GAIT ANALYSIS AND SPINAL ROTATION

A Major Qualifying Project Report:

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New Balance

Abstract

Understanding the gait cycle of runners as well as how the muscles and forces are affected when different running techniques are used is an area of interest for many athletes, coaches, and physical therapists. This study looks to correlate the effects of spinal rotation with the impact forces on the feet and knees while jogging. For a group of 12 runners, two force sensors were placed in the left insole of their shoe to measure the vertical forces upon landing during heel strike and toe off. The forces were correlated to amount of spinal rotation (which was measured using a potentiometer device) during a jog with normal form, with exaggerated spinal rotation, and with restricted spinal rotation. Musculoskeletal models of the leg and foot along with dynamic equations were used to solve for the forces in the appropriate muscles and bones. It was expected to see the initial contact force and the calculated knee loads decrease with greater rotation of the spine. The results were not conclusive as to the reduction of force as a direct result of increased spinal rotation for the entire sample size.

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Executive Summary

  Locomotion is an essential part of people's daily lives. Studying a person's walking and running form is important because it allows for a better understanding of the different roles body segment have in the gait cycle. The ability to diagnose injury can stem from assessing imperfections in a person's form. The gait cycle is an important concept that was developed to describe the cyclic motions that occur when walking or running.

  Gait analysis has been researched for many years. Early studies relied on means of observation as the only way in which information was gathered. With the progression of technology, better methods of collecting quantitative and qualitative data were developed. Types of devices that helped to advance the study of gait analysis include force measurement devices, accelerometers, and video analysis. These devices enhanced the knowledge of proper walking motion and form. The progression in the field of biotechnology has developed a new level of understanding of how the human body functions from a mechanical standpoint. The gait cycle has so many applications in today's society and so there is a large need to further expand the knowledge of the walking and running form.

  The goal of this project was to study the effects spinal rotation has on the magnitude of the impact forces in the foot and to analyze the ankle and knee joints to determine what part of the human body is more prone to injury. The experiment was conducted by attaching foot sensors in the test subjects shoe to record the ground reaction forces. A spinal rotation device was attached to the test subject's back to record the degree of rotation. The data was stored remotely to the subject using a data logger. This allowed the experiment to be done on a track without external wiring, thus freeing the subject and allowing for a natural gait cycle. A video camera and fixed tracking markers were used to record the subject so that body segment angles could be found as well as dynamic forces calculated. The data was collected and analyzed for the use of calculating the forces

in the ankle and knee. Force body diagrams were developed to solve for the unknown muscle and joint force. The expectation was that the forces in the knee will be reduced due to the absorption of force from fluid upper body rotation.

Several preliminary designs were developed for the spinal rotation device and the foot sensors. The foot sensors consisted of two force sensors on the left insole of a shoe. The spinal rotation device was two wooden shafts connected in the middle by a linear taper potentiometer. Both the lower and upper wooden shafts had a thin aluminum bar attached that allowed for accurate spinal rotation measurements to be collected. The aluminum bars had Velcro straps fixed to them which held the device in place while the test subject is in motion. The device was placed between the T1 and T12 vertebrae of the subject to allow for accurate readings of spinal rotation during the gait cycle.

The hypothesis that spinal rotation is directly related to the forces experienced during gait was not proved in this study. Due to a small sample size and variations in gait there were no conclusive or statistical evidence to support our theory.

1   Introduction

Locomotion is an essential part of everyday life and analyzing the gait cycle can shed light on the complexity of how people move. Proper upper body rotation is an important part of avoiding injury and so the dynamics of a person's form can be studied. To generate an efficient gait cycle, joints have to be capable of sufficient movement and muscles have to provide a sufficient amount of force. If joints are stiff and have a limited range, the body has to compensate for the problem, leading to biomechanical abnormalities. Hyper flexibility can also lead to an imbalance by moving body parts in a direction that impedes forward progression.

The goal of this project was to investigate the relationship between upper body rotation and the impact forces distributed in the foot while jogging. In order to study this relationship, two devices were needed: one to measure the degree of spinal rotation and the other to measure the ground reaction forces. Experiments were conducted in which test subjects were equipped with measuring devices that collect and store information while they jog. The test subjects were asked to perform several different movements while they were jogging. They were asked to jog normally, jog with an exaggerated spinal rotation, and jog with a restricted spinal rotation. The data was stored and analyzed to see if there was a correlation of more spinal rotation and smaller forces acting on the feet and knees.

Computational biomechanics was applied to study the translation of impact forces to the ankle and knee joints. This information will be useful, not only to competitive runners and sports teams, but anyone who wishes to exercise properly and avoid injury.

2   Background (Literature Review)

Gait analysis has been researched for many years; however, the study of gait analysis and the correlation to spinal rotation and the forces experienced in the feet, knees, and hips is more recent. A brief history of gait analysis, the correct running form and spinal rotation, the gait cycle, and the different instrumentation and devices that are used to measure gait are discussed in the following sections.

2.1   History of Gait Analysis

The study and analysis of the human gait cycle has been researched and hypothesized for centuries, however, only now with the advent of modern technology can quantitative data be obtained. Original analysis of the gait cycle was based exclusively on observation until the introduction of force measurement devices. These devices allowed biomechanical and computational force analysis.

The first known scholar to study the gait cycle was Aristotle. He developed basic theories based on human and animal movement. Aristotle published his theories in "De Motu Animalium" which contained simple mechanics models of joints, motion, and the gait cycle based entirely on observation (Baker 2007). Numerous scholars since Aristotle have studied the gait cycle each utilizing new technology and techniques that have been developed over time. Some of the most notable scholars are Giovanni Alfonso Borelli, Willhelm and Eduard Weber, and Willhelm Bruane and Otto Fisher.

The analysis of the gait cycle began with Aristotle, but the analysis of the forces produced during the gait cycle was not studied until Giovanni Borelli began to observe tendons and muscles from a biomechanical approach. Borelli became known as the father of biomechanics due to his extensive research in the field (Baker 2007). Borelli's research laid the foundation for Herman

Boerhaave who then applied Newtonian mechanics to the body for the first time. Willhelm and Eduard Weber, who based their research on the anatomy and mechanics of walking, expanded on Borelli and Boerhaave's work and published their own work in 1836. This was the first published study that focused on combining the anatomy and forces that are exerted while walking.

Up until the invention of photography, observation was the only means of studying gait. In 1839, with the improvement of photography, more accurate analysis of the biomechanics and gait cycle was produced. By using photography, exact moments in the gait cycle could be analyzed. One of the first examples of this is Edward Muybridge's collage of "Horse in Motion". High speed photography allowed series of photographs to observe the gait cycle one motion at a time, highlighting previously unseen aspects of the gait cycle. After Muybridge, others began to apply photography to the human gait cycle and allowed more advanced analysis to take place (Muybridge 1878). Another cutting edge photographer at the time was Étienne-Jules Marey, who was known for recording several phases of movement on one photograph. By having multiple phases on one photograph it allowed for easier analysis of motion (Braun 1992).

Willhelm Bruane and Otto Fisher utilized photography as well as their knowledge of biomechanics and Newtonian mechanics to develop the first three dimensional analysis of the human gait cycle (Braune 1987). They assumed that the body could be simplified to a series of rigid members, which then allowed the forces throughout the body to be studied in three dimensions. By using photography and their assumption that the body could be treated as a series of rigid members, they developed and published the first three dimensional analysis of gait, laying the framework for others to follow. The largest drawback for the research that they published was the lack of accurate force with which to test their models.

In 1916, the first way to measure force accurately while walking was designed by Jules Amar. The pneumatic three-component force plate was a milestone for the biomechanical analysis

of gait; it allowed forces to be measured and applied to specific phases of motion when combined with photography. The first published study using a force plate occurred over twenty years later in 1938 by Hebert Elftman. Elftman utilized a mechanical three-component force plate, and provided one of the first analyses of the kinematics of walking using force measurements (Elftman 1938).

Until the 1970s, the majority of the work utilized these techniques. In the 1970s, piezoelectric sensors came into the biomechanical field. Studies, such as Pedotti et al, theorize that using force sensors placed in the insole of the shoe could provide accurate readings of force throughout the gait cycle (Medved 2001). Force transducers, such as the piezoelectric sensors, have become smaller, more reliable and easier to use as time has progressed. This allowed for more accurate force distribution measurements during the human gait cycle.

Current technological advancements including video analysis, force mats, and computer analysis have allowed further advancement in the biomechanical analysis in conjunction with gait analysis. The current methods are finally giving quantitative data for analysis. Systems such as the Tekscan F-Scan and the Contemplas Templo motion analysis program allow the human gait cycle and resulting forces to be measured simultaneously. These technologies combine visual recording using video cameras, which is then integrated with the force measurements in the form of force sensing insoles using computer software. The software has the ability to show maximum force exerted, force distribution on the foot, and even the pronation of the foot during the gait cycle.

2.2   The Gait Cycle

The gait cycle is another term used to describe human locomotion and is divided into two phases, the stance and swing phase. This cycle is a sequence of limb motion that moves the body forward while maintaining stance stability (1992, Perry). The stance phase is the period of time when the foot is in contact with the ground. As the base of this cycle, the foot is subject to the forces

of ground contact with every step, cushioning the body on landing and launching the frame forward immediately thereafter (2010, Pribut). The swing phase is the period of time in which the foot is off the ground and swinging forward. While walking, the stance phase comprises about 60% of the gait cycle and the swing phase about 40% (2010, Pribut). As walking speed is increased, the percentage of time spent in the stance phase decreases.

Since no two people are alike, different people display different forms of the gait cycle. The stance phase is the term used to designate the entire period during which the foot is on the ground. Customarily the beginning of the stance phase has been called heel strike, yet not every person examined will contact the ground in this manner. Similarly, initial contact may occur with the whole foot (flat foot), rather than having forefoot contact occur after the heel-only support. For analyzing the distribution of forces on the foot, the stance phase will be the main focus of study. Since velocity is a key component to the gait cycle, analysis needs to accommodate for its influence on body mechanics as well.

While walking, the stance phase can be subdivided into three component phases according to the sequence of floor contact. Both the start and end of stance involves a period of bilateral foot contact with the floor, known as double stance. Initial double stance begins the gait cycle and it is the time both feet are on the floor after initial contact (1992, Perry). Single limb support begins when one foot is lifted for swing and the body's entire weight is resting on the other extremity. The third subdivision is terminal double stance. It begins with floor contact by the swing foot and continues until the original support limb is lifted for swing. This sub phase is the exchange between the swing and support limbs. To provide consistency, the Rancho Los Amigos gait analysis committee developed a generic terminology for the function phases of the walking gait (1992, Perry). Analysis of a person's walking pattern by generic phases more directly identifies the functional significance of the different motions. There are five sequential phases during stance that

enable the lower limbs to accomplish three basic tasks. These three tasks are weight acceptance, single limb support, and limb advancement (1992, Perry). The swing portion of gait is comprised of three sequential phases and only contributes to the task of limb advancement.
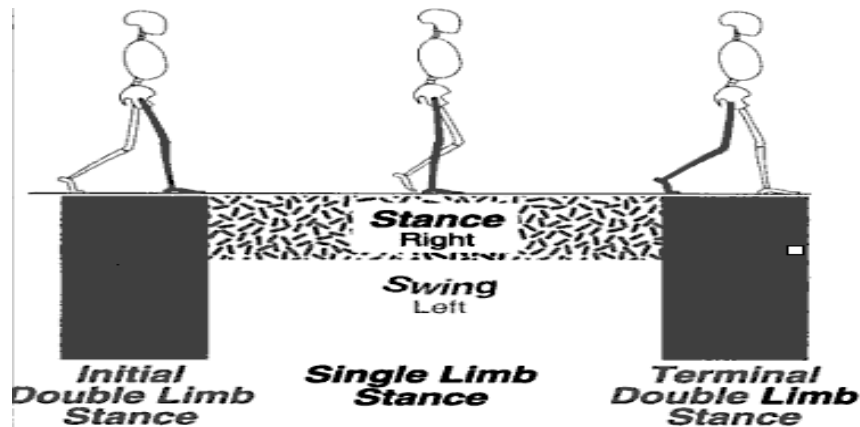
Figure 1: The subdivisions of stance and their relationship to bilateral floor contact pattern. The vertical dark bars are the periods of double limb stance. (1992, Perry)

Weight acceptance is the most demanding task of the gait cycle and has to accommodate for the abrupt transfer of body weight onto the limb that has finished swinging forward. Weight acceptance begins the stance phase and uses the first two sub phases, initial contact and loading response (1992, Perry). Initial contact is the moment at which the foot just touches the floor. The objective of this sub phase is to correctly position the heel (1992, Perry). Phase 2 is the loading response phase. In the case of walking, this is the initial double stance period. This phase begins with initial floor contact and continues until the other foot is lifted for swing. It is during this phase that the shock of the ground force is absorbed and the weight of the body is supported by the heel (1992, Perry).
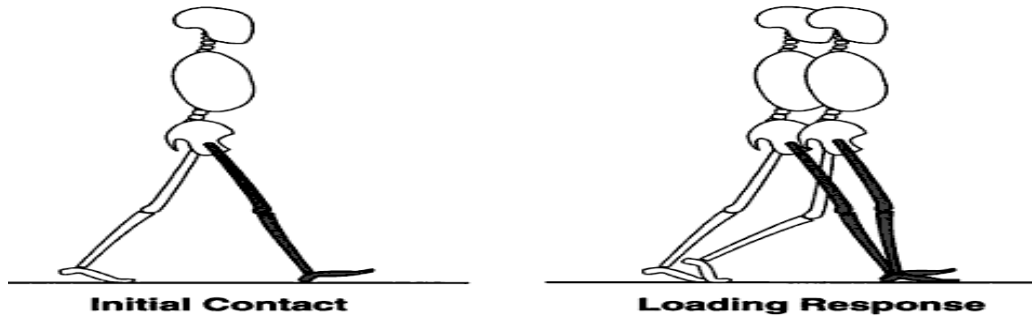
Figure 2: Initial Contact: **The hip is flexed, knee extended, and the ankle is dorsiflexed. Floor contact is made with the heel. (Shading indicates reference limb).** Loading Response**: Body weight is transferred onto the forward limb. The heel is used as a rocker and the knee is flexed for shock absorption (1992, Perry)**

Lifting the other foot for swing begins the single limb support interval and continues until the opposite foot contacts the floor (1992, Perry). During this portion of gait, one limb has the total responsibility for supporting body weight while progression is continued. The two phases involved in single limb support are: mid stance and terminal stance (1992, Perry). Mid stance is the first half of single limb support. It begins as the other foot is lifted and continues until body weight is aligned over the forefoot. Phase 4 is the terminal stance phase and it completes the single limb support (1992, Perry). It begins with heel rise and continues until the other foot strikes the ground. During this phase body weight is transferred ahead of the forefoot.
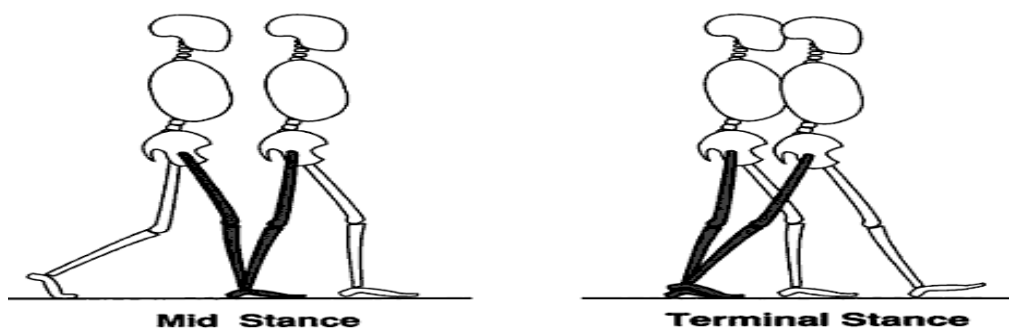
Figure 3: Mid Stance: **The forefoot strikes the floor as the ankle rocks to continue progression. The knee and hip extend.** Terminal Stance: **The heel is raised due to calf muscle action which allows tibia advancement and slight knee flexion. (Perry, 1992)**

The fifth phase, and the final one with foot contact to the ground, is the pre-swing phase. This final phase of stance is the terminal double stance interval in the walking gait cycle and contributes to the task of limb advancement (1992, Perry). It begins with initial contact of the opposite limb and ends with a final pushing force supported by the big toe and the inside third of the forefoot. An abrupt transfer of body weight occurs during this phase and promptly unloads the limb to position for swing.
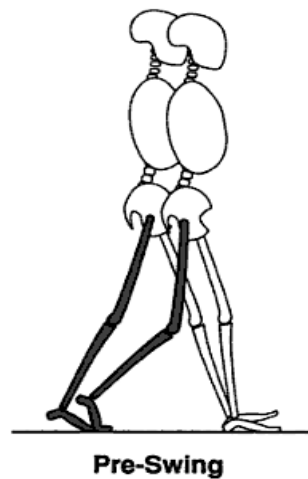
Pre-Swing

Figure 4: Pre Swing: Ankle plantar flexion and knee flexion is increased along with the loss of hip extension. (1992, Perry)

The final three phases are concerned with the foot off the ground and thus make up the swinging period (1992, Perry). Phase 6 is the initial swing which makes up approximately one-third of the swing period. Acceleration begins as soon as the foot leaves the ground when the person activates the hip flexor muscles to move the leg forward (1992, Vaughn). Phase 7 is the mid swing or the second phase of the swing period (1992, Perry). This phase begins with the initial swinging motion and ends when the limb is in a forward position where the tibia is vertical. Phase 8 is the final phase of the swing period as well as the walking gait cycle (1992, Perry). This phase begins with the tibia in a vertical position and ends when the foot strikes the floor. Deceleration

occurs when the leg slows down to stabilize the foot in preparation for the next heel strike. (1992, Vaughn)
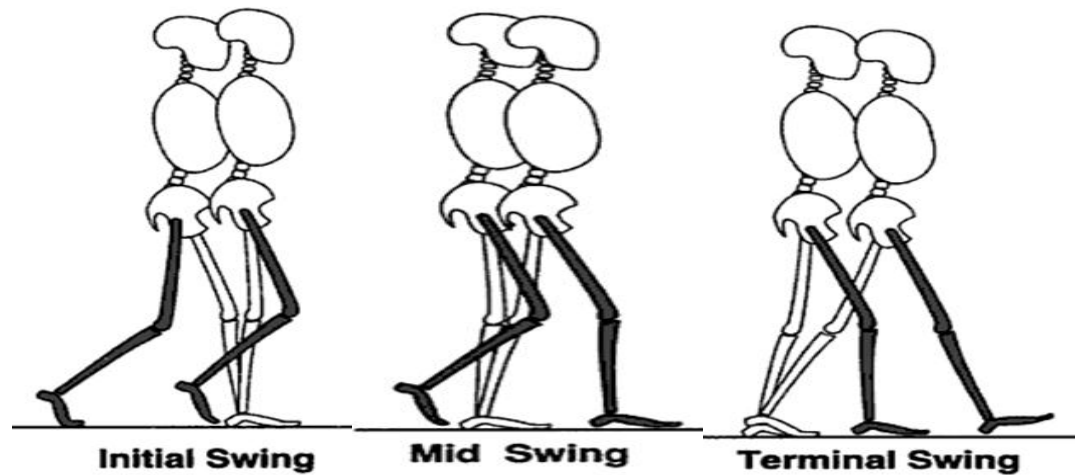
Figure 5: Initial Swing: **The foot is cleared of the floor and the limb is advanced from its trailing position by hip and further knee flexion**. Mid Swing: **Hip flexion progresses, knee extension, and dorsiflexion of the ankle begin**. Terminal Swing: **Limb advancement is completed by knee extension. The hip maintains flexion along with the ankle maintaining dorsiflexion.**

The precise duration of these gait cycle intervals varies with the person's walking velocity. The total normal distribution of the floor contact comprises 60% of the total gait cycle. The timing for walking phases of stance is comprised of 10% for each double stance interval and 40% for single limb support. Double stance occurs when there is bilateral foot contact, meaning both feet are in contact with the ground providing balance and stability. Walking faster proportionally lengthens single stance which in turn shortens the two double stance intervals.   When bilateral foot contact with the floor is omitted, the person has entered the running mode of locomotion.

During the running gait cycle, single limb support is the only form of the stance phase but it is comprised of three sub-components: initial contact, midstance, and propulsion (199, Christensen). The ball of the foot makes initial contact with the ground with most of the weight on the outer edge. A gradual shifting of weight to the inner edge follows as the foot moves down and

inward to the position of pronation (2010, Pribut). The arch then flattens to distribute the force of the heel strike. Next is midstance, which is the period when weight shifts from the posterior to the forefoot (199, Christensen). At this point in time body weight is shifted directly over the foot. Pronation ends as the foot begins to roll forward and upward. The final component is propulsion, where the foot effectively becomes a lever with the Achilles tendon providing a pulling force and the ball of the foot serving as a fulcrum (1999, Christensen). The joints in the big toe and forefoot create enough force to launch the foot off the ground and into swing phase.

2.3   Correct Running Form and Spinal Rotation

Correct upper body rotation during a person's natural running motion is a controversial topic. A person's running style tends to be roughly defined by their innate and individual biomechanics. However, it is believed that spinal motion plays a significant role in maintaining upright posture and balance as well as reducing shock transmission during gait. The position of the torso while running is affected by the position of the head and shoulders. With the head up, looking forward, and the shoulders low and loose, the torso and back naturally straighten to allow an efficient, upright position that promotes optimal lung capacity and stride length (1987, Hinrichs).

Even though running is primarily focused on lower-body activity, the hands and arms play a significant role in proper body mechanics. The hands control the tension in the upper body, while the arm swing works in conjunction with the leg stride to drive the body forward. Forward movement of one arm and the backward movement of the other create a force that rotates the upper body around a vertical axis (1987, Hinrichs). The legs move in a manner that creates a force on the lower body around a vertical axis in the opposite direction of the upper body.

Runners who have poor upper body mechanics swing their arms in a manner that does not create as much angular momentum as the lower body produces. This results in the body over rotating which alters the stride length. Thus, the runner must expend energy, not only to propel forward but to compensate for the imbalance (1987, Hinrichs). An objective during the running gait cycle is to produce enough angular momentum in the upper body to cancel out the angular momentum of the lower body. The result of this is that the total body angular momentum is zero (1987, Hinrichs).

Figure 6: The image depicts the angular momentum of the upper body being driven by the runner's right arm, where the lower body angular momentum is generated from the left leg. Both of these moments have an axis of rotation about the pelvis. (1987, Hinrichs)

The role of the spine in running is ill-defined, and the consequence of restricted spinal motion while running has yet to be explored. A study done in the Chicago Motion Analysis Research Laboratory assessed spinal motion and its contribution to the gait cycle. Ten able-bodied persons, five females and five males, were each fitted with a customized fiberglass body jacket to restrict spinal motion (2006, Konz). The protocol consisted of two successive gait analyses. The study examined each subject's conventional, unrestricted gait. The second analysis examined each

subject's gait with the spine restricted by the body jacket. The subjects were asked to walk a minimum of 5 trials with speeds ranging from 1 to 3 meters per second (2006, Konz).

Retroreflective markers, which reflect light while minimizing the light scatter, and motional analysis were used to monitor the subject's body mechanics. Ground reaction force data was retrieved using force platforms embedded in the walkway. The results of the study demonstrated that when spinal motion was restricted, pelvic obliquity and rotation were reduced across all walking speeds (2006, Konz). The reductions in pelvic motion significantly affected lower-limb kinematics resulting in shorter stride lengths as well as slight increases in ground reaction forces (2006, Konz). However the study has a few limitations, one being the small sample size making generalizations difficult. Another limitation was testing the subjects only during their walking gait. The influence of velocity on the gait cycle and the difference in reaction forces between walking and jogging could have been considered in the analysis.

2.4   New Running and Walking Styles

In recent years, new running and walking styles that focus on the use of spinal rotation have gained attention. A founder of one of these types of running forms claims that with more spinal rotation, there are fewer injuries and more of a physically stronger feeling than their previous methods of running.

2.4.1 Chi Running

Chi Running is a method of running that was recently developed by Danny Dreyer over the past 35 years. (ChiLivingInc). The principles of Chi Running are based from the principles taught in yoga and Tai Chi. The main objective of this style of running is to maintain balance and return the

body to its centerline. There are several essential principles that Chi Running follows. These principles include performing exercises before running that loosen the body, aligning the body in a vertical column when running, leaning forward when running so as to be pulled by gravity, running with a midfoot strike, opening the stride out the back, minimizing the swing of the arms, and focusing on the centerline of the body.

Dryer wrote a book called *Chi Running: A revolutionary Approach to Effortless, Injury-Free Running* (Dreyer). In the book, he explains that your shoulders, spine and hips are held in position by the ligaments and tendons surrounding them. According to Dryer, "Running created a counter-rotation between your hips and shoulders, causing your spine to twist gently. The twisting motion pulls on the ligaments and tendons in your shoulders, spine and hips, which in turn act like rubber band, wanting to return your spinal twist to its neutral position (Dryer, 59)." He continues to say that due to the rubber band effect, the legs and arms are moving as a result of the stretch and recoil of your tendons and ligaments, not because of the contraction of the muscles. He says this is practically effortless since the tendons and ligaments do not require glycogen. The more the person runs, the stronger and more flexible their ligaments become, allowing gravity to pull them forward.

2.5   Models and Dynamic Analysis of the Gait

There are certain muscles and joints that play larger roles in the gait cycle. There are also different methods of analyzing movement such as inverse dynamics. Some of the methods that are helpful in creating force body diagrams are discussed in the following sections.

2.5.1 The Musculoskeletal Model

There are two skeletal models of the lower extremity in the sagittal plane. There are three rigid bodies of importance: the thigh, the shank, and the foot (1996, Cole). The eight muscle units

considered during gait are shown in Figure 7: Two different musculoskeletal models. The masses, center of mass locations, and the mass moment of inertia can be obtained using regression equations as functions of the subject's body mass, height, thigh length, and shank length (1996, Cole). The regression equations are experimentally determined by testing large populations and then taking averages of mass, center of mass location, and moment of inertia.

1. Illiopsoas
2. Rectus Femoris
3. Glutei
4. Hamstrings
5. Vasti
6. Gastrocnemius
7. Tibialis Anterior
8. Soleus
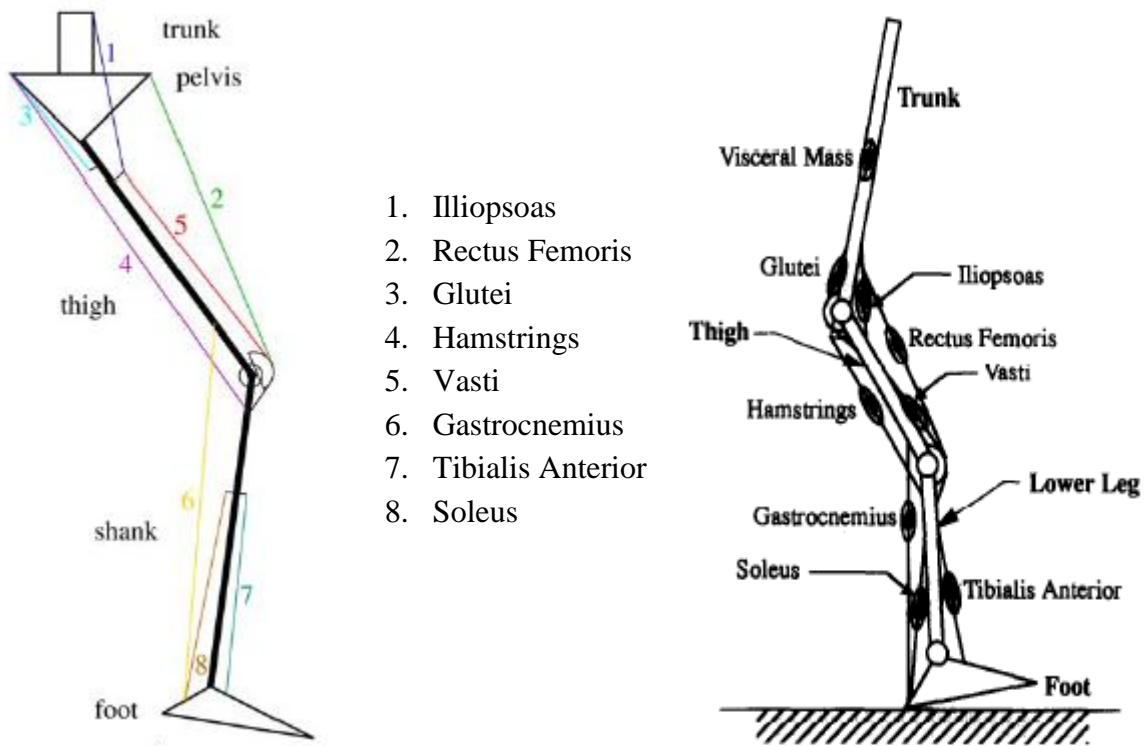
Figure 7: Two different musculoskeletal models

2.5.2 Joint Movement and Inverse Dynamics

Joint moments, also known as joint torque, may be thought of as the resultant effect of the forces exerted by the muscles. While exertion can be done voluntarily by contracting a muscle, typically it occurs in reaction to an external force (1996, Vaughan). Gravity is a common external force, but in the case of the human gait, reaction forces from the ground are also external. The

magnitude and direction of the joint torques can be established by examining the ground reaction

forces (GRF) and its line of action relative to the individual joints (1996, Vaughan).
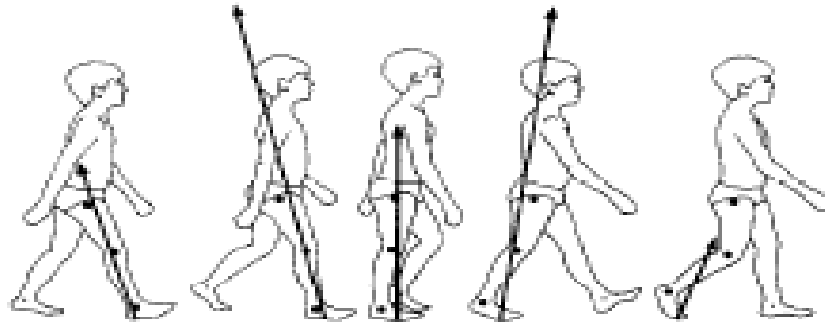
Figure 8: The ground reaction force vector, during the five segments of the stance phase, plotted with respect to the hip, knee, and ankle joint centers. The line of action of the ground force, relative to each joint, determines whether the torque is flexor or extensor. (1996, Vaughan)

Internal forces are accounted for by dissecting each body segment. While the weight of the

body segments can be calculated and the ground reaction forces measured, the bone and muscle

forces are unknown. The knowledge of the internal forces in human movement is very important

and can be addressed by creating computational models for each body segment. The estimation of

inter-segmental forces and joint moments based on external measurements is referred to as

"inverse dynamics". A specific strategy to solve for unknown variables can be done using what is

known as a "mechanical ruse" (1996, Vaughan). This involves the muscle, bone, and ligament forces

acting on the joint being reduced to a single vector with a resultant force and torque. The resultant

force and moment can be calculated based on the measured external forces and then applied

internally to determine the muscle and bone forces (1996, Vaughan). An example of the mechanical

ruse technique is applied below to the ankle joint during the terminal period of the stance phase.
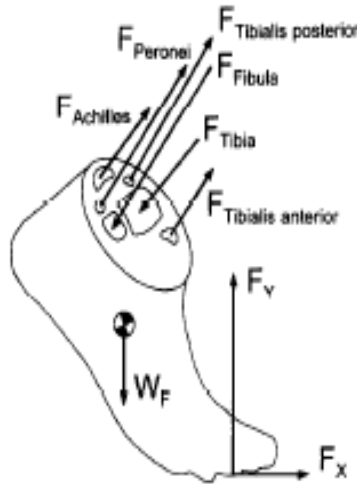
Figure 9: A diagram of the foot segment during the last portion of the stance phase. The external and internal forces are drawn as vectors. The external forces include the weight of the foot (WF) and the vertical and horizontal ground reaction forces (FY) and (FX). This model can be simplified by combining the fibula and tibia into one bone force. There are three internal unknowns of importance which are; the force of the bones (FB), the force of the Achilles (FAch), and the force of the Tibialis anterior (FTA). (1996, Vaughan)
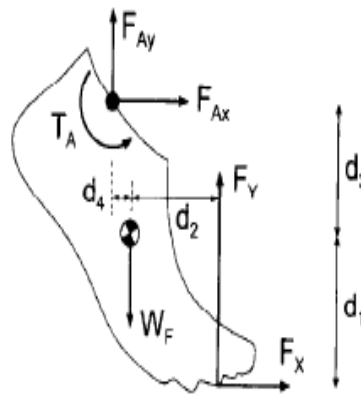
Figure 10: Shown is a free body diagram showing the muscle and bone forces at the ankle joint reduced to a single resultant force (FAY FAX) and torque (TA). These values can be calculated from the measured external ground reaction force and segment weight. Another free body diagram can be composed analyzing the internal forces using the calculated resultant force and torque. (1996, Vaughan)

2.6   *Instrumentation and Devices*

Throughout the study of gait analysis, many devices and instruments have been developed

to assist in the capture of data. Some of the possible methods that can be used to measure the data

of the gait cycle as well as spinal rotation include potentiometers, accelerometers, and video

analysis. Another method in finding data in gait analysis is though the study of computational biomechanics. These topics are discussed on the following pages.

2.6.1 Potentiometers

Potentiometers measure a range of motion (ROM). Potentiometers have had a long history but the basics principle of the device has remained the same. A potentiometer is made from a resistive element and a sliding contact (2002, Elliot). When the sliding contact moves across the resistive element, the resistance changes. A voltage is applied to the potentiometer which is a variable resistor. When an angle is changed, the potentiometer rotates and accounts for the angle change by adjusting the potentiometer voltage output accordingly.

The simplest type of potentiometer is a rotational potentiometer. When the shaft is twisted, it moves the sliding contact along the resistive element. A linear potentiometer has a shaft that moves in a linear direction and when the shaft moves linearly, it moves the contact slider along the resistive element. A more complex type of potentiometer is the string potentiometer which consists of four parts: a measuring cable, spool, string, and a rotational sensor (2008, Celesco).

Some of the advantages of potentiometers include that they are simple to use, easily integrated into different circuits, used in a variety of applications, and inexpensive. As a result potentiometers can be placed in various areas of the spine to get readings at different vertebrae.

2.6.2 Accelerometers

An accelerometer is a sensing element that measures acceleration, or a change in velocity with respect to time (Sensr). Acceleration is a vector which has a magnitude and a direction. An accelerometer measures in terms of g or 9.81m/s^2. Accelerometers measure vibrations, shocks, tilt impacts and motions of objects.

There are several different types of accelerometers that differ in the principles of their operations (Sensr). The most used types include capacitive, piezoelectric, piezoresistive, and micro electro-mechanical system (MEMS), to name a few. The principles of how each of these devices operates is different, but the idea behind all of them is that there is some force, either static or dynamic, that causes movement or vibrations which are detected by the accelerometer. The movements are changed into varying voltage, resistance, or the proper electrical characteristic depending upon the device in use. The proper electrical characteristic change depending upon the intensity of the movement. The biggest advantage of the MEMS devices is that they are relatively small. Size is important in the application of spinal and gait analysis.

Accelerometers are often used to measure the acceleration of what they are attached to; however, they can also be used to calculate velocity as well as displacement, with mathematical derivation. With the ability to measure displacement, accelerometers are useful in measuring spinal rotation. An accelerometer can be used for measurement of spinal rotation by measuring the output of the accelerometer at the spines neutral or resting position and comparing it to output measurement of the accelerometer with a rotational input of the spine. A measurement of spinal rotation can be calculated by knowing these two positions and mathematically deriving for the degree of spinal twist.

In gait analysis, accelerometers are ideal for attaching to different limbs of interest to allow for the acceleration of these joints to be determined. With accelerations of these limbs known, more complex dynamics equations can be solved. Using an accelerometer is one of several ways to solve for the accelerations of the different limbs.

As with any electronic device, there are advantages and disadvantages to using accelerometers for the application of measuring the gait cycle and spinal rotation. One of the main advantages of using accelerometers in gait analysis is that they are a less expensive alternative to

other gait and motion analysis devices. Due to their small size and little amount of extra instrumentation needed, accelerometers allow for more flexibility in testing, such that they are not restricted to a "laboratory environment" (2007, Kavangh). Another advantage to accelerometers is that they do not restrict the test subject's movement due to its size.

One disadvantage of accelerometers is there is a drift associated with them. With more expensive potentiometers, the drift may be small; however, spinal rotation and gait movements are measured with low frequency and low amplitudes. This drift can have an unwanted affect and result in more errors (2007, Kavangh). As mentioned before, gait measurements are typically low frequency and low amplitude which means that the movement may be hard to distinguish from one another (2007, Kavangh). The signal output would also have noise which may be either environmental (electronic, motion artifact, etc.) or physiological (2007, Kavangh). A method that would get rid of some of the unwanted noise is called filtering but this can also have the effect of removing some of the real signals too.

2.6.3 Video Analysis

Goniometry is an aspect of motion analysis which tries to quantify the range of motion of joints (2001, Kyriazis). One of the methods and instruments used to capture the data for goniometry is potentiometers. One way to capture movement of the body as a whole is to utilize cameras. Video analysis allows different angles of the walking/running motion to be captured for view at a later time. The two main types of camera analysis for gait cycles include systems of cinematography cameras and systems of video cameras.

Cinematography systems use markers on the subject's body so that the camera can pick up the displacement (2001, Kyriazis). The cameras are usually placed where they can get a sideways and/or frontal movement of the human gait. These systems are it is usually expensive, time

consuming to view and camera placement can cause problems due to angles causing a lack of visibility of the motion.

The main difference between cinematography cameras and video cameras is video cameras usually take more frames per second but has a lower resolution. When using video camera technology for gait analysis, markers are typically not placed on the body (2001, Kyriazis). The main disadvantage of video cameras is that it provides so much information, it makes it tough to sort through all the data. This system is also high in cost and usually requires specifically trained personnel.

One option combines both cinematography and video techniques by placing markers on the body using a separate software package to analyze and track the markers. These programs allow the locations of the markers to then be exported for analysis in programs such as MatLab or Microsoft Excel. The locations of the markers are given as pixel locations for each frame that the marker is tracked, by knowing the frame rate of the camera and a reference length the velocities and accelerations can be derived.

Video analysis may be tougher for spinal rotation tracking because video analysis typically looks at major joints and regions whereas spinal rotation requires a look at a small area with a small differential angle. Since smaller areas requires attention in spinal rotation, the use of videography would likely produce less accurate readings as opposed to larger, more noticeable movements (such as the angles between the lower leg and upper leg when walking or running).

Although video camera equipment is usually expensive, it provides an abundant amount of information that can be used towards gait analysis. Other systems that are not as high quality can be used which will reduce the cost. In addition, manual analysis will decrease the cost but will increase the length of time for the entirety of the data to be analyzed.

2.6.4 Computational Biomechanics

Computational biomechanics is the breakdown of the human body into specific rigid structures that are interconnected. Important areas such as the spine are identified and analyzed, while other regions can be simplified and assumed for the numerical calculations. The first known scholar to observe the biomechanical aspects of the human body was Borelli during the 17th century. It was not until the invention of accurate ways of measuring force and other motion that caused the surge in biomechanical analysis. Computational biomechanics takes a biomechanical model and applies the measured forces, rotational movements, or both in order to provide qualitative data. The analysis can be broken down into the different aspects of the body, such as a three dimensional analysis of the foot structure during the gait cycle (Gefen 2000). The same analysis can be applied to spinal rotation by simplifying the spine into two or three separate interconnected rigid structures. Sensors can then be placed on the separate spinal structures and the rotation between the different interconnected structures can be measured.

2.7   Summary

Locomotion is an extremely important aspect of life with studies of the human gait cycle dating back to Aristotle. The progression of modern technology has opened the door to quantitative data and new ways to obtain it. The introduction to force measurement devices, in combination with photography, revolutionized biomechanical analysis of the gait cycle. Forces could now be measured and applied to different phases of motion at certain points in time. Current technological advancements include video and computer analysis which has further increased our understanding of the gait cycle. Instruments such as force sensors and accelerometers can be attached to the foot and used to measure impact forces. A potentiometer is a device that measures the range of motion and can be used to record the degree of spinal rotation.

Because of people's individual body mechanics, correct running form is hard to define, however, it is understood that spinal motion plays a significant role in reducing shock transmission. The objective during the gait cycle is to produce the same amount of angular momentum in the upper body as the lower body, such that the total body angular momentum is zero. Runners who have poor upper body rotation cause an imbalance which leads to compensation and abnormal body mechanics. Generic models of the gait cycle have been developed to help describe the functions of different motions. It can be broken up into two phases: swing and stance. The stance phase is important because it involves a sequence of events where the foot is in contact with the floor and impact forces are distributed throughout the leg. The body's muscular system reacts to the impact of the ground in a complex manner that absorbs the force while maintaining forward limb progression.

Understanding where the force of the ground acts on the foot is crucial because it is the only external force, besides gravity. The forces in the muscles and bones are internal and activate in reaction to the impact force. Computational biomechanics can be used to breakdown and analyze specific lower limb segments. Free body diagrams and computational models can be used in conjunction with measured values to mathematically estimate the translation of force throughout the body. Over time, forces acting on the ankle, knee, and hip wear down cartilage and change the mechanical properties of the joints, resulting in significant pain and discomfort.

3   Project Strategy

Sixty-five percent of runners experience an injury in an average year (Incidence and Injury, 1993). New alternative running methods, such as Chi Running, states that with more spinal twist, the tendons and ligaments act as a rubber band to propel the body back to its natural position (Dryer, 59). The alternative running methods currently do not have numbers supporting the theory, but it is clear that there is a need for a way to reduce the injuries that occur in runners.

3.1   Problem Statement

The understanding of spinal rotation in the gait cycle and its relation to impact forces in the foot is a field of study lacking quantitative data. This experiment studied the hypothesis that increased spinal rotation reduces the forces experienced on the body during a natural gait cycle. The study aimed to measure the forces and the spinal rotation during a normal jogging gait cycle and exaggerated spinal rotation. Computational biomechanics was studied to identify the resulting forces found in the ankle and knee. In order to accomplish this problem statement, a spinal rotation device and a method of measuring the forces experienced upon landing were needed.

3.2   Objectives and Constraints

Figure 11 shows the objectives tree for the spinal rotation and foot sensor designs. These objectives were used as the basis to help decide which design options would be best for the application. The major constraint that were placed upon these deigns was a combined budget of $800.

The methods of measuring forces upon impact that are currently available include using a force plate, using a full force sensor insole, or using individual force sensors at specified places. Due to our budget constraint of approximately $800, the F-scan foot pressure system and other full foot

insoles are not possible because these systems cost approximately $12,500. When weighing the objectives, the group found allowing the test subjects to run with a natural, uninhibited gait cycle to be more important than measuring the forces with a high percentage of accuracy. For this reason, the group chose using foot sensors as the method of force measurement. The force plate is an accurate force measurement technique; however, a force plate can cause a runner to alter their stride to ensure that they land on the force plate. This would have a large effect on the resulting landing force. Knowing that the foot sensors would have less accurate force readings, they would still allow for a continuous and unaltered gait cycle.

The methods available for measuring the spinal rotation of a subject are using video analysis software, a potentiometer type device, or a simple mechanical device that would measure the maximum spinal rotation. Due to the budget constraint, video analysis software was not possible. Video analysis software that would be needed for this detailed experiment would cost upwards of $250, 000 for all the equipment. The best design for the spinal measurement device would allow for continuous spinal measurement readings. A potentiometer device would best fit this application since it could provide readings for every stride and not just a single maximum value.
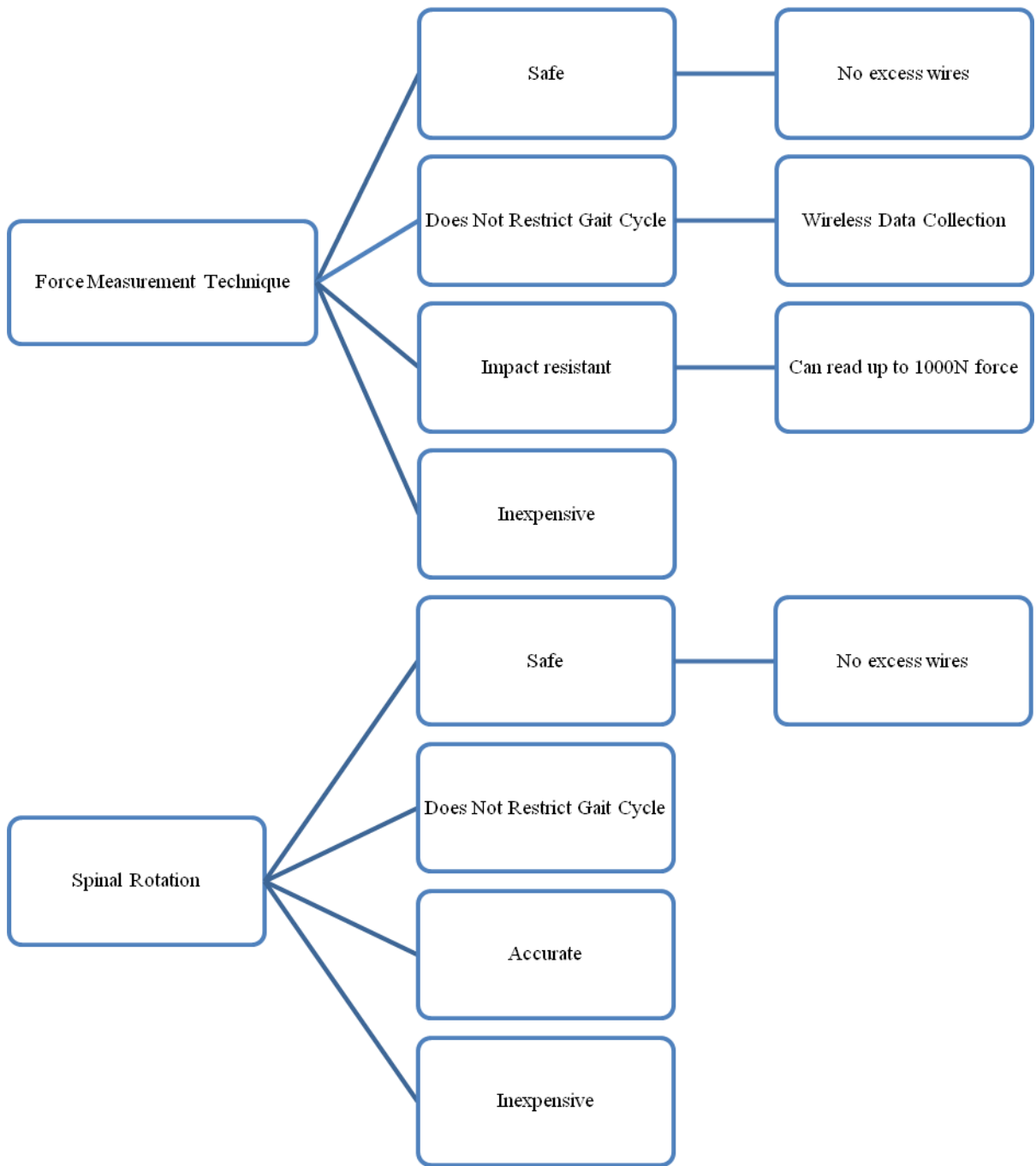
Figure 11: Objectives of Foot Sensors and Spinal Rotation Designs

4   Alternative Designs

There were two components that required a design aspect for the study: the foot sensors and the spinal rotation device. Each component had to be designed separately but with the idea of integrating each into a single data logger with four input channels, one of which would be used as an on/off switch. Before the final design of the foot sensors and spinal rotation device, preliminary concepts and designs were first developed. The alternative designs were developed and the best design chosen to test for initial results. The preliminary designs reasons the final designs were chosen are presented in the sections below.

4.1   Foot Sensors

Design 1, 2, & 3

Since the group decided upon using force sensor as the method of measuring force, each design is based on using force transducers placed into one of the insoles of the shoe. For the three designs shown below, only the right insole is depicted, the insole which will be used during the initial design had yet to be determined. The study later decided on using the left insole, which would be a mirror image of the insoles shown in Figure 12.
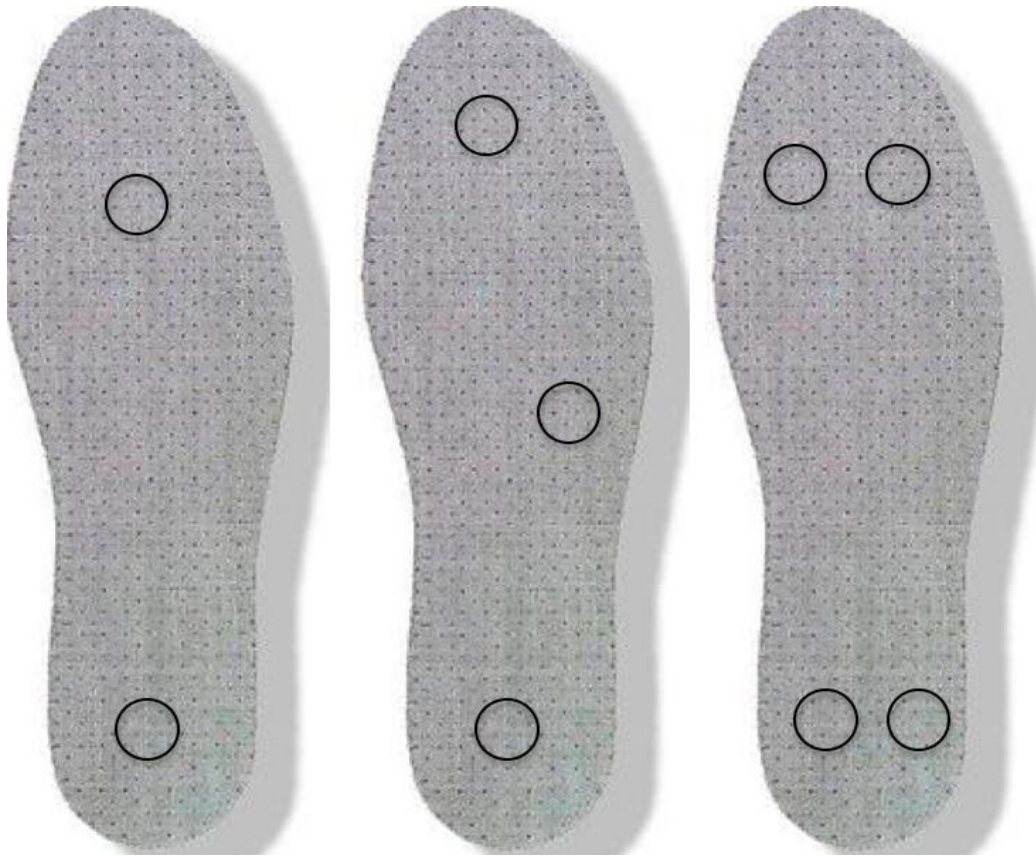
**** ◯ Denotes a force transducer

Figure 12: From left to right, Foot Sensor Design 1, Foot Sensor Design 2, and Foot Sensor Design 2

Design 1 shows two force transducers, one located in the forefoot and one in the heel, in order to accurately measure force during heel strike and toe off while in gait. The sensors will be placed on the center of pressure locations as determined by literature and using a force plate.

Design 2 shows three force transducers, similar to design one with one in the forefoot and one in the heel. One additional force transducer added along the arch of the foot to record force during mid-range of gait cycle. The transducer in the arch would allow the tracing of the center of pressure over the duration of the gait cycle.

Design 3 shows four force transducers, two located in the forefoot and two located in the heel of the insole. The pair of transducers in each location allows the study of both forces during heel strike and toe off. The advantage of this design is it is able to account better for runners with pronation of the foot during the gait cycle. The major difference in design 1 and design 3 are increase in cost of the device which correlates to an increase in the accuracy.

4.2   Spinal Rotation

Design 1:

The spinal rotation design is split into two parts: a rotation measurement device and a back brace fixation device.

The spinal rotation design is comprised of a plastic plate with holes which will allow it to be attached to the main back brace at an adjustable level. Attached to the plastic plate is a potentiometer connected to a voltage source and portable data acquisition and storage device. The potentiometer's rotating shaft is connected to a long thin strip of plastic that extends to the midpoint of each shoulder blade (this piece is also adjustable). The rotation in the potentiometer is caused by shoulder rotation, which is identical to the rotation generated by the spine. A sample of what this might look like when designed is shown in Figure 13.

The back brace fixation design is made of three components, two Velcro adjustable straps that allow proper positioning of the device and the plastic fixation plate. The two Velcro adjustable straps are needed so that the plastic fixation plate will firmly rest on the test subjects back without moving. The plastic plate has holes positioned throughout it, allowing for adjustable attachment of the spinal rotation component. The plastic plate is also slightly contoured to fit the back of the

participant and has the ability for foam padding to be added if any discomfort is felt during testing.

The possible design can be seen in Figure 13.
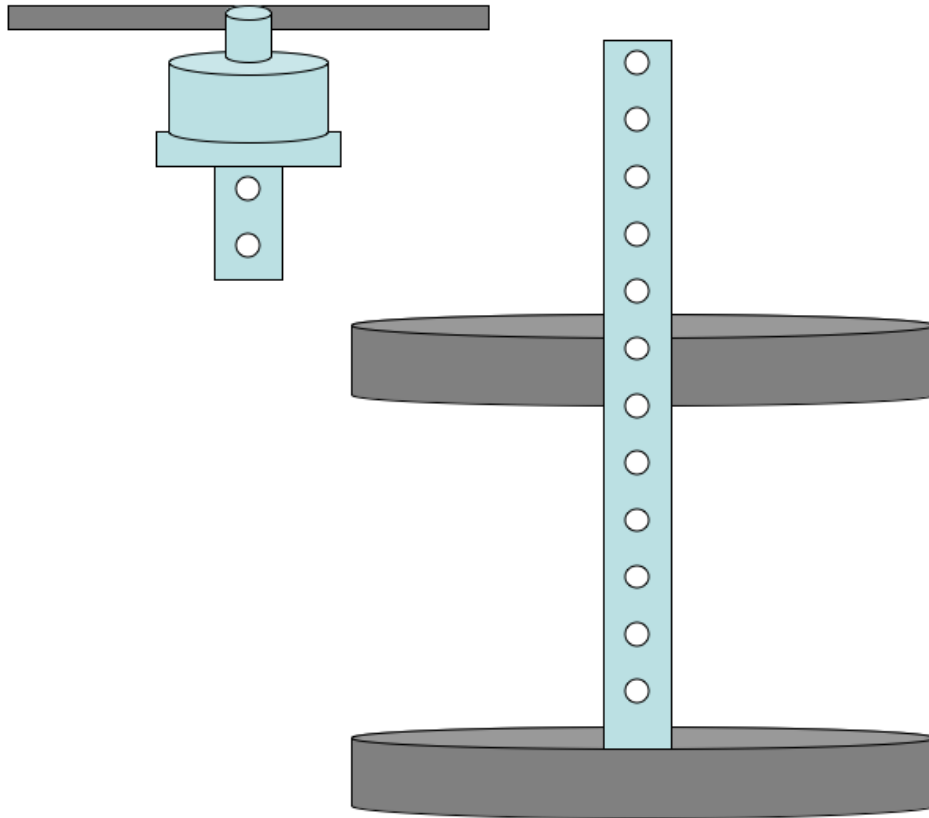
Figure 13: The spinal rotation (upper left corner) and back brace design (larger drawing)

Design 2:

The second design for the final spinal rotation is the exact same except the length of the aluminum bars that extended off the wooden dowels is longer. In this design the dimensions of the bars were 15cm in length, so the extension on each side was only 7.5cm. After preliminary testing, the design was changed for a more accurate measurement of spinal rotation.

4.3   Needs Analysis

These devices are designed to measure rotation of the spine and the forces distributed in the foot during gait. The devices need to be able to work together using a single data logger. Part of the idea of designing a new system is due to the cost of existing systems. Existing systems cost upwards of $250,000 for the computer software, video cameras, tacking markers, and foot sensors; these systems typically do not include a spinal rotation device. Spinal rotation devices range from $1,000 to upwards of $200,000 for advanced systems that measure boney prominences to accurately show spinal rotation in all three axes. The group sought to create a system that allowed the gait cycle and spinal rotation to be measured and correlated together for minimal cost.

4.4   Functions and Specifications

Functions:

- Measure reaction forces on foot during the gait cycle

- Measure rotation of the spine around the y-axis during gait cycle

- Use portable/wireless components and device

- Record voltage outputs onto a portable storage device

Specifications:

- Does not inhibit gait cycle

- High Rate of data collection (>150/sec)

- Low cost (<$850)

- Easy to use (record, download, and analyze data)

- Safe for user and technician

4.5   Preliminary Experiments

Preliminary experiments were performed to observe and test the leading preliminary designs. The designs chosen for the preliminary test were design two for spinal rotation and design one for force sensors. The subject was instructed to run and land on the force plate, placed in a 10 meter track. A camera recorded the impact at the force plate. The foot sensors were placed only under the left insole of the subject's shoe. The preliminary experiments allowed the team to ensure that the sensors were within a proper range and that both foot sensors and spinal rotation device were being recorded on the portable data logger.

The preliminary results revealed flaws in the design of the spinal rotation device, which were corrected in our final design by adding longer back pieces made of aluminum. The experiment also revealed our foot sensors were responding within the correct range for the impact of the foot. A graph from the force plate data, shown in Figure 14, was obtained through the testing. This graph was compared to the voltage outputs in the sensors to see if a distinct peak at heel strike and gradual peak at toe off was observed.

Force vs Time of Stance
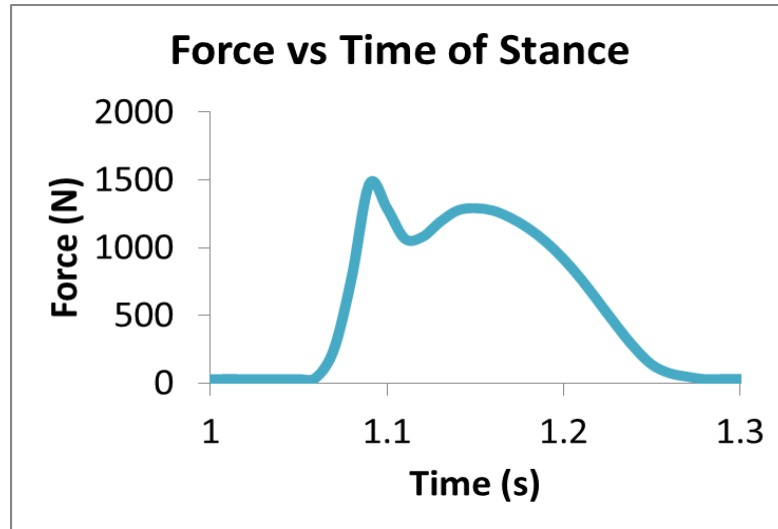
Figure 14: Preliminary Force Plate Data

All aspects of the designs were reevaluated after the preliminary tests were performed and necessary modifications were made to develop the final designs. The voltage outputs from the force sensors and spinal rotation were also evaluated at this time.

5   Final Designs and Materials

The chosen method of measuring the forces upon impact was to use foot sensors. The spinal rotation device chosen was to use a potentiometer with metal bars spanning the back. Detailed designs of the final chosen designs of the foot sensor and spinal rotation device are shown in sections 5.1 Foot sensorsand **Error! Reference source not found.**.

5.1   Foot sensors

The final foot sensor design consisted of two Tekscan FlexiForce A201 force sensors imbedded in the insole of a New Balance 509 SR athletic shoe. This was done for four different sizes of New Balance 509 SR athletic shoe: a women's size 7, a women's size 8, a men's size 9.5 and a men's size 10.5. The AMTI force plate and literature were used to find the center of pressures for heel strike and toe off phase when jogging, seen in Figure 15. This allowed for the most accurate placement of the sensors in the heel and toe region. **Error! Reference source not found.** shows the insole with the foot sensor placement.

Figure 15: Shows the center of pressures, the blue and white spots, for heel strike and toe off of the foot found in literature

Denotes a force transducer

Figure 16: Final Design of Foot Sensor

Both force sensors were attached to a circuit. The circuit schematic used is shown in Figure 16. A five-volt voltage regulator was used to regulate a nine-volt battery. Two of the circuit schematics were set up on the same breadboard to allow for both heel strike and toe off sensors to be run off of the same voltage source. The V_{out} was attached to a Pace Scientific XR440 Data Logger to allow for wireless data collection.
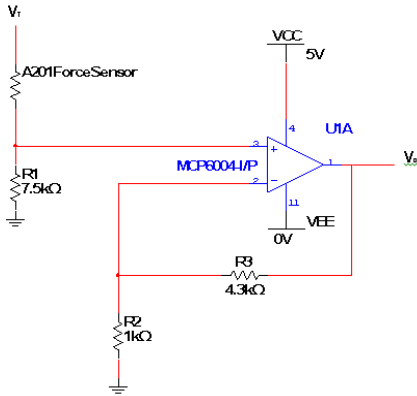
Figure 17: Schematic of the excitation circuit used for the FlexiForce A201 force sensors

5.2   Spinal Rotation

The final design of the spinal rotation device allowed for accurate measurement of spinal rotation while not restricting or limiting motion effecting the gait cycle. The device was made of two pieces, a lower and an upper portion. These were connected using a linear taper potentiometer.

The upper and lower portion of the device were made of a 7.62 cm long 2.54cm diameter wooden dowel with a 30.5cm x 3.8cm x 0.32cm aluminum bar attached in the middle so that a 15.25cm portion of the aluminum bar extended on either side of the wooden dowel. The wooden dowels were mounted in the vertical direction along the spine while the aluminum bars were mounted in the horizontal direction across the back. Velcro straps were attached to the aluminum bars to allow the device to be worn and still be adjustable for the different subjects.

The rotation of the spine was measured by attaching a potentiometer between the upper and lower portions. A 100kOhm linear taper potentiometer was used because of its size, durability, and ease of calibration. The potentiometer had a 0.63cm diameter dial that was 5cm in length. The base of the potentiometer was approximately the same size as the wooden dowel, allowing for easy

fitting. To attach the potentiometer, and connect the two portions, industrial glue was used. For the bottom portion the base of the potentiometer was glued to the end of the wooden dowel, for added strength a 2.5cm long, 2.75cm diameter PVC tubing (inner diameter 2.55cm) was glued in place over the base of the potentiometer and wooden dowel. To connect the potentiometer to the upper portion of the device, a hole was drilled in the center of the wooden dowel (0.63cm diameter and 4.5cm in depth). The dial of the potentiometer was then inserted into the hole. Friction was used to hold the dial in place and allow for the dial to rotate. This also allowed the device to be easily fixed or replaced if any problems were experienced with the potentiometer.
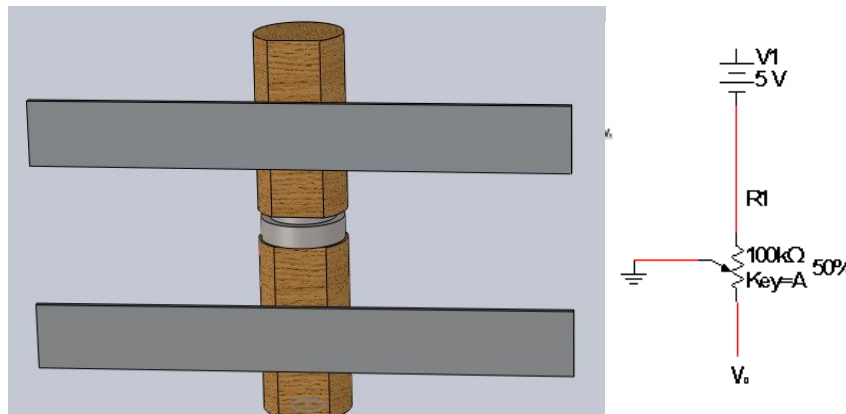
Figure 18: CAD Representation of Final Design & Final Circuit Design
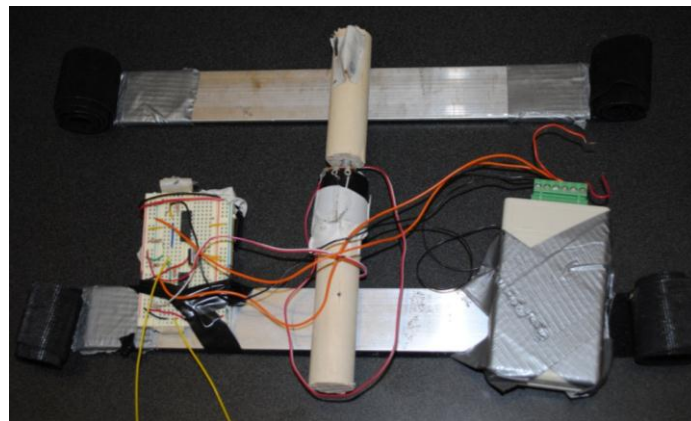
Figure 19: Final Spinal Rotation Device

The design allowed for the spinal rotation between the T1 and T12 vertebrae to be analyzed. The bottom portion of the device remained stationary while the upper portion would rotate when the subject ran or rotated their spine.

To allow for the entire system to be portable, the pocket logger, circuit, and power supply were also attached to the spinal rotation device. The pocket logger was attached on the left side of the aluminum bar on the lower portion, and the circuit and power supply were attached on the right side.

5.3   Materials

There are four main components of the system that were used to measure the forces, spinal rotation, and video analysis. The first component is the force sensors. The sensors are Tekscan FlexiForce A201 sensors that were inserted into New Balance 509SR athletic shoes. The circuit was built to allow the sensor to reach a range of 0-1000N. The circuit was built using a voltage regulator, breadboard, op-amps, resistors, and wire (a complete list can be seen in the bill of materials and circuit design found in Appendix B: Bill of Materials). The circuit was powered by a 9-volt battery. The second component was the spinal rotation device. This was made of two 7.62 cm long 2.54cm diameter wooden dowels with two 30.5cm x 3.8cm x 0.32cm aluminum bars. The potentiometer was a 100kOhm linear taper potentiometer. Two 50cm x 5cm Velcro straps were used to allow the device to be worn and adjusted easily. The spinal rotation device was assembled using industrial Gorilla Glue. The circuit for the spinal rotation device was also incorporated into the same circuit as the foot sensors. The third component of the overall system was the data logger. The Pace Scientific XR440 Pocket Logger was used because of its high data collection rate (200 samples/sec), and because of its lightweight, portable, and self-storage features. The self-storage ability allowed the device to store the data from the sensors and spinal rotation on the

device without having to attach it to a computer until after the run, making it wireless. By making

the device wireless the gait cycle was not inhibited. A laptop computer with Pocket Logger

Software was required to download the data from the pocket logger after each run. The final

component of the system was the video analysis and camera. A Casio Exilim Ex-FH100 camera and

tripod were used to record each subject's trials. The tracking markers were made of athletic tape

that was marked with an "X" for easy tracking. Adobe AfterEffects was used as the final piece of the

video analysis. Adobe AfterEffects was used to track the markers to determine accelerations,

velocities, and angles required for the computational analysis of forces.

6   Methodology

There were three distinct ways that measurements were gathered throughout this experiment. The first was using foot sensors. The foot sensors measured the force of the foot as it made impact with the ground. The next device used in this experiment was a spinal rotation device. This device measured the angle at which the spine was rotating while the test subject was in motion. The final way data was collected was through video analysis which allowed accelerations to be calculated by placing tracking markers on the center of mass of the foot, shank, and thigh. The limb accelerations along with the flexion angles of the limbs were needed to solve for the forces in the ankle and knee joints. This study tested subjects as they were jogging according to a specific guideline. The procedure and guidelines that the testing followed are explained in the following sections.

6.1   Foot Sensors

After investigating several different approaches of force measurement techniques, the group decided upon using force sensors as the method to measure the impact loads on the foot. The forces were obtained when the foot made contact with the ground when running. Since the goal of this project was to investigate the forces that the foot exerts on the ground and correlate that force to the degree of spinal rotation when running at a normal, exaggerated, and restricted form, only two force sensors were positioned in the left insole of a pair of shoes. This number of foot sensors placed in each shoe was limited by the number of input channels in the data logger; however, it was determined that two sensors would be sufficient. The force sensors were placed where the largest pressures were typically experienced during the heel strike and toe off phase of running. These exact points were decided upon by using previous studies measuring the center of pressure as well as using the force plate to find the center of pressures of one test subject per shoe

size. One image, shown in **Error! Reference source not found.**, shows the center of pressures where the FlexiforceA201 sensors were positioned.
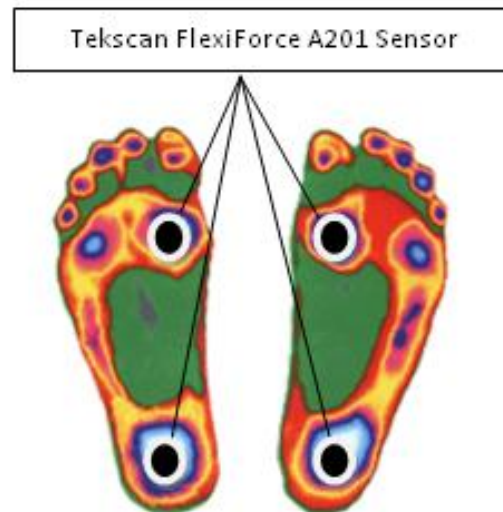
Figure 20: Foot Pressure Distributions with Ideal Placement of Force Sensors (Pedikom, 2008)

The force sensors that were used were Tekscan's FlexiForce A201 sensors which have a sensing area of 9.53 mm and the ability to measure weight up to 1000 pounds. The sensors were placed on the insole of a running shoe. They were attached to a circuit and the output voltage was stored on a Pace Scientific XR440 Pocket Logger, where the data was then available for later analyzing. Four New Balance 509 SR athletic shoes were used for testing: a women's size seven, a women's size eight, a men's size nine and a half, and a men's size ten and a half.

The concept behind the foot sensors was that, upon impact with the ground, the sensor readings would correlate to the overall force experienced upon landing. This data was then to be used in a computational biomechanics analysis which would allow for resulting forces in the ankle and knee joints to be determined.

6.2   Spinal Rotation

The spinal rotation device measured the change in voltage based on the rotation of the potentiometer. The rotation of the potentiometer was driven by the rotation of the spine and shoulders while running with a normal, exaggerated, and restricted form. The resulting motion caused the potentiometer shaft to rotate in a clockwise or counterclockwise direction, which changed the resistance of the potentiometer. The change in resistance caused a change in current which was measured and recorded using the Pace Scientific XR440 portable data logger. The device was attached to the person using two Velcro straps, which allowed for adjustable sizing and comfort while keeping it centered on the back. The spinal rotation device was positioned between the T1 and T12 vertebrae, which is the area of the spine that most spinal rotation occurs for gait and movement. The largest amount of spinal rotation occurs in the C1 to C7 region, however, this region is for rotational movement of the head and does not impact gait.

The spinal rotation device was composed of two main parts, a lower and an upper portion. Each portion was a 3 inch long, 1.25 inch diameter wooden dowel in the vertical direction. They were connected using a 100kOhm linear taper potentiometer. The potentiometer was secured to the top of the lower dowel and then inserted into a pre-drilled hole in the upper dowel and secured. On each dowel, an aluminum bar (12"x1.5"x0.125") was placed horizontally and secured. A Velcro strap was then attached to each aluminum bar to allow the device to be firmly but comfortably secured to the subject for testing. The portable data logger and circuit board for the force sensors and potentiometer were also attached to the lower portion of the spinal rotation device for increased portability. This can be seen in Figure 21.
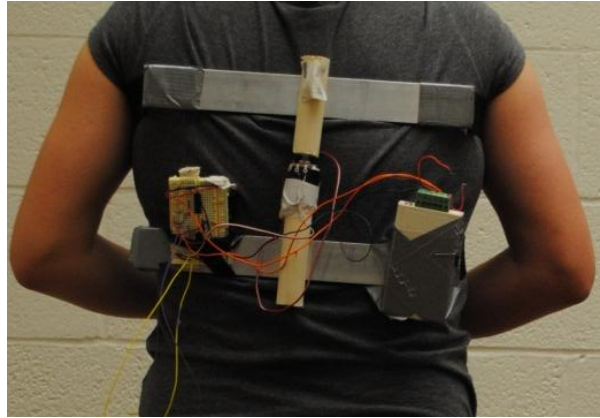
Figure 21: Spinal rotation device on test subject's back

6.3   Video Analysis

In order to calculate changes in velocity and acceleration for each body segment, a high speed, wide angle video camera was used. A Casio Exilim Ex-FH100 camera, which allowed 240 frames per second to be captured, and a tripod were used to record the subject's gait cycle. Tracking markers were placed on the subject's center of mass for their foot, shank, and thigh.

A trial run was tested in lab using a force plate to measure impact forces and moments. The subject was equipped with tracking dots placed on the center of masses of the foot, shank, and thigh. After equipped with tracking markers, they were filmed in a jogging cycle under normal running form, exaggerated spinal form, and restricted spinal form. The videos were exported to Adobe AfterEffects, which allowed for simultaneous motion tracking on the foot, shank, and thigh to be exported exportation directly to a spreadsheet. This provided the position of each point in each frame to be found. Using a known distance on the still frame and converting it to pixels, the velocities and accelerations were ultimately calculated.

Figure 22: The test subject initially made contact with his heel on the force plate. From there the body weight is shifted forward, known as load response. The third image is the beginning of mid stance.

Figure 23: Weight is shifted onto the forefoot as forward progression continues. The heel is raised off of the ground due to calf muscle action. Toe off completes the cycle as the foot is launched off the ground.

6.4   Force Diagrams

Figure 23 is a free body diagram of the foot. The segments analyzed were assumed to be rigid bodies, a common method used in biomechanics. The image shows the x and y components of the ground force in orange. This force vector acts at the center of pressure of the foot. To fit our model, we studied subjects that landed with a heel strike form. In this form, the area of the center of pressure is smaller and focused on the calcaneus (heel bone). The distances for the heel, ankle joint, and center of mass (COM) were calculated as percentages of the total foot length. To get the distance of the ankle to heel, the length from the end of the foot to the heel was subtracted by the length of the end of the foot to the ankle joint. The height of the ankle to heel was measured as a percentage of the length of the foot. The weight of the foot acts at the COM, so the distance to the ankle joint was needed. The length from the end of the foot to the COM was subtracted by the length of the end of the foot. These distances are necessary to solve for the resulting moment in the ankle joint. Three equations were used to solve for the resulting forces and moments in the ankle joint. The sum of the forces in the x and y were equal to the mass times the acceleration in those planes. The sum of the moments were equal to the moment of inertia times the angular acceleration. The three equations are show below.

$$\sum F_Y = Ma_y = Fy_{Normal} - W_{foot} + Fy_{Resultant}$$

$$\sum F_X = Ma_x = -Fx_{Normal} + Fx_{Resultant}$$

$$\sum M_{ankle} = I(\alpha) = -Fy_{normal}*(D_{a\_h}) - Fx_{normal}*(H_{a\_h}) - W_{foot}*(D_{a\_COM}) - M_{ankle}$$
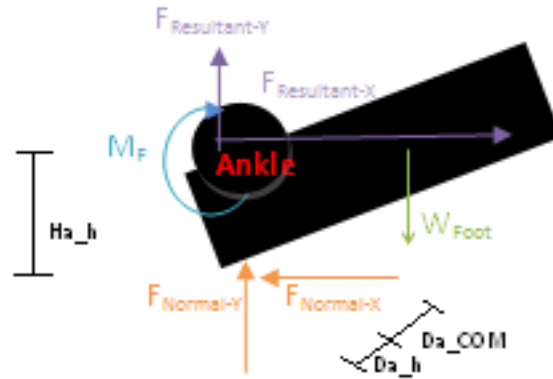
Figure 24: Free Body Diagram of the Foot

In the three equations above, the x and y resultant forces were calculated along with the resultant moment in the ankle. These three values are used to solve for the muscle forces along with the compressive force in the bone. Translating the anatomy of the foot to a free body diagram was challenging because the anatomy is not representative of how the forces act on the rigid body. A combination of models found in literature and anatomy books were used to form our model. The distances from the ankle joint to the muscles were measured as a percentage of the foot length. The tibialis anterior angle was calculated using the angle of ankle flexion, the insertion point distance on the foot, the insertion point distance on the shank, and the law of cosines. Calculating these forces also needed to account for the angle of the rigid bodies off of the x and y plane. This time the sum of the force and moment equations were equal to zero. This is because the resultant values already account for the dynamic variables. The three equations used are shown below.

$$\sum F_Y = 0 = F_{ta}*\sin((\text{angle}_{ta})-(\text{angle}_{foot})) + F_{ach}*\sin((\text{angle}_{ach})-(\text{angle}_{leg})) - F_{B1}*\cos(\text{angle}_{leg}) + F_{RY}$$

$$\sum F_X = 0 = F_{ta}*\cos((\text{angle}_{ta})-(\text{angle}_{foot})) - F_{ach}*\cos((\text{angle}_{ach})-(\text{angle}_{leg})) + F_{B1}*\sin(\text{angle}_{leg}) + F_{RX}$$

$$\sum M_{ankle} = 0 = F_{ta}*(\sin(\text{angle}_{ta}))*(D_{ta}) - F_{ach}*\sin((\text{angle}_{ach}))*(D_{ach}) + M_{ankle}$$
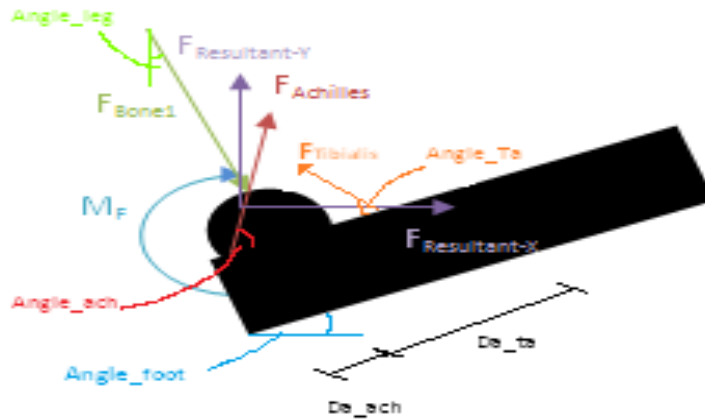
Figure 25: Free Body Diagram of the Foot with Muscles

MATLAB was used to solve the three unknown variables using matrix equations. These muscle and bone forces were then translated to the shank to solve for the hamstring, patella, and knee joint forces. The muscle vectors are always in tension and thus directed off of the rigid body. The joint force is compressive which is directed at the rigid body. When translating the vectors to the shank free body diagram, the magnitudes are equal, but the direction is opposite.

Figure 26 is the free body diagram of the shank. The only external forces acting on this rigid body is the weight of the leg. The mass and acceleration is also accounted for in the equations below.

$\sum F_Y = Ma_y = -W_{shank} + Fy_{Resultant}$

$\sum F_X = Ma_x = Fx_{Resultant}$

$\sum M_{ankle} = I(\alpha) = -W_{shank}*(D_{a\_COM}) - M_{knee}$
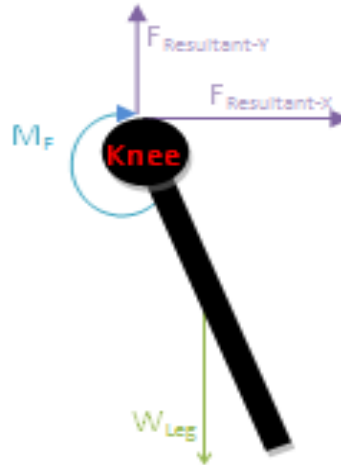
Figure 26: Free Body Diagram of the Knee

The forces that the knee joint experiences during impact are through the translation of the ankle joint force and the muscle forces. The Achilles tendon connects to two muscles, the soleus and gastrocnemius. The soleus inserts on the shank, where the gastrocnemius inserts just above the knee joint. These two muscles are fractions of the total Achilles tendon force, the soleus (2/3) and the gastrocnemius (1/3). The insertion points of the hamstring, patella, tibialis, and soleus were found using the same method that was used in the foot. The angle of the hamstring was determined using the law of cosines based on the insertion point on the shank, thigh, and the angle of knee flexion. The soleus and patella angles were measured values. Below are the three equations used to solve for the unknown variables.

$$\sum F_Y = 0 = -F_{ta}*\cos((angle_{ta})+(angle_{leg})) - F_{sol}*\cos((angle_{sol})-(angle_{leg})) + F_{B1}*\cos(angle_{leg}) + F_{RY} + F_{ham}*\cos((angle_{ham})+(angle_{leg})) + F_{pt}*\cos((angle_{pt})-(angle_{leg})) - F_{B2}*\cos(angle_{thigh})$$

$$\sum F_X = 0 = -F_{ta}*\sin((angle_{ta})+(angle_{leg})) - F_{sol}*\sin((angle_{sol})-(angle_{leg})) + F_{B1}*\sin(angle_{leg}) + F_{RX} - F_{ham}*\sin((angle_{ham})+(angle_{leg})) + F_{pt}*\sin((angle_{pt})-(angle_{leg})) + F_{B2}*\cos(angle_{thigh})$$

$$\sum M_{ankle} = 0 = F_{ta}*\sin((angle_{ta}))*D_{ta} - F_{sol}*\sin((angle_{sol}))*D_{sol} - F_{ham}*\sin((angle_{ham}))*D_{Ham} + F_{pt}*\sin((angle_{pt}))*D_{pt} + M_{knee}$$
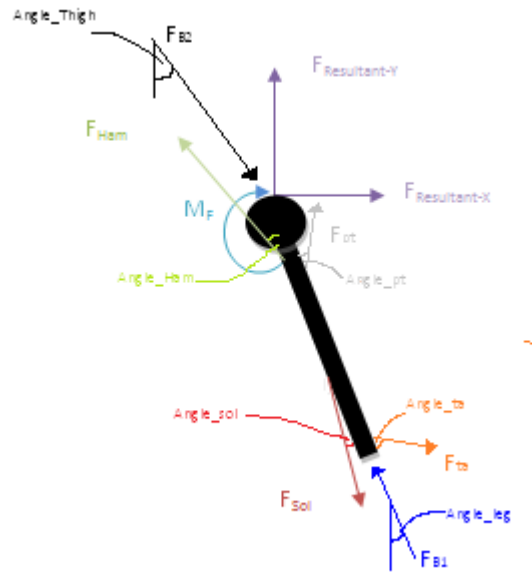
Figure 27: Free Body Diagram of the Knee with Muscles

Similar free body diagrams for the hip were developed based upon the same principles described in the foot and shank diagrams. These can be seen in Figure 28 and Figure 29. As more analysis was completed, it was discovered that, due to the complex geometry of the hip, the model could not be reduced down to only the sagittal plane.
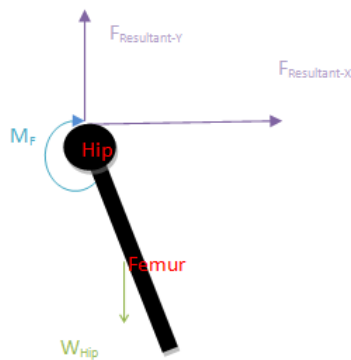
Figure 28: Free Body Diagram of Hip
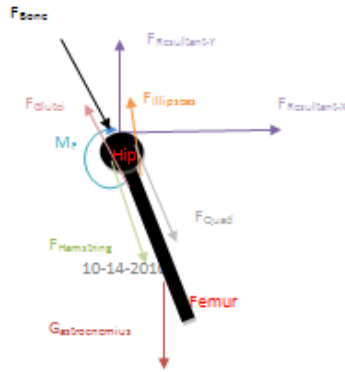
Figure 29: Free Body Diagram of Hip with Muscles

6.5   Calibration

A total of six different Tekscan FlexiForce foot sensors were used in the total study, two in the women's size 7, two in the women's size 8, and two in the men's size 9.5 (this insole was used in both the 9.5 and 10.5 shoe) . Each foot sensor was calibrated using 3 statics test and two dynamics tests. The other device that needed calibration was the spinal rotation device. The detailed method of calibration for both the foot sensors and the spinal rotation device are described in the sections below.

6.5.1 Force Sensor Calibration

In order to ensure that the force sensors values could be determined after running, the sensors needed to be calibrated. The sensors were calibrated using 3 static tests and two dynamic tests.

Static

The static calibration tests of the foot sensors consisted of placing a 4.5 kg, 11.4 kg, and a 20.5 kg weight on a small wooden dowel directly on top of the force sensing area. The sensors were

wired into the circuit with the output voltage reading into the data logger that started recording immediately upon pushing the record button. The values were plotted on a graph to show the voltage reading versus the known pressures, or force over area. Pressures were used because the voltage values read when a subject was running will be the pressure at the sensor. This was done for all six different force sensors that were used throughout this project. A sample of one of the calibration plots is shown in Figure 30.

The chart shows Voltage (y-axis, 0 to 20) versus Pressure (N/m^2) (x-axis, 0 to 1500000) with data points and trendline. The equation shown is:

$$y = 1E\text{-}05x - 0.62$$
$$R^2 = 0.93$$
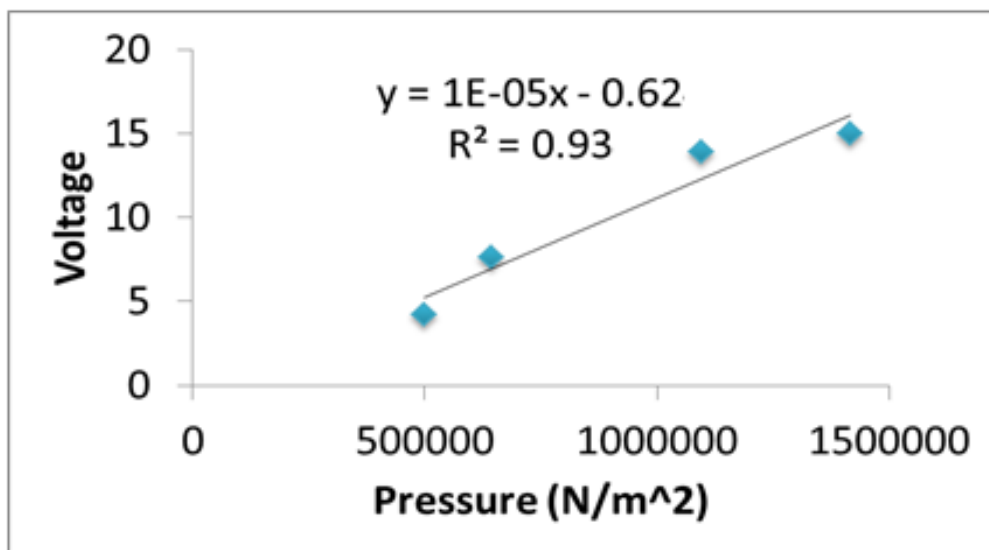
Figure 30: Sample Force Sensor Calibration Plot

Dynamic

Along with the three static tests, two dynamic tests were done to see how accurate the foot hits the sensor. The first dynamic test consisted of having a test subject place all of their weight of the foot sensor placed in the heel and remove their weight. They then placed all of their weight on the toe foot sensor quickly followed by the removal of their weight. This was repeated three times with the shoe insole placed on top of the force plate. The sensor readings and force measured by

the force plate were recorded and compared to the data found in the static calibration method to ensure that there was a close correlation in forces. This can be seen in Figure 31. From this figure, we can see that there is less accuracy in the readings. This is due to the variation in which the subject hit the sensor.
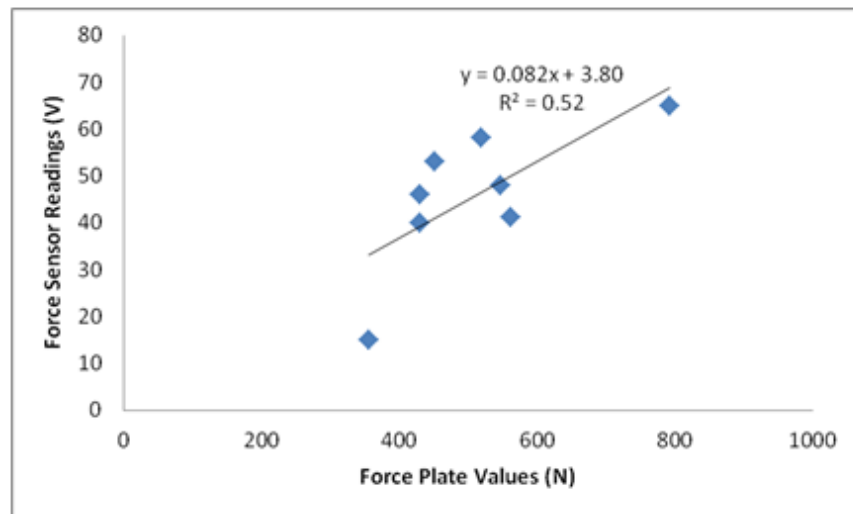
Figure 31: Force Sensors Compared to Force Plate Calibration

The second dynamic calibration method consisted of having a test subject run across the force plate while the foot sensors recorded the voltage change experienced. This test allowed the group to see the accuracy of the force sensors in the application in which the sensors were going to be used. It was found that there was more inaccuracy in the exact foot impact pressure, but a general range of the force could be determined.

6.5.2 Spinal Rotation Calibration

To calibrate the spinal rotation device, the upper rotating portion of the device was turned to known angles and the corresponding voltage output recorded. The device was wired into the portable data logger and then the device was rotated so that the potentiometer would change

position. The device was rotated in both the positive and negative direction to give a more accurate calibration based on the voltage outputs. The device was turned to 15, 30, and 45 degrees past its neutral position in both the positive and negative direction, pausing at each interval. By pausing at each interval the data logger recorded multiple voltage outputs allowing for an average to be taken. The standard deviation of each point was less than 0.03 for the voltage output, meaning the change while paused at each interval was negligible. The values and degree intervals were then exported to excel to produce a linear calibration curve. The curve was known to be linear because a linear taper potentiometer was used. The resulting calibration curve and calibration equation used can be seen in
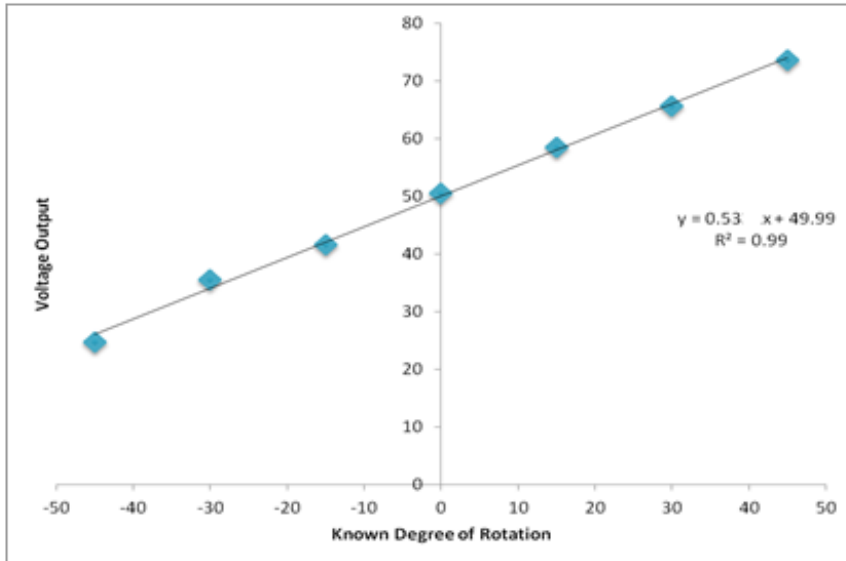
Figure 32, where "y" is the voltage output given and "x" is the resulting degree of spinal rotation.
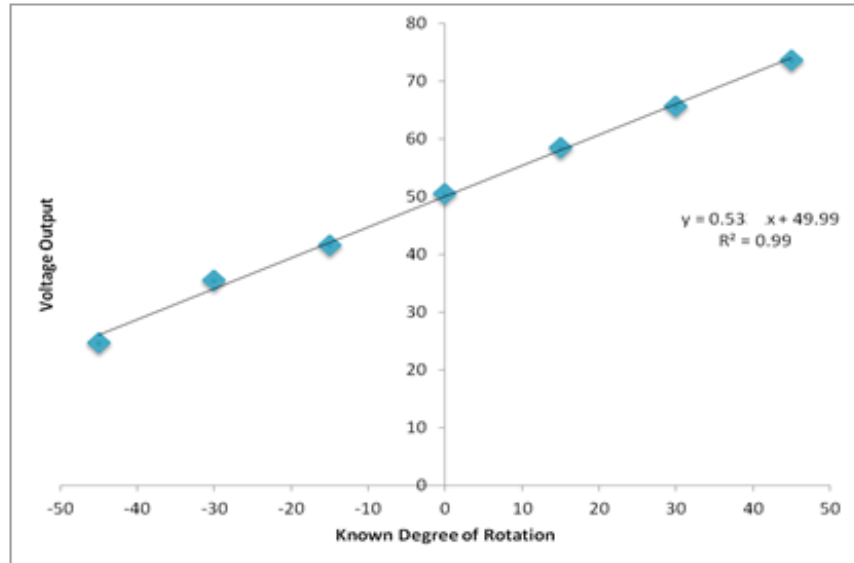
Figure 32: Spinal Rotation Calibration Curve

6.6 Testing Procedure

A Pace Scientific XR440 Pocket Logger and Pocket Logger software were used for recording the data from the force sensors and spinal rotation device during the study. The testing procedure relied heavily on the foot sensors to acquire ground reaction forces. The project used an insole device that was placed in the test subject's shoe. The insole device consisted of two FlexiForce A201 sensors placed at the center of pressure locations for gait in the left shoe. Noise in the signal was accounted for by conditioning the circuit beforehand as well as data analyzing techniques. The test subjects were assisted in putting on the equipment. The subjects also had measurements taken of their foot, shank, and thigh so that the center of mass locations were properly identified and marked using tracking markers.

The subjects were allowed time to move around and become acquainted with equipment and testing procedure. Once the subject was comfortable in the equipment, the trial runs were

begun. The subject's normal running gait was monitored over a 15 meter distance. The subject was

then asked to run in a manner that exaggerated their upper torso movement, using mainly their

arms and spinal twist to drive their stride. The subject was finally asked to run while restricting

upper torso movement. After each run, the data was downloaded off the data logger to ensure all

sensors worked properly. Based off of a visual check of the resulting graphs from the Pocket Logger

software, if the readings showed the general form expected, the subject would move onto the next

running method or repeat the same running method. This method was repeated for all three

running methods. A more detailed procedure of what the experiment followed is described in the

following section.

6.6.1 Outline of Detailed Testing Procedure

- Methodology
    - Pre-Procedure
        - Notification to participant
            - Reserve 30 minutes for testing procedure
            - Wear comfortable athletic attire (running shoes were provided)
            - Bring any medication in case of emergency
        - Equipment list

            - 4 Pairs of New Balance 509 SR athletic shoes (sizes: 7 & 8 women's, 9.5 & 10.5 men's)
            - 2 FlexiForce transducers per pair of shoes (already attached to insole)
            - Pace Scientific XR440 Pocket Logger
            - Laptop Computer with Pocket Logger software
            - Spinal Rotation Device
            - Athletic tape and marker (for tracking markers)
            - Casio Exilim Ex-FH100 camera and tripod
            - Adobe After Effects
    - Pre-test procedure
        - Description of testing procedure
        - Have participant fill out testing form and sign waiver
        - Take size measurements

- Attach tracking markers at center of mass locations on foot, shank, and thigh (athletic tape marked with a "X")
- Subject tries on shoe to ensure proper fit
- Once shoes are on, spinal rotation device attached using Velcro straps and aid from team

- Running Test Procedure
  - Start data logger and camera recording
  - Run at a constant pace (2.75-3.25 m/s)
  - Test participant runs the 15 meter track past the camera and pauses at the end, standing on right foot
  - Test participant walks back to start of track and has data downloaded from data logger onto computer

  - Test participant repeats the 15 meter run with exaggerated spinal rotation ending standing on right foot and pausing for 2 seconds before walking back

  - Test participant repeats the 15 meter run with restricted spinal rotation, ending standing on right foot and pausing for 2 seconds before walking back

- Post-procedure
  - Safely remove devices

6.7   Data Analysis

Before any testing was done, preliminary data on the test subjects was needed. The test subject's weight, height, and the length of individual limbs were measured. This information was necessary to find the center of mass of the each limb, as well as the origin and insertion points of the muscles.

There are three necessary variables that were to be acquired through experimentation: forces on the feet, rotation of the spine, and angles of limbs. The ground reaction forces were measured using foot sensors and the values were to be applied to models of the foot, knee, and hip. These calculations were be done by using free body diagrams and dynamic equations described in earlier sections. The free body diagrams can be seen in Appendix A. The free body diagrams simplified the muscles and tendons that are in the foot, knee, and hip so that the internal

translation of force through the legs were viewed. Video analysis was also important in determining the leg and joint angles which were necessary for the free body diagrams.

Once these values were solved for, the forces and spinal rotation angles were examined together. Correlations were made as to whether there was more or less force in the foot and knee joint with larger spinal rotation. From this, conclusions were made about possible injury based on the force loads.

7   Results

This study consisted of testing six male and six female subjects while jogging normally, jogging with exaggerated spinal rotation, and jogging with restricted spinal rotation. The average age of all the test subjects was 24.2 ± 12.1 years (range, 18-60 years), the average weight was 65.8 ± 9.4 kg (range, 52.2-81.6 kg), and the average weekly distance of the group was 23.1 ± 12.1 km (range, 0-80.5 km). Data from all 12 subjects was recorded for three varieties of gait cycle: normal gait, exaggerated spinal rotation gait, and restricted spinal rotation gait. The data collected was voltage readings for the spinal rotation and force sensors from the data logger as well as video that recorded the gait cycle.

7.1   *Force Sensor and Spinal Rotation Results*

The force sensors were placed at the center of pressure locations on the heel and ball of bottom of the insole for the left shoe. The insole was then cut to reveal the force sensor locations and a wooden disk was fitted in the hole directly on top of the sensors. This allowed the reading to accurately depict the impact force at that specific site during impact as opposed to spread out through the insole. The voltage output for each sensor for a sample run can be seen in Figure 33. As the figure illustrates, there are two clear spikes where for when heel strike occurred and when toe off occurred in the gait cycle. Again, because these are just voltage outputs and the sensors are all different, this data needs to be compared to the calibrated values before accurate forces can be observed. The voltage readings show a similar trend to a normalized gait cycle plot, with a distinct heel strike spike followed by a more gradual toe off region with a second peak. This form of the curves match the form of the graph received from the force plate in the preliminary results section.

Spinal rotation was measured using the voltage change that was recorded by the pocket logger. Once the pocket logger information was downloaded after each test subjects' run, the data

was input into Microsoft Excel for analysis. The raw data, seen in Figure 33, shows the voltage change over time based on the rotation of the spinal measurement device. The raw data was later analyzed using a calibration curve to give spinal rotation measurements in degrees instead of a voltage output. The voltage change observed shows a clear starting plateau, gradual decrease during the gait cycle, and gradual plateau at the end of the graph. This data is only for the left foot landing. As the data continues, the voltage output shows a reverse trend returning toward its original starting point but the values of interest are from the peak point to the minimum point to find the maximum degree of spinal rotation at this heel strike. From the calibrated data, this particular subject showed a spinal rotation of approximately 8 degrees.
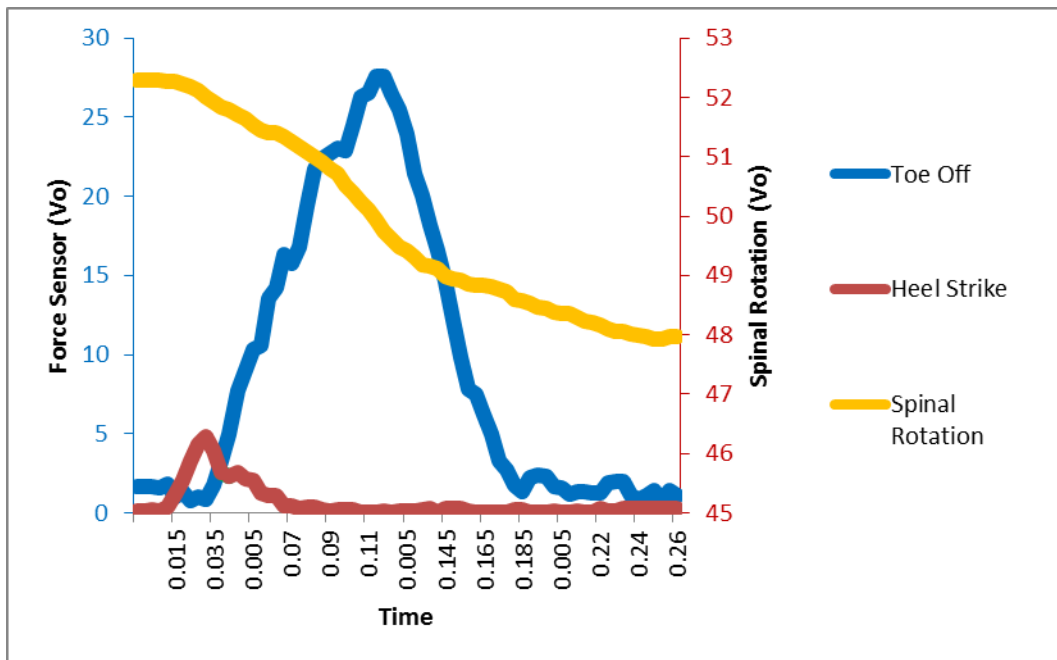
Figure 33: Voltage Output Readings

7.2 Video Results

Each test subject was recorded for each trial run that they performed. The camera recorded from 2.5 meters away to allow for a full 3 meter horizontal span to be recorded. This

allowed the subject to run with a completely unrestricted gait, because it did not force the subject to run within a frame area. The wide angle that was achieved enabled a minimum of one full gait cycle to be captured. The videos were then loaded into Adobe AfterEffects for analysis. One frame of the obtained video can be seen below in Figure 34.
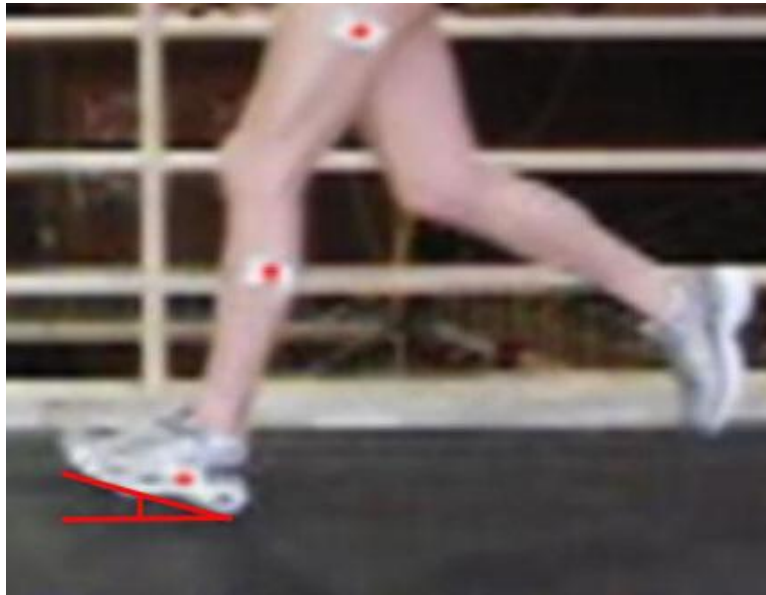
Figure 34: Example Frame from Video

8   Analysis

The analysis of the data took place in three stages: analysis of the voltage outputs from sensors, analysis of the video components, and a final computational analysis solving for forces. The voltage outputs were analyzed for both the degree of spinal rotation and forces experience in the heel and toe during gait. They were analyzed by calibrating the sensors using known values to produce a curve, then applying the curve to the voltage outputs giving rotational and force measurements. The analysis of the video components took place using a combination of Adobe AfterEffects and Microsoft Excel. The analysis revealed the angles, distances, velocities, and accelerations of limbs that were required for the computational analysis of solving for forces. The first two stages of analysis allow for the final stage to give resultant forces experienced throughout the foot and knee during gait.

8.1   Voltage Output Analysis

Each subjects output voltages for the spinal rotation device and foot sensors were downloaded from the data logger after each trial run. The data was then saved into a text file so that excel could be used to read the resulting data. Once in Excel, the calibration curves were applied to the voltage outputs to give the degree of spinal rotation and corresponding force readings.

8.1.1 Force Sensor Output

After all the tests were complete for each subject, the force sensor data was analyzed. When studying the force sensor data, the group noticed that the force values compared to the calibrated force values were significantly low for what they should have been (upwards of 2 times body

weight). When reviewing the video data for those subjects, it was noticed that, due to variations in running techniques and landing styles, subjects that landed with a toe strike, pronation, supination, or other variations upon landing forms, the force sensors were not placed in the accurate location for the maximum force experienced to be measured. For this reason, the force sensor readings were unable to be used in further analysis where the group had hoped.

8.1.2 Spinal Output

The information recorded by the data logger was exported into Excel and then analyzed using the calibration data that was performed earlier. Spinal rotation measurements were taken for all three trials of the majority of subjects. The subjects not analyzed were subjects that did not meet the requirements of heel strike and consistent velocity. The maximum spinal rotation, from peak plateau to valley, was determined by graphing the values in excel. The maximum value was then recorded on

Table 1. Subject 1 and 2 did not run with a restricted spinal rotation, and subject 3 and 4 did not

exhibit heel strike due to a mid-foot or fore-foot landing, pronation, or a combination of both. The

average spinal rotation for a normal gait cycle was 6 degrees, exaggerated gait cycle was 12.7

degrees, and restricted was 3.8 degrees. The percent change between regular gait and exaggerated

and between regular and restricted gait can be seen in the last row of

Table 1.

Table 1: Spinal Rotation

Subject	Regular	Exaggerated	Restricted
1	4.8	16.4	NA
2	3.6	6.1	NA
3	NA	NA	NA
4	NA	NA	NA
5	7.5	11.35	3.6
6	11.4	17.6	8.9
7	*4.9*	13.6	2.3
8	7.8	18.8	4.9
9	3.7	7.3	2.1
10	6.5	12.3	2.5
11	4.1	9.7	3.4
12	6.2	14.2	2.5
Averages	6.05	12.7	3.4
Percent Change from Regular		210.5	62.4

8.2   Video Analysis

Each video was analyzed separately using Adobe AfterEffects and the resulting values imported into Microsoft Excel for further analysis. The first part of the video analysis recorded the position of the tracking markers, in pixel location, during the impact of the foot during the gait cycle. Each marker was tracked separately and the corresponding pixel locations of the marker for each frame were exported into Excel.

Once in Excel, the pixel locations of the markers were identified by the frame number and the corresponding heel strike frames were analyzed. After the frames were chosen, the pixel locations for each frame were put into a formula that solved for angular velocity, angular acceleration, Cartesian velocity, and Cartesian acceleration. These components were later used in the computational bioanalysis.   The average acceleration in the x-direction and y-direction were

the two values that were closely observed. Due to poor resolution of each frame from the camera, the tracking varied from subject to subject as well as from test to test. The sensitivity of the data resulted in a range of accelerations. The four subjects that were chosen for further analysis were reanalyzed to ensure the pixel locations corresponded with an accurate location of the tracking marker. The result was accelerations that still were significantly different. The average x-acceleration was 1.2±0.9g and the average acceleration in the y-direction was 4.1±1.6g.

Foot impact angles were also calculated using this program. Due to difference in gait forms no average or statistical data could be retrieved from this. Other important angles were calculated using the positions of the tracking markers, specifically the angle of the shank and thigh. These values were then utilized during the computational bioanalysis of individual subjects.

8.3   Computational Bioanalysis

Due to the variation in each test subject's running form, it was difficult to make assumptions about the gait cycle as a population. Some runners land with heel strike rotation to mid-foot stance, while others land on their forefoot. Cases were also observed where a subject's ankle pronates and supinates in the third plane. There was no method to accurately account for these variations. The foot sensors were placed on the heel insoles of the shoes of the test subject. The curves of the voltage readings of the sensors matched the ground reaction force curves seen in literature and preliminary testing. However, the range of the voltage readings varied with each subject due to the unique land form of each individual. Even when subjects displayed the heel strike form, the center of pressure distribution for each subject is different. The voltage readings and calibration techniques used were not an accurate depiction of the impact force the test subjects were experiencing.

Due to the differences in gait and pronation, four subjects that appeared to have a clean heel strike were chosen and their data was analyzed. Two females and two males were selected since they had running velocities of 3.5±0.1 m/s with little variation between the restricted, normal, and exaggerated running form. Since the force sensors did not accurately read the maximum pressures experienced, the impact forces were assumed to be two times the subject's body weight, a value that matched previous studies. This assumption was made so that the physical effects of the altered running forms could be observed. A limitation with this assumption is that we expect variations in the impact force based on form.

The accelerations found for all four of the test subject's limbs in the x and y direction are shown in . These accelerations were used in the force and moment equations.

Table 2: Acceleration of Limbs in the X and Y Direction

		Ay Foot (g's)	Ax Foot (g's)	alpha foot (rad/sec^2)	Ay Knee (g's)	Ax Knee (g's)	alpha knee (rad/sec^2)
Subject 10	Restricted	5.3	-3.8	-440.7	2.1	3.9	10.0
	Normal	4.1	-16.1	-428.0	-1.8	-12.5	28.1
	Exaggerated	5.9	-4.2	-667.6	-2.3	-1.9	41.9
Subject 5	Restricted	3.1	-1.8	-365.3	2.7	-2.8	100.3
	Normal	2.7	-2.8	-490.2	1.9	2.4	78.6
	Exaggerated	2.6	-1.4	-308.5	1.6	0.4	-12.1
Subject 6	Restricted	4.4	-3.2	-494.7	-1.9	-2.3	5.9
	Normal	2.7	-2.4	-321.0	-1.3	-1.4	42.6
	Exaggerated	2.8	-2.4	-310.3	-2.7	2.6	70.0
Subject 11	Restricted	3.4	-1.1	-420.9	-2.3	-2.4	-7.5
	Normal	2.8	-8.6	-421.3	-1.2	-1.0	1.2
	Exaggerated	2.6	-0.3	21.7	0.8	0.2	6.8

The four test subjects used for data analysis had changes in their stride length between the restricted, normal, and exaggerated running form. Less upper body rotation was correlated to a shorter stride length. There was no correlation between ankle flexion and the magnitude of the joint forces in the ankle. These compressive forces are largely attributed to the center of pressure and where the impact force acts. However, the significance of the hamstring muscle in distributing

load in the knee was observed. Knee flexion is important in determining how the hamstring acts off of the bone. The less knee flexion a subject has upon impact, the more the hamstring can divert load from the knee joint. There was no correlation found between knee flexion and stride length alone.
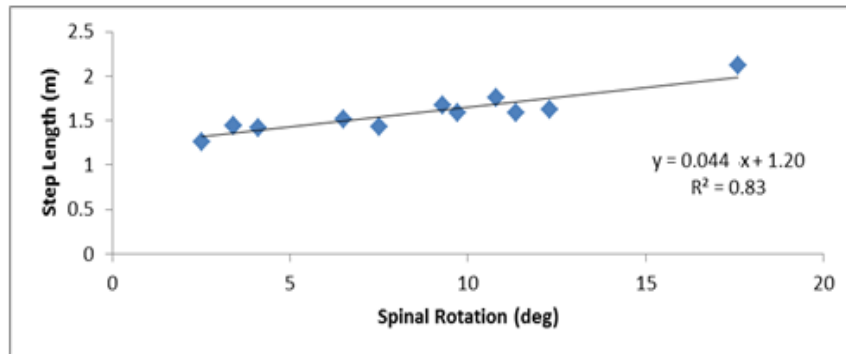

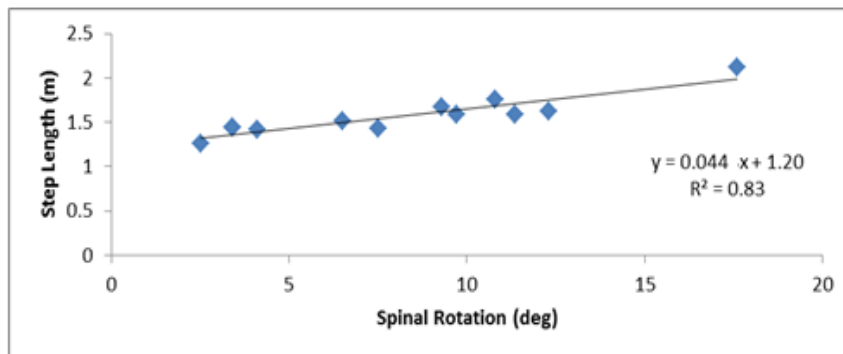**Figure 35: Stride Length due to Spinal Rotation**

Figure 35 shows four subjects that were tested with the different running forms. A linear trend shows that with more spinal rotation there is an increase in step length.

Regression analysis was performed to see if there is statistical significance between spinal rotation and the length of the subject's stride. Our null hypothesis is that the degree of spinal rotation during the gait cycle has no effect on stride length. Using the regression tool from the data analysis function in Excel, a p value of .00004 was returned. With this value, the null hypothesis is

rejected which means that the trend found is ascribed not only to chance alone. Since our p value is lower than the significance level of 0.05, our results are statistically significant.

Below are tables of the four subjects analyzed. As you can see there is no correlation between flexion angles and the magnitude of impact forces in the joints.

Table 3: Subject 10

	Spinal Rotation (deg)	Velocity (m/s)	Step Length (m)	Ankle Flexion Angle (deg)	Ankle Force (BW)	Knee Flexion Angle (deg)	Knee Joint Force (BW)
Restricted	2.5	3.63	1.26	74.0	15.5	157	13.5
Normal	6.5	3.61	1.52	74.2	15.0	155	14.1
Exaggerated	12.3	3.65	1.62	75.7	15.3	150	10.7

It can be seen that Subject 10 has no correlation to ankle flexion and the ankle force. It is also notice that there is not much change in the knee flexion angle between the restricted and normal form. It was notice, however, that the exaggerated form has a noticeably smaller knee flexion angle as well as a reduced knee joint force.

Table 4: Subject 5

	Spinal Rotation (deg)	Velocity (m/s)	Step Length (m)	Ankle Flexion Angle (deg)	Ankle Force (BW)	Knee Flexion Angle (deg)	Knee Joint Force (BW)
Restricted	3.4	3.56	1.4	81.5	15.7	158	13.7
Normal	7.5	3.31	1.4	84.0	13.3	163	12.0
Exaggerated	11.4	3.54	1.6	84.8	15.5	166	11.7

Subject 5 exhibits a progressively larger knee flexion angle with the increase in spinal rotation. The force in the knee joint decreases slightly and this shows that one component does not determine how the force is distributed. Knee flexion does not solely affect the hamstring. The angle of the thigh in terms of the y axis also has an effect.

Table 5: Subject 11

	Spinal Rotation(deg)	Velocity (m/s)	Step Length (m)	Ankle Flexion Angle (deg)	Ankle Force (BW)	Knee Flexion Angle (deg)	Knee Joint Force (BW)
Restricted	3.4	3.56	1.44	81.5	15.7	163	17.3
Normal	4.1	3.38	1.42	85.3	14.4	161	16.7
Exaggerated	9.7	3.46	1.60	82.7	12.4	157	14.2

Table 5: Subject 11 shows the data from Subject 11. Again, as illustrated in Table 5, little correlation can be pinpointed to one variable. Instead, it was concluded that the effect of the upper body rotation on stride length is significant. Stride length does not impede one variable alone but the proper increase in stride length could affect the hamstring's ability in such a way to distribute force from the knee joint.

Table 6: Subject 6

	Spinal Rotation(deg)	Velocity (m/s)	Step Length (m)	Ankle Flexion Angle	Ankle Force BW	Knee Flexion Angle	Knee Force BW
Restricted	9.3	3.26	1.7	78.0	15.1	162.7	13.4
Normal	10.8	3.34	1.8	76.3	14.9	164.2	16.8
Exaggerated	17.6	3.65	2.1	76.7	14.7	162.6	10.9

It can be noted that Subject 6 has similar compressive forces in the ankle joint for all 3 forms. Knee flexion in the normal test run is higher than the other two, thus resulting in a larger compressive knee force. The significance of knee flexion can be noticed in this test subject. The straighter the subject ran resulted in a larger magnitude of force felt in the knee.

When using the force body diagrams and computational biomechanics equations to solve for the resulting forces in the ankle and knee joints, the forces exerted on the surrounding muscles were also calculated. The bone and muscle forces for all four test subjects can be seen in

Table 7. The values for all muscles and bones forces experienced fell in ranges found in literature.

Table 7: Muscle and Bone Forces Experienced in Three Running Forms

		Fta (BW)	Fsol (BW)	Fank (BW)	Fham (BW)	Fpt (BW)	Fknee (BW)
Subject 10	Restricted	1.85	11.50	15.53	8.33	2.15	13.46
	Normal	1.83	11.16	15.01	8.18	3.40	14.14
	Exaggerated	1.80	11.33	15.33	5.81	2.15	10.71
Subject 5	Restricted	1.80	9.75	13.23	9.11	2.48	13.71
	Normal	1.77	9.81	13.32	7.33	2.33	12.02
	Exaggerated	1.78	11.22	15.48	6.19	1.90	11.66
Subject 6	Restricted	1.98	10.96	15.07	7.16	3.07	13.39
	Normal	2.01	10.88	14.89	9.10	4.61	16.79
	Exaggerated	1.91	10.75	14.68	5.42	2.30	10.88
Subject 11	Restricted	1.93	11.39	15.65	9.93	4.15	17.27
	Normal	1.77	10.61	14.39	10.01	4.21	16.74
	Exaggerated	1.86	9.24	12.41	8.59	3.80	14.17

9   Discussion and Conclusions

This study cannot conclude that the change in spinal rotation has a direct effect on the forces experienced in the foot and knee during the gait cycle. Possible future correlations were noticed. However, due to a small group size and variation between subjects, no conclusive theories were found or supported. Relationships between spinal rotation and stride length were observed as well as a reduction in knee compressive force. The trend between increased stride length and increased spinal rotation was not due to chance alone. When statistical analysis was performed, the calculated p value was lower than the .05 confidence interval.

For future studies in this area, using a full foot insert with multiple pressure sensors is recommended. Although these systems are costly, it is necessary to have many foot sensors in the insole of a shoe due to the countless variations in landing style from one person to the next. Using a full foot insert will ensure the subject does not alter their natural running form as well as provide more accurate pressure readings.

It is recommended that a three dimensional musculoskeletal model be used as opposed to a two dimensional musculoskeletal model. This will allow for fewer assumptions to be made about the lower limbs and it will also allow for hip forces to be calculated. Viewing the model in only the sagittal plane does not allow for an accurate depiction of the hip structure, which makes it difficult and very inaccurate to solve for these forces. In order to obtain accurate results for the use of a three dimensional model multiple cameras placed at different angles would have to be used. Also it is recommended that the camera have a similar if not faster frame rate and greater resolution. With greater resolution the tracking of the markers would be more accurate and decrease the sensitivity of the results for accelerations.

Based on this study, a correlation between stride length and spinal rotation was noticed. Future studies, with a larger test population, could confirm the effect of spinal rotation on a subject's stride length.

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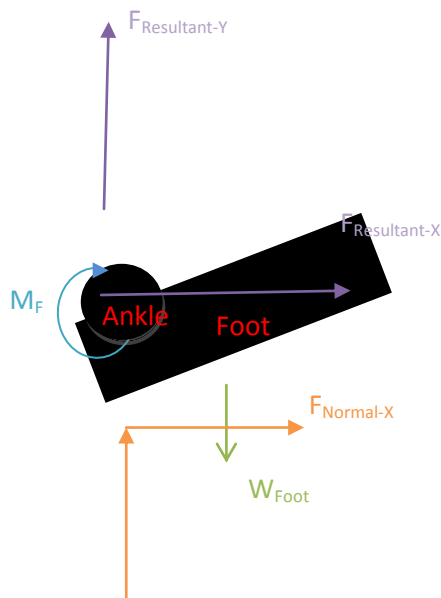
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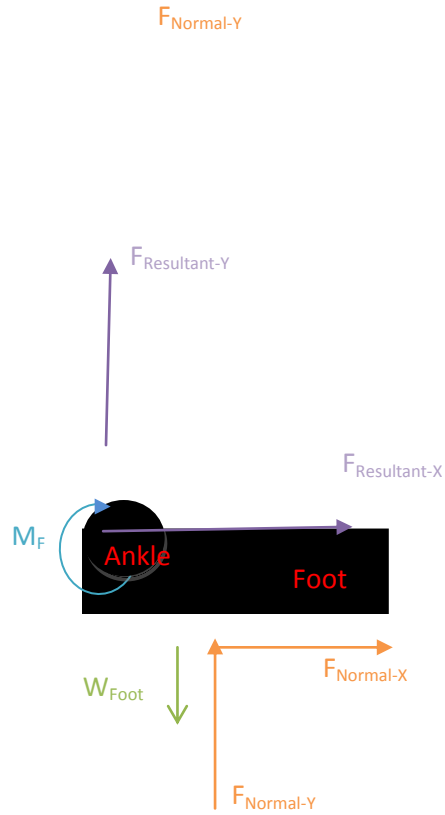# Appendix A: Free Body Diagrams

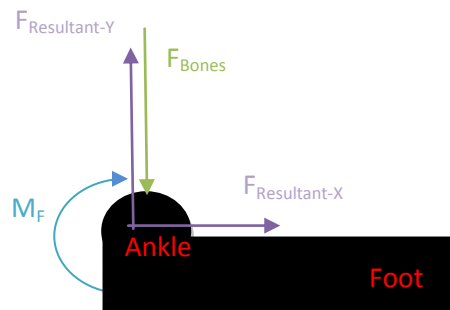## Initial contact

Foot- External forces

Foot- Internal Forces
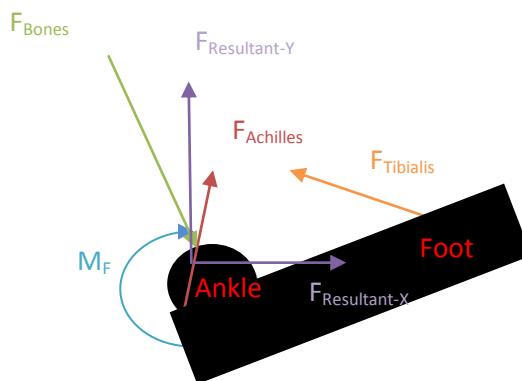
Midstance

Foot- External forces

$F_{Normal-Y}$
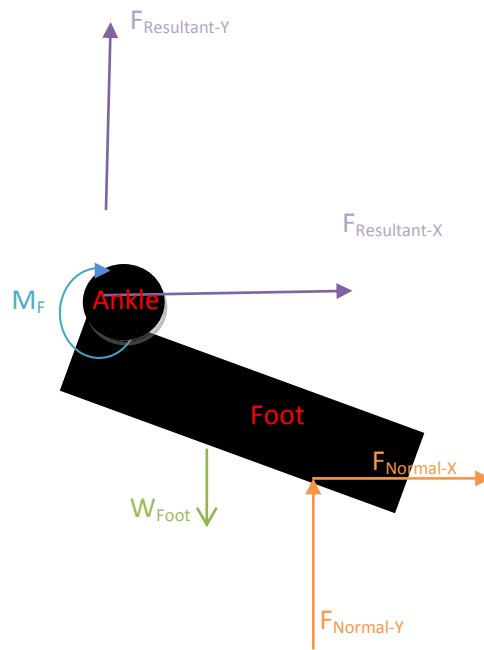
$F_{Resultant-Y}$

$F_{Resultant-X}$

M_F

Ankle

Foot

W_{Foot}

$F_{Normal-X}$

$F_{Normal-Y}$

Foot- Internal Forces

$F_{Resultant-Y}$

F_{Bones}

M_F

$F_{Resultant-X}$

Ankle

Foot

Push off

Foot- External forces

F_Resultant-Y

F_Resultant-X

M_F

Ankle

Foot

F_Normal-X

W_Foot

F_Normal-Y

Foot- Internal Forces

F_Bones

F_Resultant-Y

F_Achilles

F_Resultant-X

M_F

Ankle

F_Tibialis
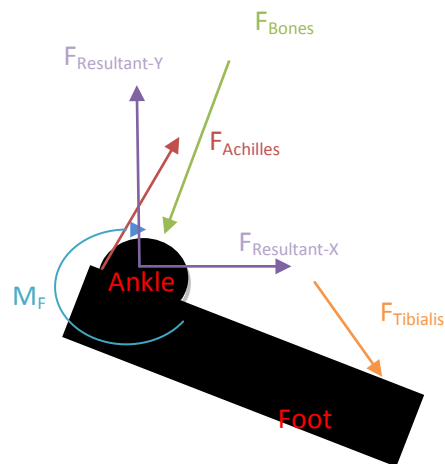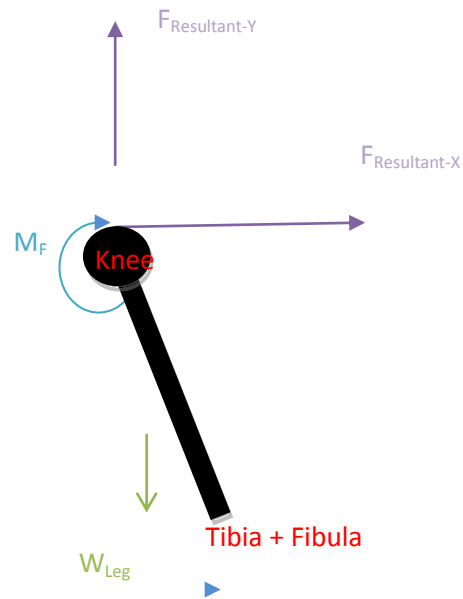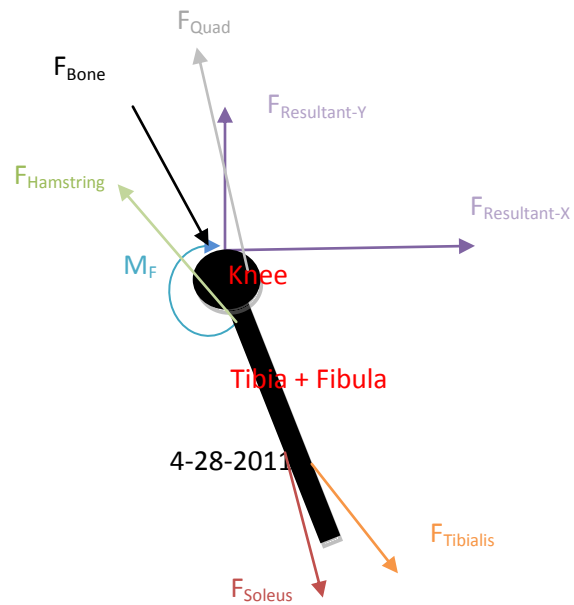
Foot

Initial contact

Knee- External forces

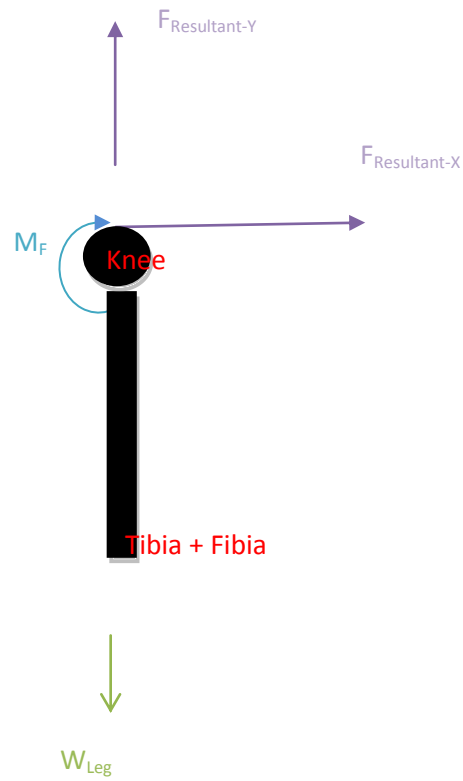

Foot- Internal Forces

Midstance

Knee- External forces

$F_{Resultant-Y}$

$F_{Resultant-X}$

M_F

Knee

Tibia + Fibia

W_{Leg}

Knee- Internal Forces

F_{Bone}

$F_{Resultant-Y}$

$F_{Resultant-X}$

M_F

Knee

Tibia + Fibula

Push off

Knee- External forces

$F_{Resultant-Y}$

$F_{Resultant-X}$

M_F

Knee

Tibia + Fibula

W_{Leg}

Knee- Internal Forces

F_{Bone}

$F_{Resultant-Y}$

$F_{Resultant-X}$

M_F

Knee

Tibia + Fibula
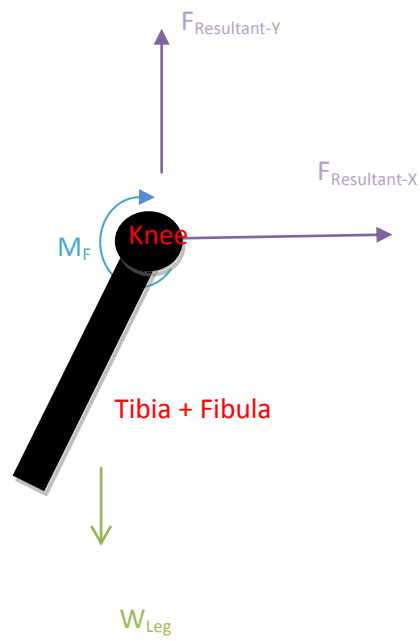
F_{Soleus}
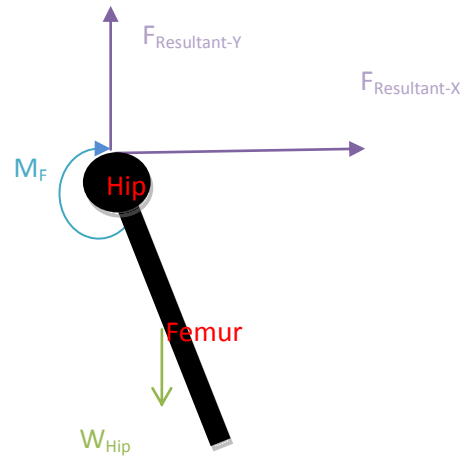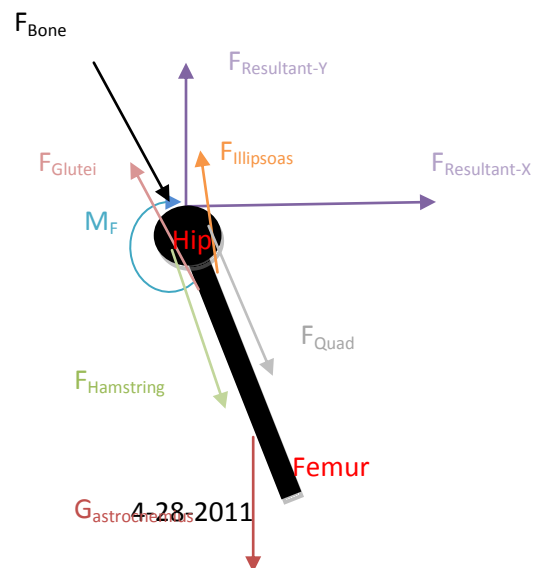
Initial contact

Hip- External forces

Knee- Internal Forces

Midstance

Hip- External forces

$F_{Resultant-Y}$

$F_{Resultant-X}$

M_F

Hip

Femur

W_{hip}

Hip- Internal Forces

F_{Bone}

$F_{Resultant-Y}$

M_F

Hip

$F_{Resultant-X}$

Femur
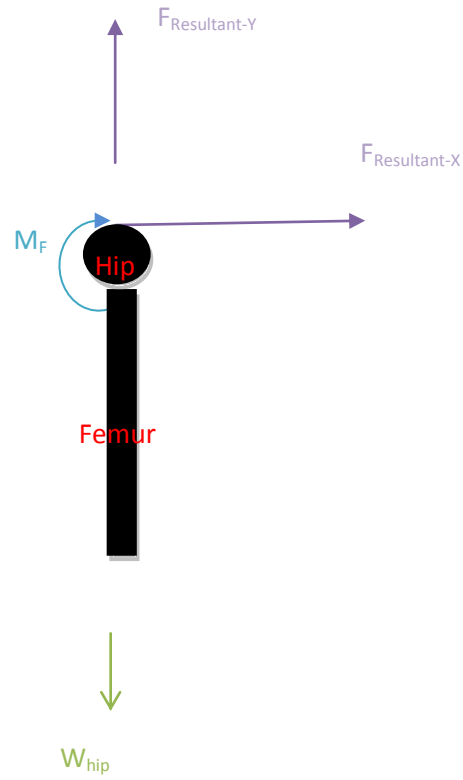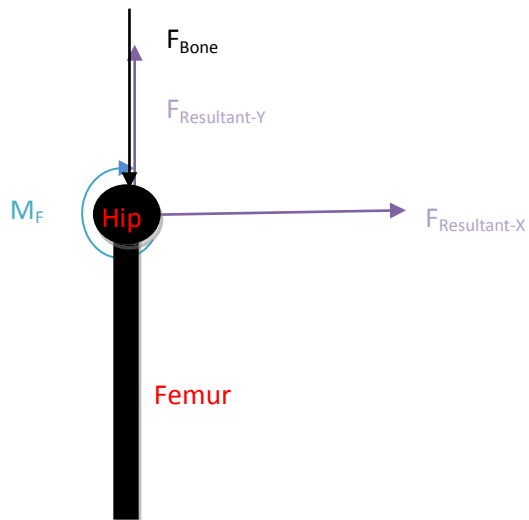
Push off

Hip- External forces

$F_{Resultant-Y}$

$F_{Resultant-X}$

M_F

Hip

Femur

W_{Hip}

Hip- Internal Forces

F_{Bone}

$F_{Resultant-Y}$

$F_{Resultant-X}$

M_F

Hip

Femur
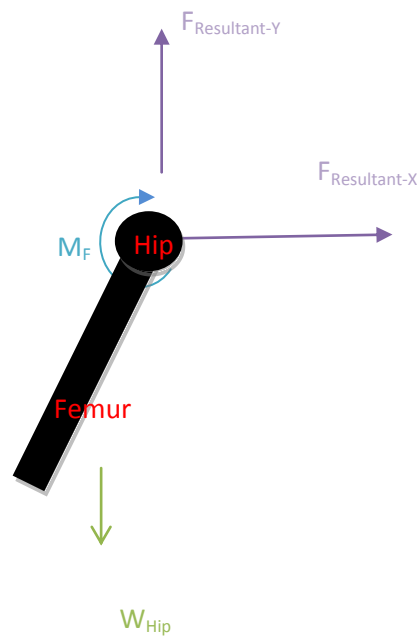
$F_{Gastrocnemius}$
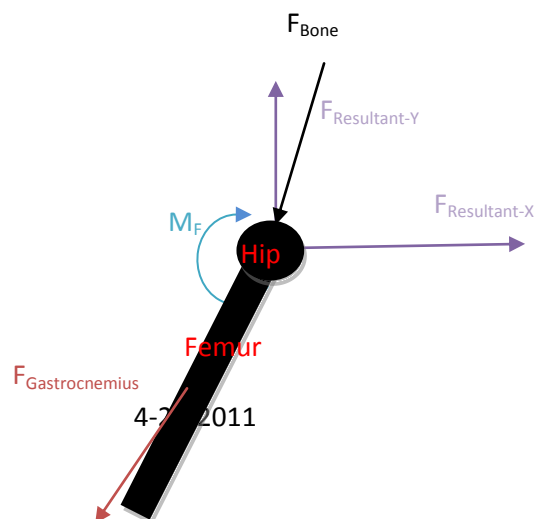
Appendix B: Bill of Materials

Subsystem	Part/Description	Cost per Unit ($)	Number of Units	Total Cost ($)
Spinal Rotation	Wooden Dowel (1" diameter, 36" length)	$3.98	1	$3.98
	Crown Bolt 1-1/2 in. x 48 in. Flat Bar 1/8 in. Thick Aluminum	$11.98	1	$11.98
	GORILLA GLUE 2 fl. oz. All-Purpose Adhesive	$4.97	1	$4.97
	100K-Ohm Linear-Taper Potentiometer	$2.99	5	$14.95
	Velcro - One-Wrap 30 Ft. x 1-1/2 in. Strap	$19.97	1	$19.97
Foot Sensors	Tekscan FlexiForce A201 sensors (8 pack)	$117.00	1	$117.00
	New Balance 509 SR (Mens: 9.5 and 10.5, Womens: 7 and 8)	Donated		$0.00
Data Logger	Pace Scientific XR440 Pocket Logger & accessories	$664.00	1	$664.00
Video Analysis	Athletic Tape (tracking markers)	$1.99	1	$1.99
Circuit	Miscellaneous Components (including voltage regulator, wire, resistors, op-amps)	$10.00	1	$10.00
			Total	$848.84

Appendix C: Consent Form

Informed Consent Agreement for Participation in a Research Study

Investigator: Benjamin Allen, Thomas Fontecchio, and Stacey Rauen

Contact Information: gait2010@wpi.edu

Title of Research Study: Gait Analysis and Spinal Rotation

Purpose of the study: To examine spinal rotation and impact forces on the feet, knees, and hips when running to determine if the amount of spinal rotation reduces injury.

Procedures to be followed:

Before starting the actual experiment, a number will be provided to conceal the identity of the subject. The subject will fill out a form that will provide important background information. After the paperwork filled out, test preparation can begin.

The experimenters will attach the spinal rotation device to the back of the subject using the two straps. The subject will also have a shoe insole with the force sensors attached inserted into the shoe that will be worn in the test. Once both the insole and the spinal rotation device are strapped into the shoe, the test is ready to begin

Each test will consist of the subjects running approximately 50 meters at a comfortable pace. The subject will start from a marked line and timed as they cross the finishing point. The data will be recorded for the entirety of the 50 meters. The subject will run this three separate times; the first time they will run this normally, the second time they will run with more exaggerated spinal rotation, and, finally, they will run with inhibited spinal rotation Each test will measure the speed, impact forces, and spinal rotation of the subject.

Spinal rotation will be measured using a potentiometer device that attaches by straps to the back. The spinal device will be measured wirelessly, allowing for no dangerous wires to be attached to the device.

Video analysis will also be used to measure the speed of each subject as well as to help determine the angle of the joints for further calculations. For measurement purposes, dots will be attached to each subject at the center of mass locations of their limbs.

Risks to study participants: Normal discomfort induced by running.

Benefits to research participants and others: Their participation will help explain the effect that spinal rotation has on the forces on the feet, knees, and hips as well as the correlation of spinal rotation, impact forces, and injuries.

Record keeping and confidentiality: All recorded data from these tests will be maintained by the members of the MQP group until the conclusion of the project. At the conclusion of the project, all data will be transferred to Professor Savilonis. No data allowing for personal identification of the subjects will be required.

For more information about this research, the rights of research participants, and/or in case of research-related injury, please contact Benjamin Allen, Thomas Fontecchio, or Stacey Rauen at gait2010@wpi.edu.

If additional assistance is required, please contact IRB Chair (Professor Kent Rissmiller, Tel. 508-831-5019, Email: kjr@wpi.edu) and the University Compliance Officer (Michael J. Curley, Tel. 508-831-6919, Email: mjcurley@wpi.edu).

Your participation in this research is voluntary. Your refusal to participate will not result in any penalty to you or any loss of benefits to which you may otherwise be entitled. You may decide to stop participating in the research at any time without penalty or loss of other benefits. The project investigators retain the right to cancel or postpone the experimental procedures at any time they see fit.

By signing below, you acknowledge that you have been informed about and consent to be a participant in the study described above. Make sure that your questions are answered to your satisfaction before signing. You are entitled to retain a copy of this consent agreement.

_____   Date: _____

Study Participant Signature

_____

Study Participant Name (Please print)

_____   Date: _____

Signature of Person who explained this study

Appendix D: Questionnaire

Subject number:	
Weight:	
Height:	
Length of lower leg:	
Length of upper leg:	
Years of running experience:	
Average weekly mileage:	
Injury history:	
Do you typically experience pain while running? Explain.	
Are you currently injured or experiencing pain? Explain.	
Do you wear orthotics/braces/tape?	
Brand and model of running shoe currently wearing?	