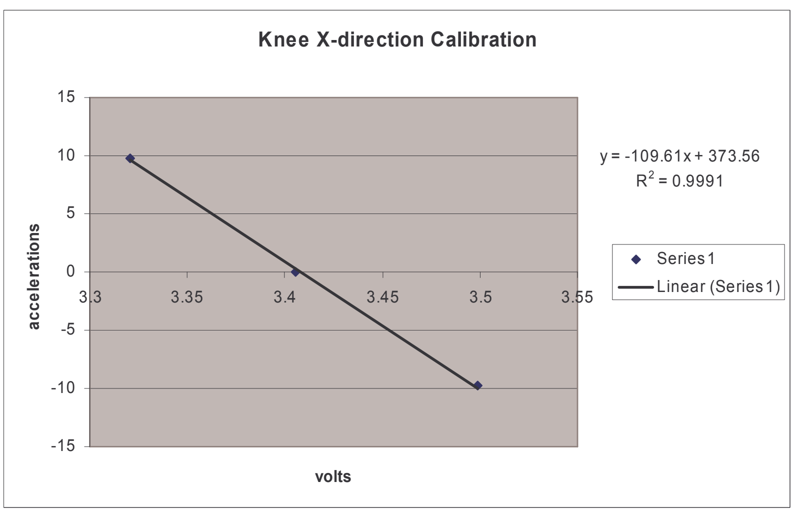

Figure 22: Knee X-direction Calibration Equation

Table 11: Knee Accelerometer Y-direction

y-direction

Volts	Accelerations (m/s^2)
3.301666	9.8
3.387361	0
3.471394	-9.8

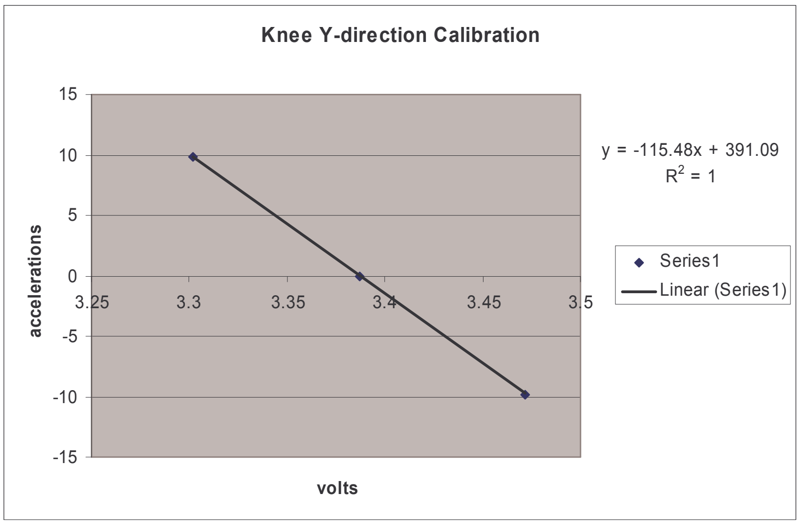

Figure 23: Knee Y-direction Calibration Equation

Table 12: Hip Accelerometer X-direction

Volts (V)	Accel (m/s^2)
3.533799	-9.8
3.444685	0
3.36094	9.8

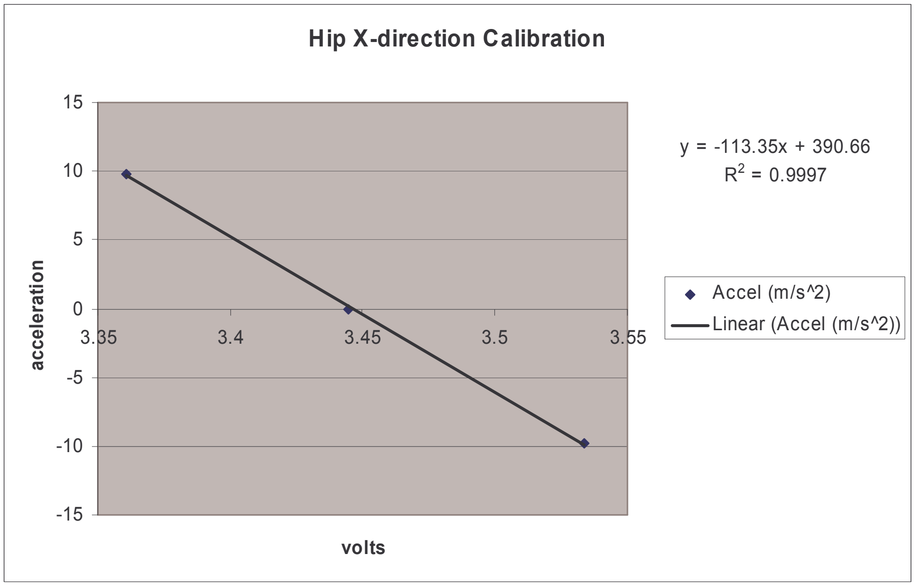

Figure 24: Hip X-direction Calibration Equation

Table 13: Hip Accelerometer Y-direction

Volts (V)	Accel(m/s^2)
3.368092	9.8
3.448771	0
3.534512	-9.8

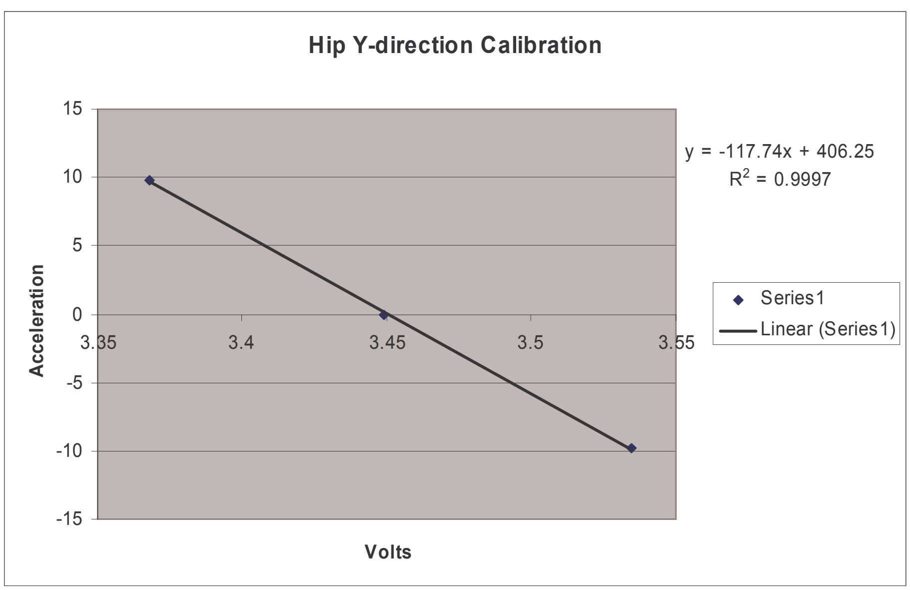

Figure 25: Hip Y-direction Calibration Equation

Table 14: Center of Gravity Accelerometer X-direction

Volts	Accelerations (m/s^2)
3.342433	9.8
3.424906	0
3.516515	-9.8

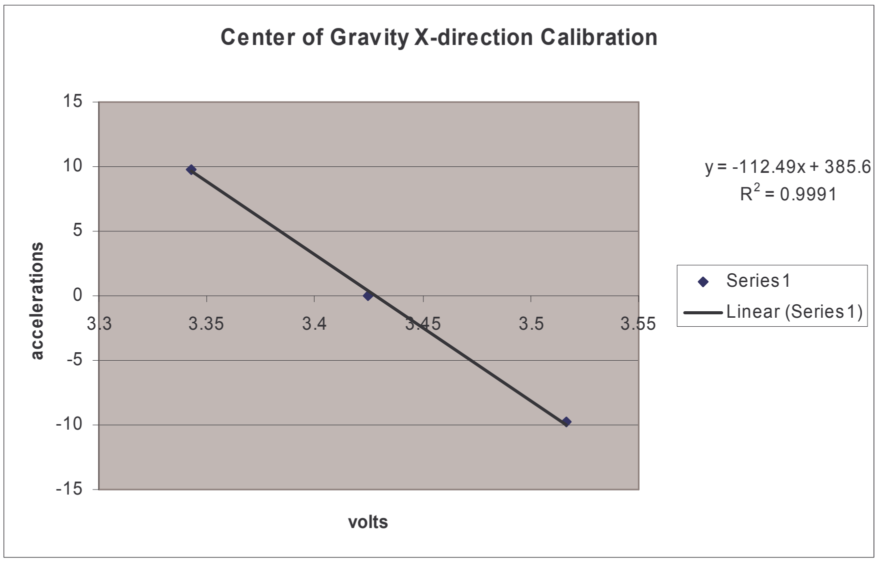

Figure 26: Center of Gravity X-direction Calibration Equation

Table 15: Center of Gravity Accelerometer Y-direction

Volts	Accelerations (m/s^2)
3.317502	9.8
3.407805	0
3.488171	-9.8

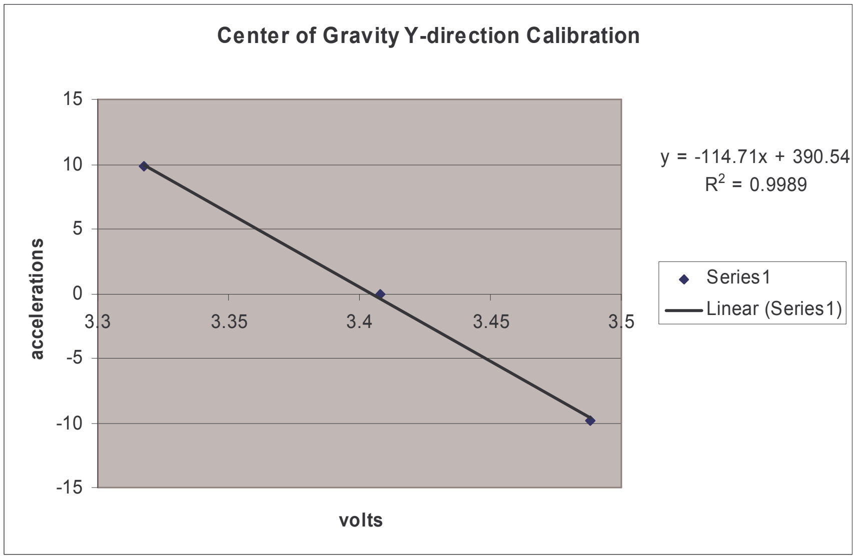

Figure 27: Center of Gravity Y-direction Calibration Equation

APPENDIX D: GAIT EFFICIENCY DATA

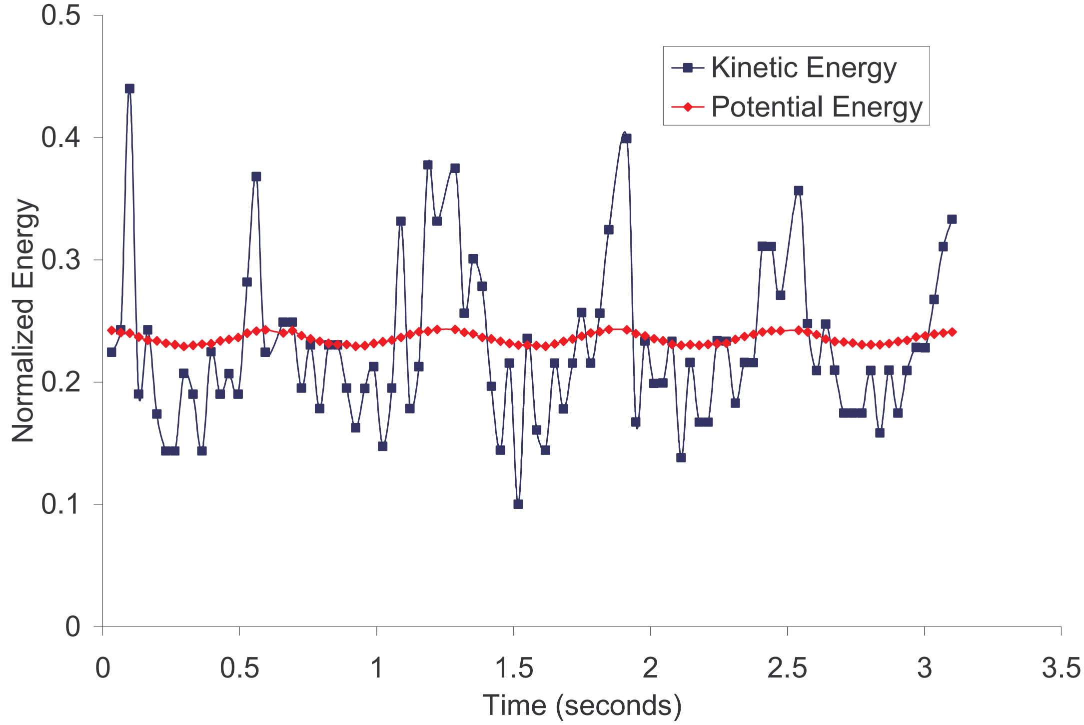

Figure 28: Normalized energy of normal walking in subject 1

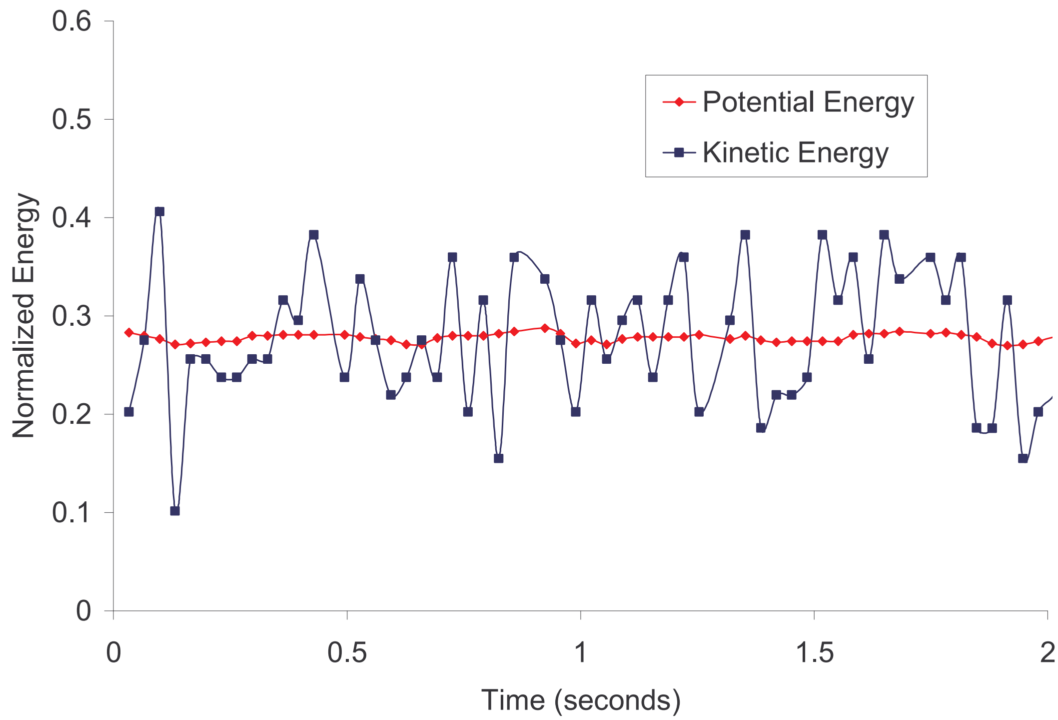

Figure 29: Normalized energy of race walking in subject 1

Table 16: Statistical t-test of correlation coefficients of subject 1

	Normal	Race
Mean	0.702	0.503
Standard Dev.	0.115	0.153
Number of sampled strides	5	
% difference	0.283570314	
α	0.05	
One-tailed t-test	0.026	
Two-tailed t-test	0.053	

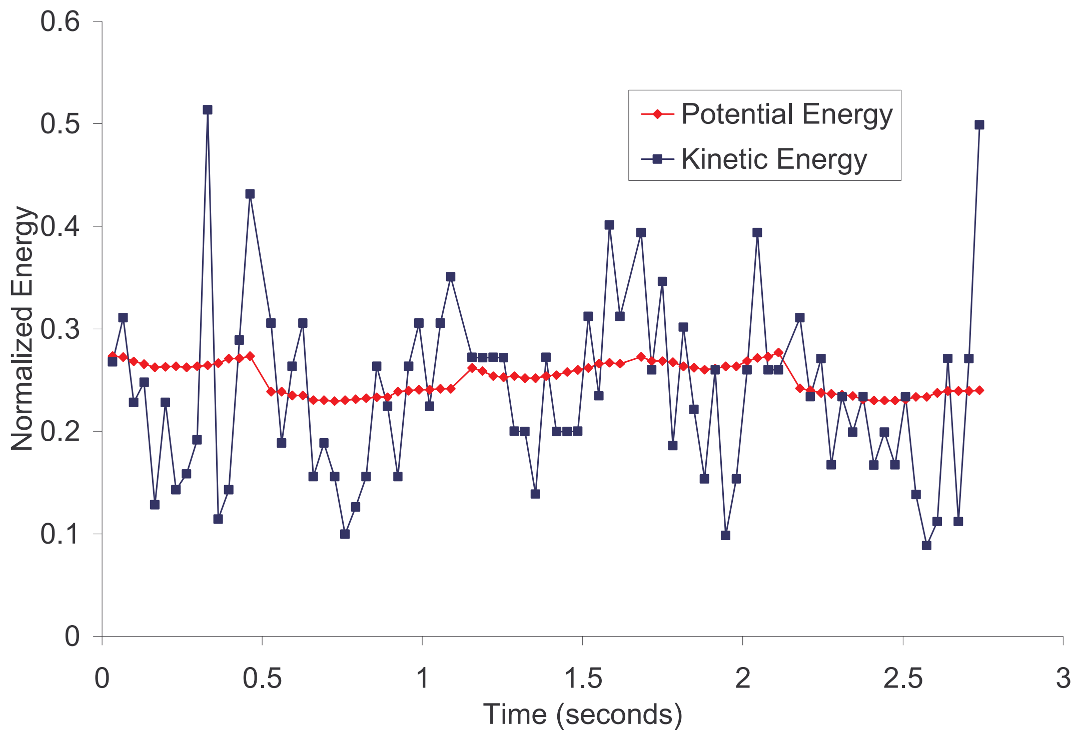

Figure 30: Normalized energies of normal walking of subject 2

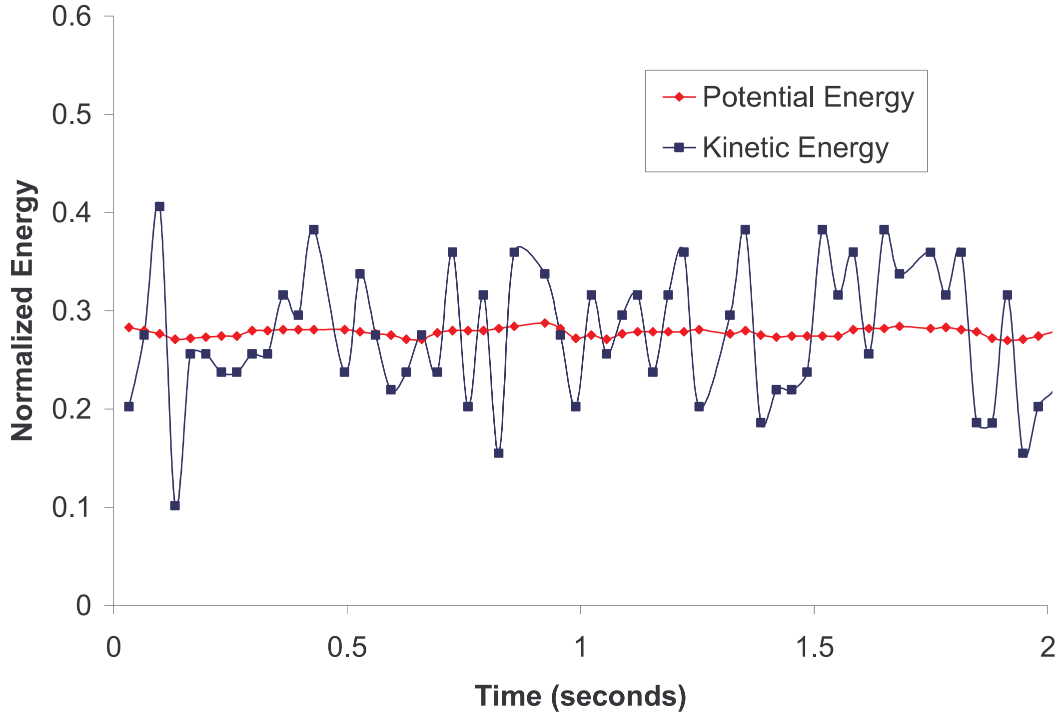

Figure 31: Normalized energies of race walking of subject 2

Table 17: Statistical t-test of correlation coefficients of subject 2

	Normal	Race
Mean	0.536	0.312
Standard Dev.	0.133	0.163
Number of sampled strides	5	
% difference	0.418724562	
α	0.05	
One-tailed t-test	0.022	
Two-tailed t-test	0.045	

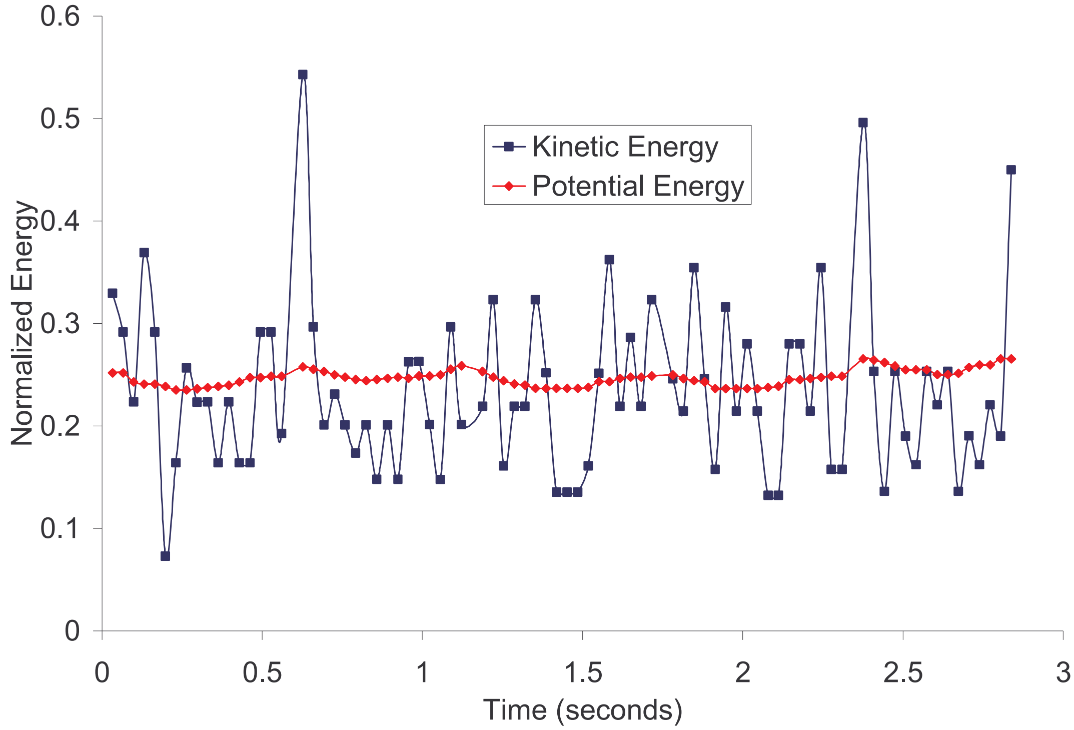

Figure 32: Normalized energy of normal walking in subject 3

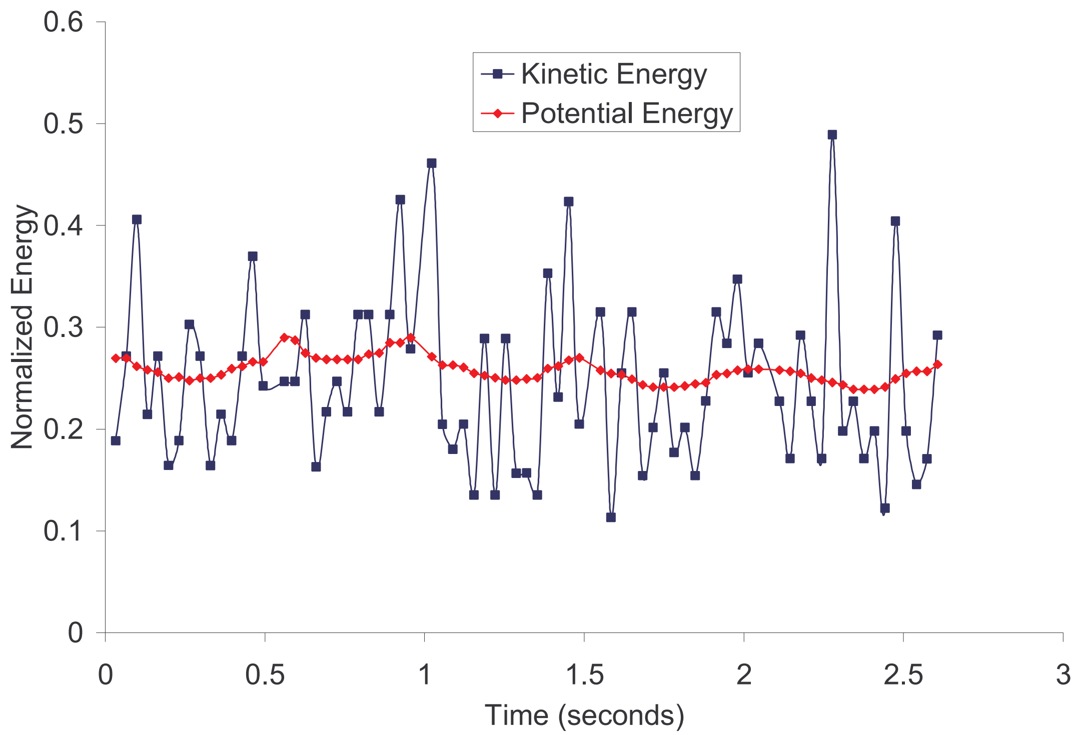

Figure 33: Normalized energy of race walking in subject 3

Table 18: Statistical t-test of correlation coefficients of subject 3

	Normal	Race
Mean	0.395	0.361
Standard Dev.	0.158	0.208
Number of sampled strides	5	
% difference	0.086162645	
α	0.05	
One-tailed t-test	0.390	
Two-tailed t-test	0.779	

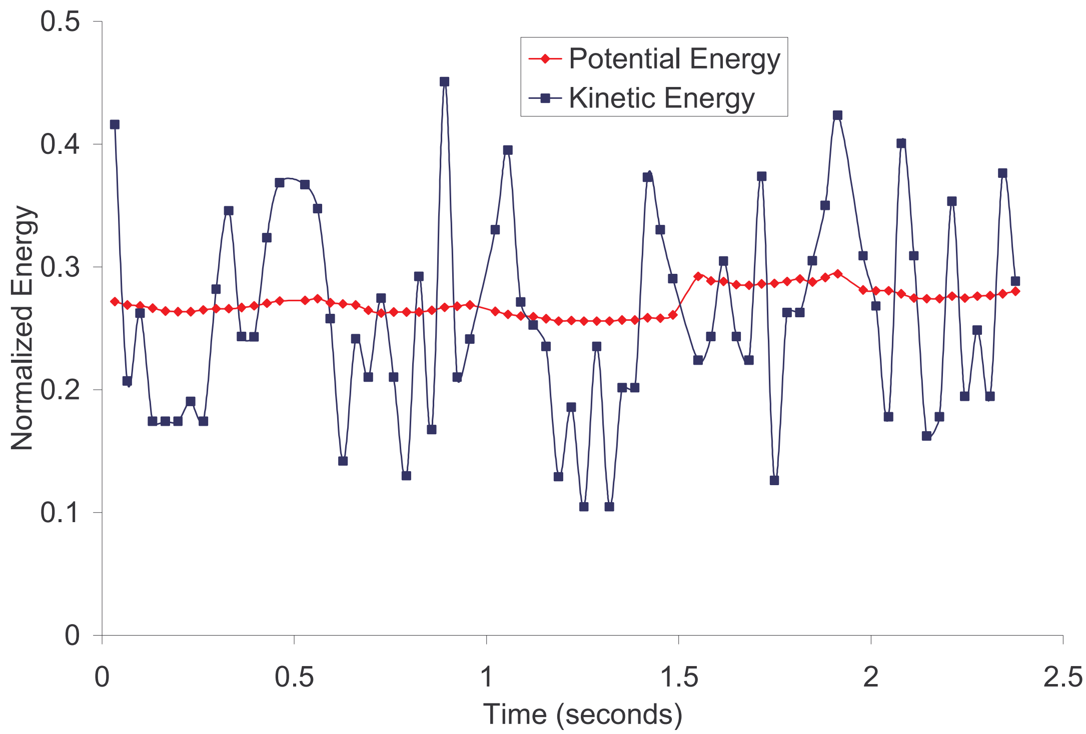

Figure 34: Normalized energy of normal walking in subject 4

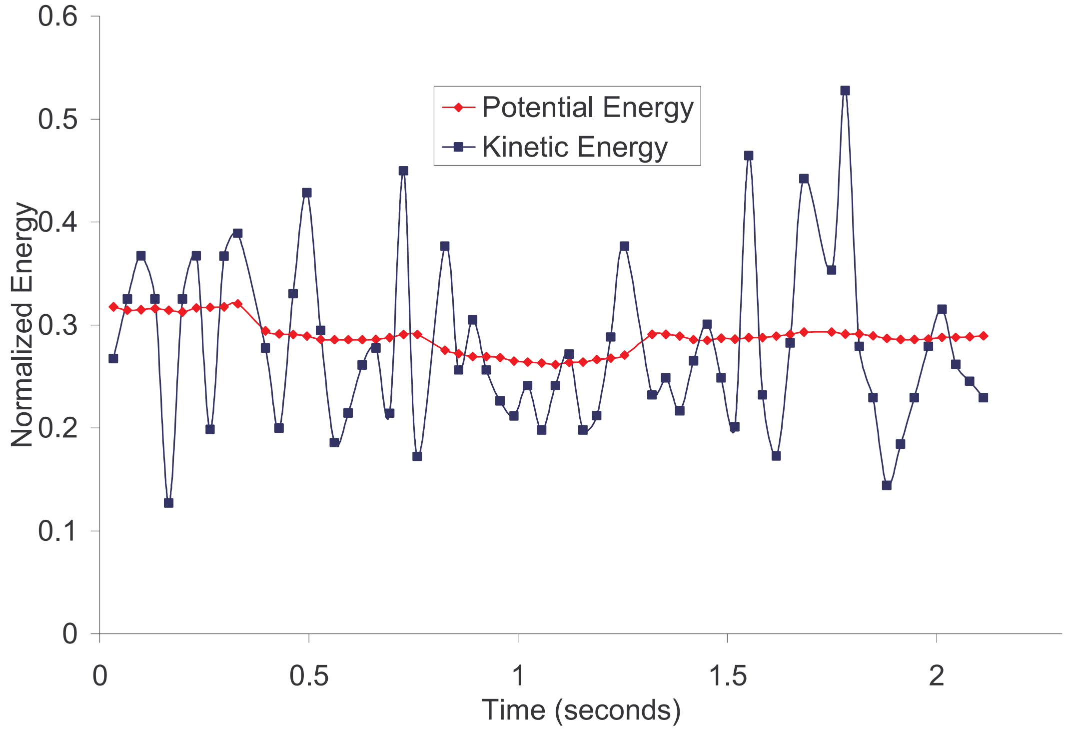

Figure 35: Normalized energy of race walking in subject 4

Table 19: Statistical t-test of correlation coefficients of subject 4

	Normal	Race
Mean	0.533	0.405
Standard Dev.	0.219	0.235
Number of sampled strides	5	
% difference	0.239346061	
α	0.05	
One-tailed t-test	0.201	
Two-tailed t-test	0.402	