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A comparative analysis of the moments of inertia of natural and above-knee prosthetic legs

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With much love to the families Tolbert and Williams and Susan, Katie, Grace, and Cory for believing in me and keeping me going when I wanted to give up.

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INTRODUCTION

Everyone expends energy to move. As people grow accustomed to their bodies, they also grow accustomed to expending a certain minimal level of energy for motion. The amount of energy expended is related to the inertial characteristics of the body and the body segments used for motion. In walking, running, and other forms of bipedal motion, the principal body segments used are the thigh, leg, and foot and the hip and ankle joints. The inertial characteristics of these segments have a direct impact on the individual's energy expenditure as well as fluidity of motion. By changing those characteristics, energy expenditure can be increased or decreased, and the resultant fluidity of motion can be improved or degraded.

When a person has an above-knee amputation, the inertial characteristics of the residual limb are changed. In losing the foot, lower leg, and a portion of the thigh, the total mass of the leg is reduced as is the amount of muscle power available for movement (Wickiewcz et al., 1983). Through the use of a prosthetic leg, the amputee regains most of the functionality of the lost leg. However, the prosthesis is considerably lighter than the natural leg so inertial characteristics are only partially restored. This change in inertial properties may be responsible for a limp or "hitch" in amputee gait (Farber and Moreinis, 1995). In effect, the symmetry of the legs has been changed through the removal and prosthetic replacement of one of the natural legs.

In the non-pathological case, a person's legs are similar enough so that some symmetry in functionality may be expected (Farber and Moreinis, 1995). It is not uncommon for one leg to be stronger than the other but, for the most part, their weight-bearing capacity and contribution to locomotion to be about equal. For pathological cases in which one leg is either physically different from the other (e.g. shorter) or damaged by injury, disease, or congenital conditions (e.g. one leg being atrophied), the symmetry of gait will degenerate if not disappear (Elftman, 1966; Soderberg, 1986). The physical manifestations range from a slight limp to an inability to walk. Although both legs are present, they are physically dissimilar enough to impair motion. The case of amputation falls into the pathological category. Amputees using a prosthesis do have enough dissimilarity in their legs to cause a limp while walking. While the limp is not due solely to the difference in inertial properties, some of it may be attributed to that difference.

Originally, the goal of prosthetists was to create legs light enough to allow the amputee to move without expending inordinate amounts of energy. In the past, amputees used prosthetic legs that were heavier than their natural legs. These heavier legs required more energy to move, often leading to premature fatigue when walking. By decreasing prosthesis weight, newer designs have decreased energy expenditure and virtually eliminated premature fatigue.

Of the many inertial characteristics of solid bodies, this study will consider only moment of inertia ($\boldsymbol{I}$). The moment of inertia of a body is a constant of proportionality relating an object's angular velocity to its rotational kinetic energy:

$$KE_{Rot} = \frac{1}{2} I \omega^2 \qquad [1]$$

where $\mathbf{KE_{rot}}$ is the rotational kinetic energy, $\mathbf{I}$ is the segment moment of inertia, and ω is the segment angular velocity. From equation 1, the amount of energy expended in rotating the leg during motion is directly related to the leg's moment of inertia. If the moment of inertia is increased, the amount of energy required to maintain or attain a given angular velocity increases. Similarly, if the moment of inertia is decreased, the amount of energy necessary to maintain a given angular velocity decreases. The effect is demonstrated by an individual running both with and without ankle weights. When using ankle weights, the mass of the leg is increased as is the moment of inertia. The result is an increase in the amount of energy needed to maintain a normal pace. The amputee, like the runner in the example, adapts to expending a different amount of energy due to changes in leg inertial characteristics. How the amputee adjusts to these new parameters varies.

Essentially, the amputee has to relearn the dynamics and mechanics of walking as the remaining muscles are asked to function in a manner to which they are unaccustomed. While learning to walk with the prosthesis, the amputee's gait changes to allow for the most comfortable, fluid motion. Theoretically and practically, amputee gait should not change to fit the prosthesis. Instead, the prosthesis should be adjusted to the amputee's gait. Although prosthetists work to decrease this compensation, it still occurs to varying degrees (Farber and Jacobson, 1995; Titianaova and Tarkka, 1995; Rose and Gamble, 1994). Some of this gait adjustment may be due to the different feel of the prosthesis as the amputee has to grow accustomed to a lack of sensation during walking-- not feeling a natural leg swing or foot contacting the ground. The inertial dissimilarity between the prosthesis and the natural leg may account for some of the different feel and subsequent abnormal gait.

The purpose of this project was to investigate how an inertially-matched prosthesis affects amputee gait and energy consumption. The research was conducted in two parts. The first study investigated the effect of changing the inertial characteristics of one leg on gait. This unilateral test simulated some of the effect of changing limb inertial properties of unilateral amputees. This part of the study examined and confirmed the hypothesis that if the inertial characteristics of one leg are changed, the change would not only affect the motion patterns of the loaded leg, but those of the unloaded leg as well. This first part acted as the basis for the second part of the project, which examined the differences between the moments of inertia of natural and prosthetic legs.

The second part of the study concerned comparing the moments of inertia for exoskeletal and endoskeletal prosthesis and human test subjects (utilizing in vivo, non-invasive means). This part of the study investigated the hypothesis that natural and prosthetic legs do not have the same or similar moments of inertia. The exoskeletal prosthesis has a customized socket incorporated into its structure such that it is homogeneous with the rest of the prosthesis. With an endoskeletal prosthesis, the socket attaches to the prosthesis and is removable just like any other component. In order to make the endoskeletal prosthesis as modular and interchangeable as possible, excluding the socket, the rest of the components need to be able to fit a range of amputees without an inordinate amount of customization (Figure 1). Since the exoskeletal prosthesis is essentially a custom fit device, its characteristics are more closely matched to a single user.

The current method for prescribing prosthetic legs to amputees involves a combination of generalized and custom fittings (Brosseau, 1995-1996; Barr, 1995). The prosthesis is prescribed based primarily on amputee weight, which is used with tables of general prosthesis parameters for knee, lower leg or pylon, and ankle-foot components. Once

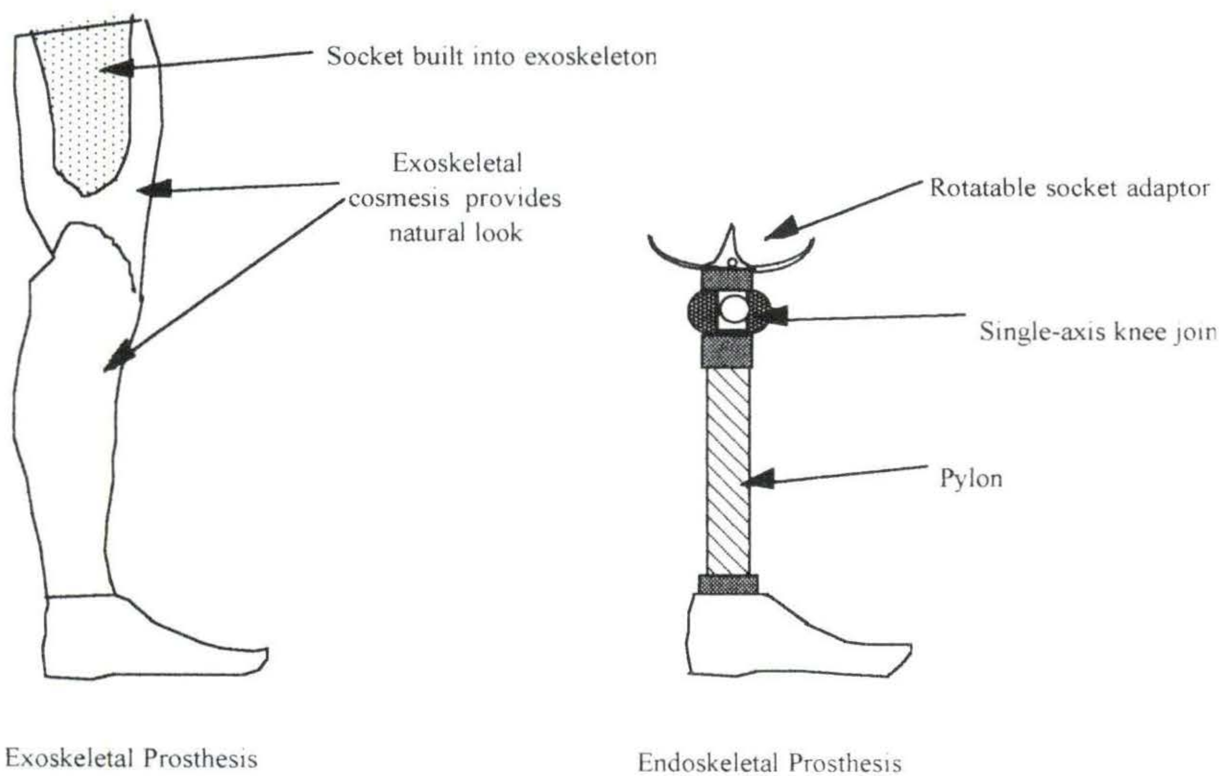


Figure 1. Anatomy of Exo- And Endoskeletal Prostheses

the leg is assembled, the prosthetist makes necessary modifications to improve the amputee's gait and overall comfort. The socket designed specifically for the amputee's residual limb is the only completely customized part of the prosthesis. Although amputee weight is taken into consideration when selecting prosthesis components, it is not used to match the inertial characteristics of the prosthetic and natural legs.

The chief criterion for prosthesis prescription is the amputee's medical insurance and personal wealth. More often than not, an amputee will use a lesser prosthesis because it is the best limb that person can afford (Brosseau, 1995-1996). This is most evident among endoskeletal prosthesis users. Because they are pieced together from standardized parts, an endoskeletal prosthesis can be customized by the amputee much in the same way one can customize a car. Wealthier amputees can afford the more expensive, higher quality components, whereas poorer amputees cannot.

An amputee's medical insurance may pay for an endoskeletal prosthesis. However, that prosthesis may not be suitable for an athletic or active amputee's lifestyle. An amputee with the money to buy higher quality prosthesis components will not have to change or give-up an active lifestyle out of danger of damaging the prosthesis. On the other hand, the amputee relying solely on medical insurance for a prosthesis may have to reduce his activity level to fit the performance capabilities of the prosthesis.

When comparing the prices of exoskeletal and endoskeletal prostheses, the exoskeletal prosthesis is usually less expensive. The actual individual price of a prosthesis varies depending on the needs of the individual amputee. Still, the general price range of an exoskeletal prosthesis is lower than the endoskeletal. This being the case, poorer amputees are more likely to use an exoskeletal prosthesis. Because an amputee may have to spend considerable additional money to get a prosthesis suitable for an active lifestyle, many amputees unable to afford such a prosthesis are forced to become less active or risk damaging or destroying the prosthesis.

The smaller mass of the endoskeletal prosthesis has advantages. A smaller mass, given the mass distribution of the prosthesis, means a smaller moment of inertia which, in turn, means a smaller rotational kinetic energy is needed to reach the same angular velocity (Equation 1). With less muscle mass available for motion and limb control, a smaller moment of inertia keeps the amputee from experiencing increased reaction moments at the hip which could lead to hip problems. The moment experienced at the hip joint is partially a function of the inertial parameters of the leg.

On a natural leg, flesh and muscle help to reduce the effective moment experienced at the hip. However, an amputee lacks the flesh and muscle to dampen or reduce the forces producing moments at the hip. Often the socket fits in a manner that prevents the residual limb from damping those moments, so the amputee runs the risk of developing hip problems. Increasing the weight of a prosthesis using the current technology only serves to increase the resultant moment at the hip which would be felt as a resistance to initiating, continuing, or stopping motion. For example, a heavy leg provides resistance to motion during swing phase and resistance to the termination of swing phase. The motion of the leg begins as an initial driving moment which serves to start the rotation of the prosthesis. A heavier leg requires a greater driving moment. Similarly, a greater stopping moment must be generated to slow and stop a heavier leg due to its greater momentum. The reduced musculature of the residual limb makes controlling a heavy prosthesis difficult as well. An amputee having a very short residual limb would have more difficulty controlling a heavier prosthesis than an amputee with a longer residual limb because the longer limb provides more muscle over which the amputee has control. A longer residual limb often also means a lighter prosthesis can be used (less components needed).

In this study, the moments of inertia were determined and compared, natural leg data to prosthesis data, to reveal that current prostheses are indeed not inertially similar to natural legs. Since this study was conducted using non-amputees, further research using unilateral amputees should be conducted to determine the effects of an inertially-matched prosthesis on amputee gait.

The study was conducted using non-amputee test subjects instead of unilateral amputees because the remaining natural limb of unilateral amputees would be adapted to the amputation. Amputation of one leg often leads to increased musculature and mass in the remaining limb as new demands are placed on the limb for motion (Soderberg, 1986; Wickiewcz et al., 1983; Titianaova and Tarkka, 1995). Non-amputees allowed for a testing situation free of the influence of this adaptation.

This report describes the methods used to evaluate the effects of unilateral loading on both the loaded and unloaded limbs and to determine the moments of inertia for the prosthetic and natural legs. The report then examines the unilateral loading results and their implications for inertia matching when combined with the results from the direct comparison of the natural and prosthetic moments of inertia. Finally, the report makes recommendations for further research into possible effects of inertial matching on actual unilateral above-knee amputees as well as into possibly changing the mass and geometry of the pylon portion of the prosthesis.

METHODS

Test Subjects

Two different sets of test subjects were used in this project. For the first study, two test subjects, one male and one female, participated (Table 1). For the second part, a total of nine test subjects, both male and female, participated (Table 2). The use of human test subjects for this study was approved by the Iowa State University Human Subjects Review Committee.

Table 1. First Study Test Subject Statistical Summary

Test Subject	Sex	Weight (kg)
1	M	81.2
2	F	54.9

Table 2. Second Study Test Subject Statistical Summary

Test Subject	Sex	Age (years)	Height (m)	Weight (kg)
1	M	24	1.88	99.8
2	M	24	1.83	69.0
3	M	27	1.78	68.5
4	M	23	1.78	66.7
5	M	28	1.80	73.9
6	F	25	1.70	59.0
7	M	25	1.75	72.6
8	F	29	1.63	59.9
9	M	24	1.79	76.7
Mean	NA	25	1.78	71.7

Experiment 1: Unilateral Load Testing

For this study, three types of data were collected: video, force plate, and electromyogram.

Videography Data

The subjects were videotaped while walking with five different single-leg weighting conditions at their normal walking rates. The normal walking rate was determined by taking five timed trials of each subject walking a pre-defined distance. The mean time was used to determine the mean normal walking rate.

Once the mean normal walking rate was determined, each subject was videotaped walking on a treadmill without additional weight to obtain control readings. The subjects had position markers placed on the hip, knee, and ankle joints of the left leg to track the paths of these joints during motion. Additional markers were placed on the heel and toe to define the foot in the digitized video.

The subjects underwent five different weighting conditions: no weight, 70.87, 135.23, and 437.15 grams and a second no weight condition (just after the heaviest weight was removed). This fifth weighting condition was included to determine the effect of the sudden lightening of the leg. The weights were strapped to the approximate midpoint of the shank of the left leg and arranged to provide an even ring of weight around the middle of the shank. The weight was applied only to the left leg and the subjects were videotaped only on their left. Figure 2 shows a diagram of the testing area.

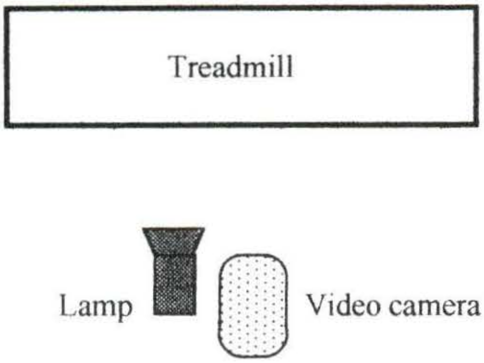


Figure 2. Diagram of Videography Test Area

Once the subjects were videotaped, the video was then digitized and transformed into Cartesian coordinate data which were smoothed and analyzed using the Ariel Performance Analysis System (APAS, Ariel Dynamics, Inc., San Diego, CA).

Force Plate

The force plate data were collected under similar conditions as the video data. The subjects underwent the same weighting conditions as in the video testing with only the left leg weighted. The subjects walked across a force plate first with the left foot then with the right. Figure 3 shows a diagram of the force plate testing area. The force plate data was sampled at a rate of 800 Hz by a computerized data acquisition system of the APAS using an external trigger controlled by the test administrator. To aid in determining the correct timing of the force plate data, the subjects were videotaped walking across the force plate. The video data provided the time information needed to indicate when the trigger was activated and when the

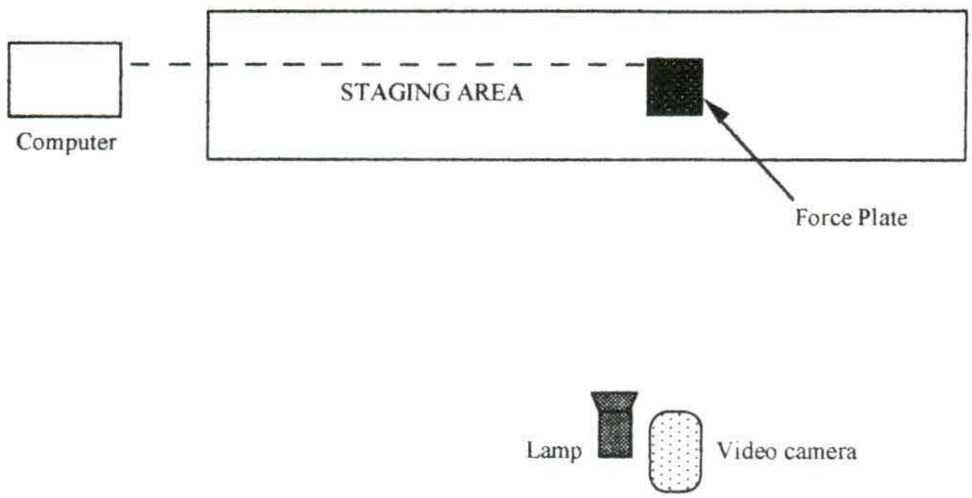


Figure 3. Diagram of Force Plate Test Area

subject actually stepped on the plate. Force plate data were transformed from raw data into force and moment data for analysis using the APAS.

Electromyogram (EMG) Data

The EMG data were again gathered by having the subjects walk on a treadmill set to the natural walking speed of the subject. Skin-mounted electrodes were placed on the vastus medialis and gastrocnemius of the right and left legs (Elftman, 1966). The signal from the vastus medialis provided information on leg movement used in determining the swing phase and heel strike portions of gait. The signal from the gastrocnemius provided information on foot flexion and the toe-off portion of gait.

The EMG signals were amplified and sampled at a rate of 950 Hz by a computerized data acquisition system, the Biopac ACKnowledge system (Biopac Systems, Inc., Goleta, CA), which used two external triggers in series. The primary trigger was controlled by the test administrator with a secondary trigger provided by a heel switch located on the foot of the test subject. As the test subjects walked on the treadmill, they continually triggered the heel switch. However, the computer did not begin sampling data until the test administrator pressed the primary trigger while the heel switch was also being triggered. EMG data were gathered for all five previously described loading conditions. Figure 4 is a diagram of the EMG testing area.

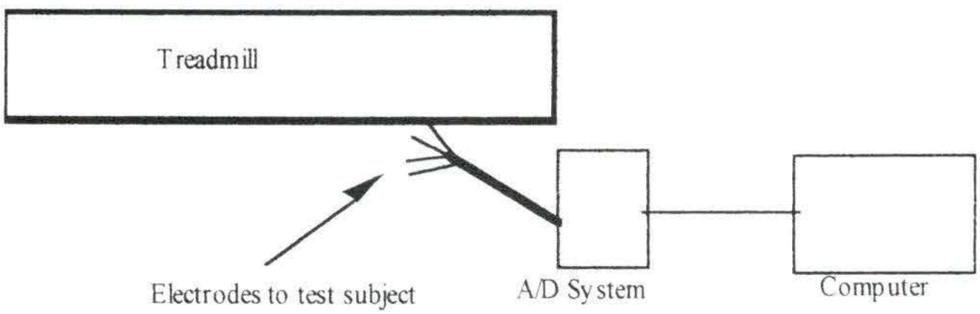


Figure 4. Diagram of EMG Test Area

Once the EMG data were gathered, it was processed for analysis by:

- passing the data through a Rectangle high-pass filter with a cut-off frequency of 20 Hz to remove motion artifacts from the data,
- taking the absolute value of the data as a form of full-wave rectification,
- passing the data through a Rectangle low-pass filter with a 3 Hz cut-off frequency to obtain a linear envelope for the data, and

- integrating the linear envelope to get a measure of the total EMG activity as well as activity throughout motion.

Experiment 2: Comparative Moment of Inertia Testing

In order to make a comparison between the moments of inertia of natural and prosthetic legs, ***I*** must be known. The ***I***s of the prosthetic legs were determined empirically, and the ***I***s of the natural legs were determined empirically using indirect measurement and anthropometric data.

Determining the Moment of Inertia of the Prosthetic Leg

Two different prostheses were examined, one exoskeletal and one endoskeletal. An exoskeletal prosthesis has a functional cosmesis which is also the functional part of the leg. An endoskeletal prosthesis has its functional parts covered with a foam rubber cosmesis whose only function is esthetic.

Since the prosthesis components are rigid, the inertial properties were assumed to be effectively constant. The moment of inertia was determined by suspending and oscillating the prosthesis as a compound pendulum. The period of oscillation is related to the moment of inertia by the relation:

$$T = 2\pi\sqrt{\frac{I_o}{mgl}} \qquad [2]$$

where **T** is the period of oscillation, $\mathbf{I_0}$ is the moment of inertia with respect to the suspension point, **m** is the mass of the prosthesis, **l** is the distance from the suspension point to the center of mass, and **g** is gravitational acceleration.

The prosthesis was suspended from the top of the knee joint on the endoskeletal prosthesis and from the top of the socket on the exoskeletal prosthesis, oscillated, and allowed to complete two full cycles per trial with a total of ten trials conducted per prosthesis (Figure 5). From the ten trials, an average period was calculated which was subsequently used to calculate the moment of inertia.

Determination of Moment of Inertia of the Natural Leg

The moment of inertia of the leg-foot complex had to be determined non-invasively and in vivo for test subjects so as to assure values accurate for living test subjects. A quick-release test apparatus described in the literature (Hatze, 1975; Winter, 1990) provided a means of doing so utilizing two measurements (Figure 6).

The moment of a rotating body is given as:

$$M = I\alpha \qquad [3]$$

where **M** is the applied moment, **I** is the moment of inertia and α is the angular acceleration of the body. Equation 3 can be rewritten for an expression of **I** in terms of **M** and α as:

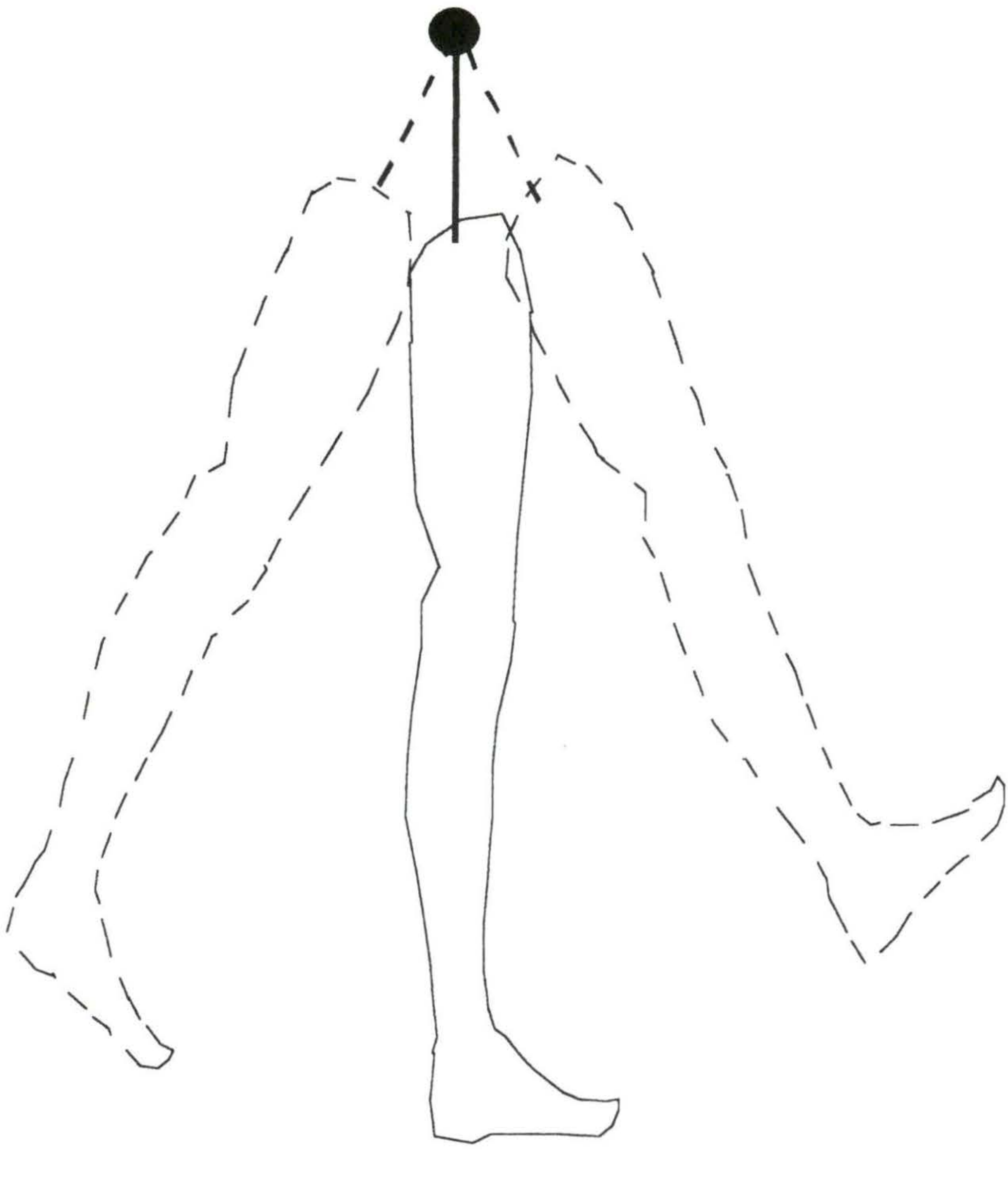


Figure 5. Diagram of Oscillatory Test Set-Up for Prosthesis

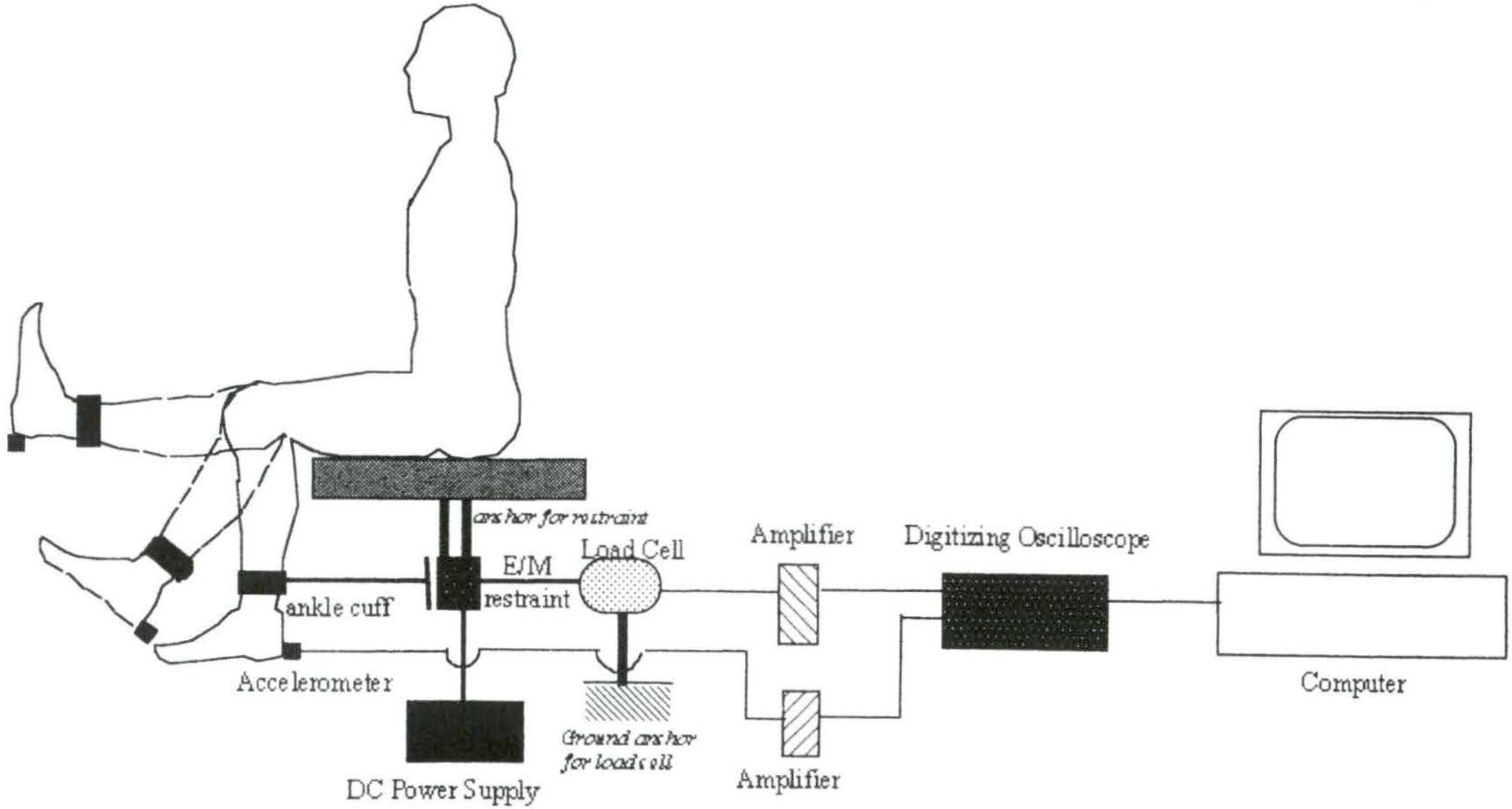


Figure 6. Testing Set-Up For Empirical Determination of the Moment of Inertia of a Natural Leg

$$I = \frac{M}{\alpha} \qquad [4]$$

From classical dynamics:

$$M_P = Fd \qquad [5]$$

$$a = r\alpha \qquad [6]$$

Equation 5 tells us that a moment about some point **P** is equal to the product of a force **F**, applied a distance **d** from point **P**, and the perpendicular distance **d**. Equation 6 tells us that the linear acceleration, **a**, of a body is given by the product of the radius of rotation, **r**, of the body and its angular acceleration, α. Equation 6 can be rewritten for an expression of α in terms of **a** and **r** as:

$$\alpha = \frac{a}{r} \qquad [7]$$

Equations 5 and 7 can be substituted into Equation 4 to express moment of inertia in terms of the force applied to a body and the resultant linear acceleration:

$$I = \frac{Fdr}{a} \qquad [8]$$

For the current project test sequence, the rotating body was considered to be the lower leg and foot, with the center of rotation approximately at the knee (femoral condyles). The applied horizontal force **F** was measured by the load cell, and linear horizontal acceleration **a** measured by the accelerometer. The distance **d** was measured as the approximate distance from the knee (femoral condyles) to the center of the restraint cuff, with the distance **r** measured from the knee (femoral condyles) to the center of the accelerometer.

During a test, the subject wore a sandal with a heel-mounted accelerometer and a cuff connected to the load cell around the ankle of the same leg. Also attached to the load cell assembly was an electromagnet which acted as a restraint for the leg. As the subject exerted a maximal force to extend the knee, the load cell measured that force. After a steady-state force measurement was recorded, the electromagnet was turned off allowing the subject to fully extend the knee. The heel-mounted accelerometer measured the acceleration of the foot during the swing phase of knee extension. The acceleration at the instant of release of the electromagnetic restraint and the force values from the load cell were used to calculate the moment of inertia of the subject's leg. As a secondary method, the moment of inertia of the subject's leg was calculated using anthropometric data tables (Miller and Nelson 1973). This provided a means of comparing individual data to the more generalized and readily available tabulated data.

Instrument Calibration

Load Cell Calibration

This project used an Interface SM-1000 (Interface, Inc., Scottsdale, AZ) load cell with a maximum loading capacity of 453.59 kg. Such a large capacity was chosen to assure that the load cell would not be damaged during testing and that the effective measurement range of the project would be within the linearly elastic range of the load cell.

Three calibration weights were applied to the load cell. Loadings of 1.27, 2.31, and 4.54 kg were applied for five trials for each weight. For each set of five trials the data were averaged and the variance calculated. The average load cell reading for each loading was then plotted against the applied load value to provide a voltage versus load calibration curve (Figure 7). The factory test specifications for the load cell indicate its linearity to be within 0.03% for loadings less than 70% of the 453.59 kg maximum.

Accelerometer Calibration

Since impact-style acceleration data were gathered, an impact calibration method was used for the accelerometer. A PCB, Inc. accelerometer was mounted to a small nut suspended from an attachment point on a wall by a length of copper magnet wire. The accelerometer-nut assembly was then raised to a height of 11 3/8 inches and released. The resulting impact with the wall registered as an acceleration trace on the digitizing oscilloscope. Only one impact was permitted in each trial (i.e. no rebounding impacts occurred). The acceleration data were numerically integrated and used to determine the calibration constant for the accelerometer (Table 3).

Determination of the Accelerometer Calibration Constant

An accelerometer provides a means of measuring the acceleration of the body to which it is attached. The acceleration as measured by the accelerometer and the actual acceleration are related by the equation:

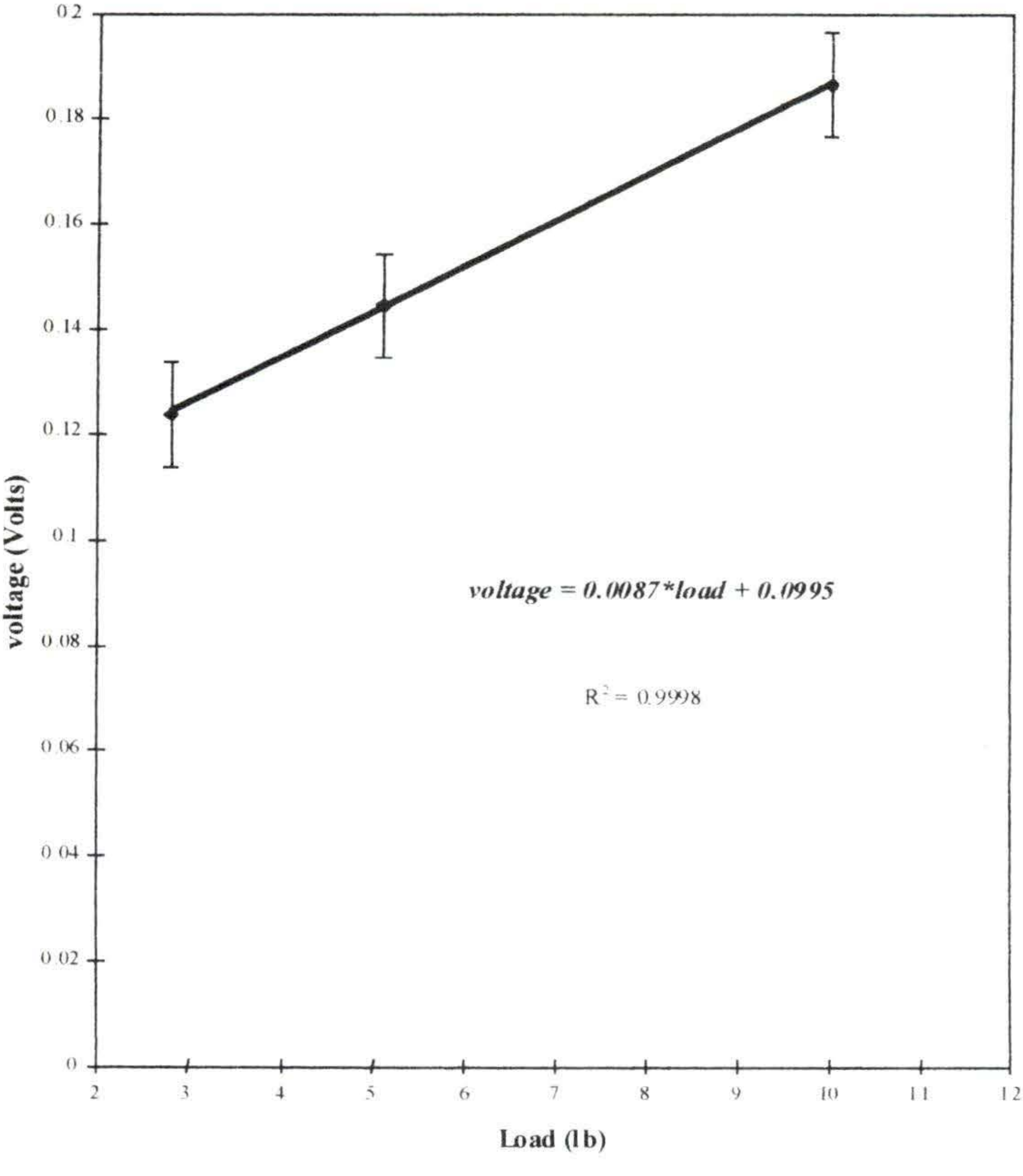


Figure 7. SM-1000 Load Cell Load vs. Voltage Calibration Curve

Table 3. Trial Data Integrals
Note: Some trials were eliminated due to errors in recalling the stored data.

Trials	Integral
Trial 2	2.55
Trial 3	2.52
Trial 4	2.51
Trial 5	2.52
Trial 6	2.52
Trial 7	2.74
Trial 10	2.76
Trial 13	1.95
Trial 15	2.39

Average: 2.50 Variance: 0.06
Accelerometer Calibration Constant: K= 2.50 ± 0.15

$$K = \frac{e}{a} \qquad [9a]$$

which can be rewritten as:

$$a = \frac{e}{K} \qquad [9b]$$

where **K** is the proportionality constant, **e** is the measured acceleration and **a** is the actual acceleration.

Since the accelerometer will measure what is effectively an impact-style acceleration, an impact calibration technique was used. The impulse-momentum equation was used:

$$M_{sys} v_{sysf} = M_{sys} v_{sysi} + \int F dt \qquad [10]$$

where $\mathbf{M_{sys}}$ is the mass of the system, $\mathbf{v_{sysi}}$ and $\mathbf{v_{sysf}}$ are the respective initial and final system velocities, and the integral is the sum of the forces applied to the system over a time **dt**.

From Newton's Second Law:

$$\sum F = \sum M_{sys} a \qquad [11]$$

Substituting Equation 11 into Equation 10 gives:

$$M_{sys} v_{sysf} = M_{sys} v_{sysi} + \int M_{sys} a dt \qquad [12]$$

The mass of the system is assumed constant throughout the experiment so both sides of the equation can be divided by $\mathbf{M_{sys}}$ to give:

$$v_{sysf} = v_{sysi} + \int a dt \qquad [13]$$

Subtracting $\mathbf{v_{sysi}}$ from both sides gives an expression for the integral of the actual system acceleration with respect to time:

$$v_{sysf} - v_{sysi} = \int a dt \qquad [14]$$

Applying the initial condition of the system being at rest and the effects of the conservation of energy, assuming negligible energy losses, the following statements can be made:

$$v_{sysi} = 0 \qquad [15a]$$

$$v_{sysf} = \sqrt{2gh_o} \qquad [15b]$$

where $\mathbf{g}$ is gravitational acceleration and $\mathbf{h_o}$ is the initial release height of the suspended accelerometer system. Substituting Equations 15a and 15b into 14 gives:

$$\sqrt{2gh_o} = \int a dt \qquad [16]$$

Substituting Equation 9b into 16 gives:

$$\sqrt{2gh_o} = \int \frac{e}{K}\, dt \qquad [17]$$

Since **K** is a constant, it can be moved outside of the integral and rearranged to yield:

$$K = \frac{\int e\, dt}{\sqrt{2gh_o}} \qquad [18]$$

where the integral is the area under the acceleration trace. Given the numerical nature of the accelerometer data, the integral was evaluated using Simpson's Rule with End Correction (Hornbeck, 1975). Since the end correction term involves derivatives, second-order Gregory-Newton Forward and Backward finite differences (Hornbeck, 1975) were used to determine the necessary derivatives.

RESULTS AND DISCUSSION

This project covered two major studies concerning unilateral loading of test subjects' legs and moment of inertia comparison.

Experiment 1: Unilateral Loading of Test Subjects' Legs

Videography

Both linear and angular displacement, velocity, and acceleration were examined. For the linear parameter analysis, the knee and hip joints were the foci while the shank and thigh were the foci for the angular parameter analysis. Linear and angular parameters were both analyzed and presented because of the findings concerning the effects of the unilateral loading. The linear data show the effects of loading on both the vertical and horizontal motions of joints; whereas, these effects are not readily apparent from the angular parameters which are useful primarily for rotation analyses of the thigh and shank segments.

For comparison, the ranges of motion and variances of the knee should be greater than those of the hip as the hip's angular motion contributes to the knee's rectilinear motion (Rose and Gamble, 1994). Thus, the knee should have greater ranges and variances for linear displacement, velocity, and acceleration. Examining the linear parameters yields some interesting results. The horizontal knee displacement was, as expected, the largest of the linear displacements. The knee's vertical motion is noticeable, but it is not as great as the horizontal motion (Figure 8).

As expected, the data show an initial decrease in knee mobility with the addition of the 70.9 g weight. This initial decrease may be due to the increase in weight without an appreciable increase in muscle power output. In essence, the subjects felt something of a "ball and chain" weighting effect. The weight of the leg had been increased, but the muscles continued to exert about the same amount of force needed for the lighter leg. As the subjects walked with the weight, they adapted to the additional weight, which can be seen primarily in Subject 2's knee data (Figure 9).

Looking at Subject 2's horizontal knee displacement shows that her mobility decreases with the first load.

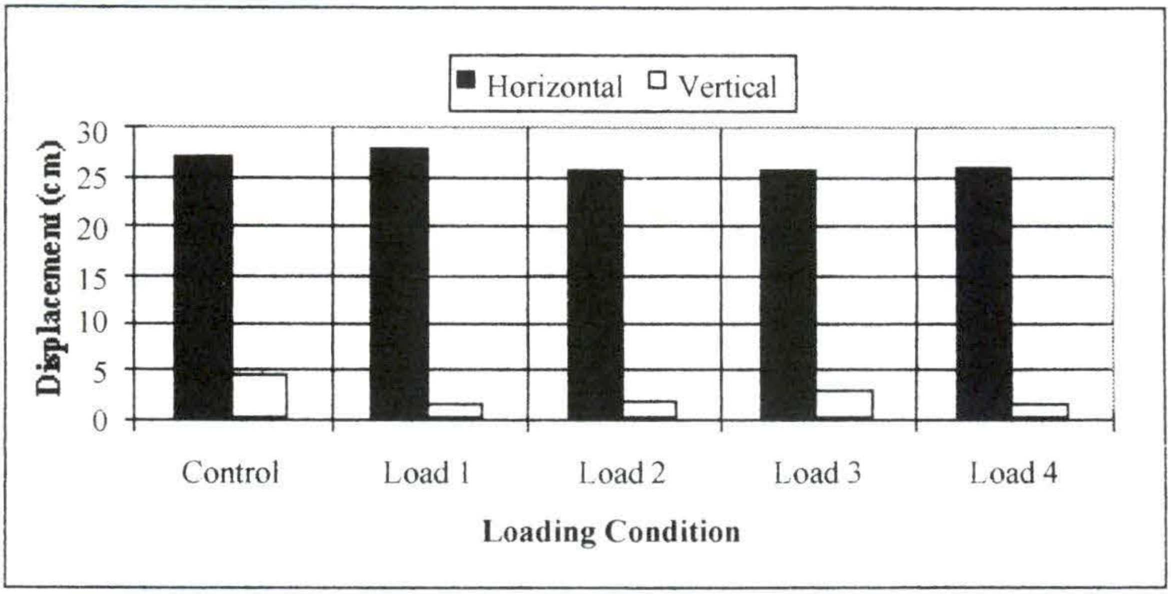


Figure 8. Representative Knee Vertical and Horizontal Linear Displacement Ranges

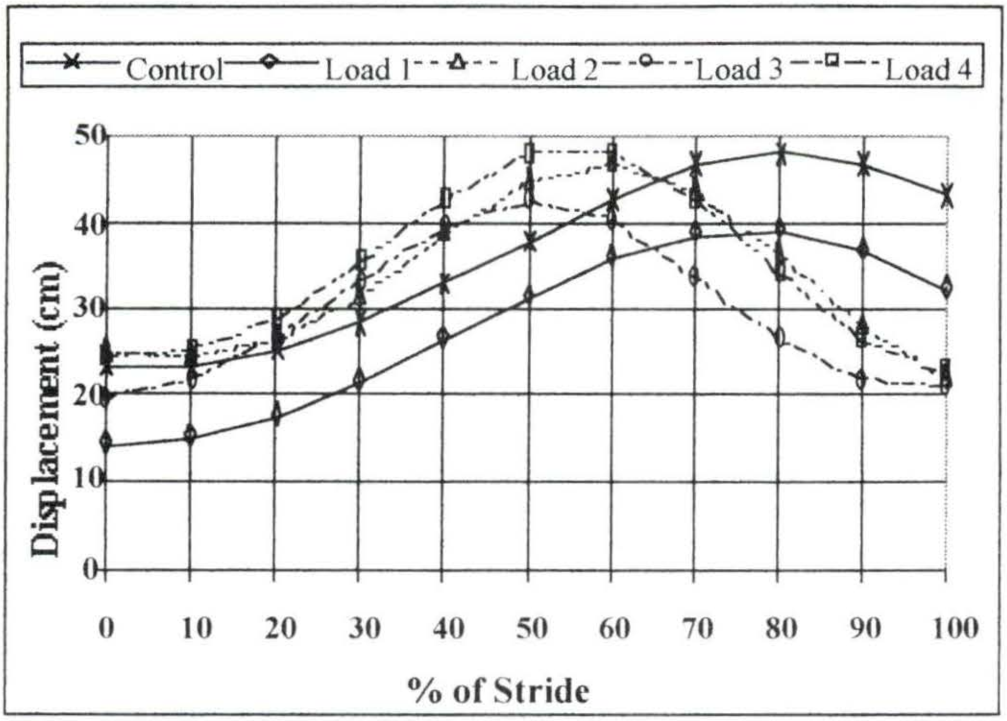


Figure 9. Test Subject 2 Linear Knee Displacement Curve

After the second load is applied, the initial displacement is greater than the initial control displacement. However, the third load causes an initial displacement less than the control yet greater than the first loading even though the third load is more than six times the weight of the first load. Subject 2 knew the third and heaviest weight was being attached to her leg but didn't know it was so much heavier than the previous weights. Interestingly, the initial displacement of Load 4, the condition just after the heaviest load is removed, is greater than the control value (Figure 9). Both conditions involve no additional weight added to the leg, but during the Load 4 case, the leg seems to still be responding as if a weight is attached to the leg. In fact, all the linear parameters show considerable deviation between the control and load 4 cases (see data tables in Appendices A and B). In Subject 2's data, the load 4 condition shows the second largest values behind the load 2 condition. Subject 1's data do show similar phenomena, but it is not as apparent as it is in Subject 2's data. This could be due to the difference in body weight, muscle mass, and leg length between Subjects 1 and 2.

A universal phenomenon was the decreased time by which the parameters reach their extreme values with increased loading. Peak displacements, velocities, and accelerations generally seem to be reached faster after the addition of weight to the leg. Also, after the heaviest weight is removed the parameters continue to reach peak values more quickly than in the control case. This decrease in time to peak values is due to the inertial effects of the added weight. As the leg is accelerated, the weight is accelerated and gains momentum. As the knee reaches its "normal" maximum displacements, velocities, and accelerations, the additional momentum is imparted to the knee to keep it moving (Figure 10). These phenomena are less pronounced in the hip linear data due to the relatively small linear translations. The important indicator of the effect of loading on the hip can be seen in the angular behavior of the thigh as the hip is the prime pivot point for the thigh (Figures 11a and 11b). The angular parameters for the thigh show the aforementioned phenomena to a somewhat lesser degree (Figure 11a). The angular parameters show relatively small variation from control values. The values do change, but the motion is fairly consistent with the control pattern. This consistency is most likely due to the difference in the types of joints and,

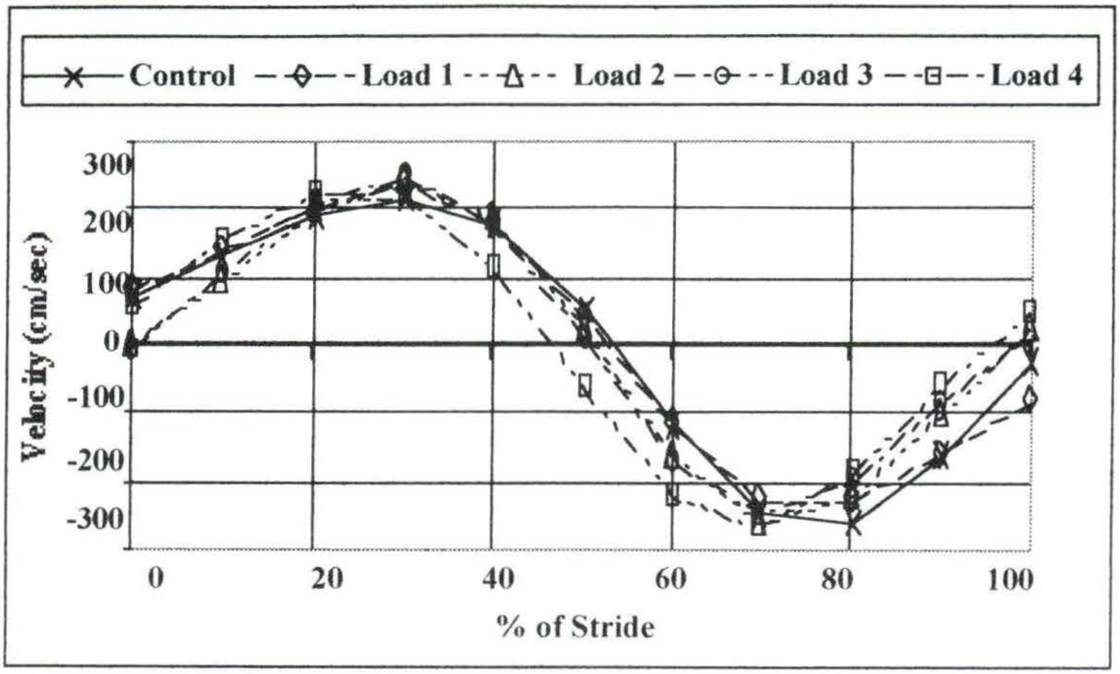
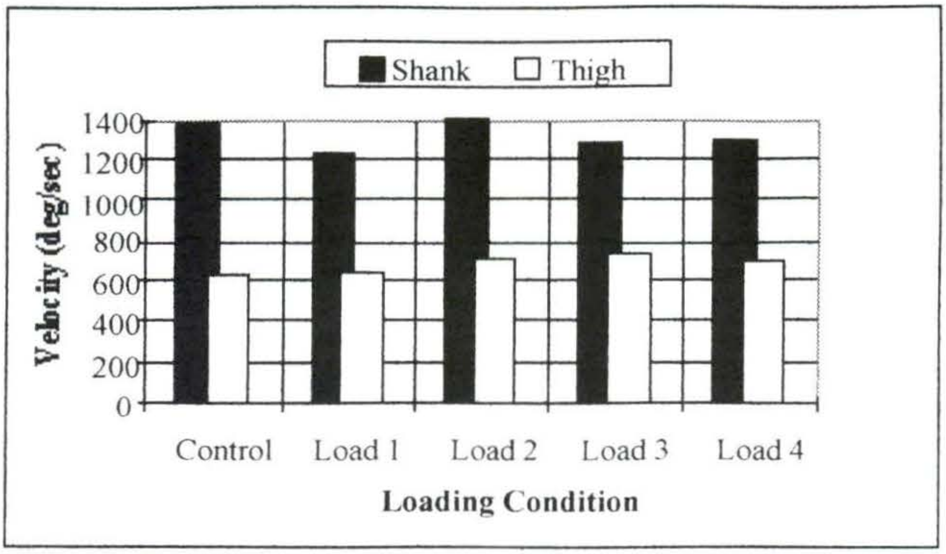
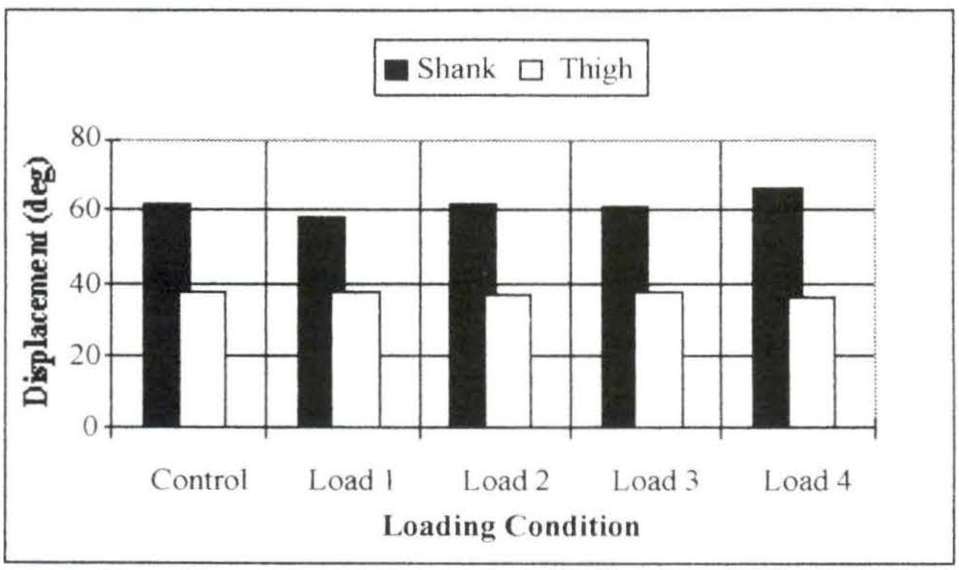


Figure 10. Test Subject 2 Knee Linear Velocity Curve

hence, the types of motion they undergo. As the hip is a ball-and-socket joint, its motion will be primarily rotational with translation coming from translation of the pelvis. That is quite different from the knee which is a hinge joint with one end fixed to the hip. Alone, the knee's hinge characteristics will allow it to undergo primarily rotation. However, the joint undergoes an arc-shaped translation due to the proximal end of the joint being attached to the hip by a long lever arm, the thigh. The longer the thigh, the greater the linear parameters will be. In fact, the hip's angular parameters are larger than those of the knee, shown by the shank (Figures 11a and 11b). That is, the pivot point for the shank is the knee so the shank's



(a) Angular Velocity Range



(b) Angular Displacement Ranges

Figure 11. Shank and Thigh (a) Angular Velocity and (b) Angular Displacement Ranges

rotational behavior is a measure of the rotational behavior of the knee. The shank values appear greater than those of the thigh but are actually a summation of the thigh's rotational behavior and its own. Thus, subtracting the thigh values from those of the shank gives a true measure of the knee's angular parameters without the majority of the hip's influence (Figure 12).

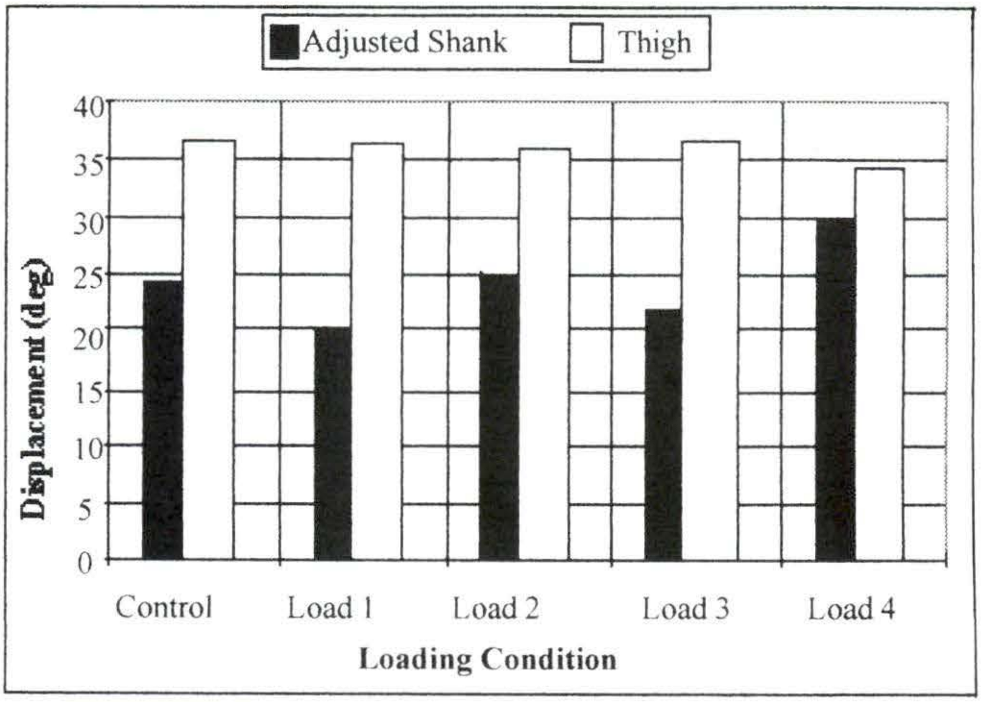


Figure 12. Adjusted Shank and Thigh Angular Displacement Range Comparison Graph

Force Plate

The force plate data show no appreciable difference between right and left foot reaction forces in the vertical direction (Figure 13). As expected, the vertical forces at heel-strike and toe-off are greater with the increased weight, but the increase at toe-off is not as great as that at heel-strike. This is due to the lesser degree of leg control at heel-strike with the added weight. During the swing phase, the leg swings forward exposed only to the action of the muscles to constrain its motion. At toe-off, the leg is constrained not only by the muscles but by friction with the plate and a considerable normal force. Because the heel-strike contact shows the influence of the weight with fewer contributing forces, it provides a good indicator of the additional control the leg muscles must provide due to the additional weight (Figure 14).

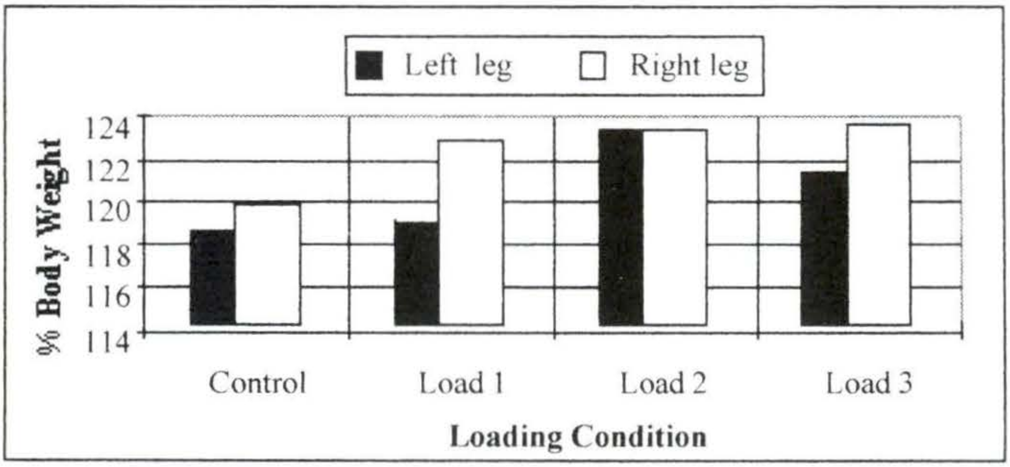


Figure 13. Representative Left Leg and Right Leg Maximum Vertical Force

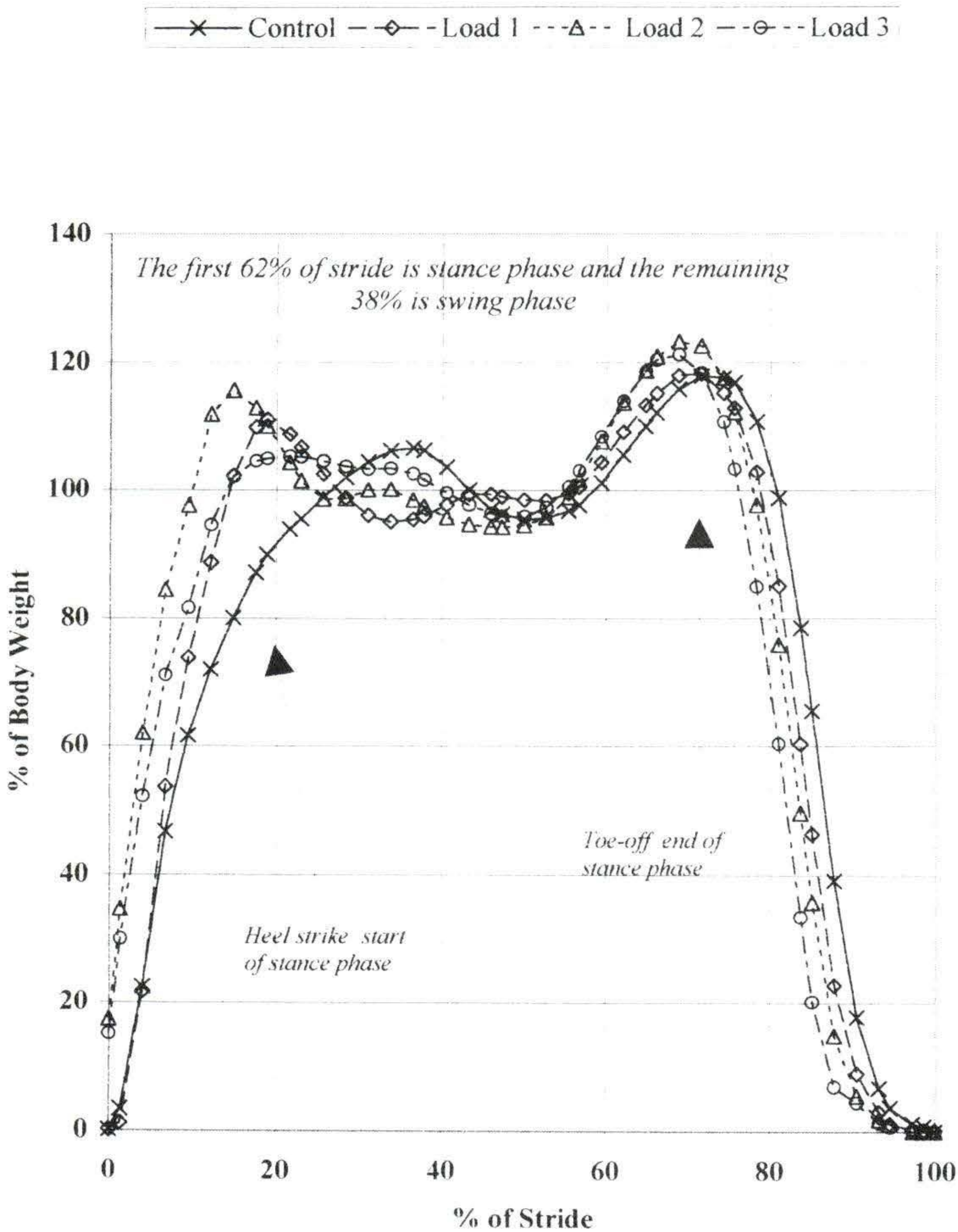


Figure 14. Representative Left Leg Vertical Force Plate Data

As the hip and knee rotate the leg, the added weight increases the leg's momentum. When the muscles act to bring the weighted leg down for heel-strike, the leg's momentum is again transferred to the hip in an attempt to continue the rotation of the leg. The muscles must then act with more force to reverse the rotation for heel-strike to occur without slowing gait. This increased muscle force brings the heel into harder contact with the ground resulting in a higher heel-strike registered force (Miller and Nelson, 1973). At toe-off, the weighted leg is under full control of the muscles. Thus, toe-off is apparently "softer" than heel-strike. The same phenomena can be seen in the medial-lateral force data (Figure 15).

The foot-force plate contact time is virtually constant with small variations. Also, the magnitude of the weighted condition forces at toe-off do stay relatively close to the control values. Again, the magnitude variation was most apparent at heel-strike. The values rise above the control level as the applied weight is increased. The exception seems to be the data for Subject 1's right foot in which the values at heel-strike were less than the control value, with the magnitudes at toe-off greater than the control value. This apparent behavior could be due to suspect data or operator error.

The center of pressure analyses for both feet show some increase in the total migration of the center of pressure (Figure 16). For both Subjects 1 and 2, the total migrations for the left leg show more variation than those for the right leg, suggesting that the

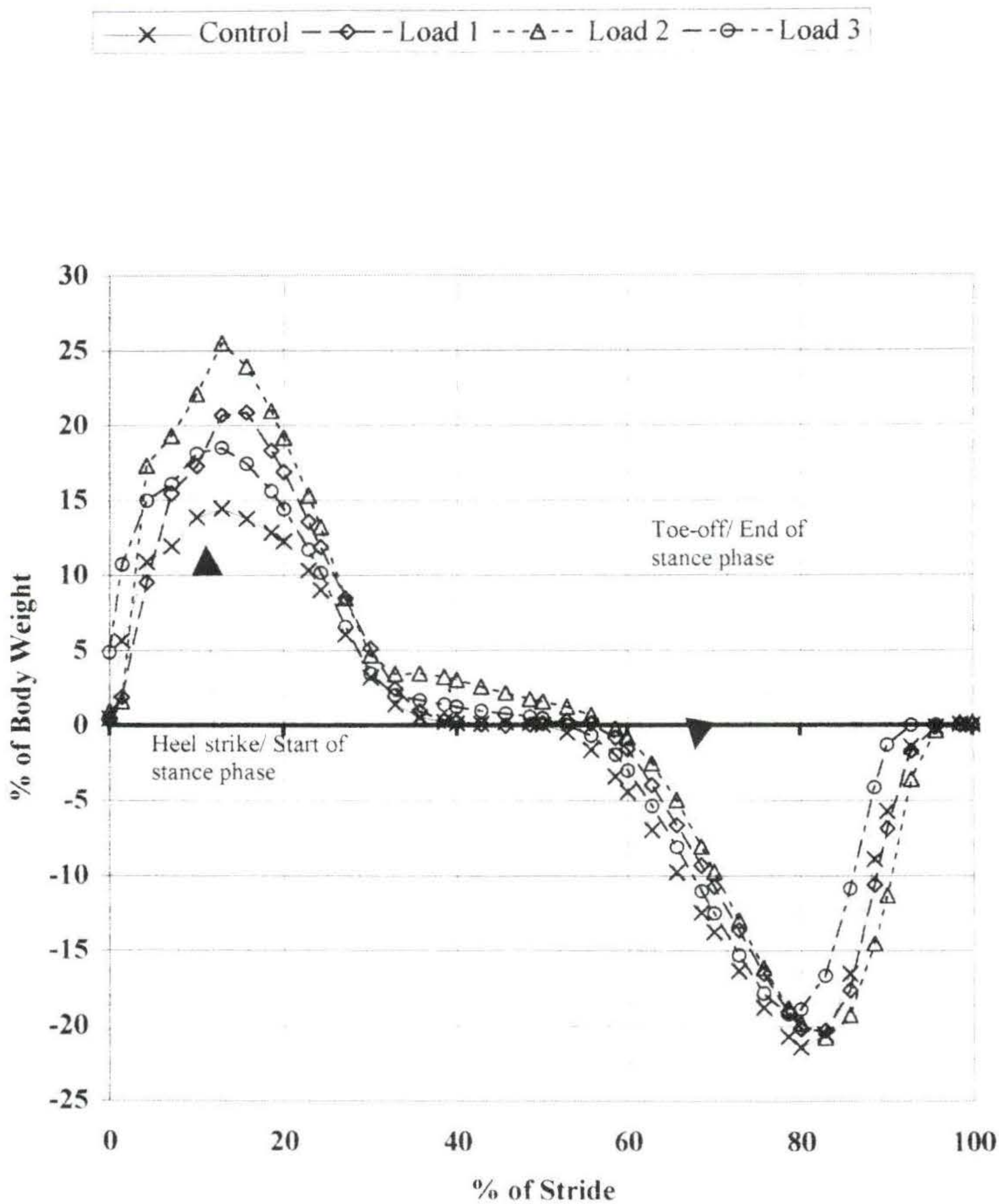


Figure 15. Representative Left Leg Medial-Lateral Force Plate Graph

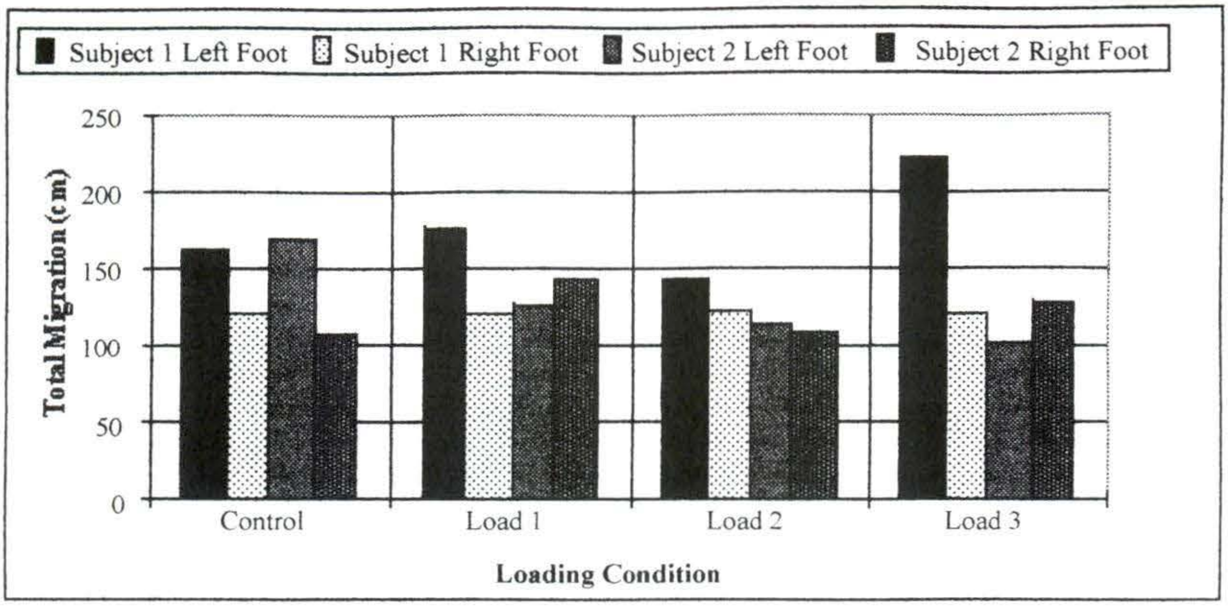


Figure 16. Center of Pressure Total Migration Comparison Graph

center of pressure of the left leg is changing position more than that for the right leg. This effect may be an additional inertial effect of the added weights. As the leg moves during the swing phase, the additional weight might cause a deviation from its regular path of motion. This would result in the foot and, thus, the center of pressure moving as well. There was less variation in the right foot center of pressure.

Another reason for the large deviations in the migrations as well as the variation in the total migrations of the right foot may be the subjects' anticipating stepping on the force plate. During this phase of the testing, several trials had to be performed for each loading condition. As a result, the subjects may have anticipated stepping on the force plate correctly. If they felt they were not going to step on the plate as close to the center as possible, they might have made minor step adjustments to assure they "hit the mark." Perhaps the best remedy to this problem is to have the force plate hidden so the subjects cannot see exactly where the plate is and, therefore, cannot anticipate stepping on it.

Electromyogram (EMG)

The EMG data provide some of the more interesting insights into the effects of the additional weight on subject gait. The integrated EMG (IEMG) data were used as measures of the total EMG activity over the time of the recording. Theoretically, the larger the IEMG value, the greater the EMG and muscle activity (Figure 17).

The data show a mixture of increases and decreases in EMG activity under the loading conditions. For Subject 1, the overall average EMG increases with loading while only the overall quadriceps IEMG increases (Figures 18a and 18 b).

The overall gastrocnemius IEMG decreases with loading. Overall refers to how the values for the various conditions behave with respect to the control value. An overall increase was defined as at least two out of three of the loading conditions resulting in an increase over the control value. For Subject 2, the data show overall average EMG increases in the left leg muscles and overall decreases in the right. The IEMG data for Subject 2's left quadriceps increases overall, while the other IEMG values show an overall decrease.

Theoretically, all the muscles should show increases in EMG and IEMG with the increases in the quadriceps being the most pronounced (Winter, 1990). The quadriceps (quads) are responsible for swinging the leg through during the swing phase in a controlled

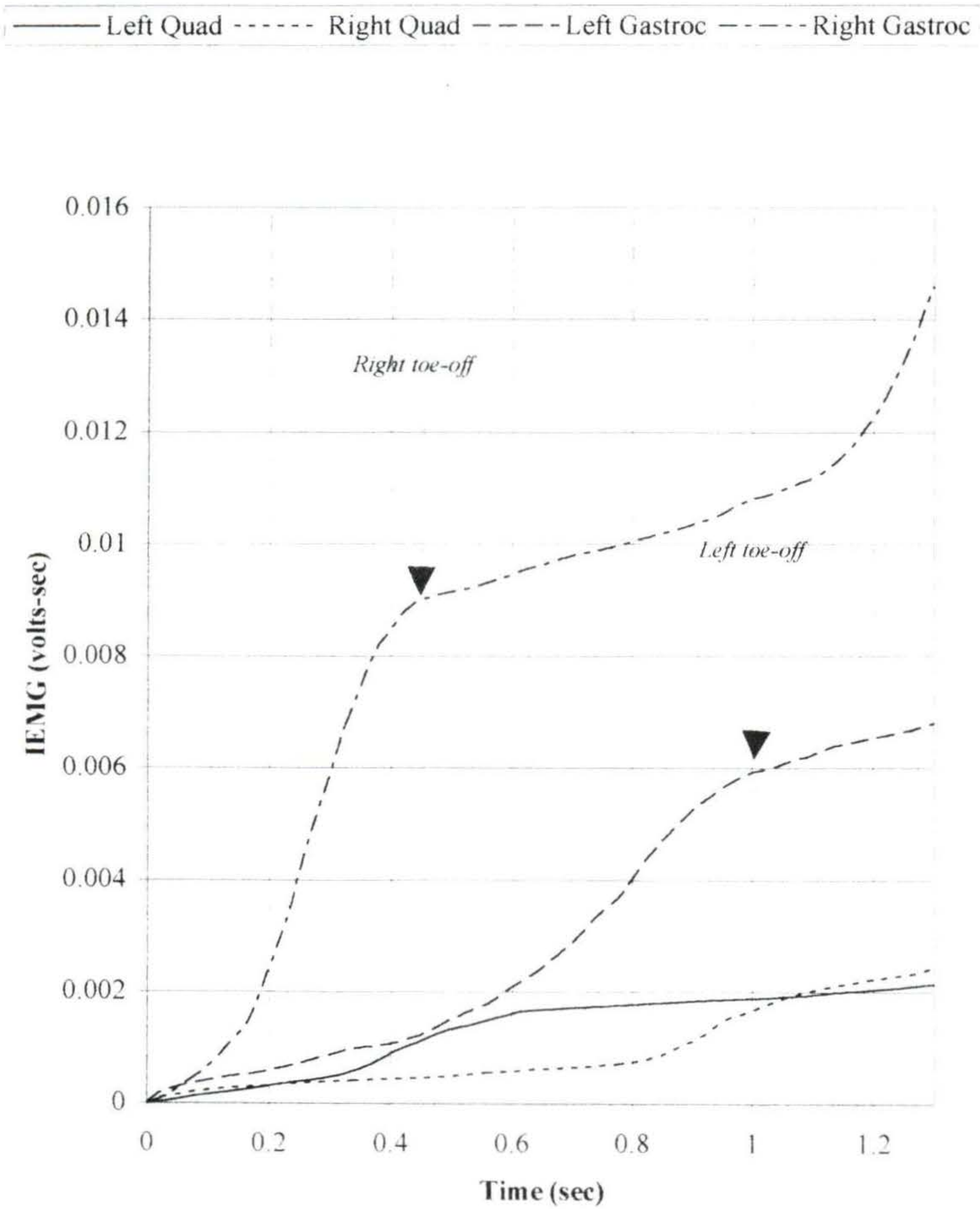
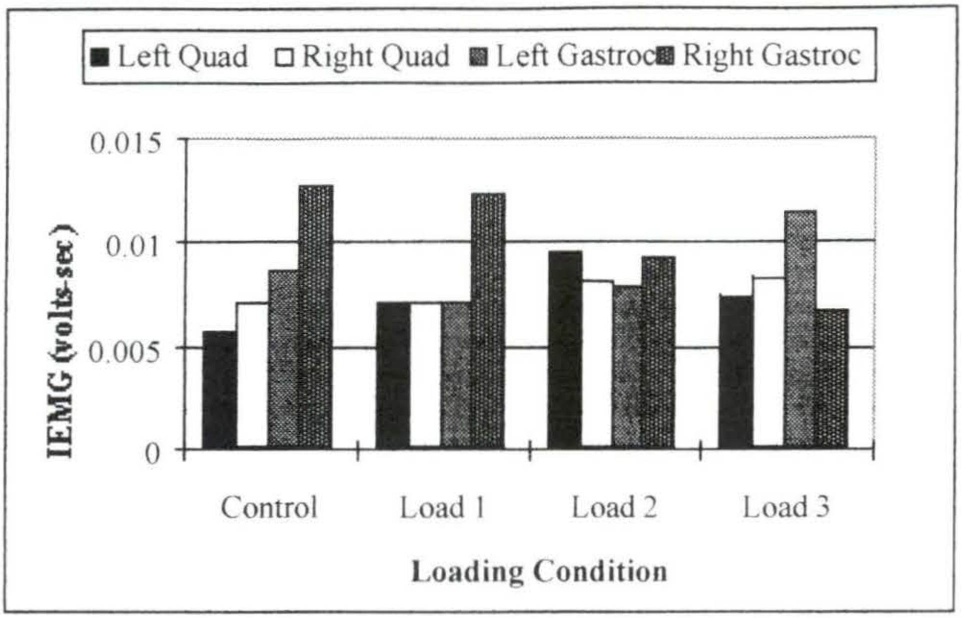
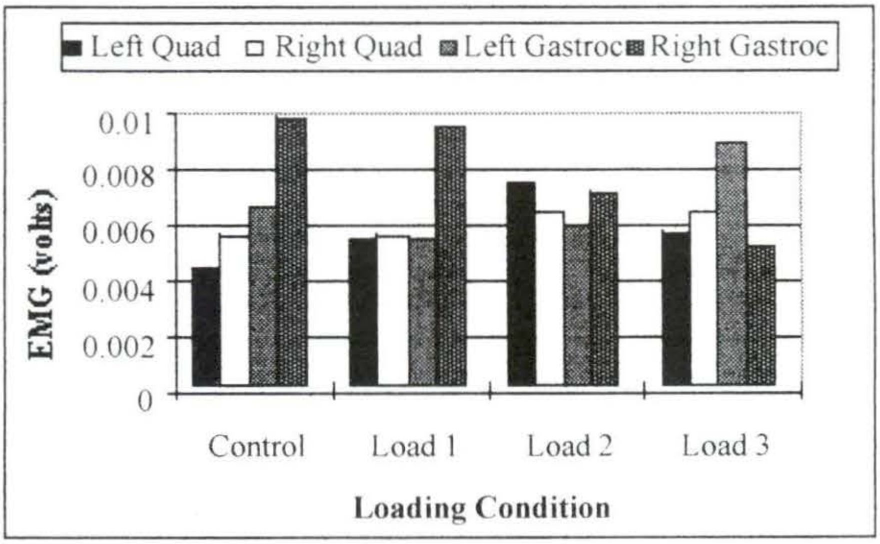


Figure 17. Representative IEMG Graph for Control Loading Condition



(a) Maximum IEMG



(b) Average EMG

Figure 18. Test Subject 1 (a) Maximum IEMG; (b) Average EMG

manner. The additional weight would make the quads work harder to start the swing of the leg, maintain control of the swing, and decelerate the leg at the end of the swing.

Experiment 2: Comparative Moment of Inertia Testing

The moment of inertia data for the prostheses and the test subjects is presented in Tables 4 and 5. The natural leg data show the empirical data are consistently larger than the anthropometric data by 14 to 54 percent. For comparison purposes, the exoskeletal prosthesis would be considered a useable prosthesis for subjects 1 through 5 and 9, with the endoskeletal prosthesis considered useable for subjects 3 through 7.

Table 4. Prosthesis Moment of Inertia Data

Prosthesis	Moment of Inertia ($kg\text{-}m^2$) (standard deviation)
Endoskeletal	0.127 (std. dev.: 0.001)
Exoskeletal	1.467 (std. dev.: 0.183)

Table 5. Natural Leg Moment of Inertia Data

Test Subject	Empirical Data ($kg\text{-}m^2$)	Anthropometric Data ($kg\text{-}m^2$)	% Difference
1	1.609	0.944	-41.34
2	0.978	0.671	-36.91
3	0.680	0.580	-14.71
4	0.655	0.564	-13.89
5	0.773	0.644	-16.69
6	0.913	0.457	-49.95
7	1.286	0.597	-53.58
8	0.622	0.424	-31.83
9	1.138	0.658	-42.18

The prostheses were matched to their respective subjects by subject height and weight. A comparison of the natural empirical and prosthetic data is presented in Table 6.

The data show the exoskeletal prosthesis fits its subject group better than the endoskeletal prosthesis fits its group. As expected, it does not fit the subjects outside of the test subject range but does fit the included range fairly well.

Table 6. Comparison of Natural Leg and Prosthesis Data

Test Subject	% Difference Exoskeletal	% Difference Endoskeletal
1	-48.84	-92.13
2	50.00	-87.04
3	115.74	-81.37
4	123.97	-80.66
5	89.78	-83.61
6	60.68	-86.12
7	14.07	-90.15
8	135.85	-79.63
9	28.91	-88.87

The endoskeletal prosthesis does not fit its test subject range as well but has considerably less variation in its percent difference with the natural data. The exoskeletal prosthesis has a percent difference range of -48.84% to 123.97% ($\sigma = 64.81$), while the endoskeletal percent difference range is considerably smaller (-90.15% to -80.66%, $\sigma = 3.87$). Leaving out the two extreme difference values for the exoskeletal prosthesis does not improve the difference variation to the same degree as the endoskeletal difference variation ($\sigma = 39.04$). The importance of the small difference variation is primarily in evaluating the compatibility between a prosthesis and a range of subjects. Taking this into consideration, the endoskeletal prosthesis is more compatible with a variety of subjects than the exoskeletal prosthesis. Examining the apparent overlap of the subject ranges for subjects 3 through 5 shows that the endoskeletal prosthesis does a better job of fitting these subjects. Its high compatibility allowed it to consistently fit those three subjects better than did the exoskeletal prosthesis. In fact, if the endoskeletal's subject range is expanded to include all the test subjects, regardless of weight and height, it still outperformed the exoskeletal prosthesis ($\sigma = 4.45$).

The main observation of this study is that both prostheses have moments of inertia which are quite different from that of the natural legs. The exoskeletal prosthesis has a moment of inertia considerably larger than most of the test subjects' legs because of its larger mass and how it is distributed throughout the prosthesis. Similarly, the endoskeletal prosthesis' moment of inertia is considerably less than the natural legs because of its smaller mass and its distribution. The increase in moment of inertia accompanied with using the exoskeletal prosthesis (for its target range excluding the first test subject) would seemingly dictate an increase in energy expenditure to successfully use and control the prosthesis during walking or any other type of motion using the legs. Similarly, the smaller moment of inertia suggests a decrease in energy expenditure when using the endoskeletal prosthesis. Whether or not the actual energy changes follow these predictions must still be determined through amputee testing.

So, the question of the effect of inertia-matching is raised once again. Inertia-matching would most likely lead to amputees using endoskeletal prosthetic legs with larger $\boldsymbol{I}$ than in current use, but exactly how much larger is not currently known. Some middle-ground between minimum prosthesis $\boldsymbol{I}$ and maximum limb control needs to be researched.

One piece to the puzzle may lie in the difference between the exoskeletal and endoskeletal prostheses. Besides a weight differential, a key difference between the two is their mass distribution. The exoskeletal prosthesis mass distribution is more like the natural leg's mass distribution (center of mass not along the natural leg's centroidal axis) while the endoskeletal limb has its mass center on the centroidal axis of the pylon.

In an endoskeletal prosthesis, approximately 83.6 percent of the weight resides almost equally in the knee and the foot. Since the knee is usually located close to the distal end of the residual limb, its moment at the hip is not as great as that of the foot. The foot is on the distal end of the pylon which acts as a very long lever arm. The long lever arm allows the foot to create a greater moment at the knee and, subsequently, the hip.

CONCLUSION AND RECOMMENDATIONS

The research demonstrated that changing the inertial characteristics of a leg will indeed change a person's gait characteristics and that prosthetic legs have very different moments of inertia from natural legs. The exoskeletal prosthesis did seem to be more closely inertially matched to the natural leg than did the endoskeletal, a result due mainly to the geometry and weight of the exoskeletal prosthesis. The geometry of the exoskeletal prosthesis more closely matches that of the natural limb than does the endoskeletal, and the majority of the weight of the exoskeletal prosthesis is in the composite cosmesis body. The endoskeletal prosthesis has its weight centered around the geometric center of the pylon using a foam rubber cosmesis of negligible weight for a natural look.

The masses of the knee and foot act to shift the prosthesis center of mass either proximally or distally from the geometric center of the pylon because of the high proportion of total leg weight the knee and foot have in addition to the exoskeletal prosthesis' mass distribution. A heavier knee leads to a net proximal center of mass shift, while a heavier foot leads to a net distal center of mass shift. A proximal shift shortens the effective moment arm just as a distal shift lengthens it. Assuming the mass of the prosthesis stays constant, modifying the mass distribution of the pylon such that the moment of inertia was decreased would decrease the resultant moment the leg imposes on the hip. Taking this into consideration, a heavier truncated conical pylon, with the knee mounted at the base and the foot at the apex, might be a suitable way to match moment of inertia without increasing the resultant moment at the hip. Figure 19 compares a theoretical truncated conical pylon to the current style being used.

Consider an endoskeletal prosthesis with a truncated conical pylon (TCP) having the same length and moment of inertia as the endoskeletal prosthesis tested. The new leg with truncated conical pylon could have a mass of approximately 5.11 kg as compared to the 1.83 kg mass of the tested endoskeletal. The different mass distribution allows the TCP to have a larger mass while having the same ***I*** as the standard cylindrical pylon of smaller mass. If the ***I*** of the prosthesis were matched to the natural leg, the geometry of the pylon could be changed to allow for an inertially matched, lightweight prosthesis.

A more comprehensive study into the effects of inertially matched prosthetics on amputee gait and health is needed. This study did point out the differences between the moments of inertia between prosthetic and natural legs. However, the next step is to conduct tests with amputees using inertially matched prosthetics. Such a study should be conducted over a time period of at least one year to allow amputees to grow accustomed to using various limbs.

An additional parameter that should be considered is amputee comfort. An inertially matched prosthesis may improve amputee gait and energy consumption, but if the amputee is not comfortable using the limb even after a suitable adjusting period, the limb might not be a viable solution. The amputee must have the most comfortable leg possible as discomfort will

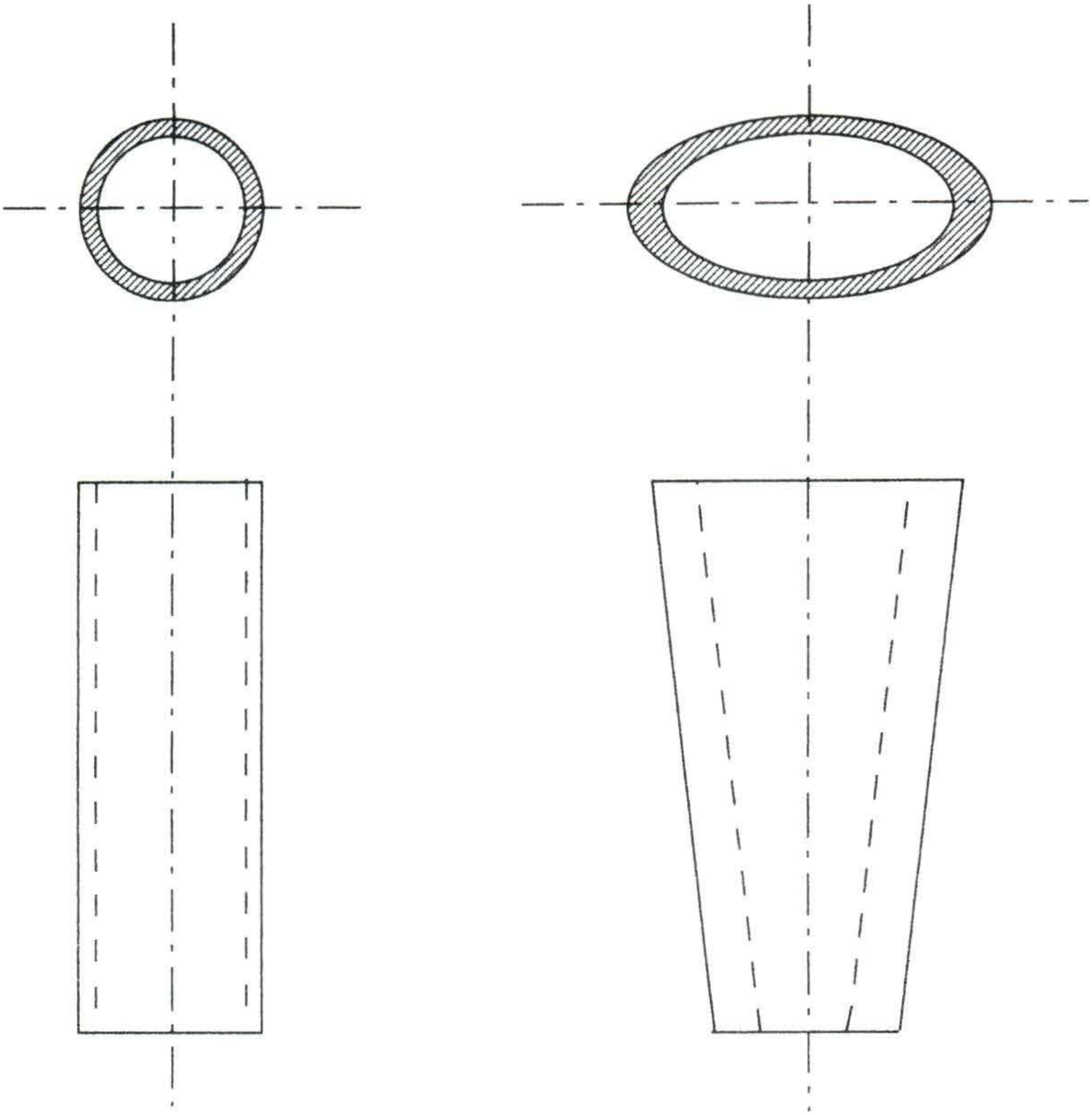


Figure 19. Standard and TCP Endoskeletal Prosthesis Pylons

most likely lead to irregular use. The situation is similar to buying a pair of shoes that provide excellent shock absorption and stability while not being comfortable. The buyer will only wear the shoes when good shock absorption and stability are needed. When good shock absorption and stability are not needed, the buyer will most likely wear more comfortable shoes. More often than not, an amputee will not have the luxury of changing to a more comfortable leg. Thus, comfort is a very important parameter that should be monitored and maximized.

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The men and women who served as test subjects for this project.

APPENDIX A. TEST SUBJECT 1 VIDEOGRAPHY DATA

Knee Displacement

Note: Displacement units are centimeters

Horizontal		Control	Load 1	Load 2	Load 3	Load 4
	Minimum	23.52	14.18	22.05	19.62	22.71
	Maximum	48.09	38.86	46.79	42.54	47.99
	Ranges	24.58	24.67	24.74	22.92	25.28
	Variances	95.76	89.89	79.76	72.64	90.85
Vertical						
	Minimum	-13.67	-14.09	-14.04	-13.66	-14.05
	Maximum	-12.50	-13.44	-11.79	-12.87	-11.76
	Ranges	1.17	0.65	2.25	0.80	2.29
	Variances	0.15	0.047	0.58	0.08	0.54

Knee Velocity

Note: velocity units are centimeters per second

Horizontal		Control	Load 1	Load 2	Load 3	Load 4
	Minimum	-148.12	-156.50	-246.43	-228.96	-257.94
	Maximum	155.70	152.19	215.82	207.16	220.95
	Ranges	303.81	308.69	462.25	436.13	478.89
	Variances	9.559E03	10.653E03	27.162E03	22.923E03	28.684E03
Vertical						
	Minimum	-4.93	1.31	-22.68	-4.62	-12.07
	Maximum	-2.08	2.58	25.56	-0.13	21.54
	Ranges	2.85	1.27	48.23	4.49	33.62
	Variances	0.89	0.18	255.90	2.21	124.31

Knee Acceleration

Note: acceleration units are centimeters per second squared

Horizontal		Control	Load 1	Load 2	Load 3	Load 4
	Minimum	-2.576E01	-2.420E01	-5.186E01	-4.786E01	-5.355E01
	Maximum	1.317E01	1.433E01	2.983E01	3.895E01	3.435E01
	Ranges	3.893E01	3.854E01	8.169E01	8.681E01	8.789E01
	Variances	2.506E06	2.267E06	8.271E06	9.434E06	10.487E06
Vertical						
	Minimum	8.56	-3.80	144.70	-13.46	100.85
	Maximum	8.56	-3.80	144.70	-13.46	100.85
	Ranges	0	0	0	0	0
	Variances	0	0	0	0	0

Hip Displacement

Note: displacement units are centimeters

Horizontal		Control	Load 1	Load 2	Load 3	Load 4
	Minimum	37.98	28.69	38.39	35.02	39.57
	Maximum	43.14	32.49	41.02	37.66	41.48
	Ranges	5.16	3.81	2.62	2.63	1.91
	Variances	4.59	2.06	0.86	0.68	0.50
Vertical						
	Minimum	24.81	24.16	24.91	24.12	24.43
	Maximum	25.14	25.39	26.40	26.15	25.34
	Ranges	0.33	1.23	1.48	2.03	0.91
	Variances	0.01	0.17	0.23	0.55	0.09

Hip Velocity

Note: velocity units are centimeters per second

Horizontal		Control	Load 1	Load 2	Load 3	Load 4
	Minimum	-22.74	-15.12	-44.57	-1.99	-25.65
	Maximum	34.26	21.34	23.42	22.46	21.33
	Ranges	57.00	36.46	67.70	24.45	46.98
	Variances	384.47	153.33	544.03	55.65	303.06
Vertical						
	Minimum	0.39	0.51	-7.38	-24.18	-8.26
	Maximum	1.60	8.66	13.74	23.78	10.03
	Ranges	1.22	8.16	21.12	47.96	18.28
	Variances	0.16	7.98	49.07	253.04	36.77

Hip Acceleration

Note: acceleration units are centimeters per second squared

Horizontal		Control	Load 1	Load 2	Load 3	Load 4
	Minimum	-393.85	-244.57	-381.29	-185.17	-480.81
	Maximum	484.88	189.10	737.05	389.27	829.16
	Ranges	878.74	433.66	1.118E03	574.44	1.310E03
	Variances	129.918E03	28.241E03	208.848E03	37.758E03	224.100E03
Vertical						
	Minimum	-3.66	-35.41	63.37	-143.89	-54.85
	Maximum	-3.66	-5.73	63.37	-143.89	-54.85
	Ranges	0	29.68	0	0	0
	Variances	0	115.65	0	0	0

Shank Angular Parameters in XY-Plane

Note: displacement units are degrees, velocity units are degrees per second, and acceleration units are degrees per second squared

Displacement		Control	Load 1	Load 2	Load 3	Load 4
	Minimum	78.86	79.85	80.49	258.96	76.78
	Maximum	132.39	133.78	134.70	313.26	133.81
	Ranges	53.53	53.94	54.20	54.30	57.03
	Variances	280.10	296.94	325.92	380.65	355.62
Velocity						
	Minimum	81.39	81.76	-756.27	-1.010E03	-925.40
	Maximum	268.49	253.45	293.20	305.85	321.68
	Ranges	187.09	171.69	1.049E03	1.315E03	1.247E03
	Variances	4.056E03	3.978E03	109.523E03	217.499E03	164.493E03
Acceleration						
	Minimum	-3.996E03	-4.261E03	-13.009E03	-15.472E03	-15.096E03
	Maximum	2.481E03	1.966E03	2.285E03	5.892E03	2.306E03
	Ranges	6.477E03	6.227E03	15.294E03	21.364E03	17.402E03
	Variances	3.189E06	3.486E03	24.982E06	42.964E03	28.631E06

Thigh Angular Parameters in XY-Plane

Note: displacement units are degrees, velocity units are degrees per second, and acceleration units are degrees per second squared

Displacement		Control	Load 1	Load 2	Load 3	Load 4
	Minimum	68.16	69.23	65.67	246.43	64.95
	Maximum	97.45	99.30	98.78	278.12	99.62
	Ranges	29.29	30.06	33.11	31.69	34.66
	Variances	127.65	135.24	144.38	142.93	165.48
Velocity						
	Minimum	-198.09	-208.59	-343.48	-331.52	-352.42
	Maximum	181.60	191.14	283.23	284.00	288.02
	Ranges	379.69	399.74	626.71	615.52	640.44
	Variances	14.487E03	17.559E03	47.345E03	44.485E03	52.090E03
Acceleration						
	Minimum	-3.180E03	-3.130E03	-6.968E03	-6.563E03	-7.079E03
	Maximum	1.292E03	1.676E03	3.629E03	4.656E03	4.178E03
	Ranges	4.472E03	4.806E03	10.597E03	11.219E03	11.256E03
	Variances	3.280E06	3.741E06	15.143E06	16.898E06	16.342E06

APPENDIX B. TEST SUBJECT 2 VIDEOGRAPHY DATA

Knee Displacement

Note: displacement units are centimeters

Horizontal		Control	Load 1	Load 2	Load 3	Load 4
	Minimum	35.55	21.15	20.43	23.68	28.11
	Maximum	62.23	48.68	45.68	49.07	53.89
	Ranges	26.68	27.53	25.26	25.39	25.77
	Variances	101.48	102.34	92.74	90.80	97.69
Vertical						
	Minimum	-18.53	-18.66	-19.11	-19.23	-18.26
	Maximum	-14.17	-17.28	-17.52	-16.68	-16.77
	Ranges	4.37	1.39	1.58	2.55	1.49
	Variances	2.26	0.22	0.18	0.55	0.25

Knee Velocity

Note: velocity units are centimeters per second

Horizontal		Control	Load 1	Load 2	Load 3	Load 4
	Minimum	-260.13	-229.79	-260.59	-250.30	-269.56
	Maximum	215.73	232.49	242.22	249.85	220.90
	Ranges	475.86	462.28	502.81	500.15	490.47
	Variances	30.708E03	30.493E03	29.476E03	28.871E03	30.295E03
Vertical						
	Minimum	-2.50E-03	-7.18	-49.04	-30.29	-17.20
	Maximum	26.21	-1.15	9.63	-2.01	14.64
	Ranges	26.22	6.03	58.67	28.28	31.84
	Variances	75.60	4.00	302.44	82.11	110.95

Knee Acceleration

Note: acceleration units are centimeters per second squared

Horizontal		Control	Load 1	Load 2	Load 3	Load 4
	Minimum	-5.055E03	-4.847E03	-5.705E03	-5.747E03	-5.557E03
	Maximum	3.924E03	2.416E03	3.999E03	3.643E03	3.854E03
	Ranges	8.978E03	7.263E03	9.704E03	9.390E03	9.411E03
	Variances	9.902E06	7.489E06	12.284E03	12.065E06	11.487E06
Vertical						
	Minimum	78.65	-18.08	-860.33	-305.81	-100.65
	Maximum	78.65	-18.08	130.53	35.42	-92.83
	Ranges	0	0	990.86	341.23	7.83
	Variances	0	0	109.538E03	17.158E03	9.21

Hip Displacement

Note: Displacement units are centimeters

Horizontal		Control	Load 1	Load 2	Load 3	Load 4
	Minimum	52.73	41.66	37.82	42.12	45.83
	Maximum	56.68	43.99	39.68	44.26	46.79
	Ranges	3.95	2.33	1.86	2.15	0.95
	Variances	2.57	0.75	0.42	0.57	0.11
Vertical						
	Minimum	21.76	20.42	19.56	21.20	21.60
	Maximum	25.24	24.57	23.83	24.27	24.23
	Ranges	3.48	4.16	4.27	3.07	2.63
	Variances	1.24	1.72	1.89	1.05	0.80

Hip Velocity

Note: velocity units are centimeters per second

Horizontal		Control	Load 1	Load 2	Load 3	Load 4
	Minimum	-22.23	-17.98	-22.64	-27.95	-0.11
	Maximum	28.85	12.27	9.25	20.41	5.82
	Ranges	51.08	30.25	31.89	48.36	5.93
	Variances	277.36	83.49	105.85	258.13	3.87
Vertical						
	Minimum	7.25	-45.11	-45.72	-34.53	-26.92
	Maximum	19.50	28.27	35.56	29.88	30.04
	Ranges	12.25	73.38	81.28	64.41	56.97
	Variances	17.97	587.30	726.72	456.39	356.97

Hip Acceleration

Note: acceleration units are centimeters per second squared

Horizontal		Control	Load 1	Load 2	Load 3	Load 4
	Minimum	-284.37	-63.37	-25.76	82.58	-17.80
	Maximum	666.40	356.70	350.23	224.19	-17.80
	Ranges	950.77	420.07	376.00	141.61	0
	Variances	122.347E03	24.843E03	20.732E03	2.884E03	0
Vertical						
	Minimum	-84.89	-257.89	-243.84	-193.24	-170.90
	Maximum	19.07	-193.69	-243.84	-193.24	-170.90
	Ranges	103.96	64.20	0	0	0
	Variances	1.572E03	557.88	0	0	0

Shank Angular Parameters in XY-Plane

Note: displacement units are degrees, velocity units are degrees per second, and acceleration units are degrees per second squared

Displacement		Control	Load 1	Load 2	Load 3	Load 4
	Minimum	70.25	71.43	249.33	252.16	249.52
	Maximum	131.77	128.67	310.88	311.56	314.50
	Ranges	61.52	57.24	61.55	59.40	64.98
	Variances	474.02	442.47	459.07	445.214	527.80
Velocity						
	Minimum	-1.021E03	-830.56	-999.67	-900.33	-876.20
	Maximum	348.44	387.22	382.43	370.06	416.99
	Ranges	1.369E03	1.218E03	1.382E03	1.270E03	1.293E03
	Variances	209.130E03	203.122E03	236.671E03	226.654E03	231.651E03
Acceleration						
	Minimum	-16.043E03	-13.595E03	-15.536E03	-13.962E03	-14.440E03
	Maximum	1.837E03	7.210E03	7.721E03	7.724E03	6.304E03
	Ranges	17.879E03	20.805E03	23.257E03	21.686E03	20.744E03
	Variances	38.492E06	40.574E06	46.359E06	41.740E06	44.798E06

Thigh Angular Parameters in XY-Plane

Note: displacement units are degrees, velocity units are degrees per second, and acceleration units are degrees per second squared

Displacement		Control	Load 1	Load 2	Load 3	Load 4
	Minimum	62.28	61.38	243.54	242.30	245.18
	Maximum	99.16	98.13	279.87	279.45	280.14
	Ranges	36.88	36.75	36.33	37.15	34.95
	Variances	174.71	185.69	182.65	189.47	178.87
Velocity						
	Minimum	-358.26	-321.26	-380.33	-361.77	-370.77
	Maximum	261.40	304.10	325.03	356.73	301.44
	Ranges	619.66	625.36	705.36	718.50	672.21
	Variances	52.703E03	56.195E03	57.151E03	59.008E03	55.537E03
Acceleration						
	Minimum	-6.737E03	-6.369E03	-7.475E03	-7.857E03	-7.523E03
	Maximum	5.229E03	3.208E03	5.360E03	4.329E03	5.188E03
	Ranges	11.966E03	9.577E03	12.836E03	12.186E03	12.712E03
	Variances	16.494E06	12.899E06	22.345E06	21.872E06	21.307E06

APPENDIX C. TEST SUBJECT 1 FORCE PLATE DATA

Right Foot Analysis

Note: units are percentage of body weight

Data Summary		Control	Load 1	Load 2	Load 3
Medial-Lateral	Minimum	-21.76	-17.36	-18.31	-19.68
	Maximum	20.27	22.02	23.46	22.57
	Range	42.02	39.38	41.77	42.25
	Variance	1.50	1.33	1.49	1.55
Vertical	Minimum	0.04	0. 35	10.62	5.09
	Maximum	118.19	116.83	116.43	120.48
	Range	118.15	116.47	105.81	115.39
	Variance	12.67	10.16	5.34	6.55

Left Foot Analysis

Note: units are percentage of body weight

Data Summary		Control	Load 1	Load 2	Load 3
Medial-Lateral	Minimum	-24.76	-24.90	-25.91	-21.29
	Maximum	12.88	14.14	10.21	22.41
	Range	37.64	39.03	36.12	43.70
	Variance	1.26	1.22	1.26	1.56
Vertical	Minimum	0.42	0.22	2.20	0.04
	Maximum	106.56	108.54	99.65	116.43
	Range	106.14	108.32	97.44	116.39
	Variance	10.84	10.68	8.60	10.72

Center of Pressure Analysis

Note: applicable units are centimeters

Left Foot		Control	Load 1	Load 2	Load 3
X-Coordinate	Mean	5.10	2.67	3.50	5.21
	Std. Dev.	1.77	2.38	2.31	4.85
Y-Coordinate	Mean	14.90	14.78	14.34	13.42
	Std. Dev.	8.93	12.58	6.18	20.24
Total Migration		160.39	173.85	142.62	219.64
Right Foot		**Control**	**Load 1**	**Load 2**	**Load 3**
X-Coordinate	Mean	4.58	6.16	0.88	2.47
	Std. Dev.	3.23	2.10	1.60	1.54
Y-Coordinate	Mean	9.47	10.99	13.23	12.07
	Std. Dev.	8.34	4.90	7.34	6.75
Total Migration		120.82	118.98	121.68	120.52

APPENDIX D. TEST SUBJECT 2 FORCE PLATE DATA

Right Foot Analysis

Note: units are percentage of body weight

Data Summary		Control	Load 1	Load 2	Load 3
Medial-Lateral	Minimum	-18.39	-16.58	-25.50	-22.15
	Maximum	22.71	22.00	20.83	20.16
	Range	41.10	38.58	46.33	42.32
	Variance	1.50	1.24	1.71	1.42
Vertical					
	Minimum	10.04	0.07	14.92	0.07
	Maximum	119.65	122.65	123.17	123.36
	Range	109.61	122.58	108.24	123.30
	Variance	7.02	11.38	6.39	12.28

Left Foot Analysis

Note: units are percentage of body weight

Data Summary		Control	Load 1	Load 2	Load 3
Medial-Lateral	Minimum	-21.43	-20.31	-20.83	-19.21
	Maximum	14.78	22.03	25.50	18.49
	Range	36.21	42.34	46.33	37.70
	Variance	1.00	1.26	1.59	1.21
Vertical	Minimum	0.07	0.13	1.14	6.58
	Maximum	118.34	118.87	123.17	121.21
	Range	118.28	118.74	122.03	114.63
	Variance	14.75	12.75	11.30	7.96

Center of Pressure Analysis

Note: applicable units are centimeters

Left Foot		Control	Load 1	Load 2	Load 3
X-Coordinate	Mean	4.18	4.18	5.44	2.63
	Std. Dev.	2.29	1.26	0.85	0.58
Y-Coordinate	Mean	15.34	12.76	11.53	11.32
	Std. Dev.	10.72	6.32	4.51	4.20
Total Migration		167.99	125.28	114.08	100.42
Right Foot		**Control**	**Load 1**	**Load 2**	**Load 3**
X-Coordinate	Mean	4.02	3.37	5.35	5.27
	Std. Dev.	1.05	2.66	0.82	1.69
Y-Coordinate	Mean	11.24	12.91	11.09	12.33
	Std. Dev.	4.70	10.05	4.37	6.62
Total Migration		104.49	142.14	106.27	126.59

APPENDIX E. ELECTROMYOGRAPHY DATA

Test Subject 1 EMG Data Summary

Loading		Lt. Quad.	Rt. Quad	Lt. Gastroc	Rt. Gastroc
Control	Maximum IEMG (volt-sec):	5.42E-03	6.85 E-03	8.37 E-03	12.50 E-03
	Mean EMG (volts):	4.17E-03	5.27 E-03	6.44 E-03	9.61 E-03
Load 1					
	Maximum IEMG (volt-sec):	6.80E-03	6.86 E-03	6.76 E-03	12.03 E-03
	Mean EMG (volts):	5.23E-03	5.28 E-03	5.20 E-03	9.26 E-03
Load 2					
	Maximum IEMG (volt-sec):	9.32E-03	7.93 E-03	7.51 E-03	8.90 E-03
	Mean EMG (volts):	7.17E-03	6.10 E-03	5.77 E-03	6.85 E-03
Load 3					
	Maximum IEMG (volt-sec):	7.18E-03	8.01 E-03	11.18 E-03	6.56 E-03
	Mean EMG (volts):	5.52E-03	6.16 E-03	8.60 E-03	5.05 E-03

Test Subject 2 EMG Data Summary

Loading		Lt. Quad.	Rt. Quad	Lt. Gastroc	Rt. Gastroc
Control	Maximum IEMG (volt-sec):	2.14 E-03	2.42 E-03	6.80 E-03	14.58 E-03
	Mean EMG (volts):	1.64 E-03	1.86 E-03	5.23 E-03	11.22 E-03
Load 1					
	Maximum IEMG (volt-sec):	2.40 E-03	2.26 E-03	6.56 E-03	10.37 E-03
	Mean EMG (volts):	2.00 E-03	1.89 E-03	5.47 E-03	8.64 E-03
Load 2					
	Maximum IEMG (volt-sec):	2.56 E-03	2.18 E-03	7.82 E-03	9.15 E-03
	Mean EMG (volts):	2.13 E-03	1.82 E-03	6.52 E-03	7.62 E-03
Load 3					
	Maximum IEMG (volt-sec):	2.24 E-03	2.07 E-03	6.53 E-03	10.04 E-03
	Mean EMG (volts):	1.87 E-03	1.73 E-03	5.44 E-03	8.37 E-03