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The Biomechanics of Leg Length Discrepancies

May 13, 1998

The Biomechanics of Leg Length Discrepancies

by

Michael D. Marotta

A Thesis
Presented to the Graduate and Research Committee
of Lehigh University
in Candidacy for the Degree of
Master of Science

in

The Department of Mechanical Engineering and Mechanics

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Chairperson of Department

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Abstract

The ultimate goal of the study of leg length discrepancies is the development of prosthetic and orthotic devices that equalize both the height difference as well as the transient shockwaves that travel up the tibia during the first 50 milliseconds of impact.

In order to achieve this goal, a preliminary study of the etiology, treatment, and effects of limb length inequality was performed. This condition has many causes, ranging from genetics to disease, paralysis, radiation damage, and trauma to the growth plates. The overall effect is the shortening, or in some cases, the lengthening, of one of the legs resulting in a tilt of the pelvis. This, in turn, causes a wide array of biomechanical problems over the person's lifetime, such as lower back and knee pain, sciatica, and osteoarthritis. Treatment for this condition depends on the severity. Shoe lifts are used for mild discrepancies, while surgical operations like bone resection and growth inhibition are used for larger ones.

The elements of the gait cycle were examined, as well as the relationships between walking velocity with cadence and impact acceleration using a non-invasive accelerometer strapped to the tibial tuberosity of the subject's leg. It was found that a direct relationship exists between walking velocity and cadence. In addition, a direct relationship between impact acceleration also exists, but this is partly due to the contribution of angular tibial motion to the total measured acceleration, which is on the order of 0.6 g at a walking speed of 2.2 m/sec.

Finally, a comparison between tibial strain measurements and impact acceleration

was made in order to establish a correlation between the two. Both non-invasive accelerometer measurements and invasive tibial strain gage measurements were used. In this study, a fairly strong negative correlation was found for both walking (-0.68) and running (-0.74).

Further work in this field will serve to verify the results obtained in this work, as well as establish a direct relationship between the degree of discrepancy and the acceleration due to impact.

Chapter 1: Introduction

The phenomenon of leg length discrepancy, otherwise known as leg length inequality or anisomelia, is a common finding in the majority of the world's population. It is defined as a bilateral asymmetry in the lower limbs which, in the majority of cases, is not significant, but can nonetheless contribute to numerous degenerative joint diseases and chronic pain over the course of a person's lifetime [2, 8, 9, 12, 14, 17, 18, 19, 26, 27, 28, 30, 32].

There are three primary categories of leg length inequality. Functional leg length inequality may be caused by a rotated pelvis or pelvic obliquity, caused by soft tissue shortening, joint contractures, or axial misalignments, such as scoliosis [2, 17, 18, 32]. As a result, the foot on the "short" side is externally rotated into the valgus position, and a collapsed arch is a common result. The posterior iliac spine is higher on the functionally shorter side, while lower on the longer side. Anatomical leg length inequality occurs when there is an actual difference in the overall lengths of the tibia and femur such that one leg is literally shorter than the other. As a result of this, the anterior and posterior iliac crests are lower on the shorter side, causing the spinal column to compensate, often causing scoliosis. In some cases, foot positioning can serve to compensate for the difference as well. Ankle pronation and collapsed arches on the longer side can occur, causing a functional shortening of the longer side to attempt to make up for some of the discrepancy. "Environmental" leg length inequality is caused by outside, man-made factors, now that running has become a popular activity in today's society. While running, the part of the road closer to the sidewalk is constantly lower than the surface closer to the middle of the road. As a result, an uneven running surface is presented to the runner, and may cause a temporary discrepancy. The treatment for this is simple; runners merely have to alternate the direction in which they travel [2, 17, 18, 32].

1.1 Etiology

The etiological factors associated with leg length discrepancies are diverse, most of which affect the person during the growth process. The vast majority of cases is congenital - caused by the individual's genetics, which, in some cases, causes one leg to be naturally longer or shorter than the other [18, 19, 27, 28, 30]. The majority of those with leg length discrepancies have a shorter left leg. It is suggested that this is a direct result of the normal position of the child *in utero*, in which the left side of the fetus is pressed against the vertebrae of the mother. In addition, approximately two thirds of third trimester fetuses have their left leg crossed over their right, increasing stress in the hip, knee, and epiphyseal growth areas [2]. Increased stress in the epiphyseal growth plates are known to affect growth [10, 18, 31].

The remainder of cases are caused by acquired means, the most common of which is physeal injury. The physis is a cartilaginous growth area for bones, located on the proximal and distal ends of both the femur and the tibia (Figure 1.1). Each area has a different growth rate. During childhood, the inner layer of the physis ossifies, while the outer layer grows more cartilage. This process continues until adulthood, when the physis fuses with the rest of the bone, causing growth arrest. Irreparable damage to any of the physeal areas during the growth process can cause premature growth arrest in that physis, stunting the growth of that leg. Most injuries of this type are caused by either severe burns or fractures along the physis, by either breakage or compression. While there are a number of different types of fractures and breaks that can occur in this area, not all of them have the potential to cause growth arrest. Only those which cross the physis are dangerous, since during the healing process they have a tendency to form a bony bridge across the area (Figure 1.2). Growth arrest will occur in these cases because the bridge ties the growth region to the rest of the bone [18, 19, 28, 30].

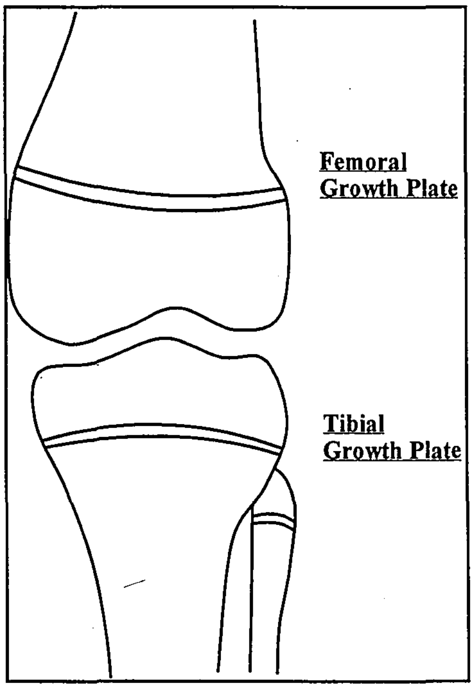

Figure 1.1 - Growth plates of the femur and tibia near the knee joint

Some types of fractures such as minimally displaced proximal fractures and

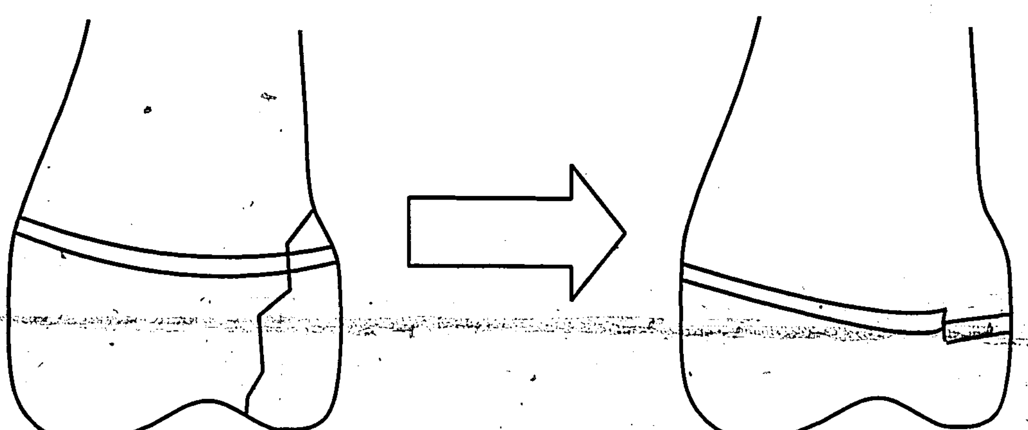

Figure 1.2 - A break across the growth plate that will lead to growth arrest.

femoral fractures in young children can cause the opposite effect: growth stimulation. It is due to the increased blood flow to the area of injury as a natural part of the healing process. This effect has been known to last for up to two years after the time of injury [19].

While trauma to the growth plate can cause physeal arrest, a number of other conditions, including disease and paralysis, can cause epiphyseal growth dysfunction. An infection of the epiphysis or areas adjacent to it, such as septic arthritis, osteomyelitis about the femur and tibia, or tuberculosis in the hip, knee, and foot can cause the destruction of physeal cells or the development of a bony bridge across the bone and physis, causing physeal arrest, or in some cases destroying the physis altogether [18, 19, 30]. In some cases of chronic osteomyelitis, however, it has been observed that the increased blood flow due to infection can cause increased growth of that leg, presumably because of the increased vascular activity [18, 19, 30].

Tumors, such as osteochondroma, giant cell tumors, Ollier's disease, and Recklinghausen have also been known to cause growth problems if present in the physeal areas. The destruction of the physis can be caused by the invasion of a tumor. If present in the cartilage cells of the growth plate, growth potential can actually be stolen away from the bone. Some tumors, especially vascular malformations that involve large portions of the limb, produce increased growth in all areas of the leg. Such vascular tumors include hemangiomatosis and Klippel - Trenaunay - Weber syndrome; other growth inducing tumors include neurofibromatosis, fibrous displasia, and Wilms's tumor. In addition, the radiation used to treat these tumors in some cases have the potential to kill the healthy osteocytes and blood vessels in the bone, halting growth in that region, usually taking years to repair [18, 19].

Muscle paralysis is yet another cause of leg length discrepancy cases, due to the prolonged amount of immobilization involved. In addition to that caused by trauma, poliomyelitis - a viral disease which affects the nervous system causing paralysis and atrophy in the legs - has long been known to contribute to the occurrence of leg length discrepancies in young children as well as adults. Bone growth depends on compression forces created by standing, walking, and running, which are severely diminished, if not absent altogether in paralysis cases. This lack of activity has the potential to cause some of the same growth difficulties described earlier [18, 19, 28].

1.2 Measurement

When a leg length discrepancy is suspected, a measurement of its degree of severity is made. Certain degrees of accuracy can be obtained, depending upon what type of measurement is used. The clinical method involves measuring the distance from the anterior superior iliac spine to the tip of the medial malleolis using a tape measure [2, 17, 19, 27, 28]. In these cases, finding inequalities of less than 1.25 cm (1/2") is very difficult because of the presence of skin and fat over these bony prominences. Another approach to this is to measure the distance form the anterior superior iliac spine to the floor, thus eliminating one of the "fuzzy" endpoints on the body [27]. In other cases, the patient is examined at arms length, and the height of the pelvic brims are compared [2, 28]. While these methods may not provide the most precise measurements, it is an inexpensive and easy way to get a rough picture of the degree of discrepancy.

In the cases in which the differences are more subtle, the radiological method is far more reliable, albeit slightly more hazardous to one's health. The patient stands with the feet 15 to 20 cm apart while X-rays of the pelvis are taken, with the beam focused on the femoral heads. The differences in their heights, in most cases, indicate a leg length discrepancy. This assumes, however, that the discrepancy is anatomic and not caused by joint contractures. The accuracy of such a measurement is within 3 mm. This is the best, most reliable method when clinical measurements would not provide enough information [17, 19, 28].

1.3 Effects

Leg length inequality can cause a wide variety of physical and biomechanical problems during any and all stages of life. Symptoms range from mild and unnoticed to severe and debilitating, depending on the severity of discrepancy. The most commonly reported problems are: back and knee pain, pelvic tilt, sciatica, stress fractures, osteoarthritis, gait asymmetry, and bone deformities [2, 8, 9, 12, 14, 17, 18, 19, 27, 28, 32].

One of the most noticeable effects of anisomelia is an asymmetry in gait, present even in people with inequalities as little as 1 cm. For the purpose of energy conservation, people with limb length inequalities must physically step down onto the shorter limb and vault over the longer one during the gait cycle. As a direct result, the time between the right - left and left - right heel strikes is not the same. Research has shown that this time difference is indeed a direct function of the degree of discrepancy in the subject. As the inequality increases, so does the asymmetry in heel strike time [14, 19]. Some patients who have had corrective surgery show some of the same asymmetry in some cases. This can be attributed to the fact that while their overall discrepancies have been corrected, their individual femoral and tibial lengths remain different, placing the knee at different levels. The leg with the knee lower on the leg naturally swings slightly faster than its counterpart, thus causing either a faster swing time for that component of gait, or an increased effort on the patient's part to equalize swing times [19].

Patients with leg length inequalities may be subject to an abnormally high incidence of stress fractures. As the impact of walking is attenuated in the legs, microfractures form. These fractures are constantly created and healed in a continuous cycle which is stable below a certain threshold. For some severe discrepancies, the greater force acting on the longer leg exceeds this threshold, thus increasing the risk of the development of stress fractures on that side [14, 17].

Pelvic tilt is a fairly common finding in cases of leg length inequalities, which unfortunately gives rise to a vast array of associative symptoms ranging from mild to severe. Like gait inequality, pelvic tilt is a direct function of discrepancy. In some cases, this tilt can cause a functional scoliosis, which is concave to the side of the longer limb. This is a compensatory mechanism that helps maintain the center of gravity. However, in approximately 1/3 of the cases, the curve is opposite to the direction of compensation, giving rise to the belief that some forms of scoliosis that develop may be a result of locomotion, like walking and running, and not just standing [19].

Anisomelia induced scoliosis, in turn, has been reported as the cause of the occurrence of sciatica and possibly lower back pain. Sciatica commonly occurs on the concave side of the scoliotic spine, in which the pressure caused by the curvature of the spine can put pressure on the dorasl sensory nerve root [17]. The pelvic tilt and spinal curve also places a constant stress on the muscles, tissues, and ligaments of the lower back, hip, and knees, causing asymmetric muscle tension in those areas. While there is conflicting evidence at this point as to whether or not leg length discrepancies are a direct cause of lower back, knee, and hip pain, it would serve to explain cases in which they are present [2, 9, 17, 18, 19, 28, 32].

The most debilitating effect of limb length inequality is the osteoarthritis that can develop over the course of the patient's lifetime due to uneven force distribution on the legs. Osteoarthritis is a degenerative joint disease characterized by the thinning and eventual disappearance of the articular cartilage. The commonly occurring pelvic tilt causes a shift of the center of gravity onto the side of the longer limb, thereby reducing the contact area of the joint surface on that side (Figure 1.3). This phenomenon is called "uncovering" [18, 19]. If this occurs, the stress per unit area increases on that joint, causing hip pain and increased wear and deterioration of the cartilaginous joint surfaces. This is theorized to be the precursor to the degenerative changes that cause osteoarthritis

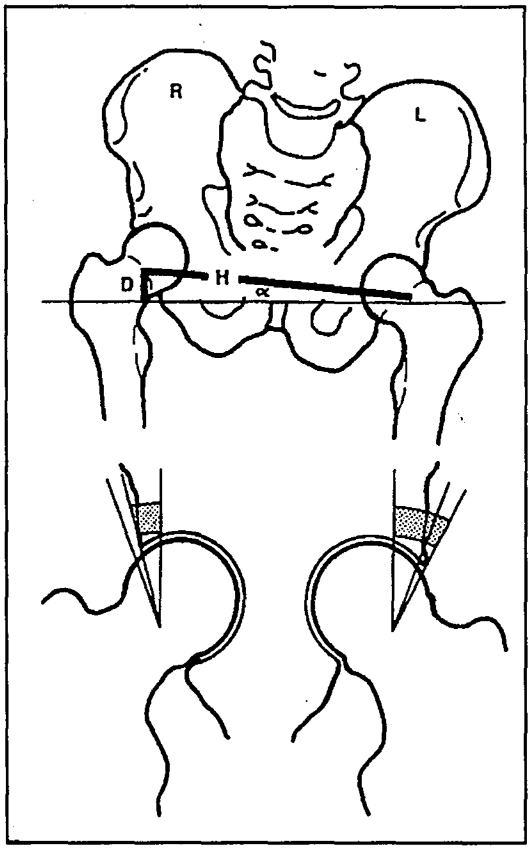

Figure 1.3 - Uncovering of the hip joints (reproduced from [20])

[2, 9, 17, 28, 30, 32]. In addition, the loss of cartilage over the course of time has been hypothesized to increase the difference in leg lengths above what was there previously, thus worsening the situation. The unequal stresses on the long side can also cause cartilage degeneration in the knee joints, thus creating the same conditions for osteoarthritis there as well [9, 12, 19]. This may also be one of the causes of knee pain in some subjects.

The formation of minor bone deformities in the spine have been directly correlated with pelvic obliquity and functional scoliosis. These are usually characterized by concavities in the lower end plates of vertebrae located in the upper lumbar region. These concavities are asymmetrically oriented towards the convex side of the curved spine. Lumbar deformities are more often found in patients with slightly more significant discrepancies of above 9 mm [8].

From the preceding information, it is relatively easy to observe the cascading trend of symptoms for people afflicted with anisomelia. Connected with the pelvic tilt which serves to compensate for the difference, there is a host of other biomechanical problems that may arise, in some cases worsening the situation. One can also easily see why there would be numerous methods of treating or eliminating this condition.

1.4 Treatment

The ultimate goal of treating leg length discrepancies is their equality at full maturity. To accomplish this, doctors must exercise a certain amount of timing in any treatment method that is chosen. In some cases, however, equalizing the leg lengths would throw the foot or pelvis out of their alignments, merely moving the problem somewhere else. In order to avoid this, the existing deformities are analyzed first, as they may affect the outcome of the treatment [8, 28, 30].

The choice of treatment depends largely on the magnitude of the discrepancy, which can be categorized by way of severity. Mild leg length discrepancies range from approximately 0 to 3 cm, moderate ranges from about 3 cm to 6 cm, while severe is anything above 6 cm [17, 27]. There is, of course, some variation among members of the medical community on what threshold lengths exist between severity levels, but they are almost always categorized in such a way that they roughly coincide with treatment methods [14, 19].

Within each severity range, several options exist which are used depending upon the age or physical condition of the patient [30]. Discrepancies of less than two centimeters are quite common in adults and usually not significantly symptomatic. Children with discrepancies such as this are usually not treated in the hope that the difference will represent such a small fraction of the total leg length that it will not pose any significant problem. Shoe lifts are prescribed for those for whom the small differences will create a noticeable limp [2, 14, 27].

For discrepancies between 2 to 5 cm, a number of options are available. Shoe lifts are commonly prescribed for smaller discrepancies, up to 5 cm. This is slightly less desirable than correction, as it does not correct the problem, but it has the distinct advantage of one of the only low - cost, non-surgical solutions capable of improving gait. Larger lifts -those greater than 5 cm - are almost always less than the discrepancy itself in order to avoid subjecting the patient' s ankle to inversion stresses, causing strains [2, 19, 26, 27, 28, 30].

There are three primary surgical procedures used to correct moderate leg length differences. The first is epiphysiodesis, or growth inhibition. The location of the operation is performed on either the distal end of the femur or the proximal end of the tibia, where 38% or 27% of the total leg growth can be stopped, respectively. The purpose of this method is to arrest the growth at one physis, thus slowing the overall growth of that leg, allowing the shorter one to catch up in such a way that when growth finally ceases, the legs will be of equal length. The procedure involves removing a block of bone from the medial and lateral parts of the growth plate, and replacing it at a 90 degree angle to create a bony bridge across the physis, arresting growth in that area (Figure 1.4). Epiphysiodesis has several advantages, such as a high degree of accuracy (within 1 cm total leg inequality), low risk, high success rate, low morbidity, ease of performance, and no postoperative immobilization. The disadvantages of such an operation include possible saphenous nerve damage, varus and valgus deformities, and unequal knee heights, resulting from the mistake of considering the total limb length and not the individual lengths of the tibia and femur individually. In addition, this operation is highly time dependent. Performing it too soon or late will result in over- or undercorrection [19, 26, 27, 30].

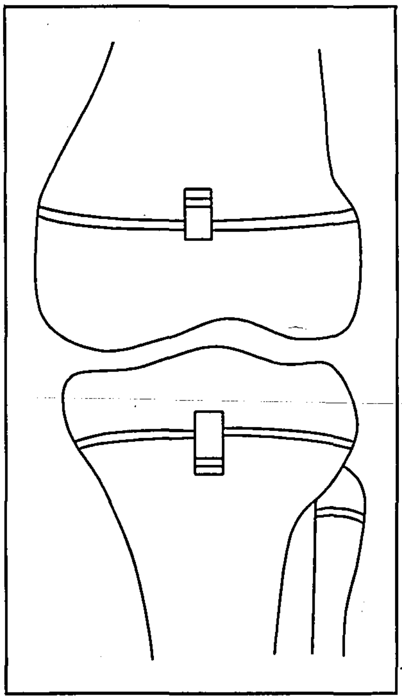
Figure 1.4 - Epiphysiodesis on the femur and tibia

Limb shortening, or resection osteotomy, is the second method used to treat moderate discrepancies. It is an option available for adults, late teenagers who have completed the growth process, or children whose growth patterns cannot be confidently predicted. In many discrepancy cases, it is more desirable to wait until full maturity before any type of corrective measures are used, thus eliminating the guesswork involved with predicting the total leg lengths. This operation is almost always performed on the femur, in which 5-6 cm can be safely removed. It can also be done on the tibia, in which 2-3 cm can be removed, but there is a higher risk of neurovascular complications and nonunion. The procedure involves the cutting of a section of either the femur or the tibia and using either a blade plate or rod to hold the bones in place until they heal. The leg then heals like a normal break wound. This is a very attractive option, especially due to its lack of guesswork, low complication, and no real need for postoperative immobilization. The negative aspects, however, include the thickening of the longer limb after shortening, bulging of muscles around the operation site, neurovascular disorders, and muscle weakness. In addition, since the shorter leg is usually the thinner one, the possible thickening of the shortened leg will accentuate this difference. Even so, resection osteotomies appear to be the safest and most reliable method for less severe discrepancies [19, 26, 27, 30].

The third method, physeal stapling, is an experimental process, using the idea that locally applied pressure inhibits physeal growth. Again, this process is performed at the growth areas, determined by how much inhibition is desired. The process involves the insertion of a number of staples across the physis, medially and laterally, to halt its growth (Figure 1.5). Once removed, growth resumes. The obvious advantages to such a method is the amount of direct control over the physeal growth. The staples can be removed as soon as the proper leg length equality has been achieved. However, there have been numerous reports of partial or total failure using this technique. Reports of shortening and angulation of the longer leg, unpredictable growth spurts, and possible physeal arrest during the insertion or removal have all been cited as the present dangers. It is thought that more research must be done before this method is to be safely considered [19, 27, 30].

There are fewer options available for those who suffer from severe leg length discrepancies. Limb lengthening, also called distraction osteogenesis, is usually one of the only courses of action open to patients who suffer from discrepancies between five and fifteen centimeters [19, 26, 27, 30]. This method is used as a last resort, when all other methods (such as a combination of epiphysiodesis and shoe lifts) have failed. It is most often done when the patient has reached adolescence but before growth has ceased. The procedure begins with an osteotomy, or cutting of the leg bone. This can be done to either the tibia or the femur, depending on the amount of lengthening desired. In most cases, up to 5 cm can be obtained from the tibia, and up to 8 cm can be obtained from the femur. Once the leg is cut, pins are inserted into their proximal and distal ends. Then, by means of an externally mounted machine, a slow distraction force is applied. There is usually a residual discrepancy of about 2 to 3 cm in most cases, and for differences of over 6 cm, this process may have to be repeated several times, or used in conjunction with resection osteotomy to ensure equalization. This process allows the height and proportions of the person to be maintained and avoids surgery on the longer side, which effectively makes the normal side abnormal in most cases. In addition, the patient is ambulatory and can apply partial weight bearing during the lengthening process. Angular deformities, refractures, physeal arrest, articular cartilage damage, long term weakness, and morbidity are the most common side effects of such an operation. Unfortunately, it is one of the only methods known capable of handling such a severe discrepancy [19, 26, 27, 30].

Discrepancies over 15 cm are usually not amenable to lengthening, shortening, or any other operative method mentioned above [26]. In the majority of cases, prosthetic devises are used to compensate for the difference. This is a method of last resort for those with deformed or functionally useless feet and those whose discrepancies exceed 15 to 20 cm. The process usually involves partial removal of the shorter limb or, in some cases, a Van Ness rotationplasty, in which the ankle is turned around and placed at the same level as the knee on the longer leg is located. Once this is completed, the prosthetic fitting can begin. The benefits of the process are quite simple. There is one hospitalization and one operation. Younger children who are chosen to have this done adapt very well to their prosthetics, and get used to them quickly, establishing a fairly normal gait. Unfortunately, it is also very emotionally trying for the entire family - especially if the children are older. However, this process has the ability to completely fix the problems caused by the severe discrepancy [19, 27].

Other methods have been theorized to stimulate growth in the physeal region without the need for surgical operations. One area, which remains unexplored at this time, is the possibility of electromagnetic growth stimulation [19, 30]. Another method, proposed by Wolff in 1892, involved subjecting the shorter leg to a constant tension, which is known to stimulate growth in the physis. This idea was extended in the 20th century into a procedure called "distraction epiphyseolysis", in which a trans skeletal distractor is mounted onto the leg to provide the distraction force, thus eliminating the need for surgical operation [19, 27, 30].

1.5 Present and Future Focus

This work dedicated to the study of leg length discrepancies will start with a description of gait analysis, moving to the investigation of the relationships among various gait parameters, such as that between walking velocity and cadence, walking velocity and impact acceleration, and the correlation factor between bone mounted strain gage and skin mounted accelerometer measurements. The work will then progress to provide a reliable way to isolate the contribution of heel impact to the total measured tibial acceleration.

Further work will be done to analyze the correlation between severity of leg length discrepancy to the force of impact on both legs. If a correlation is found, then the next step will be to find a method to equalize these forces in order to establish both a uniform gait and equal impact forces.

Chapter 2: Elements of the Gait Cycle

2.1 Introduction

For the purpose of analyzing the gait process, the human body can be divided into two sections: the passenger and the locomotor (Figure 2.1). The passenger, which consists of the head, neck, trunk, and arms, comprises approximately 70% of the total body weight, with the center of gravity located just in front of the 10th thoracic vertebra, approximately 33 centimeters above the hip joints on the average person. Coordinated muscle activity in this section maintains vertical alignment [24, 25].

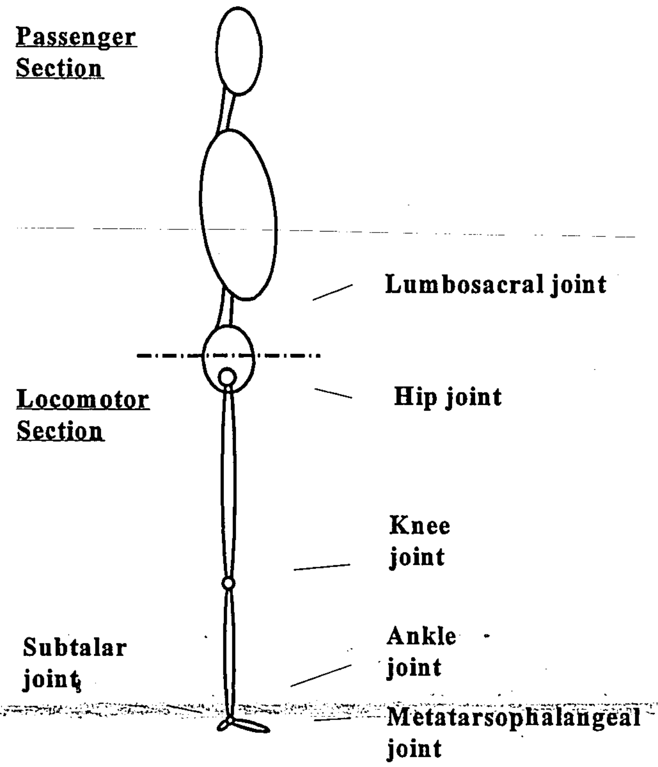

Figure 2.1 - Passenger and locomotor sections of the human body

Balance of the passenger is heavily dependent upon support from the locomotor section, which consists of the two legs and the pelvis. The pelvis can be thought of as the bridge between the two, since it serves as the support for the passenger and the link between the legs for the locomotor. There is a total of 11 joints in this half, and their motion is controlled by 57 muscles. Each of the bony segments, such as the pelvis, thigh, shank, foot, and toes, act as levers as each limb alternately assumes responsibility for the support and forward progression of the passenger section. After passing the weight of the passenger to the opposite leg, the limb swings forward to once again accept the total upper body weight during the next gait cycle [24, 25].

2.2 Functional Patterns and Objectives of Gait Cycle

Learning to walk occurs at a very early age for virtually every person such that it soon becomes an almost unconscious activity. However, there are several functional objectives that the human body' s locomotor system must fulfill each time any kind of activity such as walking or running takes place.

2.2.1 Propulsion

The first and the most obvious functional objective is that of propulsion. The main objective of walking is to move the body forward so that it may go from one place to another. The primary propelling force that causes this forward motion is the constant falling of body weight. The mobility present at the base of the supporting limb due to the presence of the heel, ankle, and forefoot is critical to the freedom of this "free fall". These joints serve as rockers, allowing the body to move forward while the knee is fully extended. The secondary propelling force is the forward swing of the free leg. The change of position and alignment provides an additional force as the momentum due to body weight decreases in the supporting limb. Once the body has moved forward, it is supported by the swing limb which has now assumed a stance position. This process continues cyclically and alternately with both limbs exchanging roles of supporter and momentum generator [13, 24, 25].

2.2.2 Stability

There are also significant challenges to stability when a person stands and walks. For example, the center of gravity is in the passenger segment of the body in which 70% of the total body weight is located. This means that only 30% of the total body weight is supporting the rest. The magnitude and direction of instability is a direct function of the extent to which the center of gravity is out of line with the rest of the joint vectors. In addition to this, the bones in the human body are long and thin, with round, smooth joint surfaces. This characteristic gives the body significant mobility at the price of stability. Due to these two factors, the upper part of the body will fall if the joints' centers of gravity are out of line with one another to any degree [24, 25].

There are a number of compensatory mechanisms present in the human body in order to counteract the unstable equilibrium inherent in standing, walking, or running. There are a total of three forces acting on the joint of the body when in any kind of weight bearing mode. There is the falling of the body weight due to instability, the tension in the ligaments, and the muscle activity in order to counteract this falling.

Because of the different mechanisms and dynamics present during walking and standing, the body has adopted different methods of handling the problems of stability.

When walking, the body lacks any form of passive stability that would be present during quiet standing. The body moves from behind to ahead of the supporting foot, and the area of support moves from the heel to the forefoot. During one complete gait cycle, different parts of the locomotor system serve to counteract the constantly changing source and direction of instability. Upon initial loading of the stance, or weight bearing, limb, the extensor muscles restrain the fall of the body weight. As the body weight progresses past the ankle, instability occurs once again. This time, the weight is restrained by the plantar flexor muscles. These muscles counteract the forces due to both gravity and forward momentum. Thus, an increase in walking speed puts a greater demand on the muscles that control deceleration and increase overall dynamic stability [24, 25].

When standing still, the entire body weight is supported by both legs. Loss of one of the legs by swinging it forward or lifting it up causes the center of gravity to become eccentric relative to the line of support. There is an instinctual contraction of the hip abductors, serving to shift the center of gravity towards the support limb, thus re-establishing stability in this position [24, 25].

2.2.3 Shock Absorption

The repetitive nature and impact associated with walking has required that shock absorption be another objective of muscle activity during the gait cycle. It takes place in three areas of the locomotor system: the ankle, knee, and the hip.

The ankle reacts immediately when the heel contacts the floor. There is a 10 degree flexion as the forefoot "free falls" to the ground. The pretibial muscle restrains this motion by slowing this rate of flexion. This action also reduces the rate of weight transfer in the forward direction [13].

The largest amount of shock absorption occurs in the knee. As the pretibial muscle restrains the fall of the foot, it forces the rest of the leg to follow the foot. This causes a forward roll in the muscle, causing knee flexion because the joint center is in front of the body vector. The quadriceps then react to this by decelerating the rate of knee flexion [13, 24, 25].

The hip reaction to impact is the immediate unloading of the other leg for swing. The removal of support causes a drop in the pelvis on that side, which is countered by the stance limb' s abductor muscles. As the hip motion and loading is countered in this way, the shock of impact is absorbed by the muscle action in the knees and ankle. Thus, the total load on the hip joint is reduced significantly [13, 24, 25].

2.2.4 Conservation of Energy

During gait, the center of gravity moves both horizontally and vertically in a somewhat sinusoidal fashion (Figure 2.2). This curve, in an average person, is about 2.2 times longer than their leg length. As the amplitude of this arc increases, so does the energy expenditure associated with the gait cycle. Several mechanisms conserve energy to the body by flattening this curve, thus reducing the amount of flexion and extension necessary in the hip and legs [13].

The first of these components is pelvic rotation. The pelvis rotates by approximately 4 degrees about the vertical axis corresponding with the line of progression, and increases in frequency linearly as speed increases. This serves to flatten the arc the center of gravity travels by elevating the ends of the arc, making the

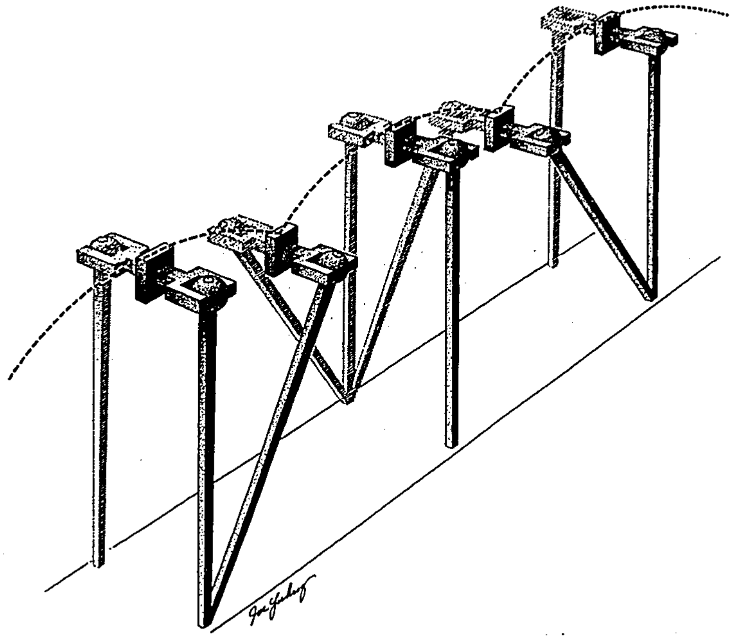

Figure 2.2 - Sinusoidal motion of the center of gravity with no energy conservation mechanisms present (reproduced from [13])

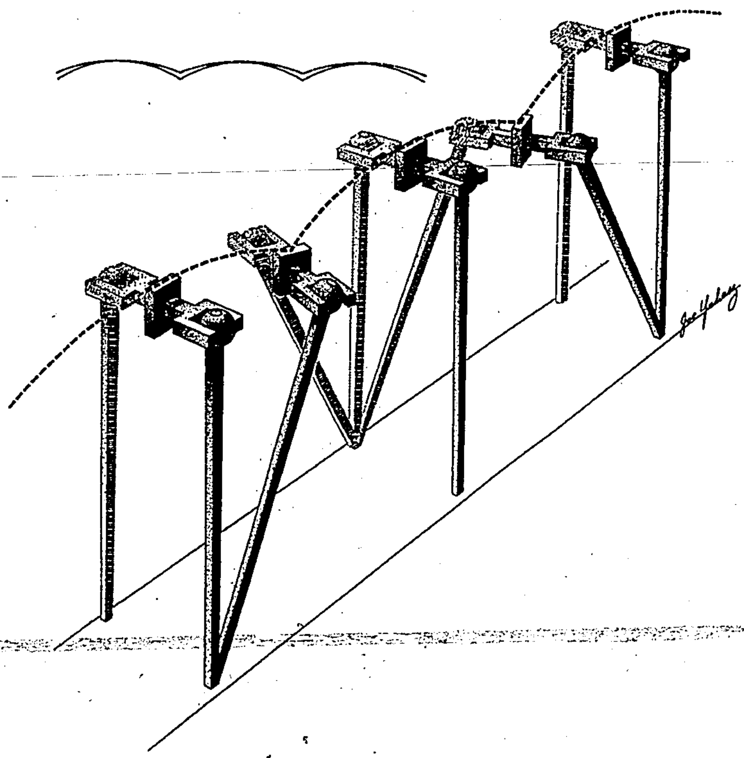

Figure 2.3 - Sinusoidal motion of the center of gravity with pelvic rotation (reproduced from [13])

intersections of the adjoining cycles less abrupt (Figure 2.3). Additionally, the magnitude of the ground reaction forces as well as the force required to change the direction of the center of gravity is decreased [13].

In addition to pelvic rotation, there is also a tilting motion in the coronal plane on the side opposite to the weight bearing limb. The magnitude of this listing is about 5 degrees on either side. In order to compensate for this, the knee joint of the swinging limb must bend in order to gain clearance from the ground. This, like the pelvic rotation, lowers the center of gravity and flattens the peaks of the arcs (Figure 2.4). It also serves as a shock absorbing mechanism by allowing some of the force due to impact to rotate the pelvis to a small degree [13].

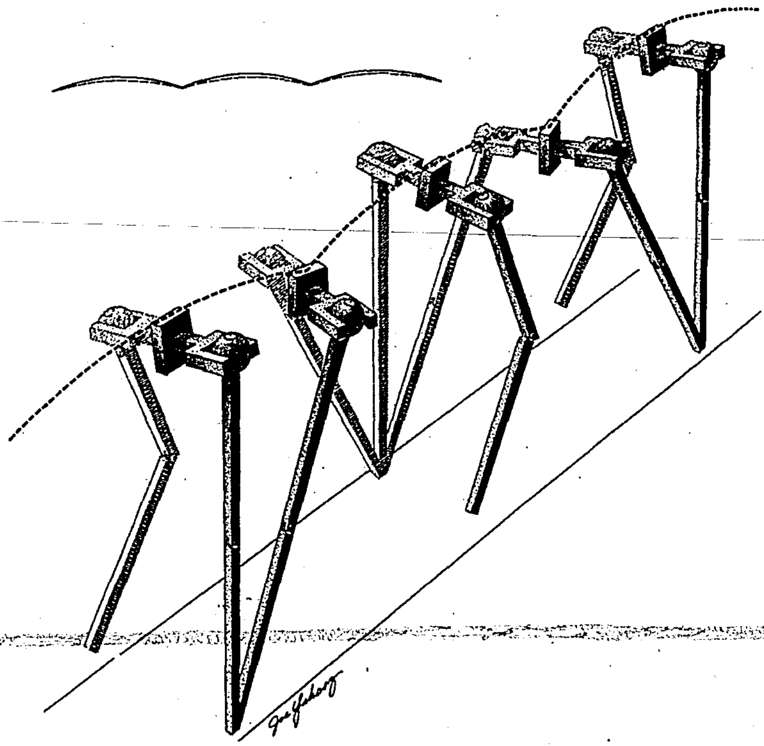

Figure 2.4 - Sinusoidal motion of the center of gravity with pelvic tilt (reproduced from [13])

When walking, the thorax and shoulders rotate back and forth, approximately 180 degrees out of phase with the hip motion. This produces an arm swing in the same shoulder which appears to have an overall balancing effect which smooths out forward progression. It has also been illustrated that thoracic rotation decreases in the overall energy expenditure [13].

Upon impact of the foot with the floor, the knee begins to flex by about 15 degrees. In addition, the presence of the foot and ankle allows the pathway of the knee to remain horizontal, allowing the knee flexion to smooth out the pathway of the hip. Thus, the flexion of the knee as well as the presence of the foot and ankle joints to serve as rockers serve to smooth out the discontinuities at the arc intersections (Figure 2.5) [13].

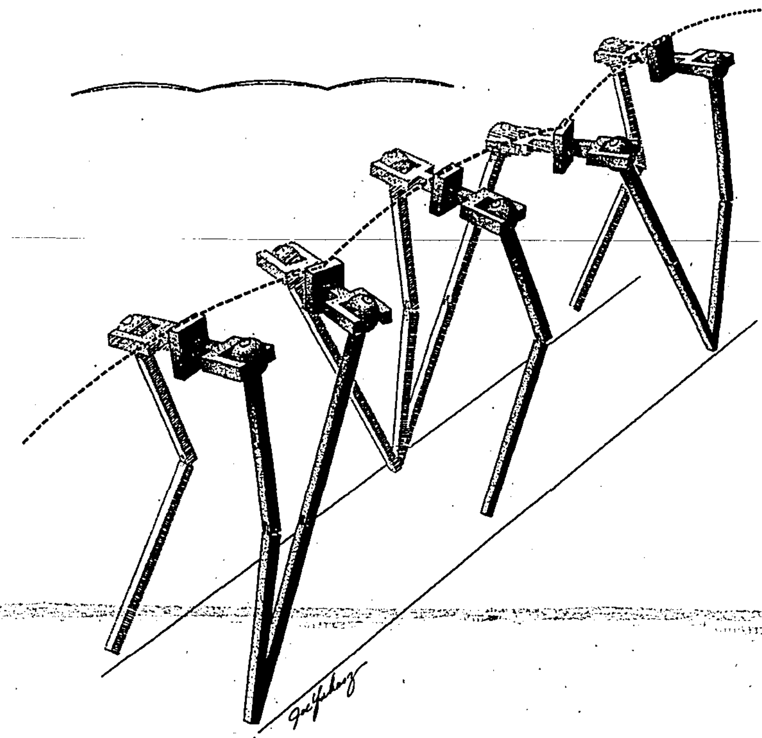

Figure 2.5 - Sinusoidal motion of the center of gravity with knee flexion (reproduced from [13])

2.3 The Gait Process

2.3.1 Cycle Divisions

The gait process can be broken up in to several components corresponding to the different loading characteristics that take place during each cycle, with minor variation among individuals. As these cycles are broken up into divisions, it is important to note that due to the symmetrical nature of the process under analysis, a complete cycle can be thought of as the actions that bring one leg from initial contact with the ground through the swing and back to initial contact again.

The stance phase is the first of two primary divisions and constitutes 60% of the total cycle. It begins with initial double stance which initiates the gait cycle (10%). Both feet are on the floor after initial contact, and there is an unequal sharing of the body weight by both feet, with more weight being supported by the front limb. What follows is a period of single limb support, while the opposite foot is lifted for its swing phase (40%). This is also called single stance, as the body' s entire weight is being completely supported by one leg. The duration of the single stance phase also gives a good indication of the support capability of that limb in a loading situation. The final part of this phase is called terminal double limb stance (10%), which begins with floor contact with the other foot. It continues until the original stance limb is lifted for swing. The swing phase makes up the remaining 40% of the gait cycle. The swing limb merely advances towards double limb stance to restart the cycle [24, 25].

2.3.2 Phases

In the previous section, the entire gait cycle was outlined in terms of phases and general actions during each of them. It is now important to examine the action and reaction of the body during each of these divisions in order to gain a more detailed understanding of how, when, and why each of the muscle actions take place.

In the double stance phase, there is an abrupt transfer of body weight from a limb of stable alignment to one of an unstable alignment. Thus, four functional patterns must be maintained, as mentioned earlier: forward progression, stability, shock absorption, and conservation of energy.

The initial contact of the foot with the ground takes place during the first 2% of the cycle, starting as soon as the heel touches the ground (Figure 2.6). During this stage, the hip is flexed, the knee is extended, and the ankle is dorsi flexed to neutral. At this time, the other limb is at the end of its terminal stance. Upon initial contact, the body

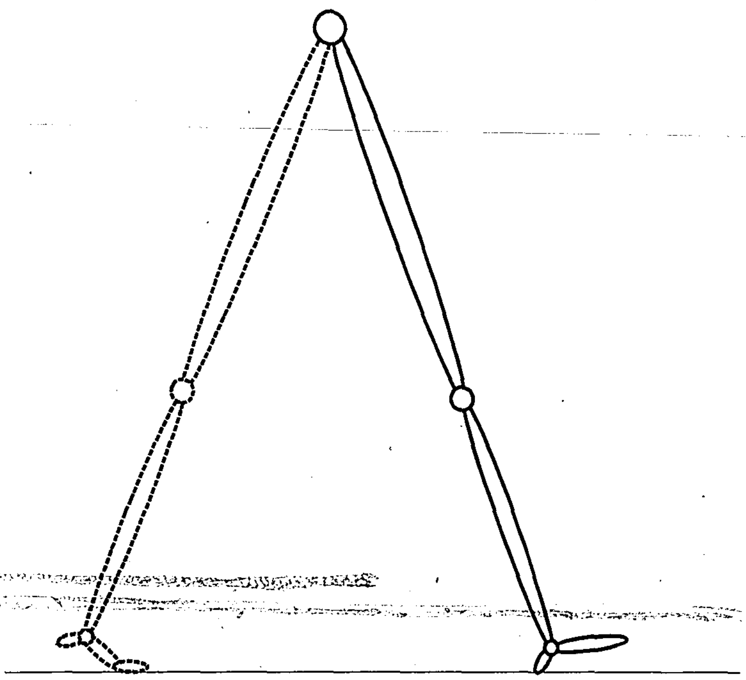

Figure 2.6 - Initial contact

responds to the change in load distribution (0-10%). Again, this occurs upon initial floor contact, and continues until the other foot is lifted to begin its swing. At this time, the body weight is transferred to the forward limb, while the knee is flexed for shock absorption. The ankle is plantar flexed in order to limit the action of the heel rocker by slowly bringing the forefoot to the floor. At this stage, the opposite limb is in its pre-swing stage (Figure 2.7) [13, 24, 25].

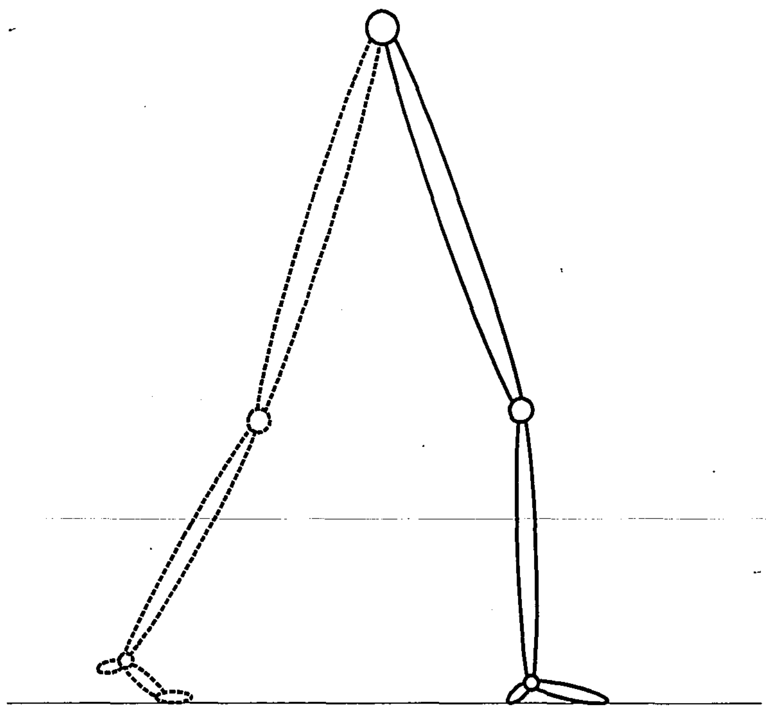

Figure 2.7 - Loading response phase

The loading response of the front leg has brought the body to the beginning of the single limb support phase. It begins when the opposite foot is lifted off of the ground, and continues until the same foot touches the floor again. During this phase of the cycle, one limb is solely responsible for supporting the entire body weight and its forward progression. During the first half of the single limb support phase, called midstance (10-

30%), the opposite foot is lifted up and continues to move forward until the body weight is aligned over the forefoot. The swing limb advances over the supporting limb by way of ankle dorsiflexion, while the knee and hip extend. The opposite limb advances through its midswing phase (Figure 2.8). In the second half of single limb support, called

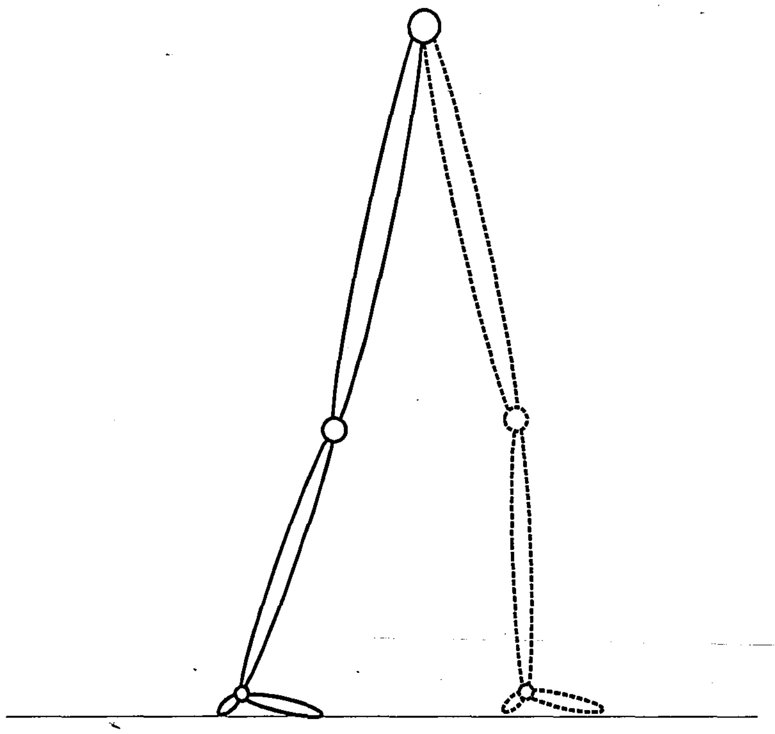

Figure 2.8 - Midstance phase

terminal stance (30-50%), the center of gravity moves ahead of the forefoot. The heel begins to rise as the limb advances over the forefoot rocker. The knee increases its extension and begins to flex slightly. The increased hip extension that takes place puts the limb further behind the progression of the body. The other limb is in terminal swing, while the weight supporting limb is now prepared for advancement (Figure 2.9) [13, 24, 25].

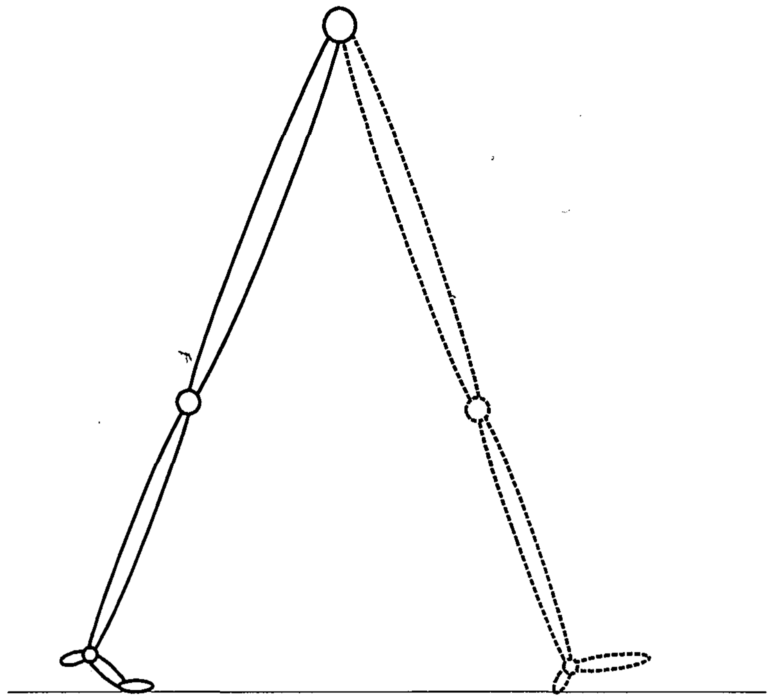

Figure 2.9 - Terminal stance phase

There are four phases of limb advancement: pre-swing, initial swing, mid-swing, and terminal swing. The pre-swing phase (50-60%) is the second interval of the gait cycle that involved double stance. It begins when the opposite foot makes contact with the floor and begins its loading response. The ankle of the swing limb plantar flexes, the knee flexes, and the hip loses its extension (Figure 2.10). During the initial swing (60-73%), the foot is lifted off of the ground and the limb is advanced by flexion in both the hip and knee. The limb advances until it is opposite the stance foot, which is now in mid-stance (Figure 2.11). Once the feet are lined up in the sagittal plane, the mid-swing phase begins (73-87%). The swing limb, which is now anterior to the body weight line, is advanced further by increased hip flexion. The knee extends in response to the influence of gravity, while the ankle dorsiflexes to a neutral position. This continues until the

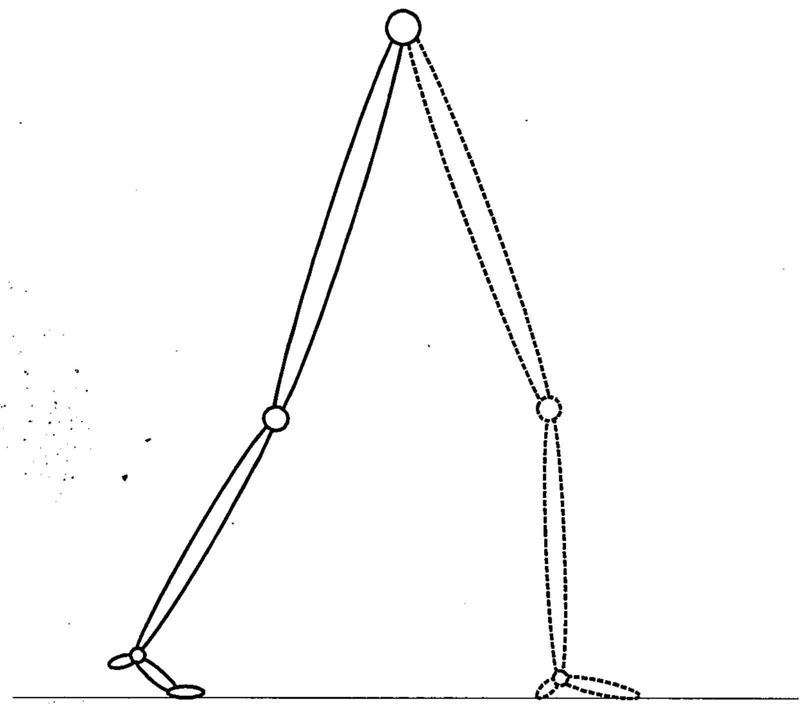

Figure 2.10 - Pre-swing phase

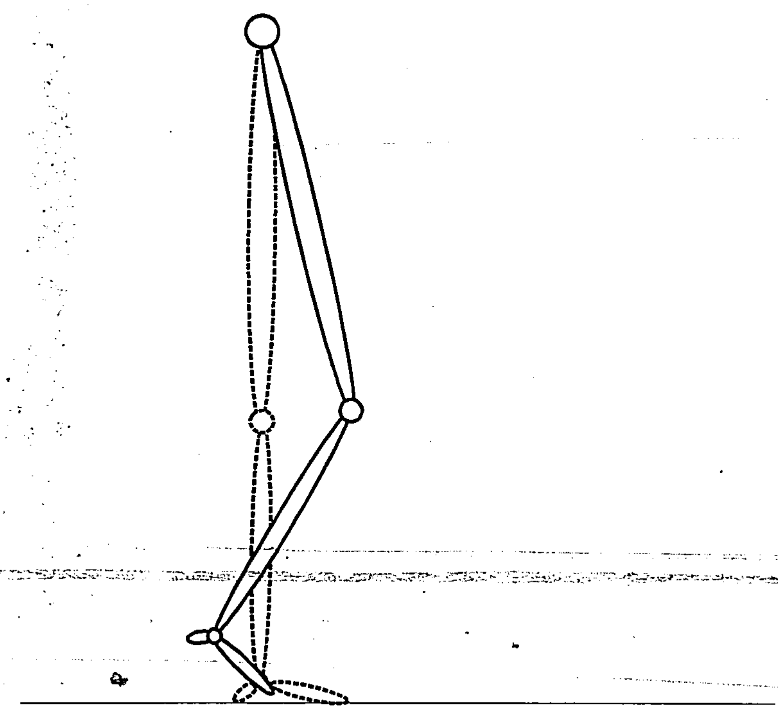

Figure 2.11 - Initial swing phase

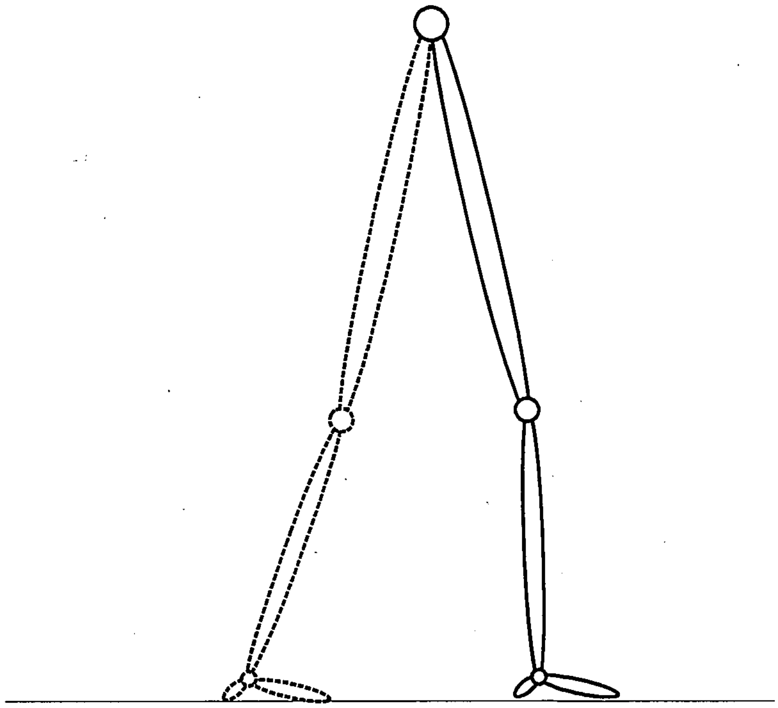

Figure 2.12 - Mid-swing phase

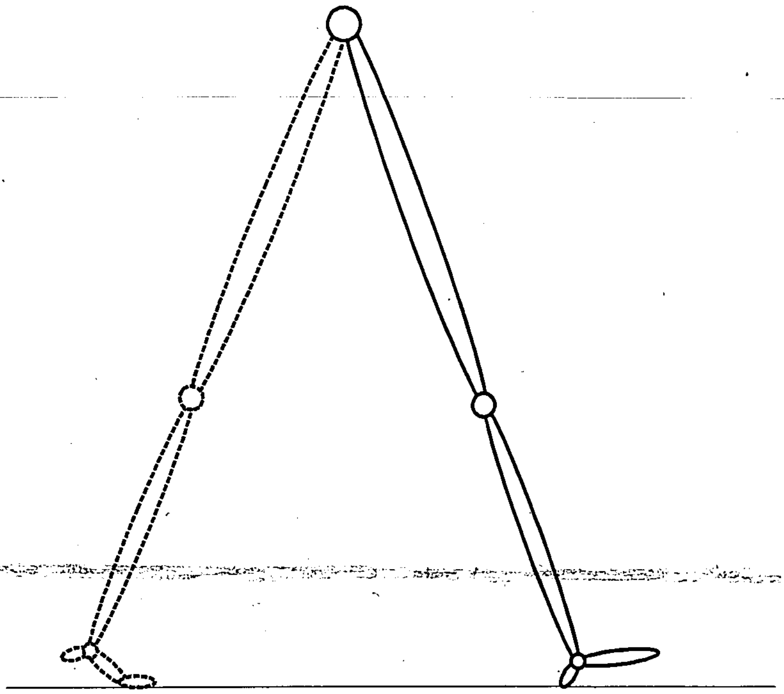

Figure 2.13 - Terminal swing

swinging limb is forward and the tibia is vertical. The opposite limb is still in its mid stance phase (Figure 2.12). The final interval of swing, the terminal swing (87-100%), begins with a vertical tibia. Increased extension of the knee causes further limb advancement. The hip maintains its flexion, while the ankle is still dorsiflexed to neutral. Limb advancement is completed when the leg is ahead of the thigh and the foot makes "free fall" contact with the floor. The other limb is in terminal stance at the end of this phase (Figure 2.13) [13, 24, 25].

Chapter 3: Experimental Methods

The underlying focus of this work is the study of leg length discrepancies. Fisrt it was important to establish the primary causes, biomechanical effects, and methods of treatment for this condition. In addition, a summary of the elements of the gait cycle was required in order to establish a basic understanding of the joint and muscle activity during the stance and swing phases. The primary goal of research, however, is to determine how the impact due to heel strike as well as the loading response of the legs changes as a result of mild, moderate, or severe discrepancies. In order to measure these quantities, acceleration measurements from a small skin mounted accelerometer will be used. It has been shown that the resulting error arising from having the measuring device mounted on the skin rather than in the bone is less than 5%. Error in this range is acceptable for this type of measurement [34].

In order to use acceleration as a basis for measurement, it is important to understand its relationships with other parameters of gait, such as walking velocity, angular motion of the tibia, and the loading response of the tibia at the time of impact. For this to be effectively determined, different methods of measurement must be used. A kinematic analysis using position markers will yield the angular motion of the tibia during the gait cycle. Surgically mounted strain gages on the upper tibia will be used in order to measure the reaction forces in the tibia during the time of impact. These will be compared to accelerometer measurements in order to find a correlation between the two.

3.1 Non-Invasive Accelerometer Measurements

The use of non-invasive techniques to obtain information about ground reaction forces and the acceleration due to impact is the most ideal, as it requires no surgical procedures on the subject and takes very little time to prepare for data collection. This allows for many sets of data to be recorded during a given session.

The use of accelerometers placed on the leg during the gait cycle allows for the measurement of the magnitude of acceleration due to heel strike that occurs once per cycle. By comparing the impact acceleration at different walking speeds, the relationship between the two parameters can be determined. It is anticipated that this method will eventually be used to compare the difference in heel strike acceleration differences in people afflicted with varying degrees of leg length inequality.

3.1.1 Experimental Apparatus and Procedure

All of the non-invasive accelerometer measurements were made in the Taylor Gymnasium at Lehigh University. A small (2 g) accelerometer was affixed to the tibial tuberosity (immediately below the patella) of the subject's right leg using a simple canvas strap. The transducer was mounted to the strap by way of a small aluminum brace, which allowed it to remain vertical during the mounting process. The transducer, which gave an output of 10 mV per g, was connected to an amplifier which boosted the signal by a factor of 100. The amplifier was powered by a 27 V DC power supply. The amplified signal was then sent to a converter which made the data readable by the computer. The computer converted the data from analog to digital at a rate of 1 kHz, or 1000 samples per second (Figure 3.1). A total of sixteen seconds of data were recorded

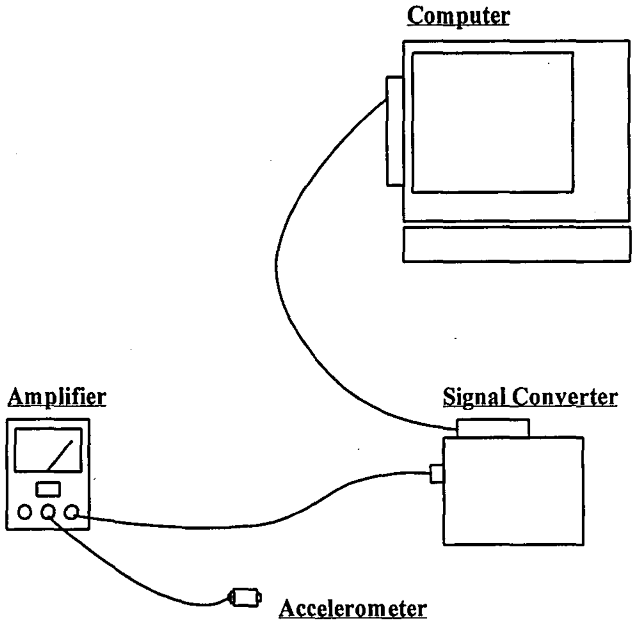

Figure 3.1 - Experimental setup for non-invasive acceleration measurements

for each trial.

A total of seventeen subjects - all healthy, unfatigued males, were chosen to participate in this experiment. Each subject walked on a treadmill at five different speeds: .89, 1.11, 1.33, 1.55, and 1.78 m/sec (2, 2.5, 3, 3.5, and 4 mph). The data taking process for one trial took approximately one minute. In this time, between 8 - 12 heel strikes were recorded onto the computer, depending on the walking speed.

3.1.2 Data Reduction and Analysis

The data was recorded as a text file on the computer, and was processed using a peak finding program. Each trial was graphed individually in an exploded view, the peaks were isolated by eye, and their locations and magnitudes were marked and recorded in a separate data file. In order to determine whether or not the differences in the time and acceleration averages were statistically significant or, in other words, from the same parent group of data, an analysis of variances (ANOVA) was performed on each individual set of data. ANOVA calculates the standard deviation among all of the data points in each set and then compares it to the average standard deviation among each set as a whole. Once the analysis was performed, the heel strike times and accelerations were each averaged, normalized for each trial in order to take into account variances among the subjects, and graphed as a function of walking velocity with error bars to take standard deviation into account.

3.2 Kinematics

Kinematic analysis using position markers is used in order to trace the path of the joints in motion during the gait cycle. From the position data which is recorded onto a computer, information about the angular and translational velocities and accelerations can be found. This method is used in order to determine the contribution of angular motion of the tibia during the moment of heel strike.

3.2.1 Experimental Apparatus and Procedure

Positional data of the subject was taken using the Ariel Performance Analysis System (APAS). Retroreflective markers were affixed to the subject' s body using athletic tape. One was placed on the rotation axis of the hip, the knee, the plantar and dorsiflexion axis of the ankle, the heel, the top of the head, and the rocker joint of the 5th toe (Figure 3.2). In order to minimize the movement of the markers relative to the underlying tissue in the hip, knee, and ankle joints, a device consisting of two wooden rods and a hinge aligned with the rotation axis of the knee was attached to the markers in

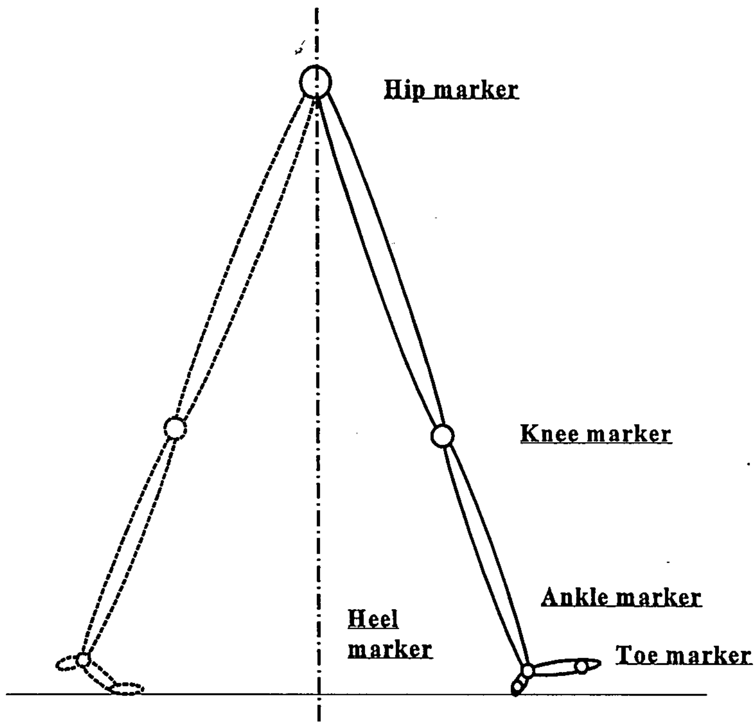

Figure 3.2 - Retroreflective marker positions on the locomotor section

order to constrain their movement. Thus, the hip-knee and knee-ankle distances were forced to be constant [3, 4].

The subject then walked on a treadmill at two speeds: 2.2 m/sec and 2.5 m/sec. The position data was recorded using a video recorder and four electronically shuttered cameras, which were zoomed into a 2 m by 1 m by 2 m viewing volume. Behind each camera was a 1000 W lamp which illuminated the reflective markers and allowed the cameras to pick up their position (Figure 3.3). Samples were taken at 50 Hz, and the positional data is recorded and interpreted using the frame grabbing module of the APAS system [3, 4].

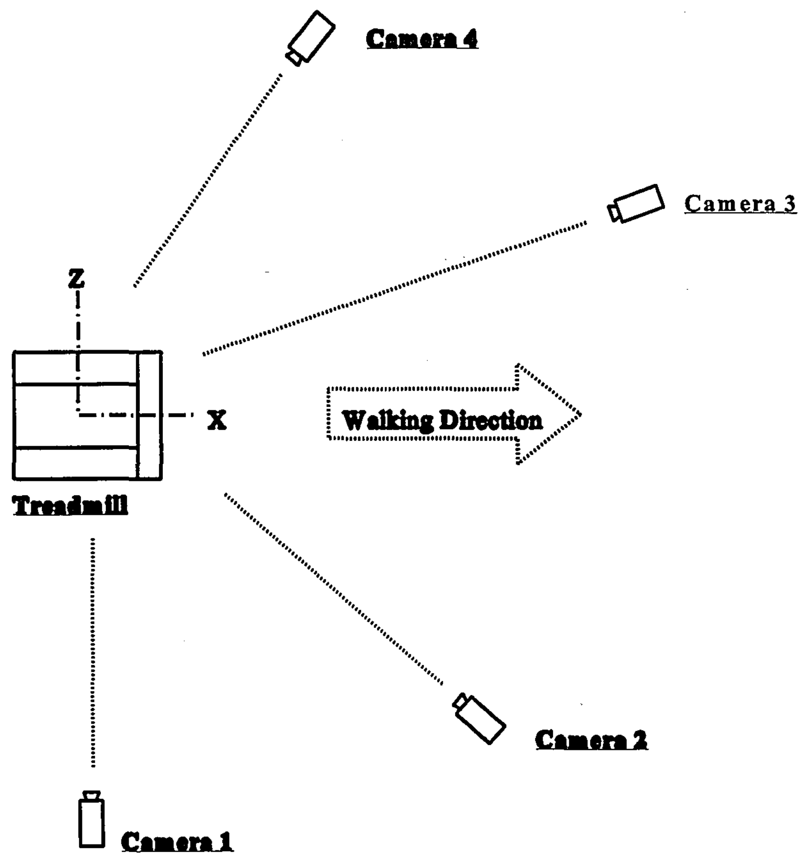

Figure 3.3 - Camera and treadmill placement for kinematic analysis

3.2.2 Data Reduction and Analysis

Once the position of each marker was plotted as a function of time, the angular position and velocity measurements were obtained. The data analysis in this section was done using Corel Quattro Pro Spreadsheet. The primary area of focus is the angular velocity of the tibia over the gait cycle. Thus, the positions of the ankle and the knee were plotted in both the x- and y- directions, and the expression for the tibial angle was obtained using the expression:

$$\theta = \tan^{-1} (x_a - x_k)/(y_k - y_a),$$

where the k and a subscripts denote the x- and y- position of the knee and ankle, respectively. A negative angle is when the hip and knee are in flexion, while the positive angle is when the hip and knee are in extension, immediately before heel strike (Figure 3.4). Each cycle of gait was then isolated,

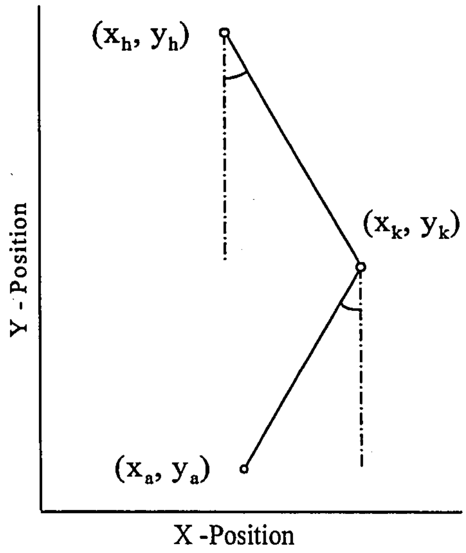

Figure 3.4 - Coordinate points for the hip, knee, and ankle

and the time of heel strike determined. This was done by interpreting the heel y - displacement. Each minimum per cycle corresponded to the heel making contact with the walking surface. The times were recorded and matched with the corresponding angle measurements at those times. From the above angle calculation, the angular velocity was measured by taking an average of the slopes adjacent to each data point. The results were plotted in order to find the instantaneous angular velocity at the time of impact.

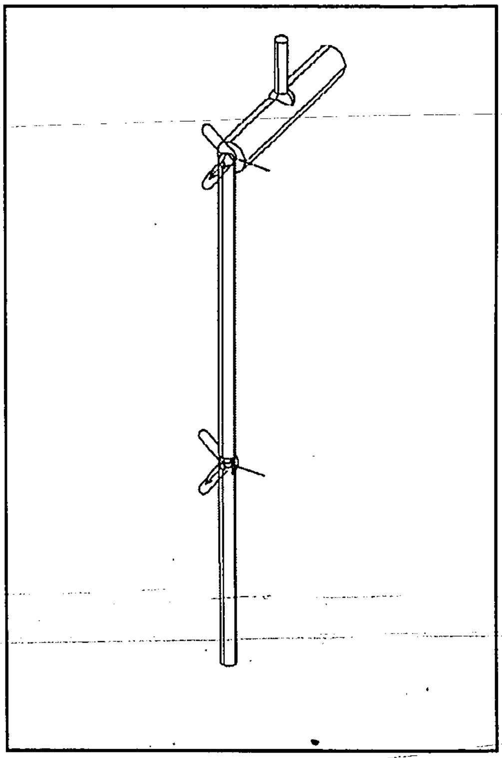

Figure 3.5 - I-DEAS rendering of human leg

Combined with the computer calculations to isolate the angular velocity, a simulation was created using SDRC' s I-DEAS CAD software. A simple model was designed (Figure 3.5), consisting of two long cylinders to reproduce the femur and tibia, with a smaller, thicker cylinder for the hip. In the Mechanism Design task, all of the instances were defined as rigid bodies and the pelvis was grounded. Two revolute joints were placed at the intersection between the tibia and femur, and the hip and knee. This type of joint provides only one rotational degree of freedom. Thus the total mechanism has two degrees of freedom.

Using the experimental data, the individual knee and hip angles were calculated. In order to obtain the individual angles, the previous equation was utilized to calculate the angles of the hip

$$\delta = \tan^{-1} (x_k - x_h)/(y_h - y_k),$$

and tibia

$$\theta = \tan^{-1} (x_a - x_k)/(y_k - y_a),$$

relative to the vertical (Figure 3.6). However, since the tibia angle depends upon the angle of the hip, it does not yet represent the true angle of the knee, τ. The true knee angle can be obtained merely by subtracting the tibial angle with the hip angle:

$$\tau = \theta - \delta.$$

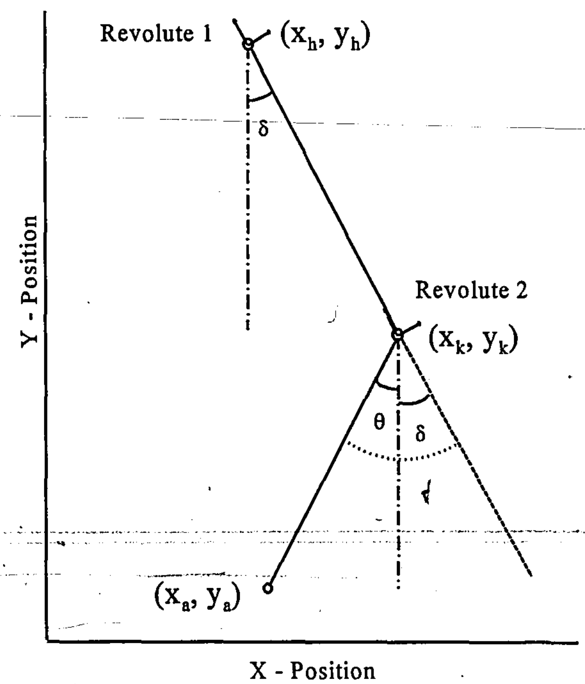

Figure 3.6 - Hip, knee, and tibial angles δ, θ, and τ respectively

A typical plot of hip and knee angles were input (with angular velocity measurements closest to the average) using the Cubic Spline option in the Function Creation command . The plots of angle vs. time for the knee (Figure 3.7, Figure 3.8) and hip (Figure 3.9, Figure 3.10) are shown. The mechanism was solved using I-DEAS' internal mechanism solver. A total of 40 data points were input for walking at 2.2 m/sec, and 37 were input for 2.5 m/sec. Each was solved using a mesh size of .1, or 10 solution points per input data point.

3.3 Invasive Tibial Strain Gage Measurements

Invasive techniques are those which require some amount of surgery for the mounting of measuring apparati on the human body. A set of three strain gages mounted onto the tibia is used in order to measure the strain present in the tibia during the stance phase of the gait cycle. This information will be used to compare with the accelerometer measurements to find a correlation between the two. If there is, then a simple, non-invasive technique can be used to find the forces present in the tibia in place of a more expensive and more painful surgical method.

3.3.1 Experimental Apparatus and Procedure

Two different measuring mechanisms were used during this phase of the experiment: a skin mounted accelerometer and a bone mounted strain gage set. The accelerometer was, as previously mentioned, mounted onto an aluminum brace and then to the tibia, immediately below the patella, by way of a canvas strap tied around the leg. The orientation was such that the wire was pointed upwards towards the belt line, making it easier to tape to the leg to avoid entangling during the data acquisition phase. The strain gage set, on the other hand, was surgically attached to the tibial bone itself,

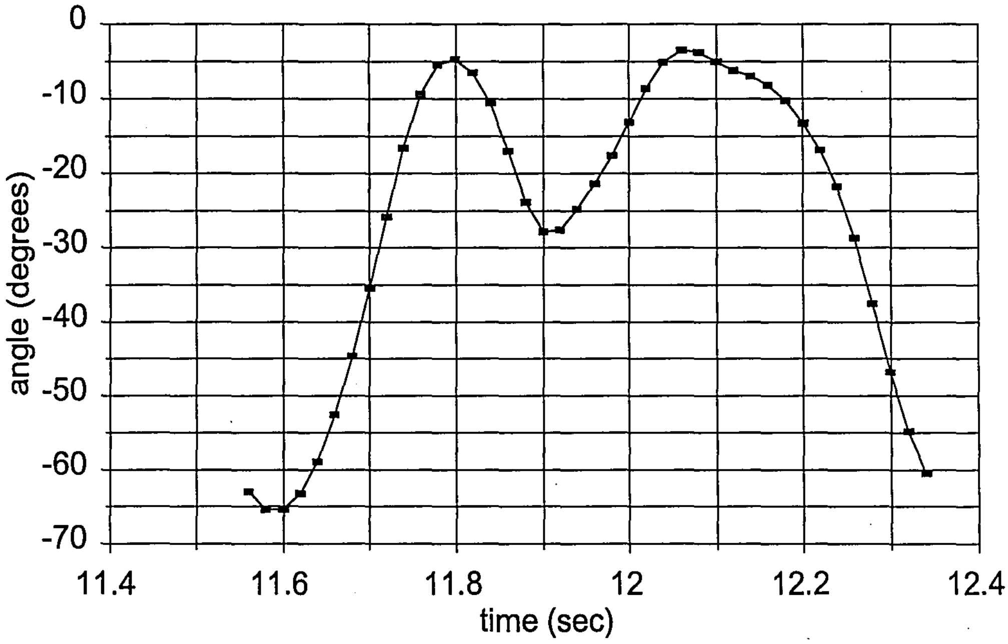

Figure 3.7 - Knee angle vs. time for a walking speed of 2.2 m/sec

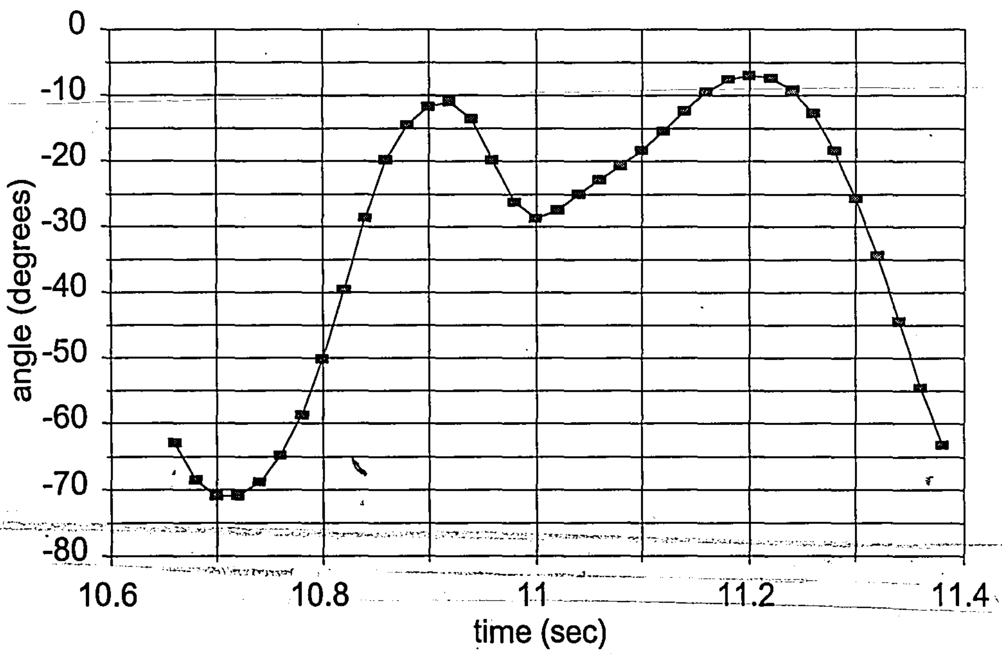

Figure 3.8 - Knee angle vs. time for a walking speed of 2.5 m/sec

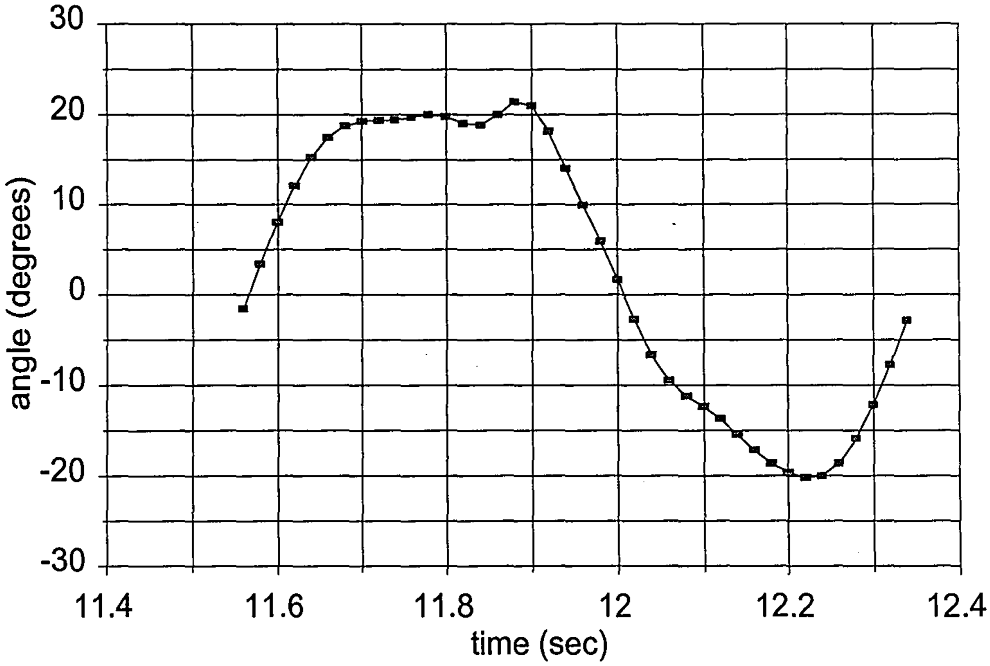

Figure 3.9 - Hip angle vs. time for a walking speed of 2.2 m/sec

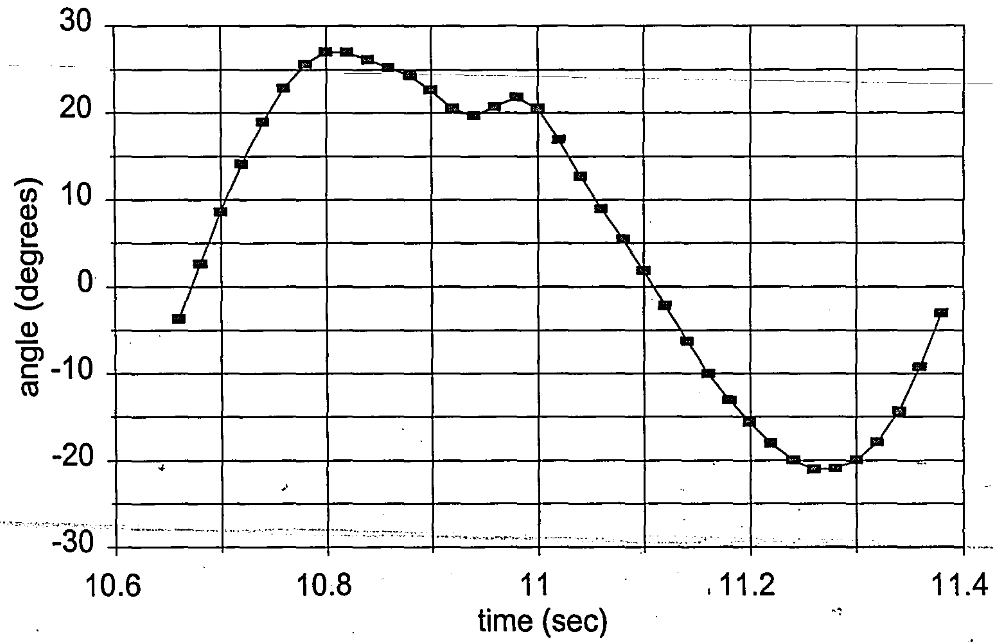

Figure 3.10 - Hip angle vs. time for a walking speed of 2.5 m/sec

immediately below the patella as well. Three strain gages were mounted this way, and positioned such that the center one was oriented vertically, while the other two were 45° from the vertical. Thus, with both the accelerometer and strain gage, both the impact acceleration due to walking and running as well as the reaction forces present in the tibia during the time of impact are measured and recorded.

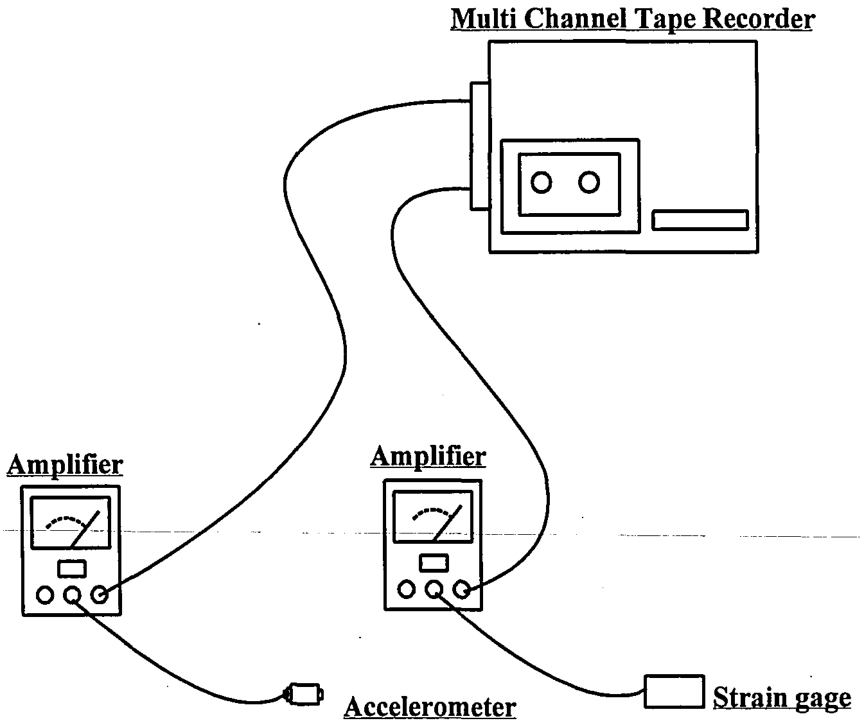

Figure 3.11 - Experimental setup for non-invasive accelerometer and invasive tibial strain measurements

The room in which the data acquisition took place contained a force plate in the floor, connected to another channel in the tape recorder (Figure 3.11). The subject walked, and later ran, over the force plate by taking a total of three steps - one before the force plate, one on it, and one after it. A total of 19 trials were performed in this manner, with each one differing only by the type of footwear used. Since the strain gage and accelerometer measurements were made simultaneously, it is not thought that the change in footwear has any effect on the correlation factor between the two.

Outputs from each strain gage and the accelerometer were sent through an amplifier, which boosted the signal by a factor of 10, thus making the conversion factor for the accelerometer 100 mV/g, and -10.43 microstrain/mV for the strain gage. The amplified signals were then sent to a multi-channel tape recorder so that they could be analyzed at a later time.

3.3.2 Data Reduction and Analysis

The multi-channel tape recorder was connected to a computer analog to digital converter, with a two channel input. The first connected the input from the vertical strain gage, while the second carried the data from the accelerometer. A computer program was used in order to transform the analog data to millivolts and write it to an external data file. Once all of the information was transferred from the tape to individual data files, it was filtered using an exponential smoothing function in Quattro Pro. This had to be done in order to proceed to peak identification, which utilized the same peak location program mentioned earlier.

The peaks were tagged and measured two ways. First, their magnitudes were measured with reference to a baseline, or zero strain or acceleration value. The peak to peak values were also recorded. This was done in order to more thoroughly establish the elements of each measurement that correlated with one another. Each set of peaks, with their magnitudes and times, were again recorded in separate data files, which were imported into Microsoft Excel for analysis.

As mentioned earlier, there was a total of three steps per trial: the step before, on, and after the force plate. Each of these steps was isolated and placed into its own subgroup, and after eliminating peaks which did not register properly or did not appear at all, a correlation analysis was performed.

Chapter 4: Dependence of Impact Acceleration and Cadence Upon Walking Velocity

4.1 Introduction and Theory

Along with the components of the gait cycle, there are also parameters that are associated with the locomotive process. The most obvious of these is *walking velocity*, which is merely the speed at which one walks. *Free speed walking* is the speed at which a person normally walks, corresponding to the lowest amount of expended energy per cycle. The *duration* of the walking cycle is the elapsed time between successive heel strikes of the same foot. By definition, it is inversely related to *cadence*, which is defined as the frequency of heel strikes per second by the same foot. *Stride length* is the distance between successive heel strikes of the same foot, while *step length* is the linear horizontal distance between the line of center of gravity and the heel upon impact with the ground (Figure 4.1). These two parameters are directly proportional to walking velocity [20, 21, 22].

At the time of impact, the heel striking the ground subjects the leg to transient shockwaves that travel up the tibia in the span of approximately 50 milliseconds. The impact forces due to heel strikes as a result of walking and running are the major cause of shin splints and degenerative joint diseases such as osteoarthritis later on in life [5, 15, 16, 33]. The total tibial acceleration is influenced by a number of different factors, such as the walking surface, the subject's fatigue, and possibly their walking or running

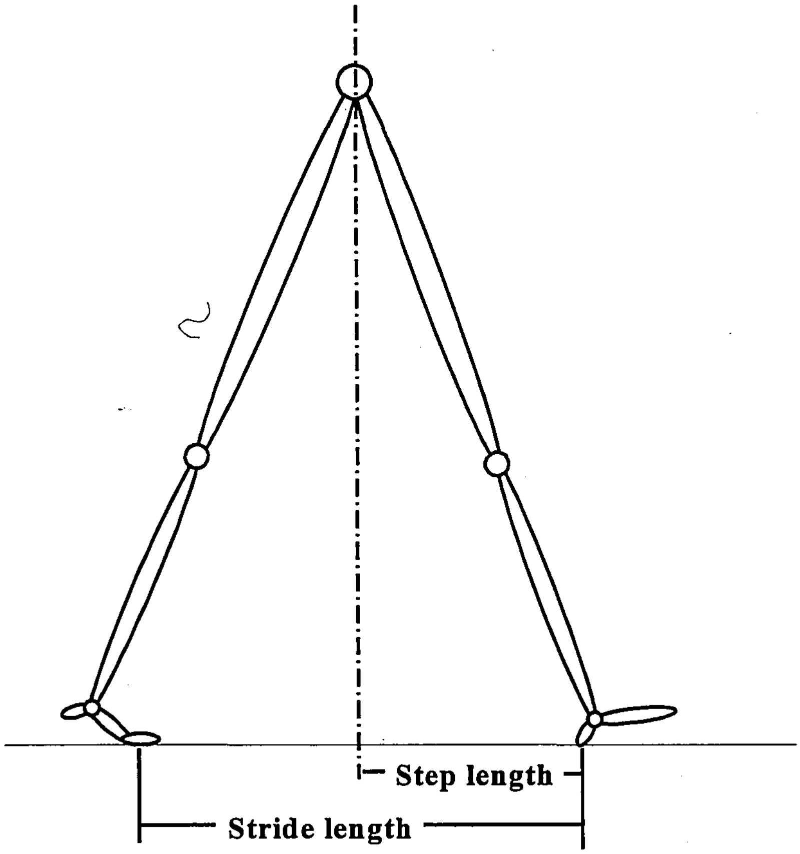

Figure 4.1 - Stride and step length

velocity, which will be shown here. There are two distinct components making up the total measured acceleration at the time of heel strike: the impact of the heel with the ground and the angular velocity of the tibia immediately prior to impact [15, 16].

The influence of walking velocity on cadence and tibial acceleration will be examined. Using a small skin - mounted accelerometer placed on the tibial tuberosity of the right leg, 15 subjects were measured at different walking speeds. The magnitude and times of their impacts were measured using a peak finder program, and from this, the velocity-cadence and velocity-acceleration relationships were found. The relationship between these two parameters is important to understand because of its use in determining the chances of the development of degenerative joint diseases in the legs and the force distribution in the joints during the gait cycle.

The angular motion of the tibia causes an acceleration in the radial direction as given by the following equation:

$$A_r = -\omega^2(t)\, r$$

The negative sign is added because the acceleration is oriented in the distal, or outward, direction. The sign convention used makes a positive value of θ to be when the knee and hip are extended, immediately before heel strike, and a negative value of θ to be when both are flexed. Thus, a positive value of θ corresponds to when the leg is swinging forward, ahead of the center of gravity.

Thus, the total tibial acceleration is the sum of the angular velocity and impact components:

$$A_{total} = A_r + A_i$$

Substituting the above expressions into the above equation, we obtain:

$$A_{total} = -\,\omega^2(t)r + A_i$$

This equation can be rewritten in order to obtain the expression for impact acceleration in terms of measurable quantities, since A_{total} and ω are both measured experimentally [16]:

$$A_i = A_{total} + \omega^2(t)r$$

The different components of total measured acceleration were isolated using a kinematic analysis system called APAS, which uses reflective markers attached to various joints on the lower half of the body; specifically, the hip, knee, ankle, heel, and toe [3, 4]. Angular position and velocity measurements were made prior to each heel strike in order to analyze the contribution of each component to the total measured quantity. From this, the true acceleration caused by heel impact with the ground was found using both computer calculation and simulation.

4.2 Results and Discussion

The graph of average normalized cadence and walking velocity in miles per hour is shown below (Figure 4.2). It can be seen that there is a linear relationship between walking velocity and cadence, with the slope of the line given by

$$C = 0.334v + 0.696,$$

where C is the cadence, normalized for a walking speed of 0.89 m/sec (2 mph), and v is the walking velocity in meters per second. The y - intercept is a significantly high value,

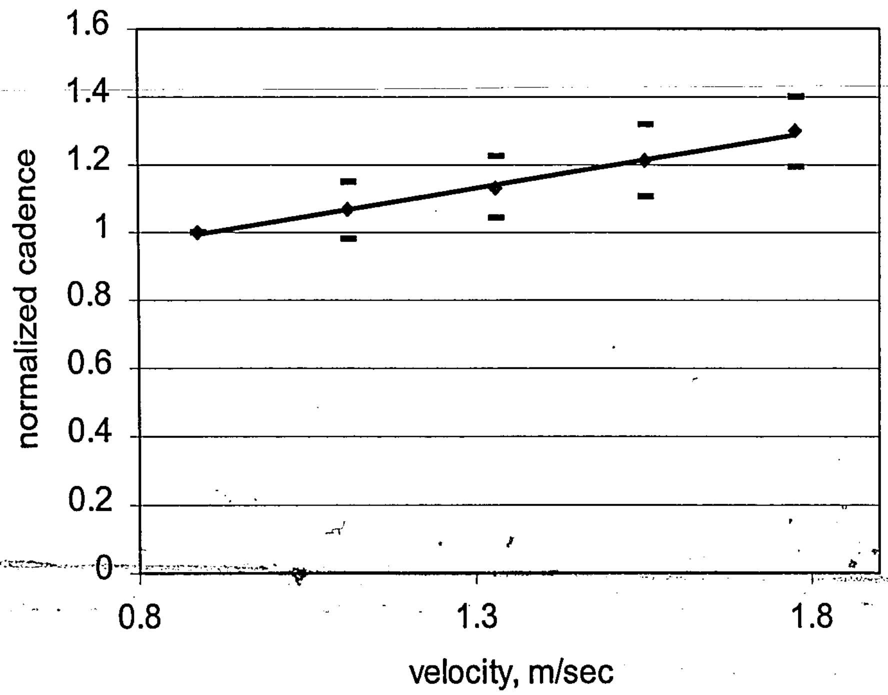

Figure 4.2 - Plot of normalized cadence vs. walking velocity

and would suggest that even with a zero walking velocity, there is a positive value for the cadence. This can be explained by the fact that at slower walking velocities, the natural gait cycle becomes distorted, with more time spent in the single stance immediately after heel strike. It is questionable whether or not the linear relationship illustrated in Figure 4.2 will be maintained as velocity is decreased further towards zero. In addition, for higher velocities, in the range of 2.44 - 2.66 m/sec (5.5 - 6 mph), the subject is no longer capable of maintaining speed by walking, and must make the transition to jogging or running. Because of the differences between these two phenomena, it does not make sense to assume that this straight line continues to a normalized cadence value significantly larger than 1.5 or 2.

The use of ANOVA served to prove that the difference between data from each trial per person was statistically significant. This verifies that the data taken per trial was not from the same parent group of data, and that there is a definite relationship between the two variables.

The results of the above mentioned averages are graphed as shown (Figure 4.3). The end result is that total tibial acceleration is linearly proportional to walking velocity, with a slope of the line given by

$$A = 1.829v - 0.613,$$

where A is the total measured tibial acceleration and v is the speed of walking in m/sec. Much like that between velocity and cadence, this equation implies a nonlinear relationship as walking speed decreases towards zero. The direct relationship between velocity and impact acceleration is similar in nature to prior research done on the

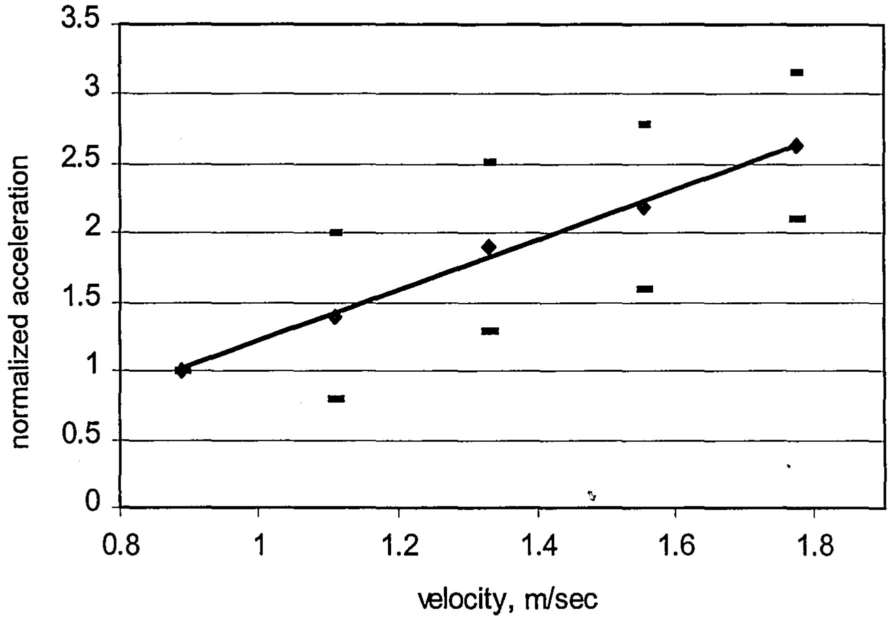

Figure 4.3 - Plot of normalized impact acceleration vs. walking velocity

relationship between walking velocity and ground reaction forces, suggesting that there may be a correlation between the two phenomena [1, 6].

The type of analysis performed with the collected data allowed for fairly consistent peak identifying and tagging for each of the trials. Some small error is present here, however, since the data reduction was performed by hand. The impact times were marked by tagging the maximum value of the shockwave that is measured by the accelerometer. While the time to reach this maximum value should be consistent, it is a strong function of the magnitude of the impact, which, as seen from the previous figure, can fluctuate significantly. The time difference due to this change in magnitude is on the order of milliseconds, which is of the same order of magnitude as the variation seen in the samples. The peak to peak value recorded may not adequately represent the heel strike that took place at that time. Thus, it is safe to assume that this is the primary source of error.

The results from the position analysis using APAS can be seen below. Angular position and velocity were plotted over the course of one complete gait cycle, with the heel strike in the middle of the graph. Results are shown for a walking speed of 2.2 m/sec (Figures 4.4, 4.5). The angle of the tibia begins at a maximum position of about -60 degrees and a velocity of approximately 1.5 sec^{-1}, representing the point at which the foot is the furthest behind the center of gravity, moving towards the vertical. The foot then crosses the vertical, directly below the center of gravity, achieving maximum positive velocity. The tibial angle increases further until it reaches a maximum of about 20 degrees. At this point, the angular velocity is zero, as the body prepares for heel strike. Impact with the ground occurs immediately after this, at an average angle of 6.079 ± 1.85 degrees and an average angular velocity of -4.115 ± 0.226 sec^{-1}. This is the point of maximum negative angular velocity for the tibia. At the time of heel strike, the tibial angle then moves back towards the vertical to a negative quantity as the ankle serves as a rocker for the progression of the body weight. The angle gradually decreases as the knee and hip are flexed and the foot is brought up to the starting position.

Measurements were also made at a walking speed of 2.5 m/sec (Figures 4.6, 4.7). While the overall trend for the progression of the tibial angle and angular velocity is the same, their values at the time of impact are different. The average angle at impact is 1.69 ± 4.15 degrees, while the average angular velocity of -3.89 ± 0.48 sec^{-1}.

All of the parameters and variables have been obtained for the time of impact.

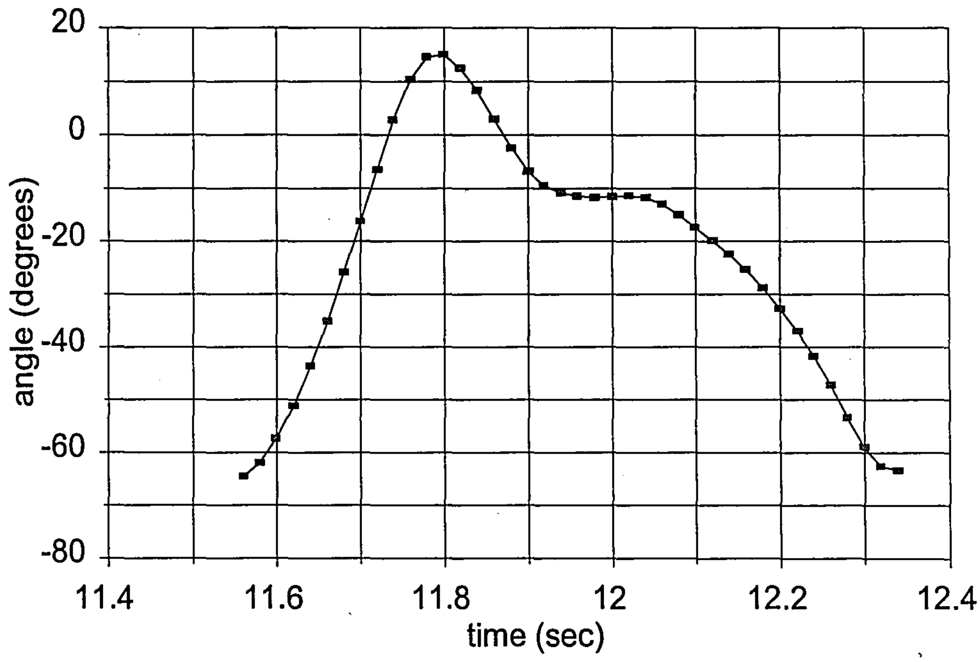

Figure 4.4 - Tibial angular position for a walking speed of 2.2 m/sec

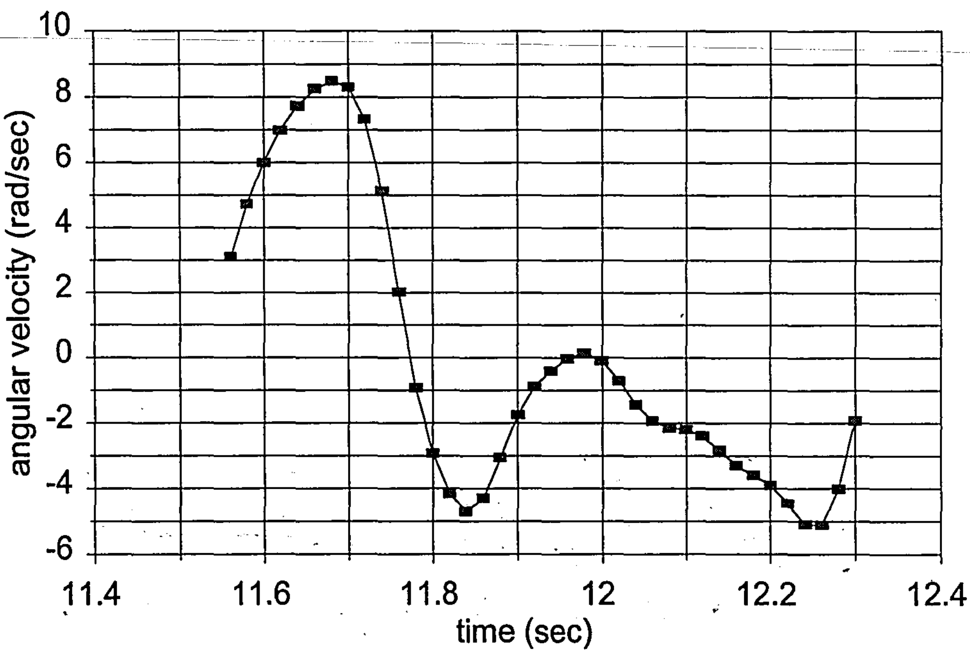

Figure 4.5 - Tibial angular velocity for a walking speed of 2.2 m/sec

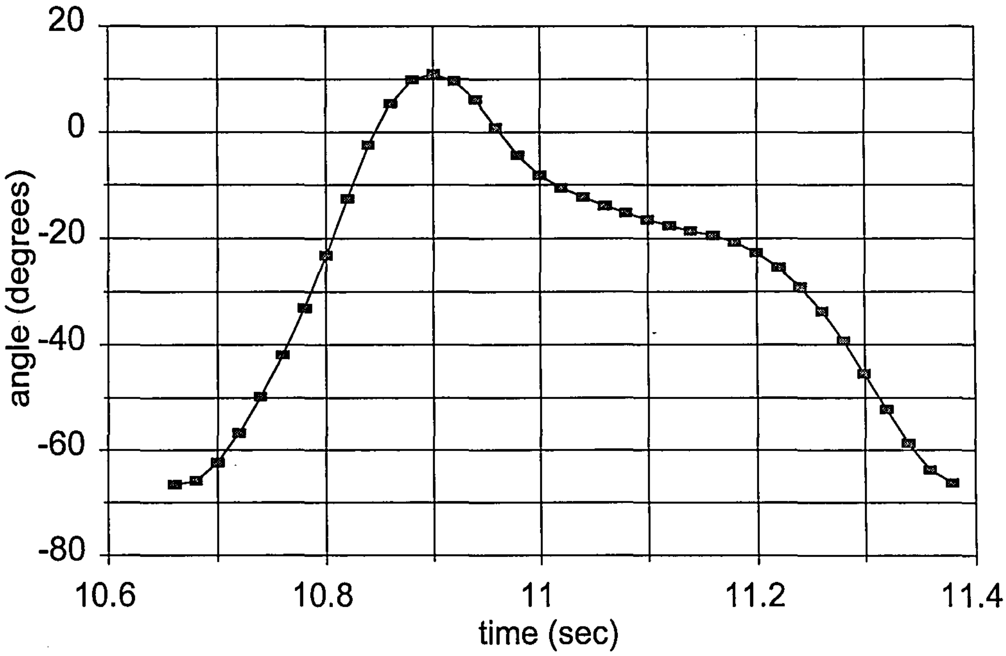

Figure 4.6 - Tibial angular position for a walking speed of 2.5 m/sec

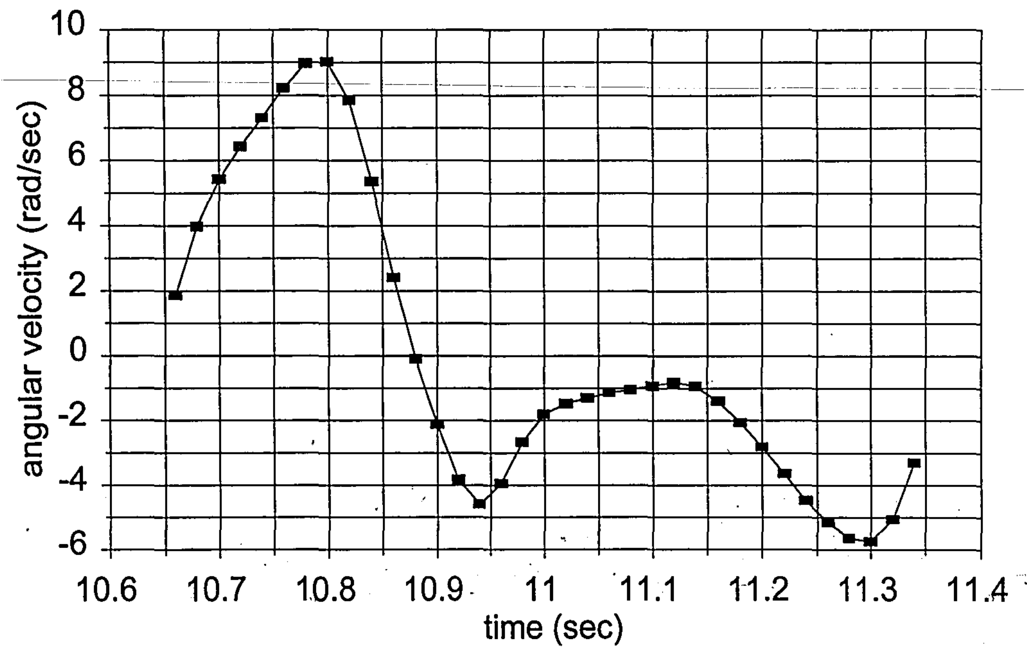

Figure 4.7 - Tibial angular velocity for a walking speed of 2.5 m/sec

The distance between the hip joint and the accelerometer is assumed to be on the order of 0.4 m, or 40 cm. Substituting ω and θ into the expression for impact acceleration, it is seen that

$$A_i = A_t + \omega^2 r(t) = A_t - 6.77 \text{ m/sec}^2$$

for walking at a speed of 2.2 m/sec. The total contribution of gravity and angular motion is 6.774 m/sec^2, or 0.69 times the acceleration due to gravity. The expression for walking at 2.5 m/sec is

$$A_i = A_t + \omega^2 r(t) = A_t - 6.06 \text{ m/sec}^2,$$

showing the contribution due to the angular motion of the tibia is 0.617 g. These contributions are significant since the impact due to walking is on the order of 2-3 g's. The values are both larger than the measured contribution of 0.44 g at a walking speed of 1.5 m/sec made by LaFortune and Hennig [15], indicating a direct relationship between walking speed and the contribution of angular motion of the tibia to total measured acceleration. The measurements taken at 2.5 m/sec, however, imply that while the contribution may increase initially, it reaches a maximum at a specific walking velocity, and then gradually decreases as speed increases until the gait cycle changes to running.

There are two primary sources of uncertainty using this type of measurement technique. The first arises from isolating the time of heel impact with the walking surface by way of finding a minimum in heel y - displacement. While this is a logical method, the sampling rate does not allow for a precise measurement of the exact time of impact. Thus, it allows the corresponding values of tibial angle and angular velocity to be erroneous. This is one of the possibilities when attempting to examine the source of the large standard deviation for the second trial (2.5 m/sec). The second source of uncertainty arises in the approximation of instantaneous tibial angular velocity. The averaging of the adjacent slopes gives an approximation, but not the exact value at that point. Both of these sources of uncertainty are significant, but not essential to eliminate in order to obtain an estimate of the contribution of each component of impact to the total measured value.

The results of the computer mechanism simulation can be seen in Figures 4.8 and 4.9 for a walking velocity of 2.2 m/sec, and Figures 4.10 and 4.11, which show the angular velocities of the hip and knee joints for a walking velocity of 2.5 m/sec. Comparing it with the results obtained from the previous section, is can be seen that the results are almost identical. Starting from the posterior position, the leg increases angular velocity until it reaches a maximum immediately below the center of gravity. It then decreases as it reaches its maximum positive angle and swings back slightly before the heel hits the ground. The angular velocity changes sign as the heel is used as a rocker for the progression of body weight and then brought back to the beginning of the measurement cycle.

At the time of impact, the angular velocity of the knee joint is -5.5 sec^{-1} and -5.25 sec^{-1} for walking speeds of 2.2 and 2.5 m/sec, while the corresponding hip joint velocities at the time of impact are 1 sec^{-1} for both walking velocities. The sum of these yield the angular velocity of the tibia at the heel strike: -4.5 sec^{-1} for walking at 2.2 m/sec, and

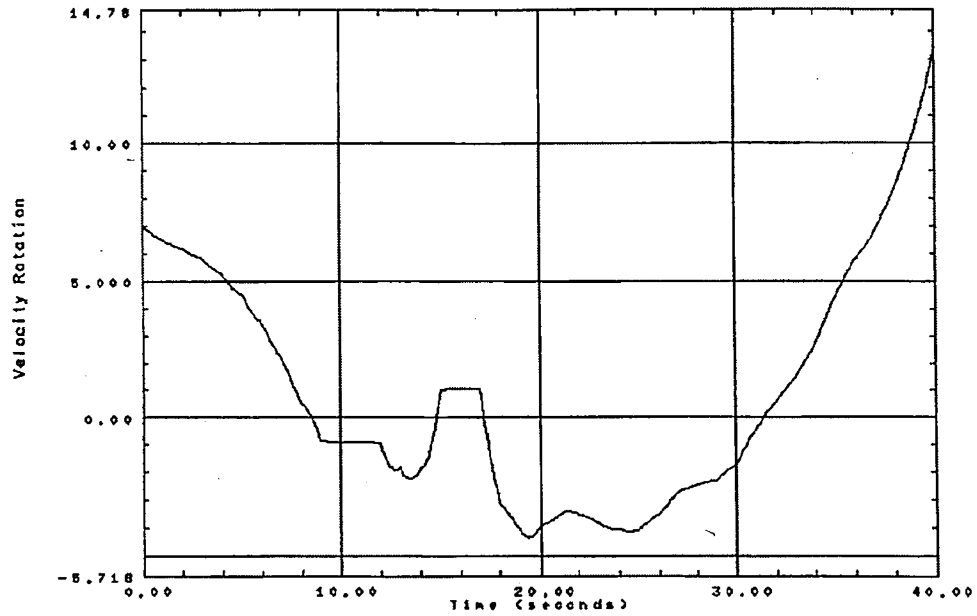

Figure 4.8 - Mechanism solution plot for hip joint angular velocity for a walking speed of 2.2 m/sec

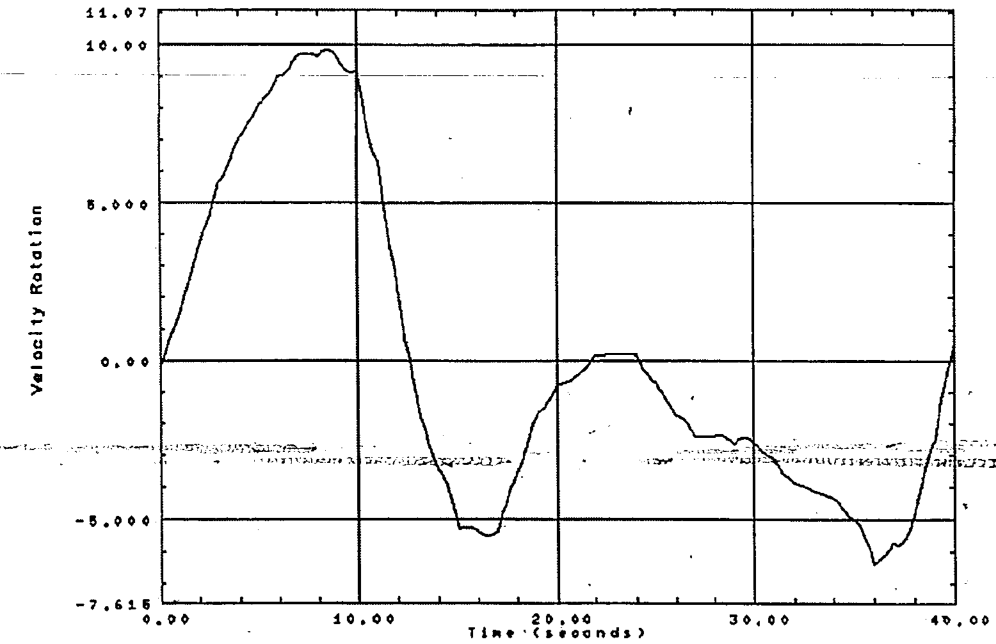

Figure 4.9 - Mechanism solution plot for knee joint angular velocity for a walking speed of 2.2 m/sec

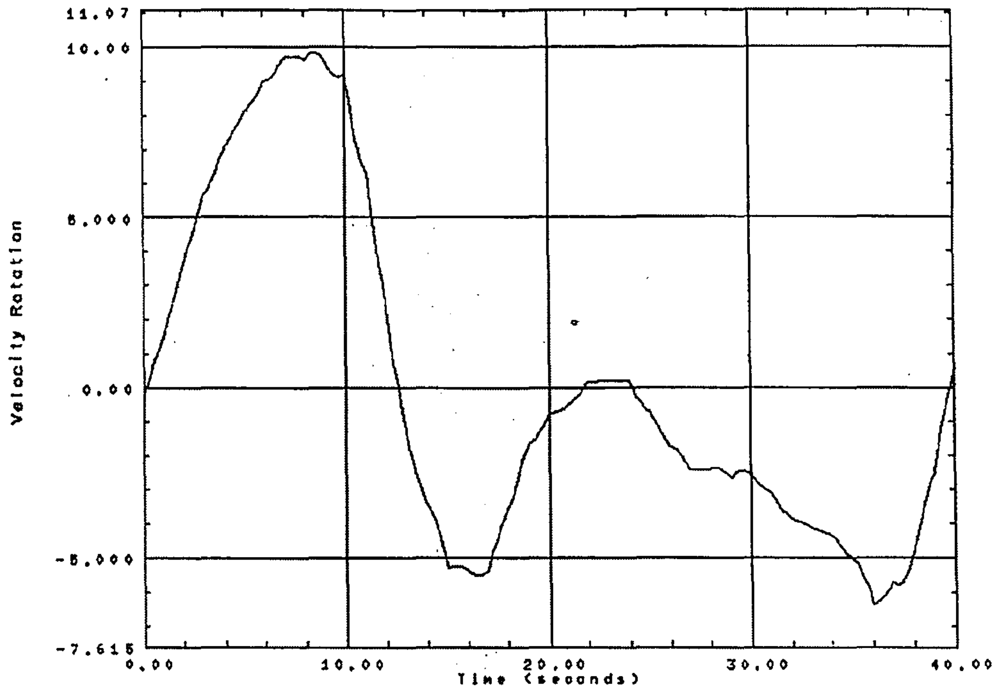

Figure 4.10 - Mechanism solution plot for hip joint angular velocity for a walking speed of 2.5 m/sec

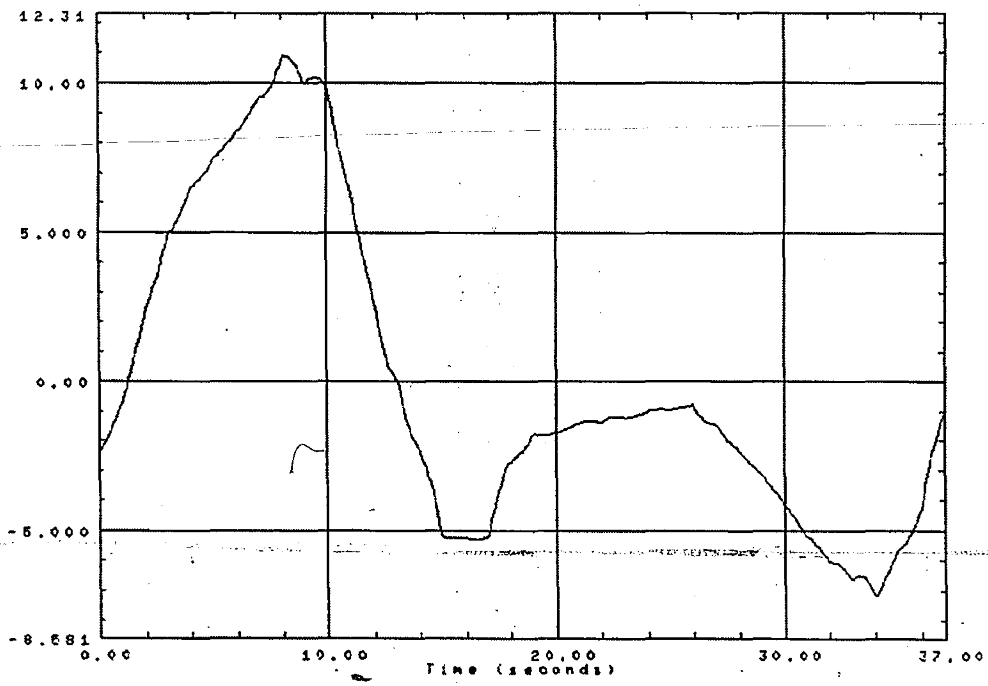

Figure 4.11 - Mechanism solution plot for knee joint angular velocity for a walking speed of 2.5 m/sec

-4.25 sec^{-1} for walking at 2.5 m/sec. These values are slightly greater in magnitude than the calculated averages of -4.12 and -3.89 sec^{-1} by factors of 8.5% and 8.47%, respectively. These discrepancies have largely been accounted for in the previous discussion of possible errors inherent in the spreadsheet analysis process.

4.3 Conclusions

There are numerous parameters associated with the phases of the gait cycle, such as walking velocity, cadence, impact acceleration, and stride length. Understanding of the relationships among them is highly beneficial in the context of experimental analysis. The relationship between walking speed and steps per second, or cadence, was examined. It was shown that a linear relationship does indeed exist between the two parameters illustrating that as walking speed increases, cadence increases as well. These results concur with prior research completed in the field [1, 7, 11, 20, 29].

During impact of the heel with the walking surface, there are two primary components that are measured simultaneously. There is the acceleration due to impact, and the acceleration caused by the angular motion of the tibia during swing and pre-stance. Using the Ariel Performance Analysis System, SDRC's I-DEAS, and simple trigonometry, the angular velocity of the tibia was isolated at the time of impact in order to estimate the magnitudes of the above components.

It was established that angular motion contributes significantly to the total measured acceleration. This contribution is somehow dependent upon the subject's walking velocity. It would seem that, given the current data, that as walking speed increases, angular acceleration decreases at the time of impact and thus its influence decreases. Further study in this field is recommended, with special focus placed on establishing a wider range of walking velocities in order to more effectively determine the effect of walking velocity on impact angle and tibial angular velocity, as well as increasing the sampling rate to 200-300 Hz, allowing for more precise identification of the time and, consequently, more precise identification of angular position and velocity at the time of impact.

Chapter 5: Correlation Between Impact Acceleration and Tibial Strain Measurements

5.1 Introduction

At the instant the heel strikes the walking surface during the gait cycle, a shockwave travels up the tibia to the femur and the rest of the body. This shockwave is highly dependent on the conditions of the walking surface, footwear, and the velocity at which one walks [5, 15, 16, 33]. Immediately after the impact occurs, the loading response takes place; the center of gravity moves over the heel rocker as the leg that made impact with the floor is now bearing the total weight of the locomotor section [3, 4, 5, 6, 24].

Both phenomena - the shockwave through the tibia and the loading response of the leg - occur at the same time. By using both a skin - mounted accelerometer and a series of bone mounted strain gages, it will be determined whether or not there is a direct correlation between the acceleration due to impact and the forces present in the tibia. If so, then it would be possible to extract force and strain measurements about the tibia by way of an inexpensive, non-invasive technique.

5.2 Results and Discussion

The significant results of the analysis are shown below, and include data from both walking and running (Figures 5.1, 5.2). It is shown that there is a fairly strong (-0.68 for walking, -0.74 for running) correlation between the output from the accelerometer and

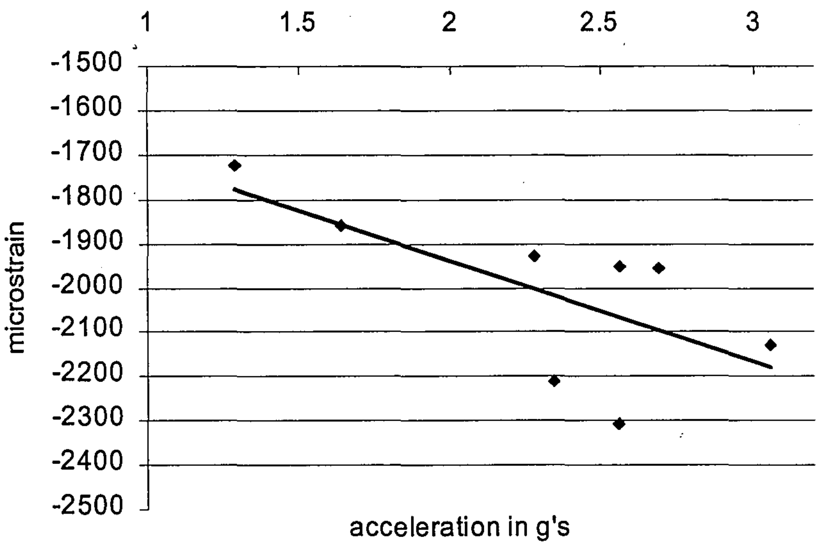

Figure 5.1 - Plot of correlation between strain gage and accelerometer output for walking

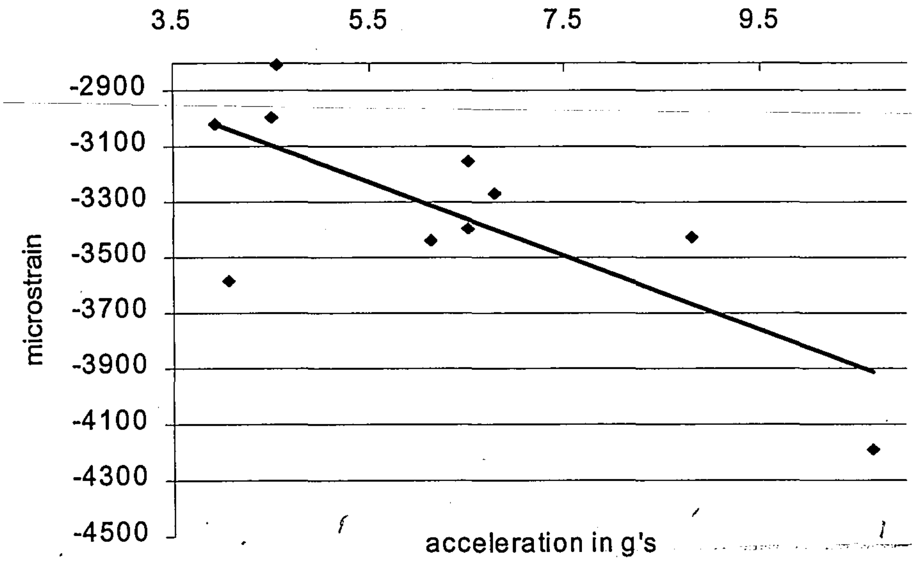

Figure 5.2 - Plot of correlation between strain gage and accelerometer output for running

the strain gage output during the first step taken prior to making contact with the force Table 5.1 shows the time delay between the measured acceleration as measured by the accelerometers and the strain measured by the strain gages. plate. Regression lines through the data show that the relationship is linear, with the equation of the line given by

$$S = -228.0g - 1484$$

for walking and

$$S = -132.0g - 2500$$

for running, where g is the accelerometer output in g's and S is the output from the strain gage in microstrains. Table 5.1 shows the time delay between the measured acceleration

walking						
data pt	**strain**	**data pt**	**accel**	**time (ms)**	**avg**	**stdev**
1138	178.43	1066	163.4983	72		
1012	221.16	987	255.57	25		
956	187.56	914	268.8057	42		
1336	187.41	1260	256.1085	76		
951	204.37	904	305.152	47		
1242	211.89	1216	234.267	26		
959	185.08	919	227.588	40		
1081	165.5	1029	128.4712	52	47.5	18.853
running						
data pt	**strain**	**data pt**	**accel**	**time (ms)**	**avg**	**stdev**
1101	269	1021	456.3261	80		
919	302	850	652.2062	69		
730	287	674	451.266	56		
951	289	891	394.4073	60		
985	329	921	881.5356	64		
993	343	928	406.7475	65		
798	330	706	613.72	92		
1275	326	1199	652.545	76		
923	313	859	678.34	64		
1077	402	982	1066	95	72.1	13.312

Table 5.1 - Time measurements for impact and tibial strain

as measured by the accelerometers and the strain measured by the strain gages for both walking and running. It can be seen that there is a clear time delay between the impact acceleration and the loading response of the tibia, with an average time difference of 47.5 msec for walking and 72.1 msec for running.

The remaining data from both the second and third steps over the force plate for walking and running, did not yield results that gave a clear correlation between strain and acceleration, even after eliminating the false readings. A number of factors had the potential to cause this condition, such as the subject's knowledge of the measuring process. During the data acquisition phase, the subject was aware of the location of the force plate and that the measurement was going to be taken at that location. As a result, it is possible that the anticipation of impact with the force plate has an effect on the overall gait pattern prior to and during the time of impact. The step after impact, since it is after the data that has been recorded, is erratic because of its perceived lack of significance by the subject. Thus, it is likely that the only aspect of gait during the measurement period that would show a correlation between the strain gage and accelerometer outputs is the first step.

5.3 Conclusions

Immediately after the heel makes contact with the ground, two phenomena take place adjacent to each other in the time domain. The first is a shockwave, traveling through the tibia and femur, as a result of the impact of the foot with the walking surface. Immediately after impact, the weight bearing phase of the leg begins. By placing an accelerometer and a set of strain gages on the tibia, both the acceleration due to impact and the loading response of the leg can be measured. It has been shown that there is a correlation (-0.68 for walking, -0.74 for running) between the peak to peak magnitude of the strain gage output and the baseline magnitude of the accelerometer measurements. This illustrates that it is possible to use a simple, inexpensive, and, most importantly, non-invasive technique to measure the reaction forces of the tibia from the accelerometer readings.

Further work is highly recommended to confirm the results obtained in this section. A method alternative to having a subject walking over a force plate taking no more than three steps per trial is desired. The ideal measurement technique that would yield the most consistent results would be to have the subject walking on a treadmill at various speeds, with both strain gage and accelerometers connected. This would allow much more than three data points per trial to be taken, and would allow for more consistent results by allowing the subject to fall into a regular, relaxed walking style. The only difficulty associated with continuing research in this area is finding subjects willing to participate in the surgical procedure required to screw in the strain gage brace to the tibia. If overcome, a more definitive correlation between these two parameters can be found.

Chapter 6: Conclusions

Leg length discrepancies are categorized as a functional or anatomic difference in the lengths of the lower limbs caused by disease, genetics, or trauma during growth. The effects of this condition differ in severity depending upon the degree of length inequality. For mild differences, there is often no significant biomechanical problem or complaint by the subject. For more severe (5 cm and up) differences, the pelvic tilt commonly associated with leg length discrepancies causes both immediate problems, such as lower back and knee pain and gait asymmetry, and more long term problems, such as sciatica, bursitis, and osteoarthritis caused by the change in the surface area of the hip joint in contact with the pelvis.

Treatment also depends on the degree of severity and the age of the patient. For mild differences (below 5 cm), shoe lifts are prescribed. While this does not take care of the cause, it compensates for the difference in height and eliminates some of the gait asymmetry. For moderate differences, such as those between 5 and 15 cm, resection osteotomy, epiphysiodesis, or physeal stapling are often used. For those patients suffering from severe discrepancies (15 cm and above), prosthetic fitting after amputation is one of the only options available.

The ultimate goal of studying leg length discrepancies is the development of orthotic and prosthetic devices to equalize the acceleration due to impact. In order to accomplish this, it was important to establish the causes, effects, and treatment of this condition. Once established, the elements of the gait cycle were examined, as well as the relationship between walking velocity with other gait parameters such as impact acceleration, cadence, and tibial strain immediately after impact. It was found that a direct relationship exists between walking velocity and impact acceleration, part of which is due to the contribution to total measured acceleration by the angular motion of the tibia. This quantity has been shown to contribute on the order of 0.6 g for a walking speed of 2.2 m/sec. In addition, a direct relationship was found to exist between walking velocity and cadence. It was found that as walking speed increased, the normalized cadence of subjects increased as well. This trend is thought to decrease nonlinearly at very low walking speeds before a normal gait cycle is reached. Through the use of surgically applied strain gages and skin mounted accelerometers, a correlation was found between impact acceleration and tibial strain at the moment of impact for walking (-0.68) and running (-0.74). These results are encouraging, as it shows that a relationship can be made between the impact acceleration and the tibial strain, thus making it possible for the use of non invasive techniques for such a measurement in the future.

Further work and experiments in each of the above mentioned subjects is highly recommended to verify the results found in this work. Additional experiments will follow, such as the measurement of the change of impact acceleration as a function of discrepancy. Orthotic devices will be individually studied, with particular focus on their ability to provide proper shock absorption and gait equalization. The ultimate goal of this will be the development of orthotic and prosthetic devices that not only restore proper leg length, but also restore symmetry to the gait cycle and equalize the acceleration due to impact.

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Vita

Michael David Marotta was born on September 9, 1974 in Belleville, New Jersey, the son of Michael Peter and Jeanne Anne Marotta. He was later raised in Forked River, New Jersey, where he attended elementary, middle, and high school, graduating as valedictorian from Lacey Township High School on June 15, 1992.

He attended Drew University starting September of 1992, where he majored in the field of physics. During this time, he was on the Dean's List for 7 semesters and became a member of Pi Mu Epsilon and Sigma Pi Sigma, the math and physics honor societies. He was graduated Magna Cum Laude from Drew University on May 18, 1996 with a Bachelor of Arts Degree in Physics.

In August of 1996, the author began his graduate studies at Lehigh University in the Department of Mechanical Engineering. He is currently fulfilling the requirements for his Master of Engineering Degree, which will be completed in May of 1998.

In June 1998, Michael will begin work in the Rotor Lifing and Structural Analysis Department at Pratt and Whitney in East Hartford, Connecticut.

END

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