BJS - AT13

Calcaneal Tendon Force Comparison Between Traditional and Minimalist Shoes

A Major Qualifying Report: Submitted to the Faculty of the Worcester Polytechnic Institute in partial fulfillment of the requirements for the Degree of Bachelor of Science by

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1) Gait analysis
2) Achilles tendon
3) Load analysis

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During data collection and analysis, Ms. Cicchetti was responsible for collection of data from the AMTI AccuSway force plate. Mr. Andruskiewicz supervised the placement of markers and the collection of data from the Casio high speed camera. Mr. Liguori was responsible for analyzing the data in MATLAB and running statistical analyses. Mr. McLaughlin conducted the tracking of all markers via Tracker.

ABSTRACT

Calcaneal tendon force while jogging was estimated using an inverse dynamics model. Subjects ran in traditional running shoes, minimalist running shoes, and barefoot. Traditional shoes had a heel-toe drop of greater than 10 millimeters, while minimalist shoes were less than five millimeters. Repeatability testing indicated accuracy to within 2.6 times subject body weight. Average maximum estimated calcaneal tendon force was 9.5 ± 2.8 for barefoot trials, 9.9 ± 3.8 for minimalist trials, and 8.2 ± 2.2 for traditional shod trials. Forces were normalized to subject body mass and are unit less. Analysis of variance (ANOVA) indicated no statistical significance between trial types ($p>.05$, $n=14$). Results indicate further study of transition from traditional to minimalist shoes.

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EXECUTIVE SUMMARY

Minimalist shoes have seen a spike in sales since the introduction of the Vibram FiveFingers in 2005 and were a $1.7 billion industry in 2010 (Billhartz-Gregorian, 2011). It is often debated whether these shoes reduce injuries or cause them. In a study on cross country running teams, 6.6% of injuries reported were directly to the calcaneal, or Achilles tendon (AT), which were commonly associated with the transition between traditional and minimalist shoes (Vita Medica, 2012). Supporters of minimalist shoes have claimed that injuries such as heel fractures are reduced due to a lower impact force between the foot and the ground (Crossman, 2011). Another study showed an increase in bone trauma in runners who switched from traditional to minimalist shoes, compared to a control group that maintained the use of traditional running shoes (Reynolds, 2013). It is unknown whether or not there is a difference in AT forces between shoe types, or if injuries are caused by another mechanism.

The goal of this project was to create a reproducible, non-invasive method of testing AT forces while running. Data for this study was gathered by a force platform, timing system, and high speed camera. A wooden walkway was designed so that the force plate was at the same level as the ground that the subjects would be running on, allowing the subjects to maintain a normal gait during the exercise. A high speed camera was placed 1.1 meters away, where it could view the subject from the knee down when crossing the force platform. The timing system consisted of two pairs of sensors placed 1.3 meters apart and arranged in a way that the subject would trigger them while running over the force plate. The subject was fitted with high contrast markers in key locations on the foot and shank. These were the two insertion points of the AT, the two insertion points of the tibialis anterior tendon, the center of rotation of the ankle, the ball of the foot, the center of mass of the shank, and the center of mass of the foot. The subjects were instructed to run through the experimental setup at a comfortable jogging pace. Three trials were recorded for each subject. The first was with traditional running shoes, the second with minimalist shoes, and the third was with no shoes. A successful run meant that all

equipment recorded data properly and the subject ran with a proper gait, which was judged qualitatively. Data for analysis was gathered for 14 different subjects. A reproducibility test was also performed by having one subject in traditional shoes run nine trials. This proved that the subjects could be compared and data acquisition for subjects could begin.

The force platform recorded landing forces and moments in three directions. The motion analysis software, Tracker, recorded the positions of all markers in pixels, which would later be converted to meters. Both the force platform and Tracker provided Excel spreadsheets which could be read by MATLAB. Using these spreadsheets, MATLAB code was used to analyze data. The output was the force in the Achilles tendon, force in the tibialis anterior tendon, and tibial bone force exerted on the ankle joint. Nine trials performed by one subject produced an average estimated AT force of 13.8 ± 2.6, which was within 20% variance. Average maximum estimated calcaneal tendon force for 14 subjects was 9.5 ± 2.8 for barefoot trials, 9.9 ± 3.8 for minimalist trials, and 8.2 ± 2.2 for traditional shod trials. Forces were normalized to subject body mass and are unit less.

A One-Way Repeated Measures ANOVA determined if there was a difference in AT force between shoe types. The maximum force exerted on the AT and the mean force exerted on the AT during contact with the ground were analyzed. The results of the statistical analysis indicated that the null hypothesis that the three shoe types yield the same AT forces can't be rejected. The maximum forces were compared ($p=.20$) along with the average values ($p=0.12$) for each shoe type. Furthermore, a Tukey's post-hoc test was conducted and revealed that there was not a significant ($p<0.05$) difference between any two of the three shoe types when looking at the means or the maximums.

Both qualitative and quantitative data illustrated that the procedure was repeatable, and that there was little or no difference between the AT forces related to different athletic footwear options. Potential sources of error include the accuracy of the 2-dimensional inverse dynamics model, the variation of shoe types considered to be minimalist or traditional, and variation in subject gait. This model

neglected lateral forces from pronation, which could change estimated AT force. Because the results of this project indicated little or no difference between shoe types, this may suggest that the method of transition between shoe types should be studied. In addition, in order to incorporate more variables and potentially acquire more accurate data, an inverse dynamics model including forces and moments up to the knee should be studied.

1 INTRODUCTION

Currently, there is a trend in the long distance running world. For various reasons, athletes are choosing to adopt the use of minimalist running shoes. These shoes are lightweight, which removes resistance from traditional running shoes. They were created with the notion of mimicking barefoot running, which is argued to be healthier for runners. Users suggest that minimalist shoes prevent injury, yet many people who wear them continue to encounter running-related injuries.

It is estimated that individuals who run regularly will sustain two injuries over the course of a year (Crossman, 2011). A recent study done in *Medicine & Science in Sports & Exercise* determined that 6.6% of running related injuries occurred directly to the subjects' Achilles tendons (Vita Medica, 2012). These injuries may not be specific to a particular running shoe or style, but the investigators chose to focus on Achilles injuries because of their predominance in running.

There are many methods of testing that can be done to determine forces in the Achilles tendon, or calcaneal tendon. Imaging techniques, like ultrasound, may be used to determine tendon length, cross-sectional area, and change in both properties over time. Ground reaction forces obtained by a force plate, along with high speed cameras, free body diagrams (FBDs), and inverse dynamics can estimate forces in the calcaneal tendon. The most accurate method of determining forces in the Achilles tendon is through invasive studies, which are beyond the scope of this project.

There is no non-invasive benchmark for determining Achilles tendon loading while running. In addition, no evidence is presented that suggests running with traditional running shoes is any different than minimalist shoes or no shoes with respect to the Achilles tendon. This project aims to fill this gap by establishing a respected procedure for testing tendon loading and determining if there is a difference between running footwear.

2 LITERATURE REVIEW

2.1 Gait Analysis

Gait analysis is the study of locomotion through the use of the human eye and various types of instrumentation (Clayton, 2001). This type of analysis aims at understanding simple body movements, mechanics, and muscle activity. It is commonly used to assess an individual's movement in order to improve posture, running motion, or other types of movement-related problems. More recently, gait analysis tools have been used to analyze such topics as the effectiveness of orthotics on gait or changes in gait due to fatigue.

In order to understand gait analysis, it is important to first understand the human gait cycle. This cycle is the repetitive pattern of walking or running which is comprised of two phases: the stance phase and the swing phase. In running, the stance phase refers to when the foot is in contact with the ground, while the swing phase refers to the rest of the cycle. During the stance phase, there are multiple ways in which an individual's foot can strike the ground. In a heel strike, the runner's heel will hit the ground first. In a midfoot strike, the runner's foot will first hit the ground between the heel and ball of the foot. In a forefoot strike, the runner's foot will first hit the ground on the ball of the foot (Boucher et al., 2012).

Though there are many instruments that can aid in gait analysis, the most primitive method involves only the human eye, a stopwatch, tape, and felt-tip pens. It is a qualitative method, but can provide insight into gait abnormalities. The felt-tip pens are taped on the back of a subject's shoes, so that just the tip reaches the floor when he or she is standing up. Four pieces of tape are placed on the floor to establish five and six meter increments, as shown in Figure 2-1 below.

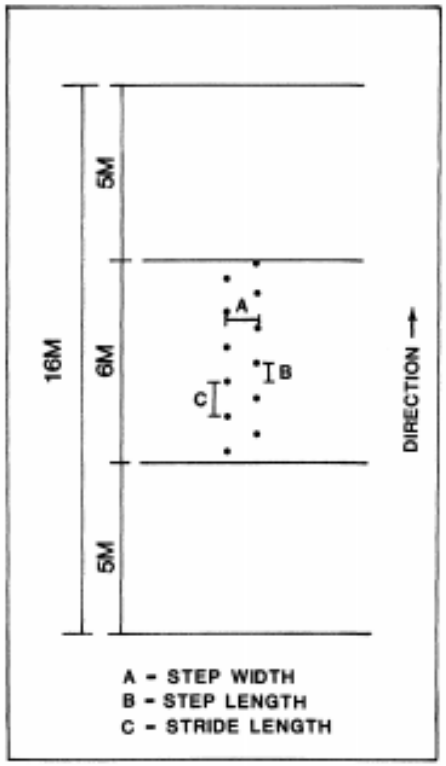


Figure 2-1: Floor layout of a primitive gait analysis method

The subject should walk normally across the floor, creating a series of dots indicating where he or she stepped. The test operator will be able to see, because of this series of dots, whether or not the subject has an uneven gait cycle. Beyond that, taking measurements between dots can allow the researcher to estimate properties such as stride length and step width, while a stopwatch can allow estimation of velocity (Cerny, 1983).

Currently, there are many advanced instruments that can aid in gait analysis. Methods such as force platforms, motion capture systems, electromyography (EMG), and portable sensors are used. Portable sensors include instrumentation such as strain gauges, accelerometers, electrogoniometers, gyroscopes, and in-sole pressure sensors. These instruments can be used to more accurately measure strain, acceleration, angles, angular momentum, and force.

Motion Capture

Video equipment can be used to capture motion in both two and three dimensions. Skin markers are placed on subjects so their position can be captured throughout the motion. In two dimensions, a camera must be placed perpendicular to the plane of interest, and in three dimensions, camera placement is not as important as long as each marker can be seen by two cameras at all times. Positioning of these surface markers can be a source of error, as skin can provide motion that is not entirely congruent with what the researcher is trying to analyze.

Similar to standard video equipment, optoelectronic systems also use marker placement, but these markers can come in both active and passive forms (Clayton, 2001). The active form of these markers emits a signal, while the passive form detects a signal from the subject of interest. These markers do not come without limitations. Use of these systems is generally limited because the markers require a hard-wired connection to the subject and may be highly dependent on lighting conditions. These systems also become less accurate as the number of markers increases (Higginson, 2009).

Force Platforms

A force platform is a plate that converts an applied force into an electrical signal, digital or analog, and then records it. Force plates may be capable of measuring forces as a function of time in three directions, as well as stance duration and centers of pressure. In gait analysis studies, force plates are often used in conjunction with video equipment. The size of many force platforms limits the region in which data can be collected. Subjects could be forced to change stride pattern, which would lead to inaccurate data.

In an attempt to improve the limitations of a force plate, force shoes were developed, allowing for the collection of data over a range of stance phases, on different surfaces, and from more than one limb in a given test (Bamberg, 2008). Some custom designed force shoes struggle to account for various

types of foot strike, potentially resulting in inaccurate data (Fontecchio, 2011). Recently, instrumented treadmills have been developed to surmount the limitations of the standard force platform. These systems allow an individual to run relatively naturally for an extended period of time, while still permitting a researcher to collect uncorrupted data. Even though this system overcomes the issues presented by standard force platforms and force shoes, gait can still be affected by running on a treadmill.

Electromyogram

EMG can measure the amount of muscle activity during gait. This can be done through either the attachment of surface or invasive electrodes. EMG can also be used to detect the amount of muscle fatigue a runner experiences throughout exercise. This technique is very sensitive and highly reproducible between individuals, but may limit the motion of the subject. Often, EMG signal is transmitted through a cable, but more recently, there have been systems developed that can wirelessly transmit signals (Higginson, 2009).

Portable Sensors

Portable sensors include accelerometers, electrogoniometers, gyroscopes, strain gauges, and in-sole pressure sensors. Accelerometers measure acceleration of the surface to which they are attached. Often, they are used on lower legs or feet to measure the acceleration during ground contact. Accelerometers reduce error generally associated with data transfer in motion capture systems. Another advantage of accelerometers is that they can continuously capture data throughout multiple gait cycles. However, skin mounted accelerometers may overestimate peak acceleration of body segments (Higginson, 2009).

Electrogoniometry is used to measure joint angle changes. This type of system has a potentiometer that measures the change in electrical resistance, which is attached to two rotating arms. The center is placed at the joint to effectively calculate the change in resistance, and in turn, the change

in the joint angle (Clayton, 2001). This type of system allows data to be captured and viewed continuously. One problem commonly associated with the use of electrogoniometers is that the actual sensor may move during testing, which can contribute to errors.

Gyroscopes can directly measure body segment angular velocity. This instrumentation is easy to attach and is insensitive to gravity. Gyroscopes are small and portable which make them an attractive method of gait analysis, but like accelerometers, can be interrupted by sensor motion throughout testing (Higginson, 2009).

Strain gauges change electrical resistance in response to an applied force. That electrical resistance is converted into voltage, which is proportional to strain. A strain gauge can be attached directly to the surface of measurement for accurate results. However, common problems in gait analysis occur when strain gauges are placed on the skin, which may cause unwanted motion.

In-shoe pressure sensors can provide a distribution of force over the surface of the foot, which is something that force platforms and shoes cannot do. They can also measure vertical forces throughout running or walking. Accuracy and precision of in-sole pressure sensors, however, may be dependent on the level and duration of pressure, as well as the age of the sensor (Higginson, 2009).

2.2 Anatomy and Physiology of the Achilles Tendon

The Achilles tendon (AT), or calcaneal tendon, is the largest and strongest tendon in the human body (Standring, 2009). Modeling the AT is important for research, but can be difficult due to variations in anatomy between individuals. The Achilles tendon complex is the main lever arm responsible for plantarflexion of the foot. The AT has insertion points at the calcaneus bone in the heel, the soleus muscle, and the gastrocnemius muscle, as shown in Figure 2. The average human AT is approximately 15 centimeters long and varies from nine to 26 centimeters (Standring, 2009). The length depends on factors such as height, heredity, and athletic ability of the subject. The AT inserts to muscle fibers on the soleus and gastrocnemius muscles, stretching almost to the plantar side of the calcaneus. The mean AT

width is 6.8 cm near the gastrocnemius muscle, decreases until just above the calcaneus, and increases to approximately 3.4 cm at its insertion point to the calcaneus (Nandra et al., 2011).

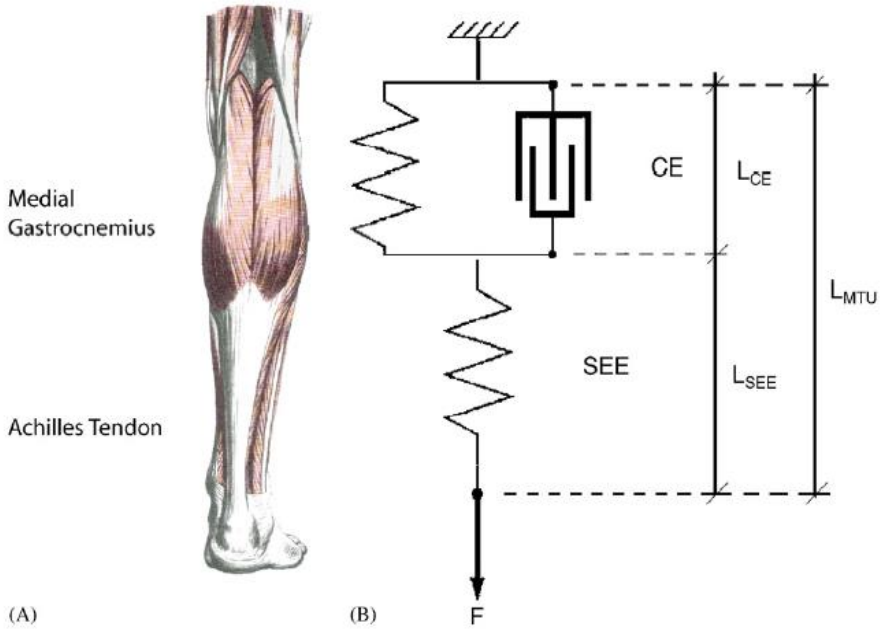


Figure 2-2: Anatomy of the Achilles tendon and the spring-dashpot model

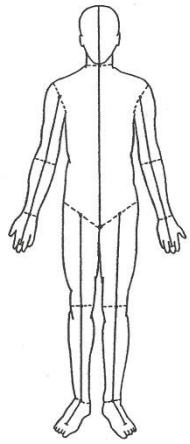
Figure 2-2 illustrates the anatomy of the Achilles tendon and one possible spring model for it. The spring-dashpot model presented by Lichtwark and Wilson represents the tendon as a simple spring, with the muscle having both spring and dashpot elements (2007). The active element in this model is the spring element. In this model, the gastrocnemius represents the contractile element (CE) of the muscle tendon unit (MTU) and the Achilles tendon represents the series elastic element (SEE). This figure models the Achilles tendon elastically, but the whole MTU as a more complicated system. The dashpot element resists a sudden force, thus protecting the MTU. Failure to decrease the dashpot's ability to resist motion by stretching prior to exercise can lead to rupture of the Achilles tendon.

The calcaneal tendon, like other tendons in the human body, is composed of more than 90% type I collagen. However, Nandra et al report that ruptured Achilles tendons form more type III collagen, which would change mechanical properties (2011).

The AT must withstand loads in excess of 9 kN in activities that require plantarflexion, such as running (Nandra et al., 2011). The collagen fibrils create a toe region in which the fibers straighten under small loads, which helps to reduce stress in the tendon. The tendon is also fan-shaped at its insertion on the calcaneus, which reduces stress concentrations at the enthesis, or tendon-bone insertion (Nandra et al., 2011). Litchtwark has modeled the Achilles tendon as purely elastic with an elastic modulus of between .67 and 1.07 GPa (2005). This model is not entirely accurate, as tendons are nonlinear viscoelastic and exhibit small amounts of hysteresis. Researchers model the Achilles tendon in different ways. While tendons are only slightly viscoelastic, a better mathematical model for the Achilles tendon would be Fung's model, as shown below. In Fung's model, A and B are constants that model stress as a function of strain in a nonlinear fashion.

$$\sigma=Ae^{(B(\epsilon-1))}$$

When modeling the Achilles tendon in a noninvasive study, one must consider the factors mentioned above that influence AT properties, including height and athletic ability. Chandler et al. have already calculated body segment lengths and center of mass data, as seen in Figure 2-3 (1975). This center of mass data helps researchers accurately identify the center of mass of body segments.



Segment	CG location (%)	Proximal end
Head	66.3	Top of the head
Trunk	52.2	Top of the neck
Upper arm	50.7	Shoulder
Forearm	41.7	Elbow
Hand	51.5	Wrist
Thigh	39.8	Hip
Leg (shank)	41.3	Knee
Foot	40.0	Ankle

Figure D.1 The body segment organization used by Chandler et al. (1975).

Regression Equations Estimating Body Segment Weights and Locations of Center of Gravity

Segment	Weight (N)	CG location (%)
Head	0.032 BW + 18.70	66.3
Trunk	0.532 BW − 6.93	52.2
Upper arm	0.022 BW + 4.76	50.7
Forearm	0.013 BW + 2.41	41.7
Hand	0.005 BW + 0.75	51.5
Thigh	0.127 BW − 14.82	39.8
Shank	0.044 BW − 1.75	41.3
Foot	0.009 BW + 2.48	40.0

Note. See Appendix D for the body segment organization used by Chandler et al. (1975). Body segment weights are expressed as a percentage of total-body weight (BW), and the segmental center of gravity (CG) locations are expressed as a percentage of segment length as measured from the proximal end of the segment.

Figure 2-3: Chandler's model

One main reason that Achilles tendons fail is over use. A second reason for AT failure is a lack of blood supply to the whole tendon. Over use and lack of blood supply are common causes of rupture in active people, and may account for as many as 17% of all running injuries (van Mechelen, 1992). Overuse of the Achilles tendon often leads to inflammation and tendonitis, or rupture (Nandra et al., 2011). Another possible reason for Achilles tendon failure in active individuals is a change in running form or footwear type, such as a transition from traditional running shoes to minimalist shoes. If this transition causes too high of a load to repeatedly be placed on the tendon, injuries can occur.

2.2.1 Runners Transitioning Between Shoe Types

Minimalist shoes come in many different forms, but they all attempt to imitate running barefoot. Minimalist shoes have a few advantages over barefoot running, while keeping the supposed

benefits. These advantages can include resistance to abrasions from asphalt, protection from objects on the ground, increased traction, and, of course, faster times. According to Roger Robinson of Runner's World, barefoot and minimalist running became popular in professional circles around the 1960's (2011).

Beginning in 1985, when Nike introduced the sock racer shoe, and continuing with the development of Vibram FiveFingers shoes in 2005, throngs of runners have switched to a barefoot or minimalist style of running (The Economist, 2011). There is much debate among practitioners and podiatrists over the benefits of this transition. People who make the switch claim that it reduces injuries, such as heel fractures, from running with traditional running shoes (Crossman, 2011). The argument is that when a person changes from traditional shoes, foot strike tends to shift from the heel to the middle of the foot or to the forefoot and the strides are shortened. Advocates say this decreases the maximum impact force of the foot upon the ground, thus reducing injury. Barefoot runners also claim that running in traditional shoes weakens the muscles of the foot. Barefoot or minimalist running strengthens these muscles. Many podiatrists argue against these claims due to the lack of data and studies supporting them. Podiatrists actually warn about switching because of the injuries that barefoot and minimalist running can cause. Stress fractures in the metatarsals, Achilles tendon injuries, and muscle strains are common concerns. Barefoot running has additional problems over minimalist shoes, namely damaged soles and puncture wounds by rocks or glass. These additional problems suggest that there is a need for differentiation between barefoot and minimalist shoes, although no studies can currently support this claim.

In order to counteract the risk of muscle injury when transitioning, some professionals have established guidelines for transitioning. Secondary muscles in traditional running shoes must be strengthened, muscles and tissues overly supported by traditional shoes must be stretched, and the runner must understand the changes in sensory feedback that will come with the switch. Running during

the first couple of weeks should be done lightly. The runner should increase the percentage of running done in minimalist shoes by 10% to 20% per week (Collier, 2011). It is also recommended to not run for consecutive days in minimalist shoes for the first four weeks. Above all, the most important thing Collier suggests is to listen to the body's reaction to the change and allow time for it to adjust (2011).

2.3 Studies Examining the Force of the Achilles Tendon

Many studies have been performed to either estimate AT force from various activities such as hopping, jumping, running, and walking. These studies generally use some combination of the equipment listed in 2.1 Gait Analysis.

In 1993, Fukashiro et al examined the AT force during vertical jumps. Fukashiro et al compared estimated AT force, calculated from the ankle joint moment, and direct AT force, measured by an implanted force transducer. Differences between methods led to correlation coefficients of between .95 and .99 for data where the foot was in contact with the ground (1993). Therefore, the calculation of the ankle joint moment from this study can be applied to non-invasive studies.

Komi et al performed a study of the ankle joint stiffness from sprint running. In this study, researchers measured change in ankle joint moment divided by change in joint angle. Komi et al measured ankle joint stiffness using electromyograms and a force platform and found the ankle joint stiffness to be constant at approximately 7 $N*m*deg^{-1}$ for sprinting efforts of 70% to 100% (2002). Researchers concluded that joint stiffness during complicated motor tasks may be controlled by individual biomechanical properties (2002).

Waggoner investigated a system of equations to accurately estimate Achilles tendon force by using Newton's second law and its rotational equivalent (2011). Waggoner's system can be found in Appendix A. The system can be rearranged to output the Achilles tendon force if all other variables are defined. Waggoner only ran computer simulated tests, rather than testing on human subjects (2011).

2.3.1 Research Gap

Many studies have been done to determine forces in the human Achilles tendon, including *in vivo*, *ex vivo*, and *in vitro* studies. However, the most accurate *in vivo* results are from invasive methods. There are no conclusive studies that examine the forces in the AT in running motions with different types of shoes. The ability to quantify the effects of wearing different shoes on running form would settle the debate between those who prefer traditional running shoes and those who argue that barefoot or minimalist shoes are better.

3 PROJECT STRATEGY

3.1 Client Statement

Initial Client Statement

Create a procedure that can accurately and consistently monitor Achilles tendon forces during running motions. Measure those forces to determine if there is a difference between running with traditional shoes, minimalist shoes, and barefoot.

Revised Client Statement

Create an accurate testing procedure to estimate Achilles tendon loads during running motions. Determine those forces to compare a possible difference between loading from traditional training shoes, minimalist shoes, and no shoes.

The biggest change in the client statement was the change from the idea of measuring the Achilles tendon forces to simply estimating or determining the forces using instruments or inverse dynamics. Because this project is geared toward researchers and not runners, the emphasis is on repeatability and accuracy.

3.2 Objectives

In order to produce a procedure or device that will satisfy the client statement, it must meet certain objectives. To evaluate which of these objectives hold priority over the others, a pair wise

comparison chart was created in Table 3-1. Above all, the most important objective is safety. This means that if a device is used, it must not be invasive, and in the testing method used, the subject must not be strained beyond a comfortable level. The next important testing parameter was found to be reproducibility. In order to have reputable data, the results must have a certain degree of consistency. To accomplish this, subject variability must be taken into account along with a set procedure. Subject variability would include things such as the size and sex of the testing subjects, as well as running speed and types of shoes. Also considering reproducibility, multiple tests will be run to assure the procedure can be repeated. Maintaining gait integrity will assure that there are no obstructions on the runway, and that a subject will not have to break stride in order to acquire usable data. It may be necessary to run multiple trials in order to achieve this. The next most important aspect for testing would be accuracy. The major point of this project was that there is little quantifiable data regarding a difference between shod and barefoot running. This procedure will be designed so that others can use it to test their experiments, so it must be user-friendly and not overly complicated to perform. Ideally, the testing method will be portable, though that is not a highly ranked objective.

Table 3-1: Pairwise Comparison Chart

	Maintain Gait Integrity	Reproducible	Accurate	User-friendly	Portable	Safe	Totals
Maintain Gait Integrity		0	1	1	1	0	3
Reproducible	1		1	1	1	0	4
Accurate	0	0		1	1	0	2
User-friendly	0	0	0		1	0	1
Portable	0	0	0	0		0	0
Safe	1	1	1	1	1		5

Below, in Figure 3-1, is the objectives tree, which expands each main objective into sub-objectives.

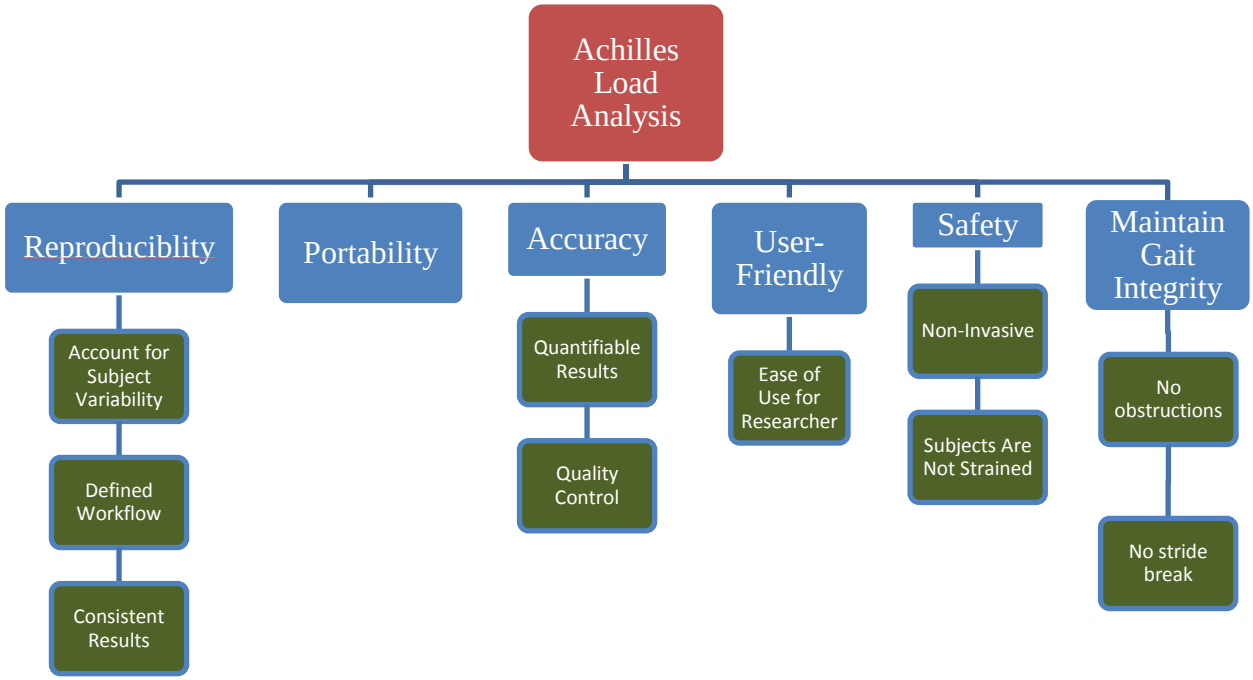


Figure 3-1: Achilles tendon load analysis objectives tree

3.3 Constraints

Constraints are limits placed on the design, which, if not met, would cause the design to be unsuccessful. The following is a list of constraints this project will face.

1. Budget ($600)
2. Time (approx. 6 months)
3. Safety
4. Institutional Review Board approval

5. Non-invasive procedure
6. Availability of subjects
7. Shoe size

The budget is $600, or $150 per team member, which has been allocated by the WPI Biomedical Engineering Department. All costs associated with parts and services must fit within that specified budget. The time limit is approximately 6 months.

The procedure must be both safe for the researcher and safe for the test subjects. It cannot pierce the skin of any subjects, or include any sharp objects that can harm either user. It must not cause any shock due to electrical activity or pinch the skin because of moving parts.

The IRB, or Institutional Review Board, monitors research and studies involving human subjects. Because this project will include the participation of human subjects, it is important that the IRB approves the testing procedure. This procedure must be non-invasive. Any devices used cannot pierce the skin of any test subjects, nor can they be inserted into the test subjects' bodies.

Another constraint is the availability of subjects. There is a limited supply of volunteers willing to take part in this study. In addition, not every shoe size will be available. Each test subject will need to take part in research done with three different footwear options (barefoot, minimalist shoe, and traditional running shoe). To acquire the most accurate data, each subject will need both minimalist and traditional shoes that fit him or her properly. Subjects may not have each type of shoe, and there may be limited availability of shoe sizes due to budget constraints.

The size of the devices used in this procedure can vary depending on which method is chosen. If the device is to be placed on the ground, it must be large enough so that a shoe may be placed upon it. If it is to be attached to the user, it must be small enough so that it does not impede the motion of the test subject.

3.4 Project Approach

In order to achieve the objectives of the project while staying within the constraints, certain technical design aspects will be required. As maintaining gait integrity and safety are the main objectives of the project, all equipment used must allow for both. For instance, one possible method of measuring the loads would be to measure ground forces exerted by the feet when running, then using anatomical free body diagrams to calculate the resultant forces required in the Achilles. Measuring said ground forces would likely require a subject to run across a force platform. With safety and gait integrity as priorities, there would need to be sufficient room for the subject to approach the force platform while gaining speed, and sufficient room to slow down following foot placement on the force platform. Also, the force platform would need to be on a level surface and flush with the running surface to ensure a normal gait, and a low chance of the subject tripping or striking the platform unnaturally. This force was measured in an MQP entitled "Gait Analysis and Spinal Rotation" to reach values just shy of 2.4 times the subjects' body weight (Boucher et al., 2012). For the sake of accuracy, any force platform used for the project would therefore need to sustain loads up to at least 2.5 times the body weight of the subjects. The force platforms available through the WPI Biomedical Engineering Department measure forces up to 400 pounds, which could equate to a 160 pound subject. In order to achieve diversity in the testing subjects, force platforms may need to be acquired from a different source or constructed by the project team.

Similar considerations will be taken for all possible testing procedures. As stated previously, all procedures will meet IRB approval and will maximize the level of safety for the subjects. Performance of the testing will meet the highest quality assurance, making sure all equipment used is rated for a broad range of possible testing outcomes, as outlined in the force platform example. In addition, gathered data will be statistically analyzed and outliers removed as to maintain a high level of research integrity.

4 ALTERNATIVE DESIGNS

4.1 Needs Analysis

There is no non-invasive "gold standard" technique for determining Achilles tendon loading while running. In addition, no evidence is presented that suggests running with traditional running shoes is any different than minimalist shoes or barefoot with respect to the forces in the Achilles tendon. This project aims to fill this gap by establishing a non-invasive, repeatable procedure for testing tendon loading and determining if there is a difference between running footwear.

4.2 Functions and Specifications

The final design for this project must have several functions and meet several specifications to be a successful design. The main functions of the final design are to determine Achilles tendon force, keep the user and the researcher safe, and identify variations in individual anatomy.

The final design must measure a variable that allows a researcher to determine the Achilles tendon force to within 10% repeatability (Hansen, 2006). Some sources estimate peak Achilles tendon force to be up to 12.5 times body weight, and local stresses up to 30 MPa (Spyrou, 2012). Every *in-vivo* study of Achilles tendon forces has led to only an estimation of the maximum force or stress because of necessary assumptions.

This design will have to make assumptions in order to reach any conclusions. If the design focuses solely on dynamics, the assumptions made will be that mass does not change with time and that markers placed on the center of mass of different body parts are accurate. A free body diagram centered approach would have to rely on the accuracy of Chandler's model and the locations of the center of mass and ankle joint (Chandler, 1975). If the design focuses on a stress and strain relationship, it may be necessary to assume that tendons are purely elastic, or that there is an equation for modeling viscoelastic stress-strain curves. A direct measurement approach would have to rely on the tested

accuracy of the transducer or ultrasound used, since there would be no way to truly verify these results. This type of approach would not be considered safe and therefore not feasible.

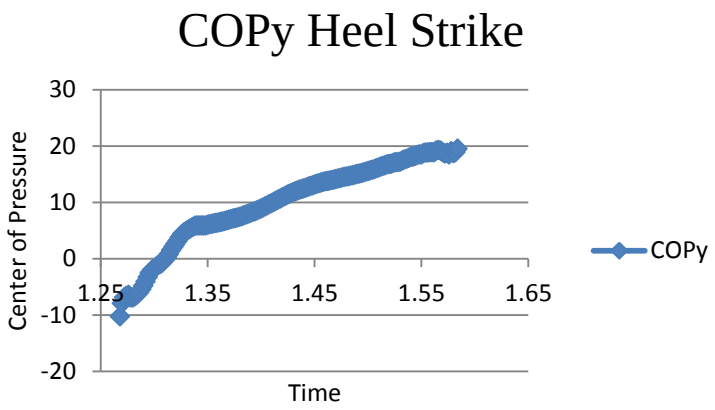
The final design must also allow a researcher to have an understanding of the variability in individual anatomy between test subjects. Specifically, the design must be able to identify individual insertion points on the calcaneus bone and gastrocnemius muscle and locations for stress concentration in the Achilles tendon (Spyrou, 2012). The design must allow comparison between subjects by normalizing body weight for all subjects analyzed.

4.3 Design Concepts

Four possible designs for the method of determining the Achilles tendon forces (ATF) were considered. Each of them utilizes different tools available to measure various spatiotemporal variables that relate to the ATF. Furthermore, each design is based on the implicit understanding that each subject will perform at least one repetition of the process wearing traditional running shoes, wearing minimalist shoes, and wearing no shoes. The first design relies primarily on the AMTI AccuSway force platform available through the WPI Biomedical Engineering Department. The second utilizes accelerometers on the foot. The third relies on a force platform measuring purely vertical forces. The fourth and final design involves Fung's model for viscoelastic material and a strain gauge. Each design requires various calculations that differ from design to design, but each can be used to determine ATF.

4.3.1 Design 1: COP Force Platform

During the stance phase of the gait cycle, the center of pressure (COP) shifts along the foot as a function of time. Data provided by Tom Fontecchio (Worcester Polytechnic Institute, Dept. of Biomedical Engineering) shows that in a subject who runs with a heel-strike, the COP begins at the heel and travels toward the toe as the stance phase continues. When a subject uses a mid-foot-strike, the COP begins at the middle of the foot and follows a much less linear shift as it traverses along the foot. This phenomenon can be observed in Figure 4-1, below.



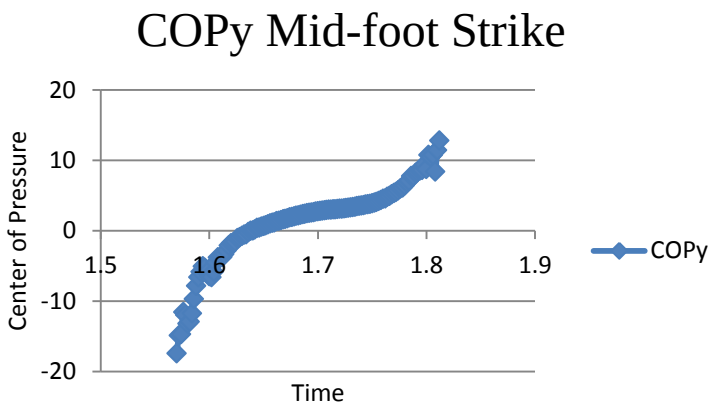


Figure 4-1: COP versus time graphs for heel-strike and mid-foot-strike runners

Since the COP varies during the stance phase, being able to measure its exact location could help determine the ATF. The AMTI AccuSway force platform, used to gather the data displayed in Figure 4-1, measures the COP along with the ground reaction forces (GRF) in both the vertical and horizontal directions. The GRF and distance from the COP to the ankle joint can be multiplied to determine the moment about the ankle. This moment is balanced almost entirely by the moment generated by the ATF, with the distance from the Achilles to the ankle serving as the moment arm. This length can be determined by using the ultrasound equipment available through the WPI Electronic and Computer Engineering Department. The length of the Achilles tendon would be defined as the distance between the location of the insertion point of the Achilles on the calcaneus and the muscle-tendon junction with

the gastrocnemius. Along with the angle between the foot and ground and the angle between the foot and shank, the moments and forces can be balanced to determine the ATF.

The first design for the testing method has three main components: (1) the COP from the AMTI AccuSway force platform, (2) the length of the Achilles measured with ultrasound, and (3) the necessary angles gathered from video analysis. The gathered data would then be supplied to MATLAB in order to compute the ATF using the balance of moment and balance of force equations described previously. A potential issue with this design is the capacity of the AMTI AccuSway force platform, which is only 400 lbs. As explained earlier, studies have shown GRF up to 2.5 times subject body weight, which would constrain subjects to less than 160 lbs.

4.3.2 Design 2: Accelerometer

The second design for a method to determine the ATF utilizes Newton's Second Law of Motion, which states that force in a direction equals mass of the object multiplied by acceleration of the object in the given direction. One accelerometer can be placed on the toe of the subject and another on the ankle joint. When combined with vertical motion data, this can describe the angular acceleration of the toe around the ankle joint. Using Newton's second law and the mass of the foot, the force applied by the Achilles on the ankle joint can be calculated. These calculations, while they appear simple, are quite complex and may be subject to outside interference. Multiple assumptions would need to be made, particularly the assumption that the ATF is the only force applied to the foot to generate the acceleration about the ankle.

4.3.3 Design 3: COP Estimation with Corn Starch

Due to weight constraints of the AMTI AccuSway force platform discussed in the first design an alternative to this platform is required. The Vernier force platform, available through the WPI Physics Department, measures only the vertical GRF. This design will utilize estimation of the COP with the understanding that while it may not be the most accurate way to determine the ATF, a similar degree of

error will exist across all subjects. Pressure sensitive films can generate a map of the COP when a subject steps on the film. These films, however, may cost up to $65 per linear foot (Sensorprod, 2012). Therefore, using a powdery substance such as cornstarch could provide a viable option. A thin layer of the powder, laid evenly across the force platform, would alter when stepped on, revealing the subjects' strike patterns. The COP at the landing and toe-off could then be estimated and used to estimate the COP throughout the stance phase. Once the COP estimations are made, the process would continue as described in Design 1.

4.3.4 Design 4: Strain Gauge

Design 4 does not require the use of a force platform. This design would utilize the stress-strain relationship of the Achilles tendon. Using a strain gauge and a viscoelastic material of similar mechanical properties to the Achilles, an estimation of the strain of the Achilles could be acquired. This estimation would be substituted into Fung's model for viscoelastic material, which states $\sigma = Ae^{B(\epsilon-1)}$, with published values for A and B for human tendon. The orientation of the strain gauge, shown in Figure 4-2, would require the material be attached to a cup on the heel of the subject and to a strap attached to the calf. The material would be approximately the same length as the Achilles tendon, and would undergo similar strains as the tendon throughout the gait cycle.

Figure 4-2: Sketch of strain gauge orientation

This design might interfere with the subject's natural gait cycle, though the material used and the way the material is strapped to the leg would take this into account. Further, if the strap or the heel-cup were to slip, the strain readings would no longer be accurate. In addition, an assumption would be made that all individuals have Achilles tendons with the same constants.

4.4 Conceptual Final Design

From the various designs that were conceptualized by the group, Design 1: COP Force Platform was chosen as the final design. An evaluation matrix as shown in Table 4-1 was used to quantitatively evaluate the final designs. Based on the objectives rankings, each of the six objectives was given a weight. Each alternative design was compared to each objective and given a rank of one through five, one meaning that the objective would barely be fulfilled, and five meaning that the design fulfilled the objective thoroughly.

Table 4-1: Evaluation Matrix

	COP Force Plate	**Accelerometer**	**Vertical Force Plate**	**Strain Gage**
Maintain Gait Integrity (x4)	5x4= **20**	5x4=**20**	5x4=**20**	4x4=**16**
Reproducible (x5)	4x5=**20**	3x5=**15**	4x5=**20**	4x5=**20**
Accurate (x3)	5x3=**15**	3x3=**9**	4x3=**12**	2x3=**6**
User-friendly (x2)	5x2=**10**	5x2=**10**	5x2=**10**	4x2=**8**
Portable (x1)	1x1=**1**	5x1=**5**	3x1=**3**	5x1=**5**
Safe (x6)	5x6=**30**	5x6=**30**	5x6=**30**	5x6=**30**
Totals	96	89	95	85

This design's key feature implemented a force plate that could measure both vertical and horizontal forces and center of pressure location. In order to gather the proper data, the force plate should be stepped on as naturally as possible during the subjects' running motions. To solve the issue of the runner hopping onto the elevated force plate, a platform will be constructed that is level with the plate. Ramps on either end are important for the runners' transition onto the platform. The subjects could skew data or injure themselves. The flat portion made for the actual running purposes should be

long enough to allow for the runner to reach the desired pace of 3.35 m/s, which would be approximately an 8 minute mile pace.

The point of focus of the project is to find the force generated in the Achilles tendon during a subject's run. To do this, several variables must be calculated along with the center of pressure and vertical force. A high speed camera with the use of markers is a suitable way to track the angles of the foot as they relate to the ground and shank. In order to place the markers on the leg and foot in the correct location, the insertion points of the Achilles tendon into the gastrocnemius muscle and into the calcaneus bone must be found. To accomplish this, an ultrasound was seen as the best method to locate these points. Along with finding these insertion points, the ultrasound device can also help determine the Achilles tendon length at rest. Using the free body diagram of the foot and shank, these variables can give the force generated in the Achilles tendon.

4.5 Feasibility Study and Experiments Methodology

To recruit test subjects for data collection, all undergraduates at WPI were emailed, thus allowing equal participation for any interested person. Data was collected for three run-throughs of each subject and 14 subjects total. Run-throughs were completed wearing traditional running shoes, minimalist shoes, and no shoes.

The measurement system included a Casio EX-ZR100 high speed camera, an AMTI AccuSway force plate, and a Brower TC timing system. Instead of a ramp, three particle board plates were positioned in front of the force plate and two boards were positioned after the force plate. These boards were included in the setup in order to allow continuity in gait of subjects.

The high speed camera was placed 1.1 meters (43 inches) from the center of the force plate. This distance was chosen to allow the subjects' legs below the knee joint to be within the field of view for the entire stride. The camera was positioned level with the force plate.

The Brower TC timing system was set up as shown in Figure 4-3. The two sets of sensors were placed 1.3 meters (53 inches) apart. One set of sensors was set up before the subject passed the force plate, and the other was set up after the force plate. The timing system was used to determine the time it took for each subject to pass through the sensors, which was used in determining the velocity of each subject.

Figure 4-3: Force platform and timing system experimental setup

Each subject was fitted with eight markers created from small neon stickers and tape, as shown in Figure 4-4. These markers were used to identify important physiological locations for analysis. Two markers identified were on the foot, which was treated as a rigid body. One marker was used for the center of mass of the foot and another was used for center of mass of the shank, as determined by Chandler's model (1975). Two markers represented the insertion points of the tibialis anterior tendon, and two more markers represented the insertion points of the calcaneal tendon. An ultrasound was not available, so determination of these points was done manually, by asking subjects to plantarflex.

Although markers were placed on the subject's skin, the data of Lichtwark and Wilson proved that the difference between elasticity of skin and tendon do not cause a significant change in marker location (2005). Complete descriptions of marker placement and determination for all three trial types can be found in Appendix B.

Figure 4-4: Marker placement

Once a subject was fitted with all appropriate markers, the force plate was zeroed. All equipment began recording data, with the high speed camera at 240 frames per second, and with the force plate at 1000 points per second. The subject was instructed to run through the experimental setup at a comfortable jogging pace. The subject was told not to alter his or her stride to land on the force plate and could take as many run-throughs as necessary. Run-throughs were recorded for each trial type until one successful stride was recorded. A successful stride was one in which the subject stepped fluidly onto the force plate in view of the camera without breaking stride, and all recording equipment worked properly. A normal gait was determined qualitatively by the investigators and subject.

Acquiring data from force platform

To acquire data from the AccuSway force platform in a fashion that would be readable by MATLAB, it was necessary to open the data in the BioAnalysis software. BioAnalysis opened by clicking 'save' and 'analyze' in the NetForce software. In BioAnalysis, clicking 'view… data,' showed a window with the data acquired by the force platform. The data had seven columns; time (sec), Fx (lbs), Fy (lbs), Fz (lbs), Mx (in-lbs), My (in-lbs), and Mz (in-lbs). In order to be able to use the data, it was copy and pasted into an Excel file, and saved as a .csv file.

Acquiring data from Tracker

To examine multiple points of the shank and foot, the video files were placed into Tracker, a video analysis and modeling tool. The proper frame range in which the subject passed the camera was isolated.

The parameter used for this analysis was a point mass. Point mass allowed a tracker to be placed onto the markers. One point mass was created for every marker tracked. Once the runner's foot was a few frames from contact, a frame was found that clearly showed all markers with little distortion. Clicking on the desired point mass in the track control window, then holding control and shift and clicking on the marker, created an initial template. The program analyzed the video and created a point on the marker at every frame. This output data to the bottom right table and the top right graph, as seen in Figure 4-5. The x and y coordinates were in pixels with the origin located at the center of the video. This step was repeated for each marker. The program looks for RGB differences in the pixels to find the best match to the template and marks a point at this spot. The dashed rectangle surrounding the circle is the area the program searches when applying a point. This can also be resized and moved for the best results. The template changes over time to adapt to shape and color changes of the marker.

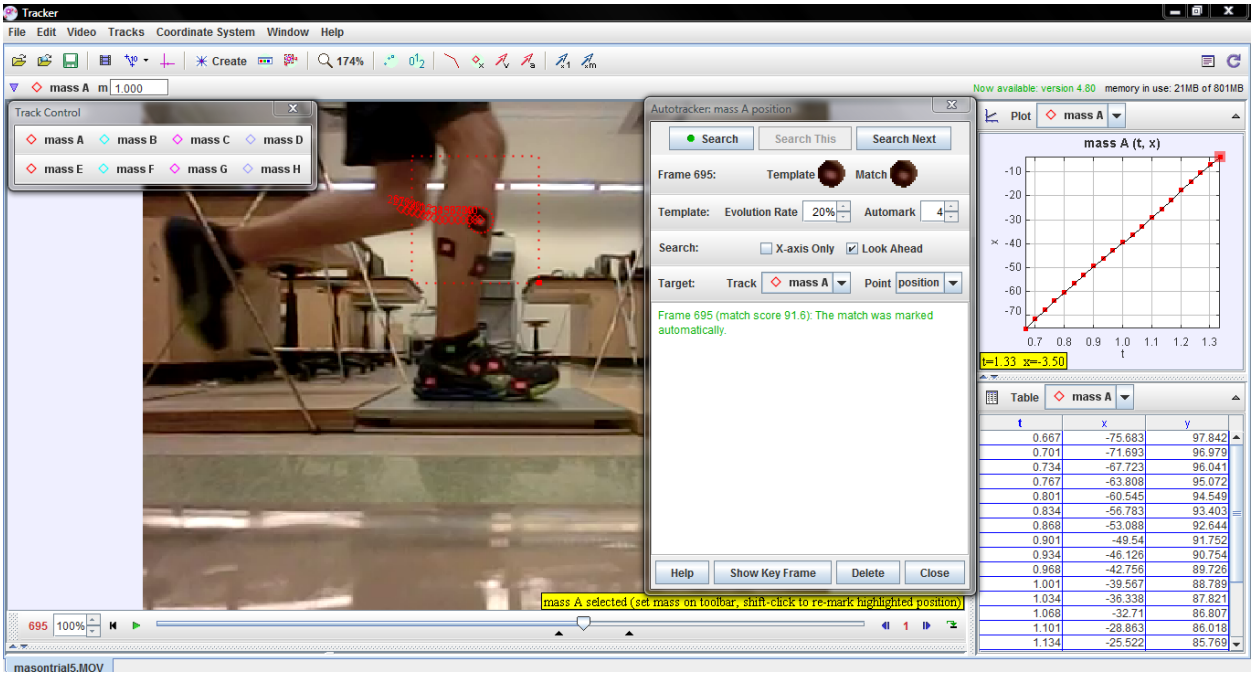


Figure 4-5: Screenshot of the Tracker program used

When tracking, if the program made tracking errors, those points were manually fixed by clicking and dragging to the desired location. Once the desired data was obtained, it was exported by highlighting the data in the table for each point mass and copy and pasting it into a spreadsheet.

Data Analysis Methods

Below in Figure 4-6 is the free body diagram used in the analysis. In order to use this diagram and solve the system of equations, there were some assumptions made.

a. The study is 2-Dimensional

b. The foot is a rigid body

c. The foot can be modeled as a right-angled wedge with moment of inertia

$$I = \frac{W}{18}(A^2 + H^2)$$

Where W is the weight of the foot, A is the distance from heel to toe, and H is the height from ground to ankle (Myers, 1962).

d. F_{tib} acts at the junction of the phalanges and metatarsals

e. Force at the center of rotation of the ankle is solely due to the force of the tibia

f. Weight of a subject's shoe does not significantly change center of mass or weight of the foot

g. The AT is subject to all force generated by the calf muscles

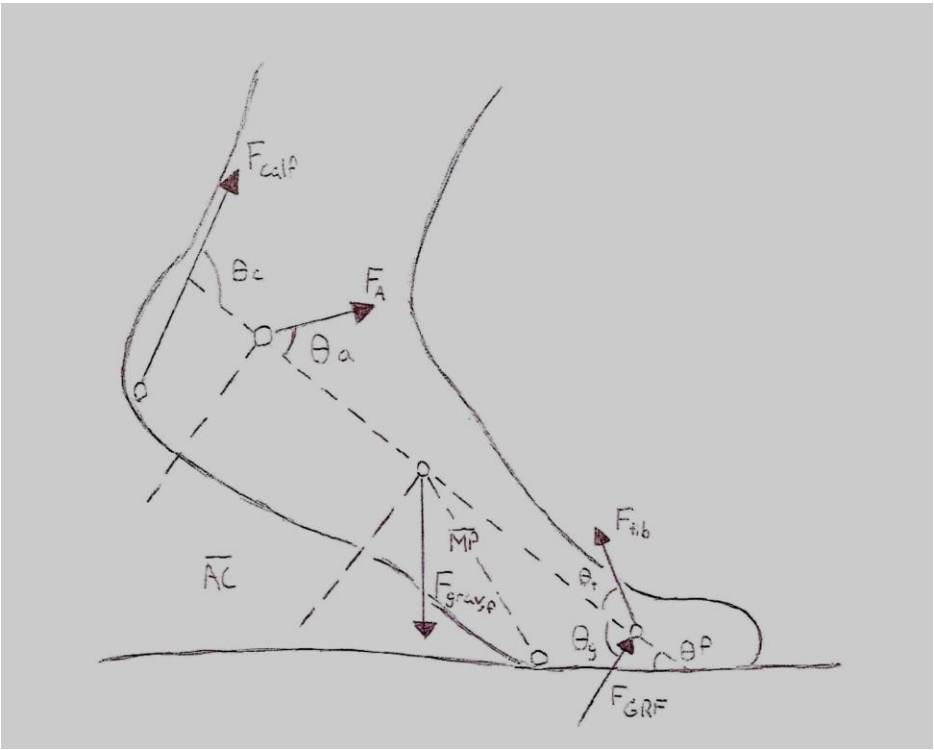


Figure 4-6: Free body diagram of the foot

Setting up a system of equations to find the force in the Achilles tendon was heavily modeled on the work done by Waggoner (2011). The system of equations is:

$$\begin{cases} \sum F_{f,x} = F_{calf,x} + F_{tib,x} + F_{ankle,x} + F_{GRF,x} = m_f a_{f,x} \\ \sum F_{f,y} = F_{calf,y} + F_{tib,y} + F_{ankle,y} + F_{GRF,y} + F_{Grav,f} = m_f a_{f,y} \\ \sum M_{f,ankle} = F_{calf}(\overline{CA})\sin(\theta_c) + F_{tib}(\overline{TA})\sin(\theta_t) + F_{grav,f}(\overline{AC})\sin(\theta_a) + F_{GRF}(\overline{MP})\sin(\theta_g) \\ = I\alpha_{f,com} \end{cases}$$

CA represents the distance from the insertion point of the AT on the calcaneus to the center of rotation of the ankle. TA is the distance from the insertion point of the tibialis anterior tendon on the foot to the center of rotation of the ankle. AC measures the distance from the center of mass of the foot to the center of rotation of the ankle. MP is the distance from the center of pressure of the foot to the center of rotation of the ankle.

F_{calf} is the force generated by the calf muscles that goes through the AT, F_{tib} is the force generated by the tibialis anterior muscle, F_{grav} is the force of gravity which is constant, F_{GRF} is the ground reaction force as recorded by the force platform, and F_{ankle} is the force present at the center of rotation of the ankle. The angles are as follows:

$\theta_c = angle\ between\ F_{calf}\ and\ line\ of\ action\ of\ foot$

$\theta_t = angle\ between\ F_{tib}\ and\ line\ of\ action\ of\ foot$

$\theta_a = angle\ between\ F_{grav}\ and\ line\ of\ action\ of\ foot$

$\theta_g = angle\ between\ F_{GRF}\ and\ line\ of\ action\ of\ foot$

$\theta_f = angle\ between\ foot\ and\ shank$

This system of equations served as the basis for the reverse dynamics used to determine the Achilles tendon forces. Data was processed using a pair of MATLAB files, which can be found in Appendix C.

MATLAB Data Processing

Two MATLAB codes were written for the data analysis, one to extract information regarding each subject from a spreadsheet. This data includes the subject's body weight, foot length, ankle height, toe to ankle length, heel to ankle length, and subject number. Also included is important information regarding the scale of the video data in terms of pixels per meter, and the distance from the center of the video frame to the horizontal center of the top of the force plate. These two measurements were acquired using ImageJ, the process for which will be discussed in the following section. The same code also generated the output file, a spreadsheet containing the maximum Achilles tendon forces for each subject and each shoe type, found in Appendix D.

The majority of the analysis took place in the MATLAB function written for the project. It used the data from the first script to determine where the data for the current subject was stored, then accessed that directory. Then, using a for loop, it looped over the data for each of the shoe types for the subject in question. Using frame numbers provided in the first script, the data from the Tracker software was extracted from an Excel spreadsheet and adjusted from pixels to meters using the previously discussed scaling factor. The locations of the eight markers in the horizontal and vertical directions were each acquired as an Nx1 matrix, where N equals the number of frames during which the foot was in contact with the force platform. Furthermore, the vertical values were adjusted such that the top of the force platform had a vertical value of zero. The time from the video data was also acquired. The camera samples at a rate of 240 frames per second, so each frame was approximately 0.0041 s. The time was adjusted such that the time at point of contact became zero.

A similar process was used to extract the data from the force platform, which provided an output of forces and moments in three directions. The important measurements for the purpose of this study were the x-direction and y-direction ground reaction forces, and the moment about the z-axis. The contact and lift-off times were determined by recognizing when there was a change in the y-direction

readings, and eliminating the time points that did not have change of at least one Newton and time points that reported a negative y-direction force. These time points were eliminated from all three of the important platform outputs. Also eliminated were the corresponding time recorded by the platform, which were set so contact occurred at a time of zero seconds. The force platform samples at a rate of 100 samples per second, which was substantially less than the camera. The data from the camera had to be fit to the different time profile. This was achieved by applying a third degree spline-fit to the position data acquired by Tracker, then evaluating the spline function at the time points used by the force platform, essentially down sampling the video results to match the time points and matrix length of the platform data. This data was used to determine the center of pressure, which was taken to be the moment about the z-axis divided by the force in the y-direction. The final center of pressure reading was set equal to the length of the foot, and all others derived from that.

As discussed previously, and seen in the system of equations, the angles between various parts of the foot and shank played a major role in determining the Achilles tendon force. Most angles were first determined in relation to the x-axis by finding the two markers that would create the line in question. Taking the arctangent of the vertical distance between them divided by the horizontal distance yielded the angular measurement. The angle of the foot relative to the ground was a helpful angle because the ground was the x-axis, so the angle did not need to be adjusted. The angles between the foot and shank were determined by how they were geometrically related. The angle of the total ground reaction force was determined by taking the arctangent of the vertical force divided by the horizontal force. It must be noted that this trigonometry was performed for each time point, and resulted in additional Nx1 matrices as discussed previously.

The final inputs to the system that required calculations were the accelerations. To determine these, a seventh order polynomial was fit to the position data for the point in question, which also included the angle of the foot. Fitting was performed using a Least Squares Approximation. A second

derivative was taken of the resulting polynomial, and the time values applied to generate Nx1 matrices of the accelerations and angular accelerations required.

With all the measurable quantities calculated, the values were applied to the system of equations, which was solved for the Achilles tendon force, anterior tibialis force, and force in the ankle joint. Plots were generated and the outputs from the function were used to generate the final output used for statistical analysis.

ImageJ Data Processing

ImageJ is a free software program downloadable from the internet and generated by the National Institutes of Health. The advantage of using ImageJ was that the scale can be manually set based on the distance separating objects within the picture. The user simply draws a line connecting the two objects, which are a known distance apart, then sets the scale so that the number of pixels traversed by the line equals the set distance. Once the scale is set, the distance of any other pair of points can be easily determined by drawing a similar line and the software will measure the distance and apply the scale. An important note is that the pixel sizes of the JPEGs used in ImageJ were a tenth the size of those in the video data analyzed by Tracker. Therefore, the pixel data provided by the Tracker output needed to be divided by one tenth of the ImageJ scale to arrive at the final distance values used. One valuable measurement was the offset, which was determined by measuring the distance from the middle of the frame to the middle of the top of the force platform, such that the top of the force platform became the horizontal axis when the values were adjusted.

4.6 Summary

The testing and analysis procedure can be summed up into a few key points, listed below:

1. Set up the board plates, force platform, high speed camera, and timing system in the proper locations
2. Locate and mark all eight key points on a test subject's lower leg

3. Perform all three subject trials
    a. The force platform will output an Excel file
4. Input the video into Tracker, locate all eight markers, and track them through the program
    a. Tracker will output an Excel file
5. Input the Tracker and force platform spreadsheets into MATLAB
6. Run a statistical analysis on the results

5 RESULTS

A consistency test was conducted, in which one subject performed nine shod trials with the intent of demonstrating repeatability from multiple trials. Data was analyzed as a normalization of force divided by body weight. Figure 5-1 shows the normalized data for the entire stance phase of the gait cycle. The average maximum value for all nine trials was 13.8±2.6 N/BW.

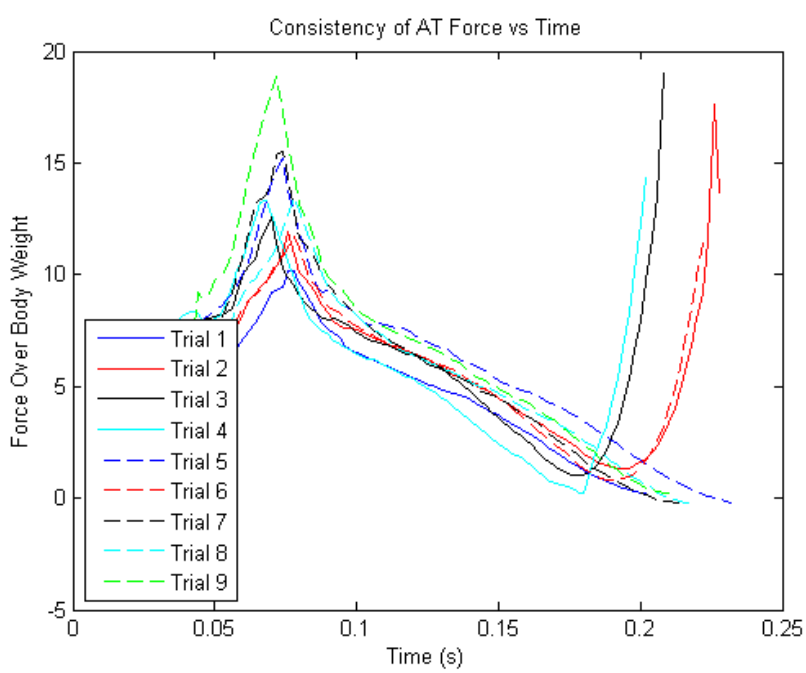


Figure 5-1: Plot of single subject running nine shod trials

The remaining subjects performed a single trial with each of the three shoe types: barefoot, minimalist, and shod. Below, in Figure 5-2, is a representative plot of the results of a single subject. Again, the data is represented as Force/Body Weight as a function of time. All fourteen individual subject plots can be found in Appendix E.

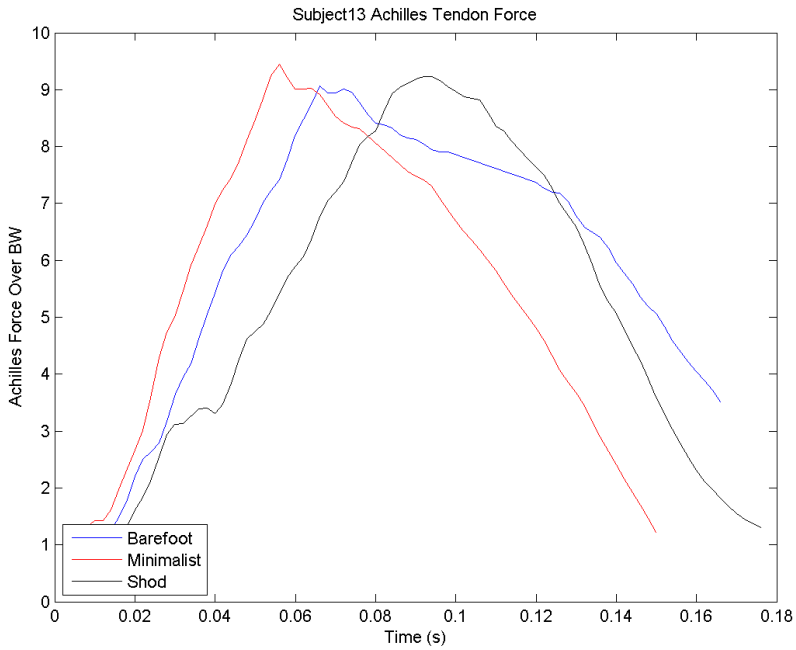


Figure 5-2: Individual plot of all three shoe type trials for subject 13

Two parameters were chosen for analysis. The first was the maximum force exhibited by the Achilles tendon during the stance phase, and the other was the mean of the Achilles tendon forces exhibited during this time. Box and whisker plots of these values for all fourteen subjects can be seen in Figure 5-3, below.

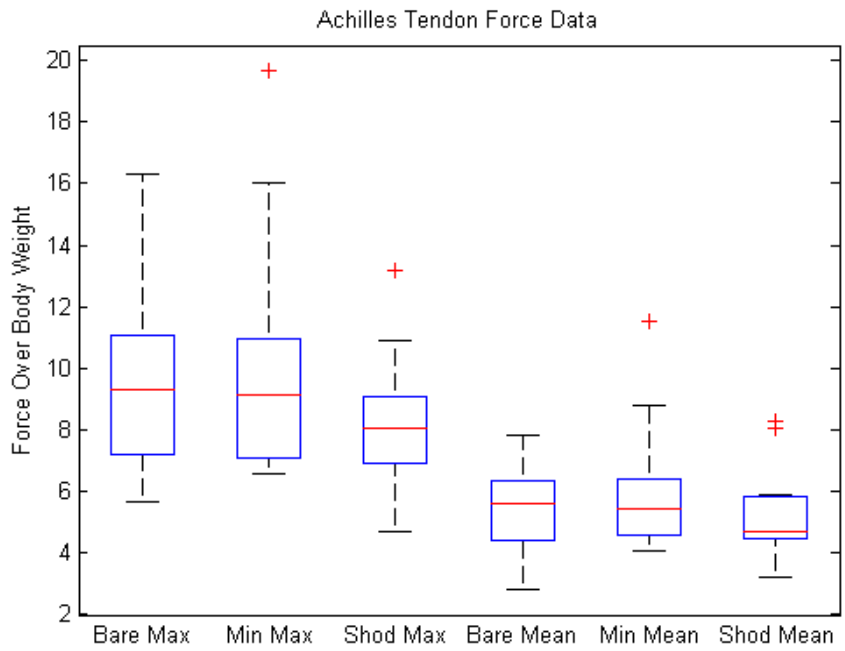


Figure 5-3: Box and whisker plots of the maximum and mean values for all 14 subjects arranged by shoe type

Both quantitative and qualitative analyses were performed Table 5-1, shown below, displays the average values for 14 subjects and each shoe type for the three forces calculated: Achilles tendon, tibialis anterior, and bone force on the ankle. All values are normalized.

Table 5-1: Average values for all shoe types for all calculated forces

Force (N/BW, n=14)	Barefoot	Minimalist	Shod
Average max AT force	9.5 ± 2.8	9.9 ± 3.8	8.2± 2.2
Average mean AT force	5.4 ± 1.3	6.0± 2.0	5.1 ± 1.5
Average max compressive tibialis force	-3.9 ± 1.3	-3.8 ± 1.9	-3.1 ± 0.77
Average max compressive bone force	-8.0 ± 1.8	-8.7 ± 2.1	-7.8 ± 1.7

In order to determine if there was any statistical difference between the shoe types for each of the calculated forces, One-Way Repeated Measures Analysis of Variance (ANOVAs) were performed on each of the data sets described in Table 5-1. These can be found in Table 5-2 through Table 5-5. Note that the null hypothesis is that the means of the three data sets analyzed for each ANOVA are equal, so a p-value greater than 0.05 indicates that there is not a statistical difference between the data sets.

Table 5-2: Results of One-Way Repeated Measures ANOVA for maximum Achilles tendon forces

Source	df	SS	MS	F	P
Treatment	2	22.4425	11.221	1.71	0.20
Error	26	170.5684	6.599		
Subjects	13	179.4273			
Total	41	372.4382			

Table 5-3: Results of One-Way Repeated Measures ANOVA for mean Achilles tendon forces over the stance phase

Source	df	SS	MS	F	P
Treatment	2	5.3999	2.7	2.29	0.12
Error	26	30.7209	1.182		
Subjects	13	73.4634			
Total	41	109.5473			

Table 5-4: Results of One-Way Repeated Measures ANOVA for maximum anterior tibialis forces

Source	df	SS	MS	F	P
Treatment	2	4.9391	2.47	1.54	0.23
Error	26	41.6842	1.603		
Subjects	13	24.5457			
Total	41	81.169			

Table 5-5: Results of One-Way Repeated Measures ANOVA for maximum bone forces upon the ankle

Source	df	SS	MS	F	P
Treatment	2	6.4389	3.219	1.54	0.23
Error	26	54.3992	2.092		
Subjects	13	81.2842			
Total	41	142.1224			

A Tukey's post-hoc pairwise comparison test indicated that in all four above cases, no combination of any two shoe types displayed significantly different results.

6 ANALYSIS

The results of the reproducibility test shown in Figure 5-1 show that the shape and values of the AT forces exhibited by the subject during the nine shod trials are fairly similar. This test indicated that a subject will display consistent results when running, which allowed for each subject to only run one trial per shoe type. This expedited the testing procedure, and minimized the amount of running required by the participants. Furthermore, it helps to validate the results. The average of the maximum values displayed in the consistency test was 13.7±2.6. The standard deviation of 2.6 was similar to the deviation of 2.8 shown in the maximum estimated Achilles tendon force from the overall shod testing.

Analyzing the data required looking at both the maximum AT force calculated for each trial and the mean value of the forces calculated for each trial over the stance phase of the gait. The maximum values were explored to determine if altering shoe type alters the load applied to the AT, with the understanding that frequent exposures to high forces would lead to injury. The mean values were analyzed to explore if there is a difference in the forces sustained by the AT throughout the stance phase, which would indicate possibility of overuse injury.

In looking at box and whisker plots, found in Figure 5-3, of the max forces and mean forces, there is a clear similarity in the distribution of the forces across the shoe types. While the shod data appears to be slightly lower than the minimalist and barefoot data, the medians in both cases still fall well within the second quartile of the other two. This is further shown in Table 5-1, which shows that the average of the shod data for both the mean and maximum forces is within half of a single standard deviation of the averages for the other shoe types. This table also includes data for the calculated forces in the tibialis anterior tendon and the force of the tibia on the ankle. These values are appropriately negative, indicating that they are both showing compression to oppose the tensile forces displayed by

the AT. These values do make sense, given the orientation of the system of equations designed to estimate the forces.

These similarities were further confirmed by performing a One-Way Repeated Measures ANOVA on the maximum and mean values across the three shoe types for the fourteen subjects tested. In general, an ANOVA is used to determine if there is a significant variation in the averages values of each shoe type. The repeated measures ANOVA takes into account that the measurements were made on the same subject under different conditions. The null hypothesis of the ANOVA is that the three data sets are statistically similar. The test statistic for an ANOVA is the f-statistic, which is taken to be the ratio of the mean square of the values within the groups to the mean square of the values between the groups. The f-statistic can then be translated to a p-value based on the number of degrees of freedom. If the p-value is less than 0.05, then the difference between the groups is taken to be statistically significant and the null hypothesis is rejected.

Four ANOVAs were performed on the data collected for this study, the results of which are shown in Table 5-2 through Table 5-5. The four data sets analyzed were the maximum AT values, mean AT values over time, maximum anterior tibialis force, and maximum ankle bone force. In all four of these cases, the p-value was greater than 0.05, indicating that the shoe type did not alter the forces in any of the analyzed locations. A Tukey's post-hoc pairwise comparison test was used to further analyze the data. This test compares every combination of two treatment groups. This test was redundant, since if any one of the treatment groups had significantly different values, this would have become apparent in the ANOVA results. Nonetheless, it upheld the findings of the ANOVAs, that there were no significant differences between the force values across shoe type.

The final product for this project was a testing and analysis procedure using instrumentation commonly available to scientists and engineers. As such, there are no foreseeable environmental, political, or ethical ramifications.

Results of this study indicate that there is no significant difference in Achilles tendon forces as they relate to various types of running footwear options. Due to this, it is possible that an increasing number of runners may purchase minimalist shoes. This could impact those companies that manufacture and sell various types of minimalist shoes, as they may experience a noticeable increase in sales. On the other hand, if runners do not see a benefit in using minimalist shoes, they may choose to not purchase them, and the opposite effect could take place.

This study did not have any effect on the safety of those who participated. If future studies similar to this one are conducted, it may be possible that injury could occur. Runners who are inexperienced in minimalist or barefoot running have a higher risk of injury. This study did not produce a specific reason for Achilles tendon injury in new minimalist shoe users. This study also did not produce any recommendations for runners who decide to transition. Therefore, this study does not directly affect runners at risk for injury.

Indirectly, this study has the potential to influence runners. There is great opportunity for the results of this project to reach a large number of individuals via online blogs and forums where runners can communicate about current events. If the results of this study were to reach one of these forums, there is potential that numerous runners may transition between shoe types.

The analysis software programs MATLAB, Tracker, and Excel are also easily attainable by any individual. The testing and analysis procedures are easily reproduced, so there are no anticipated negative effects with respect to manufacturability. Similarly, sustainability of the procedure designed for this study is simple. The materials used were not dependent on any conditions beside the body weight

of the test subject. This entire procedure could be easily replicated, modified, or improved in many biomechanics labs.

7 CONCLUSIONS AND FUTURE RECOMMENDATIONS

7.1 Conclusions

Many runners claim that the transition to minimalist shoes reduces injury, such as heel fractures (Crossman, 2011). Many users have experienced AT injuries during this transition period. Little is known about the actual forces in the AT while running. The current gold standard for testing AT force is an invasive method. The investigators interviewed several runners who have recently transitioned from traditional running shoes to minimalist shoes. The investigators determined that Achilles tendonitis and soreness was a negative side effect of the transition. Therefore, the group hypothesized that force in the Achilles tendon is greatly decreased in traditional running shoes.

The results of the repeatability testing validated the non-invasive testing procedure used. Both quantitative and qualitative analysis proved that there was no statistical difference between estimated Achilles tendon forces relative to various footwear options. This was valid for both maximum forces and average forces over the stance phase of the gait cycle. The p-value from the ANOVA on 14 subjects' data showed that the study could not reject the null hypothesis. However, the p-value was still low. This indicates that further improvements to this procedure could produce the hypothesized result.

These results indicate that further tests and studies should be done. A closer look at the transition to minimalist shoes should be taken in order to determine if that is where the cause of injuries lies. Additionally, the limitations of this study did not allow it to be done in 3-dimensions which would provide for accurate data.

7.2 Future Recommendations

There are several possibilities for sources of error that can be examined and further research that can be done to further understand the differing forces seen in the AT with the use of different shoe types. Some of the potential errors that could have occurred might be seen in the accuracy of the 2-dimensional inverse dynamics model of the foot, the accuracy of the Tracker program, changes in the subjects' gaits during running trials, improper MATLAB coding, and the variation of shoes used. Further research should be done by looking into other ways of modeling the AT forces during running motions, such as calculating lateral forces from pronation and including variables up to the knee. According to the results, there is no significant change in AT forces among the different shoe types. This translates to the reports about AT damage with minimalist shoe use may be attributed to other factors besides the AT forces taken into account during this study.

The accuracy of the 2-dimensional model would have one of the largest effects on the results of the study. Some computational factors could be overlooked, as the ankle is a complex system of bones, tendons, and muscles. Missing any of these could result in significant changes to the data. The coding in MATLAB for this complex system could have possible errors as well, due to the large amount of variables involved. However, one of the majoring limitations of the model is that it is only in two dimensions and not three. The results of this study state that the different forces along the AT when wearing different shoes were not statistically relevant. Lateral forces could have significantly different forces when the subject uses different types of footwear, as it is clearly evident in reviewing the trial videos that subjects do vary in the location of where their foot will strike the ground when wearing different shoes. The forces on the AT could be loaded in a different way than it is accustomed to due to the variation in lateral pronation. This unfamiliar loading would mean that a certain degree of exertion could cause the reported pains and injuries seen with minimalist shoes. To accomplish this type of study, not only would

the model have to be updated, but additional cameras would have to be used to fully analyze the foot's three dimensional properties while running.

Some data could lack some precision as well, due to inconsistencies in testing and analysis. Shoe types were classified on their heel to toe drop, a less than 5 millimeter drop for minimalist and greater than 10 millimeter drop for traditional running shoes. Every subject did not run in the same type of shoes for their traditional and minimalist trials. Since different companies design their shoes in different ways, subjects with different shoes considered to be in a certain classification of shoe type could possibly have enough variation to alter the resulting outputs from the run. Purchasing traditional and minimalist shoes within a certain size range could remove this variation, but it lessens the amount and variation of subjects that can be recruited. Another source of variation error could have occurred with the motion tracking program. Occasionally Tracker cannot follow a marker through the entire contact phase of the video, so the points being marked must be placed manually. Another source of error in the tracking program is the pixel area that the program searches for in the video changes over time. This change can sometimes move the area where the points are being applied and thus cause an improper pixel location in the output. The final variation is seen in the gait of the subjects during their run. People have different ways of running, and this difference could change the force that the plate reports. Gait is not entirely consistent among subjects and thus could have some effect on data. A subject's gait could also be altered during the running motion due to the limited lab space after the force plate. Subjects have been seen to turn after hitting the force plate in an attempt to slow down before running into a wall. Much of the variations could be compensated for with a larger subject population. This could then have the possibility of leading to a different statistical outcome.

Future studies may wish to look into other factors involved with injuries to the AT while wearing minimalist shoes. One method would be to include more variables in the leg to improve the accuracy of the results. Instead of only modeling the foot and ankle region, the forces and moments up to the knee

could also be examined. Another factor to take into account is the effect that transitioning between shoe types has on the AT. Some studies have shown that this transition period is when people are most susceptible to lower leg injury (Vita Medica, 2012). Finding the exact reason as to why this transition is harmful would be an important study in understanding the role of minimalist shoes in AT injuries.

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APPENDIX A: Equations Used by Waggoner, 2011

$$
\begin{aligned}
m\vec{a}_f &= \vec{F}_{calf} + \vec{F}_{tib} + \vec{F}_{ankle} + \vec{F}_{gravity} + \vec{F}_G \quad , \\
m\vec{a}_s &= -\vec{F}_{calf} - \vec{F}_{tib} - \vec{F}_{ankle} + \vec{F}_{gravity} - \vec{F}_{knee} - \vec{F}_{quad} \quad , \\
m\vec{a}_{qb} &= \vec{F}_{gravity} + \vec{F}_{knee} + \vec{F}_{quad} \quad , \\
\vec{I}_f\alpha_f &= \frac{-\vec{l}_f}{2} \times \vec{F}_{calf} + \frac{\vec{l}_f}{2} \times \vec{F}_{tib} + \left(\alpha_h - \frac{1}{2}\right)\vec{l}_f \times \vec{F}_{ankle} + \vec{T}_G \quad , \\
\vec{I}_s\alpha_s &= \frac{-\vec{l}_s}{2} \times \vec{F}_{calf} + \left(\frac{1}{2} - \alpha_a\right)\vec{l}_s \times \vec{F}_{tib} \\
&\quad + \frac{-\vec{l}_s}{2} \times \vec{F}_{knee} + \left(\frac{1}{2} - \alpha_{qs}\right)\vec{l}_s \times \vec{F}_{quad} + \frac{\vec{l}_s}{2} \times \vec{F}_{ankle} \quad , \\
\vec{I}_{qb}\alpha_{qb} &= (-\alpha_b)\,\vec{l}_b \times \vec{F}_{knee} + (\alpha_{qb} - \alpha_b)\,\vec{l}_b \times \vec{F}_{quad} \quad .
\end{aligned}
$$

APPENDIX B: Locations of all Markers Used in Running Trials

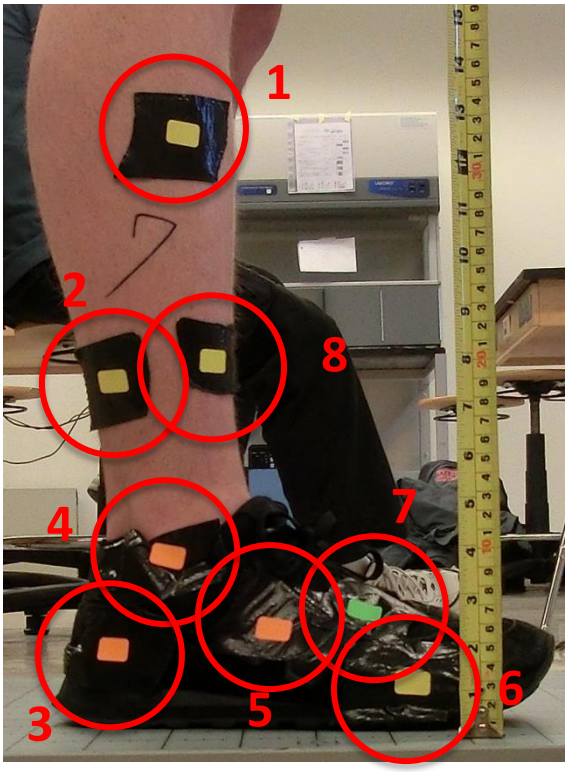

Figure B-1: Marker location on a single subject

1- Center of mass of the shank
2- Junction of the calcaneal tendon and the gastrocnemius muscle (determined during maximum voluntary plantarflexion)
3- Insertion point of the calcaneal tendon on the calcaneus bone
4- Center of rotation of the ankle (the lateral malleolus)
5- Center of mass of the foot
6- Junction between phalanges and metatarsals
7- Insertion point of the tibialis anterior tendon on the metatarsals
8- Junction of the tibialis anterior tendon and the tibialis anterior muscle

APPENDIX C: MATLAB Coding

The following MATLAB function is designed to analyze the measured values associated with AT forces. It can be adjusted to analyze AT forces or the various other forces looked at for this project.

```
function [max_bare meanbare max_min meanmin max_shod meanshod] =
CurrentSystem(Subject,BW,imageJ,contacton,contactoff,lfoot,hankle,achilles2an
kle,toe2ankle,yoffset,xoffset,plotflag)
%function [min_tibbare mean_tibbare std_tibbare min_anklebare mean_anklebare
std_anklebare min_tibmin mean_tibmin std_tibmin min_anklemin mean_anklemin
std_anklemin min_tibshod mean_tibshod std_tibshod min_ankleshod
mean_ankleshod std_ankleshod] =
CurrentSystem(Subject,BW,imageJ,contacton,contactoff,lfoot,hankle,achilles2an
kle,toe2ankle,yoffset,xoffset,plotflag)


indir = ['\\filer\home\.nthome\desktop\MQP\' Subject '\'];

 xlsx = dir([indir '*.xlsx']);
 csv = dir([indir '*.csv']);

for k = 1:3
%% Extract Data from Video

ForceData = fullfile(indir,csv(k).name);
VideoData = fullfile(indir,xlsx(k).name);

start = num2str(contacton(k)); finish = num2str(contactoff(k));

time = xlsread(VideoData,1,['B' start ':B' finish]);
shankCOMx = (xlsread(VideoData,1,['C' start ':C' finish])-xoffset)./imageJ;
shankCOMy = (xlsread(VideoData,1,['D' start ':D' finish])+yoffset)./imageJ;
calfx = (xlsread(VideoData,1,['G' start ':G' finish])-xoffset)./imageJ;
calfy = (xlsread(VideoData,1,['H' start ':H' finish])+yoffset)./imageJ;
anklex = (xlsread(VideoData,1,['K' start ':K' finish])-xoffset)./imageJ;
ankley = (xlsread(VideoData,1,['L' start ':L' finish])+yoffset)./imageJ;
heelx = (xlsread(VideoData,1,['O' start ':O' finish])-xoffset)./imageJ;
heely = (xlsread(VideoData,1,['P' start ':P' finish])+yoffset)./imageJ;
footCOMx = (xlsread(VideoData,1,['S' start ':S' finish])-xoffset)./imageJ;
footCOMy = (xlsread(VideoData,1,['T' start ':T' finish])+yoffset)./imageJ;
toex = (xlsread(VideoData,1,['W' start ':W' finish])-xoffset)./imageJ;
toey = (xlsread(VideoData,1,['X' start ':X' finish])+yoffset)./imageJ;
tibx = (xlsread(VideoData,1,['AA' start ':AA' finish])-xoffset)./imageJ;
tiby = (xlsread(VideoData,1,['AB' start ':AB' finish])+yoffset)./imageJ;
tibshankx = (xlsread(VideoData,1,['AE' start ':AE' finish])-xoffset)./imageJ;
tibshanky = (xlsread(VideoData,1,['AF' start ':AF' finish])+yoffset)./imageJ;
COM2anklex = abs(anklex(1)-footCOMx(1));

if strcmp(Subject,'Subject1') == 1
    tibshankx = tibx;
    tibshanky = tiby;
    tibx = toex;
    tiby = toey;
```

```
elseif strcmp(Subject,'Subject2') == 1 ||strcmp(Subject,'Subject3') == 1
||strcmp(Subject,'Subject4') == 1 ||strcmp(Subject,'Subject5') == 1
||strcmp(Subject,'Subject6') == 1
    tibshankx = shankCOMx;
    tibshanky = shankCOMy;
end


time = (time-min(time))/10;
%% Extract Force Data and set to time from video

GRFx = csvread(ForceData,1,2,[1 2 1000 2]); %N
GRFy = csvread(ForceData,1,3,[1 3 1000 3]);  %N
Momz = csvread(ForceData,1,5,[1 5 1000 5]); %N-m
timef = csvread(ForceData,1,0,[1 0 1000 0]);

for i = 1:length(GRFy)-1
    if abs(GRFy(i)-GRFy(i+1)) < 1 || GRFy(i) < 0
        GRFy(i) = 0;
        GRFx(i) = 0;
        Momz(i) = 0;
        timef(i) = 0;
    end
end

GRFx(GRFy == 0) = [];
timef(GRFy == 0) = [];
Momz(GRFy == 0) = [];
GRFy(GRFy == 0) = [];

GRFx = GRFx*(-4.44822162); %N
GRFy = GRFy*(4.44822162);  %N
Momz = Momz*(-4.44822162*.0254); %N-m

timef = timef-min(timef);

GRFx(timef > time(length(time))) = [];
GRFy(timef > time(length(time))) = [];
Momz(timef > time(length(time))) = [];
timef(timef > time(length(time))) = [];

GRF = sqrt(GRFx.^2 + GRFy.^2);



%% Calculate accelerations and alpha

% Acceleration of the foot in x-dir
P = polyfit(time,footCOMx,7);

Footposx = zeros(size(timef));
for i = 1:length(Footposx)
    Footposx(i)= P(1)*timef(i)^7 + P(2)*timef(i)^6 + P(3)*timef(i)^5 +
P(4)*timef(i)^4 + P(5)*timef(i)^3 + P(6)*timef(i)^2 + P(7)*timef(i) + P(8);
```

```
end

afx = zeros(size(timef));
for i = 1:length(afx)
    afx(i) = 42*P(1)*timef(i)^5 + 30*P(2)*timef(i)^4 + 20*P(3)*timef(i)^3 +
12*P(4)*timef(i)^2 + 6*P(5)*timef(i) + 2*P(6);
end

%Acceleration of foot in y-dir
P = polyfit(time,footCOMy,7);

footposy = zeros(size(timef));
for i = 1:length(footposy)
    footposy(i)= P(1)*timef(i)^7 + P(2)*timef(i)^6 + P(3)*timef(i)^5 +
P(4)*timef(i)^4 + P(5)*timef(i)^3 + P(6)*timef(i)^2 + P(7)*timef(i) + P(8);
end

afy = zeros(size(timef));
for i = 1:length(afy)
    afy(i) = 42*P(1)*timef(i)^5 + 30*P(2)*timef(i)^4 + 20*P(3)*timef(i)^3 +
12*P(4)*timef(i)^2 + 6*P(5)*timef(i) + 2*P(6);
end

rads = atan((ankley-toey)./(anklex-toex))+pi;

P = polyfit(time,rads,7);

angle = zeros(size(timef));
for i = 1:length(angle)
    angle(i)= P(1)*timef(i)^7 + P(2)*timef(i)^6 + P(3)*timef(i)^5 +
P(4)*timef(i)^4 + P(5)*timef(i)^3 + P(6)*timef(i)^2 + P(7)*timef(i) + P(8);
end

deltaang = zeros(size(timef));
for i = 1:length(deltaang)
    deltaang(i) = 7*P(1)*timef(i)^6 + 6*P(2)*timef(i)^5 + 5*P(3)*timef(i)^4 +
4*P(4)*timef(i)^3 + 3*P(5)*timef(i)^2 + 2*P(6)*timef(i) + P(7);
end

alphaf = zeros(size(timef));
for i = 1:length(alphaf)
    alphaf(i) = 42*P(1)*timef(i)^5 + 30*P(2)*timef(i)^4 + 20*P(3)*timef(i)^3
+ 12*P(4)*timef(i)^2 + 6*P(5)*timef(i) + 2*P(6);
end

footCOMx = spline(time,footCOMx); footCOMx = ppval(footCOMx,timef);
footCOMy = spline(time,footCOMy); footCOMy = ppval(footCOMy,timef);
shankCOMx = spline(time,shankCOMx); shankCOMx = ppval(shankCOMx,timef);
shankCOMy = spline(time,shankCOMy); shankCOMy = ppval(shankCOMy,timef);
calfx = spline(time,calfx); calfx = ppval(calfx,timef);
calfy = spline(time,calfy); calfy = ppval(calfy,timef);
anklex = spline(time,anklex); anklex = ppval(anklex,timef);
ankley = spline(time,ankley); ankley = ppval(ankley,timef);
heelx = spline(time,heelx); heelx = ppval(heelx,timef);
heely = spline(time,heely); heely = ppval(heely,timef);
```

```
tibx = spline(time,tibx); tibx = ppval(tibx,timef);
tiby = spline(time,tiby); tiby = ppval(tiby,timef);
toex = spline(time,toex); toex = ppval(toex,timef);
toey = spline(time,toey); toey = ppval(toey,timef);
tibshankx = spline(time,tibshankx); tibshankx = ppval(tibshankx,timef);
tibshanky = spline(time,tibshanky); tibshanky = ppval(tibshanky,timef);
thetaf = spline(time,rads); thetaf = ppval(thetaf,timef);


COPx = Momz./GRFy;
COP2ankle = sqrt((ankley).^2+(COPx-anklex).^2);

%% Mass of foot/shank and I

mf = (0.009*BW+2.45)/9.81;
I = mf*9.81/18*(hankle^2 + lfoot^2);

%% Determine angles

thetaground = -abs(atan(GRFy./GRFx));

thetagrf = thetaf-thetaground;

thetacalf = abs(atan((calfy-heely)./(calfx-heelx)));
thetac = thetacalf - thetaf + pi;
thetashank = abs(atan((shankCOMy-ankley)./(shankCOMx-anklex)));
thetatibialis =  atan((tibshanky-tiby)./(tibshankx-tibx));

for i = 1:length(thetatibialis)
    if thetatibialis(i) > 0
        thetatibialis(i) = -thetatibialis(i);
    end
    thetatibialis(i) = thetatibialis(i)+pi;
end

thetat = thetaf-thetatibialis;
thetafootshank = thetashank - thetaf + pi;
%% Set up system

fachilles = zeros(size(GRFx));
ftibialis = zeros(size(GRFx));
fankle = zeros(size(GRFx));


    ialpha = zeros(size(GRFx));
    GRFmom = zeros(size(GRFx));
    springmom = zeros(size(GRFx));
    gravmom = zeros(size(GRFx));
    Momr = zeros(size(GRFx));

for j = 1:length(fachilles)

    ialpha(j) = I*alphaf(j);
    GRFmom(j) = GRF(j)*COP2ankle(j)*sin(thetagrf(j));
```

```
    springmom(j) = 401.070457*thetafootshank(j);
    gravmom(j) = mf*9.81*COM2anklex;

left = [cos(thetacalf(j)) cos(thetatibialis(j)) cos(thetashank(j));
        sin(thetacalf(j))  sin(thetatibialis(j)) sin(thetashank(j));
        achilles2ankle.*sin(thetac(j)) toe2ankle*sin(thetat(j)) 0];

right = [mf*afx(j)-GRFx(j);
         mf*afy(j)-GRFy(j)-mf*9.81;
         ialpha(j)-GRFmom(j)-gravmom(j)];

solution = left\right;

fachilles(j) = solution(1);
ftibialis(j) = solution(2);
fankle(j) = solution(3);


Momr(j) = ialpha(j)-gravmom(j)-GRFmom(j);

end


if k == 1
    bare = fachilles(4:length(fachilles)-3);
    tibbare = ftibialis(4:length(ftibialis)-3)/BW;
    anklebare = fankle(4:length(fankle)-3)/BW;
    min_tibbare = min(tibbare);
    min_anklebare = min(anklebare);
    mean_tibbare = mean(tibbare);
    mean_anklebare = mean(anklebare);
    std_tibbare = std(tibbare);
    std_anklebare = std(anklebare);
    bare = bare/BW;
    meanbare = mean(bare);
    stdbare = std(bare);
    timebare = timef(4:length(timef)-3);
    max_bare = max(bare);
    Mombare = Momr;
    barealpha = max(alphaf);
    baregrf = max(GRFy);
elseif k == 2
    minimalist = fachilles(4:length(fachilles)-3);
    tibmin = ftibialis(4:length(ftibialis)-3)/BW;
    anklemin = fankle(4:length(fankle)-3)/BW;
    min_tibmin = min(tibmin);
    min_anklemin = min(anklemin);
    mean_tibmin = mean(tibmin);
    mean_anklemin = mean(anklemin);
    std_tibmin = std(tibmin);
    std_anklemin = std(anklemin);
    minimalist = minimalist/BW;
    meanmin = mean(minimalist);
    stdmin = std(minimalist);
    timemin = timef(4:length(timef)-3);
```

```
    max_min = max(minimalist);
    Mommin = Momr;
    minalpha = max(alphaf);
    mingrf = max(GRFy);
elseif k == 3
    shod = fachilles(4:length(fachilles)-3);
    tibshod = ftibialis(4:length(ftibialis)-3)/BW;
    ankleshod = fankle(4:length(fankle)-3)/BW;
    min_tibshod = min(tibshod);
    min_ankleshod = min(ankleshod);
    mean_tibshod = mean(tibshod);
    mean_ankleshod = mean(ankleshod);
    std_tibshod = std(tibshod);
    std_ankleshod = std(ankleshod);
    shod = shod/BW;
    meanshod= mean(shod);
    stdshod = std(shod);
    timeshod = timef(4:length(timef)-3);
    max_shod = max(shod);
    Momshod = Momr;
    shodalpha = max(alphaf);
    shodgrf = max(GRFy);
end


end

if plotflag == 1
a = figure('Visible','on');
else
a = figure('Visible','off');
end

plot(timebare,bare,'b',timemin,minimalist,'r',timeshod,shod,'k')
legend('Barefoot','Minimalist','Shod','Location','Southwest')
title([Subject ' Achilles Tendon Force'])
xlabel('Time (s)')
ylabel('Achilles Force Over BW')
saveas(a,[indir 'AchillesForce.png'],'png')

end
```

The following script is used in conjunction with the function above to perform the specified analysis on all fourteen subjects.

```
clear all; close all; clc

plotflag = 0;
subjectinfo = csvread('subjectinfo14subs.csv',1,0,[1 0 14 8]);

% Input where contact and liftoff occur in video data sheets
touch = [9 11 11;
         6 7 6;
         6 15 17;
         7 9 6;
         9 7 10;
         8 10 7;
         11 9 7;
         7 9 9;
         8 8 9;
         10 7 8;
         6 6 9;
         9 9 11;
         6 8 7;
         8 7 7];
lift = [63 65 68;
        66 61 68;
        61 72 74;
        78 75 71;
        61 63 66;
        57 66 66;
        65 57 64;
        75 77 81;
        75 77 78;
        72 74 76;
        58 53 64;
        71 73 73;
        73 74 77;
        71 70 83];


min_tibbare = zeros(size(touch,1),1);
mean_tibbare = zeros(size(min_tibbare));
std_tibbare = zeros(size(min_tibbare));
min_anklebare = zeros(size(min_tibbare));
mean_anklebare = zeros(size(min_tibbare));
std_anklebare = zeros(size(min_tibbare));
min_tibmin = zeros(size(min_tibbare));
mean_tibmin = zeros(size(min_tibbare));
std_tibmin = zeros(size(min_tibbare));
min_anklemin = zeros(size(min_tibbare));
mean_anklemin = zeros(size(min_tibbare));
std_anklemin = zeros(size(min_tibbare));
min_tibshod = zeros(size(min_tibbare));
mean_tibshod = zeros(size(min_tibbare));
std_tibshod = zeros(size(min_tibbare));
min_ankleshod = zeros(size(min_tibbare));
```

```
mean_ankleshod = zeros(size(min_tibbare));
std_ankleshod = zeros(size(min_tibbare));
max_bare = zeros(size(min_tibbare));
max_min = zeros(size(min_tibbare));
max_shod = zeros(size(min_tibbare));
meanbare = zeros(size(min_tibbare));
meanshod = zeros(size(min_tibbare));
meanmin = zeros(size(min_tibbare));


for i = 1:size(touch,1)

    % Extract subject info from .csv
    subnum = num2str(subjectinfo(i,1));
    BW = subjectinfo(i,2);
    imageJ = subjectinfo(i,3);
    lfoot = subjectinfo(i,4);
    hankle = subjectinfo(i,5);
    achilles2ankle = subjectinfo(i,6);
    toe2ankle = subjectinfo(i,7);
    yoffset = subjectinfo(i,8);
    xoffset = subjectinfo(i,9);

    contacton = touch(i,:);
    contactoff = lift(i,:);


    Subject = ['Subject' subnum];

    sprintf('Current subject number %d of %d',i,size(subjectinfo,1))


   %[min_tibbare(i) mean_tibbare(i) std_tibbare(i) min_anklebare(i)
mean_anklebare(i) std_anklebare(i) min_tibmin(i) mean_tibmin(i) std_tibmin(i)
min_anklemin(i) mean_anklemin(i) std_anklemin(i) min_tibshod(i)
mean_tibshod(i) std_tibshod(i) min_ankleshod(i) mean_ankleshod(i)
std_ankleshod(i)]=
CurrentSystem(Subject,BW,imageJ,contacton,contactoff,lfoot,hankle,achilles2an
kle,toe2ankle,yoffset,xoffset,plotflag);
   [max_bare(i) meanbare(i) max_min(i) meanmin(i) max_shod(i) meanshod(i)] =
CurrentSystem(Subject,BW,imageJ,contacton,contactoff,lfoot,hankle,achilles2an
kle,toe2ankle,yoffset,xoffset,plotflag);

end

% space = zeros(size(min_tibbare));
%
% % Output max achilles forces to csv
%  allforces = [min_tibbare min_tibmin min_tibshod space mean_tibbare
std_tibbare mean_tibmin std_tibmin mean_tibshod std_tibshod space space
min_anklebare min_anklemin min_ankleshod space mean_anklebare std_anklebare
mean_anklemin std_anklemin mean_ankleshod std_ankleshod];
% csvwrite('boneandankle.csv',allforces);
```

```
allforces = [max_bare max_min max_shod meanbare meanmin meanshod];

figure('visible','on')
%maxgroups = [repmat(cellstr(sprintf('Bare')),1);
repmat(cellstr(sprintf('Min')),1); repmat(cellstr(sprintf('Shod')),1)];
%meangroups =
[repmat(cellstr(sprintf('Bare')),1);repmat(cellstr(sprintf('Min')),1);repmat(
cellstr(sprintf('Shod')),1)];
allgroups = [repmat(cellstr(sprintf('Bare Max')),1);
repmat(cellstr(sprintf('Min Max')),1); repmat(cellstr(sprintf('Shod
Max')),1); repmat(cellstr(sprintf('Bare Mean')),1);
repmat(cellstr(sprintf('Min Mean')),1); repmat(cellstr(sprintf('Shod
Mean')),1)];
%boxplot(allforces(:,1:3),maxgroups); title('Achilles Tendon Max Force
Data');
%figure
%boxplot(allforces(:,4:6),meangroups); title('Achilles Tendon Mean Force
Data');
figure('Visible','on')
%}
boxplot(allforces,allgroups); title('Achilles Tendon Force Data');
ylabel('Force Over Body Weight')
```

The following script was used to perform the repeatability study. It contains much of the same syntax as the original function.

```
clear all; close all; clc
 Subject = 'Repeatability';

indir = ['\\filer\home\.nthome\desktop\MQP\' Subject '\'];

 xlsx = dir([indir '*.xlsx']);
 csv = dir([indir '*.csv']);

plotflag = 1;

BW = 691.8; %N

contacton = [7 10 8 9 9 9 11 10 11 8];
contactoff = [70 81 73 72 78 81 80 77 79 74];


imageJ = 285.1; %px/m

lfoot = 0.286; %m
hankle = 0.114; %m
achilles2ankle = 0.092; %m
toe2ankle = 0.173; %m
yoffset = 36.3;
xoffset = 7.6;

colors = ['b' 'r' 'k' 'c' 'g' 'b' 'r' 'k' 'c' 'g'];
trials = zeros(87,10);


for k = 1:10
%% Extract Data from Video

ForceData = fullfile(indir,csv(k).name);
VideoData = fullfile(indir,xlsx(k).name);

disp(['Force is ' ForceData ' and Video is ' VideoData]);

start = num2str(contacton(k)); finish = num2str(contactoff(k));

time = xlsread(VideoData,1,['B' start ':B' finish]);
shankCOMx = (xlsread(VideoData,1,['C' start ':C' finish])-xoffset)./imageJ;
shankCOMy = (xlsread(VideoData,1,['D' start ':D' finish])+yoffset)./imageJ;
calfx = (xlsread(VideoData,1,['G' start ':G' finish])-xoffset)./imageJ;
calfy = (xlsread(VideoData,1,['H' start ':H' finish])+yoffset)./imageJ;
anklex = (xlsread(VideoData,1,['K' start ':K' finish])-xoffset)./imageJ;
ankley = (xlsread(VideoData,1,['L' start ':L' finish])+yoffset)./imageJ;
heelx = (xlsread(VideoData,1,['O' start ':O' finish])-xoffset)./imageJ;
heely = (xlsread(VideoData,1,['P' start ':P' finish])+yoffset)./imageJ;
footCOMx = (xlsread(VideoData,1,['S' start ':S' finish])-xoffset)./imageJ;
footCOMy = (xlsread(VideoData,1,['T' start ':T' finish])+yoffset)./imageJ;
```

```
toex = (xlsread(VideoData,1,['W' start ':W' finish])-xoffset)./imageJ;
toey = (xlsread(VideoData,1,['X' start ':X' finish])+yoffset)./imageJ;
tibx = (xlsread(VideoData,1,['AA' start ':AA' finish])-xoffset)./imageJ;
tiby = (xlsread(VideoData,1,['AB' start ':AB' finish])+yoffset)./imageJ;
tibshankx = (xlsread(VideoData,1,['AE' start ':AE' finish])-xoffset)./imageJ;
tibshanky = (xlsread(VideoData,1,['AF' start ':AF' finish])+yoffset)./imageJ;
COM2anklex = abs(anklex(1)-footCOMx(1));

time = (time-min(time))/10;
%% Extract Force Data and set to time from video

GRFx = csvread(ForceData,1,2,[1 2 1000 2]); %N
GRFy = csvread(ForceData,1,3,[1 3 1000 3]);  %N
Momz = csvread(ForceData,1,5,[1 5 1000 5]); %N-m
timef = csvread(ForceData,1,0,[1 0 1000 0]);

for i = 1:length(GRFy)-1
    if abs(GRFy(i)-GRFy(i+1)) < 1 || GRFy(i) < 0
        GRFy(i) = 0;
        GRFx(i) = 0;
        Momz(i) = 0;
        timef(i) = 0;
    end
end

GRFx(GRFy == 0) = [];
timef(GRFy == 0) = [];
Momz(GRFy == 0) = [];
GRFy(GRFy == 0) = [];

GRFx = GRFx*(-4.44822162); %N
GRFy = GRFy*(4.44822162);  %N
Momz = Momz*(-4.44822162*.0254); %N-m

timef = timef-min(timef);

GRFx(timef > time(length(time))) = [];
GRFy(timef > time(length(time))) = [];
Momz(timef > time(length(time))) = [];
timef(timef > time(length(time))) = [];

GRF = sqrt(GRFx.^2 + GRFy.^2);



%% Calculate accelerations and alpha

% Acceleration of the foot in x-dir
P = polyfit(time,footCOMx,7);

Footposx = zeros(size(timef));
for i = 1:length(Footposx)
    Footposx(i)= P(1)*timef(i)^7 + P(2)*timef(i)^6 + P(3)*timef(i)^5 +
P(4)*timef(i)^4 + P(5)*timef(i)^3 + P(6)*timef(i)^2 + P(7)*timef(i) + P(8);
```

```
end

afx = zeros(size(timef));
for i = 1:length(afx)
    afx(i) = 42*P(1)*timef(i)^5 + 30*P(2)*timef(i)^4 + 20*P(3)*timef(i)^3 +
12*P(4)*timef(i)^2 + 6*P(5)*timef(i) + 2*P(6);
end

%Acceleration of foot in y-dir
P = polyfit(time,footCOMy,7);

footposy = zeros(size(timef));
for i = 1:length(footposy)
    footposy(i)= P(1)*timef(i)^7 + P(2)*timef(i)^6 + P(3)*timef(i)^5 +
P(4)*timef(i)^4 + P(5)*timef(i)^3 + P(6)*timef(i)^2 + P(7)*timef(i) + P(8);
end

afy = zeros(size(timef));
for i = 1:length(afy)
    afy(i) = 42*P(1)*timef(i)^5 + 30*P(2)*timef(i)^4 + 20*P(3)*timef(i)^3 +
12*P(4)*timef(i)^2 + 6*P(5)*timef(i) + 2*P(6);
end

rads = atan((ankley-toey)./(anklex-toex))+pi;

P = polyfit(time,rads,7);

angle = zeros(size(timef));
for i = 1:length(angle)
    angle(i)= P(1)*timef(i)^7 + P(2)*timef(i)^6 + P(3)*timef(i)^5 +
P(4)*timef(i)^4 + P(5)*timef(i)^3 + P(6)*timef(i)^2 + P(7)*timef(i) + P(8);
end

deltaang = zeros(size(timef));
for i = 1:length(deltaang)
    deltaang(i) = 7*P(1)*timef(i)^6 + 6*P(2)*timef(i)^5 + 5*P(3)*timef(i)^4 +
4*P(4)*timef(i)^3 + 3*P(5)*timef(i)^2 + 2*P(6)*timef(i) + P(7);
end

alphaf = zeros(size(timef));
for i = 1:length(alphaf)
    alphaf(i) = 42*P(1)*timef(i)^5 + 30*P(2)*timef(i)^4 + 20*P(3)*timef(i)^3
+ 12*P(4)*timef(i)^2 + 6*P(5)*timef(i) + 2*P(6);
end

footCOMx = spline(time,footCOMx); footCOMx = ppval(footCOMx,timef);
footCOMy = spline(time,footCOMy); footCOMy = ppval(footCOMy,timef);
shankCOMx = spline(time,shankCOMx); shankCOMx = ppval(shankCOMx,timef);
shankCOMy = spline(time,shankCOMy); shankCOMy = ppval(shankCOMy,timef);
calfx = spline(time,calfx); calfx = ppval(calfx,timef);
calfy = spline(time,calfy); calfy = ppval(calfy,timef);
anklex = spline(time,anklex); anklex = ppval(anklex,timef);
ankley = spline(time,ankley); ankley = ppval(ankley,timef);
heelx = spline(time,heelx); heelx = ppval(heelx,timef);
heely = spline(time,heely); heely = ppval(heely,timef);
```

```
tibx = spline(time,tibx); tibx = ppval(tibx,timef);
tiby = spline(time,tiby); tiby = ppval(tiby,timef);
toex = spline(time,toex); toex = ppval(toex,timef);
toey = spline(time,toey); toey = ppval(toey,timef);
tibshankx = spline(time,tibshankx); tibshankx = ppval(tibshankx,timef);
tibshanky = spline(time,tibshanky); tibshanky = ppval(tibshanky,timef);
thetaf = spline(time,rads); thetaf = ppval(thetaf,timef);


COPx = Momz./GRFy;
COP2ankle = sqrt((ankley).^2+(COPx-anklex).^2);

%% Mass of foot/shank and I

mf = (0.009*BW+2.45)/9.81;
I = mf*9.81/18*(hankle^2 + lfoot^2);

%% Determine angles

thetaground = -abs(atan(GRFy./GRFx));

thetagrf = thetaf-thetaground;

thetacalf = abs(atan((calfy-heely)./(calfx-heelx)));
thetac = thetacalf - thetaf + pi;
thetashank = abs(atan((shankCOMy-ankley)./(shankCOMx-anklex)));
thetatibialis =  atan((tibshanky-tiby)./(tibshankx-tibx));

for i = 1:length(thetatibialis)
    if thetatibialis(i) > 0
        thetatibialis(i) = -thetatibialis(i);
    end
    thetatibialis(i) = thetatibialis(i)+pi;
end

thetat = thetaf-thetatibialis;
thetafootshank = thetashank - thetaf + pi;
%% Set up system

fachilles = zeros(size(GRFx));
ftibialis = zeros(size(GRFx));
fankle = zeros(size(GRFx));


    ialpha = zeros(size(GRFx));
    GRFmom = zeros(size(GRFx));
    springmom = zeros(size(GRFx));
    gravmom = zeros(size(GRFx));
    Momr = zeros(size(GRFx));

for j = 1:length(fachilles)

    ialpha(j) = I*alphaf(j);
    GRFmom(j) = GRF(j)*COP2ankle(j)*sin(thetagrf(j));
```

```
    springmom(j) = 401.070457*thetafootshank(j);
    gravmom(j) = mf*9.81*COM2anklex;

left = [cos(thetacalf(j)) cos(thetatibialis(j)) cos(thetashank(j));
        sin(thetacalf(j))  sin(thetatibialis(j)) sin(thetashank(j));
        achilles2ankle*sin(thetac(j)) toe2ankle*sin(thetat(j)) 0];

right = [mf*afx(j)-GRFx(j);
         mf*afy(j)-GRFy(j)-mf*9.81;
         ialpha(j)-GRFmom(j)-gravmom(j)];

solution = left\right;

fachilles(j) = solution(1)/BW;
ftibialis(j) = solution(2)/BW;
fankle(j) = solution(3)/BW;


Momr(j) = ialpha(j)-gravmom(j)-GRFmom(j);

end


thistrial = fachilles(4:length(timef)-4);
if k == 1
    trials(:,k) = thistrial;
    basetime = timef(4:length(timef)-4);
elseif k == 5
    trials(:,k) = 0;
else
    spfit = spline(timef(4:length(timef)-4),thistrial); trials(:,k) =
ppval(spfit,basetime);
end


if k == 1 || k == 2 || k == 3 || k == 4
    plot(basetime,trials(:,k),colors(k));title('Consistency of AT Force vs
Time'); xlabel('Time (s)'); ylabel('Force Over Body Weight');
    hold on
elseif k == 6 || k == 7 || k == 8 || k == 9 || k == 10
    plot(basetime,trials(:,k),['--' colors(k)]);title('Consistency of AT
Force vs Time'); xlabel('Time (s)'); ylabel('Force Over Body Weight');
    hold on
end


end
legend('Trial 1','Trial 2','Trial 3','Trial 4','Trial 5','Trial 6','Trial
7','Trial 8','Trial 9','Trial 10','Location','Southwest')
hold off
```

APPENDIX D: Spreadsheet of Subject Parameters

Table D-1: Data pertaining to the 14 subjects tested

Subject	BW (N)	imageJ	lfoot	hankle	achilles2ankle	toe2ankle	yoffset	xoffset
1	676.50	243.24	0.30	0.09	0.07	0.25	29.4	10.3
2	536.90	263.99	0.26	0.11	0.09	0.18	28.8	10.3
3	584.10	224.86	0.35	0.14	0.10	0.20	32.4	2
5	528.90	283.58	0.23	0.09	0.07	0.19	24	10.9
6	622.50	299.32	0.29	0.09	0.08	0.16	25.2	11.3
7	711.80	299.32	0.26	0.09	0.06	0.15	25.2	9.3
8	646.90	283.90	0.28	0.11	0.08	0.15	23.4	14.8
9	634.00	315.04	0.28	0.11	0.08	0.16	21.6	5.3
10	656.40	320.62	0.28	0.11	0.09	0.17	21.9	9
12	699.30	340.59	0.25	0.09	0.07	0.15	22.1	4.1
13	645.60	322.54	0.26	0.10	0.09	0.14	36	20.3
16	727.20	320.00	0.26	0.09	0.07	0.15	23.3	3.8
17	481.50	300.07	0.24	0.09	0.07	0.14	24.4	15
20	558.50	290.31	0.25	0.09	0.08	0.13	29.6	0.9
Average	622.15	293.38	0.27	0.10	0.08	0.17		
STDEV	74.27	32.08	0.03	0.01	0.01	0.03		

APPENDIX E: Plots of Estimated Achilles Tendon Forces for 14 Subjects

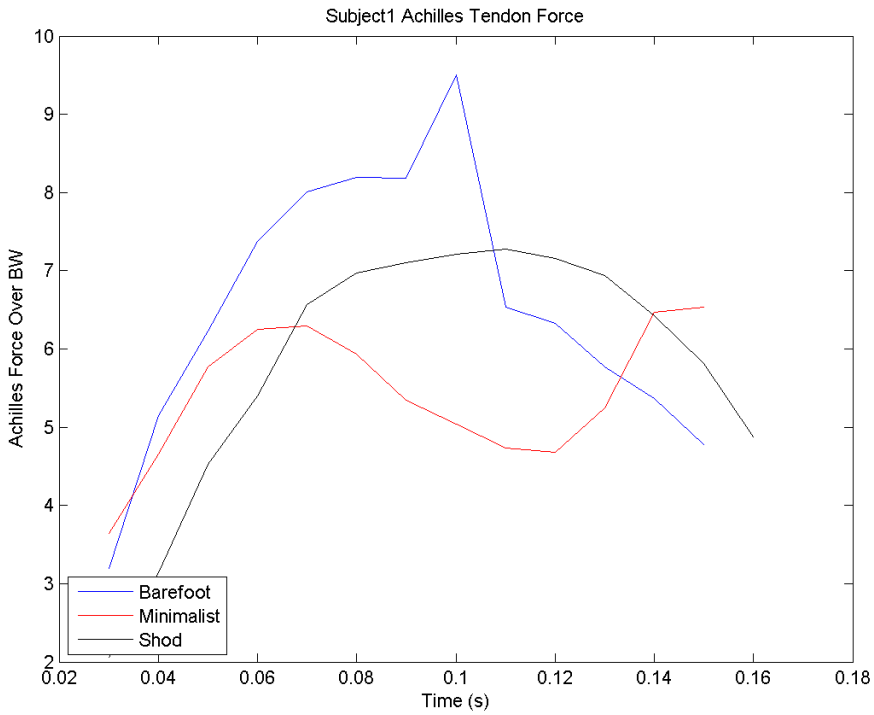

Figure E-1: Achilles tendon force data for subject 1

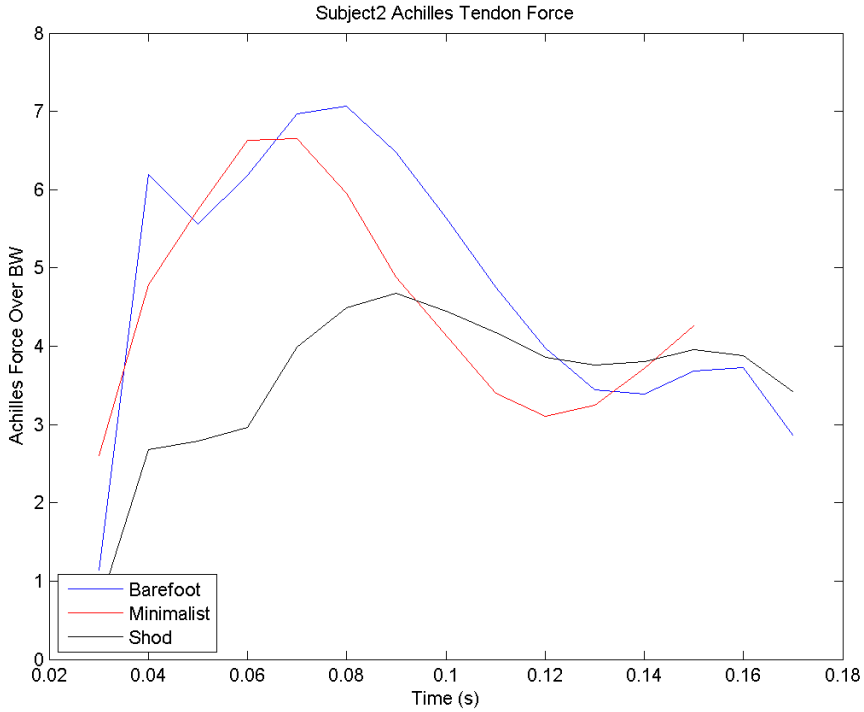

Figure E-2: Achilles tendon force data for subject 2

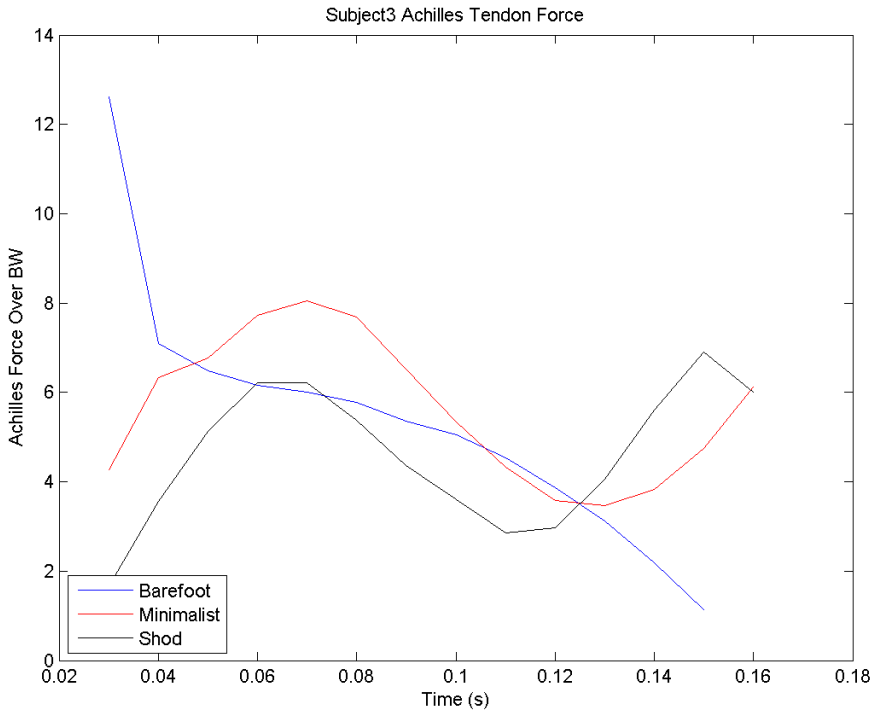

Figure E-3: Achilles tendon force data for subject 3

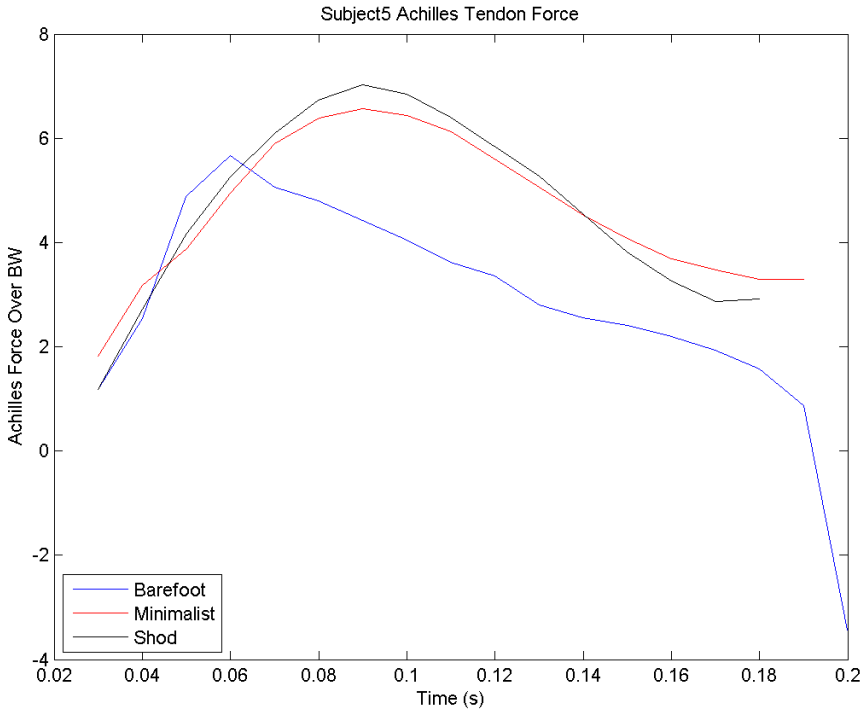

Figure E-4: Achilles tendon force data for subject 5

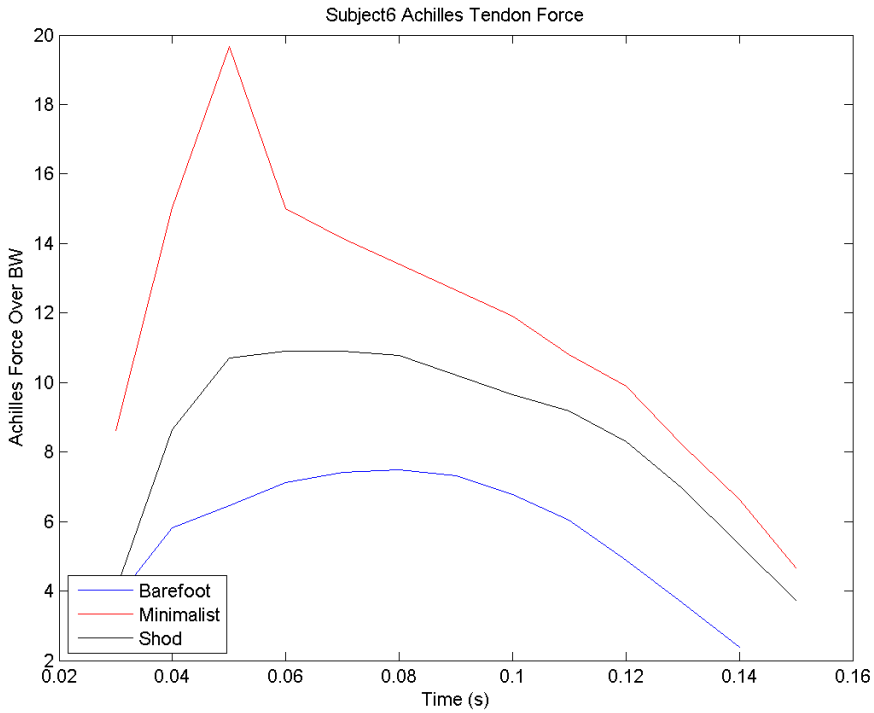

Figure E-5: Achilles tendon force data for subject 6

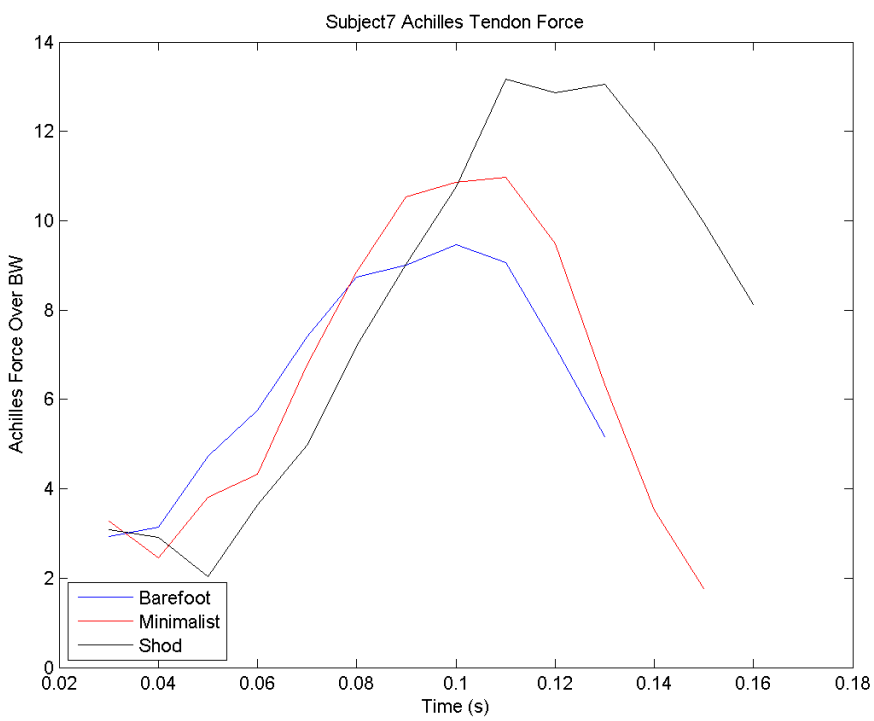

Figure E-6: Achilles tendon force data for subject 7

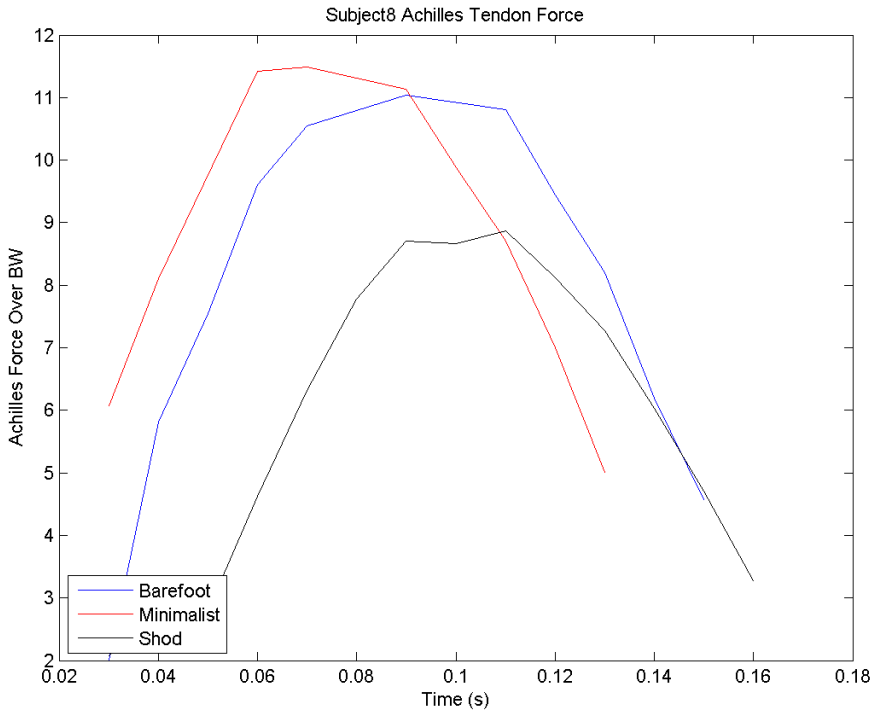

Figure E-7: Achilles tendon force data for subject 8

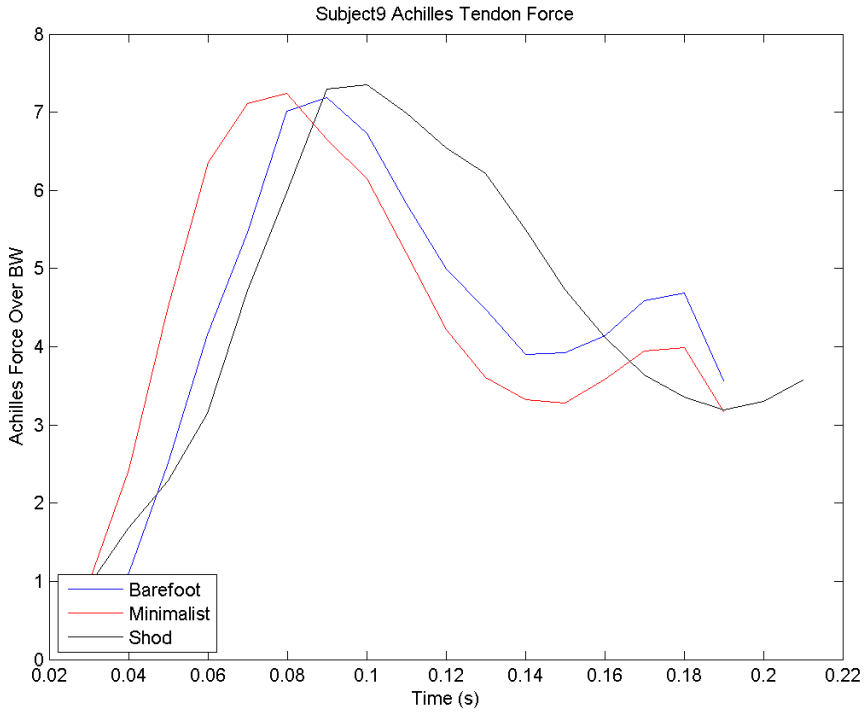

Figure E-8: Achilles tendon force data for subject 9

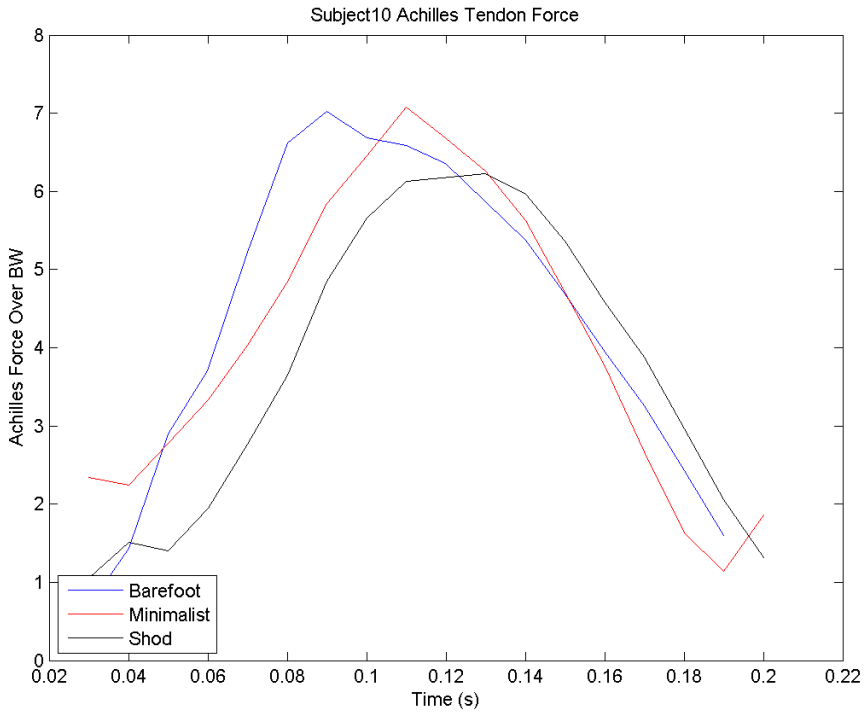

Figure E-9: Achilles tendon force data for subject 10

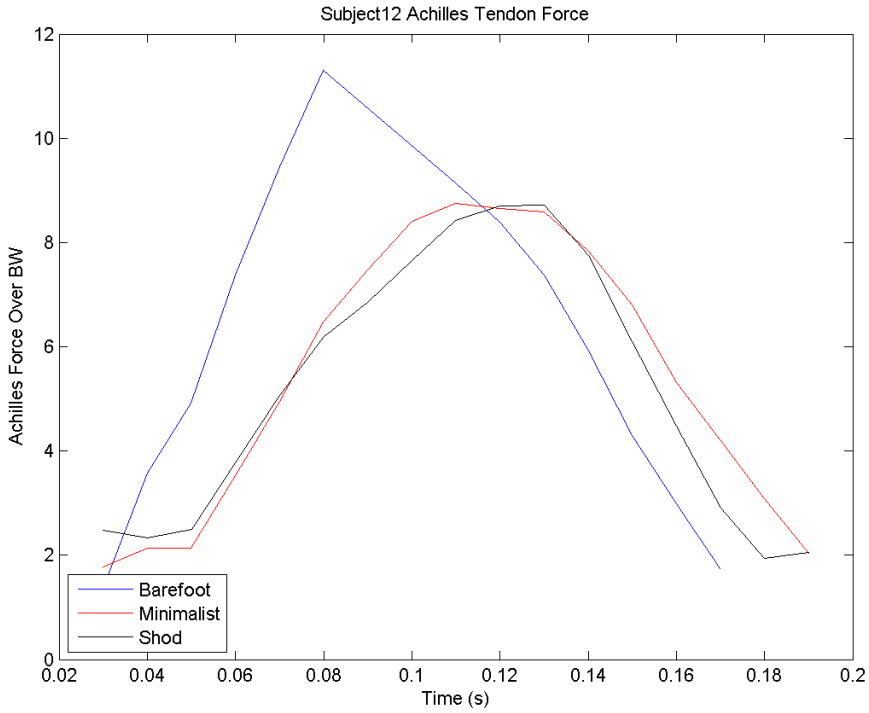

Figure E-10: Achilles tendon force data for subject 12

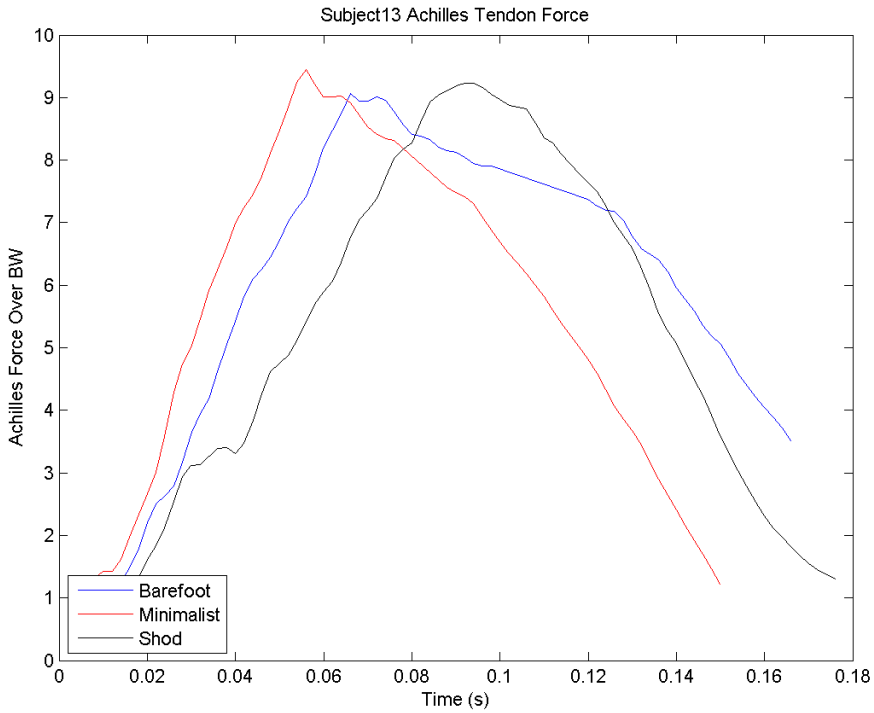

Figure E-11: Achilles tendon force data for subject 13

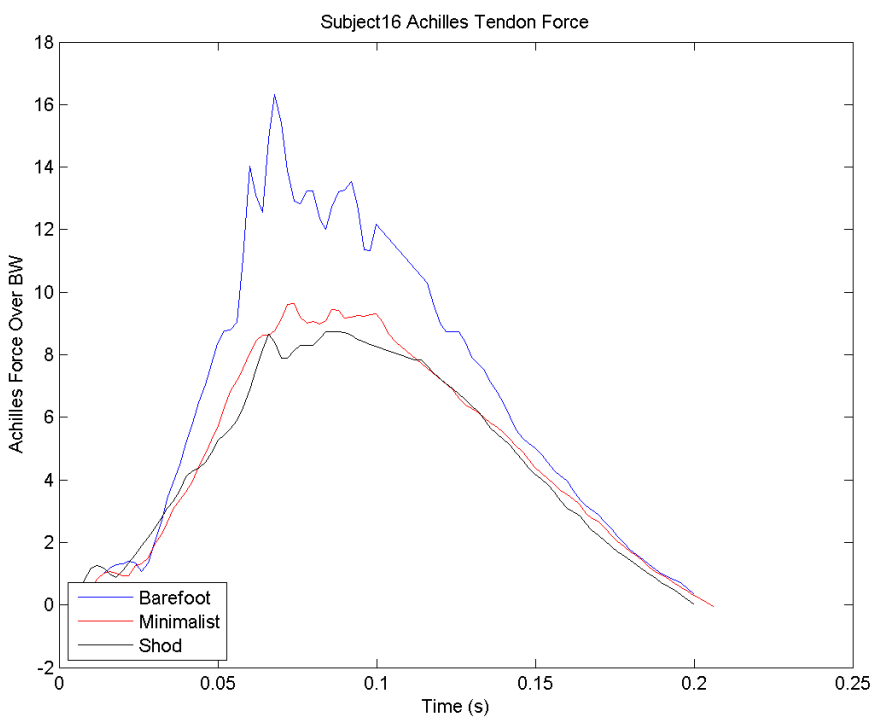

Figure E-12: Achilles tendon force data for subject 16

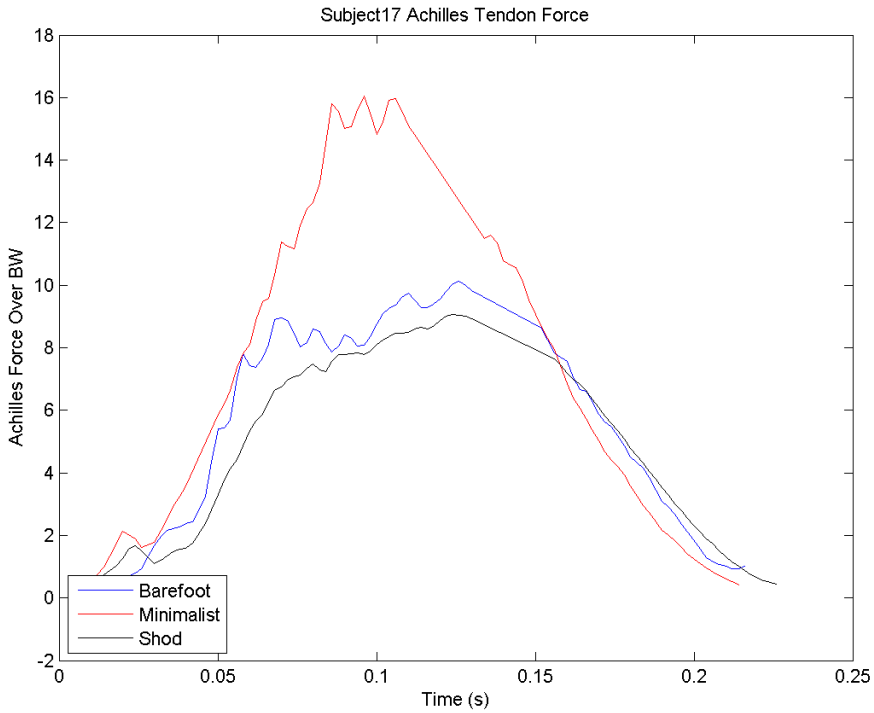

Figure E-13: Achilles tendon force data for subject 17

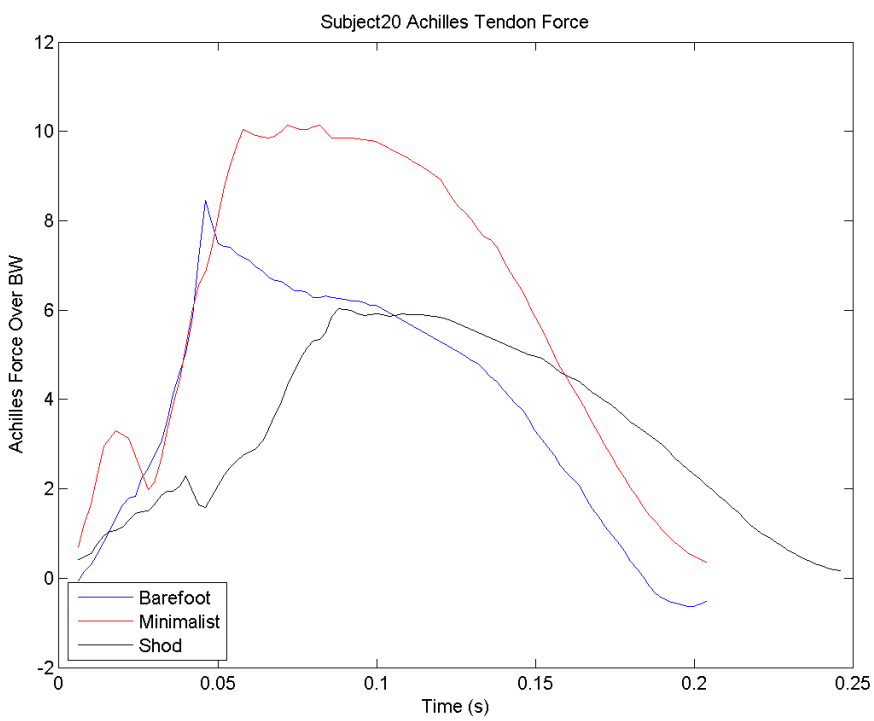

Figure E-14: Achilles tendon force data for subject 20

APPENDIX F: Estimated Forces for all Subjects

Table F-1: Maximum forces estimated in the Achilles tendon

Subject	Max Barefoot	Max Minimalist	Max Shod
1	9.5	6.5	7.3
2	7.1	6.6	4.7
3	12.6	8.0	6.9
5	5.7	6.6	7.0
6	7.5	19.7	10.9
7	9.5	11.0	13.2
8	11.0	11.5	8.9
9	7.2	7.2	7.3
10	7.0	7.1	6.2
12	11.3	8.7	8.7
13	9.1	9.5	9.2
16	16.3	9.6	8.7
17	10.1	16.0	9.1
20	8.4	10.1	6.0
Average	9.5 ± 2.8	9.9 ±3.8	8.2 ±2.2

Table F-2: Mean forces estimated in the Achilles tendon

Subject	Mean Barefoot	Mean Minimalist	Mean Shod
1	6.5 ± 1.7	5.4 ± 0.9	5.8 ± 1.6
2	4.7 ± 1.7	4.6 ± 1.4	3.6 ± 1.0
3	5.3 ± 2.8	5.6 ± 1.6	4.7 ± 1.6
5	2.8 ± 2.1	4.5 ± 1.4	4.6 ± 1.8
6	5.8 ± 1.7	11.5 ± 4.2	8.3 ± 2.7
7	6.6 ± 2.4	6.4 ± 3.4	8.0 ± 4.1
8	7.8 ± 2.9	8.8 ± 2.4	5.9 ± 2.5
9	4.4 ± 1.9	4.5 ± 1.8	4.5 ± 1.9
10	4.4 ± 2.1	4.1 ± 1.9	3.6 ± 1.9
12	6.0 ± 3.1	5.4 ± 2.6	5.2 ± 2.5
13	5.8 ± 2.4	5.4 ± 2.6	5.2 ± 2.8
16	6.3 ± 4.7	4.6 ± 3.2	4.5 ± 2.8
17	5.4 ± 3.4	7.1 ± 5.2	4.7 ± 3.0
20	3.6 ± 2.7	5.7 ± 3.4	3.2 ± 1.9
Average	5.4 ± 1.3	6.0 ± 2.0	5.1 ± 1.5

Table F-3: Maximum compressive forces estimated for the tibialis anterior tendon

Subject	Max Barefoot	Max Minimalist	Max Shod
1	-2.8	-2.3	-2.2
2	-2.8	-2.2	-2.3
3	-6.3	-3.2	-2.3
5	-2.3	-2.0	-2.5
6	-2.7	-8.8	-4.3
7	-3.5	-3.6	-4.7
8	-4.4	-4.1	-3.2
9	-2.9	-2.8	-2.9
10	-3.2	-3.0	-2.4
12	-4.3	-2.8	-3.2
13	-4.0	-4.1	-3.8
16	-6.7	-3.8	-3.0
17	-4.4	-6.9	-3.5
20	-4.1	-3.4	-3.1
Average	-3.9 ±1.3	-3.8 ± 1.9	-3.1 ± 0.8

Table F-4: Maximum compressive forces estimated for the bone force

Subject	Max Barefoot	Max Minimalist	Max Shod
1	-10.1	-7.3	-8.1
2	-7.4	-7.1	-5.3
3	-7.9	-7.9	-7.8
5	-4.8	-7.3	-7.3
6	-8.0	-13.4	-9.7
7	-8.4	-9.6	-11.7
8	-9.1	-10.1	-8.0
9	-6.5	-6.7	-6.6
10	-5.9	-6.2	-6.0
12	-9.5	-8.3	-8.0
13	-7.5	-8.1	-8.1
16	-11.8	-8.1	-8.1
17	-8.8	-11.9	-8.7
20	-6.1	-9.5	-5.3
Average	-8.0 ± 1.8	-8.7 ± 2.1	-7.8 ± 1.7