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The effect of fatigue on shock absorption in the human body

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The Effect of Fatigue on Shock Absorption in the Human Body

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ABSTRACT

The repetitive loading found in running sends impact shock waves through the body. These shock waves have been related to various types of injury. Proper attenuation and absorption of impact shock waves is critical to keeping the body healthy and injury-free. In this investigation, the role of fatigue in the capacity of shock absorption was examined. Skin-mounted accelerometers monitored a number of recreational runners for changes in the shock wave accelerations which occur after foot-strike as they were fatigued on a treadmill under normal workout conditions. The results found that approximately two-thirds of the subjects experienced reduced shock absorption capacity as they fatigued while the remaining subjects showed an increase in shock absorption capacity. No changes due to fatigue were found in the frequency response either in magnitude of power spectrum density or frequency. Possible causes for the reduction in shock absorption capacity were discussed. Other correlations between physical characteristics of the subjects are reduced shock absorption capacity were noted and discussed.

1.0 INTRODUCTION

Ever since humans have been a bipedal creature, there has been a yearning to run faster, longer and for farther distances, either because of the advantages in hunting and safety from other predators or because of the desire to compete and excel in competition. Yet despite such an interest in improving running performance, published literature of running has only happened recently in terms of the history of the activity (Cavanagh, 1990). This may be due in part to the complexity of running, the uniqueness of style of individuals and the lack of suitably accurate measurement techniques. Early studies of running were limited to qualitative observances and rudimentary measurements and calculations involving time and distance. Not until Etienne Jules Marey (Cavanagh, 1990) was there instrumentation capable of accurately recording basic characteristics of running as stride frequency and stride length. With the advancement of technology, there was increased interest in studying human locomotion and accompanying factors, such as muscle strength and capacity, economy of technique and kinematic analysis.

Recently, there has been an increase in the popularity of running as a sport (Cavanagh, 1990), and there are more people participating in running activities than ever before. While there are exceptional runners, the vast majority of people can be considered recreational runners. Many recreational runners train without supervision, and consequently, may expose themselves to a number of injuries which can be the result of overuse. Health problems which may accompany a running routine may not be associated with other activities (Nigg, 1985). This may be a result of the continuous, repetitive loading nature of running as an activity (James and Jones, 1990). Studies involving animals showed a relationship between repetitive impact loading and several long-term injuries, including osteoarthritis, stiffening of bones and reduction of shock-absorbing capability in joints (Radin *et al.*, 1972). The increased interest in running combined with a rise in risk of injury has inspired more research in the biomechanics of running.

One area suspected of being associated with locomotive injuries is the foot strike and collision with the ground. This collision produces an impact shock wave which begins at the heel and propagates throughout the body (Voloshin and Wosk, 1982; Valiant, 1990). As it moves through the body, the shock wave is absorbed by the body as it is transmitted through the bones (Valiant, 1990). It is suspected, although not confirmed through clinical study, that the transmission and attenuation of shock waves is a cause of injury in humans.

A number of different methods to describe and measure the heel strike shock wave have been used. These methods include ground reaction force studies, which uses the principle that every force has an equal and opposite reaction. Ground reaction force studies generally investigate the initiation of the shock wave through the body by recording the shock that is imparted to the ground by the body at foot strike. A number of studies (Valiant, 1990; Valiant and Cavanagh, 1985; Light *et al.*, 1980; Nigg, 1983) have found factors which significantly reduce the magnitude of heel strike shock waves. Some of these factors are the type and grade of the running surface, the orientation of the foot at impact and the construction of the running shoe, including cushioning in the heel and in the midsole.

A second method used to measure the heel strike shock wave is the placement of lightweight accelerometers on or near the bones of the body. These areas of application include the shank of the tibia, the tibial tuberosity (near the front of the knee), the inner section of the femur at the knee joint, the forehead and the jaw joint (Voloshin and Wosk, 1982). These locations of investigation are able to track the shock wave as it propagates and is attenuated through the body, and are indicators of the shock absorption qualities of the body. Some of these absorption qualities are the heel pad, the style of running and orientation of foot at impact, and the flexibility and compliancy of joints.

One aspect of running which has not been addressed adequately is that running is a dynamic exercise and has the capacity to change and adapt to varying conditions. It is known that muscles under exertion or in a state of fatigue suffer from a reduction in capacity (Darcus, 1953). Such a reduction will affect the performance of running and the capacity of shock absorption, either directly or indirectly. Other tissues in the body may also see a reduction in their capacity as a result of fatigue and may affect the running performance and shock absorption capacity of the body as a whole.

A previous study by Dickinson *et al.* (1985) used the force plate technique to examine the ground reaction forces of subjects with respect to the effect of fatigue. Subjects were fatigued on a treadmill and periodically stopped to collect force plate data. Results from this study showed that the magnitude of forces occurring at foot strike increased as subjects moved from a rested state to a fatigued state. The conclusions were that fatigue had no apparent influence on frequency response and that there was a possible relationship between the height of the subjects and the time that the heel strike spike reached its maximum.

In the series of experiments of this thesis, the impact shock waves generated at heel strike during running were monitored by measuring the accelerations transmitted through the bones of the leg. Lightweight accelerometers attached tightly to the surface of the skin covering the tibia and femur were able to measure shock accelerations of subjects as they ran on a treadmill at a typical workout speed. Measurements were taken at regular intervals over the course of the session to examine gradual changes in accelerations in the leg over time. These changes in accelerations indicate a change in shock absorption capacity of the leg due to fatigue.

2.0 THEORY

2.1 HEEL STRIKE SHOCK WAVES

During running, there are several functions of the support extremity (James and Jones, 1990). These include maintaining forward motion, accelerating the body against resistances (internal and external), supporting the body's weight and absorbing the impact of foot strike. Unfortunately, the support extremity is not completely effective in absorbing the impact shock of foot strike. Several studies (Voloshin and Wosk, 1982; Valiant, 1990) have shown that after impact a shock wave beginning at the heel propagates throughout the body. This shock wave is attenuated as it progresses away from the heel, with up to ninety-nine percent of the energy absorbed in the heel in some cases (Valiant and Cavanagh, 1985). Over the course of time, the shock absorption capability of the human body may decrease, which could lead to injury.

2.2 FACTORS AFFECTING ABSORPTION OF IMPACT SHOCK WAVES

There are many features in humans which serve to attenuate shock waves. Some of these features can be considered active, in which the part changes or reacts to the shock wave as it goes by the area. Other attributes absorb shock in a more passive manner dependent on their construction or design. Some of these features are discussed below.

2.2.1 HEEL PAD

Beneath the calcaneus in the foot, there is a fatty tissue whose function is to absorb shocks initiated by impacts and other excessive pressures (Valiant, 1990). This combination of fat and fibrous tissue is approximately 20 mm thick and extends from the bottom surface of the calcaneus (heel bone) to the outer edge of the skin (Valiant, 1990). By the nature of its composition, the heel pad has a viscous quality, which allows it to absorb a high percentage impact energy very quickly (Valiant, 1990).

During testing, the heel pad hardens quickly under compression (Valiant and Cavanagh, 1985). The ensuing resistance to compression is a result of deflection of the fat and fibrous tissue composite material. The viscoelastic quality of the heel pad is shown when the load is removed and the pad does not return to its original shape as quickly as it is compressed (Valiant and Cavanagh, 1985). Although there is no conclusive indication that running leads to deterioration of the heel pad, there is documented evidence that a reduction in the capacity of the heel pad does lead to injuries such as heel pain (Marr and Pod, 1980), plantar faciitis (Sewell *et al.*, 1980) and Achilles tendinitis (Jorgensen, 1985).

2.2.2 JOINTS

Joints are generally defined as the bearing surfaces between adjacent bones (Kroemer *et al.*, 1990). There are three types normally found in humans: a) fibrous, which allow no appreciable relative motion; b) cartilaginous, which provide a limited amount of relative movement between the bones, and c) synovial, where joints have a space between the bones. These synovial joints are surrounded by elastic ligaments and usually contain articular cartilages. Sometimes, as in the knee joint, a fibrocartilage disk or wedge is present (Kroemer *et al.*, 1990).

Synovial joints are encased in a web of connective tissues (Kroemer *et al.*, 1990), which include ligaments, tendons, fascia and cartilage. Ligaments are dense tissues which connect bones and serve to stabilize the joint during intense strain. Tendons are similar to ligaments but connect muscles and bones. Fascia wraps organs or muscles and generally have no appreciable effect during movement. Cartilage is a "translucent material of collagen fibers embedded in a binding substance" (Kroemer *et al.*, 1990). By supplying firm but elastic mobility, cartilage may play an important part in the shock absorption capabilities of a joint.

In synovial joints, fluid can be secreted to lubricate the adjoining bone surfaces (Kroemer *et al.*, 1990). During intense strain, synovial fluid is absorbed by the articular cartilage. This fluid assists in the distribution of pressure across the bearing surfaces, reducing local pressures and risks of danger. Because it can increase the thickness of cartilage up to ten percent (Kroemer *et al.*, 1990), synovial fluid may play an important role in shock absorption.

2.2.3 MUSCLES

The last soft tissue element which might affect shock absorption capacity is muscle. Skeletal muscles provide voluntary motion as well as assist in the stability of joints. They are composed of hundreds, sometimes thousands, of muscle fibers bundled in fascia (Kroemer *et al.*, 1990). Each muscle fiber is made up of myofibrils, which are arranged in parallel to allow for contraction in one direction only. Each myofibril contains a number of myofilaments, which are protein rods also arranged in parallel. Myofibrils are cross-linked to provide the contractile movement.

In each muscle, a nerve fiber controls several hundred, sometimes thousands muscle fibers (Kroemer *et al.*, 1990). Each set of these muscle fibers is referred to as a motor unit and react to the same nerve signal. Each motor unit is distributed over the muscle, resulting in a weaker contraction over the length of the muscle rather than a strong localized contraction. In general, only about two-thirds of all fibers in a muscle can be voluntarily excited and contracted. If more than two-thirds react, a muscle strain or tear may result (Kroemer *et al.*, 1990).

A twitch is a single contraction resulting from a single instantaneous stimulus followed by complete relaxation (Kroemer *et al.*, 1990). An entire cycle takes between 75 and 220 ms, consisting of a latent period, a period of shortening, a period of relaxation and a period of recovery. Typically, the latent period is about 10 ms; the shortening period lasts about 40 ms; relaxation takes about 40 ms; and the recovery period is about 40 ms.

Occasionally, stimuli trigger twitches before the end of the previous cycle. As a result, twitches take on a summation effect (Kroemer *et al.*, 1990). This superposition of twitches allow for higher contractile tension. Generally, superposition happens at stimulation frequencies greater than ten cycles/second. When stimuli frequencies reach 30-40 cycles/second, successive contractions fuse together to result in a maintained contraction referred to as tetanus.

Muscles also change their behavior on a macroscopic scale as they are used under static loading or phasic movement at constant speed and/or load. As effort is maintained, there is a "progressive increase in activity of muscles or more contraction to produce the same output" (Darcus, 1953). The increased contraction of one muscle begins to involve the surrounding muscles, which may lead to the actions of the activity ceasing to be smooth. Also, in addition to amplitude increases in muscular activity, there is an increase in the duration of individual contractions. Eventually, contraction cycles may overlap and complete muscle relaxation stops.

2.2.4 BONES

The transmission heel strike impact shock waves through the body is primarily done by bones (Valiant, 1990). Bones are by far the most rigid objects in the body and are capable of resisting high strain (Kroemer *et al.*, 1990). However, bones do have certain elastic properties which may contribute to the absorption of impact shock. The shock absorption capacity of bones is small (Valiant, 1990), especially when compared to soft tissues. Also, the effect of fatigue of shock absorption capacity of bone is assumed to be very small when compared to the other components of shock absorption.

2.2.5 SHOES AND RUNNING SURFACES

The collision of the foot with the shoe and the ground at heel strike generates the impact shock wave that travels through the body. Obviously, the conditions of the shoe and running surface will affect the shock wave in magnitude and rate of generation. In fact, the insertion of highly viscous shock-absorbing heel inserts can reduce the peak acceleration on the tibia shank by 50 percent and reduce the rate of rise of acceleration significantly (Light, McLellan and Klenerman, 1980). Also, running on different surfaces may reduce peak accelerations (Nigg, 1983).

2.2.6 RUNNING STYLE

Another factor which determines the shock absorption characteristics of humans is the technique, style or gait with which they run. The runners can be classified into one of three categories: rear-foot strikers, mid-foot strikers, or fore-foot strikers (Cavanagh and Lafortune, 1980). Each type of runner encounters a different impact condition and consequently different shock-wave magnitudes and patterns.

One of the most compelling features of running techniques as a factor in shock absorption is its ability to change from one style to another. Often, this is done subconsciously, as when one runs on a non-level grade (Hamill *et al.*, 1984). However, technique can be controlled by conscious effort of the runner. With proper training and feedback, heel strike can be reduced or eliminated altogether (Dickinson, Cook and Leinhardt, 1985). A general feeling of fatigue may contribute greatly to a change in running technique either on a conscious or subconscious level.

2.2.7 OTHER FACTORS

In addition to the factors listed above, there are various other physical and psychological factors which may contribute to the performance of a runner, both in a positive or negative way. Some general positive influences on performance and fatigue include feedback of results, instructions on proper technique, arousal of ego, hypnosis and drugs. Negative or uncertain influences include fear of injuries, incentives or goal setting, verbal encouragement, competition and spectators (Kroemer *et al.*, 1990).

2.3 ROLE OF FATIGUE IN ABSORPTION

Over the course of an exercise routine, fatigue will begin to affect the performance of the activity. Fatigue will generally affect the muscles (Darcus, 1953; Kroemer *et al.*, 1990; Cooper *et al.*, 1988; Beelen and Sargeant, 1991) and, presumably other soft tissues of the body that assist in movement in running. These soft tissues are responsible for a portion of impact shock absorption and consequently, the amount of impact shock which is safely absorbed by the body may decrease over the course of an exercise routine. Decreased shock absorption capacity has been linked to various types of injuries (MacLellan and Vyvyan, 1981; Voloshin and Wosk, 1982).

2.4 INDICATIONS OF FATIGUE EFFECT

Changes in shock absorption capacity would result in different magnitudes of shock waves propagating through the bones of the leg. For example, increasing shock absorption capacity by placing a shock absorbing heel insert in the shoe reduces accelerations measured on the bone due to impact accelerations (Light, McLellan and Klenerman, 1990). Therefore, one indicator of reduction in shock absorption capacity due to fatigue is greater shock waves occurring at later times in the workout. This indication is measured by larger heel strike peak accelerations in the tibia and femur later in the workout (Dickinson, Cook and Leinhardt, 1985). In addition, the shock absorption capacity of the knee joint can be examined by comparing the transfer of shock from the tibia to the femur.

Impact shock waves also may be characterized by their power spectrum density, i.e. the amount of energy transmitted at various frequencies. As fatigue becomes a factor in shock absorption capacity, the amount of energy transmitted at various frequencies may change. Therefore, a second indicator of the effect of fatigue on the capacity of shock absorption may be a change in the magnitude of power transmitted at particular frequencies over the course of the workout. In a similar way, a shift in the maximum power frequency over time may be a third indicator of fatigue affecting the shock absorption capacity.

3.0 EXPERIMENTAL APPARATUS AND PROCEDURE

All data collection was done in the Fitness Center of Taylor Gym on the campus of Lehigh University. Also, the Fitness Center facilities were used with the permission and cooperation of the Fitness Center staff.

3.1 APPARATUS

All equipment, except for the treadmill, was portable and brought on-site for each of the collection sessions.

3.1.1 ACCELEROMETERS

The accelerometers used in this experiment were lightweight quartz accelerometers, with a nominal weight of two grams. When mounted appropriately, the accelerometers converted vibration and shock motion into a high-level, low-impedance voltage signal. Each accelerometer was capable of reliably measuring over a frequency range of 1 to 10000 Hz with a sensitivity of 0.01 g. The resonant frequency of the accelerometers when mounted according to specifications was approximately 70,000 Hz. As mounted in the experiment, the resonant frequency was reduced.

The accelerometers were provided with power over a single coaxial cable and returned a voltage signal to an appropriate amplifier. This amplifier was able to multiply the signal it received by 1, 10 or 100 and transmit the signal to a recording device. Each accelerometer was recorded as its own separate data stream and had a separate channel dedicated to it in the software program.

3.1.2 RECORDING DEVICE

The recording device used in this experiment was a portable computer using an analog-to-digital signal converter board. An oscilloscope was mounted in-line with the amplified signal to give immediate approximate feedback on the operation of the measuring equipment.

The portable computer was manufactured by Micro Express and was operated by an INFORMTECH286/SUPRA processor running at 8 Mhz. The computer had data storage capacity of 16 megabytes and operated with 1 megabyte of random access memory.

The analog-to-digital (A/D) signal converter board had twelve bit capacity and a signal range of +/- 10 volts.

3.1.3 RECORDING SOFTWARE

The software used to acquire the data in this experiment was Streamer v 3.1. This software enabled the computer to store digital data directly onto a hard drive storage space at a high speed sample rate. This data was stored on the hard drive in an encoded form in a contiguous file. Before any data processing could take place, the "streamed" data was "unpacked" into any ASCII data format.

Streamer data acquisition software is available through MetraByte Corporation, 440 Myles Standish Blvd., Taunton, MA 02780.

3.1.4 TREADMILL

The running surface used in the data acquisition for this experiment was a Star Trac 3000 Programmable treadmill. The treadmill was capable of performing a variety of factory-installed workout routines as well as user-programmed workout routines. Also, the treadmill was equipped with an emergency stop (panic-type) button as well as front and left-hand side safety handrails.

The treadmill was capable of variable speed control, with a range of 1.0 - 10.0 miles/hour (1.6 - 16 km/h). The speed could be varied in 0.1 mph increments and was able to maintain a constant speed with +/- 0.2 mph. Also, the Star Trac 3000 was able to provide inclines to simulate running uphill. The range of inclines was 0 - 15 % with increments of 1 percent.

Output available from the treadmill included elapsed time, distance, total calories burned, pace (minutes/mile), and heart rate. Also included on the control panel was a display showing the factory-installed programs, incline, and a graphical representation of a 1/4 mile track which showed equivalent distance travelled.

3.2 PROCEDURE

Before any data was collected, a Lehigh University Human Subjects Questionnaire was submitted to the Institutional Review Board for Human Subjects Research in accordance with the Lehigh University Human Subject Policy. After evaluation in the limited review category, approval was granted for one year from the approval date. A copy of the approval letter is in Appendix A.

Recruitment of volunteers was done through the use of posters distributed at Taylor Gym and Packard Laboratory, electronic bulletin board posting on the Lehigh University Network Server, and public information sessions held at Taylor Gym. Active recruitment through word-of-mouth was also used during early data collection sessions. Subjects volunteered to perform in the experiments, and were under no obligation to Lehigh University or others to participate in part or in whole.

Once a volunteer expressed interest in participating, a time was arranged for the subject to perform the experiment at his convenience. At the beginning of the session, the subject was instructed to review the release form and questionnaire provided. A copy of these documents are contained in Appendix B and Appendix C. When both forms were completed, the subject was ready to begin the experiment.

3.2.1 SKIN-MOUNTED ACCELEROMETER

The accelerometers used to monitor the impact shock waves were mounted to small, lightweight brackets as shown in Figure 3.1. The brackets had threaded connections to provide the accelerometers with a stable surface to mount onto the skin. The brackets were attached to each subject by an elastic band wrapped around the leg at the appropriate locations (see Figure 3.2). The tibia measurement location was in the front of the leg just beneath the kneecap, also known as the tibial tuberosity. The location for the femur measurement was on the bony area inside of the thigh just above the knee. In order to reduce error inherent in skin-mounted accelerometers (Valiant, McMahon and Frederick, 1987), the elastic band was tightened as tight as possible for each subject.

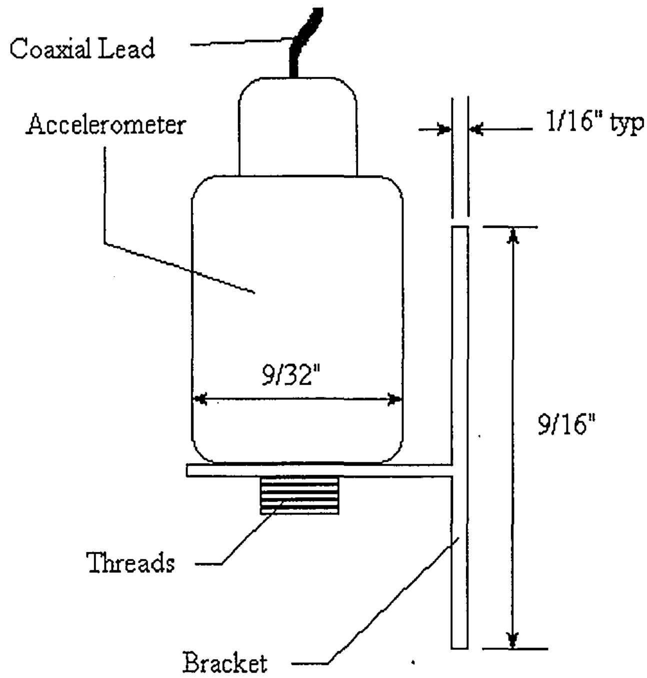


Figure 3.1 Accelerometer Attached to Bracket

During each session, the connections leading from the accelerometer to the computer were secured to some fixed object (usually the treadmill side safety guard) to prevent the wires from interfering with data collection. Without this fixturing, the wires were able to dangle near the accelerometer, occasionally jarring it. The external impact of the wire against the accelerometer would make the measurements invalid. Whenever possible, the wires were secured by taping the wire to the subject himself at the hip of the leg being measured. This action also assisted in preventing wire interference. In all cases, enough slack was allowed in the wires so that additional

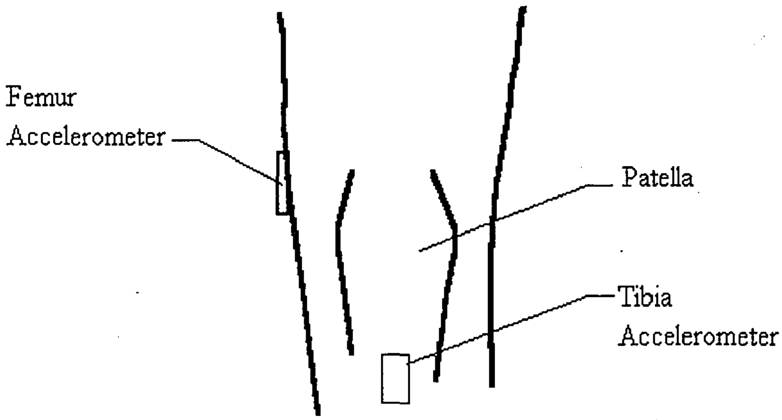


Figure 3.2 Location of Accelerometers on Left Knee

tensions did not provide damping to the accelerations.

3.2.2 TREADMILL OPERATION

During each data acquisition session, the treadmill was operated in manual mode. The incline was constant throughout the entire experiment and set to zero percent. No other modifications were made to the treadmill.

Before each session, the subject was instructed on the operation of the treadmill, including emergency stop procedures. The subject was reminded that participation in the experiment was voluntary and that he was under no obligation to Lehigh University or any other entity to complete the experiment. Also, each subject was instructed to be aware of conditions which indicate a physical problem (such as dizziness, numbness, shortness of breath, chest pains, etc.) that should stop the exercise.

The experimenter began each trial by starting the treadmill at its lowest setting, 1.0 mph. Over the first two minutes, the experimenter gradually increased the speed to the desired speed, usually in 2 mph steps in order to acclimate the subject to the feel of the treadmill. Once the desired setting was reached, no changes were made to the conditions of the treadmill until data collection was finished.

Upon completion of data collection, the speed of the treadmill was reduced gradually, usually in steps of 2 mph. Once the subject felt capable of reducing the speed himself, the experimenter allowed the subject to do so.

3.2.3 SUBJECTS

Volunteers were recruited based on certain criteria of running performance. Each subject was to be a "consistent" runner, i.e. the person was already in a regular running routine of twenty or more minutes per session at least three times a week for a number of years. These criteria were chosen so that: subjects would be familiar with distance running and when they would fatigue; subjects would be conditioned well and have a low risk of physical danger when participating; and subjects' bodies would be accustomed to the rigors of straight-ahead running, as opposed to lateral direction conditioning found in court or field sports.

All subjects were in good overall health. None were recovering from any recent leg injuries, although some had previous injuries which had long since healed. No subjects reported any detrimental long term effects from injuries, and one subject actually had better performances after recovery from an injury.

Subjects were instructed to treat the experimental session as though it were a typical workout. They were not to purchase any new equipment nor vary their warm-up routine in order to improve their performance. Consequently, a variety of shoes were used by the subjects, and warm-up routines varied from no warm-up to over 10 minutes of stretching and light exercise.

From the pre-session questionnaire, a speed for each subject's typical training session was calculated. The subject was consulted about this "normal" workout speed, and slight adjustments were made to the speed once the subject reached this speed on the treadmill.

3.2.4 DATA ACQUISITION

Before starting the treadmill, the overall length of running time was agreed upon by both the subject and experimenter. The total length of each data acquisition session was 30 minutes. In a few cases the time was somewhat less than 30 minutes because of subject capacity limitations.

Once the treadmill was started, the speed was increased to the "normal" workout speed as described above. Because most subjects did not know the exact speed, distance or time of a typical workout, the speed was adjusted within the first two minutes to fit their workout pace. If the subject was comfortable with the idea, the speed was increased slightly (0.2 - 0.4 mph) to insure a fatigue condition near the end of the session. A fatigued condition for this experiment is defined as where the subject feels tired but not exhausted at the end of the trial.

The first data sample was taken at two minutes after the start of running, usually within one minute of achieving the final experimental speed. This reading can be considered a "rested" state and was used to compare all ensuing measurements. After the initial measurement, data was collected at three minute intervals, i.e. five minutes, eight minutes, 11, 14, etc. This sampling interval was short enough to provide a somewhat continuous indication of shock absorption and long enough to allow the computer and software to perform the necessary conversion and file storage. There was adequate time between data collection for the experimenter to make notes on the conduct of the experiment or adjust equipment if necessary.

Each data sample consisted of approximately 30 seconds of data. This length of time provided between 25 and 40 heel strikes, depending on the subject. Each data sample was recorded at a one kHz sampling rate over two channels (one for each accelerometer), or 500 Hz per channel.

During all sessions, the subject was not allowed to view the oscilloscope used to monitor the accelerometer signals. Subjects were allowed to view only the treadmill informational displays. No verbal feedback was given to the subjects, but the subjects were not discouraged from talking about other topics during the trial.

3.2.5 DUPLICATE SUBJECTS

In order to determine if the effect of fatigue was constant within an individual over multiple exercise events, some subjects were asked to participate in a second or third trial.

4.0 DATA REDUCTION

The raw data saved to hard disk by the Streamer software package was in a compressed, unreadable form. In order to retrieve the data, the files must be "unpacked" into an ASCII format, which can then by manipulated by other software. In the execution of this experiment, data was unpacked into two columns of numbers ranging from 0 to 4095, each representing the voltage signal of one channel. Streamer identified which column represented which channel by the first row in the columns. Plots of sample raw data can be found in Figures 4.1 and 4.2. Because the accelerometer signal amplification was adjusted so that voltage never exceeded the maximum voltage of the A/D board, the extremal values of 0 and 4095 were never reached.

4.1 IDENTIFYING HEEL STRIKE PEAKS

The data recorded by the computer was a representation of the acceleration of the leg of the subject as a function of time. In order to determine the effect of fatigue on shock absorption, the peak acceleration of each heel strike needed to be identified and separated from the other data. This was accomplished by writing a FORTRAN program. A copy of the source code of this program can be located through the Mechanical Engineering and Mechanics Department of Lehigh University.

The FORTRAN program read the data from both channels of a single time trial into memory, taking care to use the first row to identify which channel was which. Once this was done, the absolute maximum value and absolute minimum value of the

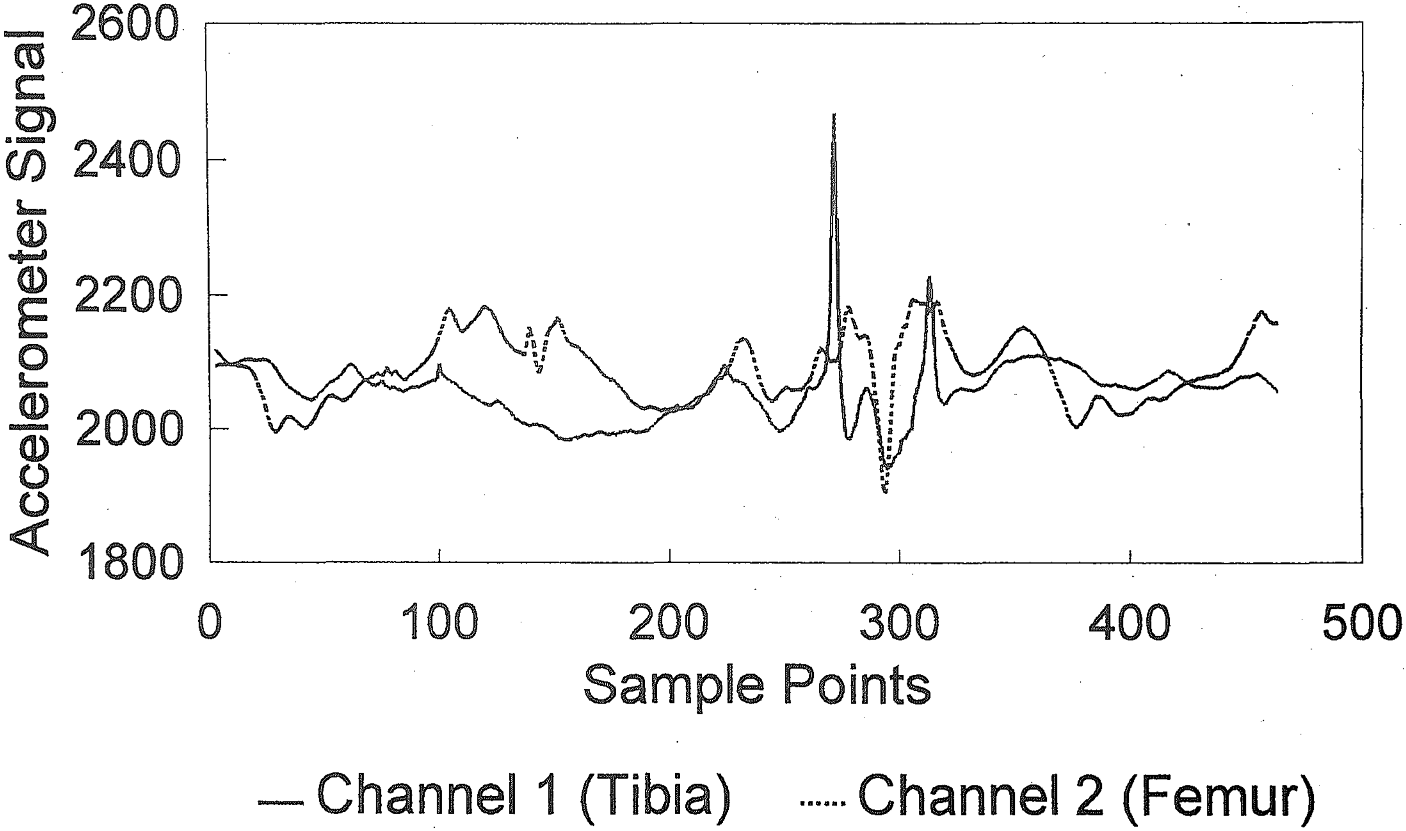


Figure 4.1 Raw Data vs. Time, One Heel Strike

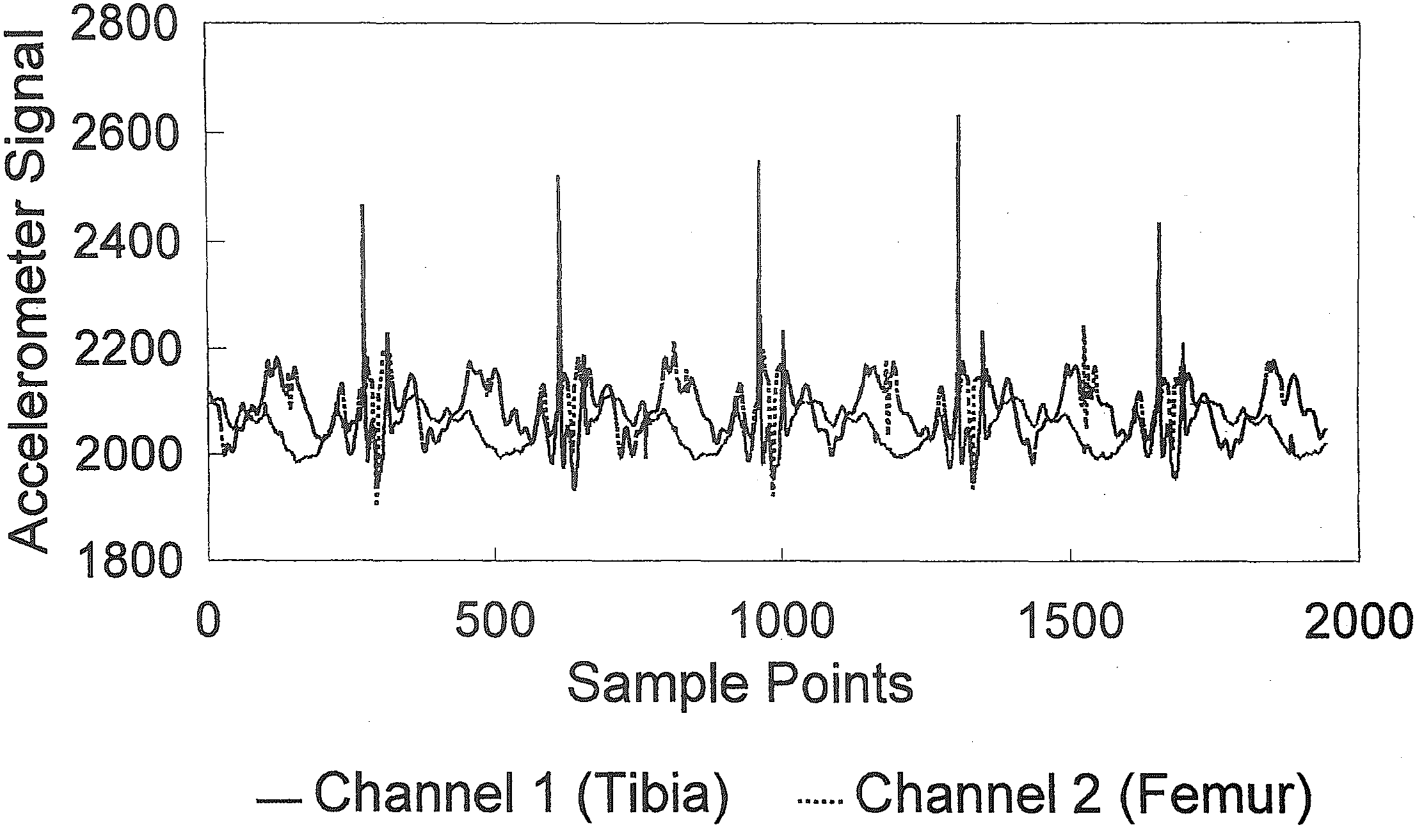


Figure 4.2 Raw Data vs. Time, Five Heel Strikes

tibia signal was identified to determine the range of acceleration values. The tibia was selected because in general the tibia signal gave a better and more distinctive indication of the impact shock wave.

It was confirmed that the accelerations caused by the heel strike were the maximum and minimum values recorded by observing the oscilloscope during collection of the data. Although not confirmed for every time trial of every subject, it was assumed that all heel strike accelerations were approximately equal in value for a specific time trial. Knowing this, each heel strike maximum acceleration could be identified by searching for a localized maximum (Figure 4.3).

In order to establish a range in time to search for the maximum acceleration, a jump criterion was established. This criterion identified the point in time when the signal would change more than a predetermined percentage of the absolute acceleration range. The large, sudden change in acceleration found by this criterion was generally caused by a heel strike shock wave.

When the jump criterion identified a suspected heel strike, the program called a subroutine which would plot the 400 data points surrounding the jump. The values correspond to a plot for 0.8 s around a suspected heel strike. The plot allowed for a visual inspection of the signal. A prompt asking for confirmation of the heel strike allowed the user to verify that the identified point was indeed a heel strike. If the jump was a false reading, the user could bypass to the next suspected heel strike.

Once a true heel strike was identified, the program then searched the data from the plot range to identify the localized maximum acceleration. After the maximum was

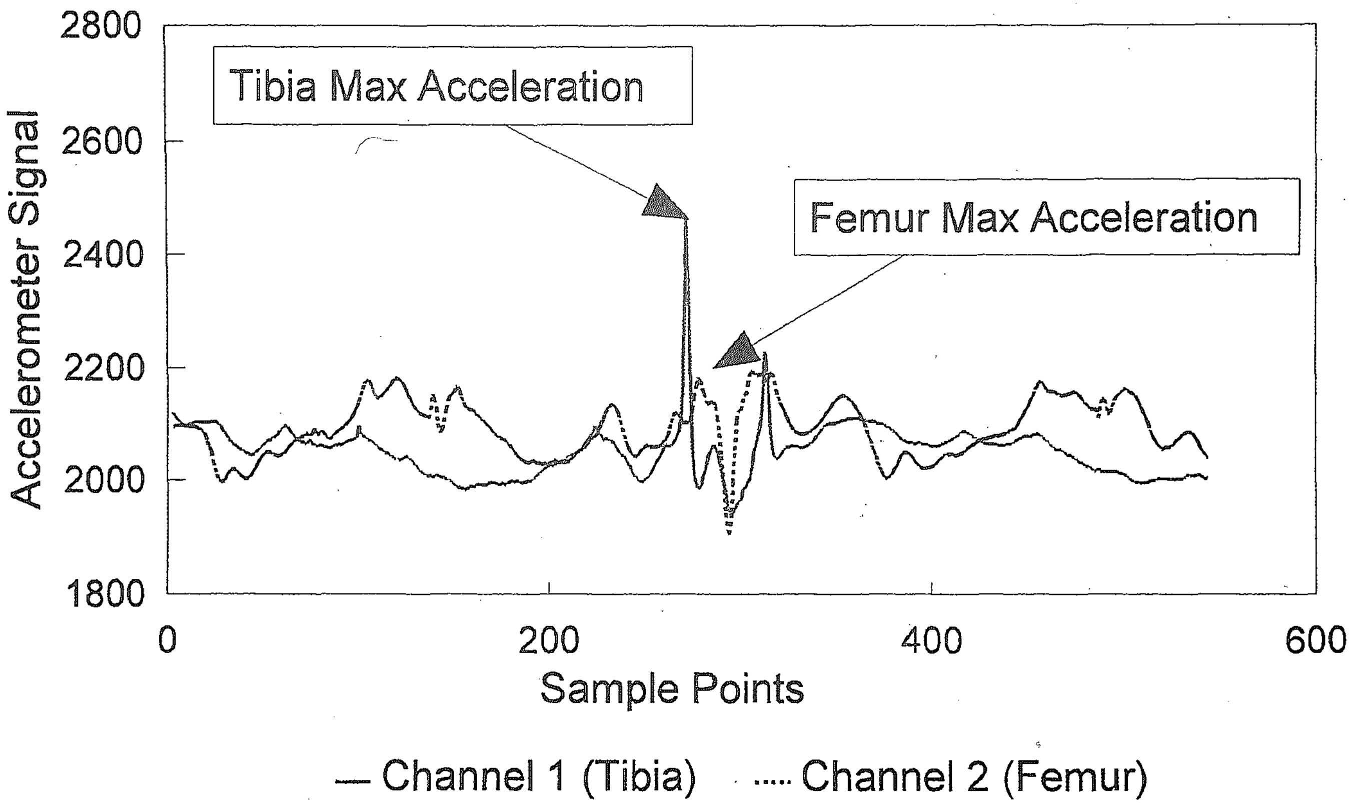


Figure 4.3 Maximum Accelerations at Heel Strike

determined, another prompt asked if the localized maximum represented the maximum acceleration of the heel strike. If the maximum acceleration did represent the heel strike, the program then plotted the identical time period for the femur accelerations and searched the femur accelerations in an area 10 points (20 ms) before to 10 points (20 ms) after the time of heel strike peak tibia acceleration. The program prompted the user for confirmation of the localized femur maximum acceleration. If the acceleration was correct, the maximum tibia acceleration, its corresponding time index, the maximum femur acceleration and its corresponding time index were written to a separate data file.

4.2 FREQUENCY ANALYSIS

To determine the power spectrum density of the heel strikes, another FORTRAN program was written. This program retrieved the heel strike time indexes identified in the initial program and prepared data arrays for use in the subroutine SPCTRM. SPCTRM was retrieved from Numerical Recipes and uses fourier transforms to perform a spectrum analysis of data. A copy of the source code of this program can be located through the Mechanical Engineering and Mechanics Department of Lehigh University.

For the discretely sampled data of this experiment, the sampling interval was 0.002 seconds, which corresponded to a Nyquist critical sampling frequency of 250 Hz. The number of discrete frequencies to examine between 0 Hz and the Nyquist frequency 250 Hz was chosen to be 128. This value gave a power value for every multiple value of 1.95 Hz up to 250 Hz. Also, the number of partitions of original sampled data, which is equal to the number of fast fourier transforms and the number of frequency periodograms which were averaged to determine the power spectrum density, was chosen to be two. In order to produce the smallest spectral variance per data point (Press *et al.*, 1989), the data point segments were overlapped.

From the conditions listed above, the proper amount of data points needed in the SPCTRM subroutine for each spectrum analysis was 640. A total of 60 points, 20 (or 40 ms) preceding maximum tibia acceleration and 40 (80 ms) after, were retrieved from the raw data. The remaining data points were zero-padding, with 30 zero-points preceding raw data and the remaining zeroes following the recorded data. An example of a data array is located in Figure 4.4. This padding of the data array with zeroes helps to minimize the end effects normally associated with the fourier transform method used in SPCTRM. This process was used for the first ten heel strikes of each time trial for each subject, and the resulting power spectrum densities were saved to a separate data file.

4.3 SPREADSHEET ANALYSIS

Once both FORTRAN programs were applied to the raw data, two sets of reduced data files were generated, one set for the maximum heel strike accelerations and one set for the power spectrum densities. Both of these sets needed to be organized into a format which allowed for quick analysis with regard to many variables. Consequently, two spreadsheet templates were created.

The spreadsheet template for the maximum accelerations converted the raw data

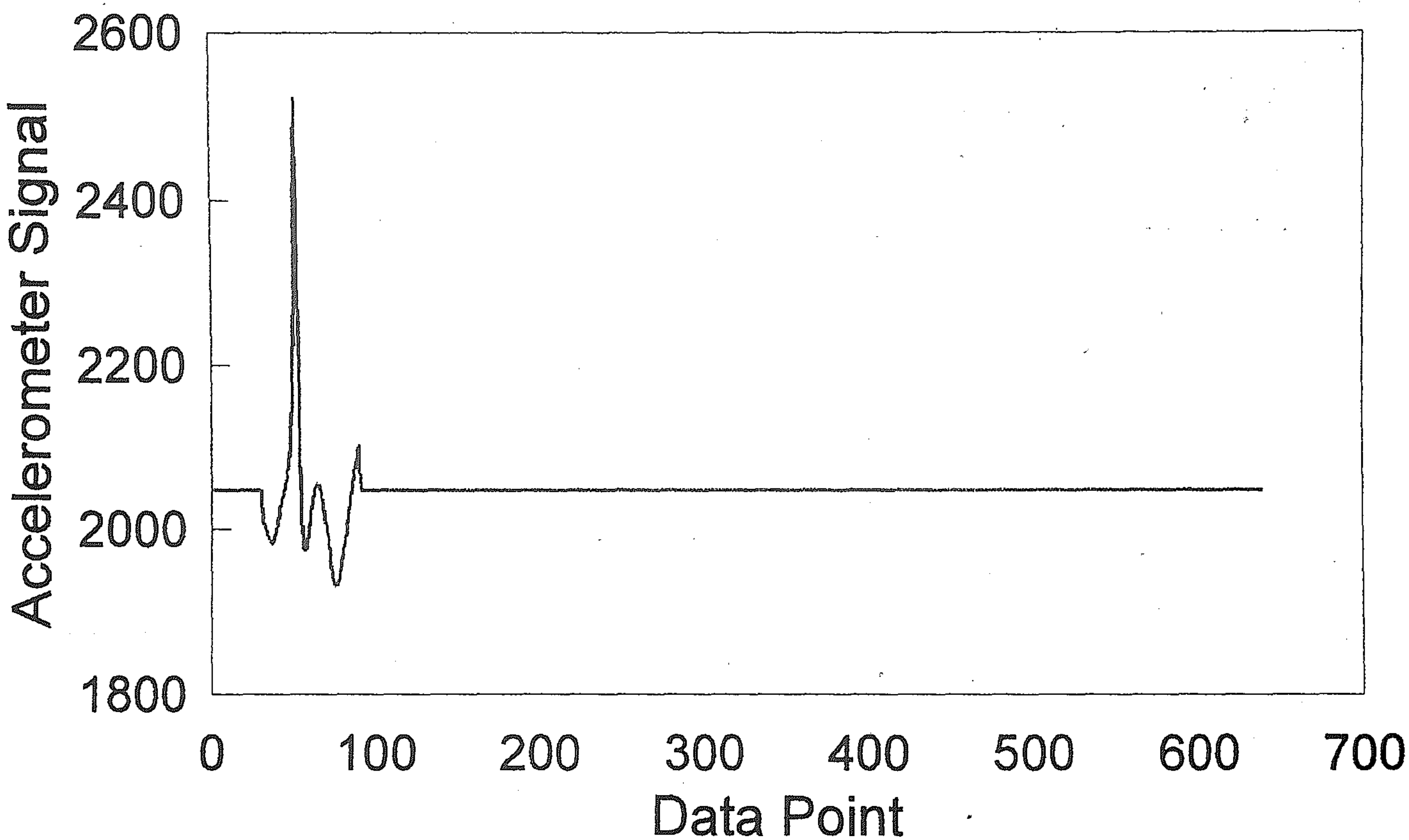


Figure 4.4 Data Array for SPCTRM Subroutine

voltage readings to accelerations and calculated the average and standard deviation for the maximum accelerations measured at the tibia and femur. A transfer function was defined as the ratio of tibia to femur accelerations. Average and standard deviations for the transfer function of each time trial were also calculated. All of the above averages and deviations were then plotted versus time to determine trends in the data.

To quantify the existence of a relationship between maximum accelerations and workout time, a linear regression analysis was applied to the plots to determine slope and correlation coefficient. A t-distribution test was performed on the slopes to determine the likelihood that the slopes were statistically variant from a zero slope.

The spreadsheet template for the power spectrum densities compiled the average of the ten power spectrums for each time trial. These average spectrums were then plotted against the appropriate frequencies and examined for trends with respect to time. The maximum value of power and its corresponding frequency were determined for each time trial. Changes in frequency or maximum power value were noted, and linear regression analysis was applied to the maximum power value versus time.

Numerical data accumulated from the above templates was also organized into a spreadsheet format. In addition, information gathered from the questionnaire was placed in the spreadsheet so that a single source of data for each session could be consulted or otherwise observed.

All standard and statistical analyses were performed by the spreadsheet (Lotus 1-2-3, v5.0) calculation functions.

5.0 RESULTS

5.1 SUBJECT BACKGROUND INFORMATION

A variety of data concerning the physical attributes and training regimen of the subjects was collected before each data acquisition session. Information included the subject's height, weight, age, gender, years of running experience, frequency of running workouts, typical distance and amount of time for each workout, previous injuries, current treatments, and any abnormal lasting results from injuries. This data was collected in anticipation of a relationship between fatigue and shock absorption so that any other correlations might also be observed.

A basic subject profile can be found in Table 5.1. A total of twenty-one different subjects were tested, with a few subjects volunteering for a second or third trial. This resulted in twenty-eight different data collection sessions, divided into eighteen male and ten female sessions. Only twenty-three data sets were able to be analyzed, due to particularly erratic data or, in one case, the subject not qualifying as a consistent recreational runner.

TABLE 5.1 Profile of Subjects Used in Study

	Male	Female	Total	Total Standard Deviation
Number of Samples	18	10	28	--
Average Height (m)	1.80	1.62	1.73	0.11
Average Weight (N)	777.9	516.0	684.7	167.8
Average Age (yrs)	29.7	28.0	29.1	8.9
Average Running Experience (yrs)	11.1	8.3	10.1	7.4

Histogram distribution graphs of height, weight, age and running experience time are shown in Figures 5.1, 5.2, 5.3 and 5.4.

Seven different subjects listed previous leg injuries, but all subjects considered themselves healed and able to return to normal activities. Some injuries listed included shin splints and stress fractures, chronic knee soreness, dislocated kneecaps and twisted ankles. None of the subjects were able to directly relate their injury to their running exercise routine. Only one subject was receiving treatment for injury. The treatment was a pair of customized orthotic shoe inserts which the subject had only begun using within the month prior to testing.

Of the seven subjects, four reported lasting effects from the injuries, ranging from periodic minor pain to stiffness. One subject was limited in the frequency of running which could be done in any one week. Another subject reported that his running times had improved after the injury.

5.2 EXPERIMENTAL RESULTS

5.2.1 TIME DOMAIN

Raw data accelerations from the tibia and femur accelerometers were examined for the local maximums at heel strike. The values of these accelerations were recorded and averaged for each time segment of each session. For each session, the average maximum accelerations were plotted against the corresponding time values. A sample plot of the maximum accelerations vs. time is shown in Figure 5.5. A linear regression analysis was performed, and the y-intercept estimate and error, the slope estimate and

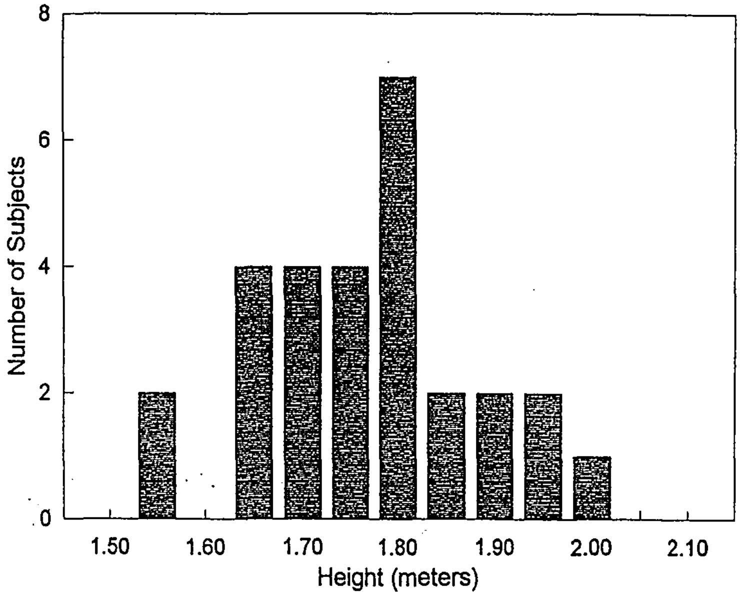


Figure 5.1 Height Distribution of Subjects

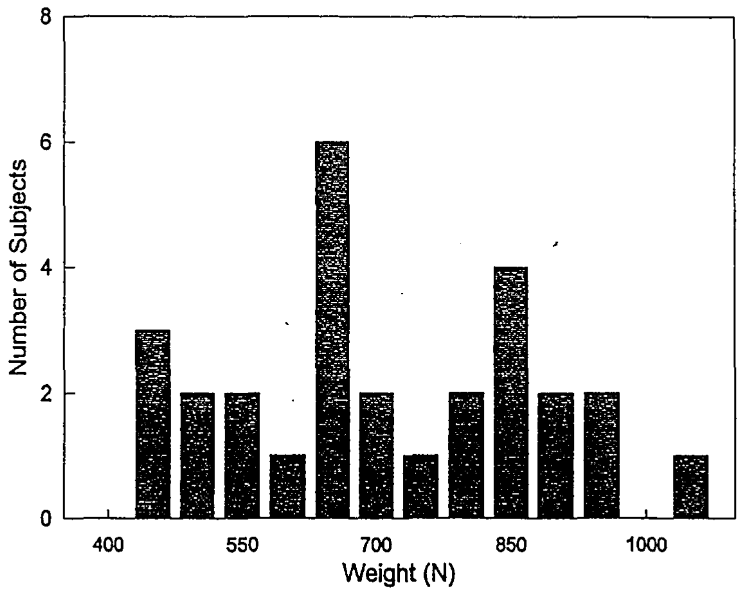


Figure 5.2 Weight Distribution of Subjects

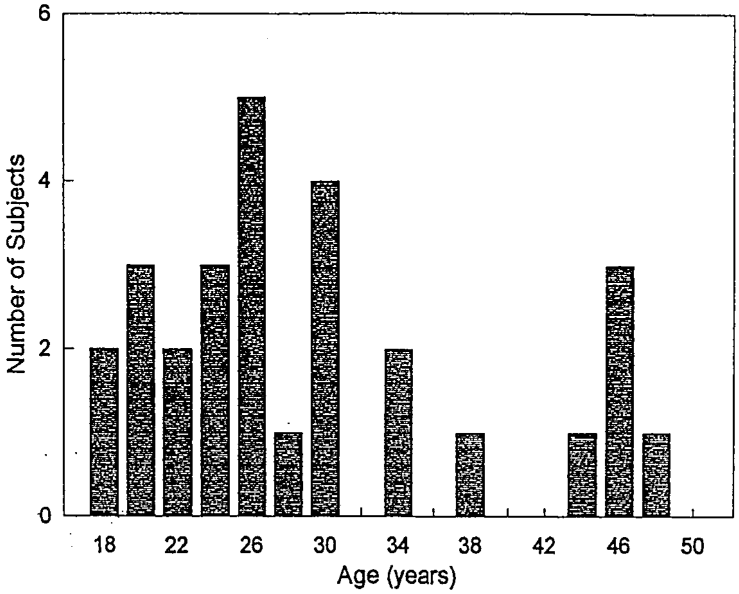


Figure 5.3 Age Distribution of Subjects

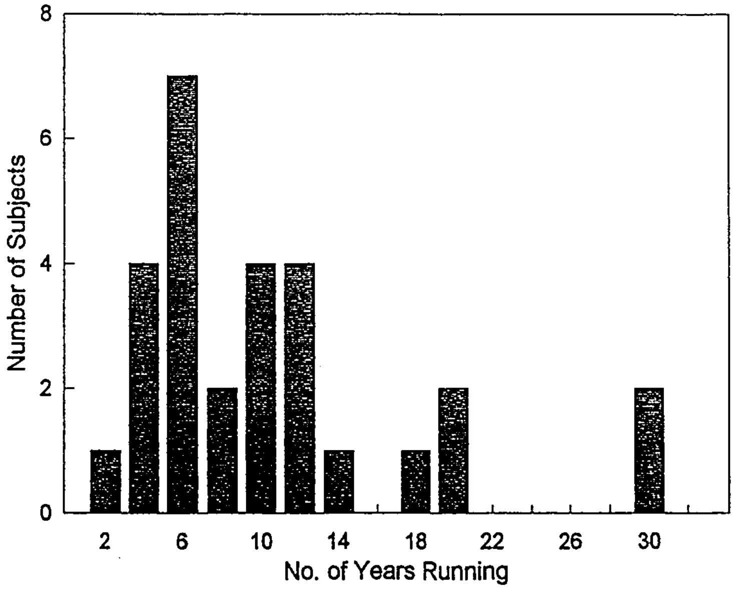


Figure 5.4 Running Experience Distribution of Subjects

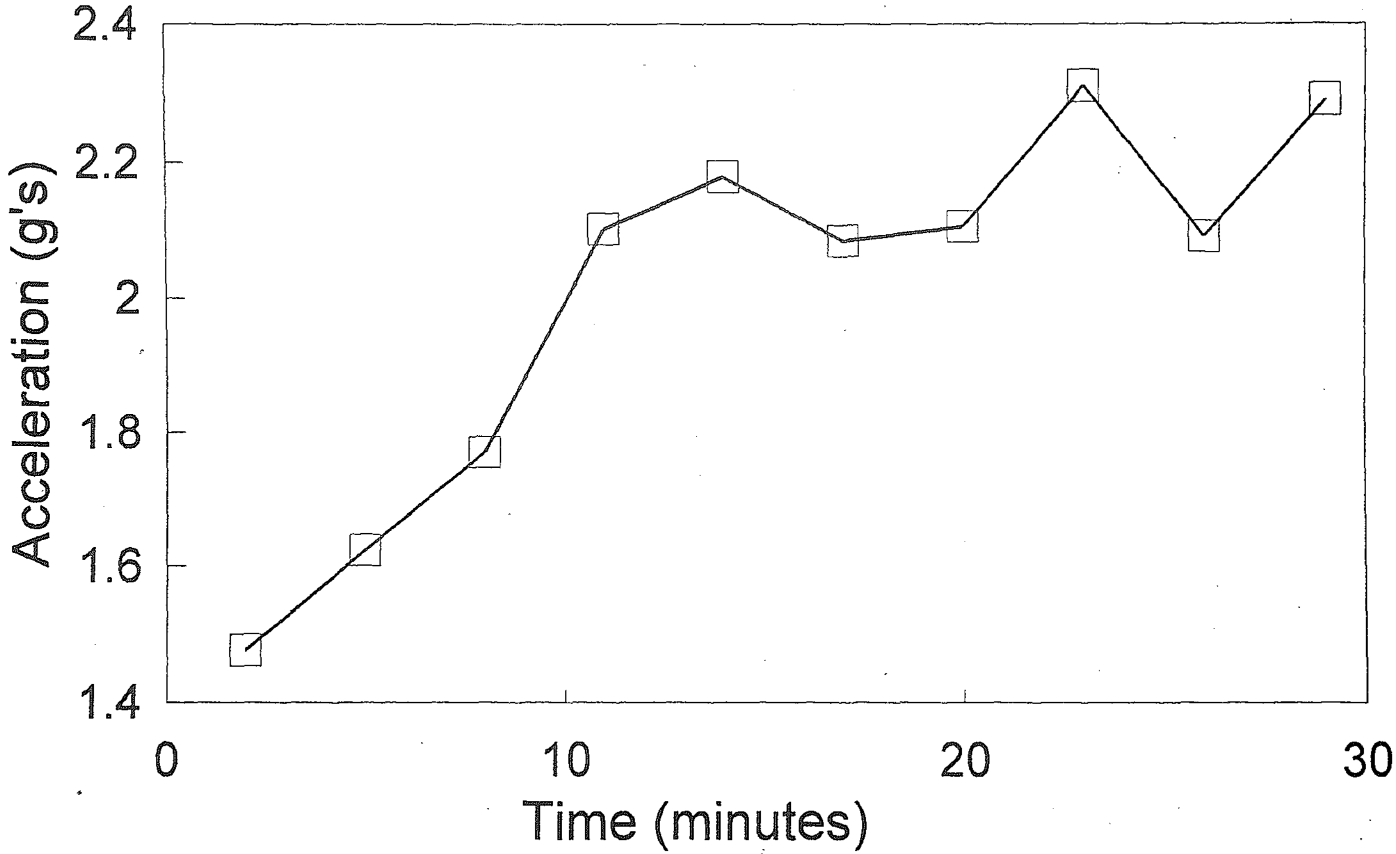


Figure 5.5 Example of Increasing Acceleration with Time

error, the correlation coefficient (r^2), the number of samples and the degrees of freedom were determined. Table 5.2 contains the slopes of the maximum accelerations vs. time and transfer function vs. time for each subject as gathered from the linear regression analysis. Greater than 65 percent of the subjects show a positive slope at the tibia and more than 69 percent show positive slopes for the femur, indicating that as the subject ran longer and became more fatigued, greater accelerations were being transmitted through the subject's body. Larger accelerations are a sign that shock absorption capacity is decreasing. A histogram distribution graph of both the tibia and femur slopes is provided in Figure 5.6 for clarity.

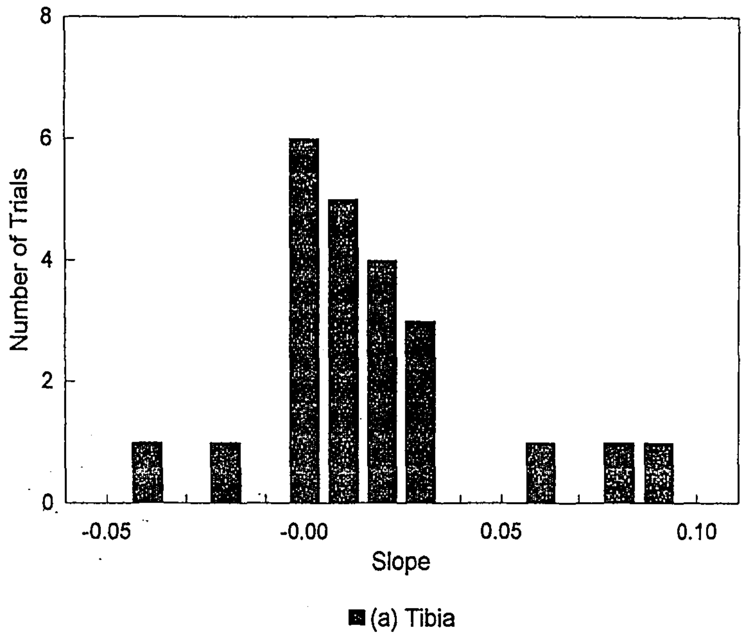
Table 5.2 includes the slopes of the transfer functions for each subject. The transfer function is an indication of the shock absorption characteristics of the area between the measuring points, in this case the knee. The transfer functions do not show any trend in slope for the group, therefore results about the effect of fatigue on shock absorption of the knee in general are inconclusive.

The correlation coefficient of the linear regression (r^2) and the t-distribution probability for each session are contained in Table 5.3. A correlation coefficient approaching a value of one indicates a strong linear trend in the data, while lower values indicate stronger non-linear trends. Values for r^2 averaged 0.4 for tibia and femur readings. These values show a rather weak linear tendency, although some individual subjects approached values of 0.85. Examples of a strong linear trial and weak linear trial can be found in Figure 5.7.

The t-distribution probability is a measure of the possibility that the calculated

TABLE 5.2 Maximum Acceleration vs. Time Slopes

		Maximum Acceleration vs. Time Slopes		
Subject	Session	Tibia	Femur	Transfer
1	1	-0.0420	-0.1380	0.0110
2	1	0.0170	0.0060	0.0370
4	1	-0.0270	-0.0050	-0.0130
4	2	0.0100	0.0050	-0.0002
6	1	0.0900	0.0300	0.0280
6	2	0.0220	0.0230	0.0010
6	3	0.0130	0.0180	0.0000
8	2	-0.0008	0.0000	-0.0080
9	1	0.0060	0.1140	0.1220
10	1	0.0060	0.0110	-0.0090
11	1	0.0230	0.0250	-0.0040
11	2	0.0710	0.0200	0.0140
12	1	0.0120	-0.0020	0.0210
13	1	0.0030	0.0090	-0.0180
14	1	-0.0001	0.0080	-0.0120
15	1	0.0270	0.0520	-0.0110
16	1	0.0090	0.0130	-0.0710
17	1	0.0060	0.0110	-0.0040
17	2	-0.0007	-0.0050	0.0040
18	1	-0.0080	-0.0100	1.4700
19	1	0.0580	-0.0070	0.0810
20	1	-0.0020	-0.0070	0.0210
21	1	-0.0072	0.0092	-0.0048
Average		0.0124	0.0078	0.0720
Std Dev		0.0281	0.0401	0.3002



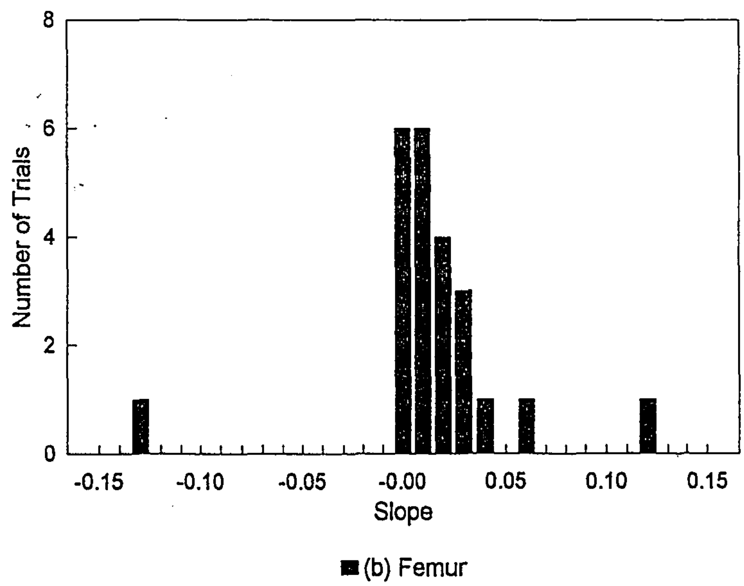
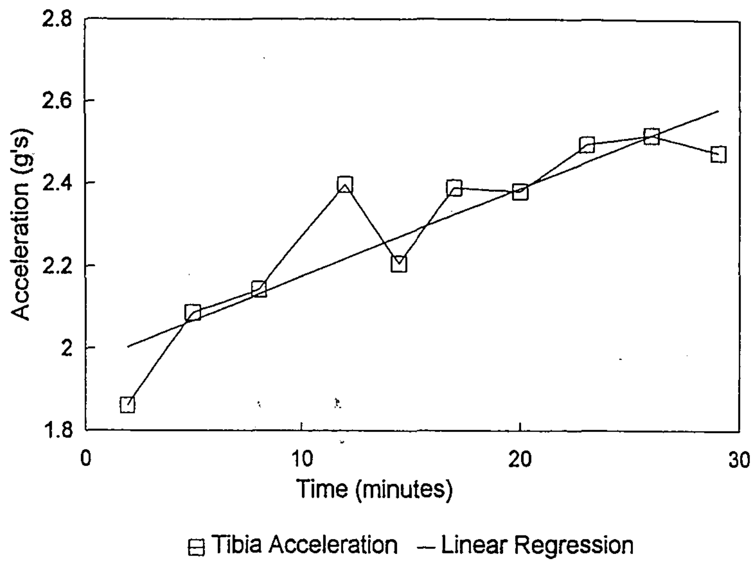


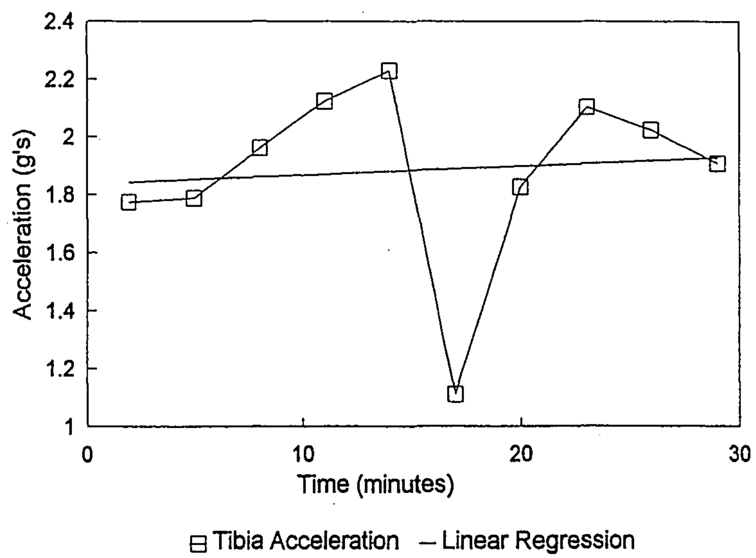
Figure 5.6 Slope Distribution for Maximum Acceleration vs. Time

TABLE 5.3 Correlation Coefficients and t-Distribution Probabilities

		Tibia		Femur		Trans.	
		Corr Coef	t-Dist Prob	Corr Coef	t-Dist Prob	Corr Coef	t-Dist Prob
Subject	Session						
1	1	0.05	62.2	0.24	26.9	0.60	4.0
2	1	0.45	11.8	0.18	28.4	0.38	13.8
4	1	0.64	1.5	0.15	35.6	0.13	35.6
5	1	0.59	2.4	0.69	1.0	0.55	3.1
6	3	0.25	17.7	0.36	8.6	0.09	44.3
7	1	0.32	8.1	0.58	0.6	0.32	5.5
8	1	0.72	0.2	0.75	0.1	0.36	0.4
8	2	0.54	1.7	0.15	34.7	0.50	1.7
8	3	0.01	80.9	0.30	11.0	0.15	26.4
9	1	0.00	98.1	0.34	8.1	0.42	4.3
10	1	0.74	0.2	0.91	0.0	0.00	92.8
11	1	0.50	1.7	0.75	0.3	0.71	0.2
11	2	0.39	8.1	0.36	5.9	0.17	21.9
12	1	0.53	2.9	0.85	0.0	0.40	5.1
13	1	0.54	2.3	0.14	35.1	0.82	0.1
14	1	0.03	70.1	0.22	20.4	0.22	20.4
15	1	0.82	0.1	0.53	2.1	0.06	34.7
16	1	0.01	70.0	0.00	99.4	0.07	44.7
17	1	0.00	18.8	0.14	27.8	0.52	1.7
17	2	0.52	3.2	0.56	1.7	0.00	95.4
18	1	0.80	0.1	0.79	0.1	0.46	4.8
20	1	0.05	51.0	0.22	13.4	0.28	8.1
21	1	0.14	29.9	0.33	13.4	0.00	97.4
Ave		0.38	23.6	0.42	16.3	0.31	24.6
Std Dev		0.28	30.6	0.26	21.7	0.23	30.7



(a) Strong Linear Correlation, Subject 6-2



(b) Weak Linear Correlation, Subject 13-1

Figure 5.7 Maximum Acceleration vs. Time Slope values are statistically significant (Press *et al.*, 1989). A t-distribution test assumes that the population from which the data came has a slope of zero and then calculates the probability of taking data which results in the slope which was actually found. Values for the data show that over half of the data samples taken at the tibia and femur have less than a 10 percent chance of coming from a population which would have a slope of zero. Over two-thirds of the data samples have less than a 20 percent chance of being taken from a zero-slope population. These results are a good indication that there is a relationship between fatigue and shock absorption capacity.

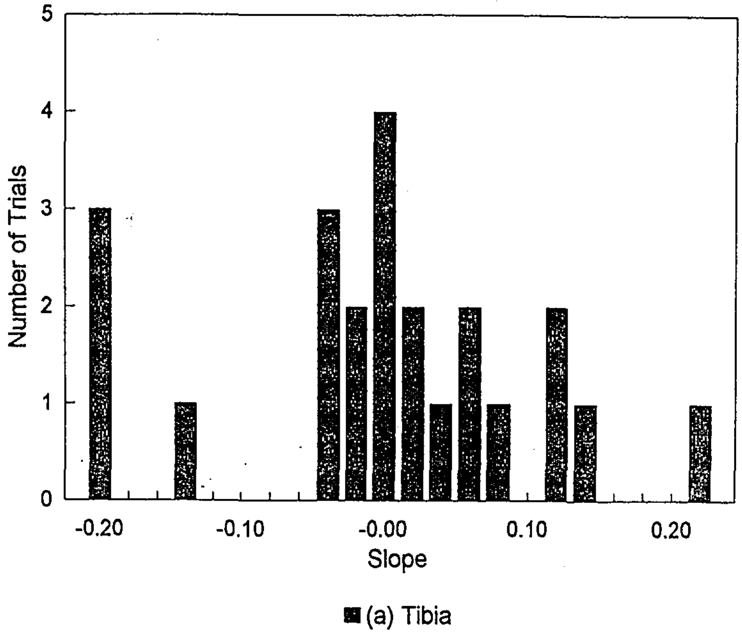
5.2.2 FREQUENCY DOMAIN

The slopes of the maximum power density value vs. time for each session are listed in Table 5.4. The slopes do not show any general trend in value. Therefore, the effect of fatigue on magnitude changes in power density is not conclusive. A histogram distribution graph is included as Figure 5.8 for clarity. The average value of r^2 for tibia and femur measurements is about 0.3, also showing a poor linear trend for the slopes.

Analysis of the data indicated that the frequency at which maximum power occurred did not change with time. Although frequency values of maximum power were not constant from subject to subject, the frequencies for each session were constant. This information indicates that power density frequencies do not have any dependence on fatigue.

TABLE 5.4 Maximum Power Density Value vs. Time Slopes

		Maximum Power Value vs. Time Slopes		
Subject	Session	Tibia	Femur	Transfer
1	1	0.080	-0.196	
2	1	-0.048	-0.632	-0.175
4	1	-0.010	-0.447	-1.970
5	1	-1.940	0.155	
6	3	0.136	145.000	-1.050
7	1	0.051	-0.059	-0.104
8	1	-0.018	0.175	2.710
8	2	0.005	0.139	-0.154
8	3	0.109	-0.270	-0.616
9	1	-0.011	-0.057	3.318
10	1	-0.220	0.467	-0.886
11	1	0.030	0.070	-0.121
11	2	-0.031	0.059	0.722
12	1	-0.383	-7.182	
13	1	-0.142	0.081	-3.850
14	1	0.016	0.338	-0.239
15	1	0.113	0.005	-0.209
16	1	-0.055	0.079	-0.234
17	1	-0.028	0.084	-0.428
17	2	0.042	-0.141	-0.305
18	1	0.219	0.005	0.663
20	1	-0.044	-0.047	-0.057
21	1	-0.016	0.238	1.194
Average		-0.093	5.994	-0.090
Std Dev		0.411	29.673	1.450



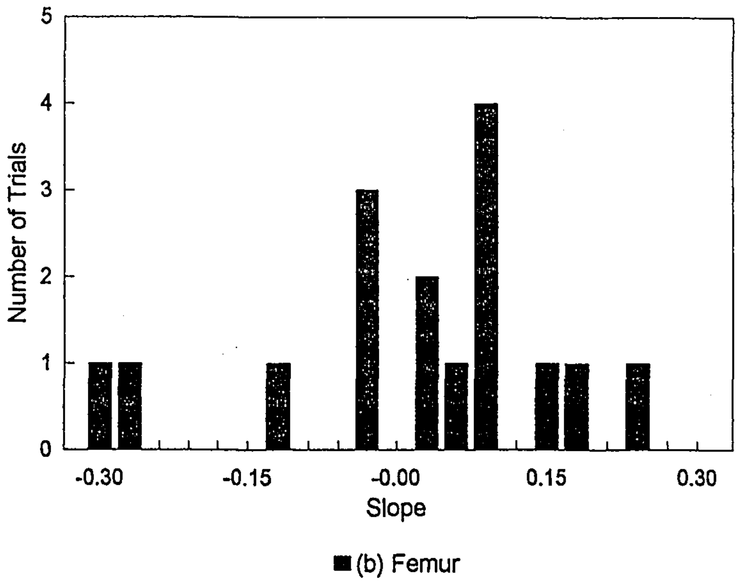


Figure 5.8 Slope Distribution for Maximum Power Density Value vs.Time

6.0 DISCUSSION OF RESULTS

6.0.1 TIME DOMAIN

The maximum accelerations experienced at heel strike by the subjects increased as the length of running time increased. With increasing running time, the amount of fatigue which each runner endured also increased. Therefore, a positive correlation existed between fatigue and heel strike shock waves. This was consistent with the hypothesis that fatigue reduces the body's capacity to perform physical work to attenuate impact shock waves.

As a first estimate, a linear relationship between fatigue and maximum heel strike acceleration was assumed. This assumption grossly simplified a number of independent and dependent factors in running, fatigue and shock absorption into one independent variable, time. Consequently, the results of the linear regression analysis performed on the data did not show a strong correlation coefficient, indicating a weak linear tendency. Also, the values of the slopes of maximum heel strike acceleration vs. time, while positive, were very small. Extremal values in the tibia were 9 and -4 percent, while the femur showed extremes of 11 and -14 percent (Tables 5.2 and 5.4). Because of the limited number of trials, the values for slope were confirmed to be statistically different from a zero slope condition by performing a t-distribution test. This test calculated the probability that the data was taken from a sample which had zero slope. The results of the t-distribution test (over two-thirds of the trials had less than 20 percent probability of being zero-slopes) confirmed the existence of the positive relationship between fatigue and shock absorption capacity.

Positive slopes in the relationship between peak acceleration at heel strike vs. time showed that fatigue tends to reduce the shock absorption capacity of the legs and most likely the body in general. There are many possible causes for this effect, and this study does not intend to attempt to identify all of them. A few possibilities which have been introduced in earlier sections are examined below.

The heel pad in the foot has been shown to be a major shock absorption factor (Valiant and Cavanagh, 1985). When the heel pad does degenerate for whatever reason, heel pain and other assorted injuries can result. Based on the construction of the pad, however, it is unlikely that fatigue from any one normal exercise session will affect its absorption capability. The fatty tissues and fibers, while displaying the viscoelastic quality of delay in returning to its original form after deformation (Valiant and Cavanagh, 1985), do not actively respond to any stimulus. The shock absorption capability seems only to be dependent on deformation.

Similarly, the role of shoes and running surfaces in shock absorption are not dependent on fatigue, but on the condition of the material used. The overall capacity of these factors is unlikely to change over the course of one session. However, shoes and surfaces may greatly contribute to the absorption factors. The cumulative effect of greater stress and load may be detrimental to the runner over the course of months or years (Nigg, 1985).

Another factor in the capacity of shock absorption is the construction of the joints between bones. In the joints which provide large ranges of motion, an elastic mesh of connective tissue provides stability, strength and shock absorption. Ligaments of a joint are not working actively to absorb shock and therefore probably do not show large changes in shock absorption capacity due to fatigue in any one particular running session. Tendons most likely are affected by fatigue, and consequently may lose shock absorption capacity in a short amount of time. During exercise, the cartilage material of a joint quickly engorges with synovial fluid, which probably aids in shock absorption. However, the condition of the cartilage may change as the body fatigues, dehydrates and requires fluid in other areas of the body. In this way, the capacity of shock absorption of cartilage may reduce over the course of a workout session. The cumulative effects of these factors may have a significant impact on the overall shock absorption capacity of the joints.

The factors which are probably most responsible for the reduction in shock absorption capacity over time are muscle fatigue and style of running. These factors are not independent of one another. As a muscle fatigues, the twitch characteristics decrease in magnitude and increase in time. Individual muscle fiber twitches begin to overlap and the full tensile capability as well as frequency response may be compromised. Also, in order to maintain work output, other surrounding muscles are recruited to assist in the task. This "recruitment" leads to involuntary changes in the execution of an activity. In runners, changing the execution of each stride may alter the striking characteristics of the heel, resulting in poorer shock absorption capacity.

In a force plate study (Dickinson, Cook and Leinhardt, 1985), an example of the effect of changing stride on shock absorption was presented when a subject consciously controlled his running style so as to eliminate the heel strike acceleration peak. Over the course of a running workout, the body subconsciously compensates running style for changes in conditions, such as grade, speed, pain and/or fatigue. Unfortunately, a subject under the effect of fatigue may not recognize subtle subconscious adjustments in running style. Without proper training, a runner may persist in exercising through fatigue and subject himself to increased risk of injury.

6.0.2 FREQUENCY DOMAIN

The maximum power density values of the heel strikes were not constant with respect to length of time run. However, no pattern was noticeable to the relationship. A linear regression performed on each trial did not produce a significant trend in slopes of maximum power vs. time (Figure 5.8). One would expect that since maximum accelerations increased with time, the maximum power transmitted at the dominant frequency would also increase with the time of running. However, closer examination of frequency mechanisms involved in running support the data.

At the moment of heel strike, the long bones of the leg can be thought of as long slender bars. In their examination of longitudinal impacts of slender bars, Timoshenko and Goodier (1982) found that the compressive stresses in the bars, which are proportional to the forces and the accelerations in the bars, are dependent only on the velocity at impact (or the initial conditions of impact), the modulus of elasticity and the density of the bar. These compressive stresses are reflected back and forth across the length of the bar, dependent on the end conditions of the bar. In the case of the bone, the amount of reflection of the compression waves is dependent on the joints at either end of the bone.

In the instants immediately following heel strike, the long bones of the leg can also be thought of as a slender beam under a varying axial force. If the beam is subjected to a small lateral impulse, as the bone is by the muscles, there will be a lateral frequency response which will be dependent on the modulus of elasticity, the moment of inertia, the effective length of the beam and the varying forces on the beam (Timoshenko and Gere, 1961). The effective length of the beam is dependent on the end conditions of the beam, which in the case of the bones is the joints at either end of the bone.

The measurement of heel strike shock waves in this experiment was done on bones, which do not change their material properties significantly over the course of a workout session and may be thought of as long slender rods. One area which may change over the course of a workout is the end conditions of the bones. The ends of bones are encased in elastic meshes of connective tissue, as mentioned previously. This connective tissue may vary the end conditions of the bone if the connective tissue is affected by fatigue. Also, the connective tissue may serve as a minor damping agent to the bone, causing a variability in maximum power value. End damping may also affect the power spectrum bandwidth. A qualitative analysis of the data suggests that this effect may be present and contribute to the variability of maximum power transmitted at the dominant frequency.

Another cause for the variability in maximum power value over time may be the initial conditions of the frequency response. With each step, the magnitude and orientation of the impulse caused by the collision of the foot with the ground changes. These changes are due to typical variations in stride caused by a number of factors, possibly including fatigue, and result in variable amounts of impact shock introduced into the system. Also, changes in orientation of the legs and body will affect the transmission of the shock wave through the bones of the body.

The change of the frequency of the maximum power value with respect to time was investigated as well. There was no significant change in the dominant power frequency over the time of the trial. This is expected because the mechanical properties of the bones do not change significantly over the course of the trial.

A third area examined on a qualitative basis only was the change of the shape of the power spectrum density curve with respect to time. In most subjects, the overall shape of the frequency curve did not change a great deal and did not show a general trend over time. An example of the shape of a frequency curve response over time can be seen in Figure 6.1. The variability of the power spectrum bandwidth may be attributed to the changes in end conditions of the bones and initial conditions of the shock wave.

A conclusion which can be reached from the above analysis is that the frequency characteristics are not dependent upon factors which fatigue. The results of this study indicate that a possible model for further theoretical analysis is that the long bones of the leg behave similar to a longitudinal beam with variable elastic end conditions subjected to an impulsive compressive force.

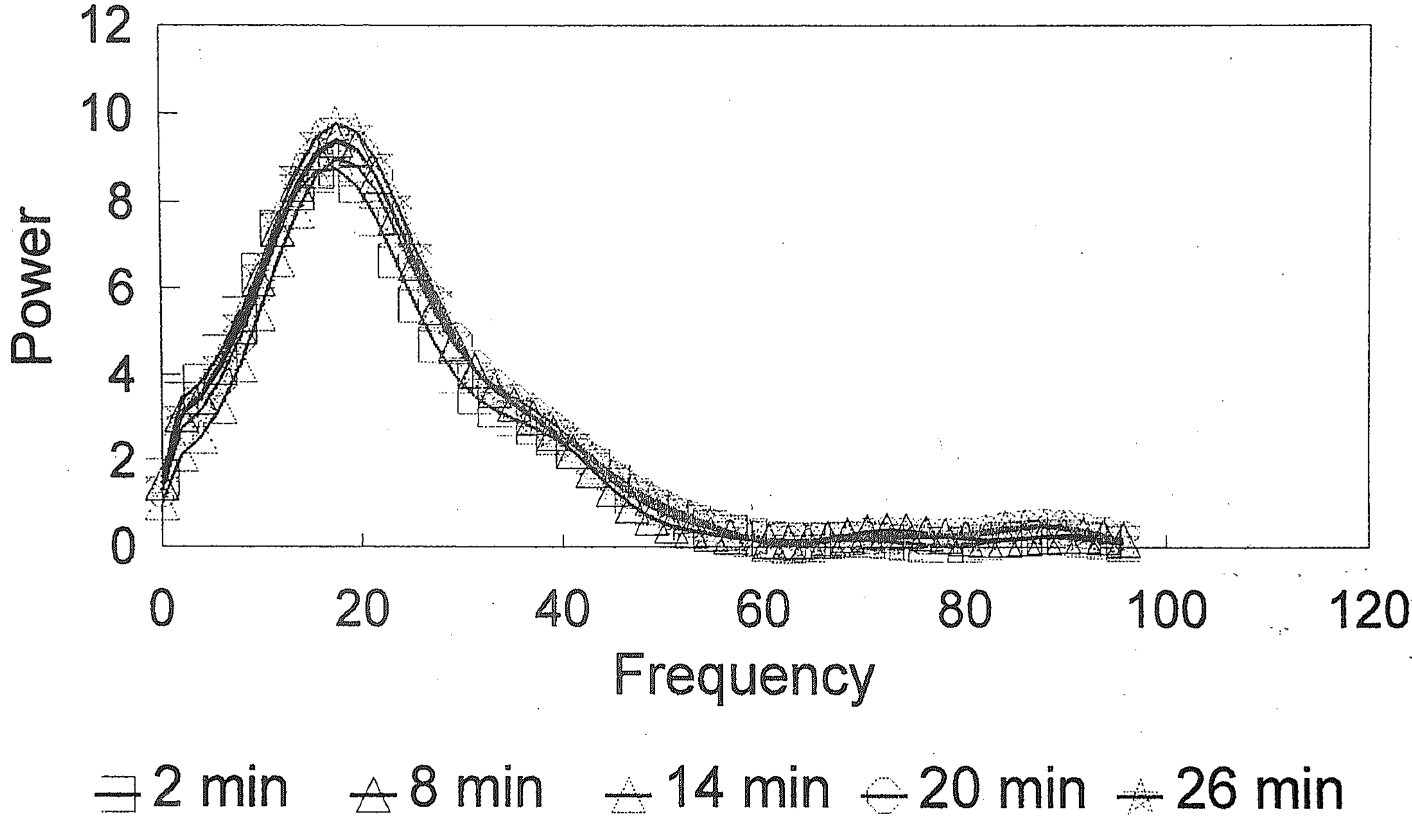


Figure 6.1 Shape of the Frequency Curve Over Time

6.1 SOURCES OF ERROR

In previous studies (Hennig and Lafortune, 1988) on the accuracy of externally mounted accelerometers, the results of skin mounted accelerometers were compared to results from internally bone mounted accelerometers. These studies found that the skin mounted accelerometers could produce readings which were up to 50 percent greater than the internal accelerometers. Convention is that the bone mounted readings are more accurate (Hennig and Lafortune, 1988). These differences can be attributed to the relative motion between skin and bone. During heel strike, the external accelerometer reacted less rapidly immediately following contact, then increased dramatically resulting in overshoot. In addition, the stretched skin stiffened to produce an elastic recoil (Hennig and Lafortune, 1988).

Two methods have been suggested for reducing this inherent error in measurement. One factor which reduced the error was the reduction of the mass of the accelerometer (Hennig and Lafortune, 1988). When 6 gram accelerometers were used, error was approximately 50 percent. Using 4.4 gram accelerometers in a linear spring and dashpot model, a 17-24 percent overestimation resulted (Valiant, McMahon and Frederick, 1987), and reducing the mass to 1.0 grams lowered the error to 8 percent (Gross and Nelson, 1988). In general, the skin mounted accelerometers were most accurate when the mass is small (Valiant, 1990).

The second method for improving accuracy for externally mounted accelerometers is to increase the tension in the strap holding the accelerometer to the leg (Valiant, McMahon and Frederick, 1987). In fact, best results were found when the strap was as tight as could be tolerated by the subjects. This method serves to reduce the relative motion between the skin and bone, resulting in a truer reading of the bone's accelerations. Some subjects were more heavily muscled than other subjects. This increased the thickness of soft tissue between the bone and accelerometer, thus increasing the error.

Another source of error in reading accelerations was orientation of the accelerometer along the length of the bone. Without proper orientation, the accelerometer measured only a portion of the acceleration, resulting in underestimation of true values. No attempts were made to quantify the amount of error given by the accelerometer due to misalignment because no accurate estimations of misalignment could be determined when mounting the accelerometers on the subjects. Care was taken to approximate the direction of the bone, but because of inability to observe the bones, especially the femur, accuracy may have been compromised.

In the mounting of the accelerometer to the subject, care was taken to fix the slack connections to some object. This action helped to reduce, but did not eliminate, motion imparted to the accelerometer by the leads. Such external motion would either supply some damping or some external vibration, which would damage the accuracy of the measurements. Again, no attempts were made to quantify the amount error introduced by this action.

Finally, there was no attempt to quantify the amount of fatigue each subject experienced in this experiment. As a result, there is no guarantee that all subjects, or for that matter, any subjects were fatigued at all. Upon completion of each data session, the experimenter briefly questioned the subject as to the degree of fatigue he was experiencing. Most subjects felt that the workout was normal for the subject. Few subjects considered themselves "exhausted" and most subjects recovered to an acceptable state of rest within ten minutes.

While there are ways to determine the state of fatigue a person is experiencing, the methods were beyond the skill, knowledge and equipment used in the experiment. As a consequence, the degree of fatigue of the subjects is the largest unknown source of error in the experiment.

No attempt was made to test subjects at similar times of the day. This could be significant in the magnitudes of shock which are absorbed.

6.2 DUPLICATE SUBJECTS DISCUSSION

In an effort to measure the effect of fatigue within subjects, some subjects were tested a second or third time on different dates. Subjects were selected based on willingness and availability of the subject. No preliminary or final calculations were given to the returning subjects prior to the duplicate testing. A total of five subjects were tested multiple times. Of those five, one subject had a heel strike signal which was too erratic to determine the heel strike peak acceleration in more than one trial.

A qualitative analysis of the raw data showed that each subject had a distinctive heel strike in that factors such as rate of acceleration increase and decay were consistent between trials. Figure 6.2 shows an example of the heel strike pattern of one subject on two different dates, with the data vertically shifted for clarity. This information

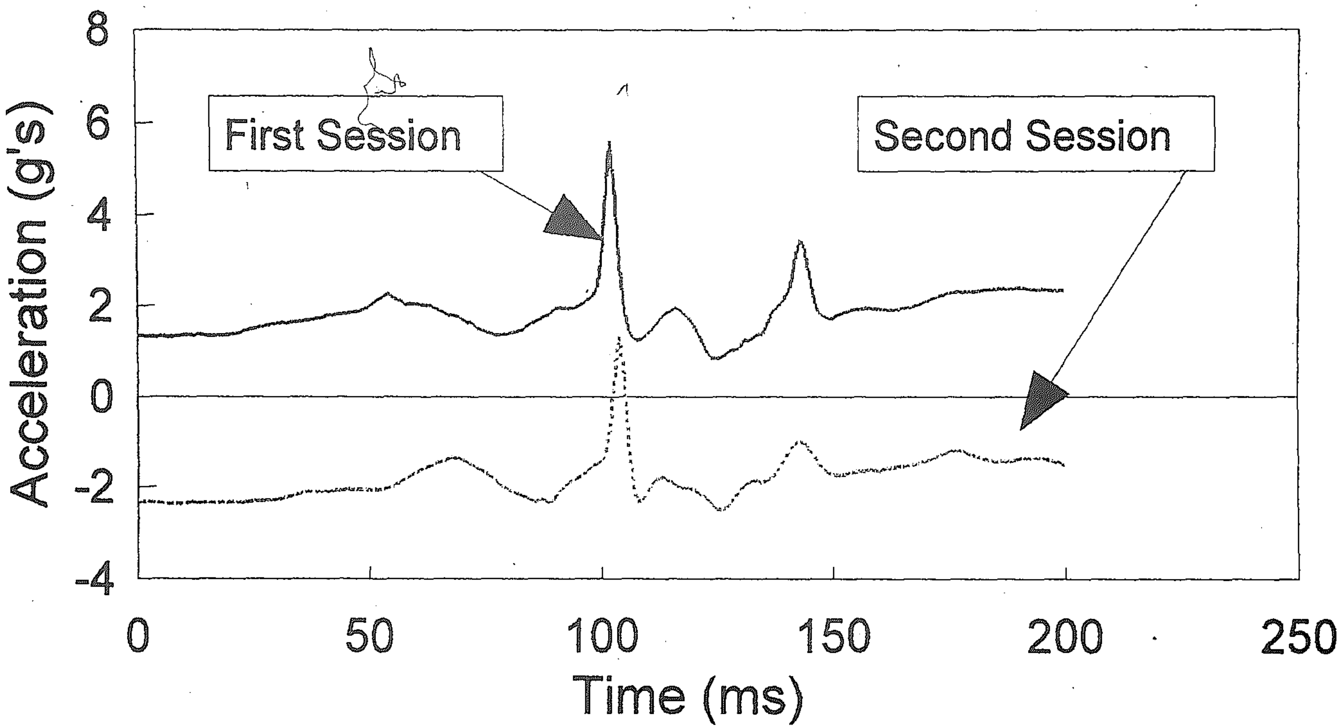


Figure 6.2 Heel Strike Patterns From One Subject on Two Different Dates

indicates that each individual had a consistent running style identifiable by heel strike pattern. Consequently, consistent shock absorption mechanisms can be assumed to be used by each subject from trial to trial.

Comparison of each subject's trials on a quantitative basis showed that the magnitudes of initial and subsequent accelerations were not similar between trials. Also, the rate of change of acceleration magnitude was different from trial to trial. Since shock absorption mechanisms did not change between trials (as evidenced by the similar heel strike acceleration patterns), the capacity of shock absorption varied from trial to trial. There could be many reasons for this variation, including but certainly not limited to changes in warm-up and/or stretching, amount of exertion prior to testing, changes in locations of accelerometers and not testing at similar times of the day.

Each subject had a consistent shape to their power spectrum density curve from one trial to the next (Figure 6.3). Also, the maximum power frequency was consistent for each subject from one trial to the next. However, the magnitudes of maximum power value were not similar for multiple trials. Timoshenko and Goodier (1982) showed that the compressive stresses in a slender bar are dependent only on the velocity at impact, the modulus of elasticity and the density of the bar. Also, Timoshenko and Gere (1961) showed that the lateral frequency response of a slender beam under a varying axial force subjected to a lateral impulse is dependent on the modulus of elasticity, the moment of inertia, the effective length of the beam and the varying forces on the beam. If one considers the bone to be an elastic beam and the joints elastic end

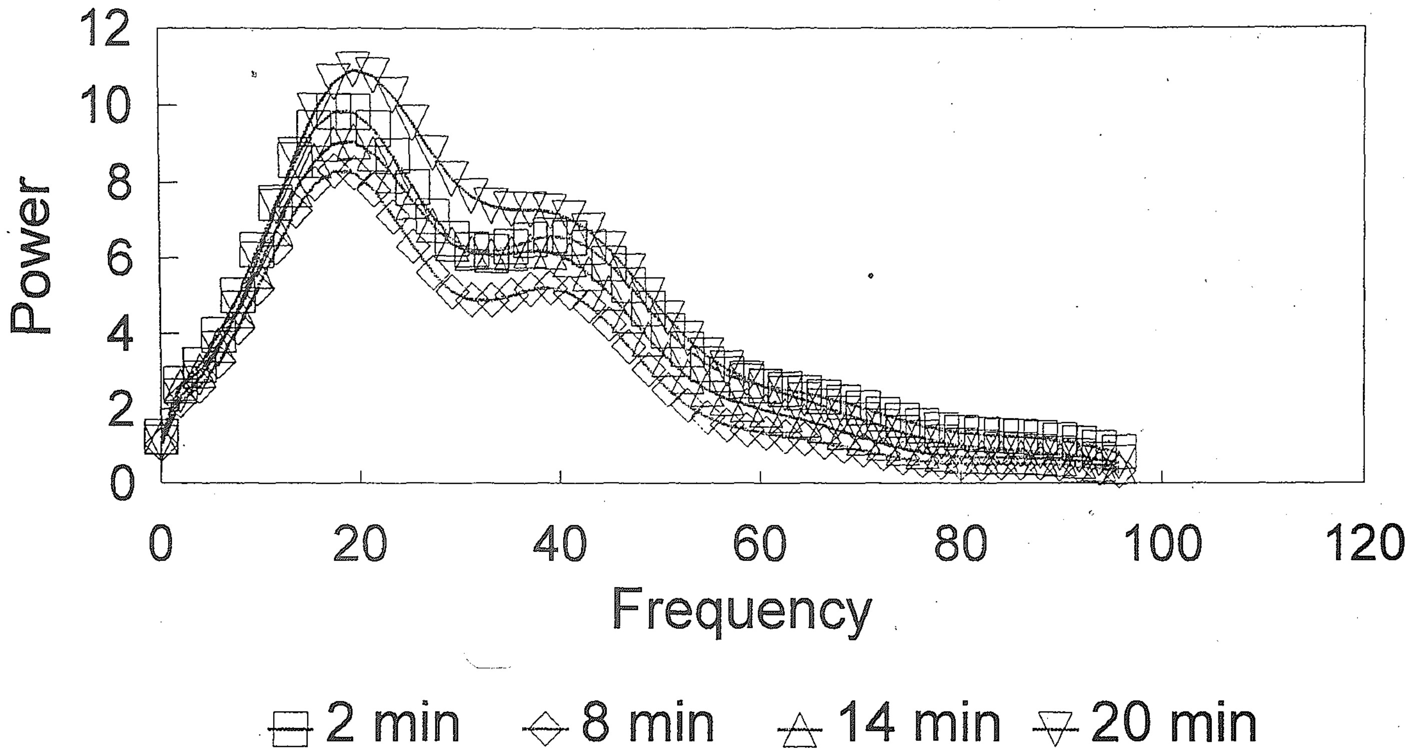


Figure 6.3 (a) Power Spectrum Density Curve
Subject 4, Session 1

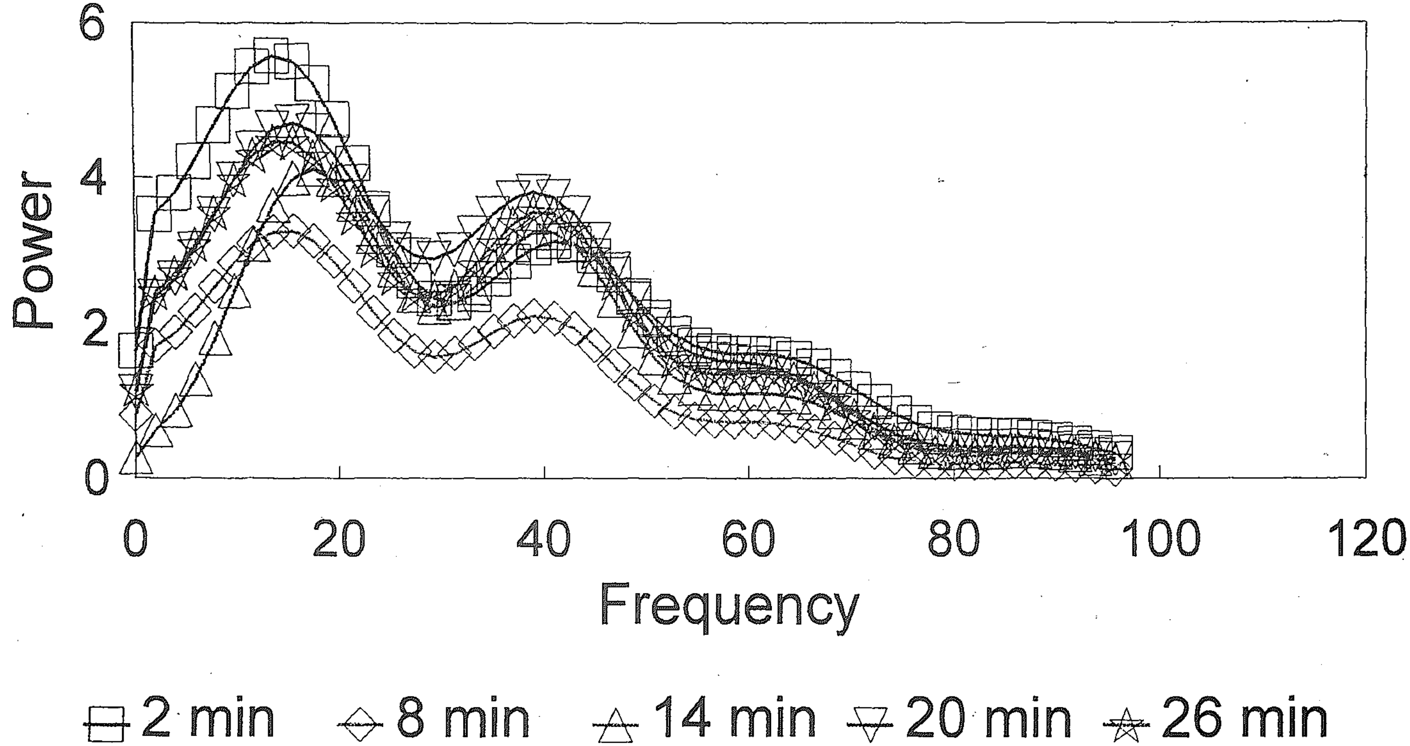


Figure 6.3 (b) Power Spectrum Density Curve
Subject 4, Session 2

conditions, the solution would result in a natural frequency and magnitude based on initial conditions and mechanical properties.

6.3 FURTHER CORRELATIONS

6.3.1 MAXIMUM ACCELERATION VS. HEIGHT

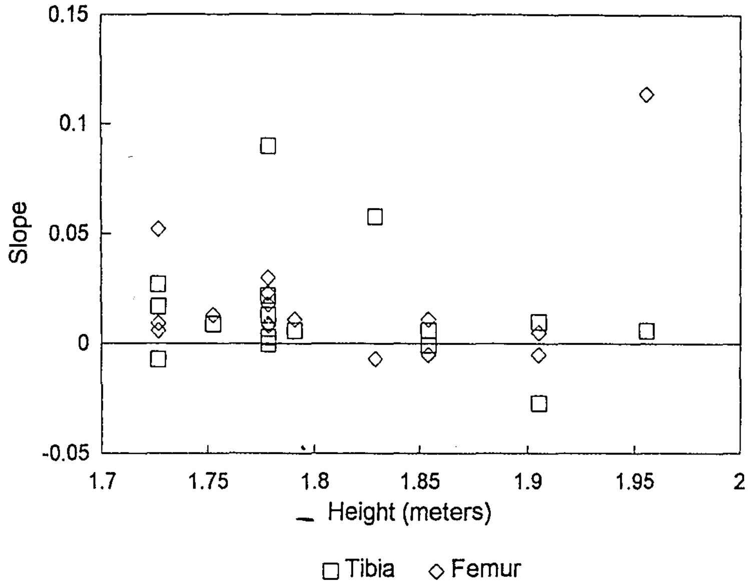
As the height of a subject increases, factors which may affect shock absorption take different values. Some of these factors may be leg length, heel pad thickness and center of mass. A comparison of the relationship between the effect of fatigue on shock absorption and subject height can be seen in Figure 6.4. Upon examination, slopes of maximum accelerations vs. time in males tend to become smaller as height increases. In contrast, slopes in females tend to be more constant or even slightly increase as height increases.

6.3.2 MAXIMUM ACCELERATION VS. WEIGHT

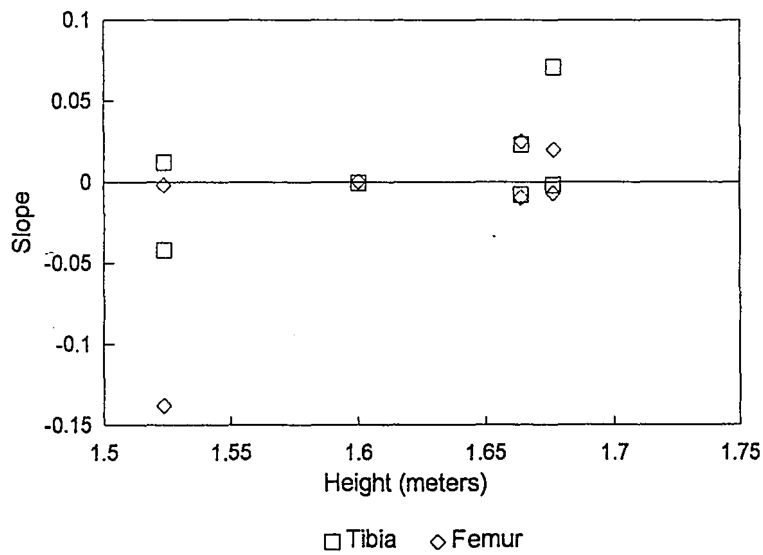
The effect of fatigue on shock absorption may also be affected by the weight of the subject. A comparison of this relationship can be seen in Figure 6.5. From observation, no apparent relationship exists between weight and effect of fatigue on shock absorption in either males or females.

6.3.3 MAXIMUM ACCELERATION VS. AGE

Because humans generally decrease in physical capacity as they age, another possible relationship may exist between fatigue, shock absorption, and age of the

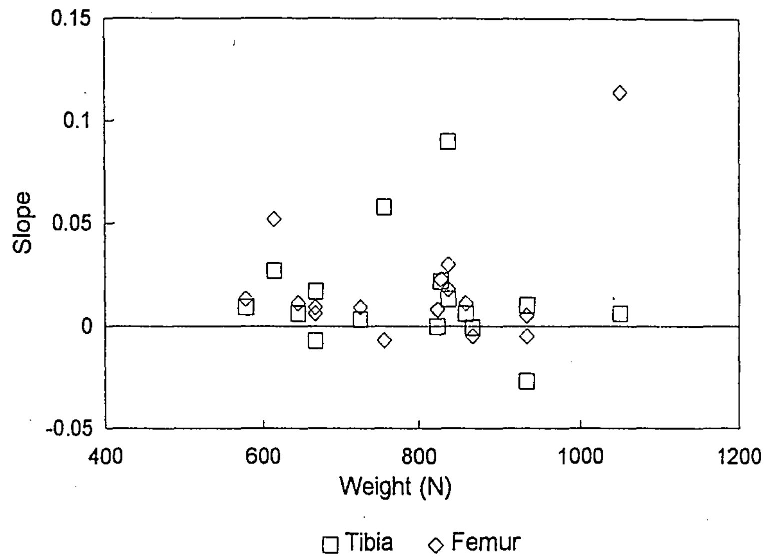


(a) Males

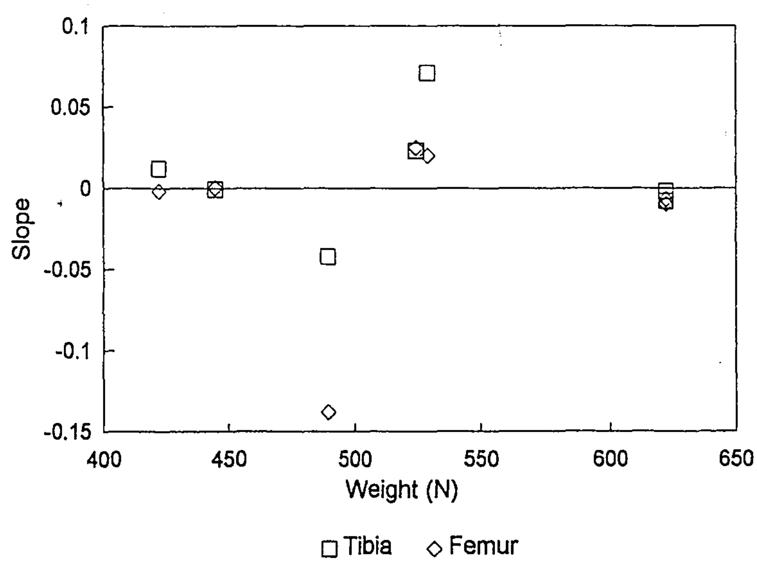


(b) Females

Figure 6.4 Maximum Acceleration Slopes vs. Height

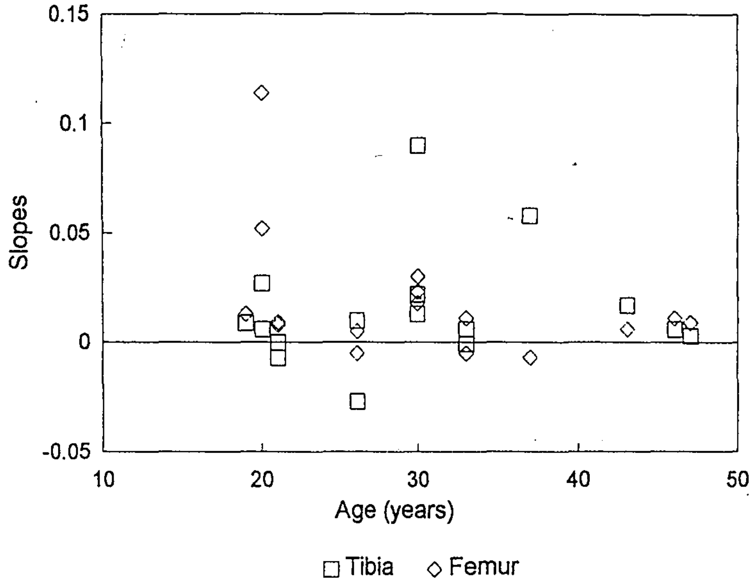


(a) Males

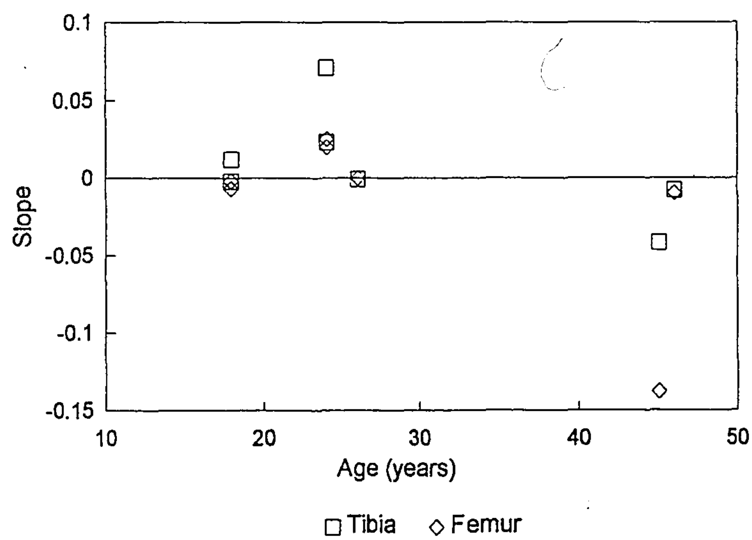


(b) Females

Figure 6.5 Maximum Acceleration Slopes vs. Weight



(a) Males



(b) Females

Figure 6.6 Maximum Acceleration Slopes vs. Age subject. A comparison of these factors is shown in Figure 6.6. Once again, no apparent relationship can be observed in either males or females.

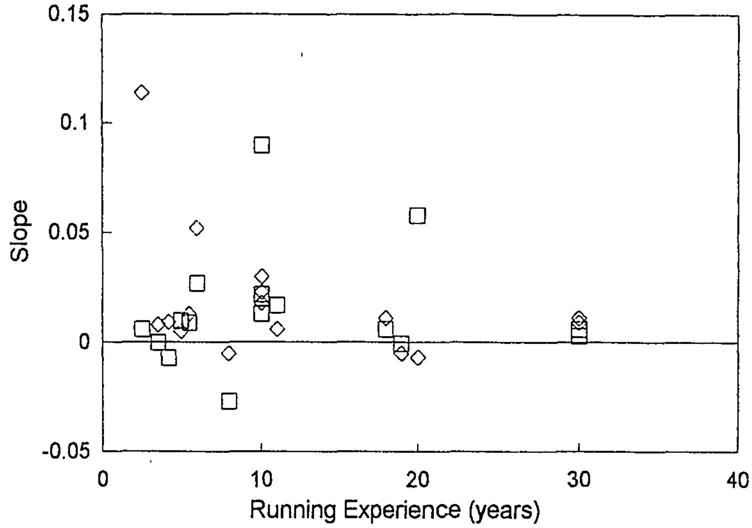
6.3.4 MAXIMUM ACCELERATION VS. RUNNING EXPERIENCE

Humans, as machines, have the unique quality in that movements or particular activities have a tendency to strengthen or otherwise enhance the performance of that movement or activity. Therefore, as subjects continue to run on a consistent basis, their response to fatigue and shock absorption would have a tendency to improve as experience increases.

In Figure 6.7, a plot of the slope of the maximum heel strike accelerations over time vs. the amount of years running experience for each subject is shown. Through examination, a trend can be identified in both males and females showing the slopes decrease as subjects accumulate running experience. This would indicate that consistent recreational runners tend to improve their resistance to fatigue in shock absorption capacity.

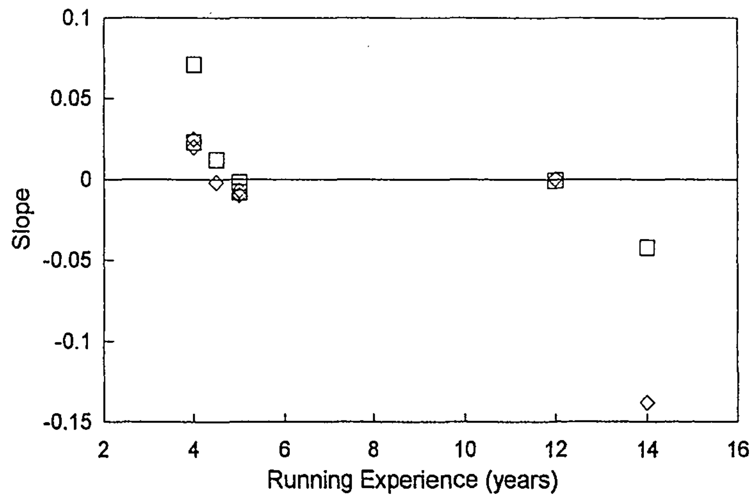
6.4 COMPARISON WITH FORCE PLATE FATIGUE STUDY

In an experiment by Dickinson, Cook, and Leinhardt (1985), subjects were tested for their heel strike transient response to fatigue. The testing area consisted of a measured runway leading to a force plate. Each subject practiced striking the plate with their left foot as they ran by at their normal running gait. Initially, subjects in a rested state struck the plate both in bare feet and with protective footwear. The



□ Tibia ◇ Femur

(a) Males



□ Tibia ◇ Femur

(b) Females

Figure 6.7 Maximum Acceleration Slopes vs. Running Experience subjects were then fatigued by treadmill running (with shoes) at a pace constant for all subjects. At 15 minute intervals up to 45 minutes, the subjects were taken off of the treadmill and tested on the runway in bare feet. Passes over the force plate were limited to three per interval to prevent recovery from fatigue during data taking. Also, all subjects were exposed to identical time and distance parameters while fatiguing instead of based on their typical running routine.

Some results found by Dickinson, Cook and Leinhardt (1985) are consistent with the results of this report. For example, Dickinson *et al.*. found that maximum of the initial force spike increased from 186 percent of body weight (BW) at the rested state to 203 percent BW for the averaged values of fatigued data. This is an indication of reduced shock absorption capacity due to fatigue.

Also, Dickinson *et al.* (1985) found that each subject consistently reached the maximum force at the same time during foot strike, although the time was different form individual to individual. This lead them to conclude that there was no change in vibrational frequency, assuming that the time-to-max-force is a valid indicator of frequency.

Other correlations done by Dickinson *et al.*. (1985) indicate a relationship between the height of the subject and the time of the maximum heel strike peak and a lack of a relationship between weight of the subject and magnitude of heel strike. These correlations are similar to the results found in this experiment.

6.5 IMPROVEMENTS AND AREAS OF FUTURE STUDY

6.5.1 IMPROVEMENTS

There are many improvements which can be made to this type of study. For example, there are more variables involved in shock absorption than were recorded during this experiment. These include but are not limited to: variability in the speed of the treadmill; location and orientation of the accelerometers; leg length; and foot striking style, either rear-foot, mid-foot, or fore-foot striker. Efforts need to be made to reduce the influence or account for the effect of these factors. Such improvements will enhance the correspondences between fatigue and shock absorption reported in this study.

Also, there are many more effective methods of measuring fatigue while running. Some of these can include a measure of resting heart rate vs. fatigued rate, VO_2 measurements, lactic acid build-up in the muscles, etc. Typically, these are more difficult to measure, but such methods would give a better indication of amount of fatigue than the method used in this study. This improvement would help to verify that fatigue is truly an independent variable with respect to shock absorption.

One of the most difficult matters in experiments involving humans is controlling factors which may influence the outcome of the experiment. Unknown outside variables, especially those of a psychological nature, may significantly affect the manner in which a subject responds to the desired independent variable and interfere in the acquisition of data. For example, just having the elastic bands, accelerometers and wires around the leg may cause some subjects to favor that leg in their running style resulting in poor data. Ideally, data acquired with human subjects should be as isolated from as many distractions as possible.

Other improvements can reduce variability in the "normal" condition of the subjects. Subjects may be required to run in bare feet to eliminate the effect of different shoe conditions. (On the other hand, these may cause subjects to alter their typical running style.) Subjects may be required to perform the same amount and type of stretching and warm-up routines. Subjects may be required not to work out for at least twenty-four hours prior to the testing, or possibly test within four hours of waking up. Also, subjects may be subjected to the same running routines for weeks prior to testing. All of these factors may contribute to the effect of fatigue on impact shock absorption.

6.5.2 AREAS OF FUTURE STUDY

One area which could be pursued is to develop an accurate and consistent method of mounting and remounting the accelerometers on any one subject. The mounting of the accelerometer is one of the most crucial elements in the collection of heel strike acceleration data. Location, orientation and tension in the strap all contribute to the error of the signal, resulting in errors larger than the changes due to fatigue in some subjects. Such inaccurate measurements make inter- as well as intra-subject comparisons invalid.

Another area which should be considered is a case-study style examination of one or more subjects, using a significant number of trials for each individual. This information could be used to determine the effect of changes in running routine on fatigue and shock absorption. Variables to be considered could be stretching and warm-up, the effect of training with variable speeds and grades, the effect of recovery time from previous running episodes and examination of various degrees of fatigue.

7.0 CONCLUSIONS

Because running is increasing in popularity as a sport and fitness activity, greater attention needs to be paid to areas which have a risk of injury associated with them. Increased magnitude of heel strike shock waves has been shown to be associated with an increase of injury in recreational runners. Before more effective reduction in shock wave magnitude can be possible, a clearer understanding of the factors which affect shock wave magnitude must be reached.

Fatigue has been shown to affect the capacity of the body to do physical work, including running. As fatigue affects the performance of the running activity, the capacity of shock absorption may also be affected by fatigue. In order to determine the relationship between fatigue and shock absorption, experiments were performed on a group of recreational runners. The shock wave accelerations being transmitted through the tibia and femur of the subjects were monitored as the subjects were fatigued by running on a treadmill. The shock wave accelerations increased over the course of the workout for over 65 percent of the subjects tested. The most likely reason for this increase in shock wave magnitude is the fatigue of the muscles affecting the technique of the runner, making him less effective in shock absorption capability. Statistical analysis showed that the data was very likely to have a non-zero slope with respect to time, even though the data had a poor linear tendency.

Power spectrum analysis was also performed on the data, but no significant change in frequency or power value with respect to fatigue were noticed. This result indicates that frequency response during running is most likely a function of the mechanical properties of the body and independent of fatigue.

Other correlations between the increasing magnitudes of heel strike shock waves and physical characteristics of the subjects showed a possible relationship between the height of the subject and increasing effect of fatigue. Also, a possible relationship can be seen between the amount of running experience and decreasing effect of fatigue.

In future studies involving heel strike accelerations, a consistent and accurate method of remounting the accelerometers will need to be developed. Without such technical advancements, conclusions about any information recorded after removing and replacing the accelerometers would be invalid. Future work with the effect of fatigue should explore the effect of changes in running routine on fatigue and shock absorption.

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APPENDIX A Lehigh University Human Subjects Policy Approval

Lehigh University

Office of Research and Sponsored Programs
telephone (610) 758-3020
fax (610) 758-5994

526 Brodhead Avenue
Bethlehem, Pennsylvania 18015-3046

MEMORANDUM November 10, 1994

TO: Michael Slusser

FROM: Linda F. Cope, Executive Secretary IRB

SUBJECT: Proposed Human Subjects Research: "Effect of Fatigue on Shock Absorption in Humans"

This is to advise you that your protocol has been approved by the Institutional Review Board for Human Subjects Research of the limited review category.

This approval is good for one year. If you wish to continue beyond that time, you must again submit your proposal to the committee for review. Also, if during the ensuing year you have made changes in your approved protocol, please submit these changes immediately to the committee for further review.

LFC:sab

cc: A. Voloshin

APPENDIX B Experimental Release Form

INFORMED CONSENT FORM

I,________________________ , hereby agree to participate as a subject in the research experiment on The Effect of Fatigue on Shock Absorption in Humans conducted by Michael Slusser under the supervision of Dr. Arkady Voloshin.

It has been explained to me that the purpose of the study is to learn the relationship between shock absorption in a rested state and shock absorption in a fatigued state.

The procedures which will be used in this experiment are as follows: the subject will run in a controlled method on a treadmill. During the session, small, lightweight accelerometers applied to the subject's leg with elastic bands will measure the impact shock as it travels through the subject's body.

My participation in the experiment will involve one running session of approximately
_____ minutes in length. I understand that I may end the session at any time and for any reason without penalty.

I understand that possible risks to me associated with the study are:

(a) the operation of the treadmill may fail and cause a sudden fall

(b) the wire connections from the accelerometers may become loose or tangled and cause a sudden fall

I understand that I may not receive any direct benefits from participating in this experiment, but participation may help to increase knowledge that may benefit others in the future.

I understand that any data or answers to questions will remain confidential with regard to my identity.

I understand that my participation is voluntary and that I am free to withdraw from this experiment at any time without jeopardizing my relationship with Lehigh University.

I have any questions about this study and what is expected of me in this experiment, I may call Michael Slusser at (610) 944-0645 (h) or (610) 758-4622 (LU).

Problems that may result from my participation in this experiment may be reported to Linda F. Cope, Office of Research and Sponsored Programs, Lehigh University, (610) 758-4861.

I have read and understand the foregoing information.

Date	Subject's Signature

I, the undersigned, have defined and fully explained the investigation to the above subject.

Date	Investigator's Signature

APPENDIX C Experimental Questionnaire

LEHIGH UNIVERSITY
EXPERIMENTAL QUESTIONNAIRE

The Effect of Fatigue On Shock Absorption in Humans

Subject No. Session No.

Subject's Sex M / F Height Weight Age

How long have you been running? years months

About how often do you run? times/week

About how many miles do you run per session? miles

About how long does it take for you to complete a session? minutes

Approximate your average speed for any average session (units?)

Have you ever had any significant leg injury?

if yes, what was it?

how long ago?

does it require any medical treatment now?

have you been cleared by a physician to run again?

are there any permanent effects (limp, reduced time, chronic pain)?

additional comments?

Are you in good health now?

Do you have **ANY** reason (physical or otherwise) why you may not want or be able to participate in this experiment?

if yes, why?

VITA

Michael J. Slusser was born in Reading, Pennsylvania on July 11, 1968 to Jack and Marie Slusser. He is a graduate of Governor Mifflin High School, Shillington, Pennsylvania and Widener University, Chester, Pennsylvania, where he received a Bachelor of Science degree, cum laude, in Mechanical Engineering in May, 1990.

After receiving his undergraduate degree and his Engineer-in-Training Certificate, he accepted a position with a Reading, PA-based consulting engineering firm. After about three years as a Project Engineer, he enrolled in the Lehigh University Graduate School Mechanical Engineering Department in August, 1993. While at Lehigh, he was employed under a Teaching Assistant scholarship.

In November, 1993, Michael and the former Marci Mayer were married. Upon graduation, he will begin a career with Air Products and Chemicals, Inc., Allentown, PA as a Career Development Program engineer.

END
OF
TITLE