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Theoretical Moment and Energy Requirements for a Prosthetic Ankle Joint

by

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Abstract

Current prosthetic legs on the market are passive and do not allow function similar to an intact leg. The purpose of this study was to determine the requirements for a theoretical prosthetic ankle joint to have the same energy and moment characteristics as an intact ankle joint during gait.

Kinematic and kinetic data were collected for three children with an able-body while walking and two while running. Results indicated that more energy is generated from the system than absorbed. This indicates that a passive prosthesis cannot achieve moment parity with a sound limb. It was determined that a non-linear spring with the characteristics of $M=e^{(A\varphi+B)}$ and an auxiliary energy source with the characteristics of $M=At^2e^{(-Bt)}(\varphi-\varphi_0)$ would effectively reproduce the ankle moment. These results provide information regarding the characteristics for a prosthetic ankle joint to function like an intact ankle joint during gait.

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1. Introduction

Kinematic (Lewallen et al., 1986; Engsberg et al., 1990) and kinetic asymmetries (Engsberg et al., 1991, 1993a, b) have been reported for the gait of children with a below-knee amputation (BKA) when compared to those of children with an able-body (AB). The asymmetry reported in the gait of BKAs seems to be related to degenerative changes in the lumbar spine and the knees of adults with a BKA (Perry, 1975; Burke et al., 1978; Brouwer et al., 1989). There is also a reported increase in the energy needs of children with a below-knee amputation while walking as compared to children with an able-body (Herbert et al., 1994). It has been hypothesized that the loading asymmetry and metabolic inefficiency cannot be reduced until the prosthetic leg functions similarly to an intact leg (Engsberg et al., 1993a, b). To accomplish this task, at least three areas require investigation: 1) socket design, 2) foot/ankle assembly, and 3) prosthetic alignment. This project focuses on the foot/ankle assembly.

The conventional prosthetic foot/ankle assembly is a passive structure, with any deformation being accomplished by the prosthetic foot. In an intact limb the gastrocnemius and soleus undergo an energy absorbing and energy producing phase (Winter, 1983a). The energy absorbing and generating mechanism is minimal or absent in a BKA (Prince et al., 1993). In an attempt to replace the dynamic function of the foot and ankle, "energy storing designs" (flexible keel designs) such as the Flex foot and Seattle foot have been developed. Michael (1987) describes the mechanical concept as a cantilevered spring which will absorb and return energy. Prince et al., 1994 also reported that the SACH foot should be added to the list of energy storing designs because even though it has a rigid keel, it's cosmetic material is capable of storing and recovering energy. Schneider et al. (1993) investigated kinematics and kinetics of the SACH foot and Flex foot for walking at a freely chosen speed and also at a fast walking pace. It was reported that there were asymmetries in the ground reaction forces, joint moments, and joint powers when comparing the prosthetic limb to the intact limb. However, the asymmetries were less pronounced with the Flex foot as compared to the SACH foot. Rigid and flexible keel prostheses were investigated in BKA runners by Prince et al., 1992. Prince et al., 1992 reported in both that flexible keels had less asymmetry than rigid keels with regard to vertical and anterior-posterior forces and corresponding impulses. However, they suggested that further improvements were necessary to reduce the running gait asymmetry (Prince et al., 1992). Regardless of the terminal device, forces produced by the prosthetic legs were less than those of an intact limb (Czerniecki et al., 1991; Engsberg et al., 1993b). It could be speculated from these data that current foot/ankle assemblies are unable to provide the forces necessary to achieve parity with intact legs. It would appear that the addition of an active prosthetic ankle complex may be an attribute necessary to allow the prosthetic leg to function similarly to an intact leg. An active prosthetic ankle complex would utilize a spring, but also have energy contribution by an auxiliary energy source and will be referred to as a spring system. To improve the design of the prosthetic ankle joints' components the following information is required: 1) investigation of the energy and moments of the ankle joint, 2) exploration of the various possibilities for generating those forces, and 3) determination of a method of interfacing the prosthetic ankle with the amputee. The purpose of this investigation was to determine the energy and moment requirements for a spring system incorporated into a theoretical prosthetic ankle joint. The theoretical prosthetic ankle joint would be a prediction of the experimental response of an intact ankle. This theoretical prosthetic spring system would have similar energy and moment characteristics as an intact ankle during gait.

2. Literature Review

2.1 Overview

The literature review provides an overview of scientific studies directly related to this investigation. The first section briefly discusses the structure and function of an intact foot and ankle joint as well as the motion of the foot and ankle joint during gait. The second section briefly discusses the design and function of the major below-knee prostheses on the market. The third section includes kinematic, kinetic, EMG, and oxygen consumption results from individuals with below-knee amputations. This section also presents studies which discuss energy generation and absorption of the ankle joint, and ankle moment results for individuals with able-bodies and individuals with below-knee amputation.

2.2 Structure and Function of the Foot Joints and Ankle Joint

2.2.1 Structure of the Foot and Ankle Joint (Figures 2.1 & 2.2)

The functional joints of the foot and ankle are the talocrural (also referred to as the ankle joint), the subtalar, the midtarsal (also referred to as the transverse tarsal), the tarsometatarsal, and the metatarsophalangeal joints (Donatelli, 1990). The ankle joint is a synovial hinge joint between the distal ends of the tibia and fibula, and the talus bone (Hay & Read, 1982; Crafts, 1985; Donatelli, 1990). The primary motions provided by the muscles acting on the ankle joint are dorsiflexion and plantarflexion (Hay & Read, 1982; Crafts, 1985; Condie, 1988). The subtalar joint allows rotary movement of the lower limb to occur without subsequent movement of the foot and provides shock absorption (Donatelli, 1990). The talocrural and subtalar joints comprise the ankle joint complex (Engsberg, 1987). The talocrural, subtalar, and midtarsal joints allow for motion in the three primary planes of the body; frontal (inversion/eversion), sagittal (dorsiflexion/plantarflexion) and transverse (adduction/abduction) (Donatelli, 1990). The tarsometatarsal joints permit flexion, extension and a certain degree of supination and pronation (Donatelli, 1990). The metatarsophalangeal joints provide flexion, extension and some abduction and adduction. The complex structure and function of the ankle and foot are difficult to reproduce. However, for a simplified artificial ankle/foot device for walking, Condie (1988), suggests that duplication of the ankle and subtalar joints would be sufficient.

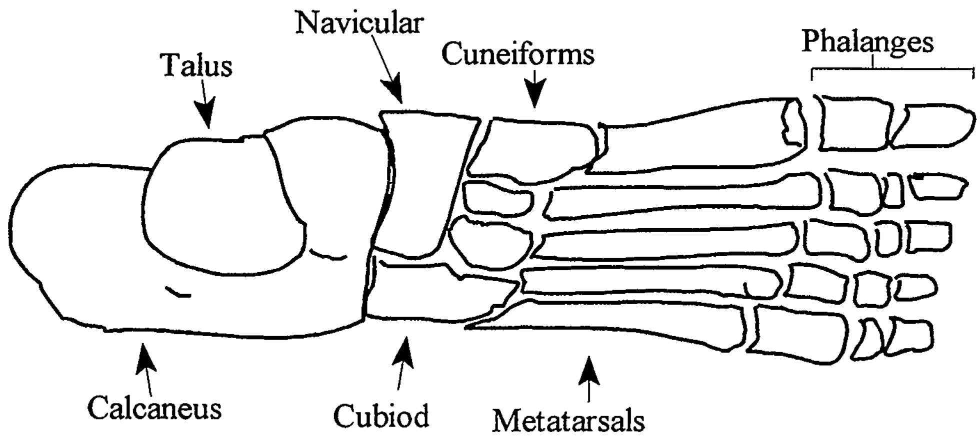

Figure 2.1: Dorsal View of the Bones of the Foot

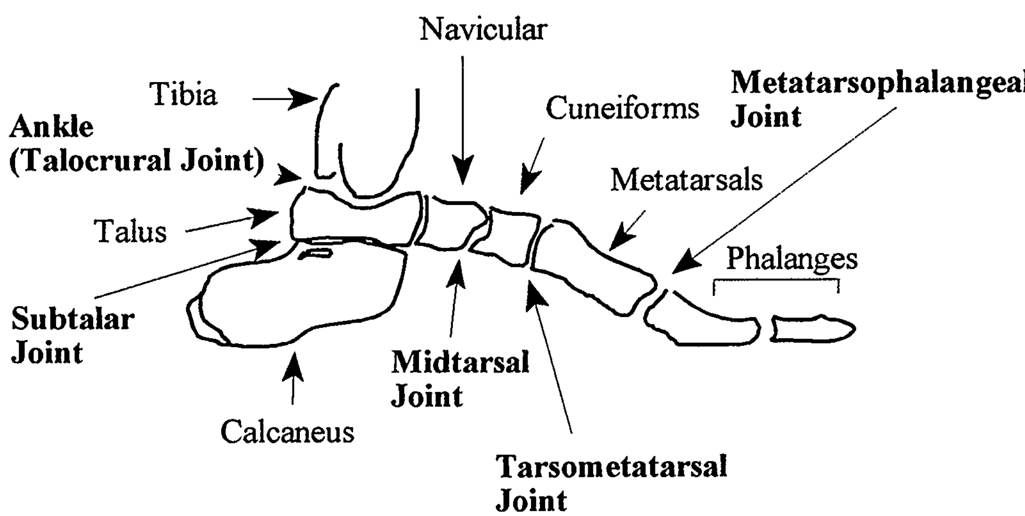

Figure 2.2: Medial View of the Bones of the Foot

2.2.2 Motion of the Foot and Ankle During Gait

The ankle motion during a gait cycle for an individual with an able-body has four major phases. The first phase occurs from initial contact to foot flat. The second phase starts at the onset of foot flat and continues until midstance. The third phase is from double support until toe-off. The first three phases comprise the stance phase of the gait cycle. The fourth phase is the swing phase.

During the first phase the foot is in a neutral or slightly dorsiflexed position. The pre-tibial muscles (dorsiflexors) contract eccentrically to control the lowering of the foot and decelerate the rate of plantarflexion (Donatelli, 1990; Ounpuu, 1990; Perry, 1992). In the second phase the ankle motion reverses to dorsiflexion. The dorsiflexion is controlled by eccentric contraction of the soleus muscle, assisted by the gastrocnemius (Perry, 1992). During the second phase the center of mass progresses in front of the ankle joint. In the third phase rapid plantarflexion of the ankle occurs as the body is propelled forward. The gastrocnemius and soleus contract concentrically (Perry, 1992). Plantarflexion occurs until toe-off. Perry, 1992 suggests that toe-off is a rolling over motion and not an active push-off. However, several researchers investigating kinematics of a prosthesis suggest there is an active push-off (Czerniecki et al., 1991; Prince et al., 1992; Schneider et al., 1993; and Allard et al., 1994). This investigation assumes there is an active push-off. The fourth phase is the swing phase where the primary motion is dorsiflexion. A small amount of plantarflexion exists during terminal swing, controlled by the anterior tibialis muscle in preparation of heel strike (Ounpuu, 1990).

The ankle joint motion during the gait cycle is associated with energy absorption and generation in the ankle joint complex (Winter, 1983a, b). The plantar-flexor group of an intact ankle undergoes two energy specific phases (Winter, 1983b). The first is an energy absorbing phase that occurs in the second phase of the gait cycle (from foot flat to double support) and consists of eccentric contraction in the gastrocnemius and soleus. The function of the energy absorbing phase is to control the rate of progression in the tibia and therefore control the extent of knee flexion during stance (Mann, 1981). Another function of the energy absorbing phase is converting kinetic energy to potential energy and storing it until it can be reconverted to kinetic energy at toe-off (Inman, 1993).

The second phase of energy generation occurs from heel-off to toe-off (the third phase of the gait cycle). Concentric contraction occurs in the plantar-flexor muscles and is the major source of energy for the lower extremity during toe-off (Winter, 1983a, b). The anatomical structure of the ankle/foot complex permits the necessary motion to achieve an efficient gait.

2.3 Prosthetic Feet Currently on the Market and Their Effects on Gait

A prosthetic limb attempts to simulate, by mechanical design, the complexity of the intact limb's system. Although the prosthetic designs vary in their complexity, the function of the intact foot has not yet been duplicated. The following text will briefly discuss the four major designs; Uniaxial, SACH, Multiaxial, and "energy storing" feet, and their general function.

2.3.1 Prosthetic Feet

The uniaxial design is one of the older prosthetic foot designs (Figure 2.3). It consists of a standard wood or plastic foot blank with the ankle bolt rotating in plain bushings (Condie, 1988). The design also implements rubber bumpers to resist and "restore" ankle moments (Condie, 1988). It is a passive structure, and not very efficient with energy return (Prince et al., 1993).

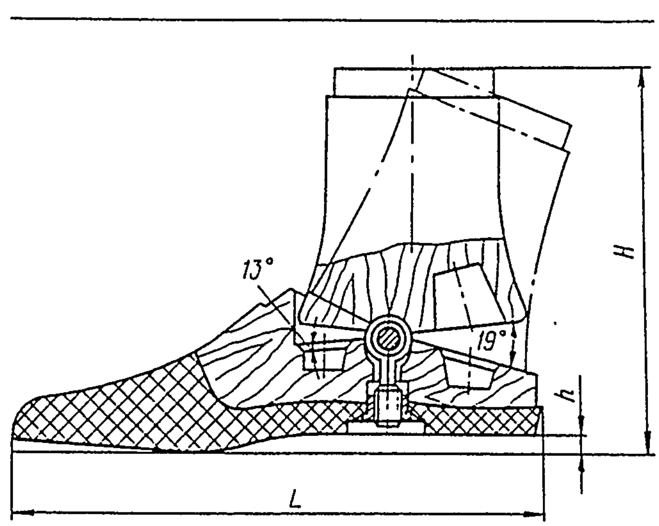

Figure 2.3: Single-Axis Foot (Michael, 1990)

The SACH (solid ankle cushioned heel) as its name suggests, consists of a rigid ankle and a rubber heel bumper (Figure 2.4). The SACH design allows for a smooth transition from heel strike to foot flat (Michael, 1990). The design, however, is a passive structure that responds to ground reaction forces. The SACH foot requires the BKA to develop enough momentum to ride over the rigid keel (Michael, 1990). Also, the SACH foot is not very efficient since much of the energy absorbed appears to dissipate in the viscous material of its design during walking (Winter and Sienko, 1988).

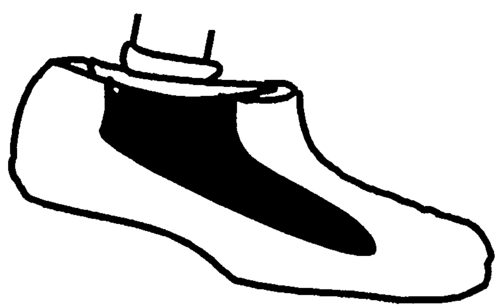

Figure 2.4: Solid Ankle Cushioned Heel (SACH) Foot (Michael, 1990)

Multiaxial feet permit mediolateral motion and transverse rotation (Figure 2.5). One multiaxial design is the Greissinger. It consists of a wooden ankle section and a molded rubber block recessed into plastic. The foot and ankle are attached by a U-bolt and a yoke-type assembly. The Greissinger foot design permits dorsiflexion, inversion, eversion and transverse rotation (Condie, 1988). Although this design is good for ambulation on uneven ground, it is not efficient at storing and returning energy (Michael, 1990).

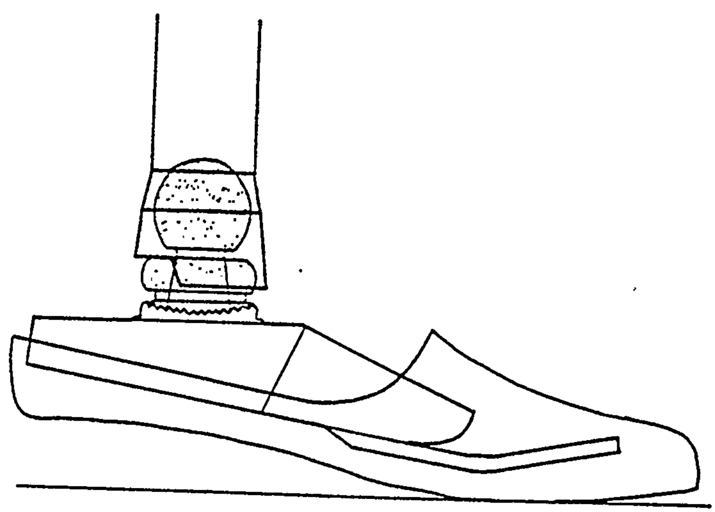

Figure 2.5: Multiflex Ankle-Foot (Condie, 1988)

In an attempt to improve the efficiency of the prosthetic feet on the market, "energy storing" designs have been manufactured. The intent of the "energy storing" designs is that the foot will store energy during the heel-strike and later return it for toe-off. Two designs marketed as “energy storing designs” are the Seattle foot and the Flex foot. Each design consists of a cantilevered spring which is attached to a resilient pylon or to a rigid pylon.

The Seattle foot consists of a keel constructed of Delrin, a Kevlar toe pad (to prevent the keel from penetrating the sole of the foot), and cosmetic external foam (Figure 2.6) (Lehmann et al., 1993). The prosthesis allows for 10 to 15 degrees of plantarflexion at the ankle joint (Lehmann et al., 1993). However, even though the Seattle foot is marketed as an “energy storing design” Prince et al., 1994 suggested that the viscoelastic material around the keel of the Seattle foot limited the recovery of energy and resulted in important dissipation (Prince et al., 1994). The design of the Flex foot varies from the Seattle by structurally integrating a laminated carbon fiber composite foot and shank segments (Figure 2.7) (Lehmann et al., 1993). The designs are customized to the subjects height, weight, and length of shank. The "energy storing" designs permit a more normal range of motion during stance phase and give the amputee a sense of active push-off (Michael, 1990). Schneider et al. (1993) reported that the Flex foot had a greater potential for reducing energy cost for walking at freely chosen speeds and fast speeds for children with a below-knee amputation. However, these designs have only resulted in minimal or no improvement in metabolic savings (Nielsen et al., 1989; Torburn et al., 1990; Barth et al., 1992).

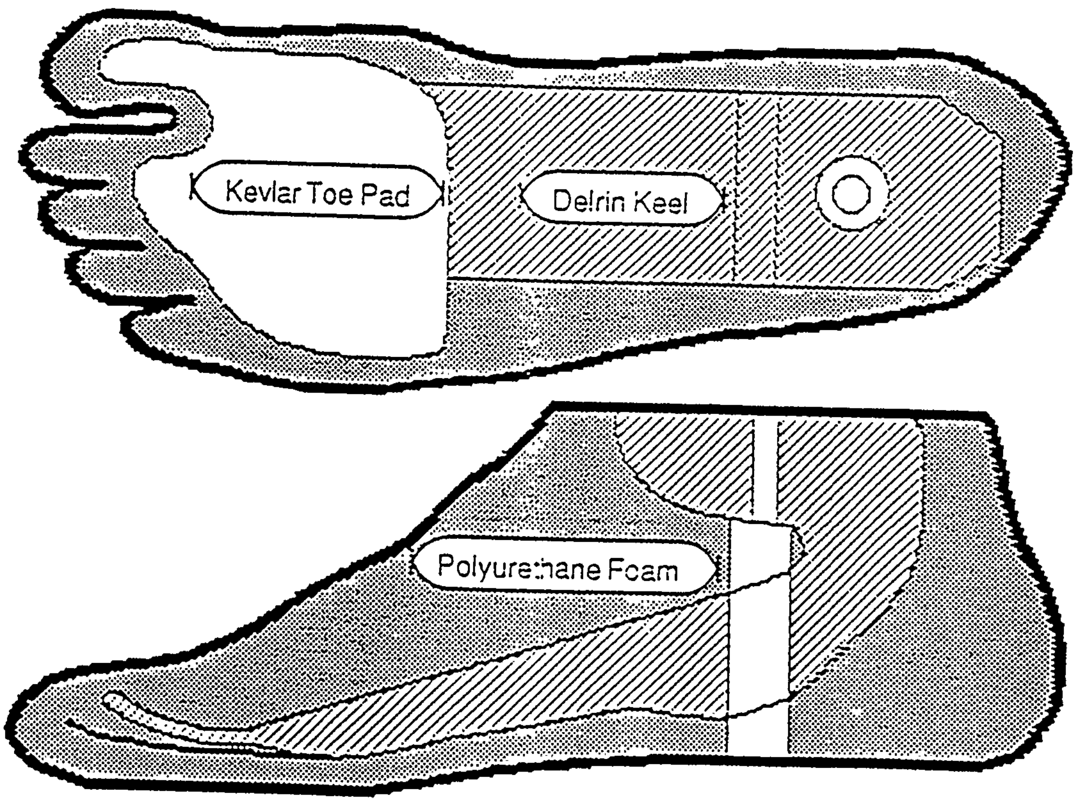

Figure 2.6: Seattle Foot; note cantilevered plastic spring keel to store energy (Michael, 1987)

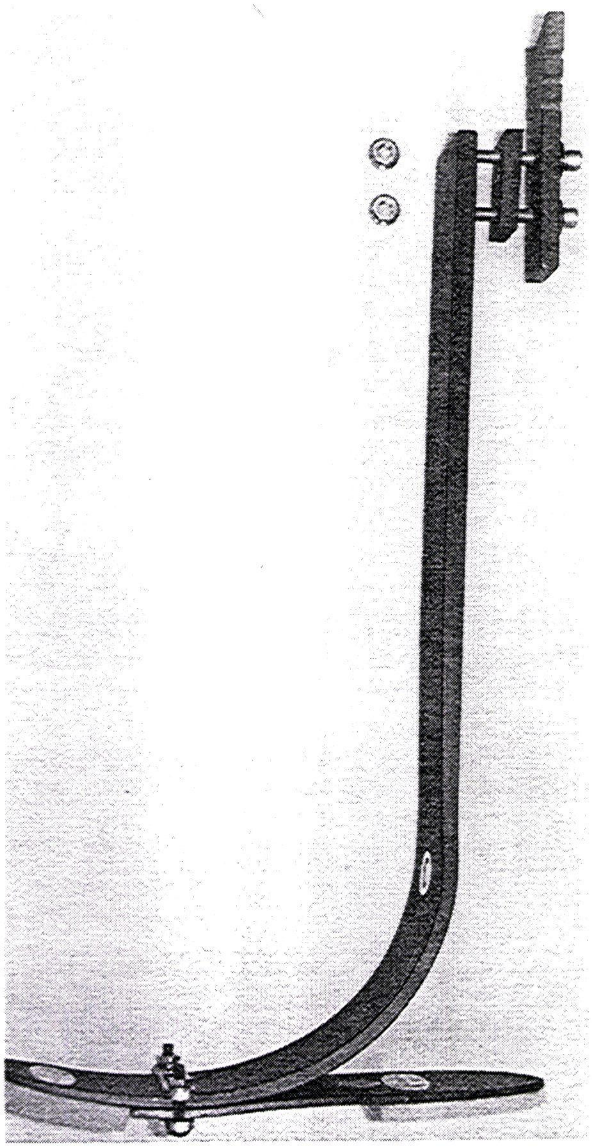

Figure 2.7: Modular Flex-Foot (Michael, 1987)

2.3.2 Summary Prosthetic Feet

The SACH and Uniaxial feet do not attempt to replace the dynamic function of the foot and ankle (Czerniecki et al., 1991). The multiaxial designs attempt to replace some of the range of motion of an intact ankle joint. The energy storing foot designs permit smooth transitions through stance. The energy storing designs attempt to partially reproduce the energy absorption and generation of the normal foot/ankle complex. They are inefficient, however, in regard to energy storage and return. The improvements in prosthetic design have not significantly lowered energy consumption (Barth et al., 1992; Nielsen et al., 1989; Torburn et al., 1990).

2.4 Literature Relevant to the Function of Prosthetic Foot/Ankle

There is a paucity of data available which characterize the gait of adults or especially the gait of children with below-knee amputations. The publications available typically have the limitation of a small number of subjects with BKAs. This becomes more limiting when comparing the effects of different prosthetic feet on the gait of an individual with a BKA. As a result, caution must be used when reviewing the results of these studies. The following section will present a brief overview of studies which have been conducted analyzing, kinematically and kinetically, the gait of individuals with BKAs, in contrast with individuals with an AB. The effects of different prosthetic feet on BKA gait will also be discussed. The aim of this section is to show the quantitative inadequacies of the present prosthetic ankle/feet designs, and derive a basis for the importance of this investigation.

Mizuno et al. (1992) conducted a study with 10 male adults with a BKA and 5 adults with an AB. Each subject, with a BKA, had data collected with 12 different feet. The 12 feet tested included 7 non-axial, 3 single axis, and 2 multi-axial. They looked at five variables and compared the results from the subjects with BKAs to the subjects with ABs. The step length was greater for the prosthetic limb than the non-prosthetic, and walking velocity were slower for the subjects with a BKA as compared to the subjects with an AB. There was a significantly ($p < 0.01$) lower vertical component in ground reaction force in the prosthetic limb as compared to the non-prosthetic and able-body limb. Mizuno et al. (1992) studied the efficiency of deceleration and acceleration, which was determined by adding the absolute values of the maximum deceleration force and the maximum acceleration force. They reported that the prosthetic limb was inefficient at decelerating and accelerating compared to the non-prosthetic and able-bodied limb. Mizuno et al., also reported irregular fore-aft components for two of the feet tested, the SACH and the Seattle, and suggested that the roll-over of the foot was not occurring smoothly (Mizuno et al., 1992)

Prince et al. (1992) studied the running gait patterns, at speeds from 2.8 m/s to 3.2 m/s, of 9 individuals with BKAs (mean age 16.4 years±3.8 years) and 6 individuals with ABs (mean age 22.2 years±3.5 years). They tested five of the subjects with BKAs wearing a rigid foot (SACH foot), four of the subjects with BKAs with flexible keel designs (2 Seattle and 2 S.A.F.E.), and retested 2 of the subjects with the Flat Spring foot. The investigators studied the vertical and the anterior/posterior ground reaction forces, as well as the impulses for these curves (Kistler force plate sampling at 600 Hz.). They reported asymmetry present between the prosthetic limb and sound limb, with all prosthetic feet. They noted that the asymmetry was less pronounced with the flexible keel designs. The flexible keel designs' asymmetry was associated with the force line modulation rather than magnitude alone, as reported with the other prosthetic feet.

Barth et al. (1992) studied 6 adult subjects with BKAs (3 traumatic and 3 vascular) with 6 separate prosthetic feet; SACH, S.A.F.E. II, Seattle Lightfoot, Quantum, Carbon Copy II, and the Flex-Walk feet. All of the feet were attached with an ankle bolt. Therefore, this study did not examine the effects of dynamic ankle or shin components. Each subject wore each foot for 3 weeks prior to testing. The Seattle Light foot exhibited no significant difference between the sound limb and prosthetic limb. The other five feet each had unique characteristics. The SACH foot had greater late-stance stability, the S.A.F.E. II foot would be preferred for walking on inclines or uneven terrain, the Quantum and Carbon Copy II showed greater weight acceptance forces than the sound limb, and the Flex-Walk had a greater range of motion. There was no significant difference in energy cost relative to the foot selected and there was also no significant change in speed due to foot selection. It could be interpreted, then, that no foot was more efficient than the other.

Engsberg et al. (1990) investigated the gait of 11 children with an AB and 3 children with a BKA. All the children with a below-knee amputation used a SACH foot. This investigation found an asymmetrical gait related to the timing of stance, swing, and double support. In 1992, Engsberg et al. reported greater flexion in the trunk angle of a child with a BKA than a child with an AB. There was reported a tendency for the center of mass to stay on the non-prosthetic side throughout the gait cycle in a child with a BKA.

Engsberg et al. (1993b) studied the walking gait of 225 children with an AB and 22 children with a BKA (12 wore a SACH foot, 4 a flex foot, 5 a Seattle, and 1 wore a single axis). They investigated the normative ground reaction forces from two consecutive foot falls, while walking at a speed of 1.2 m/s±10%, on Kistler force plates (sampling at 1000 Hz.). Engsberg et al. (1993b) reported a significant difference between the prosthetic limb and the non-prosthetic limb, and between the children with a BKA and the children with an AB for the discrete vertical, anterior/posterior, and medial/lateral forces. There were kinetic asymmetries in the gait of children with a BKA. They reported an increase in the rate and magnitude of loading in the non-prosthetic legs as compared to the legs of children with AB. Support time and impulse in the non-prosthetic limb were also increased. The authors suggest that BKA research should focus on eliminating or reducing the loading differences between the prosthetic and non-prosthetic limbs, and the limbs of BKA with limbs of an AB. One solution would be to develop a prosthesis to function more like an intact limb than present designs.

Herbert et al. (1994) investigated oxygen consumption for 10 children with a BKA and 14 children with an AB. The subjects walked for two minutes at four different speeds (freely chosen walking speed (CWS), 20% below CWS, 20% above CWS, and a fixed speed of 1.2 m/s). The results indicated that oxygen consumption was 14% greater for children with a BKA than those with an AB. The CWS was not significantly different between the BKAs and the ABs. This suggests that the biomechanical differences between children with a BKA and children without an amputation could significantly affect physiological function. Again, to allow a BKA to walk like a child with an AB, the authors suggest that the prosthetic limb would have to function similarly to an intact limb. The new prosthesis would need to permit dorsiflexion and plantarflexion movements and provide similar propulsive forces to those of an intact limb.

Scheirman (1982) studied the running gait of 5 adult BKAs. Their speed ranged from 2.0 m/s to 4.3 m/s. Two of the BKAs had a SACH foot, one used a S.A.F.E. foot, one a Greissinger, and one a leaf spring foot. Kinematic information was collected with a Locam 16 mm camera at 100 fps, and ground reaction force data were collected with a Kistler force platform at 1000 Hz. They reported that the intact limb segments consistently produced more mechanical energy than the prosthetic limb segments. They saw a trend that the "energy storing" designs transferred more energy and performed more work. This was attributed to the enhanced ankle mobility and increased "energy storing" capabilities in the keel. They noted, however, that due to the small numbers, they could not conclude whether differences were significant. Scheirman (1982) suggested that the optimal prosthetic foot would accept kinetic energy during impact and restore the energy for propulsion.

Torburn et al. (1990) conducted a pilot study comparing the "energy storing" designs (also referred to as dynamic elastic response [DER] prosthetic feet) with the SACH foot on five adults with a BKA. All subjects were tested, while walking, using four different "energy storing" designs. The four prosthetic feet used were the Flex foot, Carbon Copy II, Seattle, and the STEN. The subjects were randomly fit to one of the prosthetic feet following manufacturer guidelines and given a one month adjusting period before testing. The same procedure was followed until each subject had been tested using all 5 prosthetic feet. EMG, kinematic, kinetic, and oxygen consumption data were collected. The results showed stride characteristics similar for all feet. The flex foot was the only foot with different gait dynamics, with an increased dorsiflexion motion at the ankle and torque at the end of stance. However, the flex foot did not show significant improvements in energy cost. Subjectively, the subjects preferred the dynamic feet as compared to the SACH. The authors concluded though, that there were no clinically significant changes found in gait among the 5 prosthetic feet.

Lewallen et al. (1986) conducted a 2-dimensional study on 6 children with a BKA and 8 children with an AB while walking at their freely chosen speed. All of the children with a BKA wore a SACH foot. The study investigated the kinematics and kinetics using a high speed cine film system and a force plate. Inverse dynamics was used to study the moments at the ankle, knee, and hip. They reported that ankle plantarflexion was 10 degrees greater in the intact limb than that found in the subject with able-bodies. The knee kinematics showed no significant differences. However, kinematic differences were reported for the hip. Slight hip flexion was reported throughout stance in the children with a BKA. The children with a BKA had less hip flexion during swing phase than their counterparts with an AB. The joint moments were more consistent in the able-body controls than in the subjects with BKAs. In the non-prosthetic limb of the BKA there was an increase in the ankle dorsiflexion moment. The knee and hip joint moments were normal or below normal for the prosthetic limb. They concluded that the forces in the sagittal plane of the joints in the intact limb could not be implicated as a major cause in degenerative joint disease. Lewallen et al. (1986) also reported a decrease in walking speed and step length for the children with a BKA. An increase in the double support phase and stance phase for the child with a BKA, and this may contribute to the lower, non-degenerative forces. The researchers suggest that the adaptation of slower gait reduced the loading on the joints. One should be cautious with the authors' conclusion, however, since the mechanism for degenerative joint disease is not fully understood.

Lewallen et al. (1986) also, reported the ankle plantarflexion moments, normalized to body weight and height, were approximately .55 Nm/kg.cm for the prosthetic limb and approximately .83 Nm/kg.cm for the able-bodied limb. The authors note that the moments of the amputated extremity were lower than that of the intact limb, therefore acting as a passive support. The moment curves in the subjects with a BKA and those with an AB did parallel each other.

Winter and Sienko (1988) investigated walking, at a freely chosen speed, of five below-knee amputees. They compared the BKA results with profiles from populations of able-bodies. All of the subjects wore SACH feet. Two of the BKAs were retested with uniaxial feet and one subject was also retested with a Greissinger foot. Winter and Sienko (1988) reported kinematics and kinetics on all five subjects. EMG was studied on three subjects. It was reported that the below-knee amputee subjects had hyperactive hip extensors and seemed to coincide with above normal energy generation in the concentrically contracting muscles. It was hypothesized that this was a partial compensation for the lack of energy generation during push-off, due to the absence of the plantarflexors. The ankle moment in the three types of prosthetics was 2/3 of that seen in populations with AB. The ankle moment followed a similar pattern for all three prosthetic feet. Figure 2.8 shows a comparison between the ankle moment for a typical subject with a BKA who wore a SACH foot, with the average data from 18 subjects with ABs. Figure 2.8 was a typical pattern seen in a BKA.

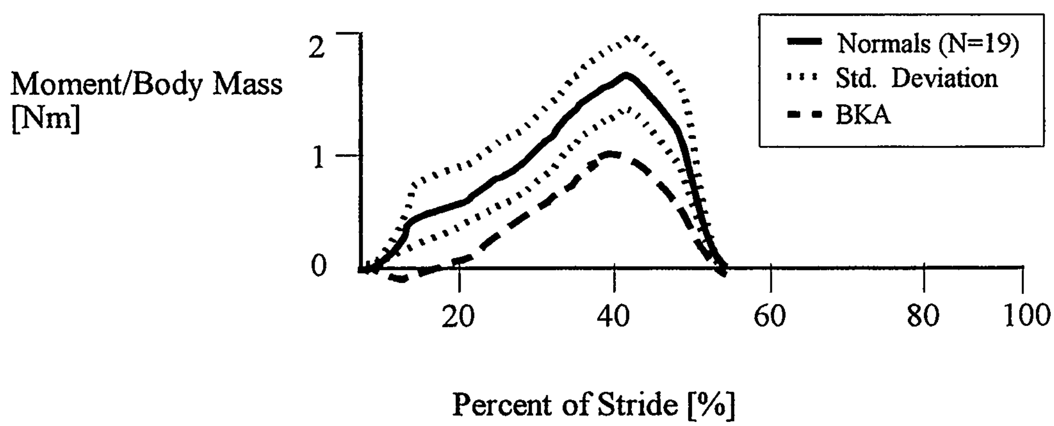

Figure 2.8: Moment of Force Patterns For One of The Amputees Fitted With a SACH Prosthesis Overlaid on Top of Intersubjects Averages From Normals (Winter & Sienko, 1988).

Winter and Sienko (1988) investigated the prosthetic limb's efficiency as percent of return of energy storage. Negative power reflects the rate of absorption by any built-in mechanism (i.e., damper or spring) while positive power reflects the return of stored energy from any spring mechanism. The percent of return was positive power divided by negative power (Winter and Sienko, 1988). The uniaxial feet showed a 20% efficacy from energy storage to release in walking (Winter and Sienko, 1988). The plantarflexion moment during toe-off, in the uniaxial foot and the SACH, was approximately 2/3 of that seen in subjects with an able-body (Winter and Sienko, 1988). The SACH foot had a reported 30% energy recovery (Winter and Sienko, 1988). Similar to the uniaxial and SACH feet, the plantarflexion moment reported for the Greissinger prosthetic foot was 2/3 of that of an intact limb (Winter and Sienko, 1988). They hypothesized that any human system with major structural asymmetries in the neuromuscular skeletal system cannot be optimal when the gait is symmetrical. They suggested that a new, non-symmetrical optimal should be sought by the amputee within the constraints of the residual system and the mechanics of an amputee's prosthesis.

Miller (1987) studied the resultant running flexion/extension moment in the ankle of four adults with BKAs. Two of the subjects had Greissinger feet and two had SACH feet. The subjects ran (all subjects were rear-foot strikers) at a self-selected running speed (ranging from 2.5 to 5.7 m/s) on a 30 meter runway. Kinematic information was collected with a 16 mm Locam camera at 90-100 fps. Kinetic information was collected with a Kistler force platform. The position-time data were smoothed using a second order symmetric Butterworth filter. The cut-off frequency was selected using the procedure by

Jackson (1979), and ranged from 4 to 8 Hz. Differentiation of the filtered data was accomplished with a first order central finite difference technique. The ankle angle used was defined as the angle described by the estimated knee joint center (lateral femoral condyle), the estimated ankle joint center (lateral malleolus), and the metatarsal-phalangeal joints. During running, the ankle went through a brief dorsiflexor moment, then displayed a dominant plantarflexor moment for both the intact and the prosthetic limb. The moment of the intact limb was greater. The experimental peak values for the plantarflexion moment were not reported, but the pattern was similar in all the subjects. The peak occurred in the prosthetic limb earlier than in the intact limb (Figure 2.9). Figure 2.9 shows that after the peak, the ankle moment in the prosthetic limb gradually diminished during the remainder of stance. This pattern was similar in the four subjects. The ankle moment in the prosthetic limb reflected the moments transmitted between the shank and foot. Miller (1987) conducted a sensitivity analysis that suggested that greater difference in the running joint moments during stance resulted from changing the center of force, rather than the inertial characteristics of the leg.

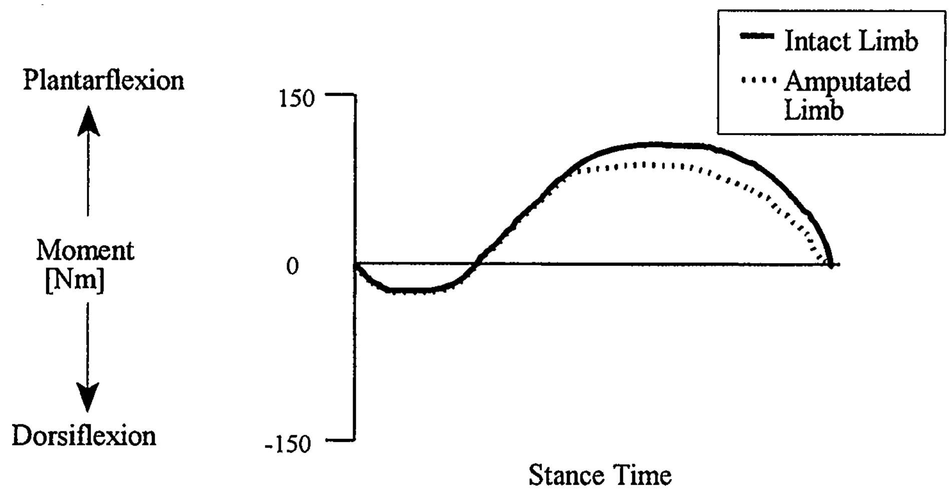

Figure 2.9: Resultant Ankle Joint Moment-Time Histories During Stance for an Individual with a BKA Running at Speed Averaging 2.6 m/s (Miller, 1987).

Czerniecki et al. (1991) studied the stance phase moments, muscle power outputs, and mechanical energy characteristics while running at 2.8 m/s. The volunteers included 5 subjects with ABs and 5 with BKAs. The subjects with BKAs were given three different feet in a random order (SACH foot, Seattle foot, and Flex foot), and were given 1 week to adjust to each foot prior to testing. The subjects ran on a 20 m runway and kinematic data were recorded with a video camera, later digitized at 60 Hz. Kinetic information was collected with a Kistler force platform (500 Hz) located in the middle of the runway. Coordinate data were smoothed with a second order, zero lag, Butterworth digital filter at 7 Hz. One of the subjects with a BKA was not included in the energy storing design due to considerable deviation from the other amputees. The peak power output for the Seattle foot was 2.5 times greater than that of the SACH foot, and the Flex foot had a peak power 3.4 times that of the SACH foot (Table 2.1).

Table 2.1: Stance Phase Peak Power Outputs (Watts) in Subjects With ABs and BKAs (Czerniecki et al., 1991)

	Muscle Group	AB Mean±SD (Watts)	SACH Mean±SD (Watts)	Seattle Mean±SD (Watts)	Flex Mean±SD (Watts)
Ankle	Eccentric Plantarflexion	433±109	331±141	515±155	422±139
	Concentric Plantarflexion	876±239	96±31	260±93	344±107

The power output pattern of the lower extremity in subjects with a BKA wearing a conventional prosthetic foot (SACH) was considerably different from the subjects with

ABs. The prosthetic foot generated a negligible amount of energy, and lacked a source of energy generation. The energy storing components, especially the Flex foot, reduced the asymmetry in the gait pattern of four of the five BKAs. The total work done by the subjects with a BKA while wearing the SACH foot was 49.5% of that of the subjects with ABs. When the BKAs used the flex foot, it increased to 70% of that of the ABs. This suggests that the amputees absorbed and generated less energy with their SACH prosthetic limb. Czerniecki et al. (1991) reported that the flex foot was a major energy generator as compared to other prosthetic feet. The SACH foot generated 31% of the energy it absorbed, the Seattle foot 52%, and the Flex foot 84%. Therefore 69%, 48%, and 16%, respectively, of the energy was lost in the system, while a similar analysis of an intact limb shows that the plantarflexors generate approximately 241% of the energy absorbed. This implies that more energy was generated than could be absorbed into the system. This study's calculations of power were limited by the underlying assumption, of a link segment model, which states that the body is a series of rigid segments linked by frictionless joints. The prosthetic feet used in this study were rigidly fixed to the shank, not linked to the prosthetic shank by a frictionless joint. The center of rotation changes in a prosthetic foot, as there is deflection in the keel or in its foam elements. Czerniecki et al. (1991) however, noted that the power outputs were reasonable approximations.

The following two investigations introduce other methodologies that have been used in determining energy return in prosthetic feet. Prince et al. (1994) studied the energy efficiency of the SACH, Seattle, and Flex foot with 5 subjects with BKAs. Kinematic and kinetic data was collected and processed with an inverse dynamic analysis.

However, unlike the methods used in this investigation, Prince et al. (1994) separated the joint power equation, calculating the power of the foot and the power of the leg. Prince et al. (1994) compared the results of this new methodology, to results attained by using the widely accepted method of calculating energies at the ankle using the time integral of ankle joint powers. Prince et al. (1994) concluded that their new methodology allowed them to account for the energy stored by the heel, ankle, and forefoot, and produced 2 phases of energy absorption and 2 of energy generation. This new information was attained when studying a prosthetic foot that has no center of rotation, and has not been confirmed by other researchers. As previously discussed it has been accepted that there are one phase of energy absorption and one phase of energy generation during stance. (Miller, 1987; Winter and Sienko, 1988; Czerniecki et al., 1991). Since the methodology has not been applied to human feet, it is unknown how the information reported by Prince et al. (1994) affects, if at all, the results to be reported in this present investigation, it may be important information to consider when designing a prototype.

Allard et al. (1994) compared energy results calculated for a prosthetic foot when using the ankle joint powers and using a finite element model. Allard et al. (1994) reported that the energies calculated with ankle joint powers had a high correlation in energy variation as those calculated with the finite element model. However, the ankle joint powers resulted in a 4.2 overestimation of the energies. Allard et al. (1994) conducted this study on a prosthetic foot, and this current study focus is on an intact foot. It would be extremely difficult to use a finite element model on an intact foot. These results may indicate that energies determined with joint powers will be an overestimation, and this would be important information when testing a prototype.

2.4.1 Summary:

In summary, there are kinematic and kinetic differences between populations of ABs and BKAs. These differences appear to be significant in the lives of amputees due to a possible greater propensity to the prevalence of degenerative joint disease in the BKA population (Burke et al., 1978). There also is an increase in energy requirements reported for both children and adults with BKAs (Herbert et al., 1994; Gonzalez et al., 1974). It is unknown whether an increase in energy cost inhibits the lifestyle of a BKA. It has been hypothesized that the loading asymmetry and metabolic inefficiency cannot be reduced until the prosthetic leg functions similarly to an intact leg (Engsberg et al., 1993a, b). To accomplish this task, an area which requires investigation is the design of the foot/ankle assembly. The current designs of prosthetic feet and ankles use a spring for energy absorption and generation. Power absorption and generation can determine if a spring alone can produce sufficient energy for push-off. Czerniecki et al. (1991) studied power input-output, and concluded that more energy was generated than absorbed in an intact limb. Miller et al. (1987) showed that the prosthetic limb had a lower peak moment than the intact limb. The moment did parallel the intact moment, but at a lower value (Miller et al., 1987). Lewallen et al. (1986) and Winter & Sienko (1988) reported a plantarflexion moment 2/3 of that seen in an able-bodied individual. These results imply that the current prosthetic designs which only implement a spring (a passive system) will not produce the required energy necessary for adequate plantarflexion during propulsion phase of gait. This suggests that the spring must be aided by an auxiliary power source.

The purpose of this investigation was to determine the energy and moment requirements for a spring system incorporated into a theoretical prosthetic ankle joint. With the spring system permitting the theoretical prosthesis to have the same energy and moment characteristics identical to an intact ankle joint during gait. The spring component of a below-knee prosthesis can be described by investigating the moment with respect to ankle position. However, there were no published data specifically investigating the relationship between ankle moment and ankle position. The methods will discuss the techniques used to determine whether a spring alone could provide the energy characteristics similar to an intact limb and determine the characteristics of a theoretical prosthetic ankle joint.

3. Methods

3.1 Overview

As discussed in the literature review, current prosthetic designs rely on a passive spring system to absorb and generate energy. However, these passive systems appear to be insufficient in generating the energy produced by an intact limb. The lack of energy results in inadequate force production during plantarflexion, phase 3 of the gait cycle. This inadequate force production, in turn, may contribute to an asymmetrical gait cycle that is different from that of children with able bodies. Thus, if adequate energy were present in phase 3 of the gait cycle, adequate force production could be present and the asymmetries in a child with a below-knee amputation could be minimized if not eliminated. The purpose of this investigation was to determine the energy and moment requirements for a theoretical spring system for a prosthetic ankle joint. This spring system would permit the theoretical prosthesis to have the energy and moment characteristics identical to an intact ankle during gait.

This chapter is divided into two main sections. The first section (3.2) describes the methods of collection of experimental kinematic and kinetic data. This includes the subject characteristics, equipment setup, subject preparation, data acquisition, and data processing, including all kinetic formulas. The results from the first section were then used to determine the energy and moment requirements for a theoretical spring system for a prosthetic ankle joint. These methods are described in section 3.3. This section will first discuss the methods to quantify the amount of experimental energy absorbed and generated at the ankle joint during gait. These results would confirm or contradict the results of Czerniecki et al. (1991) in determining if a spring alone could produce the necessary ankle moments in a below-knee prosthesis to reproduce the ankle moments of intact ankle joint during gait. If the amount of energy absorbed was greater than that generated, a spring could be a sufficient system, however if the energy absorbed was less than that generated an auxiliary energy source would be added to the spring system. The supplementary energy would permit the theoretical prosthesis to have energy characteristics identical to an intact ankle during gait. The energy data will define the mechanism (i.e., spring or spring and auxiliary energy source) for the theoretical prosthesis. Then the methods used to theoretically reproduce the ankle moment data of a child with an able-body, will be discussed along with the methodology for evaluating the theoretical spring system. The theoretical spring system utilized regression equations to estimate the ankle moment of a child with an able-body from their kinematic information. First the optimal spring constants were determined via a non-linear regression equation. Then if the energy data showed that a spring could not produce sufficient energy to duplicate the ankle moments in an intact limb, an auxiliary energy source that would provide a moment generator with the necessary energy to aid the spring during plantarflexion, prior to toe-off. The auxiliary source would be a mechanism for producing energy (e.g., motor). An ideal system would allow the same theoretical system to be used for walking and running. This would indicate that one could walk and run with the same prosthetic leg, and achieve ankle moments identical to those in an intact limb. These results are considered theoretical because it presents an idea of how the ankle moments could be reproduced, but it is unknown if in experimental gait conditions this theory would be feasible. That aspect is beyond the scope of this project.

3.2 Experimental Data Collection

3.2.1 Subjects

Three children with able-bodies, all female subjects, volunteered for this investigation (Table 3.1). All three subjects had walking data collected, subjects 1 and 2 also had running data collected. Testing was conducted on a single day at the Human Performance Laboratory at the University of Calgary, Calgary, Alberta, Canada. All children were familiarized with the laboratory and testing procedures prior to collection of data.

Table 3.1: Subject Characteristics

Subject Number	Age (Yrs)	Height (cm)	Mass (Kg)
1	8	129.5	24.0
2	10	146.0	31.5
3	14	164.5	56.4

Children with able-bodies were used as subjects to determine the energy and moment characteristics of an intact ankle system during gait. The energy and moment information attained were then used to recreate the intact system, in a theoretical prosthetic ankle joint system. The results provided us with the information necessary to achieve our purpose of determining the energy and moment requirements for a spring system incorporated into a theoretical prosthetic ankle joint.

3.2.2 Equipment Setup

Figure 3.1 shows the laboratory setup that was used for data collection. The force data (1000 Hz) were collected by a Kistler force platform (Model 9287; 60 cm x 90 cm). The force platform was located halfway along a 10 meter walkway. The video information was recorded in three dimensions using four high speed video cameras (200 Hz; Motion Analysis Corporation, Santa Rosa, CA.), but only a two dimensional, sagittal plane (x and z axis) was used for this study. The cameras were arranged in an umbrella formation (Figure 3.1), with lights directly above or beside the camera to illuminate the reflective markers. All cameras were focused on the data collection volume, and thresholds were adjusted according to guidelines by Motion Analysis. The gait speed was monitored by 2 photocells placed 2.4 meters apart. Prior to testing, the cameras were calibrated. Four strings with three reflective balls on each string, with the balls set at a height 0.2000 m, 0.4000 m, and 0.7000 m were used to describe a control volume 42 cm long, 70 cm high, and 70 cm wide. Calibration was performed with ExpertVision software, by Motion Analysis Corporation (Santa Rosa, CA.).

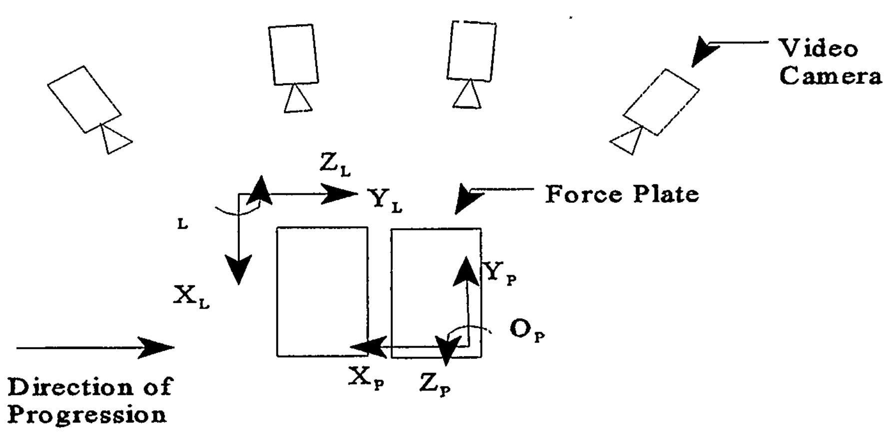

Figure 3.1: Schematic of laboratory setup, O_L is the origin of the laboratory coordinate system. The origin of the plate coordinate system was centered on the force plate with X_P running opposite Y_L, and Y_P running opposite of X_L. The axis in the laboratory coordinate system (Z_L) points up and Z_P points down.

3.2.3 Subject Preparation

Before testing, the subjects and their parents/guardians were informed of the purpose of the test and the testing procedures. All subjects wore shorts and the standard lab shoe of the Human Performance Laboratory. Three spherical reflective markers were secured on each subject's left leg and left foot. One marker was placed at the fifth metatarsal, one on the lateral malleoli, and one on the lateral head of the fibula (Figure 3.2). For 2-Dimensional calculations, the lateral malleoli has been used as the estimated ankle joint center, by several investigators (Miller, 1987; Winter and Sienko, 1988; Czerniecki et al., 1991; and Prince et al., 1993). Noting however that the lateral malleoli does not represent the true axis of rotation in 3-dimensions.

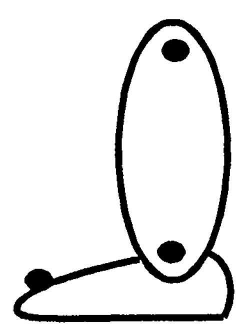

Figure 3.2: Placement of Reflective Markers on Leg and Foot.

3.2.4 Data Acquisition:

Data were collected by the "KinTrak" program by Motion Analysis. The subjects walked at their freely chosen walking speed (Table 3.2) and data from at least five successful walking trials were collected for all three subjects. Running data were then collected for subjects 1 and 2. Subjects 1 and 2 ran at their freely chosen running speed (Table 3.2) and data from at least 5 successful trials were again collected for each subject. A successful trial was one in which the left foot landed on the force platform within the space described by the force transducers and corresponding video data were collected.

Table 3.2: Freely Chosen Walking and Running Speed

Subject Number	Age (Yrs)	Average Walking Speed (m/s)	Average Running Speed (m/s)
1	8	1.68	3.24
2	10	1.69	3.04
3	14	1.86	N/A

3.3 Data Processing:

Figure 3.3 shows a flow chart of the complete data processing procedure. The data went through four programs for processing: "KinTrak" (Motion Analysis Corporation), "Expert Vision" (Motion Analysis Corporation), "GaitSync", and "Ankle". "KinTrak" was used to collect the force and video data. "KinTrak" was used to export the relevant experimental force plate data to ASCII format. This file included: time in seconds, force in the x, y, and z direction (Fx, Fy, and Fz respectively), location of the force on the force platform (Px, and Py), and the moment about the z axis (Mz). The video trials were tracked in "Expert Vision" to obtain xy coordinates of the surface markers as a function of time. Then "KinTrak" was used to export the video data to an ASCII file. This file included: the frame number, the time, and the x, y, z coordinates of the 3 markers throughout the gait phase. No smoothing was done by "KinTrak". If any of the trial data were missing, it was left absent; necessary interpolation was done by the "GaitSync" program.

A custom program called "GaitSync" synchronized the force and kinematic data for each trial. "GaitSync" filtered, interpolated, and resampled all the data with a cubic spline function at intervals of 1% of the support phase. The force information consisted of: time in seconds, the force in the x, y, and z direction, the location of the force; Px, and Py, and the moment about z. The kinematic data file consisted of the coordinates (x, y, z) of marker one, two, and three. The data were filtered with a cubic-spline with a cut-off frequency of 60 Hz and then resampled from -10% to 110% of stance, at intervals of 1% (Woltring, 1986). The information before heel-strike and after toe-off was included to ensure a complete record of the events between initial contact and the end contact. The "GaitSync" program produced one file which consisted of: time in percent of support phase, time in seconds, Fx, Fy, Fz, Px, Py, Mz, X coordinates marker 1, Y coordinates marker 1, Z coordinates of marker 1, X coordinates of marker 2, Y coordinates of marker 2, Z coordinates of marker 2, X coordinates of marker 3, Y coordinates of marker 3, and Z coordinates of marker 3.

The output data from the "GaitSync" were then input into a custom program, "Ankle". The following calculations performed by "Ankle" were simplified by limiting the study to 2-Dimensions, sagittal plane, where the majority of ankle motion occurs. Noting, however, there is ankle joint motion in the coronal and transverse planes. Another limitation in the calculations performed by "Ankle" was the omission of the mass and inertial forces on the foot. This omission has been reported to have minimal effects on the results, due to the fact that mass and inertial forces are so small in comparison to the external forces (Morlock and Nigg, 1991; Stauffer et al., 1977).

The "Ankle" program performed the following calculations; the ankle joint angle in the sagittal plane (Figure 3.4), the joint angular velocity of the foot, the moments about the ankle joint, the power at the ankle joint, and the total energy requirements of the ankle joint. The equation for these calculations will be discussed next. The angular position was determined as the angle described on the foot segment by markers 1 and 2, and the shank segment described by markers 2 and 3 (Figure 3.4).

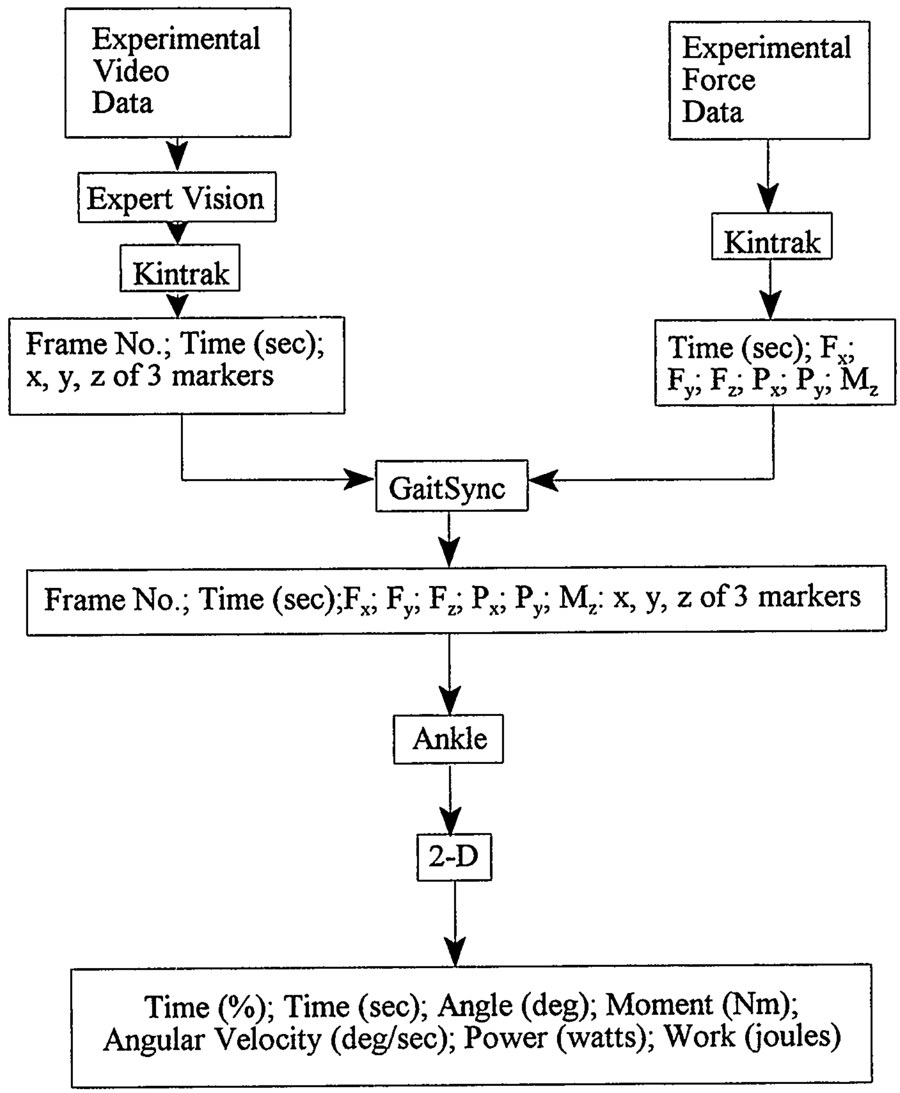

Figure 3.3: Flow Chart of Data Processing

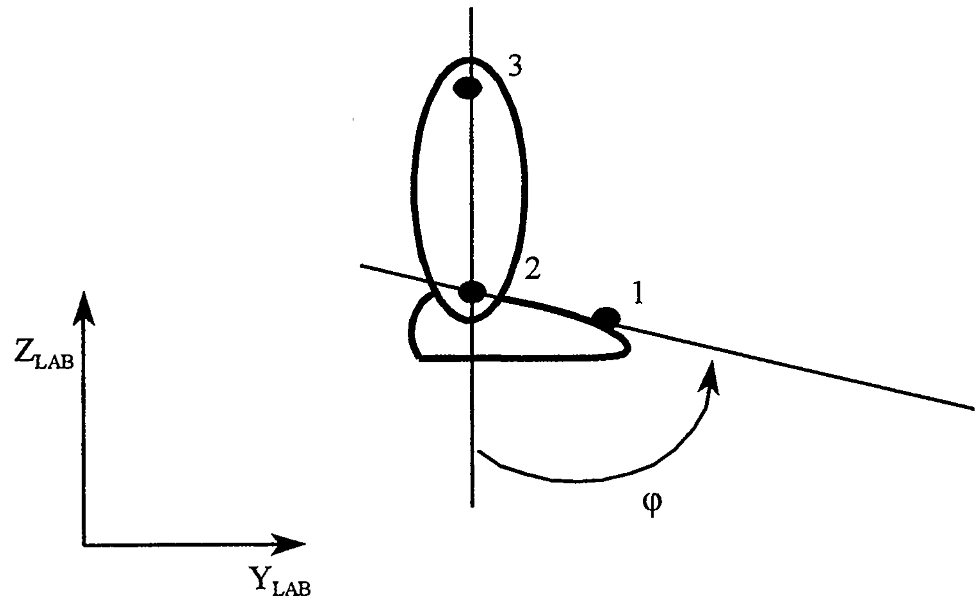

Figure 3.4: The Measured Angular Position, Measured Angle φ and the Corresponding Lab Coordinate System.

The joint angular velocity (Equation 3.1) was calculated using a cubic spline, with a 20 Hz cut-off as described by Woltring, 1986.

$\omega = d\varphi/dt$ (Equation 3.1)

where: ω: angular velocity (units - deg/sec)

φ: angular position of the foot segment with respect to the shank

t: time

The moment (Nm) (Equation 3.2) was defined as the moment about the axis created by marker 2, which was placed on the lateral malleoli head. Dorsiflexion was in the positive direction and plantarflexion was in the negative direction and the unit of measure was Newton-meters (Nm).

$M= (x_F-x_J)\, F_z - (z_F-z_J)\, F_x$ (Equation 3.2)

where: M: the moment about the ankle

F_z: the ground reaction force in the z direction

x_F: the location of the point of application of the force in the x direction

x_J: the location of the joint center in the x direction

F_x: the ground reaction force in the x direction

z_F: the location of the point of application of the force in the z direction

z_J: the location of the joint center in the z direction

Instantaneous power (Watts) was calculated with equation 3.3. This equation assumed a rigid joint model. However, Allard et al. (1994) and Prince et al. (1994) report that this is an incorrect assumption because a rigid model does not account for energy losses and recovery in the viscoelastic material of the heel. Allard et al. (1994) reported though that there was a high correlation established when using muscle power (Equation 3.3) and the finite element model, yet reported that muscle power results must be evaluated with caution. Energies calculated for a prosthetic foot by integrating muscle power (Equation 3.4) were about 4 times higher than energies calculated by a finite element model (Allard et al., 1994). Noting that this investigation uses an intact foot.

$P=M\omega$ (Equation 3.3)

where: P: Instantaneous power

M: ankle moment

ω: angular velocity (rad/sec)

The energy (Joules) (Equation 3.4) required to produce the ankle moment was calculated for each trial by integrating the power over the duration of the support phase.

$E = {}_{0}\int^{T}(M^{*}\omega)\, dt$ (Equation 3.4)

where: E: Energy

T: duration of the support phase

M: ankle moment

ω: angular velocity

The methods have now been described for attaining the experimental kinematic and kinetic results. These data were then used to determine the energy characteristics of the ankle joint during gait. The energy characteristics along with the other kinematic and kinetic data collected from the able-bodied children will help define the requirements for the theoretical spring system of a prosthetic ankle joint.

3.3.1 Theoretical Spring System

The kinematic and kinetic data were then used to determine the requirements for the theoretical spring system. The first requirement in the development of the theoretical spring system was to determine the amount of energy absorbed and generated. This determined the mechanisms for the spring system to produce ankle moments similar to an intact ankle (i.e., spring alone, or spring coupled with an auxiliary source). Then a theoretical spring system model was designed which would duplicate the ankle joint moments during gait. Regression equations were used to determine appropriate values for the ankle moment from time and foot angular position using averaged experimental trials for each subject. Figure 3.5 shows a flow chart of the regression equations used, for walking and running. The equations will be discussed later.

The software used to calculate the regression equations was SPSSx (Statistic Package for Social Sciences). The experimental ankle moment was the dependent variable in the regression equations and the experimental time and foot angular positions were independent variables. The experimental data (time and foot angular positions) were then input into the theoretical model comprising the calculated regression equations. The quantitative differences between the experimental and the theoretical model results were then examined, and quantitative evaluation of the theoretical models ability to predict the experimental response of an intact ankle could then be made. Due to the small number of subjects in this present investigation additional research would be necessary to confirm these preliminary results.

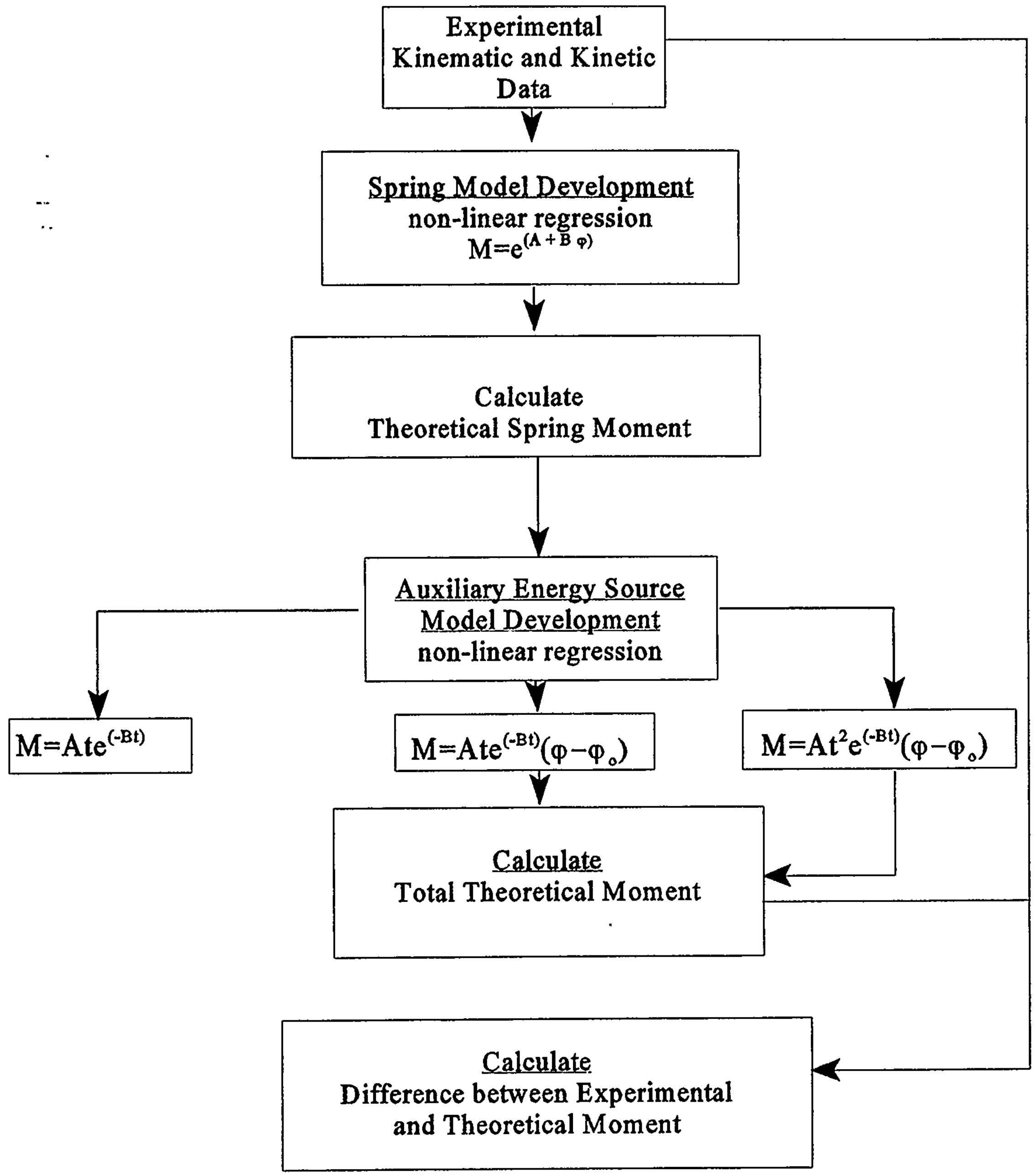

Figure 3.5: Equation and Flow Chart for Model Development

3.3.2 Energy at the Ankle Joint

Equation 3.4, $E = {}_0\int^T (M*\omega)\, dt$, was used to quantify whether additional energy was needed above that which a spring could provide in a theoretical prosthetic ankle. As previously stated, if the results showed more energy absorbed then generated, a spring system could provide sufficient ankle moment for phase 3 of the gait cycle. However, if the generated energy was less than that absorbed, a spring system could not provide sufficient energy for push-off, during gait. An auxiliary energy source would then need to be added to the spring system. The auxiliary energy source plus the spring could create a theoretical ankle moment similar to an intact ankle joint. The supplemental energy source would aid the spring during plantarflexion to offset the springs' inability to provide adequate plantarflexion during phase three of the gait cycle.

The results from the experimental ankle moment showed that the ankle moments during the energy generation phase were greater than those during the absorption phase. The additional energy needed for the auxiliary source is illustrated by the shaded area in Figure 3.6, the result section will provide the amounts of energy needed. The graphical representation the ankle moment as a function of angular position of the foot, determined experimentally, were similar for all subjects. Figure 3.6 illustrates how shortly after initial contact the ankle moment begins to increase with an increase in the angular position of the foot. This represents the energy absorption phase. After the peak angle, the energy generating phase begins until end contact. The ankle moments in Phase 3 have higher values for the same angular position in phase 1. Therefore, more energy is generated from the system than absorbed. Suggesting that it would be necessary for an auxiliary energy source to aid the in phase 3 of the gait cycle to enable the theoretical prosthetic ankle to have the same force generation, and ankle moment characteristics as an intact ankle joint. These are also similar results as Czerniecki et al. (1991). Visual examination of the moment/angle curve (Figure 3.6) determined that the optimal reproduction of the curve would be with a non-linear model.

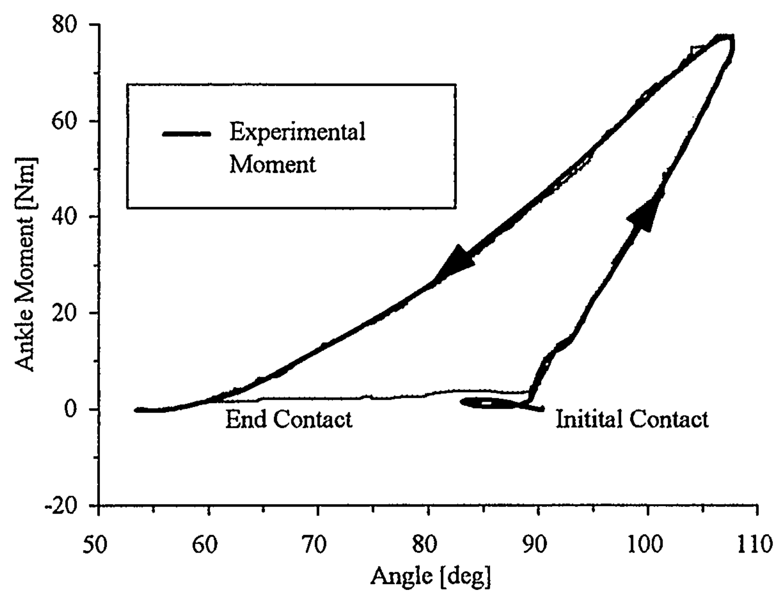

Figure 3.6: Experimental Ankle Moment Versus Angular Position of the Foot

3.3.3 Overview

Section 3.3.2 indicated that a spring alone could not provide sufficient energy for push-off during gait. Thus a spring system consisting of a spring coupled with an auxiliary energy would be the necessary theoretical mechanism to duplicate the experimental ankle joint moment. The following section discusses how the experimental information was used to derive the specifications for the theoretical spring system for gait. The specifications included the amount of moment needed to be produced throughout stance in the theoretical system to duplicate those in the experimental system during walking and running gaits. Regression equations provide a basis for estimating the values of one variable from the knowledge of another. For instance, if a good relationship existed between ankle moment and angular position of the foot then a regression equation could be developed that would predict the ankle moment from the foot position. In this investigation regression equations were the tool used to reproduce the ankle moments of an intact ankle joint in a theoretical prosthetic ankle joint. As mentioned previously, the ankle moment was the dependent variable, and time and foot angular position were independent variables.

3.3.4 Modeling Prosthetic Ankle System

To simplify regression equation calculations the experimental data were modified into three phases of stance, identical to the three phases of the gait cycle (Figure 3.7). Phase One was defined as the very beginning of the trial when the angular velocity of the ankle was decreasing. Phase One represents the period following heel strike when controlled lowering of foot was occurring. Phase One was not used in calculating the regression equations. Phase Two, energy absorption phase, began at the instant the angular velocity began to increase, and continued until the ankle angle began to decrease again, representing dorsiflexion. Phase Three, energy generation phase, began when the ankle angle began to decrease and continued to the end of stance. Phase Three represents plantarflexion.

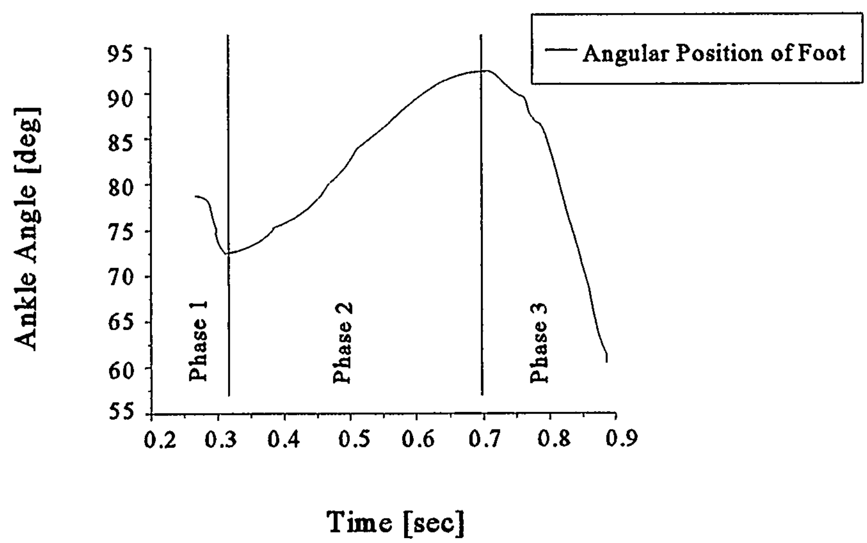

Figure 3.7: Three Phase Delineations of Data File Shown on Angular Position of the Foot with Respect to the Shank Versus Time Graph

3.3.4.1 Theoretical Spring

Phase Two was used to define the theoretical spring constants. To determine the spring constants the non-linear regression, $M = e^{(A+B\ \varphi)}$ was used (Equation 3.5). The input into the equation was the experimental ankle moment during Phase One, and experimental angular position of the foot. The output was the spring constants A and B.

$$M = e^{(A+B\ \varphi)} \qquad \text{(Equation 3.5)}$$

where:

M = Moment Phase One

e = exponential

A = Spring Constant

B = Spring Constant

φ = Angular Position of the Foot

3.3.4.2 Theoretical Auxiliary Energy Source

Figure 3.8 illustrates the difference between the experimental Phase Two and the calculated ankle moments throughout the stance phase. The calculated moment was attained by inputting the measured angles into the theoretical spring equation 3.5. The solid line is the experimental ankle joint moments. The dotted line indicates the theoretical spring. The shaded portion represents the difference between the experimental Phase Two and the calculated theoretical ankle moment. This difference was the amount the theoretical auxiliary energy source needed to provide. Non-linear regression equations were used to calculate the formula for the theoretical auxiliary energy source. Recalling that the theoretical auxiliary energy source would supplement the energy stored by the theoretical spring during Phase 2 (plantarflexion). The theoretical auxiliary energy source would provide the necessary energy to supplement the theoretical spring during plantarflexion and result in a theoretical ankle moment similar to an intact ankle moment.

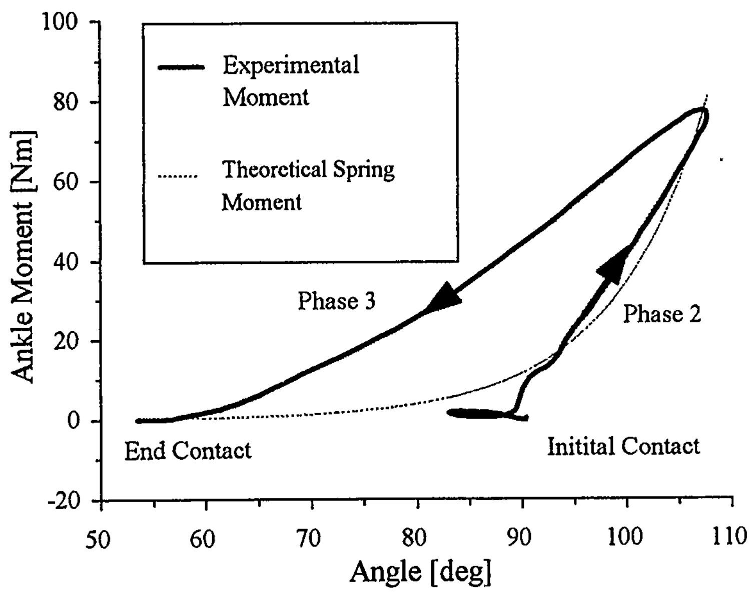

Figure 3.8: Ankle Moment Versus Angular Position of the Foot. The Experimental Moment is Shown by a Solid Line and the Theoretical Spring Moment is Shown by the Dotted Line. The Shaded Portion Represents the Energy the Spring is Unable to Provide to the System (the Quantity of the Auxiliary Moment).

The exact point when the spring would have zero angular velocity was determined. This is the same point as when the ankle angle begins to decrease and plantarflexion begins, Figure 3.7. Time was reset to zero at this point, to determine the regression constants to reproduce Phase Three. The first regression equation attempted was equation 3.6.

$$M = Ate^{-Bt} \qquad \text{(Equation 3.6)}$$

where:

M= Moment difference between Experimental and calculated

A = Constant

t = Time in seconds (reset to zero at the beginning of Phase Three)

e = exponential

B = Constant

Equation 3.6 did not allow the theoretical system to return to zero at the end of stance. The ankle moment returning to zero allows the ankle joint to prepare for the swing phase, allowing the foot to clear the floor. To aid the system into returning to zero (φ - φ_0) was added to equation 3.6 deriving equation 3.7. Equation 3.8, squares time, in an attempt to improve the simulated ankle moment. For Equations 3.7 and Equation 3.8, the dependent variable was the necessary auxiliary moment, and the independent variables were time and angular position of the foot.

$$M = Ate^{-Bt}(\varphi - \varphi_0) \qquad \text{(Equation 3.7)}$$

$$M = At^2e^{-Bt}(\varphi - \varphi_0) \qquad \text{(Equation 3.8)}$$

where:

M= Moment difference between Experimental and calculated

A = Constant

t = Time in seconds (reset to zero at the beginning of Phase Three)

e = exponential

B = Constant

φ = Angular Position of the Foot

φ_0= Angular Position of the Foot at the beginning of Phase Two

The theoretical spring system model consists of the spring model plus the auxiliary model. This coupled system permits the theoretical prosthesis to have ankle moment characteristics similar to an intact ankle during gait.

3.3.5 Integration of Results from both Walking and Running

The previous section solved for the optimal theoretical spring system. This section investigates how the theoretical spring system for walking would work when experimental running data was the input, and how the theoretical spring system for running would work when experimental walking data was the input. The purpose for inputting the experimental data from either walking or running into the others' theoretical spring system, is to determine if the theoretical system could be interchangeable between the two different cadences.

The simplest solution for an amputee would to have one prosthetic system which could be used for both walking and running with no alterations. This would allow ankle moments, similar to that reported in an intact limb, for both running and walking with the same prosthetic leg. This infers that the regression equations for walking would be used for walking and running, and/or the regression equations for running could be used for both walking and running. The second best solution would be to make a minor alteration in the system (i.e., flip a switch) when changing between walking and running. This would imply that the spring would be constant for both walking and running, but the regression equation for the auxiliary source would be modified. The last choice would be to have a totally separate system for walking, as for running. This would imply the prosthetic leg would need to be physically changed for walking and running to achieve ankle moments similar to that of an intact ankle joint.

3.3.5.1 Completely Interchangeable Theoretical System

To determine if the running system could be used for walking the following methods were used. The ideal spring and ideal auxiliary source constants that where previously determined for each subject were used. The variables, from the walking data, were entered into the optimal spring regression equation for running and the optimal auxiliary source regression equation for running. This produced a calculated ankle moment for walking using the running theoretical system. The calculated theoretical moment was compared against the experimental moment and also the theoretical moment using the ideal walking system.

The next step was to investigate how the walking system would work in running situations. To determine how the walking ideal theoretical system would work during running, methods similar to the above were followed. The constants from the ideal spring and ideal auxiliary source for walking were used. The variables, from the running data, were entered into the spring regression equation and the walking auxiliary source regression equation. This produced a running theoretical ankle moment while implementing the walking theoretical system. This calculated moment was compared with the experimental moment, and also the calculated theoretical moment from the ideal running system.

3.3.5.2 A Partially Interchangeable Theoretical System

The second best solution was to implement the optimal spring and modify the auxiliary source. The calculated difference was determined between the experimental moment and the calculated moment from the spring. This calculated moment was regressed with Equation 3.8, to determine new constants for the auxiliary source. Walking or running data was then input into the theoretical auxiliary equation, to ascertain the effectiveness of the system. The same procedures were followed for running and walking.

3.3.5.3 Evaluation of Regressions

Evaluation of the regression models result's was performed two ways; 1) qualitatively and 2) quantitatively. Qualitative analysis was performed by visual inspection of moment versus time graphs, and moment versus angular position of the foot curves comparing theoretical to experimental. Quantitative analysis was performed by examining the standard error of estimate. The auxiliary moments were added to the calculated spring moment to get total calculated moments of the ankle, for each subject. The difference was taken from the experimental net moment and the total calculated moment of the ankle. This difference was used to determine the standard error of estimate (equal to the square root of the sum of all the differences squared divided by the number of data points minus 2) [Equation 3.9] to determine which equation would be best.

$$S_{y \cdot x} = \left(\sum (Y-\mathit{Y})^2/(n-2)\right)^{1/2} \qquad \text{(Equation 3.9)}$$

where:

$S_{y \cdot x}$ = standard error of estimate

Y = observed values

Y = computed values

n = number of data points

The sum of deviations was divided by (n-2) because, this resulted in $S^2_{y \cdot x}$ an unbiased estimator of the variance about the true regression line (Hamburg, 1985). The same process was used, for walking and running, for all subjects.

3.4 Summary

The kinematic and kinetic data were collected. From the information it was determined that energy generated was greater than that absorbed. This information determined that a spring could not produce the necessary ankle moments and an auxiliary energy source would need to aid the spring during plantarflexion (Phase 3 of the gait cycle). Non-linear regression equations were then determined for the spring and auxiliary source for walking and running. These regression equations provided a theoretical system that would permit the energy and moment characteristics similar to an intact ankle during gait. It was then examined if the theoretical systems could be interchanged between walking and running.

4. Results

4.1 Overview

It was determined during the process of establishing the methods that an auxiliary source was needed to aid the spring to achieve energy levels similar to that reported in children with an able-body. This theoretical spring system would permit a prosthetic ankle joint to have moment characteristics similar to an intact ankle joint. These data also suggest that the current designs, which consist only of a spring, cannot provide adequate moment generation for push-off and need to be modified to allow BKAs to walk/run similar to ABs. This is supported by previously discussed literature (Czerniecki et al., 1991; Engsberg et al., 1993b). The first section of the results will report the quantity of additional energy needed for the theoretical spring system to achieve parity with an intact limb. The second section will present the spring constants and the auxiliary constants, for walking and then running. A qualitative and quantitative evaluation of the effectiveness of the regression equations at reproducing the ankle moments theoretically was then presented. Then the results of subject 1, will be presented for the total interchangeable system, and partial interchangeable theoretical spring systems and a corresponding evaluation. These results will provide the necessary energy and moment requirements for a theoretical prosthetic ankle joint to react similarly to an intact ankle during walking and running gaits, therefore achieving the purpose of this investigation.

sufficient ankle moments during Phase 3 of gait, to equal ankle moments produced by children with able-bodies. The quantity of additional energy needed was determined by Equation 3.4, $E = {}_{0}\int^{T}(M*\omega)\,dt$ and is presented in Table 4.1. All results are the additional energy, above that absorbed, needed to be generated during phase 3 of the gait cycle.

Table 4.1: Average Additional Energy Requirements For Plantarflexion For Walking and Running

Subject Number	Age (Yrs)	Mass (Kg)	Ave. Walk Energy (Joules)	Ave. Run Energy (Joules)
1	8	24.0	3.41	4.73
2	10	31.5	8.00	17.93
3	14	56.4	16.96	N/A

Table 4.1 shows that additional energy of 3.41 to 18.96 Joules were needed to assist the theoretical spring, during walking, to provide adequate plantarflexion to duplicate the ankle moments of an intact foot/ankle joint. There appeared to be a trend of an increase in energy with an increase in age and/or mass. Also the amount of additional energy increased when subjects 1 and 2 ran (Table 4.1), with the values ranging from 4.73 to 17.93 Joules. Note however, that due to the small number of subjects this trend could not be confirmed statistically. The additional

Table 4.2 presents the rate at which the work was done (Power) for the three subjects during stance. This was calculated by dividing the quantity of energy during stance phase by the length of time of the stance phase. This quantity of work is overestimated since it does not consider the amount of energy during the swing phase. No calculations were made during swing phase that would allow the determination of the average work done for the entire gait cycle. In normal walking it is accepted that the average stance phase is 60% of the gait cycle and swing phase is the remaining 40% of the gait cycle, and for running the stance phase is 45% of the gait cycle and the swing phase 55% (Ounpuu, 1990 & Perry, 1992). Noting that these percentages change with alterations in gait velocity. Therefore, acknowledging swing phase, the quantity of Watts is probably overestimated by a factor of 2. A battery used for portable VCRs can provide approximately 24 Watts for one hour. The highest energy requirement reported in this study was for subject 2 while running and was approximately 62 Watts. Assuming that this value is overestimated by a factor of 2 the battery would last for about 45 minutes for subject 2 while running. Table 4.2 indicates with an increase in age and/or mass there is an increase in the amount of power necessary to accomplish the same task, walking or running. The results also show that with an increase in speed (i.e., walking versus running) there is an increase in the amount of power needed to accomplish the task. Again, due to the small numbers of subjects this could not be tested statistically

Table 4.2: Average Power Requirements in the Ankle For Walking and Running

Subject Number	Age (Yrs)	Mass (Kg)	Ave. Walk Power (Watts)	Ave. Run Power (Watts)
1	8	24.0	6.14	18.49
2	10	31.5	12.80	61.58
3	14	56.4	29.87	N/A

Table 4.3 shows the approximate additional energy necessary to walk/run 1 Kilometer. These calculations were done by determining the amount of energy required if the subjects walked at their reported average walking/running speed for 1 Kilometer (Table 3.2). The amount of energy to walk 1 kilometer increases with age, mass, and speed. However, again due to the small number of subjects this trend could not be verified statistically. These quantities of energy can help determine the requirements of an auxiliary source. However, the determination of a physical auxiliary source was outside the realm of this investigation.

Table 4.3: The Approximate Energy Requirements For Walking and Running 1 Kilometer

Subject Number	Age (Yrs)	Mass (Kg)	Ave. Walk Energy (Joules)	Ave. Run Energy (Joules)
1	8	24.0	3655	5707
2	10	31.5	7574	20,468
3	14	56.4	16,059	N/A

4.3 Ankle Moment Reproduction

The kinematic and kinetic data collected from the children were used to determine

4.3 Ankle Moment Reproduction

The kinematic and kinetic data collected from the children were used to determine a theoretical spring system which would reproduce the ankle moment. The relationship between the experimental ankle moment and experimental foot/ankle position determined the regression equation for the spring, $M = e^{(A+B\,\varphi)}$ (Equation 3.5). The regression equations for the auxiliary source were determined by regression equations which utilized the relationship between the experimental ankle moment and the experimental ankle/foot position with respect to time $M = At^n e^{-Bt} (\varphi - \varphi_0)$ (Equations 3.7 & 3.8 where n represents 1 & 2 respectively). These equations define the relationship between the ankle moment (dependent variable) and angular position of the foot (independent variable) for the spring, and angular position of the foot and time (independent variables) for the auxiliary source. The next results presented will include the constants for the regression equations and the effectiveness of these equations at reproducing ankle moments similar to an intact ankle joint system for walking and then running. This will provide the necessary information to determine the ankle moment for a given angular position of the foot for the spring.

4.3.1 Results for Walking

Figure 4.1, 4.3, and 4.5 shows the average experimental moment and the two total calculated moments (the optimal spring moment [Equation 3.5] plus the auxiliary moment [Equation 3.7 (referred to as T1) & 3.8 (referred to as T2)]) versus the angular position of the foot for Subject 1, 2, and 3 respectively. Figure 4.2, 4.4, and 4.6 shows the average experimental moment and the two total calculated moments (the optimal spring moment

[Equation 3.5] plus the auxiliary moment [Equation 3.7 & 3.8]) versus time for Subject 1, 2, and 3 respectively. The regression equation constants for all subjects are shown in Table 4.4. These are the respective formulas for each subject, in which the experimental data was input. The standard error of estimates results, in Nm, are presented in Table 4.5.

Figures 4.2, 4.4 & 4.6 graphically shows how for approximately the first 20% of the stance phase the total theoretical moments overestimated the experimental ankle joint moments. It can be observed, in Figures 4.1 - 4.5 where T1 was used, Equation 3.7 ($Ate^{-Bt} (\varphi - \varphi_0)$), an overestimation of the experimental peak moment occurred. Where as when T2 was used in Equation 3.8 ($At^2e^{-Bt} (\varphi - \varphi_0)$), the estimate of the experimental peak moment reproduced the actual peak moment with greater accuracy. Following the peak moment the auxiliary T1 source underestimated the experimental moment followed by an overestimation of the experimental moment. This pattern was also seen in auxiliary T2, however to smaller degree. In Figures 4.1 through 4.6, one can see that the end calculated total moment goes below zero, whereas the experimental moment does not. This would need to be corrected, to permit the foot to prepare for the swing phase. The graphical results show that T2 reproduced the experimental moment better than T1. This was confirmed quantitatively in Table 4.5. The smaller the standard error of the estimate, the closer the theoretical model reproduced the experimental results. Table 4.5 shows quantitatively that the optimal spring was aided slightly better by the auxiliary source T2 (Equation 3.8), for all three subjects.

Figure 4.7 is a typical representation of the ankle moment production for the spring, the auxiliary T2 (Equation 3.8), the total calculated moment, and the experimental ankle moment, versus time. This figure is representative of all participating subjects. Figure 4.7 graphically shows the timing of the theoretical spring system. As the spring starts to decrease in production of ankle moment near the peak at a faster rate than the experimental moment, the auxiliary source begins to produce an ankle moment, to minimize the difference between the experimental moment and theoretical moment. Figure 4.7 shows how the spring coupled with the auxiliary source (spring system) effectively provides ankle moments during stance similar to that of a child with an AB.

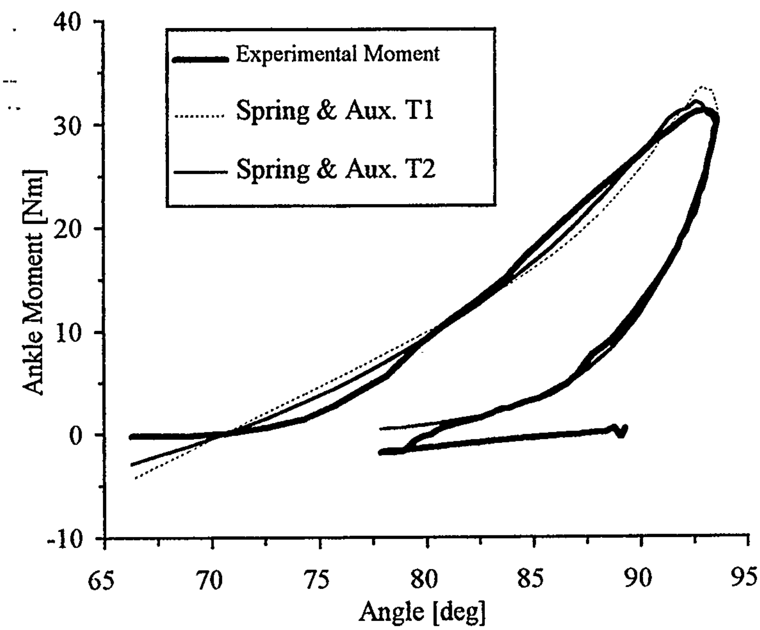

Figure 4.1: Subject 1 Experimental Ankle Moment, Total Calculated Moment (Spring and Auxiliary Moment T1 [Equation 3.7]), and Total Calculated Moment (Spring and Auxiliary Moment T2 [Equation 3.8]) Versus Angular Position of the Foot For Walking.

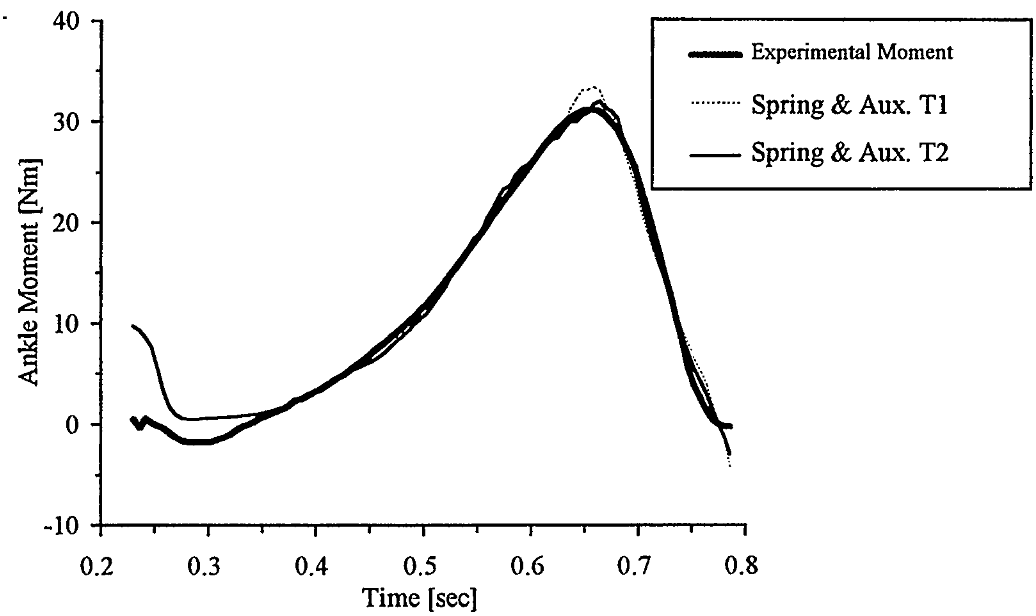

Figure 4.2: Subject 1 Experimental Ankle Moment, Total Calculated Moment (Spring and Auxiliary Moment T1 [Equation 3.7]), and Total Calculated Moment (Spring and Auxiliary Moment T2 [Equation 3.8]) Versus Time For Walking.

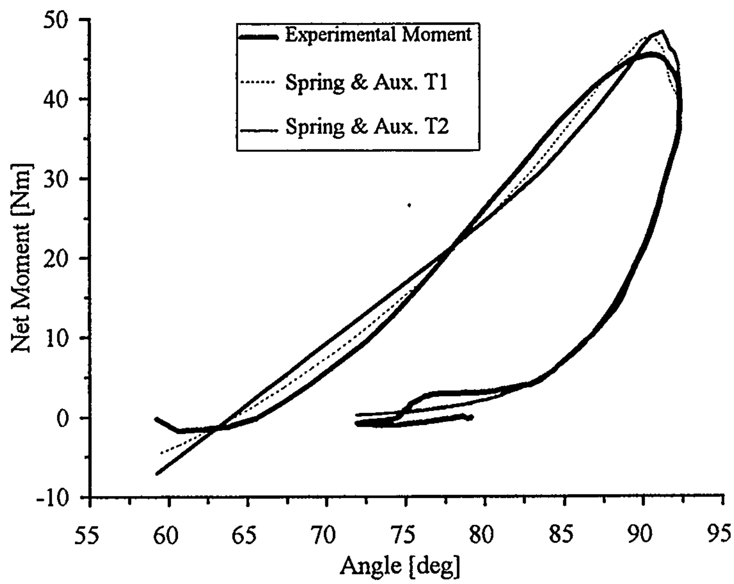

Figure 4.3: Subject 2 Experimental Ankle Moment, Total Calculated Moment (Spring and Auxiliary Moment T1 [Equation 3.7]), and Total Calculated Moment (Spring and Auxiliary Moment T2 [Equation 3.8]) Versus Angular Position of the Foot For Walking.

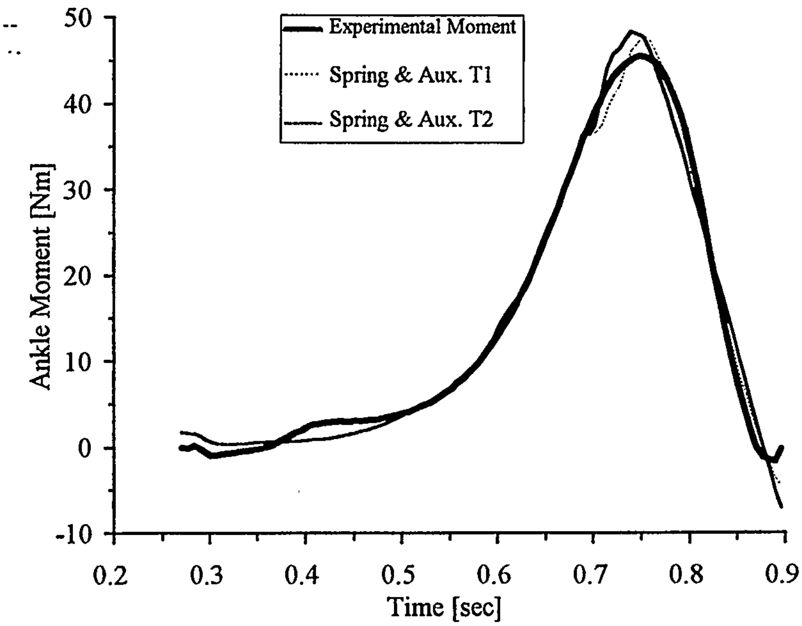

Figure 4.4: Subject 2 Experimental Ankle Moment, Total Calculated Moment (Spring and Auxiliary Moment T1 [Equation 3.7]), and Total Calculated Moment (Spring and Auxiliary Moment T2 [Equation 3.8]) Versus Time For Walking.

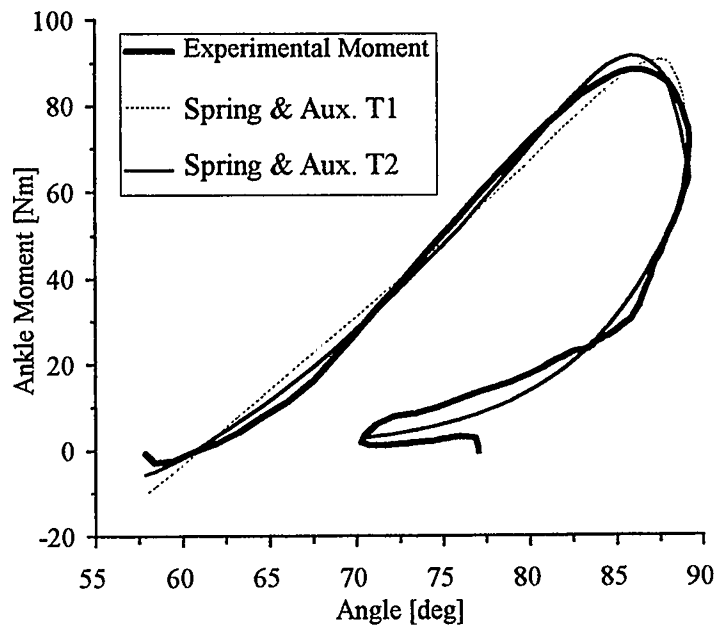

Figure 4.5: Subject 3 Experimental Ankle Moment, Total Calculated Moment (Spring and Auxiliary Moment T1 [Equation 3.7]), and Total Calculated Moment (Spring and Auxiliary Moment T2 [Equation 3.8]) Versus Angular Position of the Foot For Walking.

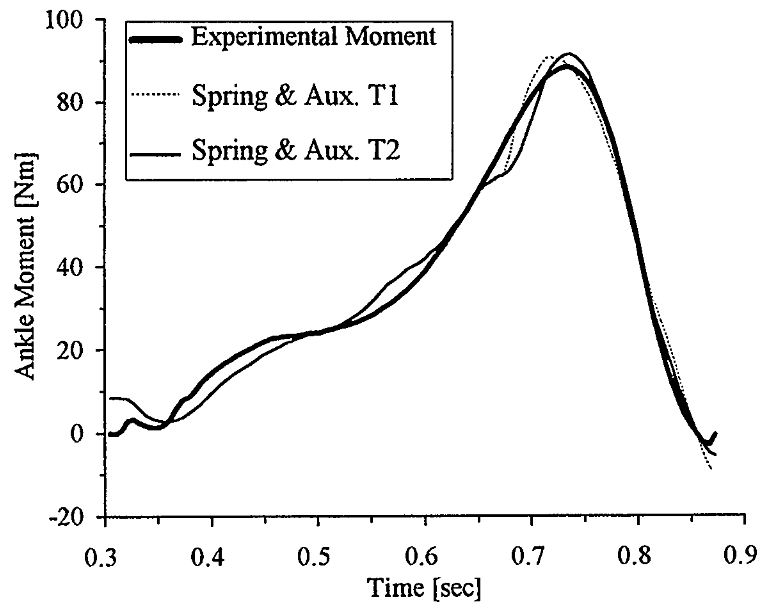

Figure 4.6: Subject 3 Experimental Ankle Moment, Total Calculated Moment (Spring and Auxiliary Moment T1 [Equation 3.7]), and Total Calculated Moment (Spring and Auxiliary Moment T2 [Equation 3.8]) Versus Time For Walking.

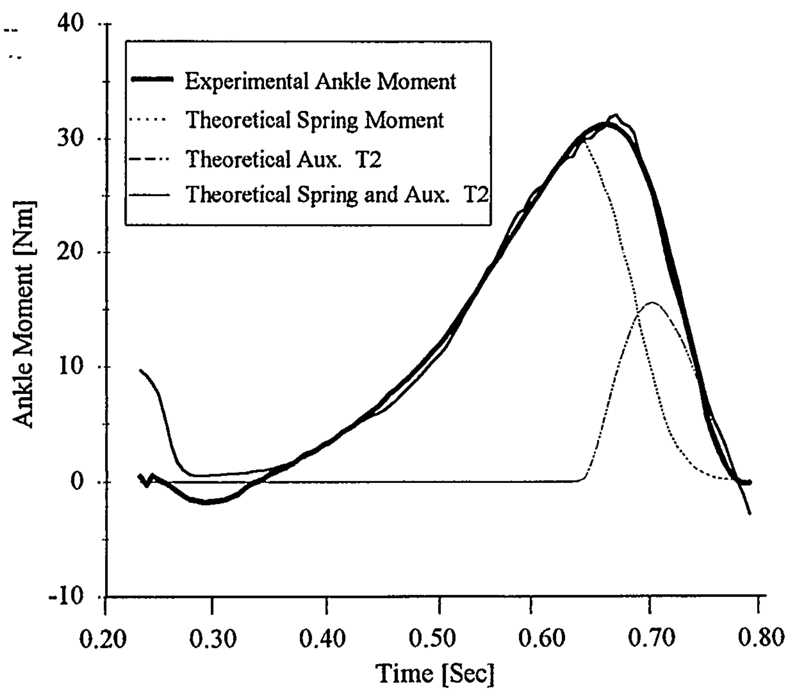

Figure 4.7: Experimental Moment, the Moment Provided by the Spring, the Moment From the Auxiliary Source (Aux.T2), and the Total Calculated Moment (Spring & Aux. T2) Versus Time For Walking.

Table 4.4: Regression Equations Including the Calculated Constants For Walking. The moment (M) is measured in Nm and angular position of the foot with respect to the shank (φ) is measure in degrees.

Subject Number	Spring Equation	Auxiliary Equation (T1)	Auxiliary Equation (T2)
1	$M=e^{(-20.52+.26\varphi)}$	$M=18.17te^{-6.67t}(\varphi-\varphi_0)$	$M=849.82t^2e^{-22.23t}(\varphi-\varphi_0)$
2	$M=e^{(-17.45+.23\varphi)}$	$M=23.43te^{-5.74t}(\varphi-\varphi_0)$	$M=883.13t^2e^{-18.33t}(\varphi-\varphi_0)$
3	$M=e^{(-10.44+.16\varphi)}$	$M=49.12te^{-5.16t}(\varphi-\varphi_0)$	$M=1871.16t^2e^{-18.56t}(\varphi-\varphi_0)$

Table 4.5: Standard Error of Estimate Results For Walking

Subject Number	Auxiliary Equation (T1) (Nm)	Auxiliary Equation (T2) (Nm)
1	0.23	0.22
2	0.53	0.47
3	0.38	0.37

4.3.2 Results for Running

Subject 1 & 2 participated in the running study. The experimental moment is graphically compared to the calculated total moments ([Equation 3.5+ Equation 3.7] and [Equation 3.8 + Equation 3.7]) in Figures 4.8 through 4.11. Figure 4.8 & 4.10 present the various moments during running versus the angular position of the foot for subjects 1 & 2 respectively. Figures 4.9 & 4.11 include the experimental moment, and the calculated moments (Spring & Aux. T1; and Spring & T2) versus time during running, for subject 1 & 2 respectively. The regression equation constants for both subjects are shown in Table 4.6. These are the respective formulas for each subject's experimental data to be input to determine the theoretical moment. The results from the standard error of estimates, in Nm, are presented in Table 4.7. The smaller the standard error of the estimate the more similar the theoretical model matched the experimental results. Table 4.7 shows that T2 provided a better solution and subject 2's model was a closer representation of the experimental moments than subject 1.

As with walking the spring overestimates the experimental moment for approximately the first 15% of stance (Figure 4.8 & 4.10). This would need to be corrected for in an actual design to permit proper slowing of the foot as it lowers to the floor following heel strike. Then the spring underestimates the experimental moment until just prior to the experimental peak moment. For running the peak moment is overestimated by the spring. This is because just prior to the experimental peak moment the slope (moment versus angle) decreases and the regression equation does not correct for this. How this would affect gait is unknown at this time. Equation 3.7, using T1, decreased more rapidly than the experimental ankle moment. Equation 3.8, T2, reproduced the additional ankle moment necessary more effectively. Similar to that reported for walking, the optimal spring was best aided by the auxiliary source T2 (Equation 3.8), which is also shown in Table 4.7. In Figures 4.7 through 4.11 the end calculated total moment goes below zero, whereas the experimental moment does not. Similar to walking, this would need to be corrected for in an actual design, to allow the foot to clear the floor during swing. The influence of these reported differences, between the experimental and theoretical ankle moments, on the gait of an individual with a BKA must be determined in future experimental investigations.

Figure 4.12 is a typical representation of running results for subjects 1 and 2 in this investigation. Figure 4.12 shows the timing of the spring and auxiliary source (Aux. T2). Figure 4.12 shows how as the theoretical spring moment begins to decrease that the auxiliary source begins to aid the spring to provide a total calculated moment similar to the experimental ankle moment during running. Figure 4.2 indicates that the spring coupled with an auxiliary source provides a theoretical system similar to an intact system.

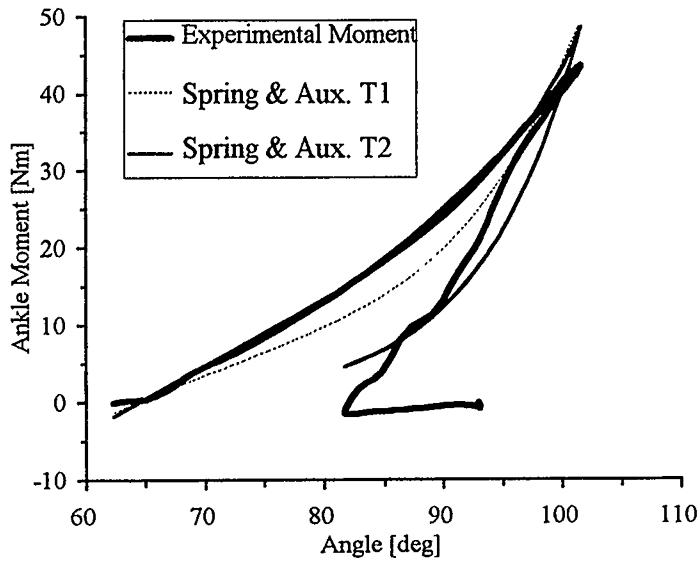

Figure 4.8: Subject 1 Experimental Ankle Moment, Total Calculated Moment (Spring and Auxiliary Moment T1 [Equation 3.7]), and Total Calculated Moment (Spring and Auxiliary Moment T2 [Equation 3.8]) Versus Angular Position of the Foot For Running

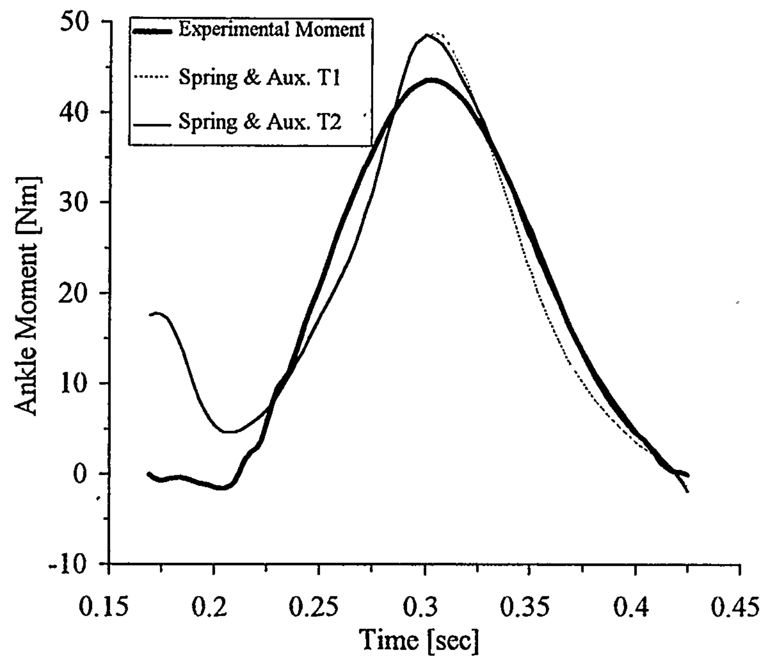

Figure 4.9: Subject 1 Experimental Ankle Moment, Total Calculated Moment (Spring and Auxiliary Moment T1 [Equation 3.7]), and Total Calculated Moment (Spring and Auxiliary Moment T2 [Equation 3.8]) Versus Time For Running.

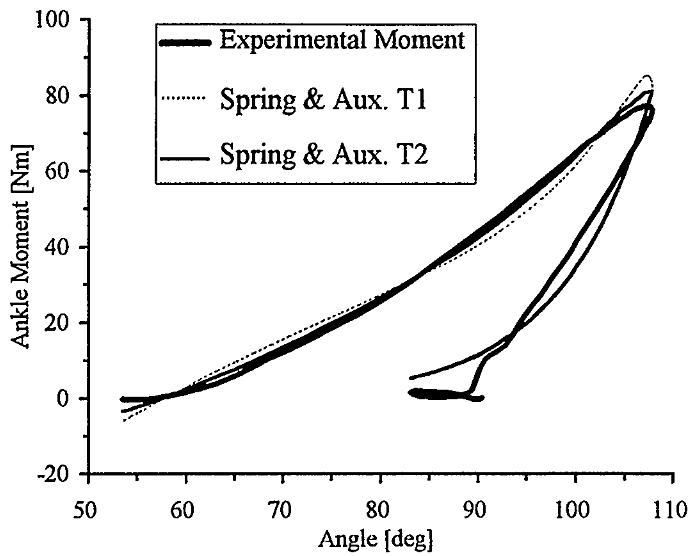

Figure 4.10: Subject 2 Experimental Ankle Moment, Total Calculated Moment (Spring and Auxiliary Moment T1 [Equation 3.7]), and Total Calculated Moment (Spring and Auxiliary Moment T2 [Equation 3.8]) Versus Angular Position of the Foot For Running

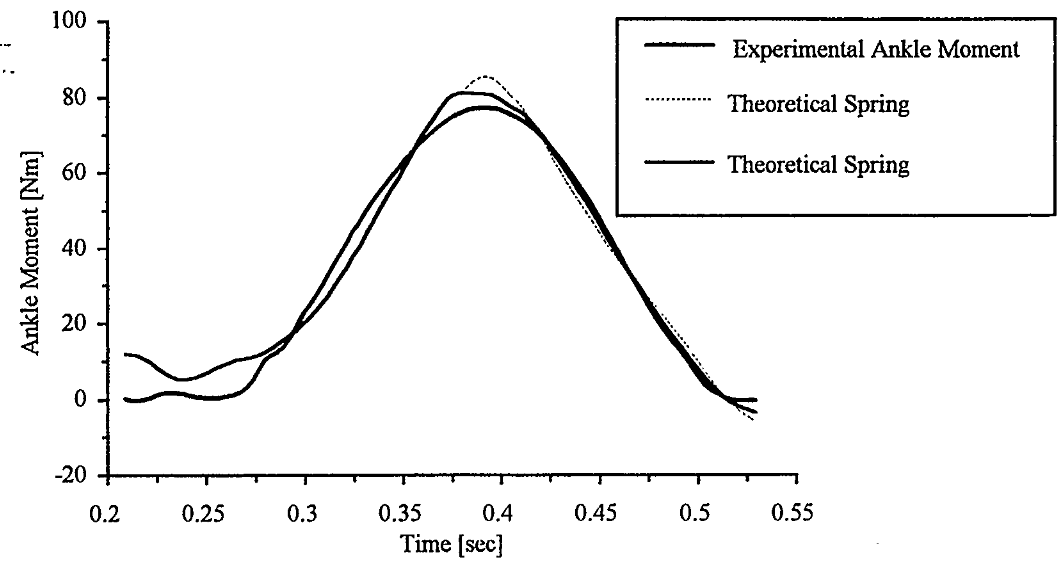

Figure 4.11: Subject 2 Experimental Ankle Moment, Total Calculated Moment (Spring and Auxiliary Moment T1 [Equation 3.7]), and Total Calculated Moment (Spring and Auxiliary Moment T2 [Equation 3.8]) Versus Time For Running.

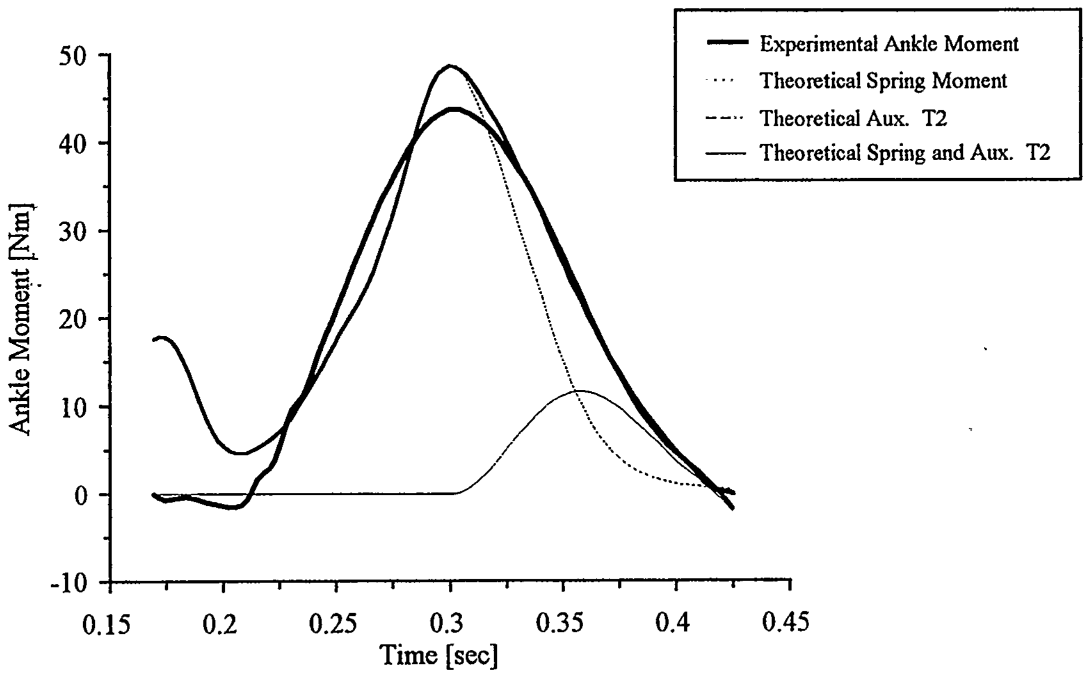

Figure 4.12: The Experimental Moment, the Moment Provided by the Spring, the Moment From the Auxiliary Source (Aux.T2), and the Total Calculated Moment (Spring & Aux. T2) Versus Time For Running.

Table 4.6: Regression Equations Including the Calculated Constants For Running. The moment (M) is measured in Nm and angular position of the foot with respect to the shank (φ) is measure in degrees.

Subject Number	Spring Equation	Auxiliary Equation (T1)	Auxiliary Equation (T2)
1	$M=e^{(-8.14+.12\varphi)}$	$M=6.83te^{-3.05t}(\varphi-\varphi_0)$	$M=408.70t^2e^{-16.91t}(\varphi-\varphi_0)$
2	$M=e^{(-7.45+.11\varphi)}$	$M=15.82te^{-3.50t}(\varphi-\varphi_0)$	$M=816.22t^2e^{-20.72t}(\varphi-\varphi_0)$

Table 4.7: Standard Error of Estimate Results For Running

Subject Number	Auxiliary Equation (T1) (Nm)	Auxiliary Equation (T2) (Nm)
1	0.62	0.59
2	0.18	0.13

4.3.3 Results from Interchanging Theoretical Running System with Walk Data

The results have now provided the optimal solution for each subject for walking and then a separate solution for running. This suggests that an individual would need to have a prosthetic leg for walking and a separate one for running. This would seem to be an inconvenience and also increase the expense to a BKA. To determine theoretically if the walking spring and it's corresponding auxiliary source could be used for running, the experimental running data (angular position of the foot with respect to the shank and time) was input into the subject's respective walking model. To allow a partial interchange the spring constants would be identical for walking and running, but the auxiliary source was modified. This would represent a system in which the hardware (spring) would stay constant but the control of the auxiliary source would be changed. These results will be presented for one subject and are representative of both subjects who participated in both the running and walking experiment.

Figure 4.13 & 4.14 represents the results of using the run regression equations with walking data (angular position of the foot with respect to the shank and time), for one subject. Figure 4.13 presents the average experimental moment, and the calculated moment using the ideal solution for walking versus angular position of the foot. Figure 4.14 is a graph of the average experimental moment, and the calculated moment using the ideal solution for walking versus time. The result from the standard error of estimate, 0.63 Nm, was over twice that calculated using the optimal running system with running data for subject 1. The results show that the optimal running system was not as efficient at reproducing the ankle moments with walking data, as it was with running data. The calculated moment overestimated the experimental moment for approximately the first 45%, and for approximately the last 10% of stance (Figure 4.13 & 4.14). Indicating that the system is overproducing energy. From approximately 45% of stance to 90% of stance the calculated moment underestimated the experimental moment (Figure 4.13 & 4.14). Indicating that the spring's stiffness is too low in the theoretical system to provide an adequate ankle moment. The peak moment was underestimated with the running theoretical system in a simulated walking situation. As mentioned previously, it is unknown in actuality how the theoretical model would affect the gait patterns of an individual with a BKA.

Calculating an optimal auxiliary source to aid the run spring resulted in an improvement in the theoretical ankle moments' ability to mimic the experimental data (Figure 4.15 &4.16). Optimizing the auxiliary source to the specific data aided the spring more efficiently during plantarflexion. The standard error of estimate, 0.58 Nm, reflected this improvement. This partially interchanged theoretical system still did not reproduce the ankle moment as well as the theoretical system designed specifically for running. Again it is unknown how this would affect gait in actual conditions.

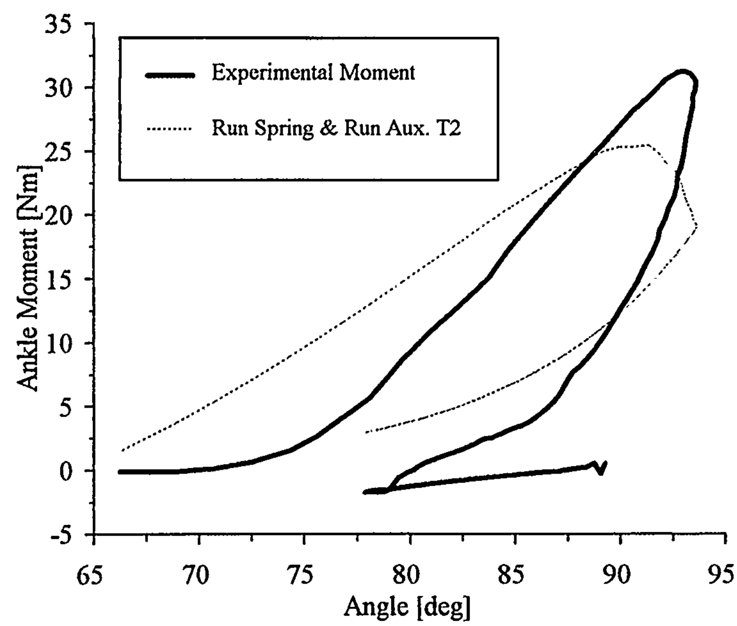

Figure 4.13: Subject 1 Experimental Moment During Walking and the Calculated Ankle Moment (Optimal Run Spring and Run Auxiliary Source Used With Walking Data) Versus Angular Position of the Foot.

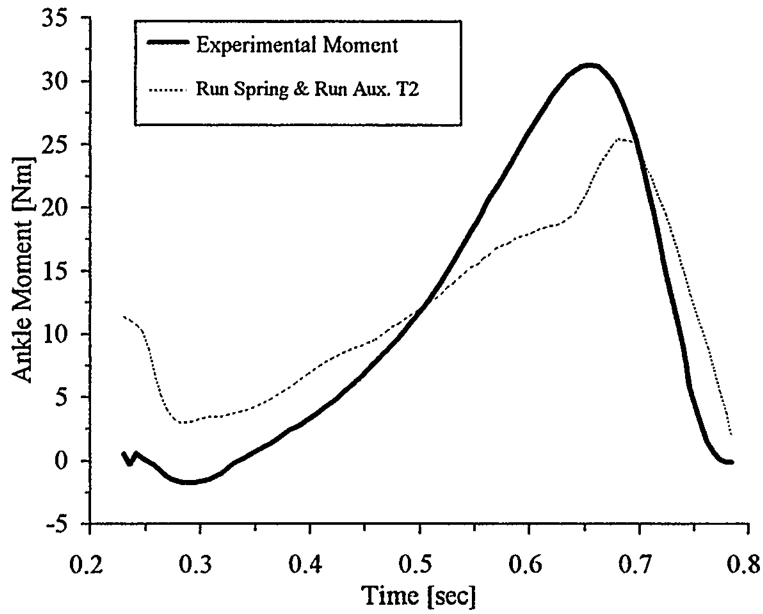

Figure 4.14: Subject 1 Experimental Ankle Moment During Walking and Calculated Ankle Moment (Optimal Spring and Auxiliary (Aux. T2) For Running Used With Walking Data) Versus Time.

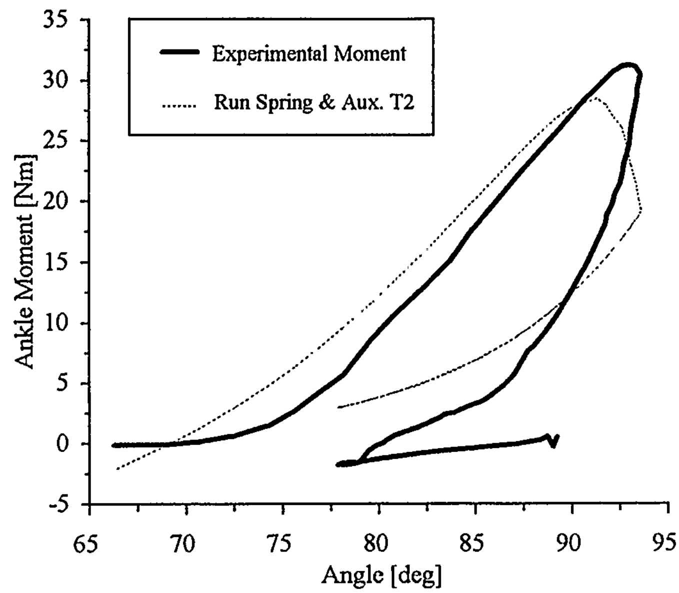

Figure 4.15: Subject 1 Experimental Ankle Moment During Walking and Calculated Ankle Moment Using Walking Data (Optimal Spring For Running and the Optimal Calculated Auxiliary Source Used With Walking Data) Versus Angular Position of the Foot.

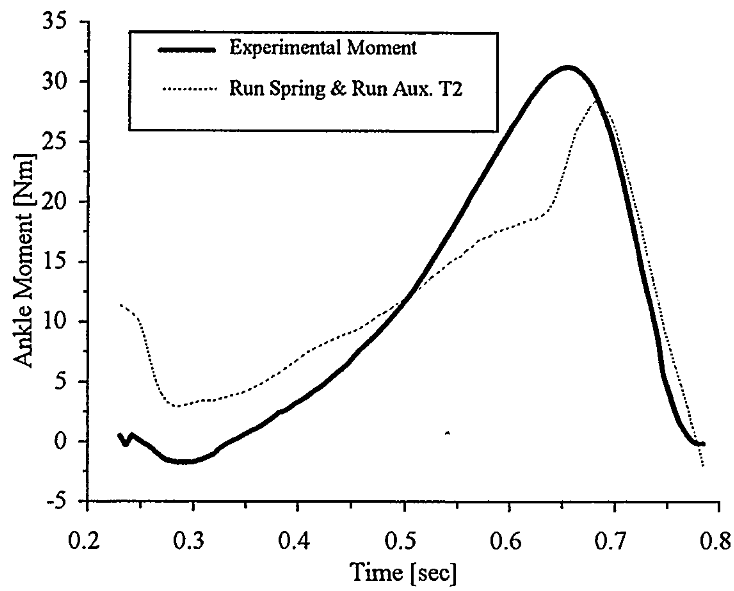

Figure 4.16: Subject 1 Experimental Ankle Moment and Optimal Run Spring and Optimal Auxiliary Source With Walk Data Versus Time.

4.3.4 Results from Interchanging Theoretical Walking System with Run Data

After examining if the running theoretical system could be used in part or in whole for walking, the same methods were conducted to determine if the walking theoretical system could be used in part or in whole for running.

The experimental moment and the calculated moment using the optimal solution for walking with running data are present in Figure 4.17 and 4.18. Figure 4.17 shows moments with respect to angular position of the foot. Figure 4.18 presents the experimental and calculated moment with respect to time. This solution did not reproduce the ankle moment as effectively as the system specifically designed for running. The initial ankle moments were overestimated. This is a result of similar angular positions in Phase 1 and Phase 2 of the gait cycle, but the regression equations based on Phase 2, where the ankle moments had higher values. Figure 4.17 and 4.18 shows graphically that the theoretical moment overestimated the ankle moment during the middle portion of the gait. This included the theoretically peak moment being approximately 4.5 times greater than the experimental peak moment. These overestimates are also shown quantitatively with a standard error of estimate of 6.50 Nm. The first third and final third theoretical moments closely resembled the experimental ankle moment.

Figure 4.19 presents data from a partially interchangeable walking system used with experimental running data. The partial interchanged system consisted of the walking spring, and an auxiliary source. The auxiliary source for the partial interchanged system was determined by the difference between the theoretical results (the running data input into the walking spring equation). Figure 4.19 is a graphic representation of the partially interchanged system and the experimental moment with respect to angular position. The results showed a slight improvement from the totally interchangeable system, at the end of stance during plantarflexion. This slight improvement was confirmed numerically with a standard error of estimate of 6.46 Nm. However, the solution still did not reproduce the ankle moment as effectively, as the running theoretical solution used with the running data. It is unlikely the spring would be able to be loaded in an experimental situation to produce ankle moments 4 times greater than those reported experimentally.

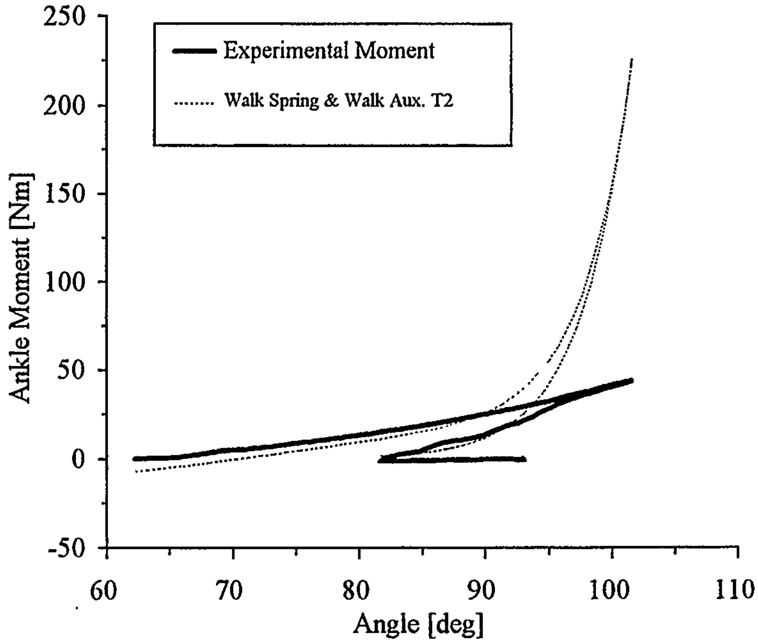

Figure 4.17: Subject 1 Experimental Ankle Moment During Running and the Total Calculated Moment While Running (the Optimal Walk Spring Plus the Optimal Walk Auxiliary Source (Aux. T2) With Running Data) Versus Angular Position of the Foot.

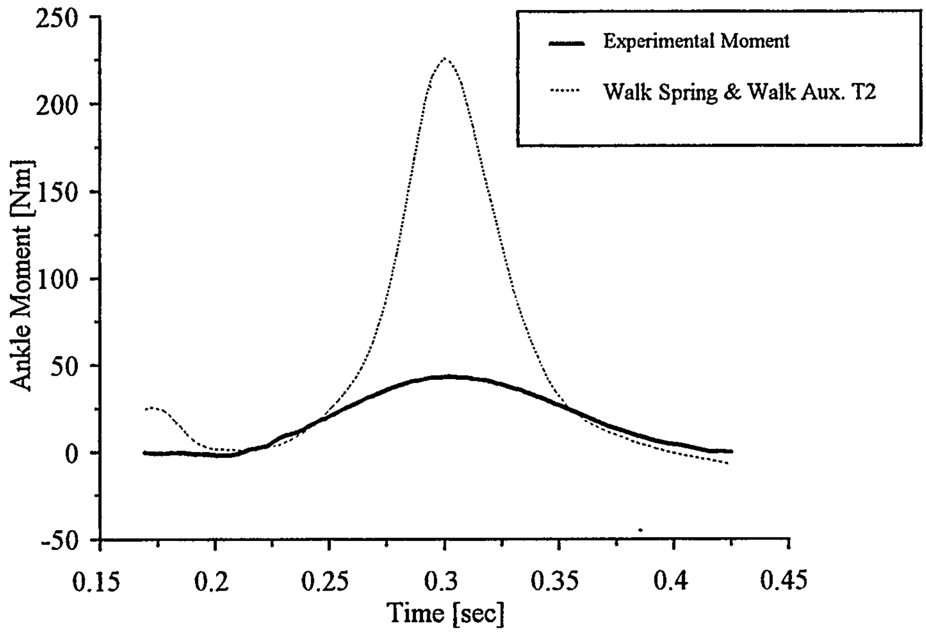

Figure 4.18: Subject 1 Experimental Ankle Moment During Running and the Total Calculated Moment While Walking (the Optimal Walk Spring Plus the Optimal Walk Auxiliary Source (Aux. T2) With Walking Data) Versus Time.

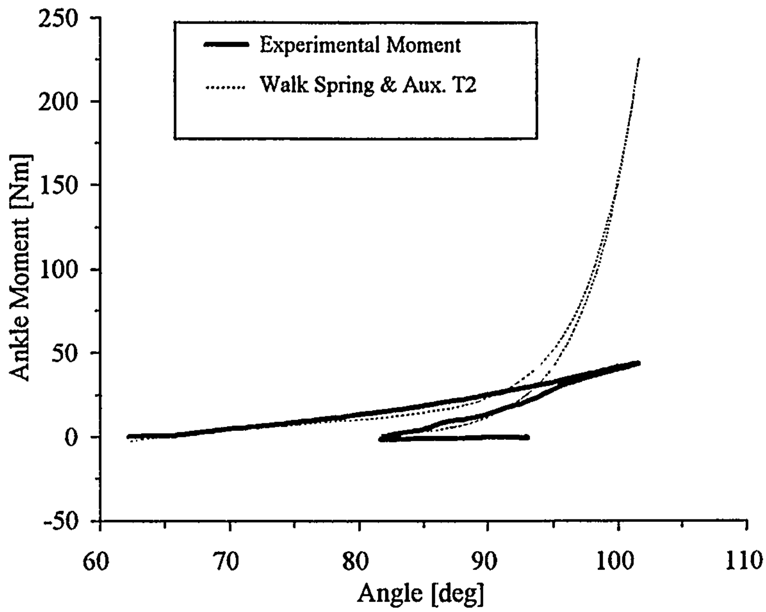

Figure 4.19: Subject 1 Experimental Ankle Moment During Running and the Total Calculated Moment While Running (the Optimal Walk Spring Plus the Optimal Auxiliary Source (Aux. T2) With Running Data) Versus Angular Position of the Foot.

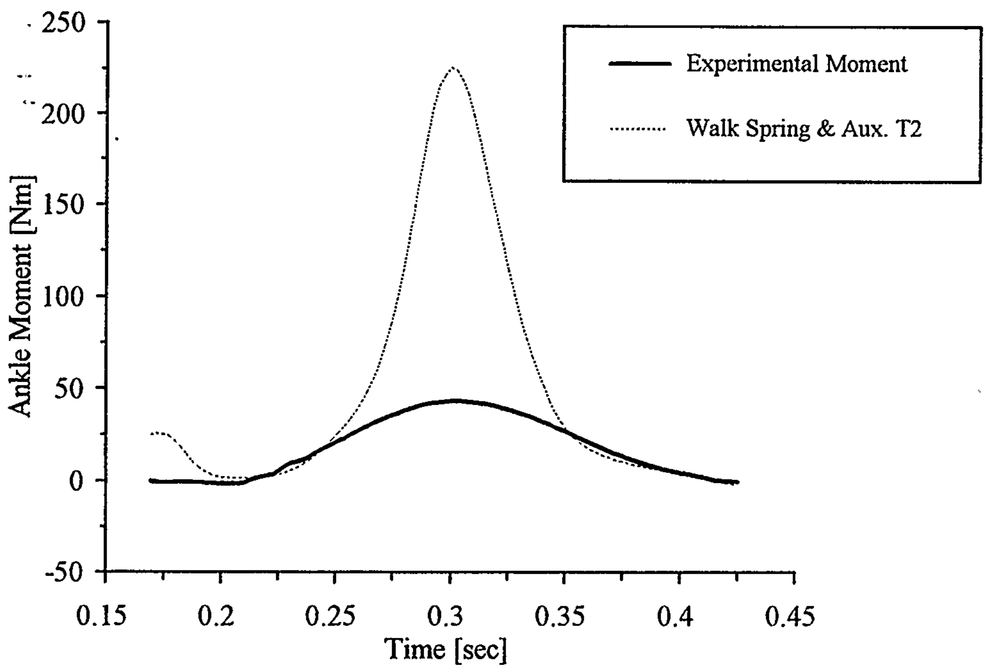

Figure 4.20: Subject 1 Experimental Ankle Moment During Running and the Total Calculated Moment While Running(the Optimal Walk Spring Plus the Optimal Walk Auxiliary Source (Aux. T2) With Running Data) Versus Time.

4.4 Summary

The results presented have given us the necessary data to investigate the requirements for a prosthetic ankle joint which could behave similarly to an intact ankle joint during gait. This data will also help determine the optimal solution, for walking and running and if the systems could interchange the springs and/or auxiliary sources.

5. Discussion

5.1 Overview

The purpose of this study was to investigate the requirements for a terminal device to allow a child with a below-knee amputation to walk similar to a child with an able-body. To achieve this purpose a methodology was designed to attain the necessary data to quantify the requirements for a prosthetic ankle joint to act like an intact ankle joint. The experimental methodology, limitations, and results have previously been discussed. This chapter will begin with a brief discussion of the limitations and will continue by comparing these results to other similar studies. The discussion will then continue with an interpretation of the significance of these results. The final statements in the discussion will include what the next step is for improving the prosthetic ankle joint.

5.2 Limitations

The first limitation of this study was the number of subjects. The second limitation was that we only investigated the propelling and restraining forces in the sagittal plane. A third limitation was that a rigid link model was assumed. It is unknown how these limitations affect the results of this study but it appears that this study was successful at investigating the propelling and restraining forces generated by the ankle joint.

5.3 Result Evaluation

The methodology for this study was specifically designed to investigate the requirements of a prosthetic ankle joint. This methodology is unique in that sense. However, some of the variables investigated, have been investigated by other researchers.

As discussed in the literature review there are no known studies which investigated the ankle moment with regard to the angular position of the foot. But one could interpret that if the ankle motion was similar to previous studies and the corresponding moment appeared reasonable that the comparison of the ankle moment to foot angular position would be accurate. During walking Stauffer et al. (1977) reported a total sagittal ankle joint motion of 24.4 degrees in adults. Perry (1992) reported an average of 30 degrees of ankle motion during walking, with a range of 20 degrees to 40 degrees. In this current study the calculated total motion of the ankle ranged from a low of 27.3 degrees for subject one to a high range of motion of 33.1 degrees for subject two. These results seem reasonable for healthy children.

Another result which can be verified by previous literature was the determination of a spring's inability to provide necessary ankle joint moments for push-off. To determine if a spring alone could produce adequate ankle moments, the energy absorbed and generated were investigated. It was reported in Table 4.3 that the energy generated was greater than that absorbed. This finding was supported by findings by Czerniecki et al. (1991). Czerniecki et al.(1991) however reported maximum values and did not report the experimental number of additional Watts needed for sufficient plantarflexion during push-off. Therefore, we were unable to verify with other reported data the quantity the auxiliary source would need to produce.

The next variables this investigation studied were the experimental ankle joint moments. Winter and Sienko (1988) (Figure 5.1) and Stauffer et al. (1977) reported similar ankle moment time curves for a freely chosen walking speed as that calculated in this study (Figure 4.2, Figure 4.4, and Figure 4.6). The figures presented a slight phase of plantarflexion in the beginning of stance, followed by dorsiflexion, and the final ankle motion was plantarflexion. The peak ankle moments normalized to body mass were 1.3 Nm/kg, 1.5 Nm/kg, and 1.6 Nm/kg for subject one through three respectively. Winter and Sienko (1988) reported a peak ankle moment of approximately 1.5 Nm/kg.

Winter (1983b) reported ankle moments while adult subject jogged slowly. The ankle moment time curve (Figure 5.2) closely resembled those presented in this study (refer to Figures 4.9 and 4.11). By normalizing Winter (1983) results, the approximate peak ankle moment was 2.21 Nm/kg. The present results yielded a peak moment of 1.82 Nm/kg and 2.46 Nm/kg for subject one and two respectively. The present ankle moment results seem reasonable when compared to other previously reported data.

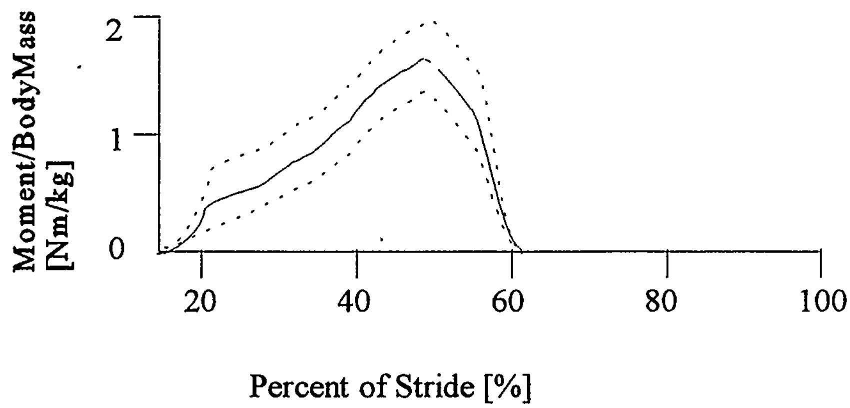

Figure 5.1: Moment of Force Inter-Subject Averages for Normal Subjects Walking at their Natural Cadences. Stride Period was set to 100% with Stance at 60%. The Moment of Force for each Subject was Normalized to Body Mass. Solid Line is the Average Curve, and the Dotted Line is plus or minus one Standard Deviation (Winter and Sienko, 1988).

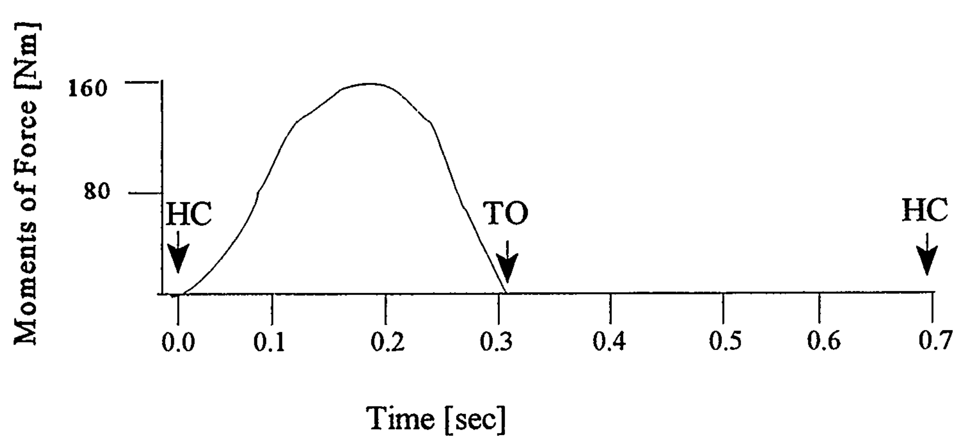

Figure 5.2: Moments of Force for One Subject Calculated During Slow Jog. HC Represents Heel Contact and TO Represents Toe-Off (Winter 1983b).

The previously published literature supported the results reported in this study. The importance of these results will be discussed next. For walking and running, it was determined that a spring would be aided better by Auxiliary T2 (Equation 3.8). These results were confirmed with the standard error of estimate results with the standard error of the estimate resulting in smaller numerical values (Table 4.2 and Table 4.4). Time squared (Equation 3.8) resulted in a more rapid moment producing theoretical system. All of the calculated ankle moment curves showed that the initial ankle moments were higher than that desired. The initial low moments are essential during heel strike to control the foot. The theoretical system could be adjusted so that the spring would be detached until the foot started increasing angular velocity, therefore preventing it from producing positive ankle moments at initial heel contact. For the remainder of the stance phase the theoretical control system closely reproduced the experimental moment values for walking and running. At the end the theoretical moment went below zero. This is undesirable and would produce difficulty for the foot to clear the floor. The calculated moments going below zero near the end of stance could be corrected for by preventing the auxiliary system from producing negative moments, therefore allowing the foot to clear the floor during swing.

Ideally the theoretical systems for walking and running would be interchangeable. However, when the walking system or running system was partially or totally interchanged, the theoretical system did not work as efficiently as the individual system. The main cause for this may be due to angle-moment relationship being different for walking and running. Walking resulted in a more concave dorsiflexion curve then that reported for running. Indicating a slower, less powerful movement. Running also had a larger range of motion of the ankle joint than walking. These differences resulted in regression equations for the springs which were not as effective when interchanged. By recalculating the regression constants for the auxiliary source, the auxiliary source was able to improve the springs' contribution for the final phase of plantarflexion. These results lead to the recommendation that there should be two separate systems, one for walking and one for running, for optimal results.

5.4 Future Research

This data provided us with preliminary information necessary for the design requirements for a better prosthetic leg. However, more data should be collected to confirm the results, and determine if the reported regression constants could be determined by anthropometric information. It would then need to be determined if a spring and auxiliary source could be designed with the regression constants. The design would also have to consider other design constraints such as: weight, size, performance during temperature extremes, durability, reliability, and cost.

6. Conclusion

This project produced regression equations which would reproduce moments of the ankle joint during gait. The information, to our knowledge, has not previously been published. The information is essential to design a prosthesis that would function similar to an intact limb. The study showed that the current passive prosthesis could not produce an adequate moment for sufficient propulsion of the foot. It was shown through this investigation that an auxiliary power source was a necessary component for a prosthetic leg to function similar to an intact leg. Further research exploring the possible mechanism for the spring and auxiliary source is necessary, along with the determination of a method of interfacing the prosthetic ankle with the amputee.

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