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BELOW KNEE AMPUTEE MINIMUM ENERGY EXPENDITURE

By
Daniel James Winarski

A dissertation submitted in partial fulfillment
of the requirements for the degree of
Doctor of Philosophy
(Mechanical Engineering)
in The University of Michigan
1976

Doctoral Committee:

Professor J. Raymond Pearson, Chairman
Assistant Professor James L. Cockrell
Professor Robert C. Juvinall
Professor Joseph E. Shigley
Doctor George H. Koepke

DEDICATION

To My Wife, Donna, and Son, Tyson.

ACKNOWLEDGMENTS

In this project, little would have been accomplished without the sincere cooperation of many key individuals. Because of the considerable assistance rendered, the author feels compelled to express his deep gratitude to a great many who unselfishly gave their time towards the advancement of this disseration.

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PREFACE

As a senior at The University of Michigan, 1970, the author conducted an investigation of the influence of anterior-posterior and medial-lateral alignment adjustments on the knee torque applied to the knee of an amputated limb with a below knee, patellar tendon bearing prosthesis.

A three-dimensional, computerized model was used, consisting of five rigid parts. These parts were the pelvis and a thigh, below knee stump, prosthesis, and artificial foot. No experimental motion data was taken. Instead, various data gathered by assorted previous researchers were pieced together for the model's motion. The anthropometry was gathered from the subject's roommate, a person who had nothing to do with the gait studies of the aforementioned researchers. The anterior-posterior and medial-lateral linear alignment adjustments were then applied to the model's prosthesis without change in the model's original motion.

The result of this effort was that the model's knee torque did not change with alignment adjustment.

After some thought, the author realized that perturbing the model accomplished nothing. It was only in the perturbation of the actual system, the amputee, that the response to the perturbation could be accurately studied.

It was not long after graduating with his Bachelor of Science degree in Mechanical Engineering from The

University of Michigan that the author broke his leg. A walking cast enabled the Lieutenant to continue his duties in the 82nd Airborne Division, but not without occasional difficulty. Perturbations, such as cracks in the sidewalk, sticks, or stones forced the author to respond much like a sudden alignment perturbation would force an amputee to respond.

It was this practical experience coupled with the previous analytical experience that guided the author to return to The University of Michigan. The research was guided by the mistakes made with the previous model. Actual experimental data were taken of each subject's motion as the alignment perturbations were performed. Each subject's anthropometric measurements were used in a much more comprehensive model.

This new model was not applied blindly. The prosthesis worn in the experiment was an instrumented load cell, providing a basis of comparison between empirical and theoretical pylon loads. After the new model was shown to predict reasonably the pylon loads, it was applied to determine the mechanical energy expended during ambulation.

Studying the mechanical energy as a scalar valued function of two variables, flexion-extension alignment adjustment and time, allowed a cumulative energy expenditure surface to be constructed for each of two amputees. Each surface possessed a characteristic geometry, and the minimum mechanical energy was expended at

optimum alignment of the below knee prosthesis. Furthermore, a tightly grouped velocity distribution for each amputee showed that each surface was constructed independently of the average velocity of forward motion of each subject. Finally, this inability of the alignment angle to change the average velocity of forward motion suggested that each system minimized its energy requirement needed to respond to the angular perturbation by retaining its velocity at the optimum.

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LIST OF ABBREVIATIONS

A/D	Analog to digital
APHI	PHI component of acceleration
APHIC	PHI component of CM acceleration
APSI	PSI component of acceleration
APSIC	PSI component of CM acceleration
AR	Radial component of acceleration
ARC	Radial component of CM acceleration
ASA	American Standards Association
AX	X component of acceleration
AXC	X component of CM acceleration
AY	Y component of acceleration
AYC	Y component of CM acceleration
AZ	Z component of acceleration
AZC	Z component of CM acceleration
B/K	Below knee
B/W	Black and white
CM	Center of mass
cp	Center of pressure
CPHI	Cosine(PHI)
CPSI	Cosine(PSI)
D	First derivative operator
D2	Second derivative operator
D/A	Digital to analog
db	Decibel
dc	Direct current

FCX	X component of CM force
FCY	Y component of CM force
FCZ	Z component of CM force
FPS	Frames per second
FX	X component of force
FY	Y component of force
FZ	Z component of force
HAT	Head-arms-trunk
HS	Heel strike
MB	Memory bank
MSB	Most significant bit
MTS	Michigan Terminal System
PHI	Sagittal plane angle
ppm	Parts per million
PSI	Conical angle
PTB	Patellar tendon bearing
QN	Quarter number
R	Segment vector
RC	Segment CM vector
SACH	Solid ankle, cushioned heel
SD	Standard deviation
SI	International System
SPD	Sagittal plane displacement
SPHI	Sine(PHI)
SPSI	Sine(PSI)
*SSP	Scientific subroutine package
TO	Toe off

TX	X component of torque
TY	Y component of torque
TZ	Z component of torque
VAPC	Veterans Administration Prosthetics Center

CHAPTER ONE

INTRODUCTION

This study involved human subjects who have lost a portion of their shank and adjoining foot due to amputation. These subjects, referred to as below knee (B/K) amputees, have had an artificial leg, or prosthesis, constructed and fitted to allow ambulation. The placement of each artificial foot became a function of many considerations. Such a set of considerations might have included gait appearance, minimum energy expenditure, minimum stump normal or shear stresses, or minimum stump strain energy. This set of considerations was manipulated by the six degrees of freedom used to initially position the artificial foot with respect to the stump. Optimum prosthesis alignment was then a minimization problem over the set of considerations.

This minimization procedure was done by a trained prosthetist each time the amputee was fitted for a prosthesis. However, the prosthetist could not always put exact numbers on certain considerations he used to make his final judgement.

The energy expended during ambulation, as well as the minimum of this energy expenditure, formed the topic of this dissertation. In this manner, the engineer attempted to assist the prosthetist in his efforts by application of

modeling, mathematics, and experimentation to solve for the minimum energy solution which the prosthetist had knowledge of but could not analytically depict.

Hypotheses

1. Cumulative energy expenditure, as defined by the classical definition of work, is a minimum at optimum alignment of the B/K prosthesis. This was best visualized via an energy expenditure surface. Figure 1.1 depicts the general form of the surface as a function of the alignment adjustment and time.

2. If the subjects are left unconstrained and allowed to walk at a comfortable speed, this speed would not be a function of alignment angle.

3. This ability to perturb the energy expenditure of the system without changing its velocity of forward motion means that the system will minimize its response to an alignment change by retaining its motion along the optimum velocity.

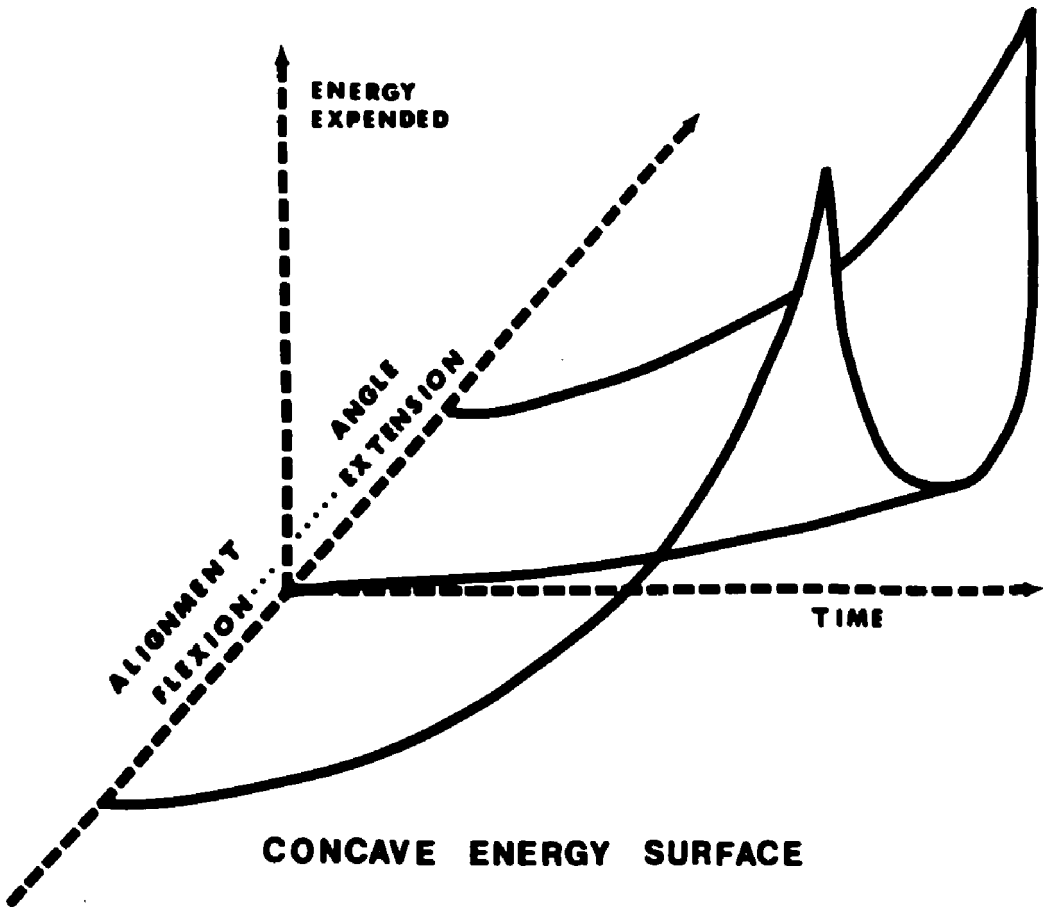


Figure 1.1.--Energy Surface

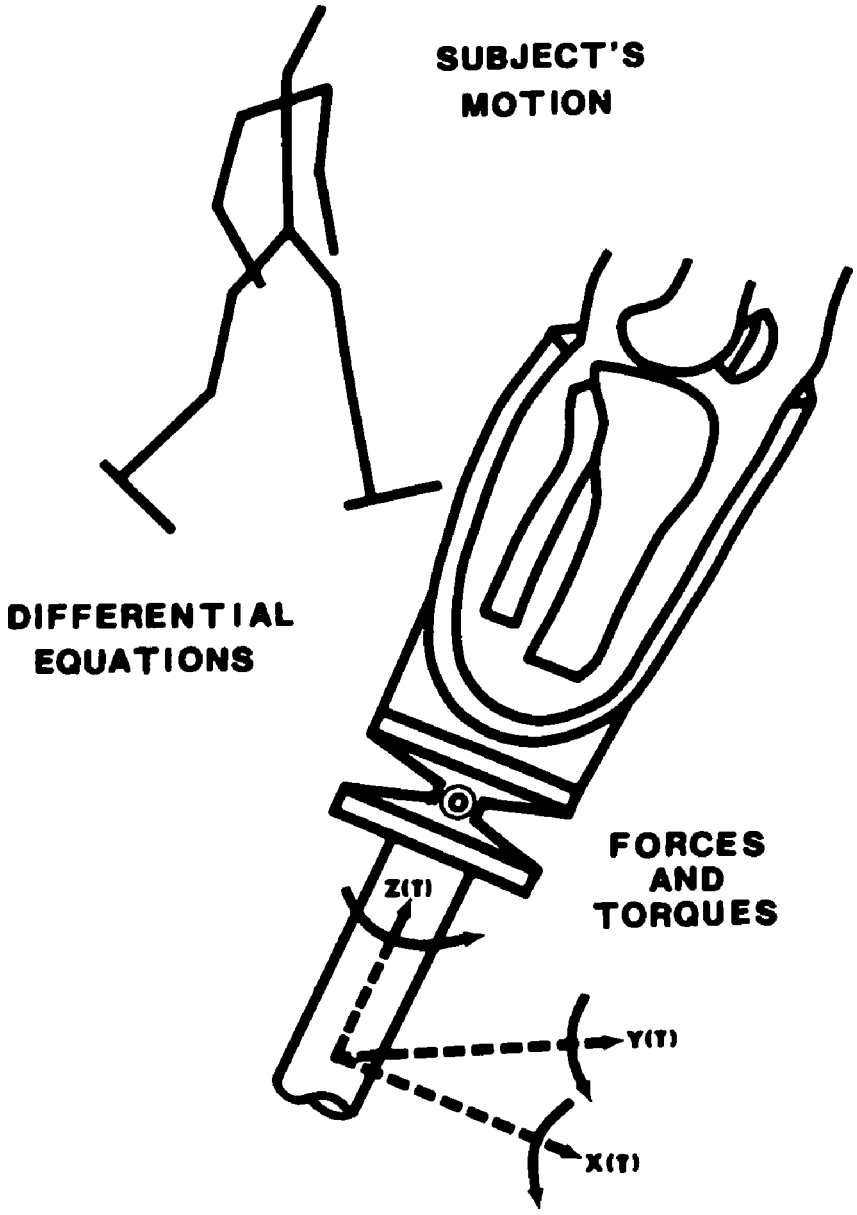


Figure 1.2.--The Theoretical Model

Modeling Process

A model of the amputee-prosthesis system was developed to calculate energy expenditure during ambulation, Figure 1.2. This model, labeled the differential equations of motion, related each subjects's motion to the forces and torques exerted.

The formulation of the mathematical model from the physical problem was a complicated procedure, Figure 1.3. Starting from the given physical situation, the amputee-prosthesis system, approximations were made. This generated an ideal physical model from which the mathematical model was derived. Historically, the model was derived before the experiment. This enabled identification of required facets of the experiment before it took place.

Not shown in Figure 1.3 were the many iterations between initial modeling efforts and experimental data gathered by other researchers. No conclusions were asserted about these comparisons. They were made simply to develop and strengthen this author's modeling skills.

The parallel paths of modeling and experiment were not totally separable. The model's theoretical predictions were used to help formulate the design of the experiment. In particular, the predicted pylon loads were used to select the proper B/K alignment unit and to select and mount appropriate strain gage transducers on the aluminum pylon.

The purposes of the experiment were two fold. Data were required for establishing the degree of applicability of the model and for the application of the model in the energy expenditure study.

Specific mathematical techniques were used to go from each amputee's observed motion to the theoretical forces and torques exerted on his prosthesis. Since the prosthesis was partly an instrumented load cell, it was possible to compare theoretical solutions to physical reality as observed in the experiment. These comparisons tested the validity of the assumptions used in making the physical approximations.

The modeling done in this dissertation was not confused with the actual physical system. The two concepts maintained their distinctions. However, statistical comparisons between the analytical predictions and empirical observations established that the degree of applicability of the physical approximations and the mathematical model was sufficient to allow their use in explaining the behavior of the physical system involved.

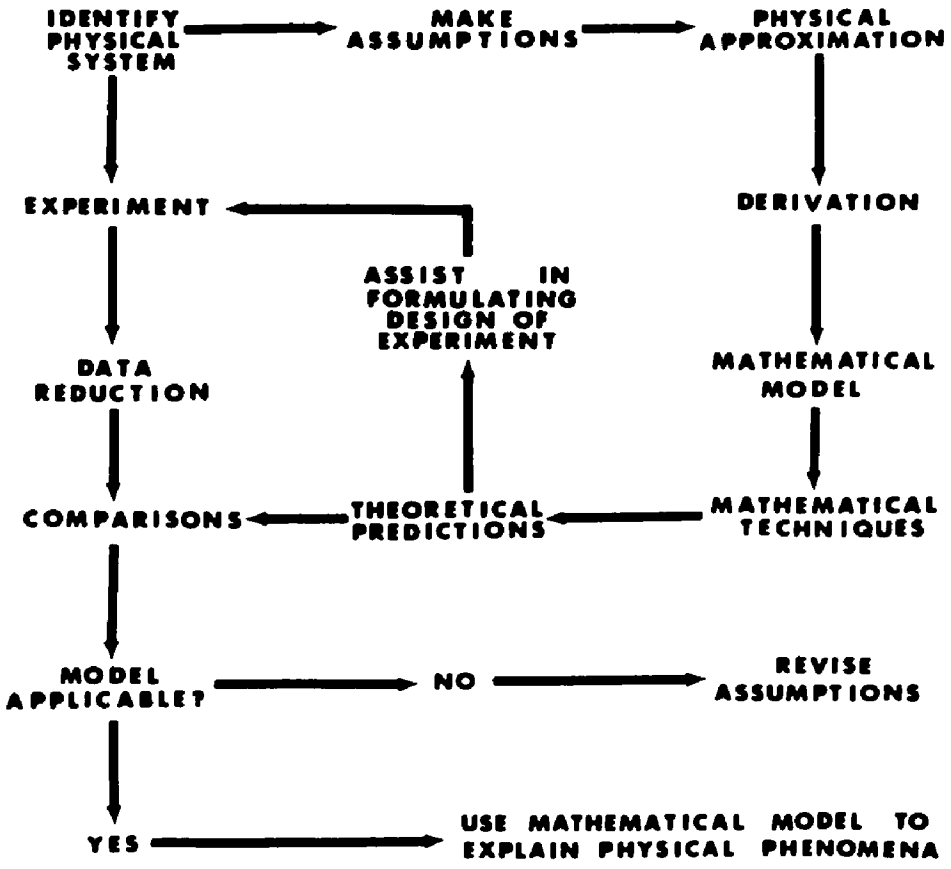


Figure 1.3.--Modeling Process

CHAPTER TWO

PHYSICAL SYSTEM

Subject Description

The first (1) subject was a twenty-six year old graduate student in the Department of Mechanical Engineering of the University of Michigan. Unlike the later subjects, he was not an amputee. His function in the research was to help design the photographic scheme described in Chapter Four, debug it before actual amputees were used, and to help provide a basis for comparison between amputees and non-amputees.

Two active, healthy, unilateral below knee amputees were then involved in this study. Elderly or diseased amputees, as well as patients recovering from their amputation, were excluded from consideration. These subjects were selected from the set of those desiring refitting of their below knee prosthesis. In this manner, the amputee subject population was restricted to not only active, healthy amputees, but to experienced walkers whose stumps had healed.

Approval of this research was obtained from the Human Use Committee, The University of Michigan Medical Center, before this selection process began.

The second (2) subject was a fifty-eight year old skin diver. He was tall, slender, and in quite good

physical condition. His B/K amputation was on the right side. The surgery was performed due to circulatory problems. Despite the reason for the surgery, his stump had healed well with no stump swelling problems.

The third (3) subject was a twenty-seven year old, tall, heavily built individual. He lost his left leg, below the knee, in a hit-and-run motorcycle accident. The surgery left him with an extremely short stump and no fibula at all. Extensive scar tissue existed on the lateral side of the stump, Figure 2.1.

Both subjects were custom fitted with a B/K prosthesis, Figure 2.2, in which some overall load bearing was allowed. This tended to favorably stimulate the distal end of the stump.

The device used to directly support the stump was the stump socket. It was constructed by a three-part molding process. The first mold is an impression (negative) of the actual stump. It is made from plaster gauze wrap. Next, a plaster positive of the stump is formed from the initial impression or negative. Finally, a laminar composite of two kinds of materials is formed around the plaster positive. In particular, two layers of 1 ounce dacron felt and two layers of nylon stocking net impregnated with polyester resin are usually used. It is this final plastic negative which is used to support the stump.

Each B/K prosthesis had a rubber sleeve suspension

and a gel liner, Figure 2.2. The purpose of the gel liner was to distribute the shear forces exerted on the stump by the plastic stump socket. This gel was silicon sheet reinforced with a nylon mesh and encased with horsehide to prevent its deterioration during gait repetitive loading. A thin sock was used to help insert the stump into the finished gel liner.

The rubber sleeve, according to Chino, et al. (1975), and Grevsten and Eriksson (1974), was used to minimize stump-socket displacements, thereby increasing patient comfort. These displacements are generated when the effects of gravity and centrifugal force tend to pull the prosthesis away from the stump during the swing phase. Of course, reverse displacement occurs when the ground reactions tend to force the prosthesis against the stump during the stance phase. The use of the rubber sleeve is that of a suction seal. Between the seal and friction effects, the amplitude of this cyclic piston action is reduced over that of the previously used cuff suspension.

Besides consideration for the patient, the rubber sleeve was used to reduce large displacement, small strain conditions for a possible finite element analysis.

The lower portion of each B/K prosthesis consisted of a mass-produced artificial foot. Its particular characteristic was that there was no time dependent articulation at the ankle. To make up for this rigid connection, the heel and the toe of this foot were

compliant. Such characteristics lended themselves to the label of solid ankle, cushioned heel (S.A.C.H.) foot.

The region between the stump socket and the artificial foot consisted of a wooden base supporting two layers of nylon stocking net impregnated with polyester resin. This connection was totally rigid except for one degree of freedom, allowing unlimited toe-in or toe-out of the S.A.C.H. foot.

Because of this limited articulation ability, the standard B/K prosthesis, as created by prosthetists at University Hospital, is herein considered as being rigid in nature.

It should be noted that the basically irreversible positioning of the artificial foot and the stump socket was determined by a conventional fitting process. This fitting process used the stump socket, S.A.C.H. foot, and a six degree of freedom alignment unit.

These articulated units, typically the Berkeley Adjustable Leg manufactured by Hosmer/Dorrance, were never included with the final product. Instead, they were constantly being reused. Once the prosthetists visually determined the optimum configuration (alignment) of the prosthesis, the appropriate information was transferred to an alignment jig. From this jig, the final (rigid) prosthesis was constructed.

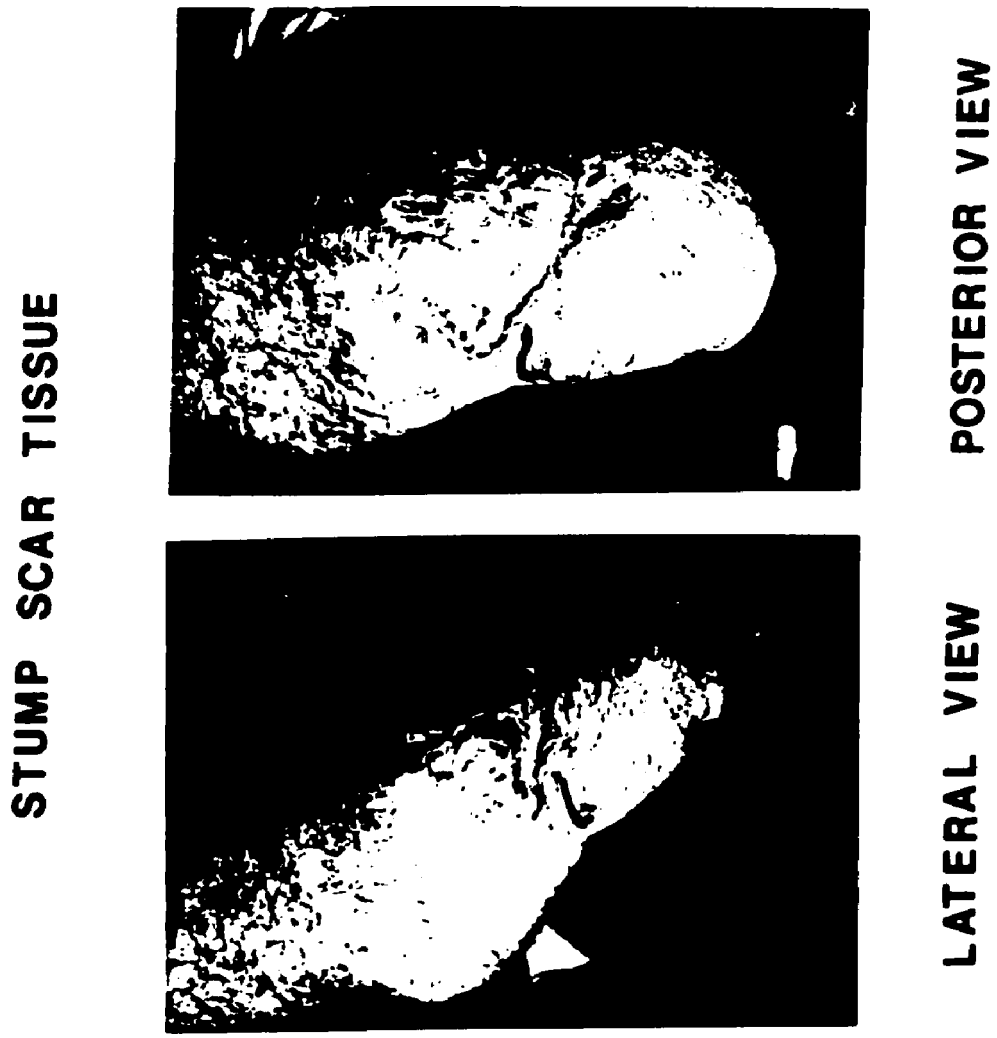


Figure 2.1.--Subject 3: Extensive Stump Scar Tissue

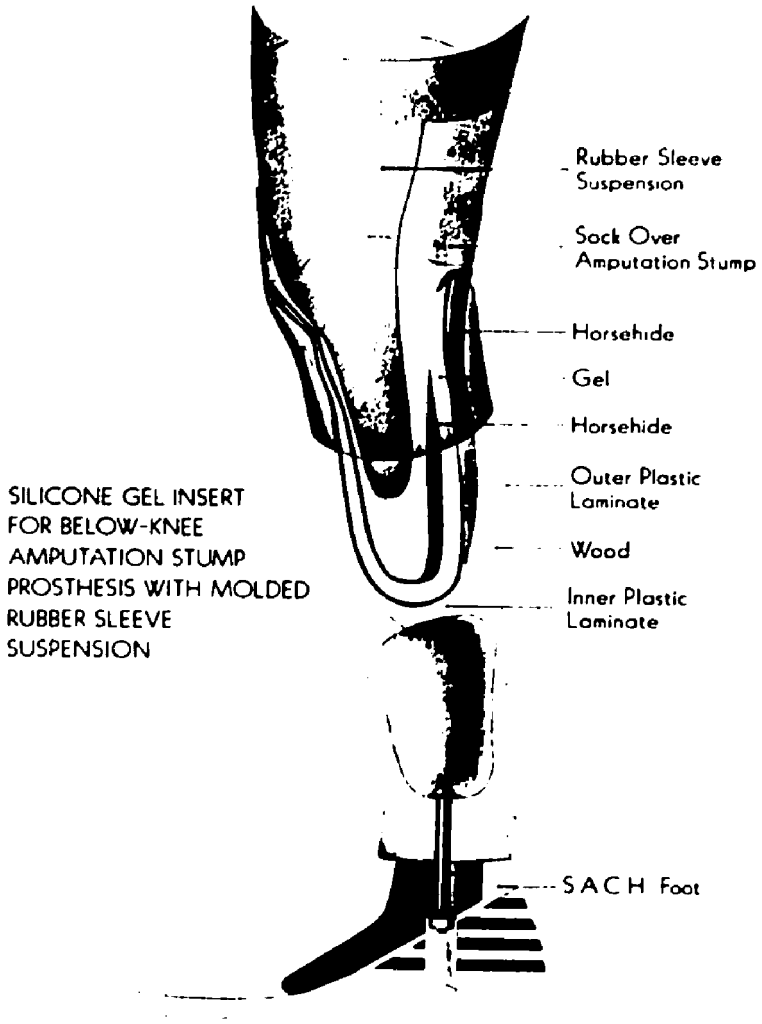


Figure 2.2.--Total Contact Below Knee Prosthesis

Anthropometric Measurements

Physical characteristics of the subjects were documented in order that as much direct information about their physical systems as possible would be known. In the following chapter, approximations of these data were used to derive specific quantities for the differential equations of motion. In order to preserve the distinction between the existing physical reality and the eventual physical approximations, the actual anthropometric measurements have been presented below.

Linear measurements were made with Swiss GPM calipers to the nearest millimeter. The purpose of these linear measurements was to obtain actual data to be compared to the predictions of the statistically derived segmental length parameters of Chapter Three. These comparisons are discussed in tabular form in that chapter.

1. Total height was measured from the floor to the top of the head. Subjects 2 and 3 wore standard one-inch heels, Subject 1 was barefoot for this and the next six measurements.

   Subject 1: 1.706 m

   Subject 2: 1.884 m

   Subject 3: 1.854 m

2. Floor to hollow of the sternum manubrium

   Subject 1: 1.382 m

   Subject 2: 1.525 m

   Subject 3: 1.485 m

3. Floor to the crest of the ilium

Subject 1:		right 0.984 m
Subject 2:	left 1.120 m	right 1.120 m
Subject 3:	left 1.130 m	right 1.127 m

4. Floor to greater trochanter of the femur

Subject 1:		right 0.860 m
Subject 2:	left 1.020 m	right 1.020 m
Subject 3:	left 0.909 m	right 0.912 m

5. Floor to crotch

Subject 1: 0.803 m

Subject 2: 0.887 m

Subject 3: 0.850 m

6. Floor to the medial condyle

Subject 1:		right 0.424 m
Subject 2:	left 0.552 m	right
Subject 3:	left	right 0.507 m

7. Floor to lateral malleolus

Subject 1:		right 0.082 m
Subject 2:	left 0.112 m	right SACH foot
Subject 3:	left SACH foot	right 0.010 m

8. Acromion to olecranon process

Subject 1:		right 0.330 m
Subject 2:	left 0.380 m	right 0.380 m
Subject 3:	left 0.332 m	right 0.343 m

9. Olecranon process to middle finger tip

Subject 1:		right 0.414 m
Subject 2:	left 0.485 m	right 0.480 m

Subject 3: left 0.451 m right 0.451 m

10. Scaphoid to middle finger tip

Subject 1: right 0.176 m

Subject 2: left 0.220 m right 0.222 m

Subject 3: left right

Circumferential measurements were made with a steel tape measure applied with sufficient pressure to make firm contact with the skin, but not to cause the skin to be indented. Measurements were made to the nearest millimeter.

At the time these measurements were taken, it was hoped they would lead to the calculation of body segmental volumes. Once the segmental volumes were known, it was hoped that the product of volume times the appropriate segmental density would yield the mass of each segment.

After considerable thought, it was decided that determination of segmental volumes by water immersion was superior to this method. Hence, the numerical calculations were not carried out in favor of directly applying statistical mass parameters obtained by water immersion.

For completeness, the circumferential data are still presented below.

11. Head at brow

Subject 1: 0.575 m

Subject 2: 0.580 m

Subject 3: 0.580 m

12. Neck - just below the "Adam's apple"

Subject 1: 0.368 m

Subject 2: 0.430 m

Subject 3: 0.420 m

13. Chest - at arm pit level

Subject 1: 0.890 m

Subject 2: 1.010 m

Subject 3: 1.100 m

14. Waist - at umbilicus

Subject 1: 0.720 m

Subject 2: 0.880 m

Subject 3: 1.040 m

15. Buttock - at greater trochanter

Subject 1: 0.850 m

Subject 2: 0.990 m

Subject 3: 1.090 m

16. Thigh at crotch level

Subject 1:		right 0.500 m
Subject 2:	left 0.600 m	right 0.600 m
Subject 3:	left 0.650 m	right 0.690 m

17. Knee - at medial condyle

Subject 1:		right 0.382 m
Subject 2:	left 0.400 m	right 0.400 m
Subject 3:	left 0.390 m	right 0.400 m

18. Calf or stump - at greatest circumferential extent

Subject 1:		right 0.341 m
Subject 2:	left 0.355 m	right
Subject 3:	left 0.310 m	right 0.420 m

19. Bicep - flexed

Subject 1: left 0.267 m

Subject 2: left 0.290 m right 0.330 m

Subject 3: left 0.385 m right 0.450 m

20. Forearm - flexed and at greatest circumferential extent

Subject 1: left 0.244 m

Subject 2: left 0.280 m right 0.310 m

Subject 3: left 0.315 m right 0.330 m

21. Subject's mass in kilograms. Subjects 2 and 3 were weighed with shoes and prosthesis, Subject 1 was barefoot.

Subject 1: 58.3 kg

Subject 2: 79.4 kg

Subject 3: 100.4 kg

Skinfold measurements were done on four regions of the body to the nearest millimeter. The purpose behind these measurements was to assist determination of whole-body density. The Pascale formulas, as listed in Contini (1969), were applied. Density was expressed in kilograms/litre. The triceps value used in formula A was the mean skinfold of the two triceps. The scapular value used in formula B was also the mean of the two scapular measurements. Both were expressed in mm. These formulas are:

$$\text{Density(A)} = 1.0923 - 0.00202*(\text{Triceps})$$

$$\text{Density(B)} = 1.0896 - 0.00179*(\text{Scapula})$$

It was noted that not all of the skinfold

measurements presented below were used in the Pascale formulas. However, at the encouragement of Dr. Merle L. Foss, The University of Michigan, the additional measurements were taken for completeness.

Each body skinfold was lifted firmly between the thumb and adjacent fingers. Following the natural contour of the skinfold, the calipers were placed around the withdrawn tissue and gradually released. The calipers were allowed to compress the skinfold about 10 mm from the point where the skinfold was lifted.

The instrument used was a Lange Skinfold Caliper. It was manufactured by Cambridge Scientific Industries in Cambridge, Maryland. The instrument was carefully calibrated with 5 mm, 10 mm, and 15 mm gage blocks before any data was taken.

The flesh was viscoelastic. This time dependent property caused the skinfold caliper reading to drop from its initial value to a lower steady state value. In order to standardize all measurements, readings were taken after a ten second wait. This was sufficient to allow the flesh to reach its steady state thickness under the caliper compression.

22. The skinfold measurement at the triceps was taken at the midpoint between the acromion and the olecranon process. The arm was hanging freely at the subject's side. The skinfold was lifted on the back of the arm parallel to the long axis of the arm about 10 mm above the site.

Subject 1: right 4 mm

Subject 2: left 12 mm right 12 mm

Subject 3: left 24 mm right 18 mm

23. The skinfold measurement at the scapula was located 10 mm below the inferior tip of the scapula, and the fold was lifted parallel to the long axis of the body.

Subject 1: right 8 mm

Subject 2: left 11 mm right 11 mm

Subject 3: left 27 mm right 28 mm

24. The suprailiac skinfold was lifted just above the crest of the ilium. The fold was lifted to follow the natural diagonal at this point, dorsally upward.

Subject 1: right 10 mm

Subject 2: left 16 mm right 16 mm

Subject 3: left 38 mm right 42 mm

25. The umbilical skinfold was lifted about 10 mm to the right of the umbilicus and parallel to the long axis of the body.

Subject 2: 8 mm

Subject 3: 30 mm

26. Frontal and sagittal stump x-rays of Subjects 2 and 3 were taken. The subjects were in a standing, load bearing position. Figures 2.3 through 2.6 display the results. The reason for these x-rays was to provide a logical basis for positioning the pressure transducers discussed in Chapter Four and to provide a jumping off point for the construction of the stiffness matrix required by the finite

element analysis mentioned in Chapter Six.

27. The S.A.C.H. feet used by the amputees had characteristics which are documented in Table 2.1. These data are referenced in Chapter Three. It should be noted that the center of mass of each S.A.C.H. foot-shoe-footswitch system was determined by the balance method. The center of mass was then located from the bolt hole in the artificial foot in the direction of the toe of the foot.

28. Subject 2 wore a number 2 rubber sleeve; Subject 3, a number 3 rubber sleeve.

29. Subject 2 had a stump volume of 1.280 litres; Subject 3, 1.500 litres. These data were determined by filling the stump socket with water. The stump volume was needed to calculate its mass. By using the whole-body density predictions of the Pascale formulas, Contini's (1969) graphs for relating segmental densities to the whole body density, and the stump volume, it was possible to calculate a stump mass. Stump center of mass was determined by balancing the water filled socket on its side. Of course, the open end of the socket was momentarily sealed.

TABLE 2.1

S.A.C.H. FOOT CHARACTERISTICS

	Subject 2	Subject 3
Manufacturer	Wagner	Wagner
Size	11.5-12R	10.5-11L
Toe stiffness	27	29
Heel stiffness	Regular	Firm
SACH mass	0.466 kg	0.407 kg
Shoe mass	0.550 kg	0.475 kg
Footswitch mass	0.046 kg	0.046 kg
Segment total mass	1.072 kg	0.928 kg
Center of mass	0.070 m	0.069 m
Chapter Three Segment #	17	13

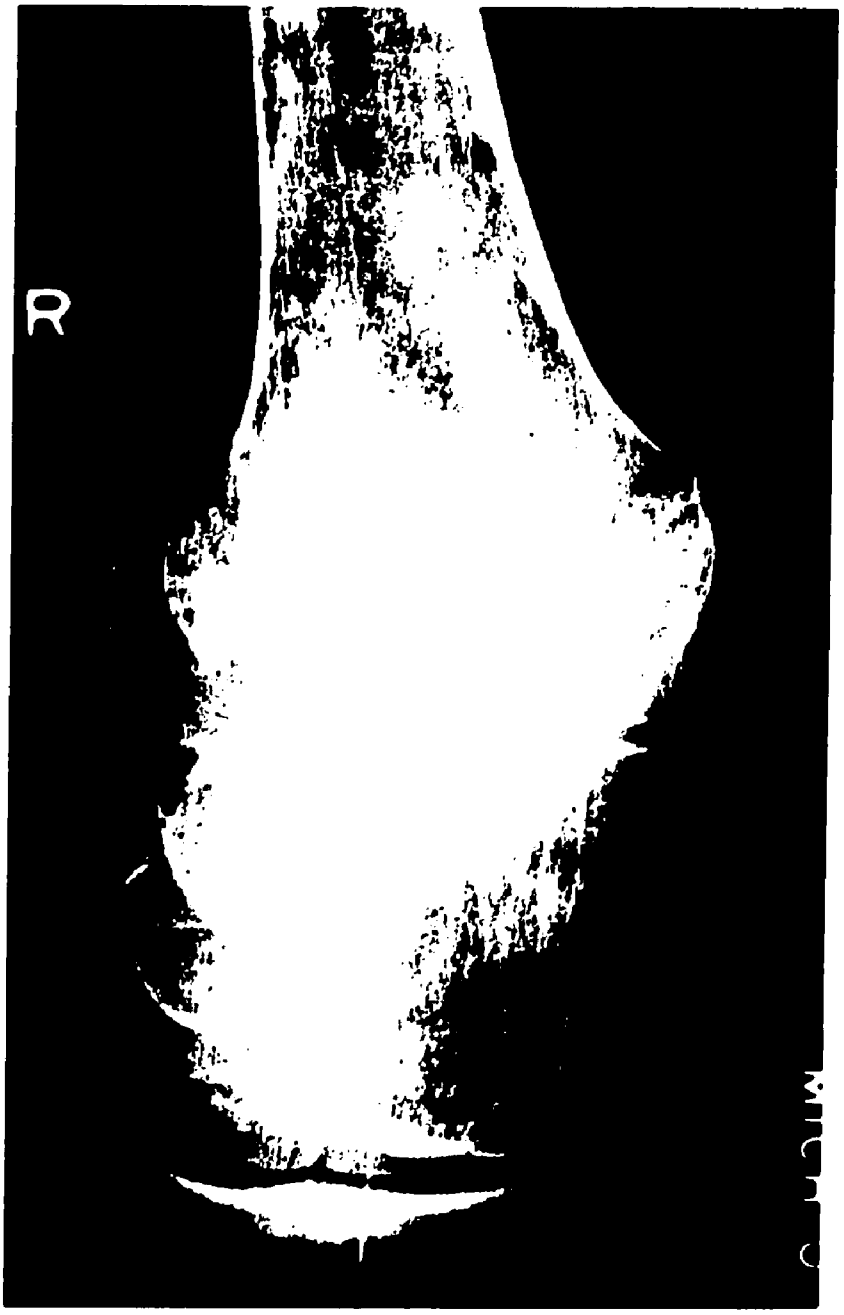


Figure 2.3.--Subject 2: Stump Frontal X-ray



Figure 2.4.--Subject 2: Stump Sagittal X-ray



Figure 2.5.--Subject 3: Stump Frontal X-ray



Figure 2.6.--Subject 3: Stump Sagittal X-ray

CHAPTER THREE

MATHEMATICAL MODEL

Physical Approximations

The three-dimensional mathematical model was composed of seventeen rigid segments, Figure 3.1. These segments were the lumbar; thorax; head; left shoulder, arm, and forearm; right shoulder, arm, and forearm; left ilium, thigh, shank, and foot; and the right ilium, thigh, shank, and foot. These segments were connected by thirteen frictionless ball joints, Figure 3.2.

This figure does not show articulation between the lumbar and the thorax. None was put into the model because the author felt he could not obtain reliable data thereof. Hence, the spine was treated as a rigid member for the purposes of this analysis. However, this pair of collinear vectors was left in the model because they could be articulated in the future.

In reality, each subject's body was not entirely articulated by ball joints. For instance, the human knee has a characteristic path of its instantaneous center of rotation, as documented by Blacharski, Somerset, and Murray (1975) and by Smidt (1973). The ankle, as analyzed by Isman and Inman (1968), is actually a two joint mechanism. The shoulder complex, as documented by Shoup (1976) is also quite complicated. However, restricting the mathematics to a real-life motion analysis made the thirteen ball joints

act like their real life counterparts. This happened because the model's joints were constrained by the filmed motion of each subject to act in the same manner as the subject's.

Different physical approximations were made by other researchers. Beckett and Chang (1968), Cappozzo (1976), Chaffin (1969), Chao(1973), Chow (1971), Ghosh and Boykin (1974), and Gubina (1974) used planar models. In particular, these models were confined to motion in the sagittal plane. The closest similarities were found in the three-dimensional work of Huston and Passerello (1971).

In the model used in this research, each segment possessed length, mass, and a center of mass. A lumped mass approximation, Figure 3.5, was assumed. This meant all mass in a segment was concentrated at its center of mass.

Huston and Passerello (1971) had a distributed mass approximation. However, human body segments do not entirely rotate as rigid bodies. In particular, they respond viscoelastically to rotations about the transverse axis. Hence, it was felt better to underestimate the small inertial contribution due to transverse rotations with a lumped mass model than to overestimate the inertial contribution with a rigidly-distributed mass model.

The values of the segment length, mass, and center of mass were required coefficients in the differential equations of motion. Contini (1970) and Dempster (1967)

were sources of this information. Actual subject segment lengths documented in Chapter Two agreed closely with Contini's and Dempster's predictions, Table 3.8. Hence, it was concluded that their mass and center of mass predictions were applicable.

The one exception to the application of these predictions was with respect to the abdominopelvic region, consisting of the left ilium, right ilium, and the lumbar. Dempster (1967) gave global mass and center of mass parameters for the abdominopelvic region. However, data for the individual portions of the region were not available. These data were obtained by dividing the abdominopelvic mass into 3 portions, 50% for the lumbar and 25% for each ilium. Based on the observation that most body segments have their center of mass close to the segment midpoint, the center of mass of each ilium was assigned as 50% of the lumbosacral-hip distance. The lumbar center of mass was then assigned as the distance necessary to make the global 3-segment center of mass fall upon Dempster's prediction.

Tables 3.1 through 3.4 have been listed to enumerate how those predictions apply to the differential equations of motion. Included in these tables are auxiliary vectors needed to "round the rough edges" of the seventeen segment model. These vectors specify the location of the heels, prosthetic alignment unit, pylon strain gages, and the perpendicular distance between the lumbosacral articulation

and a line passing through both hips.

It was assumed that the heel vector was collinear with the respective foot vector.

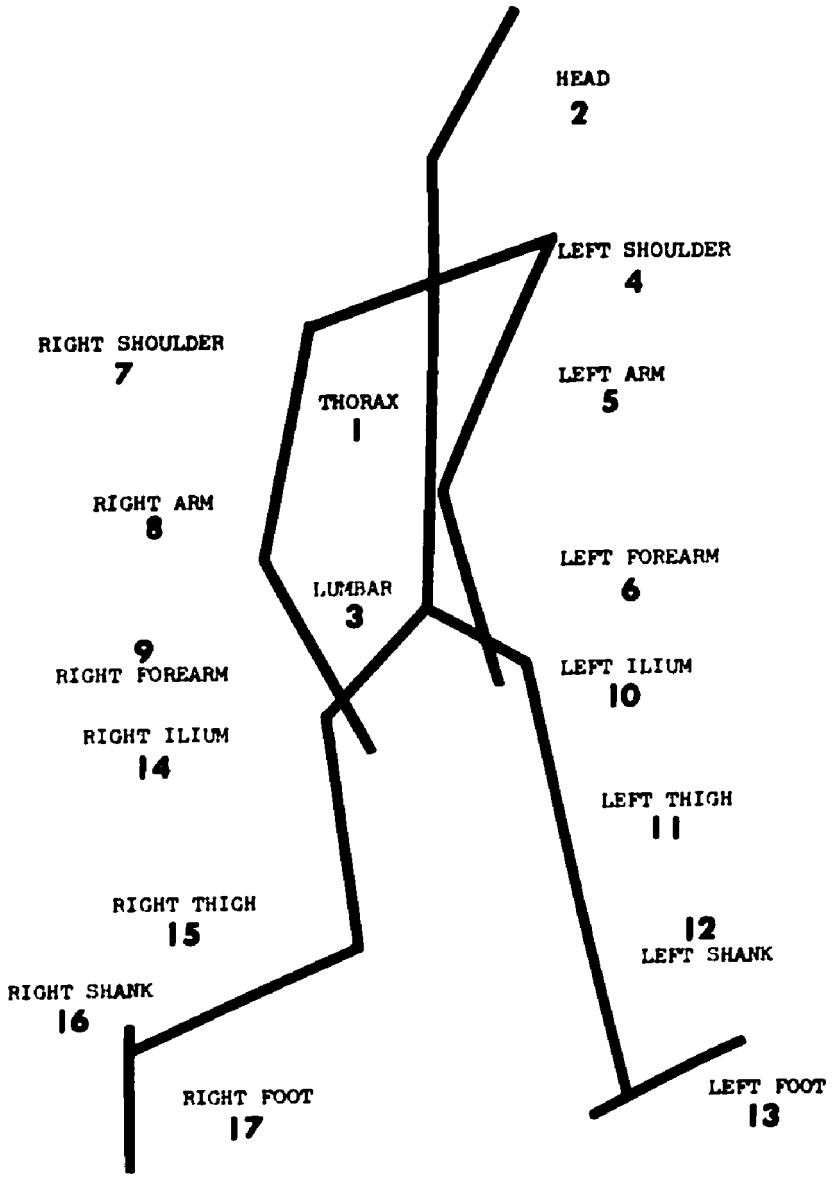


Figure 3.1.--Three Dimensional Body

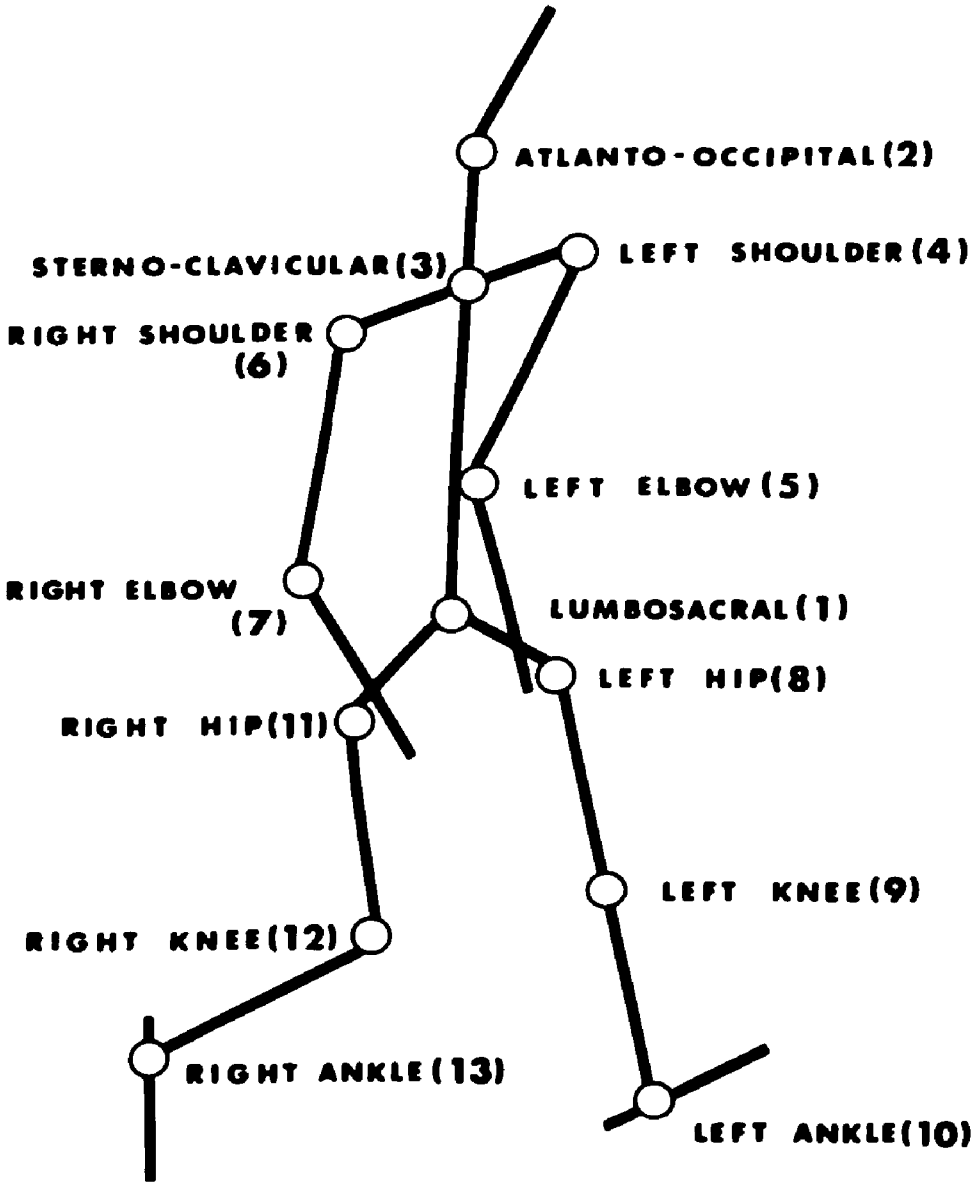


Figure 3.2.--Thirteen Articulated Interfaces

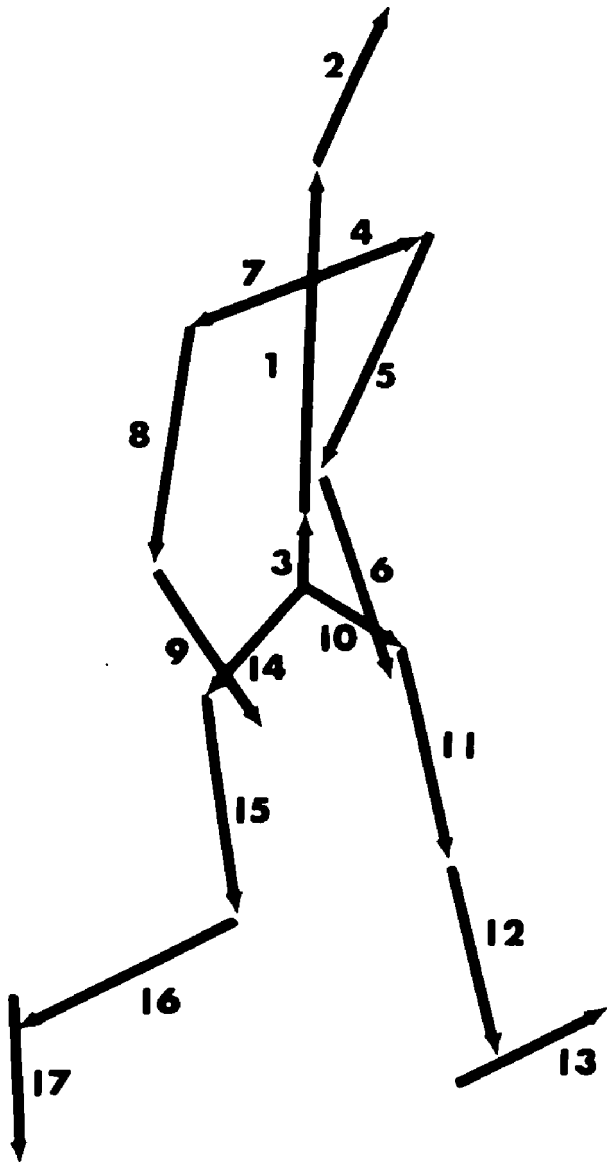


Figure 3.3.--Seventeen Segment Vectors

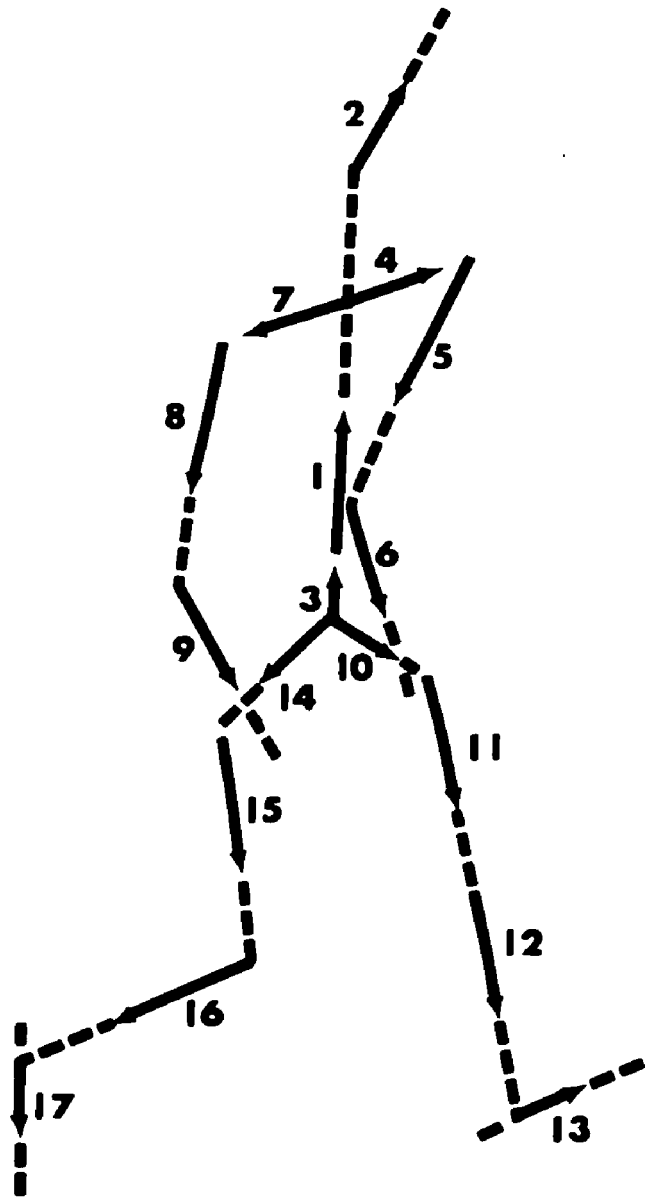


Figure 3.4.--Segment Center of Mass Vectors

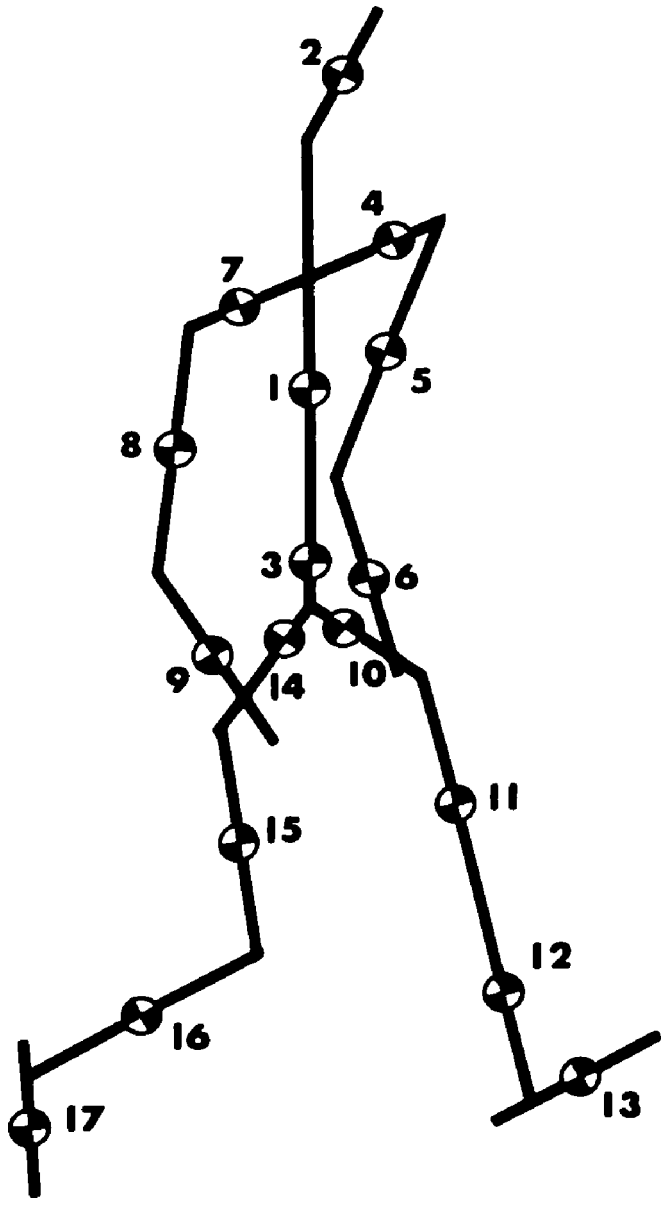


Figure 3.5.--Lumped Mass Approximation

TABLE 3.1

SEGMENT NUMBERING

Segment Number	Segment
1	Thorax and neck
2	Head
3	Lumbar
4	Left shoulder
5	Left arm
6	Left forearm and hand
7	Right shoulder
8	Right arm
9	Right forearm and hand
10	Left Ilium
11	Left thigh
12	Left shank
13	Left foot
14	Right Ilium
15	Right thigh
16	Right shank
17	Right foot
18	Left heel
19	Right heel
20	Left VAPC alignment unit
21	Right VAPC alignment unit
22	Left pylon strain gages
23	Right pylon strain gages
24	Lumbosacral to hip line

TABLE 3.2

THEORETICAL SEGMENT LENGTHS

Segment	% Barefoot height
1	26.8%
2	13.0%
3	21.6%
4	12.9%
5	18.6%
6	25.4%
7	12.9%
8	18.6%
9	25.4%
10	12.0%
11	24.5%
12	24.6%
13	10.9%
14	12.0%
15	24.5%
16	24.6%
17	10.9%
18	2.73%
19	2.73%
24	7.20%

TABLE 3.3

THEORETICAL SEGMENT CENTER OF MASS

LOCATED FROM PROXIMAL END OF SEGMENT

Segment	% Segment length
1	43.1%
2	50.0%
3	16.7%
4	84.0%
5	46.1%
6	42.0%
7	84.0%
8	46.1%
9	42.0%
10	50.0%
11	42.7%
12	40.4%
13	24.9%
14	50.0%
15	42.7%
16	40.4%
17	24.9%

TABLE 3.4

THEORETICAL SEGMENT MASSES

Segment	% Total nude mass
1	11.4%
2	8.20%
3	13.7%
4	5.36%
5	2.70%
6	2.18%
7	5.35%
8	2.70%
9	2.16%
10	6.86%
11	10.1%
12	4.68%
13	1.46%
14	6.86%
15	10.1%
16	4.67%
17	1.46%

It should be noted that the preceding four tables applied to a nude, barefoot, non-amputee. The derivation of the final anthropometric coefficients for the differential equations of motion were corrected for the fact that, during the experiment, the subjects wore shoes of known mass and these shoes had standard one-inch (25 mm) heels. Further corrections included the length, mass, and center of mass of the stump-socket-pylon system and the S.A.C.H. foot instead of a shank and real foot for Subjects 2 and 3. Table 2.1 listed the S.A.C.H. foot corrections explicitly.

Tables 3.5, 3.6, and 3.7, the final anthropometric coefficients, constitute a tabulated numerical summary of the subjects' physical being. It should be noted that the length of segment 16 on Subject 2 and segment 12 on Subject 3 were the knee joint to alignment unit distances. The alignment unit was treated as a pseudo-14th joint for modeling purposes. The prosthesis was specified by segment 20 if the amputee was a left B/K; segment 21, a right B/K. This distance was from the alignment unit to the ankle. The value zero was inserted if the leg was "normal." This pseudo-joint was frozen in place during each test and articulated as required to simulate each angular alignment perturbation of Chapter Four.

TABLE 3.5

SUBJECT 1: FINAL ANTHROPOMETRIC COEFFICIENTS

Segment	Length	C.M.	Mass
1	0.457 m	0.197 m	6.65 kg
2	0.222 m	0.111 m	4.78 kg
3	0.368 m	0.061 m	7.99 kg
4	0.220 m	0.185 m	3.12 kg
5	0.317 m	0.146 m	1.57 kg
6	0.433 m	0.182 m	1.27 kg
7	0.220 m	0.185 m	3.12 kg
8	0.317 m	0.146 m	1.57 kg
9	0.433 m	0.182 m	1.27 kg
10	0.205 m	0.102 m	4.00 kg
11	0.418 m	0.178 m	5.89 kg
12	0.445 m	0.170 m	2.73 kg
13	0.186 m	0.046 m	1.00 kg
14	0.205 m	0.102 m	4.00 kg
15	0.418 m	0.178 m	5.89 kg
16	0.445 m	0.170 m	2.73 kg
17	0.186 m	0.046 m	1.00 kg
18	0.047 m		
19	0.047 m		
20	0.000 m		
21	0.000 m		
22	0.000 m		
23	0.000 m		
24	0.123 m		

TABLE 3.6

SUBJECT 2: FINAL ANTHROPOMETRIC COEFFICIENTS

Segment	Length	C.M.	Mass
1	0.498 m	0.214 m	9.08 kg
2	0.242 m	0.121 m	6.52 kg
3	0.402 m	0.067 m	10.9 kg
4	0.240 m	0.202 m	4.25 kg
5	0.380 m	0.175 m	2.15 kg
6	0.480 m	0.202 m	1.73 kg
7	0.240 m	0.202 m	4.25 kg
8	0.380 m	0.175 m	2.15 kg
9	0.480 m	0.202 m	1.73 kg
10	0.223 m	0.112 m	5.45 kg
11	0.450 m	0.192 m	8.01 kg
12	0.530 m	0.178 m	3.71 kg
13	0.203 m	0.070 m	1.76 kg
14	0.223 m	0.112 m	5.45 kg
15	0.450 m	0.192 m	8.01 kg
16	0.220 m	0.125 m	2.90 kg
17	0.203 m	0.070 m	1.07 kg
18	0.051 m		
19	0.051 m		
20	0.000 m		
21	0.310 m		
22	0.000 m		
23	0.205 m		
24	0.134 m		

TABLE 3.7

SUBJECT 3: FINAL ANTHROPOMETRIC COEFFICIENTS

Segment	Length	C.M.	Mass
1	0.490 m	0.210 m	11.4 kg
2	0.238 m	0.119 m	8.20 kg
3	0.395 m	0.066 m	13.7 kg
4	0.236 m	0.198 m	5.34 kg
5	0.340 m	0.157 m	2.70 kg
6	0.465 m	0.195 m	2.18 kg
7	0.236 m	0.198 m	5.34 kg
8	0.340 m	0.157 m	2.70 kg
9	0.465 m	0.195 m	2.18 kg
10	0.220 m	0.110 m	6.85 kg
11	0.448 m	0.191 m	10.1 kg
12	0.212 m	0.212 m	3.10 kg
13	0.200 m	0.069 m	0.93 kg
14	0.220 m	0.110 m	6.85 kg
15	0.448 m	0.191 m	10.1 kg
16	0.522 m	0.182 m	4.67 kg
17	0.200 m	0.069 m	1.45 kg
18	0.050 m		
19	0.050 m		
20	0.310 m		
21	0.000 m		
22	0.205 m		
23	0.000 m		
24	0.132 m		

TABLE 3.8

SEGMENT LENGTH COMPARISONS

Subject	Segment	Theoretical	Empirical	Difference
1	5	0.317 m	0.330 m	3.9%
1	6	0.433 m	0.414 m	4.3%
1	11	0.418 m	0.436 m	4.1%
1	12	0.445 m	0.424 m	4.7%
2	5	0.380 m	0.380 m	0.0%
2	6	0.480 m	0.485 m	1.0%
2	11	0.450 m	0.468 m	3.8%
2	12	0.530 m	0.552 m	4.0%
3	5	0.340 m	0.332 m	2.4%
3	6	0.465 m	0.451 m	3.0%
3	15	0.448 m	0.405 m	9.6%
3	16	0.522 m	0.507 m	2.9%

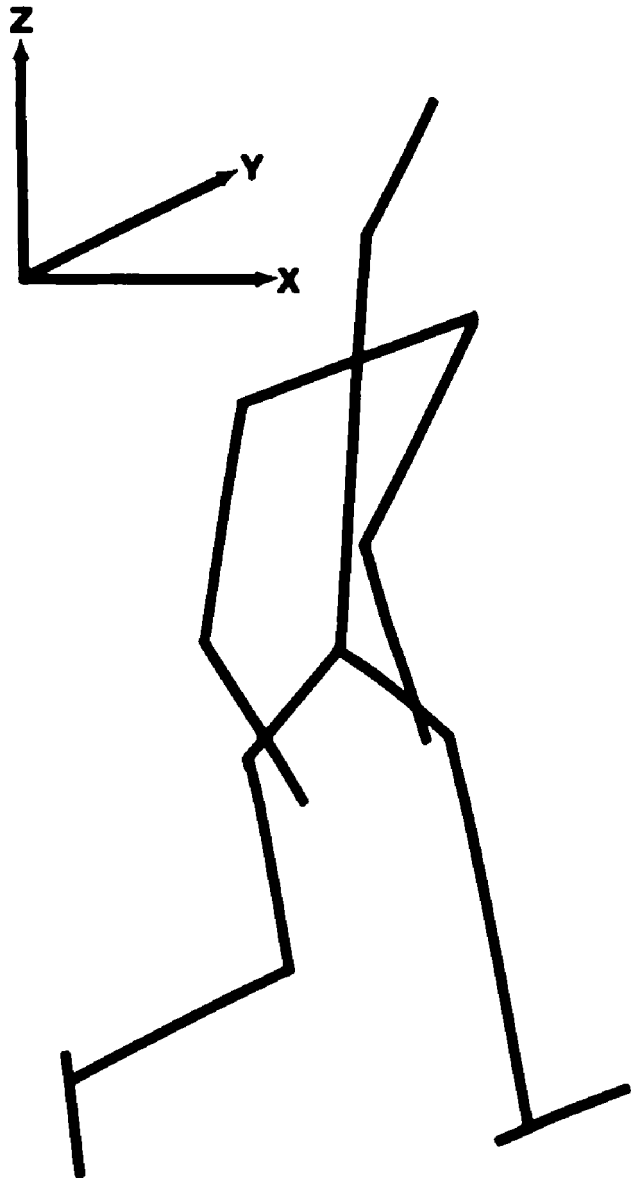


Figure 3.6.--Global Cartesian Coordinate System

Model Derivation

The first step in deriving the three-dimensional differential equations of motion was to define a global Cartesian coordinate system, Figure 3.6. The global +Z-axis direction was defined parallel to the upward vertical. The normal to the subject's sagittal plane pointing away from the sagittal plane observer defined the +Y-axis direction. Having desired a right hand coordinate system meant that the +X-axis was normal to the frontal plane and in the same direction as the left to right direction of the sagittal plane observer. The X-axis datum was defined as 9 metres away from the frontal plane observer; Y-axis datum, the sagittal plane; the Z-axis datum, the floor. Figures 4.6 and 4.7 augment this description.

The origin, as well as the orientation of this global Cartesian coordinate system, remained fixed with respect to the earth for each subject. Therefore, for all intent and purpose, it was considered an inertial reference frame.

It was from this coordinate system that the three-dimensional motion of the subjects was specified via thirty-seven time dependent coordinates. The first three coordinates were linear, the x(t), y(t), and z(t) position of the pelvis with respect to the origin of the coordinate system, Figure 3.7. This vector corresponded to the external coordinates as defined by Smith and Kane (1968). From the pelvis, the configuration of the body with respect

to the pelvis was described via two spherical angles for each of the seventeen segments, or thirty-four spherical angles in all. These angles were classified as internal coordinates by Smith and Kane (1968).

These thirty-four angles were used only to specify the orientation of the transverse axis of each segment. Rotations of a segment about its transverse axis were ignored.

The set of segmental angles was divided into two subsets. The first subset, labeled PHI, involved projections of the long axes of the segments onto the sagittal plane, Figure 3.8. The angle of the projection in the sagittal plane was measured with respect to the +Z-axis. Some researchers referred to this kind of spherical angle as an azimuth angle. The second subset, labeled PSI, consisted of the angles formed by the segments' long axes and the +Y-axis, Figure 3.9. These angles were measured with respect to the +Y-axis.

Unlike the first subset of angles, which were contained in a specific plane, the second subset was only constrained to pivot about the +Y-axis, unconfined to any fixed orientation plane. This pivoting about the +Y-axis lended itself to labeling the second subset of angles as conical. Some researchers referred to these conical angles as swing angles.

Knowing this set of linear and absolute angular motions, it was possible to mathematically define the

forces acting on each segment and their points of application as functions of time. These forces are listed below.

1. The first set of forces consisted of the weight of each segment acting through the segment's center of gravity. These gravitationally induced forces were evaluated by multiplying each segment mass (kg) by the acceleration of gravity (9.806 m/s^2). Of course, these forces were directed in the negative Z-axis direction.

2. The force necessary to accelerate the center of gravity of each segment described the elements of the second set of forces. These inertially induced forces were calculated by multiplying each segment mass (kg) by the acceleration of that center of mass with respect to the inertial reference frame (m/s^2).

   Each inertial force had three components, one directed along each axis of the inertial reference frame. This decomposition of the inertial forces allowed construction of a total segment center of mass force consisting of inertial components along the X and Y-axes and combined inertial and gravitational component along the Z-axis.

3. The ground reaction forces, consisting of a vertical and two horizontal components exerted at the center of pressure on each foot, combined to make the third set of forces. These quantities were evaluated based on the notion that the vector sum of the forces and the vector sum of the moments evaluated at the model's center of mass were zero in each component.

Most of the differential equations of motion were derived for the set of torques exerted on each of the segments. They were derived for each segment by use of the vector cross product. The cross products were functions of the forces exerted on a segment and the vectors which located their points of application with respect to the joint at the proximal end of the segment in question. By use of the concept of free body diagrams, appropriate sums of cross products were taken.

This was not the only means of derivation. The Lagrange Equations, as presented in Easthope (1964), could have been used. Similarly, virtual work, virtual power, or the Lagranges form of the D'Alembert equations, as used by Chace and Angell (1973) and Winarski (1974), were realistic possibilities. As it was, the vector cross product approach was chosen because it was easy to program the computer to derive the differential equations in addition to its solving them.

There were thirty-seven differential equations in

all, one for each coordinate used in the model. Furthermore, this set of equations had to be written separately, or piecewise, for each of the five distinct situations below.

1. Right foot on the ground; left foot, swing phase

2. Double stance with the right foot trailing

3. Left foot on the ground; right foot, swing phase

4. Double stance with the left foot trailing

5. Free flight with neither foot on the ground

The need to switch from one of the above situations to another at the proper time necessitated the use of footswitches as event markers. More information about this particular instrumentation will be discussed in Chapter Four.

The need for the fifth situation was never encountered in the experiment. However, programming the model for zero-valued contact forces during such a situation was quite trivial. Hence, this was included mostly because it was the easiest of the five situations to model.

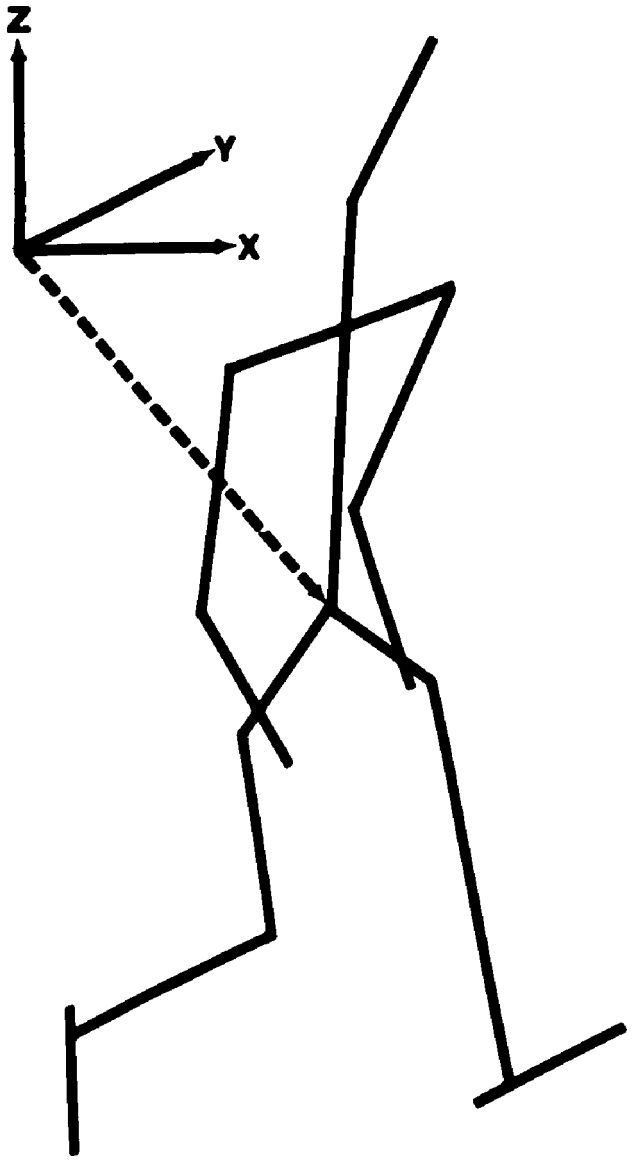


Figure 3.7.--Three Linear Coordinates

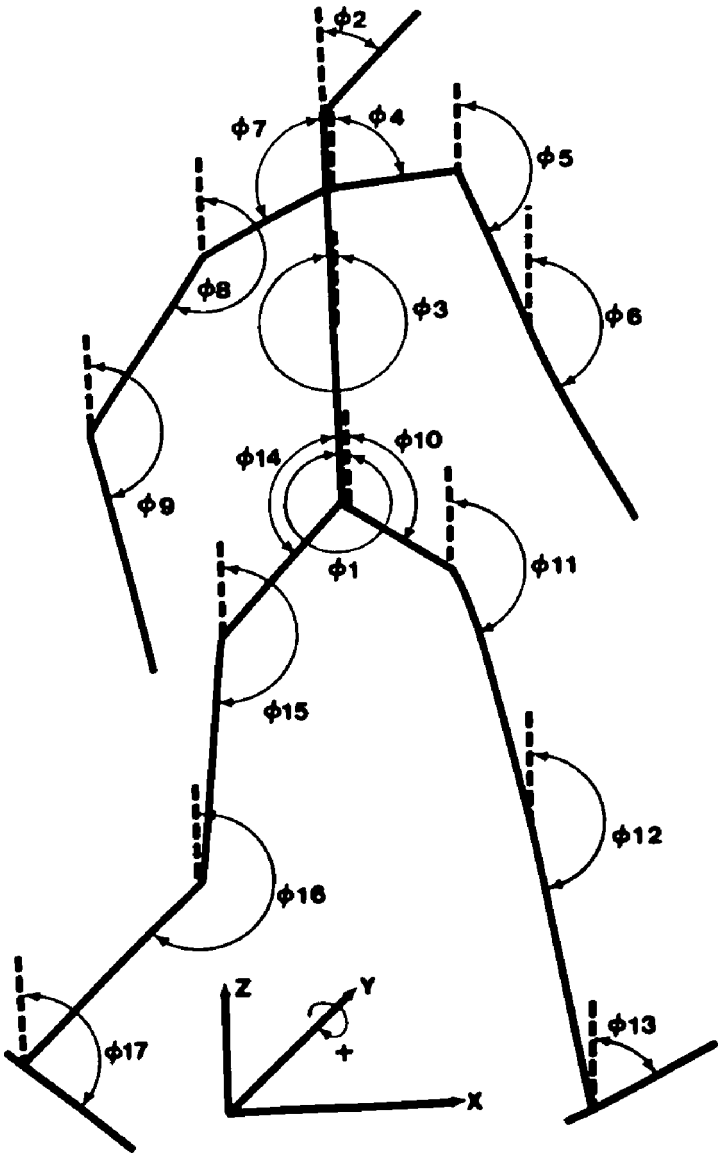


Figure 3.8.--Sagittal Plane Angles

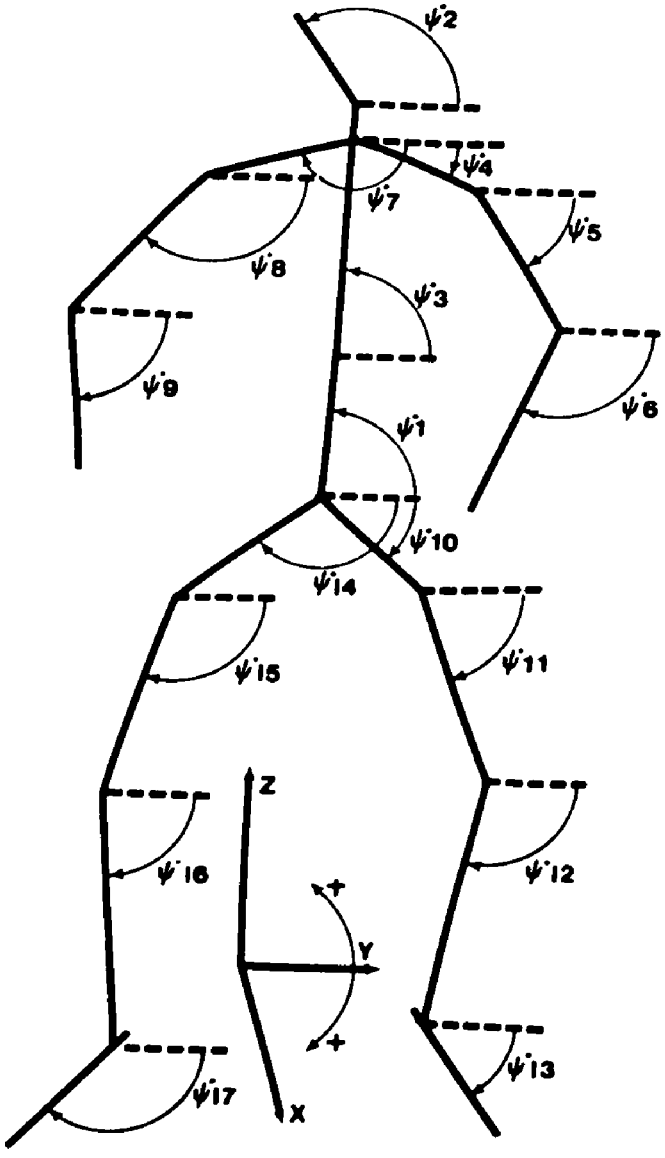


Figure 3.9.-- Conical Angles

CHAPTER FOUR

THE EXPERIMENT

The Prosthesis

Contruction and fitting of the subjects' below knee prostheses occurred at University Hospital, The University of Michigan. Those tasks were performed by the members of the prosthetic committee, Mr. Joseph P. Giacinto and Mr. Richard A. McUmber.

The latter two subjects for the experiment were selected from the group of B/K amputees who came to University Hospital for refitting. One of the primary reasons for this refitting is due to stump shrinkage. Since this process required a new stump socket to be constructed, it enabled a duplicate socket specifically for use in the experiment to be made at low expense.

In accordance with established procedure, the duplicate socket was constructed out of layers of 1 ounce dacron felt and of nylon stocking net impregnated with polyester 4110 resin. A mixture of polyester 4110 and microballoon filler was used to attach a hardwood base to the distal end of this socket. The microballoons were a phenolic plastic and were brown in color. Their purpose was to reduce the density of the adhesive, thereby saving weight. These phenolic microballoons were lighter, but harder to mix (less wettable) than the other commercially

available microballoons. This other variety is of glass and is white in color. It was to this hardwood base that the alignment unit and instrumentation were attached.

Similar to Pearson (1973), the alignment unit used was the below knee, adjustable shank designed by the Veterans Administration Prosthetics Center (V.A.P.C). It was donated for use in this project by The United States Manufacturing Company.

This unit was selected for many reasons. It was compactly designed. This maximized the available instrumented pylon length, facilitating a slender beam analysis on the thin-walled cylindrical pylon. The unit was light in weight, yet quite durable. Finally, this unit had six degrees of freedom, performing three linear and three angular adjustments independently.

This unit was not the typical one used by the prosthetists at University Hospital. They prefer to use a heavier unit, the Berkeley Adjustable Leg, manufactured by Hosmer/Dorrance. This unit is perfectly suitable for their work. However, it was not chosen for the experiment for two reasons. It was not compactly designed. There is often little room for a pylon. Furthermore, not all of the six degrees of freedom are independent. Hence, the technology of the U.S. Manufacturing Company was applied.

This author termed both of these 6 degree of freedom units as articulated in nature. This was in contrast to the rigid prosthesis the amputee normally wears.

Attachment of the alignment unit to the hardwood base required some redesign. The standard quick disconnect was removed and a flat 75 mm diameter base plate put in its place. Then the alignment unit was firmly attached to the hardwood via the base plate and a dozen wood screws.

Attachment of the instrumented pylon to the alignment unit and the S.A.C.H. foot to the pylon required no redesign. Standard V.A.P.C. attachment couplings were used.

The transition of the subject from his own rigid prosthesis to the articulated prosthesis was readily accomplished as both prostheses had sockets formed from the same plaster stump positive and both had the same rubber sleeve suspension. Since each subject's personal and duplicate sockets were identical, the subject used his own gel liner in the experiment. Not only did this represent a cost reduction, but using the same gel liner reduced the perturbation induced by the transition. The S.A.C.H. foot used on the articulated prosthesis was identical to the one on the rigid prosthesis and the subject's own shoes were placed on it. Finally, both prostheses were constructed to the same overall height, socket orientation, and to similar weight.

As mentioned before, the V.A.P.C. unit allowed independent linear and angular alignment adjustments. The range of the anterior or posterior and medial or lateral linear adjustments was 10 mm. Two power screw driven

dovetail assemblies provided these linear adjustments. The proximal-distal positioning of the S.A.C.H. foot over a 20 mm range was the third linear adjustment.

Four wedge nuts allowed motion of the two tilt plates. This provided flexion or extension and abduction or adduction adjustments of up to 10 degrees (0.175 radians). The third angular adjustment was the unlimited toe-in or toe-out of the S.A.C.H. foot.

Obviously, the above set of possible alignment adjustments was too large. To limit the set to a size small enough to accomplish during the experiment, yet still important enough to be worth doing, only flexion-extension angular adjustments were carried out.

The above choice was reasonable, since it was known that amputees tend to respond more to flexion-extension changes in their alignment than any other. This meant the perturbation that the system was probably most sensitive to was chosen for the experiment.

Initially, the prosthesis was aligned statically according to established procedures. Initially the S.A.C.H. foot was positioned so that the plane of its attachment base (ankle) was parallel to the floor. The pylon was then mounted normal to the foot attachment base. Finally, the stump socket was mounted so the weight bearing line passed about 20 mm forward of the pylon axis. No angular preload was introduced, such as mentioned in Pearson (1973). A mockup pylon and alignment unit assisted

In this effort by preventing unnecessary handling of the actual instrumented pylon. Each amputee was checked by the prosthetic committee for dynamic optimum alignment before the experiment began. This optimum alignment position became the neutral position in the set of alignment adjustments. The schedule of alignment adjustments made is documented in Table 4.1.

It was noted that extension of the pylon tended to cause the amputee to compensate by flexing his knee; flexion of the pylon, extending his knee. Many, especially the prosthetists, referred to pylon extension as knee flexion; pylon flexion, as knee extension. This train of thought was a manifestation of the desire to observe an anatomical response rather than the pylon perturbation.

TABLE 4.1

ALIGNMENT ADJUSTMENT SCHEDULE

Tests	Angle	Pylon	Description	Data
1-3	0.0^0	Neutral	Warmup	----
4-5	0.0^0	Neutral	Test	4
6-7	$+3.3^0$	Extension	Test	7
8-9	$+6.7^0$	Extension	Test	9
10-11	$+10.^0$	Extension	Test	11
12-13	0.0^0	Neutral	Re-zero	----
14-15	-3.3^0	Flexion	Test	15
16-17	-6.7^0	Flexion	Test	17
18-19	$-10.^0$	Flexion	Test	19

The set of flexion-extension alignment adjustments was selected in increments of 3.3 degrees. It was felt that an increment of this magnitude was small enough to allow a reasonable number of increments to be made, and sufficiently large to produce a noticeable effect each time it was applied.

Two kinds of continuity were important in the above schedule, subject and mathematical. By subject continuity, it was decided to begin the extension perturbations from the neutral position. This meant a smooth (ramp) entry into the experiment. A mathematical discontinuity then occurred from 10^0 extension to 3.3^0 flexion. Two data runs at the neutral position were included to smooth each

subject's transition.

Pearson, et al. (1973) treated the problem in the opposite order. There were no mathematical discontinuities. Data were taken from one extreme to the other without interruption. However, this did produce a subject discontinuity (step-function) when the first (extreme) perturbation was applied.

For safety reasons, the subject continuity was felt to have priority in this research. This enabled possible termination of the experiment if, as the alignment angle was increased, the subject was becoming dangerously unstable. This precaution was looked on favorably by the Human Use Committee.

The actual angular alignment adjustments were performed by proper positioning of the tilt nuts. A circumferential groove was cut on the outer edge of each tilt nut. The alignment adjustment was then made by aligning the nut's circumferential groove with an appropriate mark on the tilt plates. The positioning of the tilt plate marks was done by calibrating the alignment angle of the V.A.P.C. unit with a micrometer and appropriate trigonometry.

In addition to the construction of fitting of the prosthetic hardware, the members of the prosthetic committee supervised the application of the aforementioned alignment adjustments. They were careful to observe the subject's reaction to the perturbation, insuring that the

subject's safety was not jeopardized. Each amputee's verbal reactions to the perturbations were also encouraged and partially documented in Appendix D.

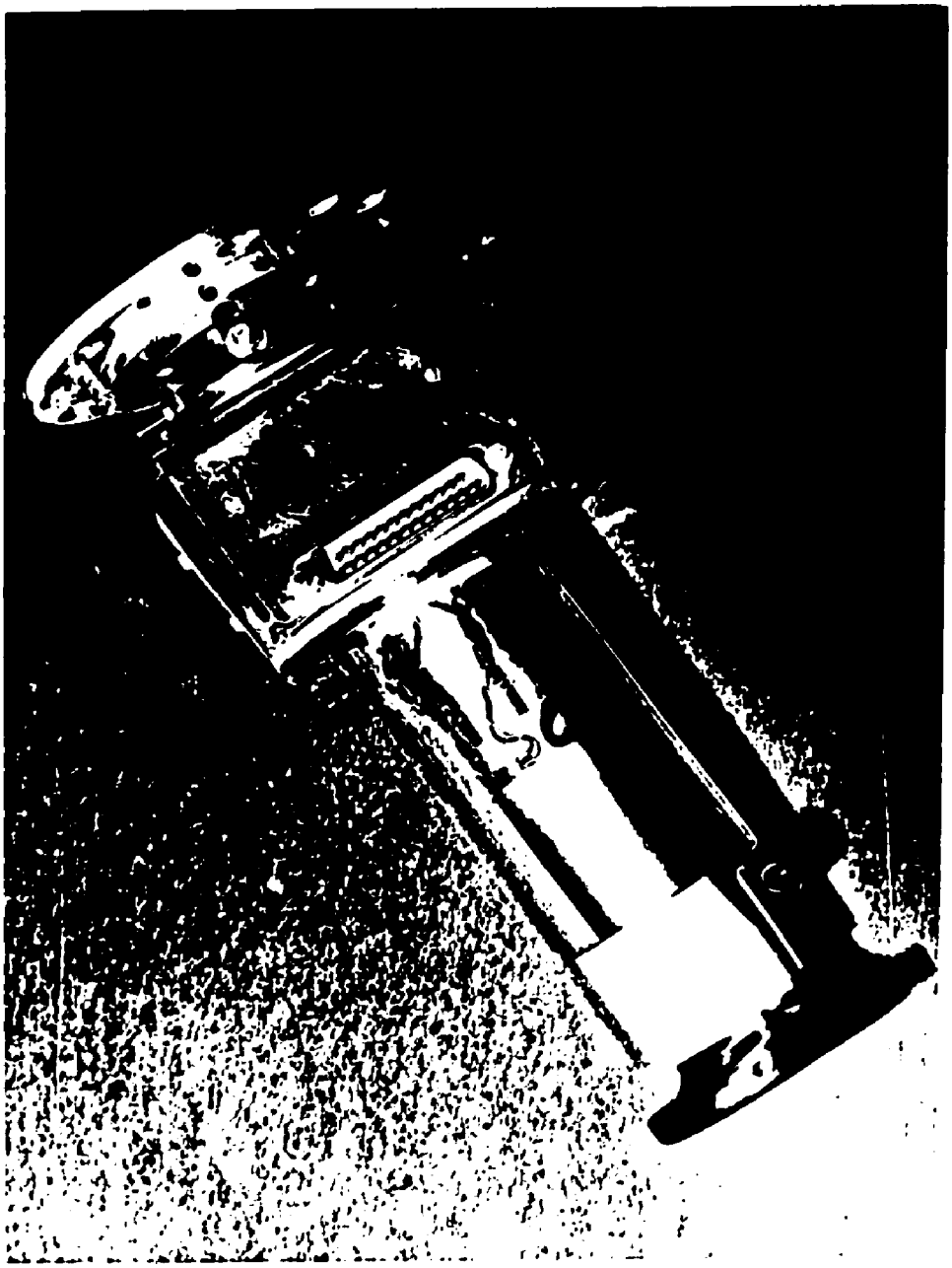


Figure 4.1.--V.A.P.C. B/K Adjustable Shank

Instrumented Pylon

The aluminum pylon of the articulated B/K prosthesis was augmented with 350 ohm strain gages to make it into a load cell. The empirical data from this load cell became the basis for establishing the degree of applicability of the model.

Other experimenters, such as Bressler and Frankel (1948), Chao (1975), Hughes (1967), Marks (1952), and Skorecki (1966), used instrumented force plates. Advantages to this latter method were avoidance of complex pylon biaxial loading and of the problems of mounting an instrumented pylon to each subject.

On the other hand, there were disadvantages. Thorton-trump and Daher (1975) discussed the possibility of misleading results due to subjects subconciously aiming at the force plate. Furthermore, a force plate instruments the floor, not the subject.

Based on the above thoughts, it was decided that instrumenting the system, in this case the B/K amputee, was the more reasonable route.

The positioning of the strain gages was patterned after Cunningham and Brown (1952). Strain gage signal conditioner, amplifier, filter, and computer storage availibility limited the pylon to four Wheatstone bridge circuits. Of the three possible forces and three possible moments, the pylon loads chosen for measurement were:

1. Flexion-extension bending moment
2. Axial torsion
3. Abduction-adduction bending moment
4. Axial force.

The particular gages used in each bridge are listed in Table 4.2.

The construction and calibration of this particular load cell was documented by Wisner (1975). His results were stored for the experiment by means of calibration resistors, Table 4.3. The particular resistors chosen, as well as the 350 ohm strain gages, were manufactured by Micro Measurements, Incorporated.

TABLE 4.2

STRAIN GAGE SELECTION

Bridge	Gage I.D.	# Gages	Type
Flexion-Extension	EA-13-125BZ-350	2	Single
Axial Torsion	EA-13-062TV-350	2	Pair
Abduction-Adduction	EA-13-125BZ-350	2	Single
Axial Force	EA-13-125BZ-350	4	Single
Axial Force	EA-13-062DN-350	4	Single

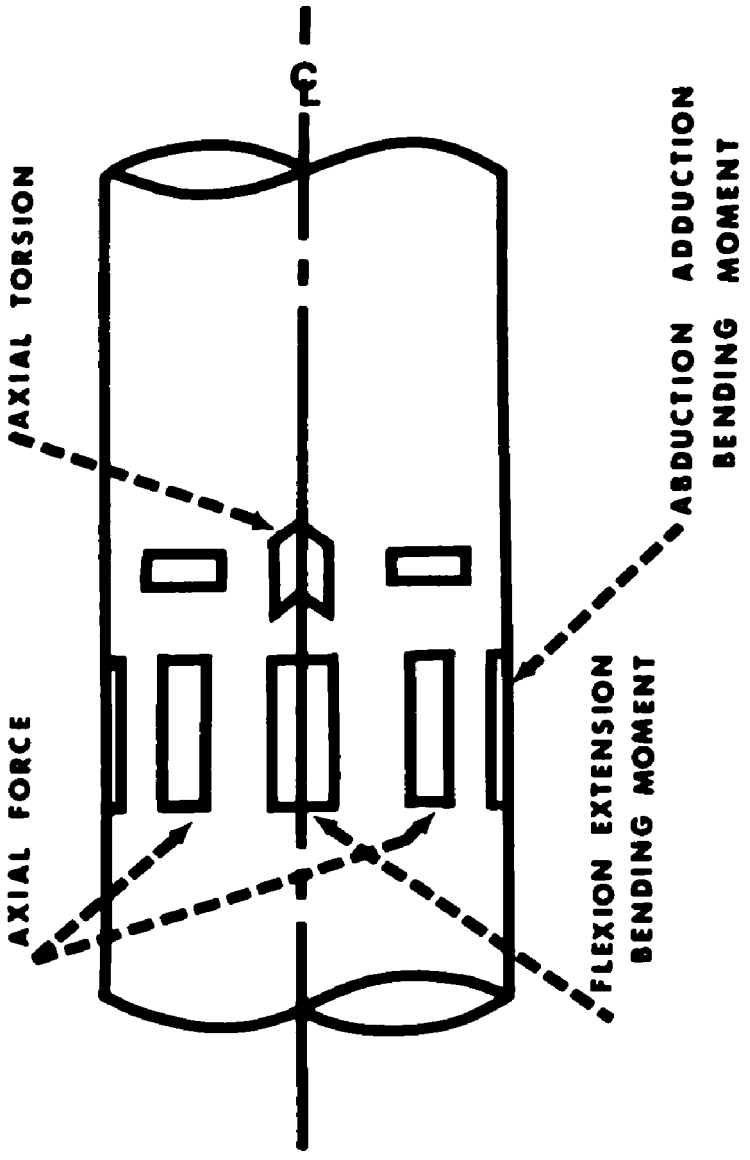


Figure 4.2.--Strain Gage Layout

TABLE 4.3

CALIBRATION RESISTORS

Bridge	Resistor	Load
Flexion-extension	57 983 ohms	150 Nm
Axial torsion	349 650 ohms	20 Nm
Abduction-adduction	349 650 ohms	25 Nm
Axial force	599 850 ohms	755 N

TABLE 4.4

BRIDGE SENSITIVITY

Bridge	Sensitivity
Flexion-extension	9.92 microvolts/volt/Nm
Axial torsion	13.1 microvolts/volt/Nm
Abduction-adduction	9.92 microvolts/volt/Nm
Axial force	0.09 microvolts/volt/N

The strain gage signal conditioners used to excite the Wheatstone bridges, balance them, and shunt the appropriate calibration resistors when required was the Daytronic 601B. The Hewlett Packard 8875A differential amplifier was then used to amplify and filter the 601B output in preparation for signal transmission to the Linc-8 data storage computer. Table 4.5 is provided to show the particular bridge direct current (dc) excitation voltage, diffferential amplifier gains, and single pole, low pass

filter breakpoint frequencies used in the pylon portion of the experiment.

TABLE 4.5

PYLON INSTRUMENTATION SETTINGS

Bridge	Voltage	Gain	Filter
Flexion-extension	4	100	200 Hz
Axial torsion	6	300	200 Hz
Abduction-adduction	7	300	200 Hz
Axial force	6	1000	200 Hz

The low pass filter was set at the same frequency as the sampling rate. This seemed to be a reasonable way to filter out aliases. The actual filter response was 4 decibels (db) off from the asymtotic Bode representation. However, this did not interfere with the data interpretation. Roll off was at 20 db/decade beyond the breakpoint.

The dc excitation voltages were sufficiently small that only 390 milliwatts had to be dissipated by the pylon. This resulted in a steady state temperature rise of 1 Kelvin with a temperature stabilization time of 15 minutes.

The heating question might not seem important, were it not for the fact that the pylon was enclosed by a clear plastic cylinder. This was to prevent damage to the strain gages. It also prevented substantial free convection.

Hence, heat dissipation requirements had to be kept low.

Perhaps this was a blessing in disguise. Although the gages were temperature compensated for the 2024-T4 aluminum pylon, forced convection, such as during the swing phase, could have given misleading results due to non-uniform cooling of the pylon surface.

Normal Pressure Transducers

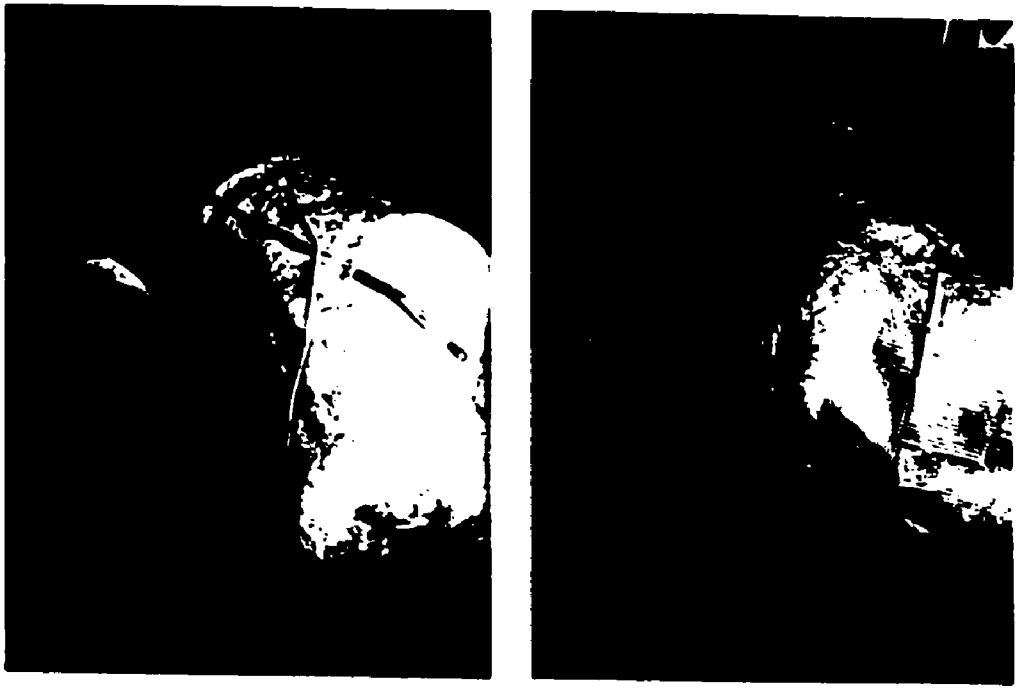
Some information was gathered concerning the manner the pylon loads were borne by the stump. Two Kulite pressure transducers, similar to the ones used by Corell (1969), were tape mounted onto the stump. Due to equipment limitations, at most two pressure transducers could have been mounted.

It was felt that the ease in which the Kulite gages were inserted between the stump and the gel liner outweighed the less expensive gages used by Appoldt, Bennett, and Contini (1968).

The regions chosen were the patellar tendon and distal portion of the gastrocnemius muscle, Figures 4.3 and 4.4. In this manner, it was hoped to follow the spirit of Radcliffe (1962) and sketch a relationship between flexion-extension bending moment and stress. Furthermore, these data were needed if the finite element analysis were ever undertaken. Results of the Patellar stress versus flexion-extension bending moment as well as the Gastrocnemius stress versus axial force are presented in Chapter Six.

The transducers were carefully recalibrated to check the factory calibration. A diaphragm pressured with air was used to compress each transducer against a metal plate, Figure 4.5. To protect against shear caused by expansion of the diaphragm's surface, a sheet of silicon gel was placed between the diaphragm and the pressure transducer as well as between the transducer and the metal plate. The results of the recalibration are given in Table 4.6. Values were within 8% of the factory predictions.

KULITE TRANSDUCER MOUNTING



PATELLAR GAGE GASTROCNEMIAL GAGE

Figure 4.3.--Subject 2: Normal Pressure Transducers

KULITE TRANSDUCER MOUNTING



PATELLAR GAGE GASTROCNEMIAL GAGE

Figure 4.4.--Subject 3: Normal Pressure Transducers

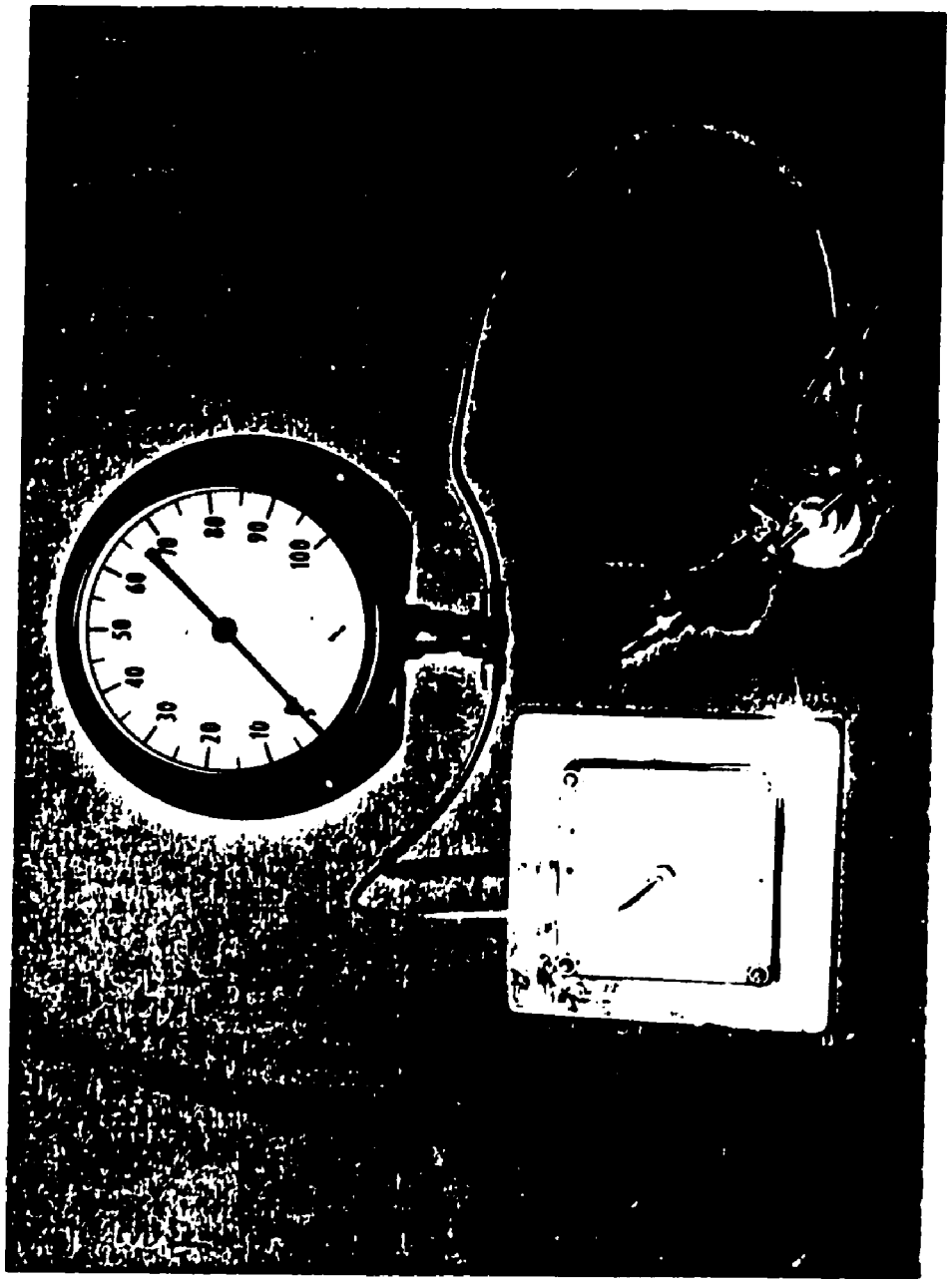


Figure 4.5.--Kulite Transducer Recalibration

TABLE 4.6

KULITE RECALIBRATION RESULTS

Model	Gage I.D.	Sensitivity	Function
LPS-125-200	1714-7-9	1.057 mV/V/kPa	Patellar tendon
LQL-125-25	2083-7-11	4.347 mV/V/kPa	Gastrocnemius

Table 4.7 is provided to show the particular transducer dc excitation voltage, differential amplifier gain, and single pole, low pass filter breakpoint frequency used in the experiment.

TABLE 4.7

PRESSURE INSTRUMENTATION SETTINGS

Gage	Voltage	Gain	Filter
Patellar tendon	1	100	200 Hz
Gastrocnemius	1	30	200 Hz

As with the pylon load cell, the low pass filter setting was at the 200 Hz sampling rate of the transducer outputs. The poor heat dissipation characteristics of human flesh required that as little heat as possible be generated by each Kulite sensor. The 1 dc volt excitation reasonably accomplished this by limiting the power output to 1 milliwatt per gage.

It was interesting to note that some years ago, a

previous researcher caused blisters to form on a B/K stump with similar gages and only a 5 volt dc excitation. Truely, the problem of adequate heat dissipation was not trivial.

Displacement Data

The subjects' linear and angular displacement data dominated the theoretical and experimental effort. It was from these data that the theoretical pylon loads were calculated for comparison with the actual pylon loads. These same displacement data were then used to study the subjects' energy expenditure surfaces, Figures 7.2 and 7.3.

Many possibilities existed for gathering the displacement data. Grieve (1969) used polarized light and special targets; Winter, et al. (1974), a television system viewing diode targets. Waters, Morris, and Perry (1973) as well as Ralston and Lukin (1969) tied cords to specific segments and measured translatory motion with displacement transducers. Morris (1973) used accelerometers; Page and Infante (1973), magnetometers.

It was decided for this experiment to use the somewhat less sophisticated but just as applicable 16 mm motion picture camera.

Three stationary 16 mm cameras were used to view two orthogonal planes, the sagittal and frontal. This was a plan similar to the one used by Hughes (1967). Figure 4.6 shows the camera layout for a left B/K; Figure 4.7, a right

B/K. The purpose of this inversion was to allow full view of the prosthetic limb by the sagittal plane camera and to always have each subject walking towards the frontal plane camera.

The characteristics of the three cameras used is listed in Tables 4.8 and 4.9.

TABLE 4.8

CAMERA CHARACTERISTICS FOR SUBJECT 1

Name	Number	Fps	Lens	F-stop	Shutter	Exposure
Cine Kodak	1	64	25mm	1.9	1	1/140
Cine Kodak II	2	64	25mm	1.4	1	1/140
Bolex Rex 4	3	64	50mm	1.4	1	1/220

TABLE 4.9

CAMERA CHARACTERISTICS FOR SUBJECTS 2 AND 3

Name	Number	Fps	Lens	F-stop	Shutter	Exposure
Cine Kodak	1	32	25mm	1.9	1	1/70
Cine Kodak II	2	32	25mm	1.4	1/4	1/280
Bolex Rex 4	3	32	25mm	1.4	3/4	1/225

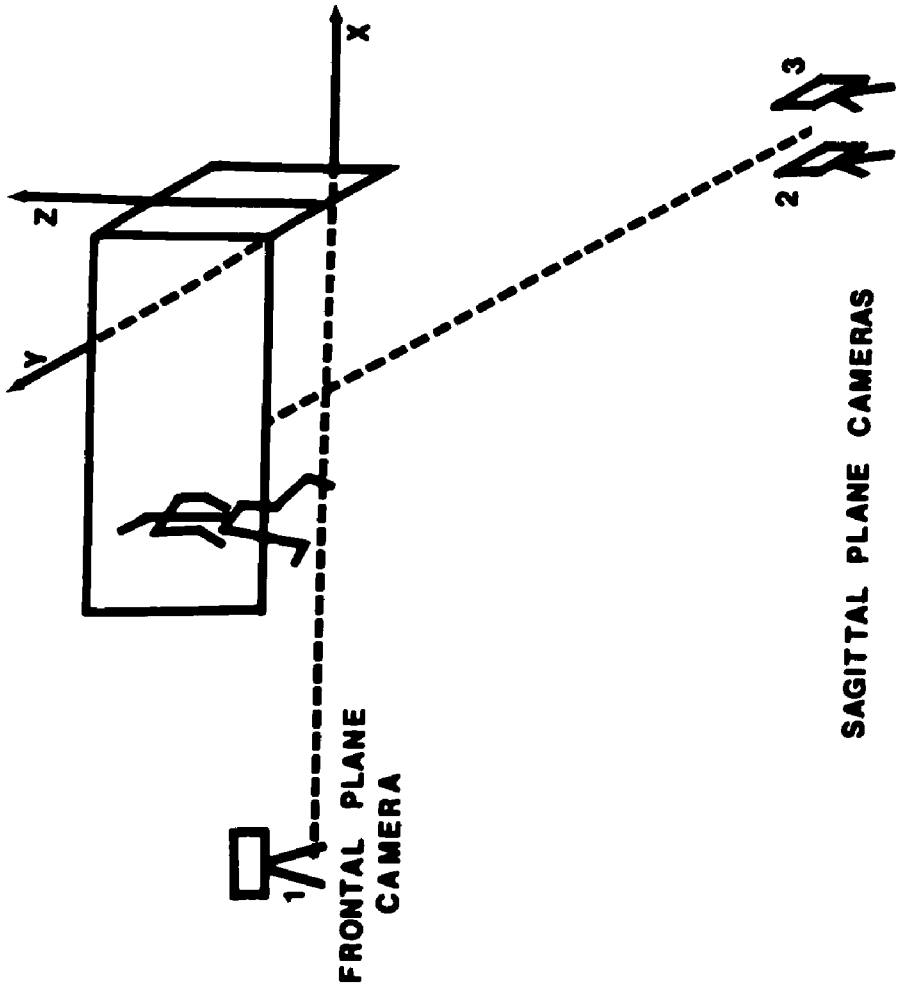


Figure 4.6.--Camera Layout for a Left B/K

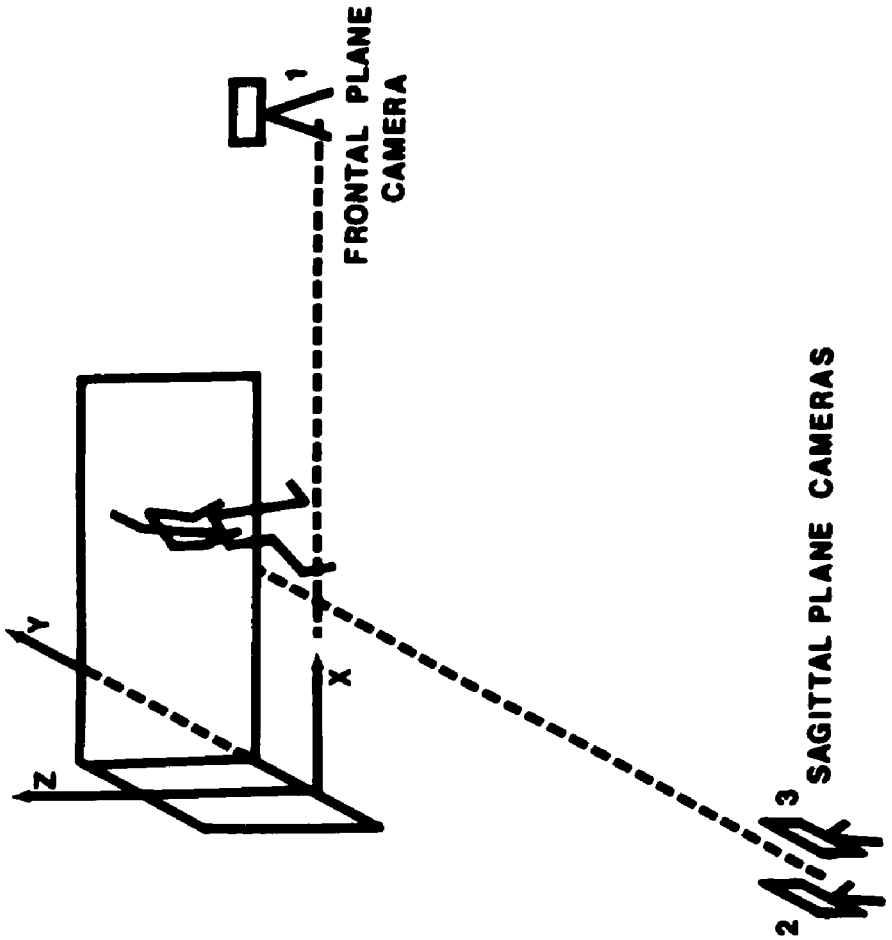


Figure 4.7.--Camera Layout for a Right B/K

The purpose of the cameras was to gather enough raw data so that the 37 time dependent coordinates required by the differential equations of motion could be determined. It was decided quite early in the experiment preparation period to use Kodak 4XR 449, black and white, 16 mm, double perforated, reversal film. This choice was made mostly in the interest of obtaining as high an ASA as possible.

The filming was done indoors in the Functional Training Gym, University Hospital. Three DP12 Smith Victor floodlights, each with a 250 watt tungsten bulb, provided illumination. Lexan diffusion screens augmented the floodlights.

Each 16 mm frame was correlated to real time and the Linc-8/PDP8E computer by placing a computer-triggered counter-timer in each camera's field of view.

These electronic counters became the lighting restriction. The lighting had to be limited to 750 watts. Increased lighting and subsequent increase in lens f-stop would have rendered the fixed illumination counters invisible and useless. Based on counter visibility, the f-stops chosen for use were the lowest ones available on each camera lens. One observation about this whole effort is that the use of light emitting diodes as discussed in Dearing (1972) would have probably prevented a lot of headaches concerning borrowing of equipment, triggering of the counters, proper positioning of the counters in the depth of field, etc.

The three floodlights were positioned with the use of a reflective light meter and an 18 percent reflective card. The article by Stidham and Colortran (1972) was helpful in this task.

The work of Sato (1975), with Subject 1, demonstrated that 32 frames per second (fps) was a good camera speed. This provided a change in data being observed greater than the error in the measurement. To have increased the camera speed would have made the change in data equal to or less than the measurement error. The measurement noise would be the same in both of the preceding situations. However, the latter would have given rise to enough noise in the first and second derivatives to have made the double differentiation of this data, as done in Chapter Five, difficult or meaningless.

Under the tungsten light, the 4XR film had an ASA of 320. This was sufficiently high that at 32 fps it was possible to partially close each sagittal plane camera's variable shutter. This partial closure favorably increased the clarity of the photographs.

Since the frontal plane motion was much less than in the sagittal plane, a fully opened shutter was used for this view. However, the frontal plane camera had a special problem. Its depth of focus had to be sufficiently great to bracket the subject's motion as he walked towards the camera. This explains why the frontal plane camera's f-stop was 1/2 stop above the sagittal plane camera's, Table

4.9.

Through this increase in f-stop, it was possible to photograph clearly the subjects' motion. Considering the small angle approximation, there was no depth of focus problem while viewing the sagittal plane.

In order that raw data be gathered from each frame of film, the subject had to be marked with targets. In general, black and white film has a higher ASA than color. This is why it was chosen for the experiment. Once chosen, it dictated that each subject be marked with white targets over black tights and leotards.

The targets used were those constructed and tested by Sato (1975). The subject wore ten plastic Cartesian coordinate systems, Figures 4.8 and 4.9. They were placed on the head; pelvis; and the left and right arms, forearms, thighs, and shanks. Styrofoam balls marked the endpoints of the plastic axes. Styrofoam hemispheres were also placed on the subjects' joints.

To assist visibility of the white styrofoam balls, a black background was placed behind the subject. To enhance visual observations without the use of special equipment, 0.250 metre squares were marked on it.

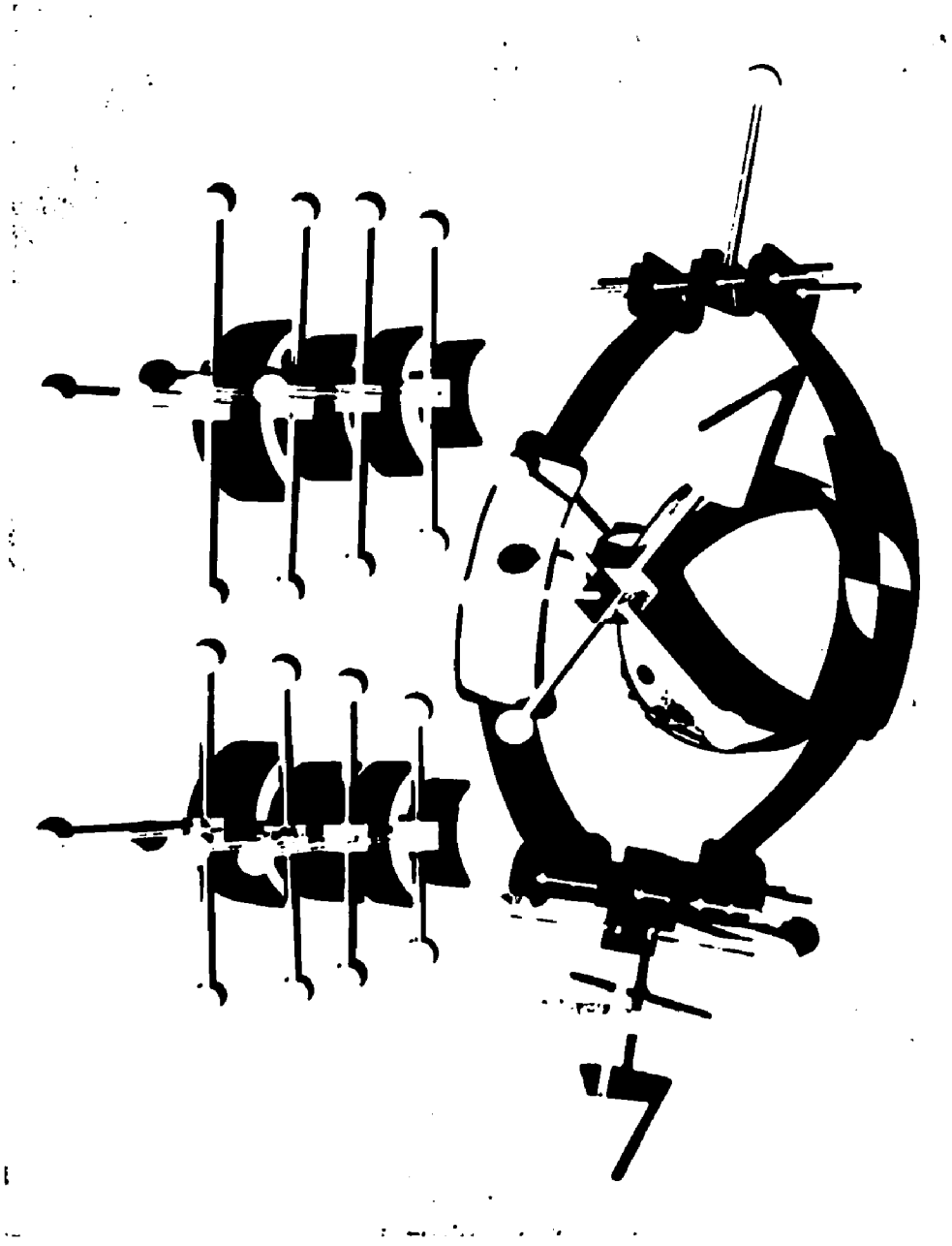


Figure 4.8.--Ten Plastic Coordinate Systems

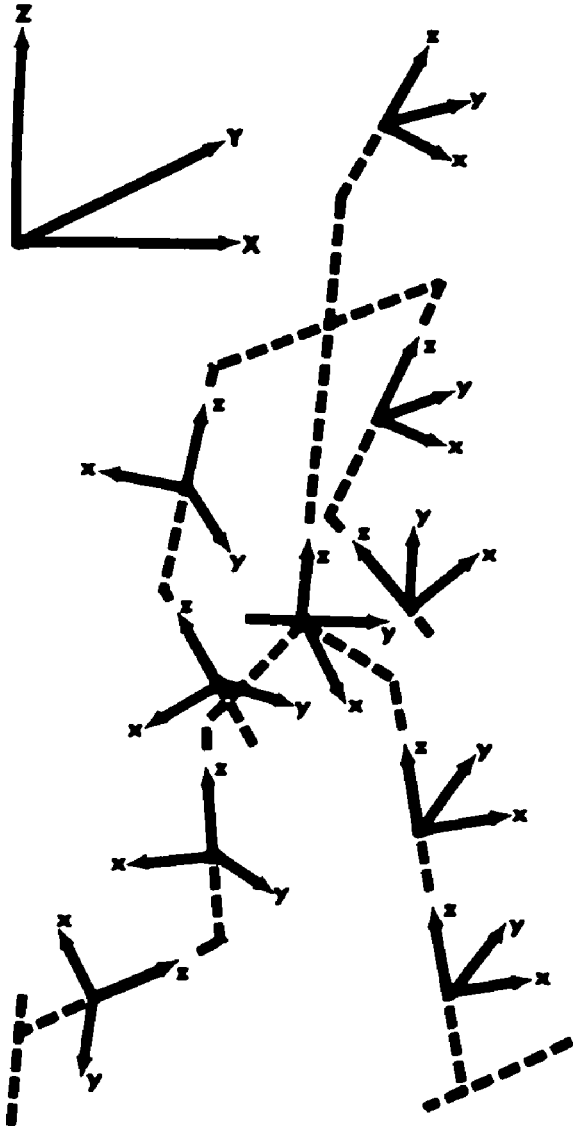


Figure 4.9.--Coordinate System Mounting

Footswitches

Footswitches were used to augment the photographed displacement data. Although it was possible to define heel-strike and toe-off times from the films, the foot swtiches also gave some idea of the traverse of the center of pressure from heel to toe. Indications of non-load bearing contact or foot scuffing during the swing phase were also made apparent by these swtiches.

After consulting with the literature of Corell (1969), Leavitt (1970), Ralston and Lukin (1969), Schwartz, Heath, and Wright (1933), and Winter, Greenlaw, and Hobson (1972), it was felt four switches per shoe would be adequate. They were mounted on the dorsal side of each shoe as is described in Figure 4.12. This figure shows switch A on the posterior heel; switch B, anterior heel; switch C, sole; and switch D, the toe.

The actual choice of ground contact transducer was made on three criteria, low cost, ease of mounting, and accessibility for repairs. This precluded the use of microswitches carefully built into each subject's shoes, as done by Winter (1972) and by Ralston (1969). Furthermore, an embedded microswtich which was shorted out or permanently jammed open would require a great effort to repair.

This analysis lead to the use of externally mounted transducers. Rather than use discs as Schwartz did (1933), it was decided to use a thin strip shape. Each strip went

from the medial to the lateral side of the shoe. This shape gave a better chance of determining the heel-strike or toe-off times than discs.

The footswitches chosen were manufactured by Tapeswitch Corporation of America. The particular model was the Controflex 121-BP. Representative samples were carefully calibrated. Results showed that the maximum load required to trigger the switch was about 100 Newtons per metre of switch length. Since this value was obtained by compression of the switch between parallel platens, Figure 4.10, it represented the maximum loading necessary. This calibrator was constructed by Chow (1974).

In walking, the initial contact between shoe and floor could hardly be described as the merging of parallel plates. Since there is some angle of merging, any load on the switch was probably applied in a small region of the switch rather than uniformly across it. Hence, the ambiguity of contact was probably not 100 N/m times the switch length. Allowing for a probable 10 mm length of initial load bearing contact along the switch, it was more like 1 Newton.

Four switches per foot required eight data channels if each switch were to have been sampled separately. Since only 2 channels were available, one 8 bit digital to analog (D/A) converter was used per foot. This enabled the digital outputs from each set of four footswitches to be sampled via one analog channel.

The resolution limit of the analog to digital (A/D) converter on the mini-computer used to store this data made it impossible to put all eight switches on one analog channel. Since the required 4 bit D/A converters were not commercially available, the larger 8 bit units were chosen.

The D/A converters, model DAC-19-8B, were manufactured by Datel Systems, Incorporated. In Figure 4.11, a D/A and footswitch are sketched.

The converter required a +15 volt dc and -15 volt dc input as well as a signal ground. The D/A had two possible output ranges as determined by the offset and the +IN or the -IN. Connection of the offset to the +IN gave a unipolar voltage range from 0 to +10 volts; to the -IN, a bipolar range from -5 to +5 volts. The latter was chosen and reduced by a resistor voltage divider to a -1 to +1 volt signal range. This signal was then sampled without subsequent passage through filter or amplifier.

Both D/A converters were 8 bit. The best resolution of the analog output was obtained by using the four most significant bits of each D/A for each set of four footswitches. Table 4.10 shows the relationships between the binary digital inputs of the four switches A, B, C, and D and the analog output. Table 4.10 is organized to count from zero to fifteen in base two by treating the four footswitches as binary digits. Figure 4.12 shows the typical mounting of four footswitches on each foot.

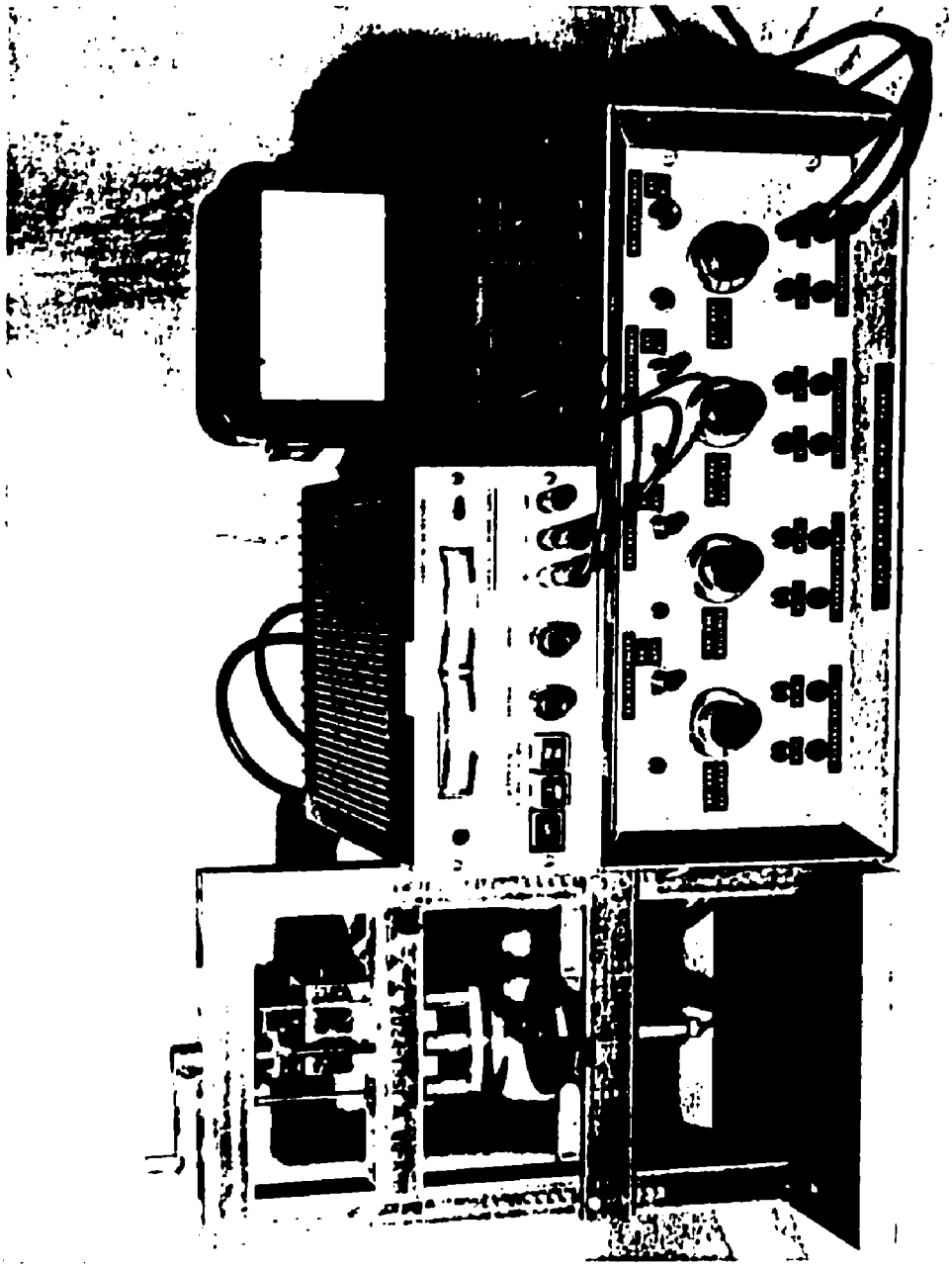


Figure 4.10.--Footswitch Calibration

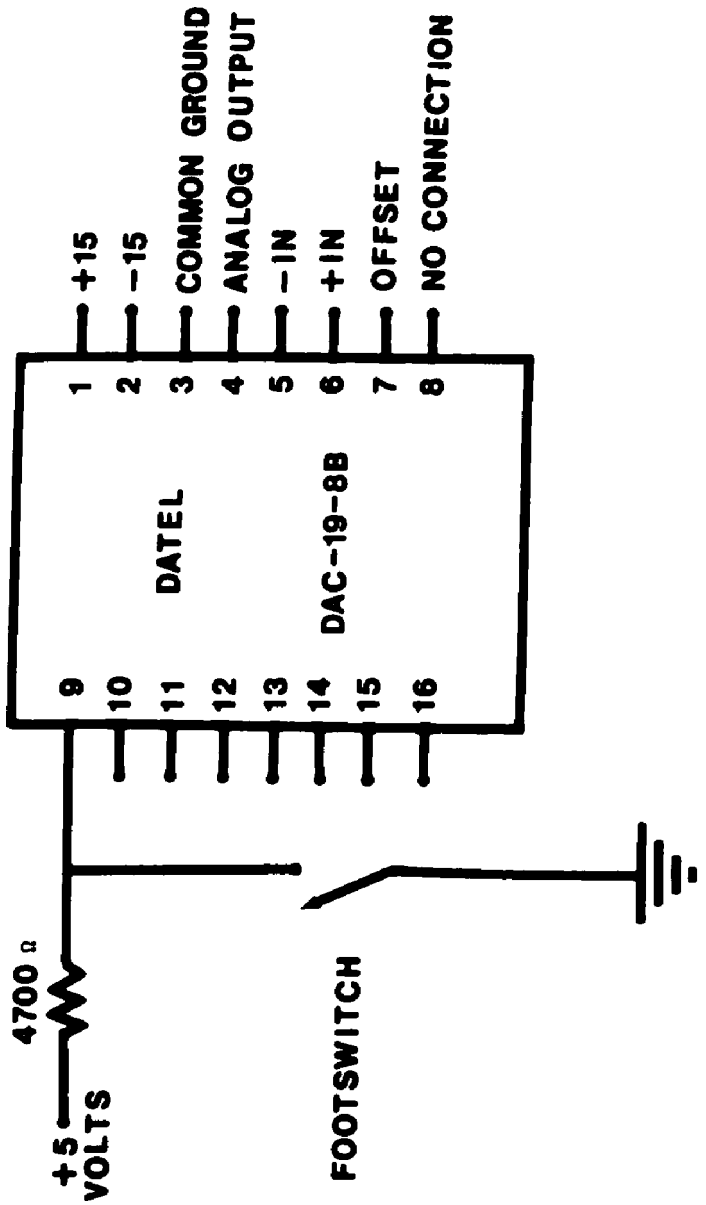


Figure 4.11.--Digital to Analog Converter

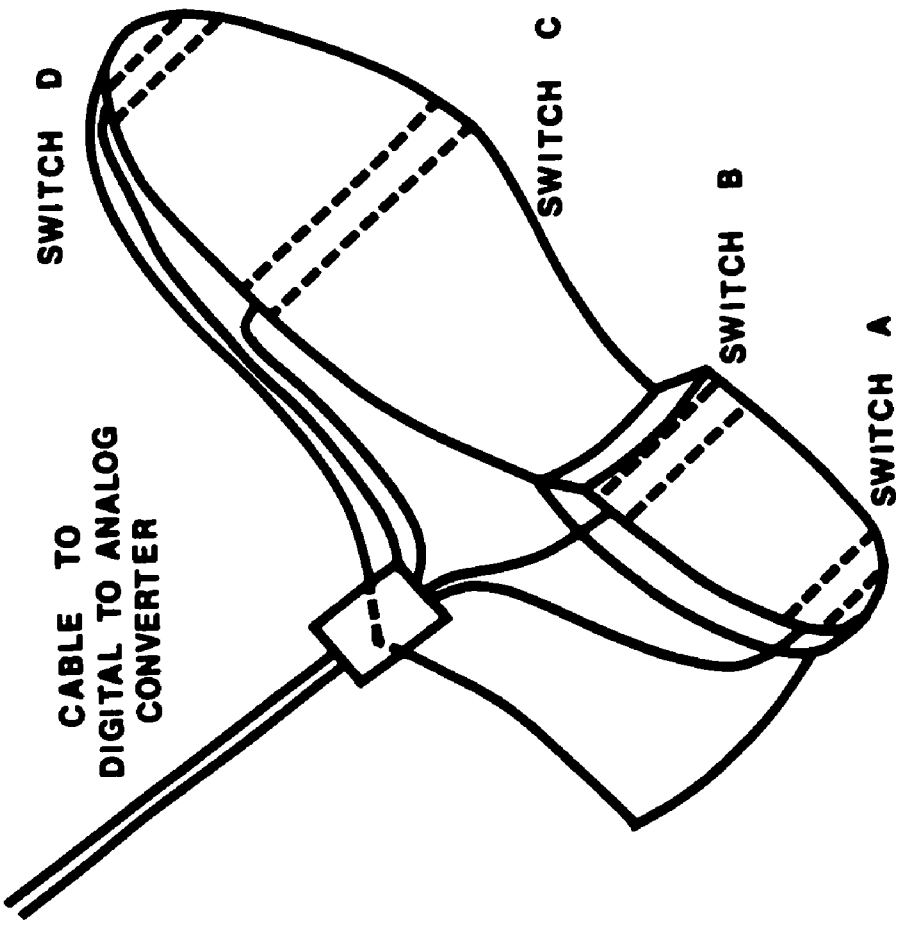


Figure 4.12.--Footswitch Mounting

TABLE 4.10

BINARY DIGITAL INPUT AND ANALOG OUTPUT

ABCD	Analog Output
0000	+0.000 volts
0001	+0.125 volts
0010	+0.250 volts
0011	+0.375 volts
0100	+0.500 volts
0101	+0.625 volts
0110	+0.750 volts
0111	+0.875 volts
1000	-1.000 volts
1001	-0.875 volts
1010	-0.750 volts
1011	-0.625 volts
1100	-0.500 volts
1101	-0.375 volts
1110	-0.250 volts
1111	-0.125 volts

Probability theory states that for n variables, the number of combinations taking 1, 2, 3, ..., n at a time is 2 to the power n. This was another way of showing that for 4 footswitches, there were 2^4 or 16 possible output states. Division of the 2 volt window by this value gave a figure of 1/8 volt. In order to calculate the analog output, it was simply a matter of adding 1/8 volt successively for each additional increment in the binary number. The only discontinuity occurred at the jump to -1.000 volt. This allowed the voltage output to remain within plus or minus 1 volt.

Linc-8/PDP8 Computer

The mini-computer used in the experiment was the Linc-8/PDP8E manufactured by the Digital Equipment Corporation, Figure 4.13. If one considered the pylon load cell, pressure transducers, 16 mm cameras, and footswitches as spokes in a wheel, then the mini-computer was certainly the hub.

All data were gathered as functions of real time. The 16 mm cameras photographed real time as displayed in their respective fields of view. This timing was triggered by the computer as it began the sampling of the pylon loads, stump pressures, and footswitches. This enabled relating the photographed displacement data to the electronically gathered data.

The software which controlled the real time, as well

as stored the sampled pylon loads, stump pressures, and D/A output, was the data acquisition program, Sample8.

The original program, Sample, was written by Corell (1969). Stribley (1974) did work to enlarge the capability of Corell's program. His fourth revision became his final version, Sample4. The main problem with both of the above was that timing between samples was controlled by millions of iterations through instructions having an uncertain microsecond duration. This author corrected the ambiguity by using the programmable clock routine written by Wilbur DeHart. This addition and the inclusion of an interactive graphics package written by the author was the eighth version of Corell's program, Sample8. The interface between Sample8 and the experimenter was the portable control panel, Figure 4.14. This most useful piece of equipment was designed by Corell (1969) and built by Wilbur J. Nelson.

The portable control panel had all the strain gage signal conditioners, differential amplifiers, and filters required to balance the pylon and Kulite strain gage bridges and amplify the bridge outputs for transmission to the computer. The labeling of portable control panel analog inputs and computer analog inputs for respective functions is documented in Table 4.11. It was important to remember that the analog voltage had to be in the range of -1 to +1 volt. Excess voltage could have caused saturation of the analog to digital converter in the mini-computer.

It was also necessary to consider the resolution of the analog to digital converter. There were 256 discrete levels from zero to +1 volt, or 511 levels throughout -1 volt to +1 volt. This gave a range of 3.9 millivolts per discrete level, or bit. Unsaturated resolution would then amount to (0.0039/2)*100% or about 0.2% ambiguity in a full scale measurement.

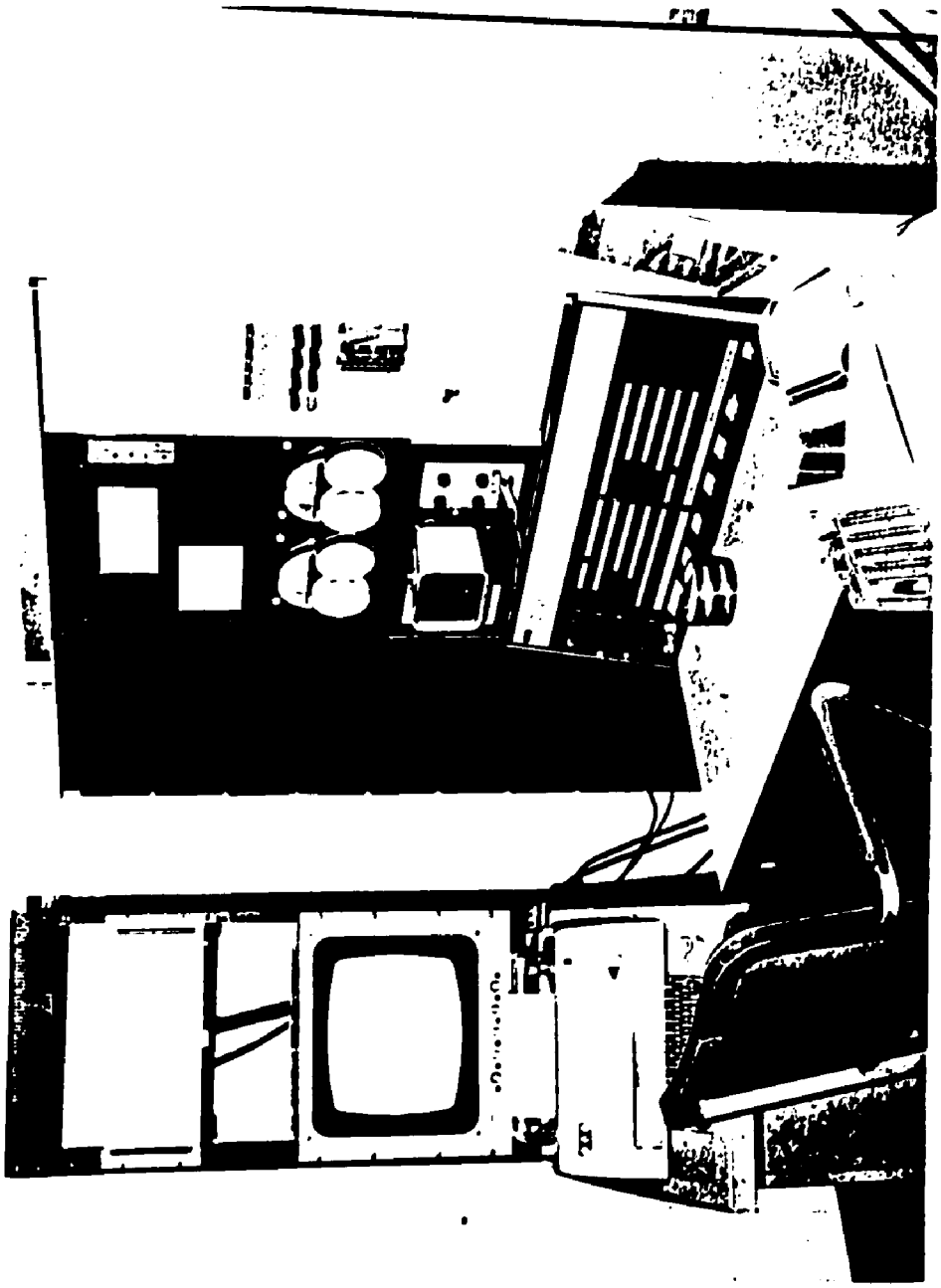


Figure 4.13.--Linc-8/PDP8 Computer

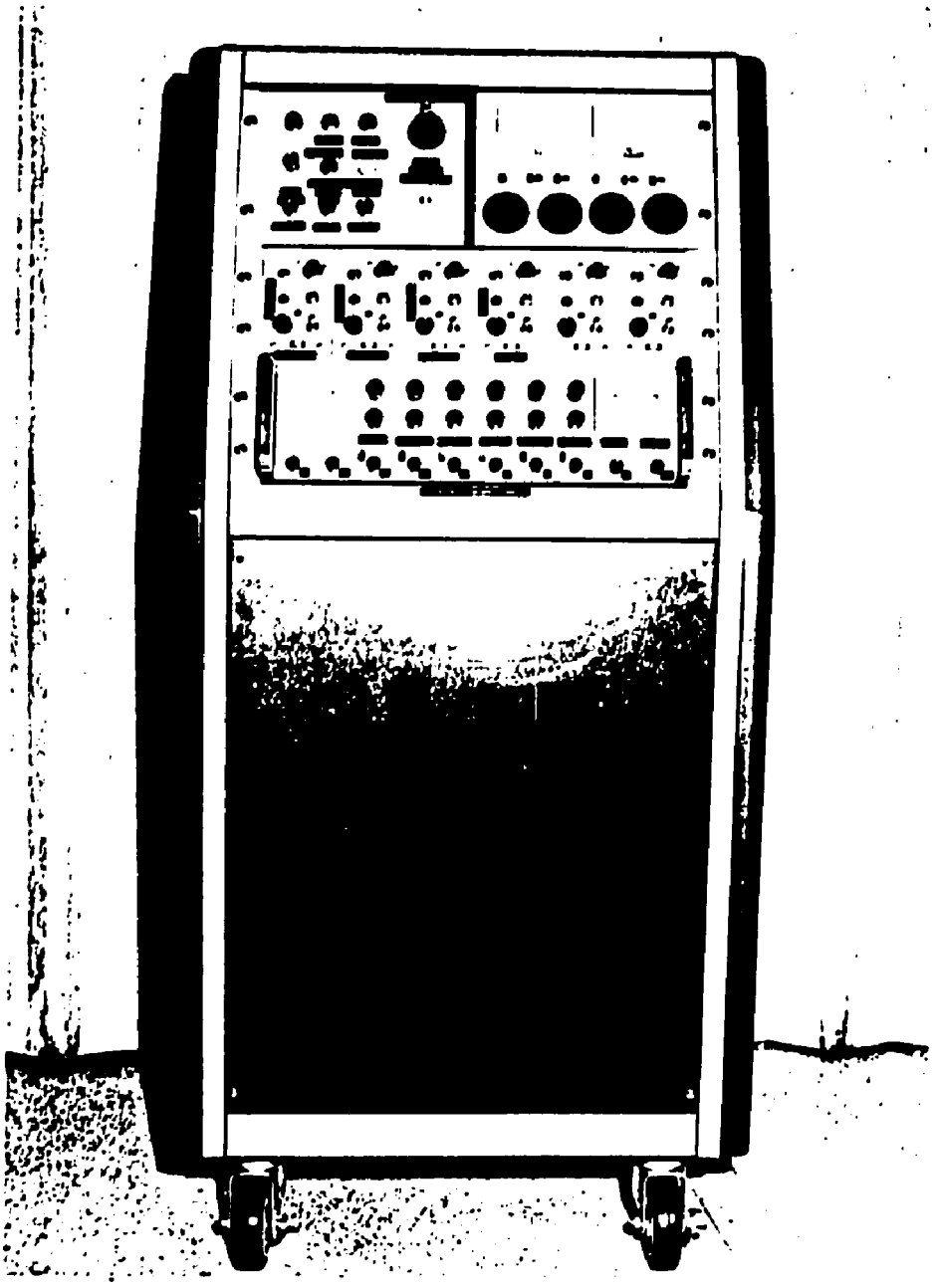


Figure 4.14.--Portable Control Panel

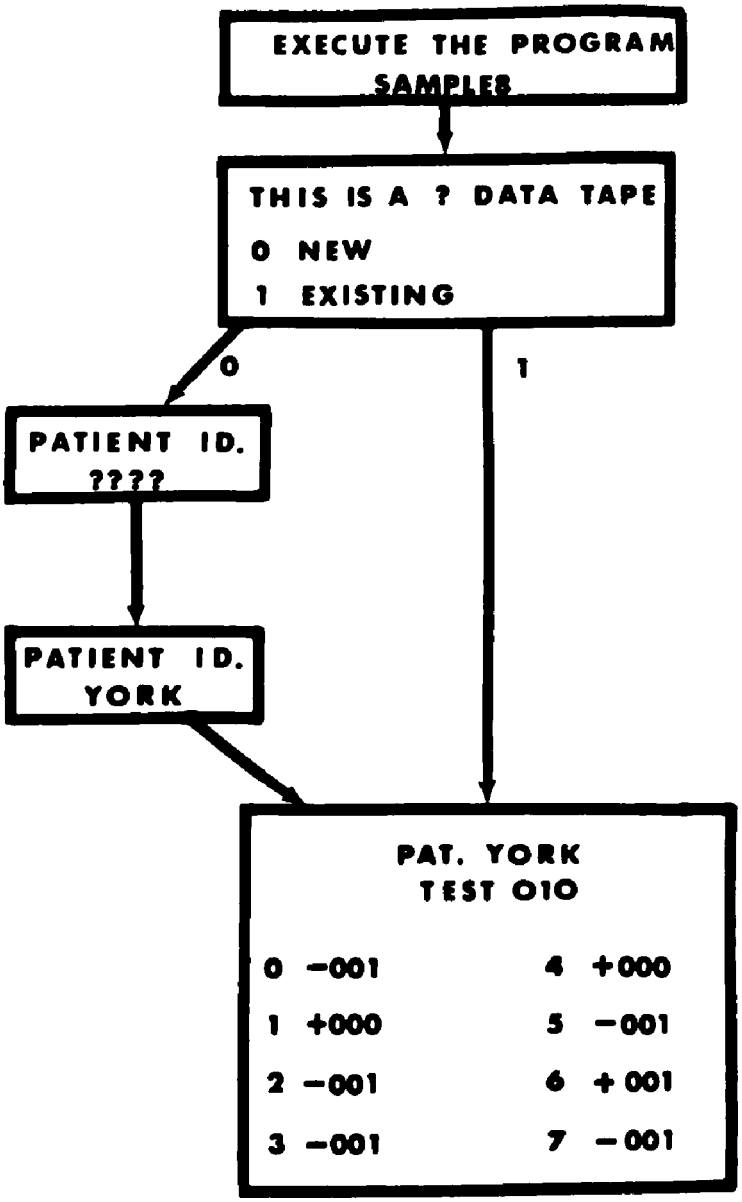


Figure 4.15.--User Entry Into Sample8

TABLE 4.11

SAMPLE LINES

Panel #	Computer #	Function
1	0	Flexion-extension moment
2	1	Axial torsion
3	2	Abduction-adduction moment
4	3	Axial force
5	4	Patellar tendon stress
6	5	Left footswitch
7	6	Right footswitch
8	7	Gastrocnemius muscle stress

Four externally controlled sense lines and four computer controlled relays concluded the interface hardware. The sense lines were used to manipulate the computer from the experiment site, a remote location 50 metres away from the computer.

The computer's internally actuated relays made it possible for the experimenter to identify the activity of the data acquisition program. This communication was achieved via four colored lights, each wired in series with one relay and mounted on the portable control panel.

Sample8 was the eight channel data acquisition program. Before experimental data were taken, questions about the data storage tape were asked, Figure 4.15. This questioning was terminated with the display of the instantaneous input of all 8 channels. This was quite

useful in observing the pre-experiment status of the various transducers and in balancing the Wheatstone bridges.

The data sampling cycle is sketched in Figure 4.16. The interval between successive samples was governed by the 20 kilohertz programmable clock. Every 100(base 10) cycles, or 5 milliseconds, the sampling process occurred, with each of the 8 samples taken and stored in memory every 28.5 microseconds. In between sampling processes, time was displayed in 5 millisecond intervals in base 8, Figure 4.16.

Since the computer did not have differential inputs for sampling, care had to be taken to avoid ground loops. All remote equipment, including the computer chassis, were grounded to the same point in a tree-like fashion. This action diminished noise to a level of 1 bit out of 256.

After the 1400(8) or 768 (10) samples were taken, the two subprograms allowed visual inspection of the data while it was still in core. The first graphed three quarters of data on the Dumont scope screen by omitting every third point. Each graph of three quarters of data was a sketch of all data gathered by a specific channel. Appropriate labels were displayed at the bottom of each picture. These were the test number, core location of the moveable cursor, function being observed, and the datum at the core location of the cursor, Figure 4.17.

In the second interactive graphics package, two

quarters of data were shown at a time. No data points were omitted. This enabled observation of every data point. However, this particular package was awkward to maneuver. Since the data storage in core was not in a particularly accessible pattern, Figure 4.18, this program was almost always omitted in favor of the first package.

As in the first package, a moveable cursor allowed pointwise inspection of the data. Control of the cursor was done by sample line 6, a dual purpose line. Primarily meant for right footswitch data, sample line 6 changed to a cursor control mode via a normally open, momentary contact switch. Failure to activate this switch resulted in the sample line maintaining its footswitch mode by default.

Once the data had been inspected, it was possible for the experimenter to exit the display mode and write the six memory banks of data onto 12 bit magnetic tape. At the end of this task, the Sample8 returned to its instantaneous 8 channel display. The user had completed one data taking cycle and was prepared to do another.

Had the user not been satisfied with the data stored on magnetic tape, it was possible to delete it during this 8 channel display mode. It was also possible to exit the sampling program at this time.

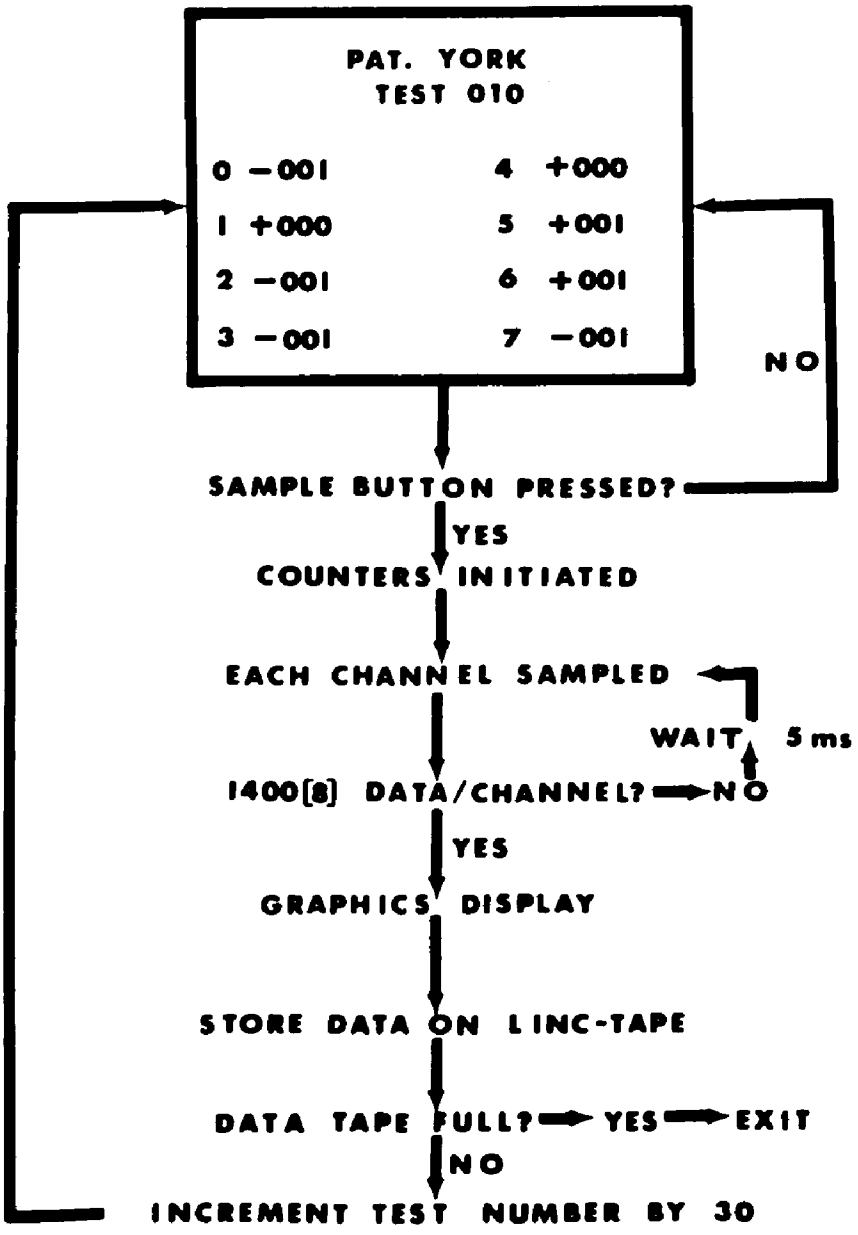
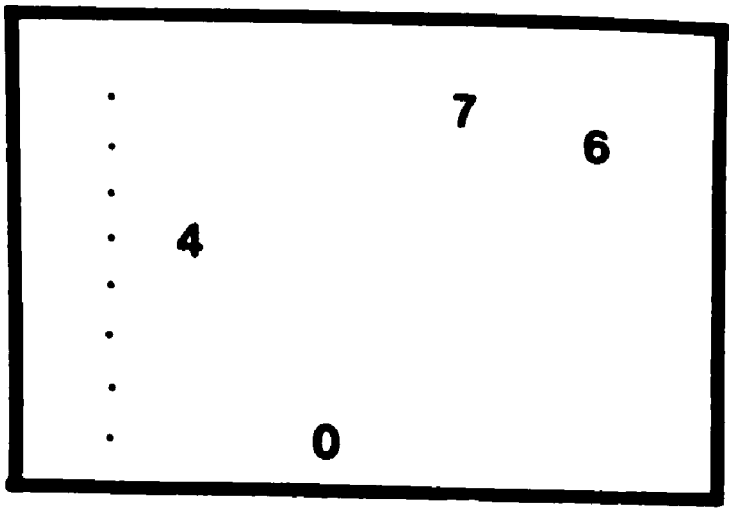


Figure 4.16.--Data Sampling Cycle



CONVERSION

$$4076[8] = 4 \cdot 8^3 + 0 \cdot 8^2 + 7 \cdot 8^1 + 6$$

4076[8] = 2110 MILLISECONDS

Figure 4.17.--Time Display

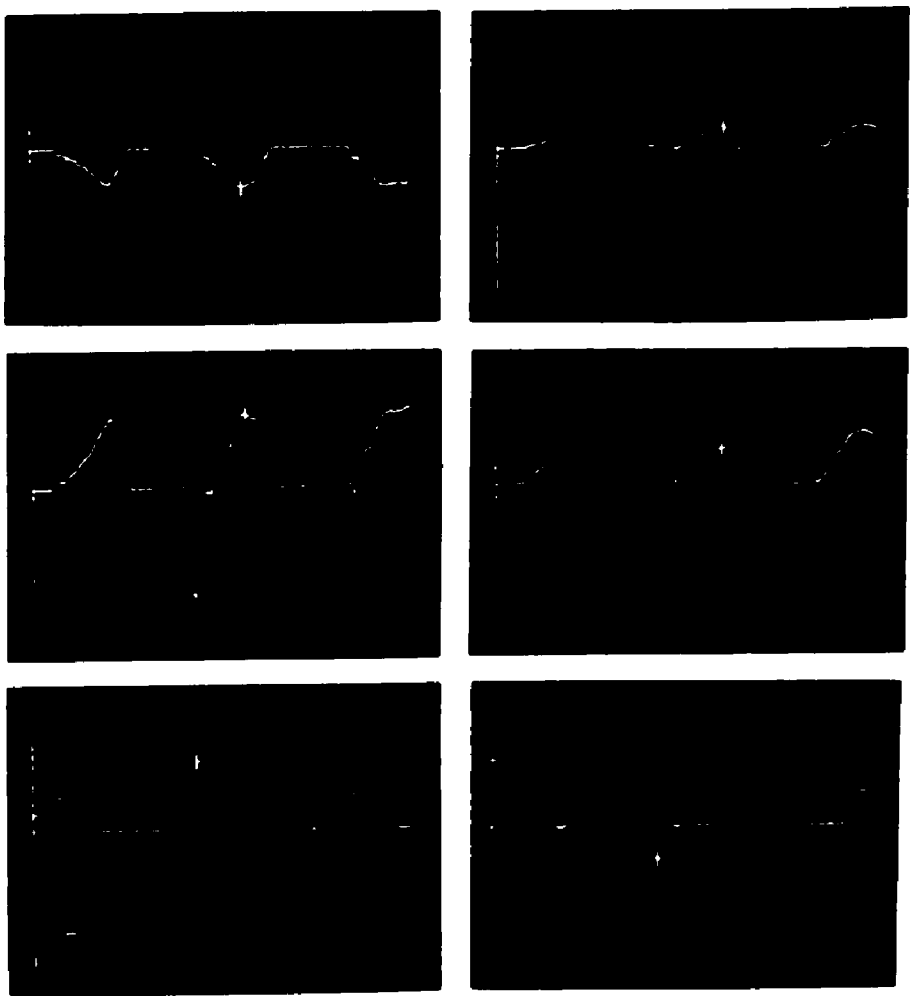


Figure 4.18.--First Interactive Graphics Routine

MEMORY BANK	QUARTER 0	QUARTER 1	QUARTER 2	QUARTER 3
7	PYLON FLEXION-EXT. MOMENT			RIGHT FOOTSWITCH DATA
6	PYLON TORSION			
5	PYLON ABDUCTION-ADD. MOMENT			
4	PYLON AXIAL FORCE			GASTROC-NEMIUS PRESSURE
3	LEFT FOOTSWITCH DATA			
2	PATELLAR TENDON PRESSURE			
1	SAMPLES			
0	PROGRAM OF OPERATION			

Figure 4.19.--Core Storage

CHAPTER FIVE

DATA REDUCTION

16 mm Film Analysis

The initial portion of the film analysis was one of the easist to neglect, yet one of the most important steps in the entire experiment. This was the developing of the film. Initially, the author took Sato's films to be processed through the Chemistry Store, The University of Michigan. Indeed, this was an unfortunate error. Processing took 4 to 6 weeks and the quality was not as good as it could have been (grainy).

After the amputees were photographed, the author decided to go to Custom Tech, Detroit, Michigan. It turned out that one of the owners was involved in a coal mine accident in his youth. He ended up with a hip replacement prosthesis. In contrast to the Chemistry Store, these films were developed in 24 hours and the quality was far better. This was due to the personal interest of the owners in this project and the normally excellent service they provide.

Raw data were obtained from the 16 mm pictures, frame by frame, by use of the Vanguard Motion Analyzer. It was manufactured by Vanguard Instrument Corporation, Melville, New York. The particular model used was M-160W.

This machine cast an enlarged image of each frame of 16 mm film onto a screen. X and Y crosshairs were available for determination of linear motion; a rotatable

grid, for determination of angular orientation. The rotatable grid was marked off in degrees rather than radians. A Vernier scale allowed resolution to quarter degree intervals.

The 16 mm film for the experiment was chosen double perforated specifically for its use in the Vanguard Motion Analyzer. The analyzer itself required only single perforation. However, the double perforation allowed flexibility in the manner in which the film was fed through the mechanism.

The analyzer lost its automated interface with a keypunch when International Business Machine switched from an 026 to an 029 model. This made it essential to record only the minimum amount of information to eventually yield the 37 time dependent coordinates for the differential equations of motion. This meant abandoning careful tracking of the Cartesian coordinate systems (Chapter Four). Instead, only the segmental frontal and sagittal projection angles and linear coordinates were measured.

After much thought, Tables 5.1, 5.2, and 5.3 were constructed. Table 5.1 is used in conjunction with Figure 4.6 to gather the left below knee amputee sagittal plane data. Table 5.2 is used in conjunction with Figure 4.7 to gather the right below knee amputee sagittal plane data.

The non-amputee was treated as a left below knee amputee for the purpose of gathering raw data from his sagittal plane and frontal plane films.

In Table 5.3, there is no difference in the basic pattern for gathering the frontal plane raw data from the left or right B/K or the non-amputee. A subtle difference was apparent in the numerical tabulation of this data for each subject. This occurred due to the change in subject orientation with respect to the global Cartesian coordinate system, Figures 4.6 and 4.7.

The method involved in taking these data was to follow one marker or one segment through the entire test, writing down one datum per frame for only that particular target. This allowed visual smoothing of the data as it was taken. It also was quite instructive, allowing the researcher to see what was happening to that particular measurement during the test. The alternate method was to take all required measurements on one frame then go to another. This was simply too tedious and it was not very instructive at all.

TABLE 5.1

LEFT B/K SAGITTAL PLANE RAW DATA

Measurement	Description
1	x(t)
2	z(t)
3	Thorax angle
4	Head angle
5	Left shoulder horizontal displacement
6	Left arm angle
7	Left forearm angle
8	Left hip horizontal displacement
9	Left thigh angle
10	Pylon angle
11	Right thigh angle
12	Right shank angle
13	Right foot angle

TABLE 5.2

RIGHT B/K SAGITTAL PLANE RAW DATA

Measurement	Description
1	x(t)
2	z(t)
3	Thorax angle
4	Head angle
5	Right shoulder horizontal displacement
6	Right arm angle
7	Right forearm angle
8	Right hip horizontal displacement
9	Left thigh angle
10	Left shank angle
11	Left foot angle
12	Right thigh angle
13	Pylon angle

TABLE 5.3

FRONTAL PLANE RAW DATA

Measurement	Description
14	y(t)
15	Thorax angle
16	Head angle
17	Left-right shoulder angle
18	Left arm angle
19	Left forearm angle
20	Right arm angle
21	Right forearm angle
22	Left-right hip angle
23	Left thigh angle
24	Left shank angle
25	Left foot toe-out distance
26	Right thigh angle
27	Right shank angle
28	Right foot toe-out distance

In addition to each subject's motion, data was needed concerning the position of the global Cartesian coordinate system with respect to the cameras. Included in this was the scale factors of linear units of measure on the Vanguard Motion Analyzer versus the actual units of measure along the global Cartesian coordinate system.

This problem was not exactly simple. The lens of each camera possessed a finite focal length. This gave the effect of a vanishing point phenomena in each frame of film.

It was necessary to correct for this vanishing point phenomena in the frontal plane raw data, Table 5.3. Observing the film showed each subject to grow larger as he approached the frontal plane camera. This growth was only apparent.

The sagittal plane camera gave the subject's distance from the frontal plane camera. This monitoring of the forward progression of the subject allowed correction for the vanishing point phenomena as filmed by the frontal plane camera. Sears (1958) was helpful in analytically describing the required linear relationship.

Although the vanishing point phenomena was visible in the sagittal plane camera, no correction was made for it. This was reasonable due to the extremely small change of subject to camera distance during each test.

In summary, a constant scale factor was used to relate the linear units of measure on the Vanguard Analyzer

to the actual units of measure in the sagittal plane. In the frontal plane, a scale factor linearly dependent on the forward progression of each subject was used to relate the linear units of measure on the Vanguard Analyzer to the actual units of measure in the frontal plane. Rather than evaluate these scale factors for each frame of film, they were evaluated for the first frame of film in each test and assumed constant for the duration of that test. This simplification explained why fixed-focal-length lenses were chosen over zoom (variable-focal-length) lenses.

Reduction of the raw data began with the knowledge that the S.A.C.H. foot was mounted normal to the axis of the instrumented pylon. Having gathered the sagittal plane angle of the pylon meant a simple arithmetic operation was all that was necessary to obtain the S.A.C.H. foot sagittal plane angle.

Assumptions had to be made about the portions of the body invisible to the sagittal plane camera.

The sagittal plane angles of the shoulder, arm, and forearm, which were invisible to the sagittal plane camera, were assumed equal to the corresponding visible limb angles with the inclusion of a phase angle in time.

Determination of the upper limb phase angle was by assuming it equal to the phase angle between appropriate lower limbs, namely the thighs.

No assumptions of symmetry of thigh angles were made. However, the thigh angle curves were not altogether

dissimilar. This, coupled with the fact that upper limbs tend to move 180 degrees out of phase with the corresponding lower limb, allowed manual ascertainment of the phase angle between visible and invisible upper limb segments. One of the final assumptions rested with the connectivity of the pelvis. Although modeled as two independent halves, the pelvis was rigid. This fact was accounted for by input of appropriate angular information to make the model's independent halves act as one rigid member.

Included in this approximation was that the portion of the ilium not visible to the sagittal plane camera was 180 degrees out of phase with the visible one.

In the frontal plane, it was assumed that the left shoulder frontal angle was equal and opposite to the right shoulder frontal angle. This was reasonable, since during the tests none of the three subjects ever moved the shoulders as independent entities.

Actual calculation of the thirty-seven coordinates for the differential equations of motion proceeded as follows. First of all, the vector (x(t),y(t),z(t)) from the global Cartesian coordinate system to the lumbosacral region was calculated via raw data from observing the pelvic target and appropriate scale factors.

The next calculations involved converting raw hip and shoulder motion into the angular coordinates desired. Figure 5.1 gives the relationship between the sagittal

plane horizontal displacement (SPD) of a shoulder about the sterno-clavicular joint, the shoulder's frontal plane projection angle, length, and the two spherical coordinates used by the model.

Figure 5.2 gives the relationship between the sagittal plane horizontal displacement (SPD) of a hip about the lumbosacral region, the left-right hip frontal plane projection angle, length, and the two spherical coordinates used by the model. The vector R(24) between the lumbosacral interface and the line passing through both hip joints is used in this figure. Values for this vector's magnitude were specified in Chapter Three.

It was interesting to note that Figure 5.2 degenerated into Figure 5.1 when R(24) assumed the magnitude of zero.

A figure similar to Figure 5.1 was used for the foot spherical angles. However, for the feet, the raw data consisted of the sagittal plane angle and the frontal plane horizontal toe-out distance.

The final calculations, Figure 5.3, related the frontal plane projection angle, the sagittal plane projection angle, and the conical angle for the thorax, head, left arm, forearm, thigh, and shank as well as the right arm, forearm, thigh, and shank.

No corrections were needed for the sagittal plane projection angles, since this subset of angles was already one of the two subsets of spherical coordinates.

Furthermore, no correction for parallax was required in either plane, due to the favorable subject to camera distances.

Variability among observers was avoided by having only one skilled operator do all of the Vanguard analysis. However, this did not preclude variability within the one skilled observer. This became possible because the joint markers were not dimensionless points. Instead, they were hemispheres with a non-trivial, 12 mm radius. Furthermore, these markers were moving during the exposure of each frame of film.

Reduction of marker blur was achieved by partially closing each sagittal camera shutter, Table 4.9. When blur did occur, especially in the swing phase, the middle of the blur was used for the measurement.

Assuming a 0.500 m shank (Tables 3.5 through 3.7) and a 1 radius error in one of the two joint markers due to the observer's not being able to locate the center of the hemisphere, the angular error is 12/500, or 0.024 radians. This is about 1.3 degrees, close to the 1 degree value of possible error given by Sato (1975).

The whole idea of tracking one marker or segment at a time during the entire test was to observe for this error. For instance, if all of a sudden a measurement jumped by 2 degrees then remained stationary, when the observer knew it was moving all along, it was possible to go back and check what went wrong. This does not mean that

the data was tampered with. It simply meant that the observer was conscious of the generally smooth motion which occurs during human ambulation.

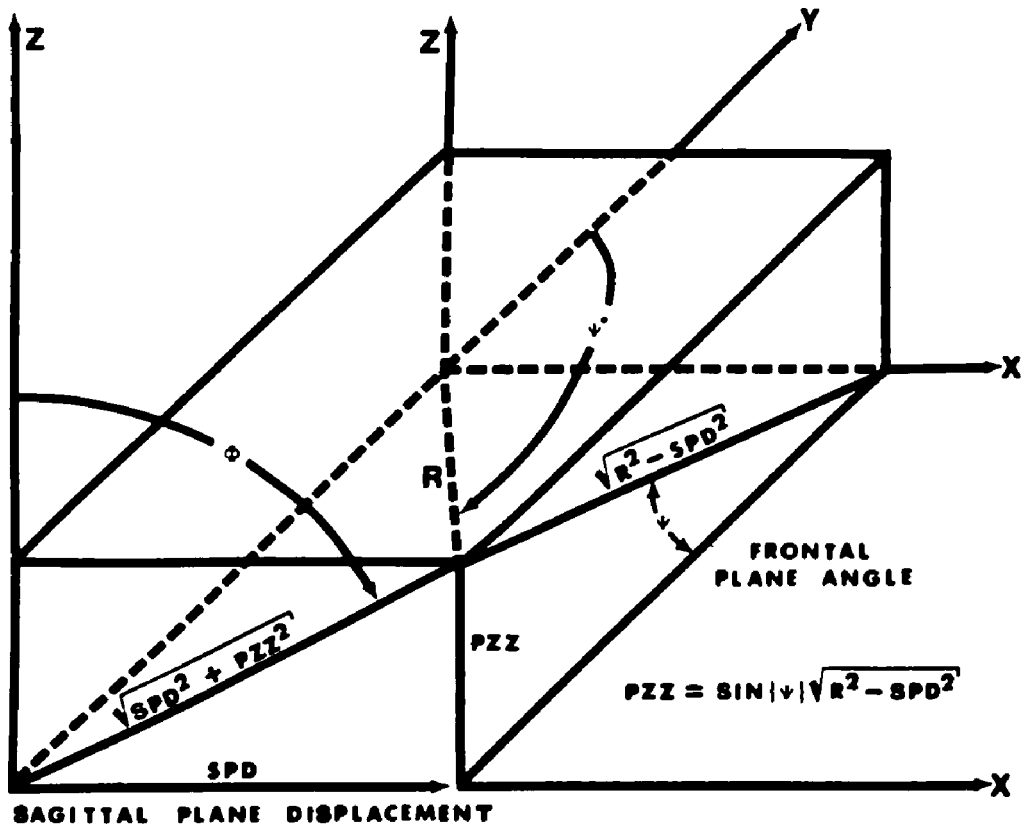


Figure 5.1.--Shoulder Spherical Coordinates

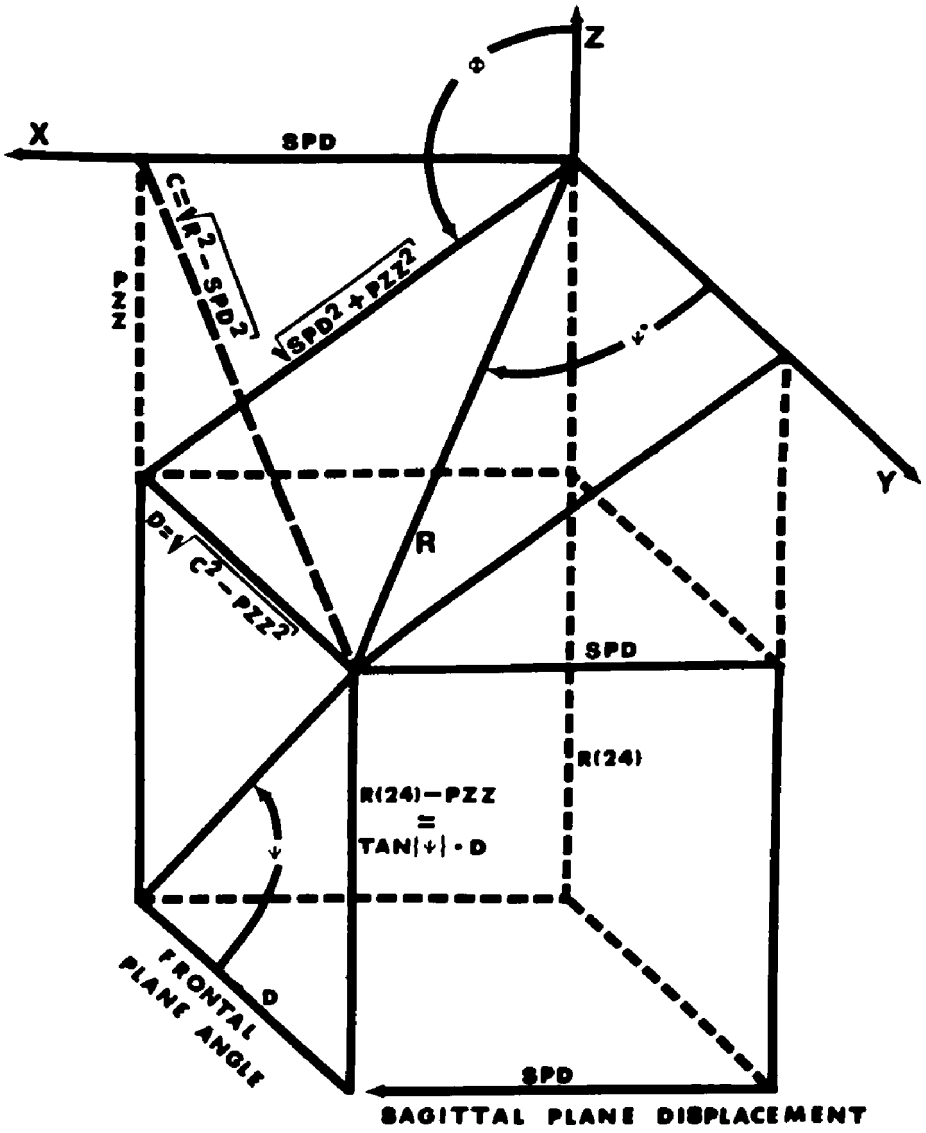


Figure 5.2.--Illum Spherical Coordinates

$$\tan(\psi^{\circ}) = \cos(\Phi) \cdot \tan(\psi)$$

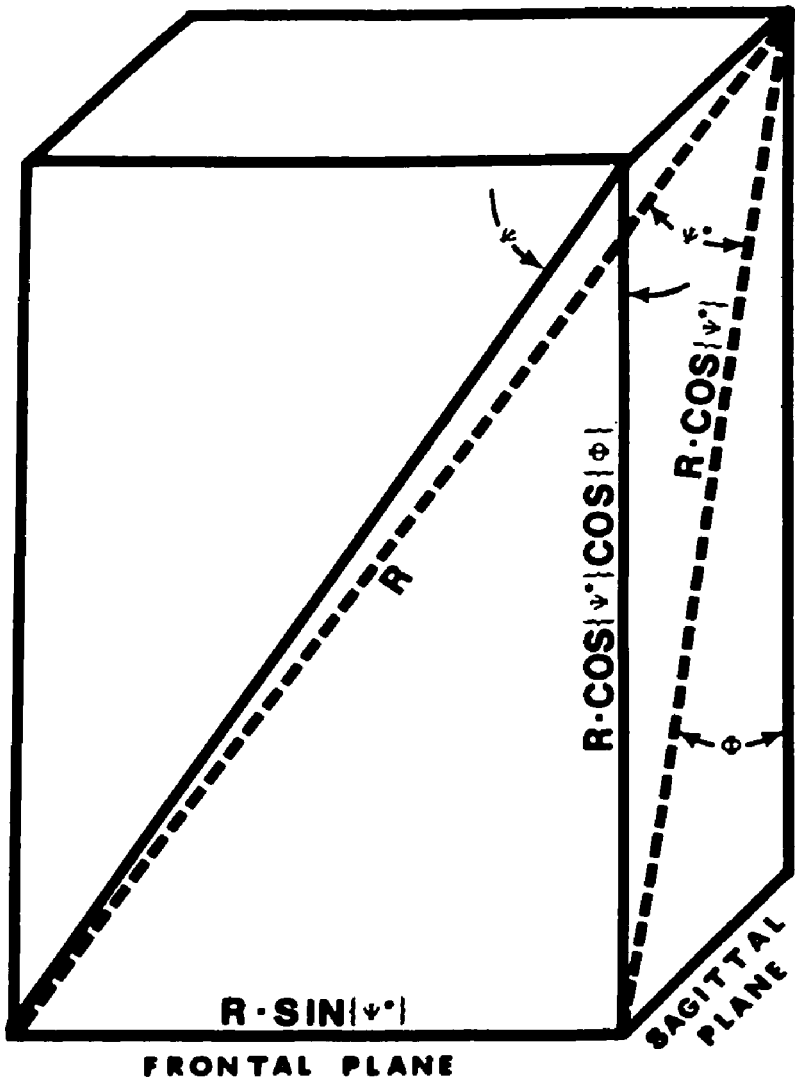


Figure 5.3.--Frontal Plane versus Conical Angles

Storage of Linc-8/PDP8 Data

Storage of the experimental data digitized by the Linc-8/PDP8E mini-computer was done initially on 12 bit Linc tape. A bit denoted one binary digit.

The Linc tape had two halves, each half formatted into 512 blocks. A block of data was the same size as a quarter of memory; 256 entries, each entry or word containing 12 bits, or 12 binary digits. Since three quarters of memory were devoted to each data channel, three blocks of magnetic tape were similarly devoted. In base 8 arithmetic, 3 blocks of data per channel times the number of channels gave 30 to the base 8, 30(8), blocks of data per test. This was the test number increment mentioned in Figure 4.16. By setting the initial test number and initial data tape number the same and incrementing them by always the same number, the experimenter was able to keep track of his initial storage of experimental data.

From 12 bit Linc tape, the data was transferred to 16 bit DEC tape. Extension of the 12 bit number to 16 bits was done by shifting of the sign bit. Block size was the same, 256 words per block.

Although the DEC tape had fewer blocks per reel, 578 in all, this did not interfere with the data transfer. Data was transferred block for block in the same order as it was stored originally on Linc tape.

The final transfer occurred when the hexidecimal entries on the DEC tape were transferred to an unlabeled

tape for use on the Michigan Terminal System, Figure 5.4. The tape itself was nine track: one track for a parity bit; the other 8, for data.

Each set of 8 bits was called a byte. This meant the hexidecimal data required 2 bytes per datum.

The Highway Safety Research Institute did the DEC to MTS transfer. Although the tape was rated at 1600 bytes per inch, B.P.I., the data were stored at 800 B.P.I. One minor problem was that the pair of bytes per datum somehow were reversed. This necessitated a small subroutine for untangling the bytes.

Program access to the unlabeled tape was by mounting it, giving it a title, such as *TAPE*, and by referencing the tape title with an input/output unit. An example was 8=*TAPE*

Before the tape was read, it had to be positioned to the correct test number. Since the tests were stored sequentially, this was a stralght forward task.

Reading the tape was done with an A (alphabetic) format. The data were stored in a matrix declared to be a 2 byte integer type. This storage matrix had 8 rows, one for each Linc-8 sampling channel, and 768 columns, one for each sampling period.

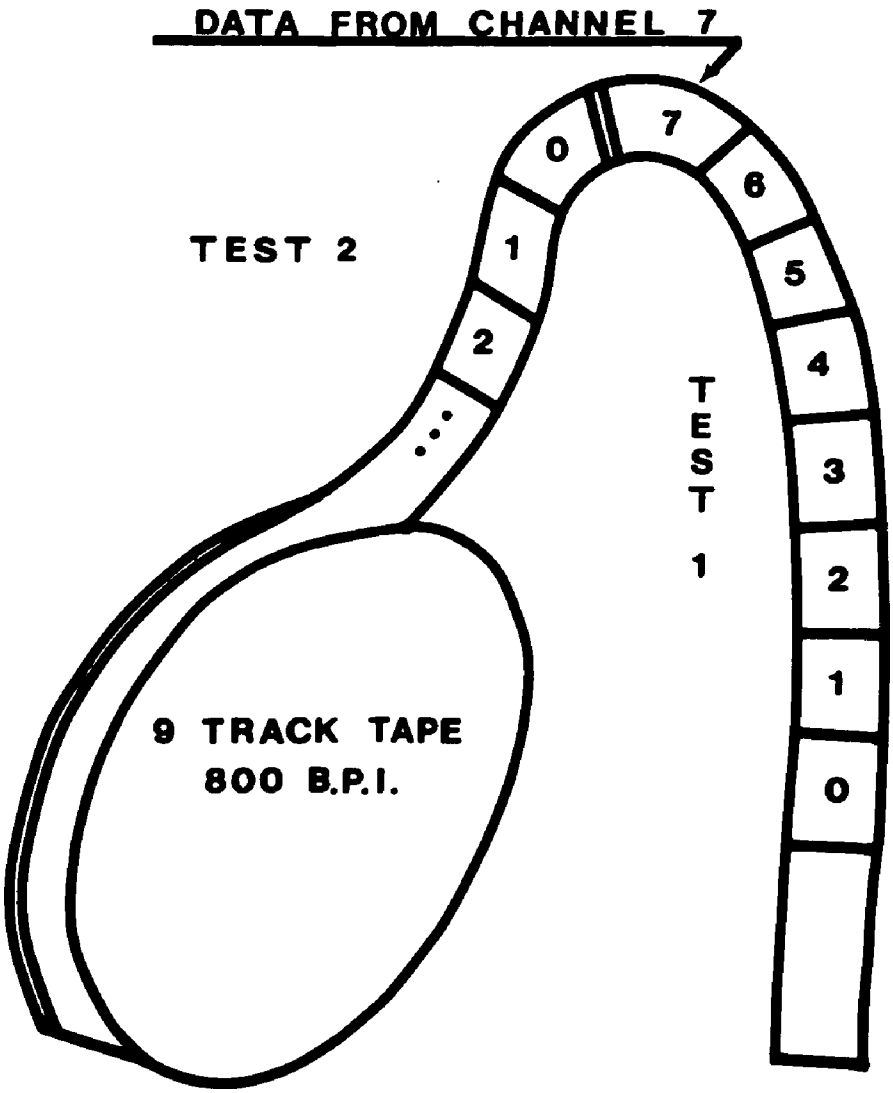


Figure 5.4.--M.T.S. Unlabeled Tape

Mathematical Techniques

The mathematical techniques involved in obtaining the first and second derivatives on the experimental displacement data and in determining the cumulative energy expenditure during the gait cycle were handled by appropriate numerical analysis methods in International Business Machine's System/370 Scientific Subroutine Package, *SSP. Figure 5.5 shows the three types of techniques used.

The displacement data were smoothed locally before each of two differentiations. Each locally smoothed datum point was obtained by evaluating at x(I), the least squares polynomial of degree 1 relevant to x(I-1), x(I), and x(I+1).

Except for the end points x(1), x(2), x(N-1), and x(N), the locally evaluated derivative at x(I) was the derivative of the Lagrangian interpolation polynomial of degree four evaluated at x(I-2), x(I-1), x(I), x(I+1), and x(I+2).

Sufficient data were collected in order to be able to disregard five points at the beginning and end of each interval. This was the amount suggested by differentiating known functions and comparing the Lagrangian evaluated derivative and the actual derivative.

The truncation error inherent in this particular differentiation process was proportional to the fifth derivative of the function. All gait cycle displacement

data were taken pointwise. Since these functions were not known in closed form, it was not possible to evaluate the truncation errors.

The sequence of smoothing and differentiating displacement data was evaluated on pointwise representation of known closed form functions. Comparisons of the pointwise functions and their first and second derivatives with the actual functions showed errors to be within 10 percent. This comparison was done at 20 and 40 Hertz sampling frequencies.

Similar techniques of obtaining derivatives of the smoothed displacement data were used by Bender (1972).

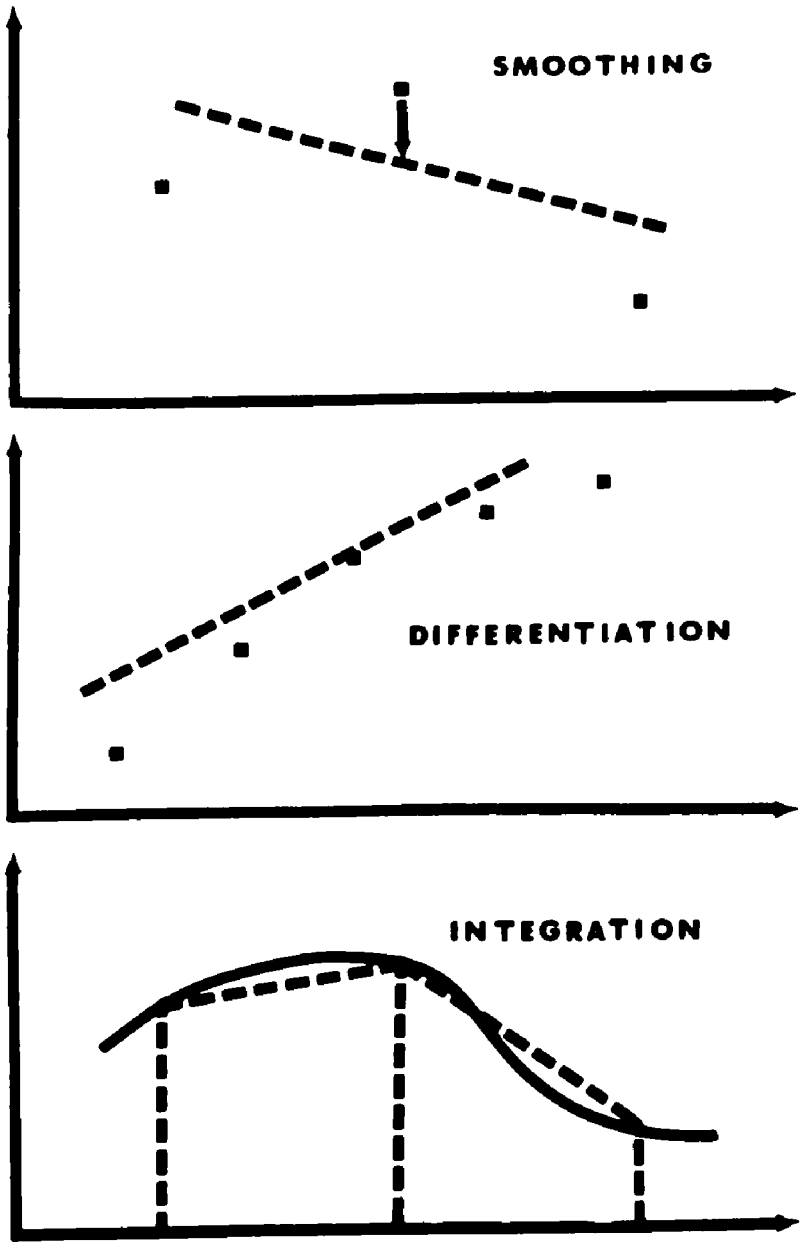


Figure 5.5.--Three Mathematical Techniques

Computer Software

This section was devoted to a naming of the various Michigan Terminal System (M.T.S.) programs used in the analysis, their descriptions, and their lengths. All of the below were written in single-precision Fortran IV for analysis on the Amdahl 470V/6. This computer replaced the slower I.B.M. 370/168 previously used. It was interesting to note that the 370/168 model lasted only one year after replacing the I.B.M. 360/67 once used by M.T.S.

MAIN: This was the central program which queued the seven subroutines, ARRAY, INTERP, FIGURE, ACC, CASE, TORQUE, and GAGE. When appropriate, it called upon the four *PLOT subroutines (MTS Volume III) for graphical output. Finally, this program did the classical definition of work calculations which yielded the cumulative energy expenditure surfaces of Chapter Seven.

There were various possibilities in evaluating this measure of energy expenditure. Vukobratovic, et al. (1973) calculated power expenditure as the integral of joint moment with respect to angular velocity. Work would then be the integral of this power with respect to time. Beckett and Chang (1968) solved for this mechanical work as the product of moment and limb relative

angular velocity integrated with respect to time.

For this research, it was decided to apply the classical definition of work directly and not involve angular velocity. It was felt that using direct experimental displacement data, rather than a calculated derivative, made the results less error prone.

There were two completely equivalent ways to write the required expression. One was to write the integrals of the joint torques with respect to relative angular displacement. The other was to integrate the segment action torques on the proximal end of each segment minus the segment reaction torques on the distal end with respect to the absolute angular motion of the particular segment. The difference between these two approaches was where one would like to subtract, either in the integrand or in the differential. The answers, of course, were the same in either case.

The method of segment torques actions minus re-actions integrated with respect to the absolute angular motion was chosen. Again, this was so because of its ease of being programmed.

The actual calculations involved determination of the difference between the action torque exerted on the proximal end of each segment

and the re-action torque exerted on the distal end. This was done for all segments, with the following exceptions. Since each S.A.C.H. foot connection was mechanically rigid, it was omitted from the calculations. Furthermore, pelvic rigidity was accounted for in the work equations.

These torques were expressed with respect to the global Cartesian coordinate system.

Next, the conical (PSI) and sagittal (PHI) absolute angular displacements were expressed in global Cartesian coordinate system components. In particular, the X, Y, Z components were:
(Cos(PHI)*PSI, PHI, -Sin(PHI)*PSI).

The final step was application of the trapezoidal rule through subroutine QTFC for the integral of each component of torque with respect to the appropriate component of absolute angular motion. The appropriate component was determined by application of the dot product.

It was 535 Fortran statements long.

Subroutine ARRAY: ARRAY handled the calculation of the 37 time dependent coordinates from the raw data described in Chapter Six.

Input to this subroutine was in a strict line format.

Line 1. The start and finish time of the sagittal

plane camera then the start and finish time of the frontal plane camera was first.

Line 2. The pylon flexion-extension angle for the left leg then for the right leg was next. Zero degrees was inserted if the leg was "normal" or if the alignment was in the neutral position.

Line 3. The next three numbers denoted the location of the X-axis datum, the Y-Axis datum, and the Z-Axis datum (Chapters Four and Five).

Line 4. This line was reserved for the linear scale factors which corrected Vanguard analyzer measurements to inertial reference frame measurements. The need for these factors was discussed in Chapter Five.

Line 5. The number of frames of frontal plane film and the number of frames of sagittal plane film in the analysis were entered here.

Line 6. Footswitch data denoting each time of left foot heel-strike and toe-off.

Line 7. Footswitch data denoting each time of right foot heel-strike and toe-off.

Line 8. Sagittal plane data were listed until exhausted. Then the frontal plane data were listed until exhausted.

The actual calculations performed in ARRAY were discussed earlier in this chapter.

ARRAY was 384 Fortran statements long.

Subroutine INTERP: This linear interpolation subroutine was called from ARRAY to coordinate the raw data from the frontal and sagittal plane cameras. This was required for two reasons. Each camera started at different times. Furthermore, the "selected" speed of each camera and the "actual" running speed were not the same. It was assumed each camera speed was an undetermined constant during the four second duration of each test. This allowed taking into account that each camera ran at a slightly different speed in each test.

INTERP was 33 Fortran statements long.

Subroutine FIGURE: This plotted the joints (J), center (C) of masses, and end points (X) of the human body. It also connected the joints and end points (*) to complete the representation of the human body as a stick figure. The final notation was (P) for a B/K prosthesis, if one was worn by the subject. This denoted the pseudo-14th joint (the alignment unit) mentioned in Chapter Three.

The primary motivation behind this symbolism was the DRAW program documented by Chace and Angell (1973). The final calculations were of the instantaneous potential and kinetic energies of the model.

The potential energy of each segment was

calculated as a function of time with respect to the Z-axis datum, the floor. The sum of the segmental potential energies was taken to yield the whole-body potential energy as a function of time.

The product of each segment's mass and square of the magnitude of the linear velocity vector of its center of mass gave the kinetic energy of each segment. It should be noted that the lumped mass model precluded kinetic energy due to the rotation of the segment about its center of mass. The whole-body kinetic energy was then obtained by summing its segmental kinetic energies. The actual calculation of the magnitude of each velocity vector was done in two steps. First, the chain rule was applied to the position vectors describing the center of mass and distal end of each segment with respect to the proximal end. Then these relative velocities were appropriately summed to yield the absolute velocity of each segment with respect to the inertial reference frame.

These absolute velocities were expressed in X, Y, and Z components, so the square of the magnitude of each velocity vector was the sum of the square of its components.

This subroutine was 409 Fortran statements long.

Subroutine ACC: The ACC subroutine calculated the segment center of gravity accelerations with respect to time and the inertial reference frame. This data was required input for the differential equations.

The actual calculations were done in three steps. The first involved calculating the acceleration of each joint and segment center of mass with respect to the proximal end of its segment. In the listing of the iteration below, D refers to the first derivative with respect to time; D2, the second derivative. The formulas used were those for acceleration in spherical coordinates, Easthope (1964).

```
SPHI(N) = SIN(PHI(N))
CPHI(N) = COS(PHI(N))
SPSI(N) = SIN(PSI(N))
CPSI(N) = COS(PSI(N))
AR(N)       =       -R(N)*(DPSI(N)*DPSI(N)          +
SPSI(N)*SPSI(N)*DPHI(N)*DPHI(N))
APSI(N)          =          R(N)*(D2PSI(N)          -
SPSI(N)*CPSI(N)*DPHI(N)*DPHI(N))
APHI(N)    =   R(N)*(2.*CPSI(N)*DPSI(N)*DPHI(N)     +
SPSI(N)*D2PHI(N))
ARC(N) = AR(N)*RC(N)/R(N)
APSIC(N) = APSI(N)*RC(N)/R(N)
APHIC(N) = APHI(N)*RC(N)/R(N)
```

The second step was to express these relative accelerations in X, Y, Z components.

```
AX(N) = AR(N)*SPSI(N)*SPHI(N) + APHI(N)*CPHI(N) +
APSI(N)*CPSI(N)*SPHI(N)
AY(N) = AR(N)*CPSI(N) - APSI(N)*SPSI(N)
AZ(N) = AR(N)*SPSI(N)*CPHI(N) - APHI(N)*SPHI(N) +
APSI(N)*CPSI(N)*CPHI(N)
AXC(N) = AX(N)*RC(N)/R(N)
AYC(N) = AY(N)*RC(N)/R(N)
AZC(N) = AZ(N)*RC(N)/R(N)
```

The final step was to express the absolute acceleration of each segment center of mass as a summation of relative accelerations, as discussed in Craig (1943) or Easthope (1964).

ACC was 40 Fortran statements long.

Subroutine CASE: This subroutine determined the leg configuration as required by the differential equations of motion in 96 Fortran statements.

The input to this subroutine consisted of the actual time that heel-strike and toe-off occurred for the left foot and then for the right foot. This timing was obtained from the footswitches as discussed in Chapter Four.

Subroutine TORQUE: Once having derived the differential equations, this subroutine calculated the joint torques and forces, the ground reaction forces and their center of pressure on each foot, as well as the theoretically exerted pylon loads.

The first portion of this program found the X, Y, and Z components of force acting on each segment's center of mass as a function of gravity and inertial acceleration. Floor reactions were initially ignored.

Next, the torques at each joint were calculated using the cross product of displacement and force. Again, floor reactions were omitted. These calculations, in inertial reference frame components, were:

```
TX(N)      =      RC(N)*(FCZ(N)*CPSI(N)      -
FCY*SPSI(N)*CPHI(N))   +   R(N)*(FZ(N)*CPSI(N)   -
FY(N)*SPSI(N)*CPHI(N))
TY(N)      =      RC(N)*(FCX(N)*SPSI(N)*CPHI(N)      -
FCZ(N)*SPSI(N)*SPHI(N))                              +
R(N)*FX(N)*SPSI(N)*CPHI(N) - FZ(N)*SPSI(N)*SPHI(N))
TZ(N)      =      RC(N)*FCY(N)*SPSI(N)*SPHI(N)      -
FCX(N)*CPSI(N))   +   R(N)*(FY(N)*SPSI(N)*SPHI(N)   -
FX(N)*CPSI(N))
```

These calculations were used to solve for

the forces and torques which had to be exerted on the instantaneous whole-body center of mass to satisfy equilibrium requirements. Of course, these required forces and torques were supplied by the floor reactions. This enabled their calculation.

Having calculated the floor reactions and their center of pressure on each foot, it was now possible to include them in the final joint torque calculations.

Once these joint torque calculations were finished, it was possible to use the theoretical floor reactions, their center of pressure (c.p.) on the S.A.C.H. foot, and the location of the strain gages with respect to the c.p. to calculate the theoretical pylon forces and torques. These were the loads exerted on the pylon at the strain gages with respect to the inertial reference frame. From here, it was necessary to rotate the reference system to that of the pylon. A direction cosine matrix accomplished this as described in Chapter Six.

Overall, 535 Fortran statements were used.

Subroutine GAGE: Written to compare the theoretical and the empirical pylon loads, GAGE had many tasks. First of all, the raw empirical data from the MTS unlabeled tape had to have its bytes unscrambled,

Chapter Five. Then the raw data was zeroed and the appropriate conversion factors applied to give the pylon empirical loads, Chapter Four. Finally, the theoretical pylon loads as calculated by TORQUE were statistically compared to the empirical loads.

This necessitated calculating the pylon loads as pointwise functions of the same time that the motion picture camera data was gathered. The computer sampling rate being much faster than the 16 mm film speed required this action.

The error was calculated as the difference between the theoretical and empirical load. It was then normalized by the largest magnitude of the particular empirical data being studied and expressed as a percentage.

The statistical analysis was formulated by assuming that the normalized error formed a Gaussian Distribution. The mean and standard deviation of the normalized errors were then calculated according to established formulas. The standardized normal variate was then used to relate mean, standard deviation, and confidence level.

This subroutine was 277 Fortran statements long.

The compiled versions of the above programs were

stored in sequential files and concantinated into one large file when the analysis was performed.

Line files were used as input files for the anthropometric coefficients, Chapter Two, and for the raw data gathered from the Vanguard Motion Analyzer, Chapter Five.

CHAPTER SIX

MODEL VALIDITY

Four Criteria

Establishing the degree of applicability of the differential equations was based on four criteria adapted from Bender (1972). These criteria came as no surprise, since they were a summary of the basic thought pattern of the modeling process, Figure 1.3. This redundancy served as a valuable check list, so that in the heat of the battle, no stones were inadvertantly left unturned.

1. The physical approximations from which the model was derived rested on sound assumptions of the physical reality. These assumptions, discussed in Chapter Three, were that the human body was composed of seventeen rigid segments. This eliminated the need of the often used HAT (head-arms-trunk), which many researchers claimed was a reasonable approximation of the entire upper half of the human body. Furthermore, the ball joint approximation to the human joint was acceptable. This was mostly due to the fact that the analysis used real-life motion (kinematics) as input to the differential equations.

   If a dynamic approach would have been undertaken, ball joints would have been totally

unacceptable.

It is interesting to note, as shown above, how the intent of the analysis affected the acceptability of the assumptions.

2. The mathematical model, as described in Chapter Three, was correctly derived by the TORQUE subroutine. The numerical analysis involving smoothing of data, double differentiation of displacement data, and numerical integration of torques and angular displacements was suitable. Finally, that the computer software which used the above quantities was thoroughly debugged.

3. Appropriate experimental data were obtained on which the degree of model applicability could be determined. These experimental data consisted of the four instrumented pylon loads (Chapter Four). The loads were the flexion-extension bending moment, the pylon axial torsion, the abduction-adduction bending moment, and the pylon axial force.

4. A suitable correlation between theoretical and empirical values existed. This fourth criterion encompasses the remainder of this chapter.

Model Predictions

Since the input to the differential equations of motion was each subject's motion, the output was the joint torques and floor reactions required to have produced that motion.

The floor reactions calculated by the model were the vertical, anterior-posterior, and medial-lateral forces exerted on each foot as a function of time. Included in this calculation was the center of pressure of the vertical force distribution on each foot. It was assumed that the anterior-posterior and medial-lateral force distributions also had their center of pressure at the vertical force center of pressure.

The torsion couple exerted on each foot during the gait was not calculated. This torsion calculation turned out to be the difference between two large quantities, the body's rotations about the vertical axis and the torsion induced by the medial-lateral force. It was felt that even with good accuracy in the calculation of both quantities, that the difference, the torsion couple, could have been well in error. For this reason, the torsion couple calculation was never done.

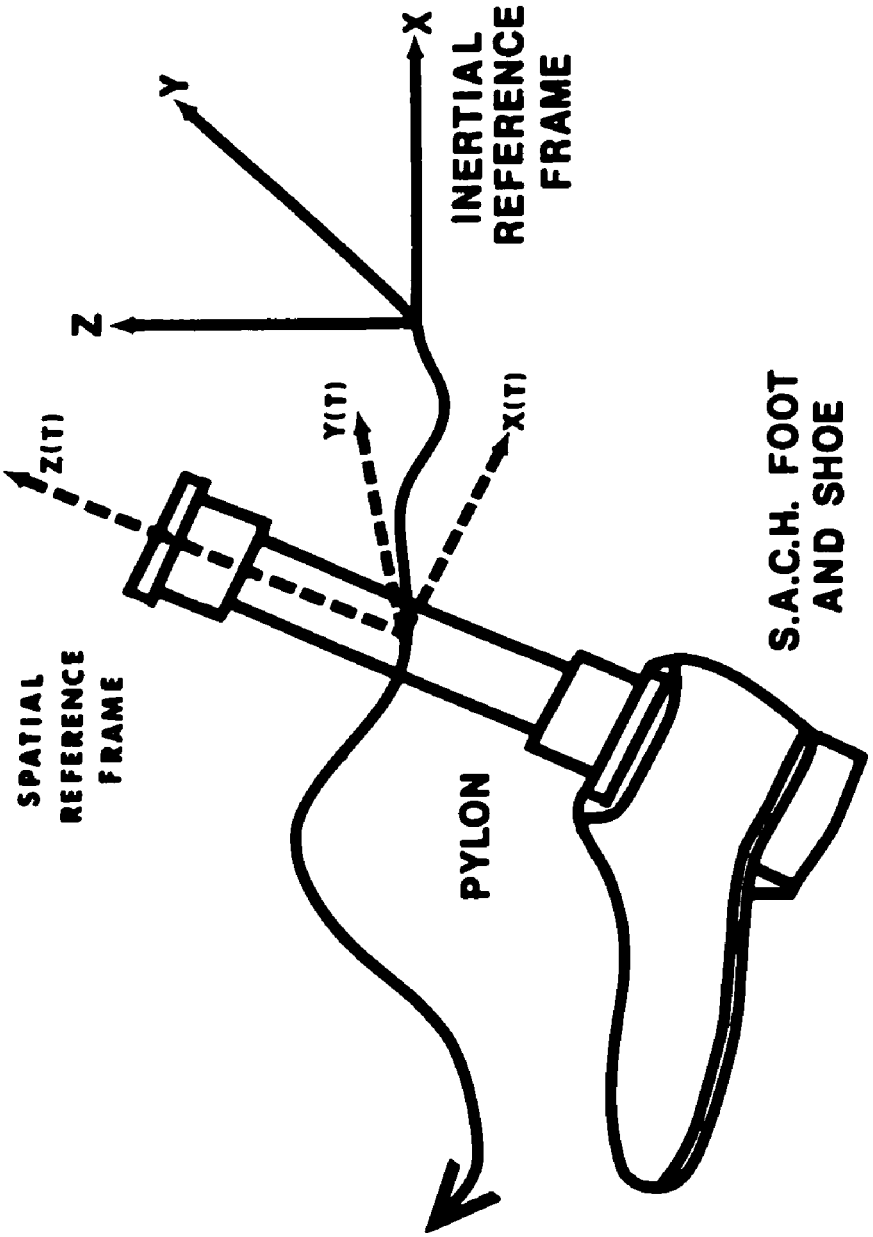


Figure 6.1.--Pylon Spatial Reference Frame

Once the floor reaction forces and their point of application were calculated for each foot as a function of time, it became necessary to calculate the forces and torques theoretically exerted on the pylon at the strain gage location. This required specification of the pylon strain gage location. Segment 22 did this for a left B/K; segment 23, a right B/K. This distance was from the mean strain gage location to the ankle. A zero entry was inserted if the leg was "normal." This also meant the orientation of a pylon spatial reference frame had to be known with respect to the inertial reference frame.

The orientation of the pylon spatial reference frame was a function of the four spherical coordinates used to specify the orientation of the pylon and the S.A.C.H. foot. The pylon itself became the spatial z(t) axis, Figure 6.1. The x(t) spatial axis was chosen normal to the pylon and parallel to the S.A.C.H. foot heel-toe line. The spatial y(t) axis was then formed by the vector cross product of the z(t) and x(t) spatial axes.

Once the orientation of the spatial reference frame was determined it was necessary to solve for the transformation matrix between vectors in the inertial and in the spatial reference frames.

Following Craig (1943), the appropriate matrix was derived in terms of direction cosines.

The floor reactions were specified with respect to the inertial reference frame. Then the force and torque

vectors were solved for at the strain gage location with respect to the inertial reference frame. Finally, the direction cosine matrix, expressed as a function of time, transformed these vectors to the pylon spatial reference frame.

The transformed vectors were the theoretical loads exerted on the pylon.

It should be noted that Figure C.1 applied to only the left B/K. The orientation of the S.A.C.H. foot and shoe were reversed for the right B/K. A subtle point is that the spatial reference frame did not rotate with respect to the inertial reference frame when the change was made between the left and right B/K. Even though during this transfer the subject rotated above the pylon and the S.A.C.H. foot rotated below the pylon, no reversal of pylon orientation occurred. This greatly simplified the derivation of conversion factors for pylon Wheatstone bridge electrical outputs to pylon load values.

Theoretical-Empirical Comparisons

The theoretical pylon loads agreed quite well with the empirically measured values. The standard applied was that at 80% confidence the normalized error would be within 20%. The abduction-adduction and flexion-extension bending moments met this standard. The axial compressive load exceeded it, with a 95% to 99% confidence that the normalized error would be within 20%.

There was not such agreement with the axial torsion load. Typically, the theoretical torsion load was half the empirical value. This was due to neglecting the torsion couple floor reaction, Chapter Six.

Figures 6.2 through 6.6 were prepared to show the typically good agreement between the theory and experiment. True, the axial torsion calculation was half what it should have been, but when the error in this component was compared to the magnitude of the torque vector, it appeared negligible.

For these reasons, the model was described as a non-linear, algebraic transformation capable of an appropriate kinematic analysis.

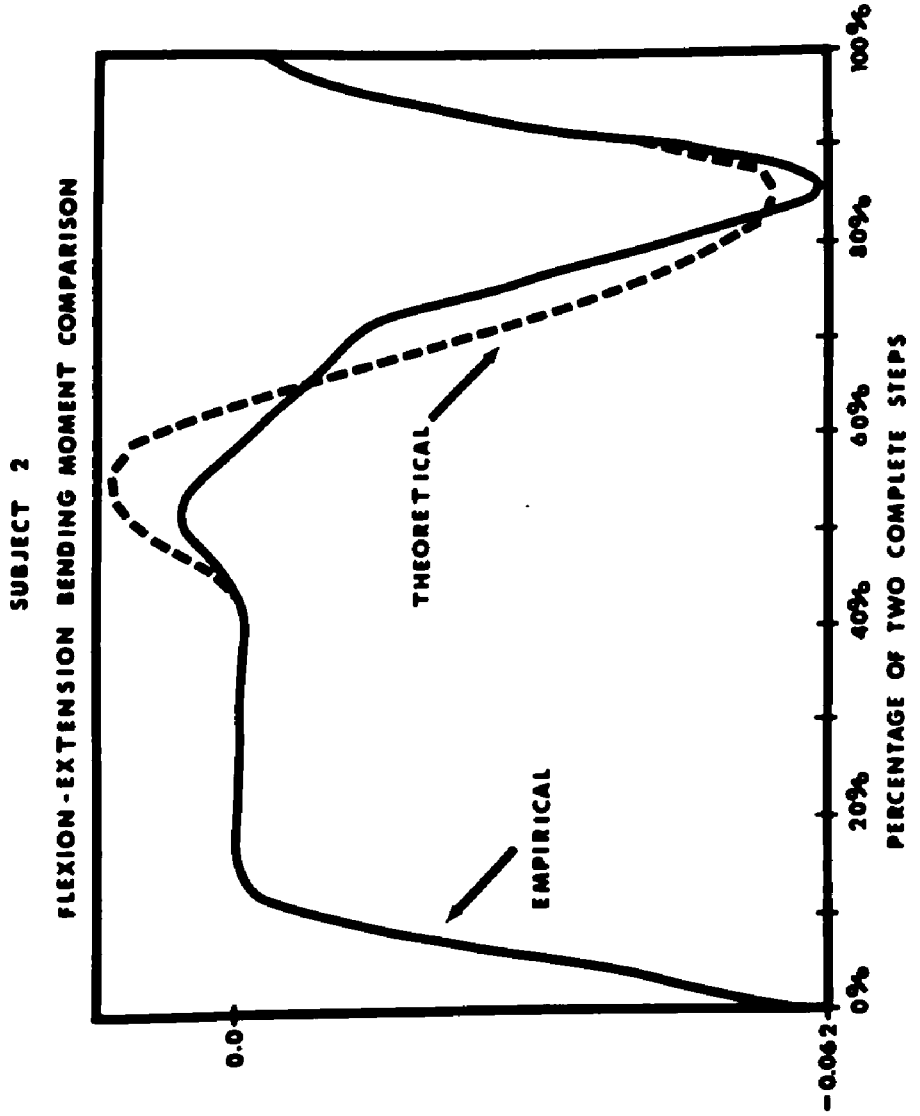


Figure 6.2.--Flexion-Extension Moment

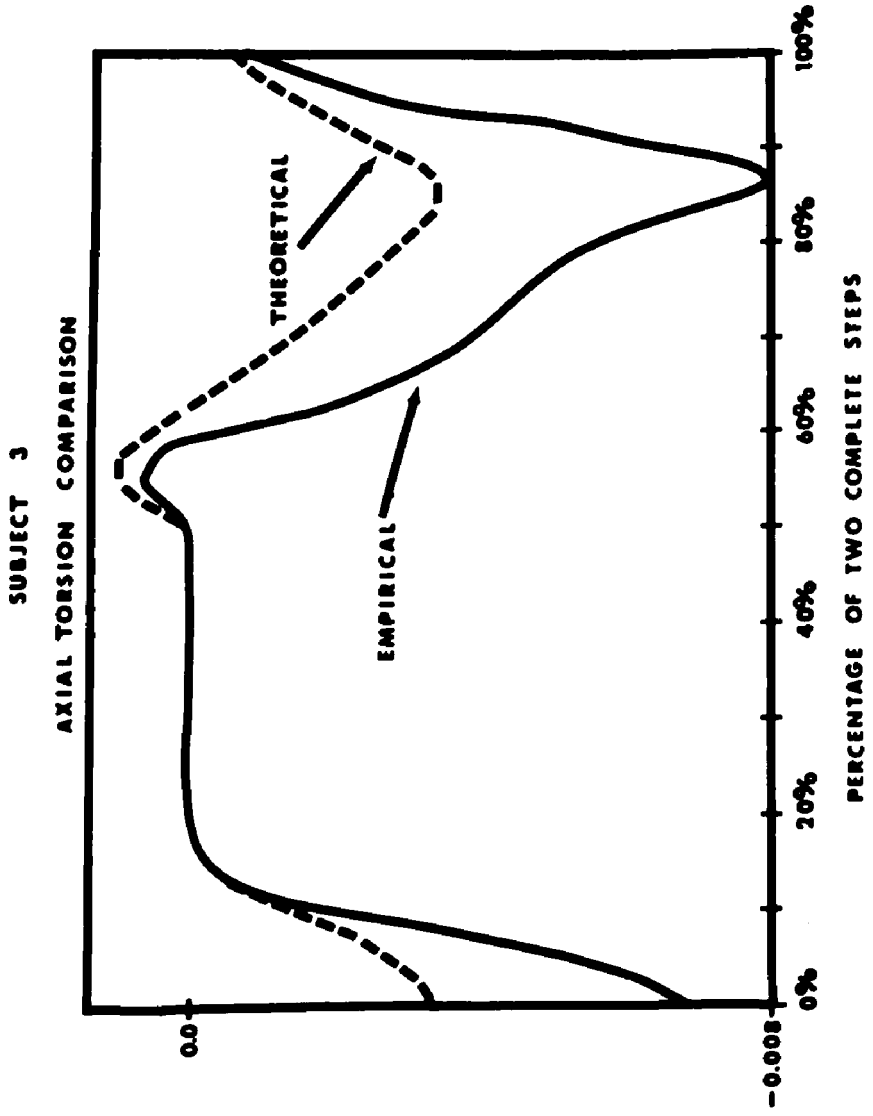


Figure 6.3.--Axial Torsion

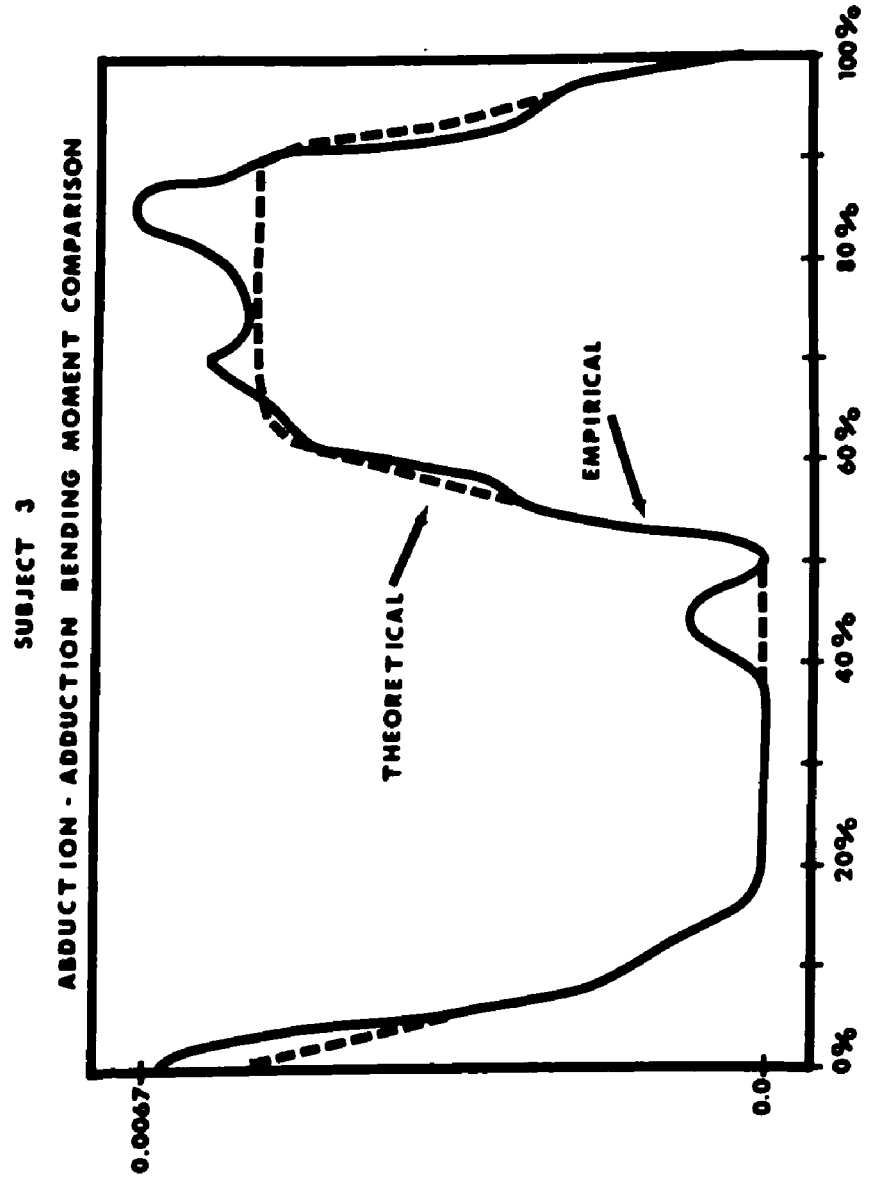


Figure 6.4.--Abduction-Adduction Moment

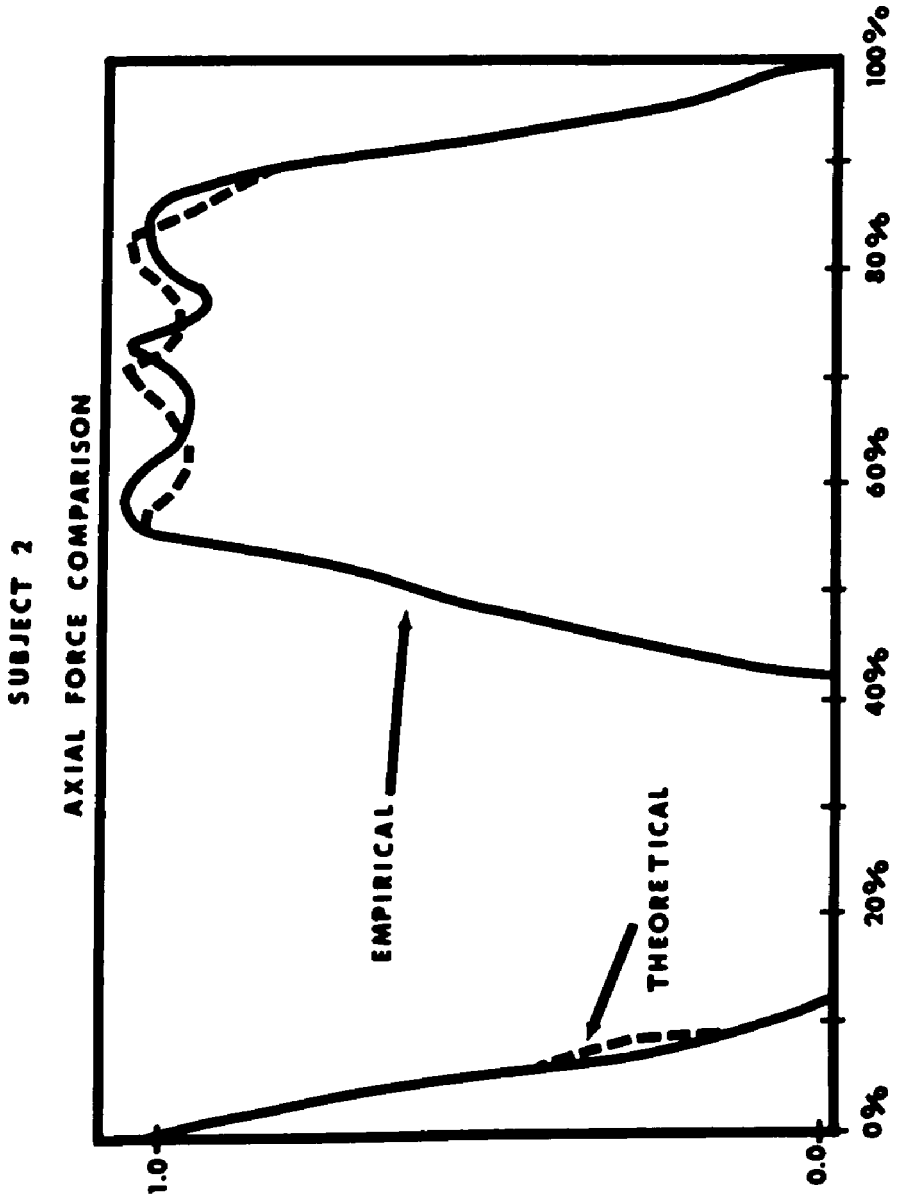


Figure 6.5.--Axial Force

Error Analysis

Besides describing the applicability of the model, the preceding section constituted a form of an error analysis. It did this by showing how closely the model predicted reality. It also showed, as in the case of the axial torsion, why differences between the model and reality occurred.

Other forms of error analysis were possible. One method was to write the differential equations in the form of functions of coordinates and errors in these coordinates. In this manner, sensitivity coefficients could have been derived to show how much an error in a measurement could have perturbed the outcome of the kinematic analysis.

This was not done, for the reason that there were more appropriate ways of determining where error occurred in the analysis.

The first problem encountered was in the calculation of segment center of gravity accelerations with respect to the inertial reference frame. An example was that the foot's position in Euclidean three-space was specified by the (x(t),y(t),z(t)) vector and the absolute angles of the ilium, thigh, shank, and foot. This meant that the position of the center of mass of this segment was a function of 11 coordinates. The acceleration of the segment center of mass, for each point in time, using the Lagrangian interpolation polynomial, was a function of 9

data points in each of the 11 coordinates. This was too many numbers.

If this analysis were ever done again, and a Vanguard Motion Analyzer with an automated interface were available, the position of the segment center of mass with respect to the inertial reference frame would be taken directly off of the 16 mm film, rather than be calculated from a series of coordinates. Each component of acceleration calculated from this set of data would involve only 9 data points from 1 variable.

The second set of errors came from the fact that the stump/socket interface was flexible, not rigid. This was visible in the 16 mm films in terms of angular displacemnt of the pylon. The linear (pumping) was not observed for. At the time, nothing was done about either. In the future, finite element analysis hopefully will lend understanding to this viscoelastic deformation.

It was in hopes of doing such an analysis, that pressure data were taken on two areas of the stump. Given the empirical loads and a stiffness matrix model of the stump, theoretical pressures could be derived. These would be compared to the actual pressures measured (see Figure 1.3 for the thought pattern involved). Once the model was validated as an appropriate stiffness matrix, it could be applied to global displacements of the stump in the socket or to a study of stump strain energy or even to a study of critical pressures in important areas on or inside the

stump.

It was noted that most finite element analysis programs were displacement oriented. This meant that the topic of stump linear and angular global displacements would not be an inappropriate starting point for this sort of work.

Besides the angular displacements visible on the 16 mm films, Grevsten and Eriksson (1974) gave linear and angular stump deformation data gathered by roentgenographic examination.

Part of the feasibility of relating load and stress can be seen in the following figures relating actual pylon flexion-extension bending moments to patellar tendon stress and pylon axial force to gastrocnemius stress. These stresses are actually gage pressures. The minus sign denotes pressures less than ambient.

It was interesting to note that the author expected the exact opposite relationship to occur. It was felt that the total contact prosthesis would still be like the patellar tendon bearing (PTB) prosthesis and that the majority of the pylon axial force would be along the patella. Furthermore, the patella would act as a hinge point, resulting in the gastrocnemius muscle sustaining a stress distribution due to the flexion-extension bending moment.

This was not the case. The total contact prosthesis did not exhibit PTB behavior.

The mounting of the gages were personally checked by the author before the experiment began. So, no mixup occurred and the empirical results are a true representation of what happened.

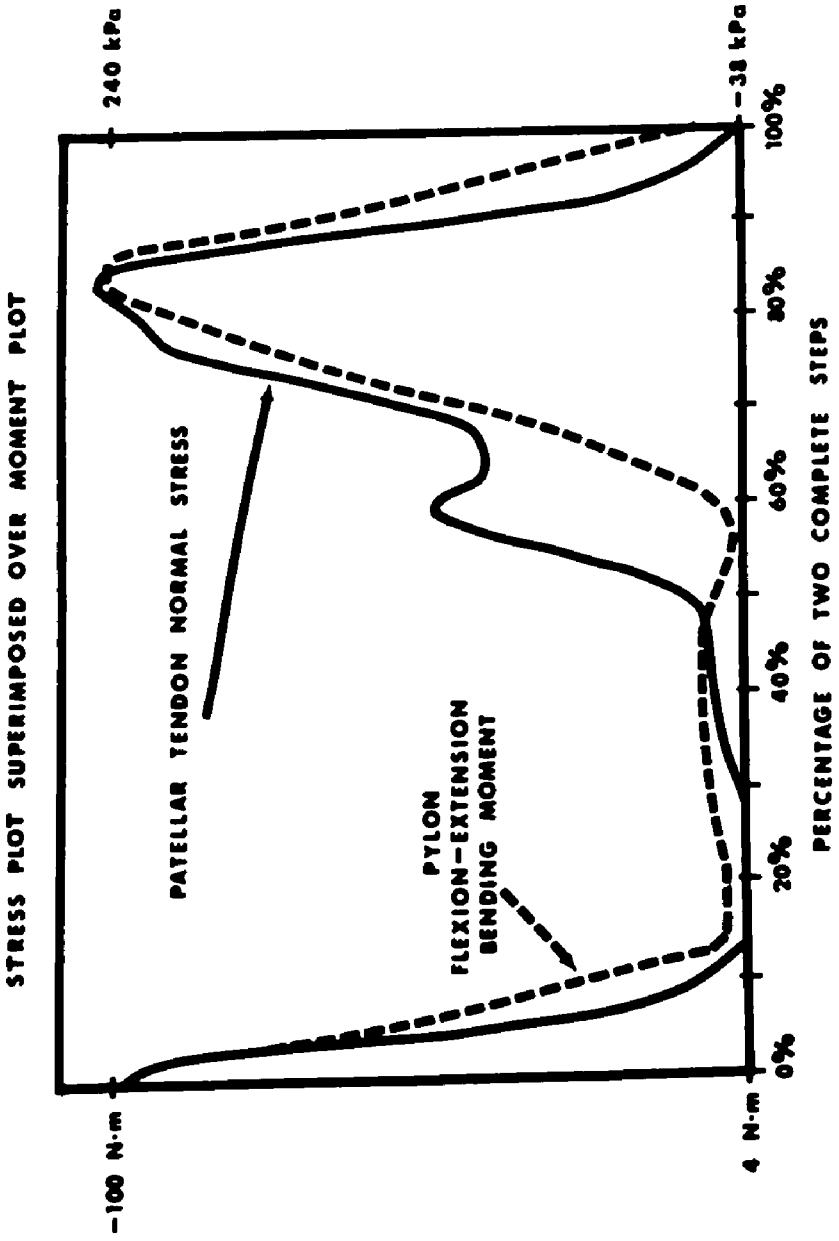


Figure 6.6.--Bending Moment vs Stress

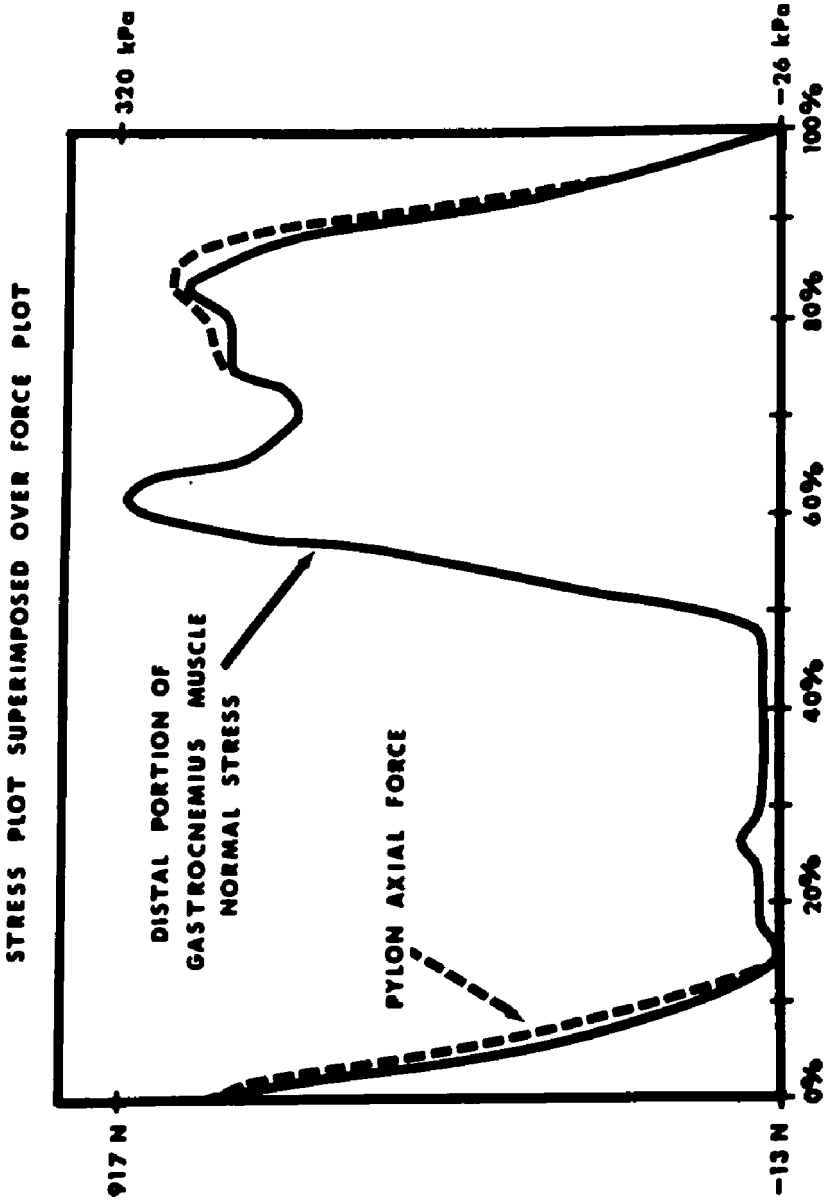


Figure 6.7.--Axial Force vs Stress

CHAPTER SEVEN

MODEL APPLICATION

Cumulative Energy Expenditure

This model was applied to solve for the cumulative mechanical energy expended by the human body. This cumulative energy was a scalar, evaluated by the classical definition of work. In particular, the joint torques integrated with respect to the segment angular displacements gave the mechanical energy expended. Similar to Beckett and Chang (1968), absolute values of the integrals were used, since all calculations represented effort that was not recoverable. Although the absolute values prevented the existence of the impossible state of negative work, they were not sufficient to preclude the possibility of regions of negative power.

To resolve this problem, the negative of this negative power was used in the analysis. The advice of Anderson and Weener (1974) was to take 0.33 times the negative of this negative power. The author felt that an appropriate factor was between 0 and 1 and that he preferred to use the full amount, 1. Whether the value of 0.33 or 1 was used, the actual difference in the cumulative energy expenditure calculations was small. This occurred because the regions of negative power rarely occurred and when they did, they were very small.

Energy, a scalar, was a function of a vector, the

alignment angle and the normalized time of two complete steps. Furthermore, this scalar was expressed in non-dimensional terms by dividing the cumulative energy expenditure by the subject's weight and distance traveled. Unlike Ganguli (1973), the weight of each amputee's prosthesis was included in this normalization.

A step was defined from heel contact, or 'heel-strike, of a foot to the toe-off of the same foot. By including two complete steps in the analysis, Figure 7.1, unnecessary discontinuities were avoided in all calculations. Furthermore, if one defined a stride from heel-strike to heel-strike or from toe-off to toe-off on the same foot, the concept of two complete steps as a gait norm contained two complete strides.

Unfortunately, the stride developed as the gait norm on a historical basis. This was really too bad, since it required a decision of whether to use the left or right stride as the norm. This choice was particularly difficult when the subject was asymmetric, such as a unilateral B/K amputee.

This argument precipitated a stride analysis for each subject. Results are presented in Table 7.1. Simply, enough work was done to show that the typical left stride time need not and usually did not equal the typical right stride time. Therefore, one should use a gait norm which encompassed both strides, the concept of two complete steps.

TABLE 7.1

STRIDE ANALYSIS

Subject	Angle	Left Stride	Right Stride	Difference
1	----	1.100 s	1.125 s	+0.025 s
3	0.0°	1.305 s	1.285 s	+0.020 s
3	+3.3°	1.290 s	1.275 s	+0.015 s
3	+6.7°	1.360 s	1.295 s	+0.065 s
3	+10.°	1.270 s	1.265 s	+0.005 s
3	-3.3°	1.230 s	1.230 s	0.000 s
3	-6.7°	1.260 s	1.235 s	+0.025 s
3	-10.°	1.265 s	1.295 s	-0.030 s

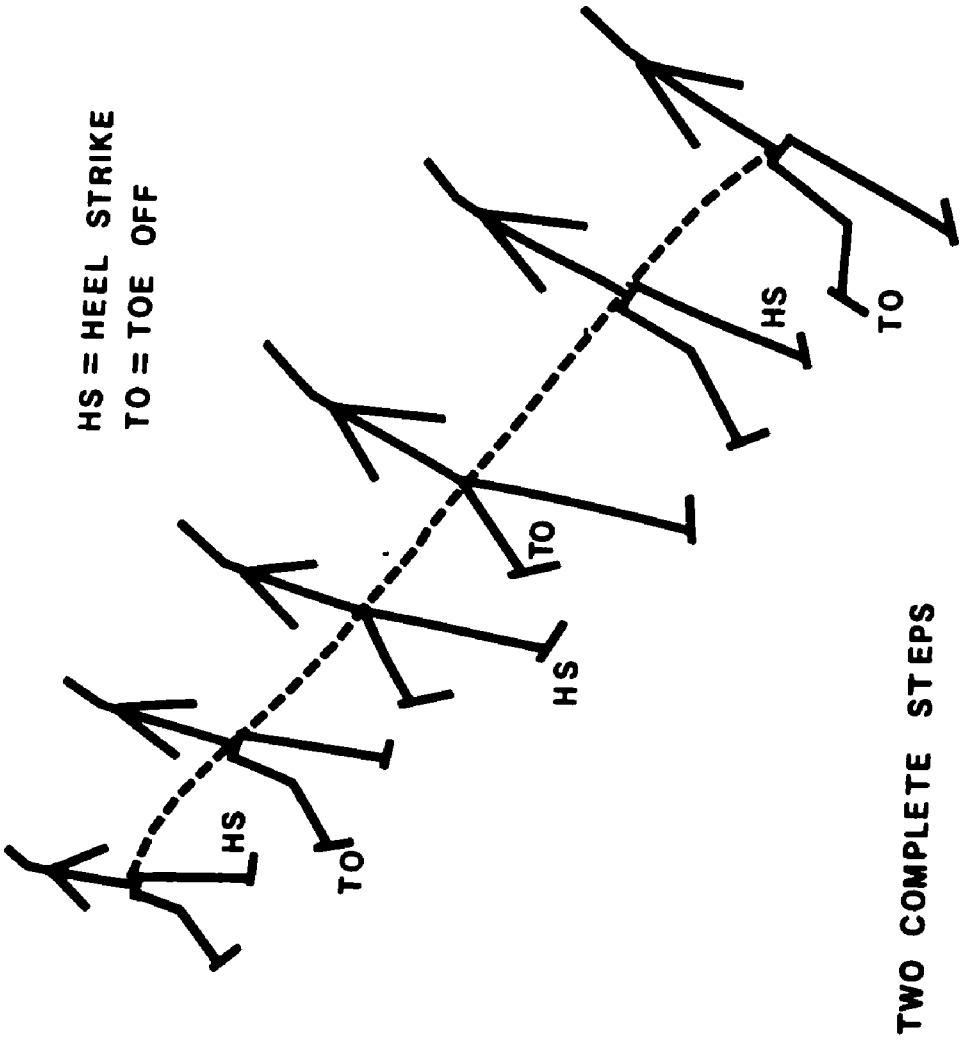


Figure 7.1.--Two Complete Steps

Cumulative energy expenditure, the scalar valued function of a vector, was expressed in terms of a surface. By drawing the iso-alignment-angle curves and the iso-percent-two-steps curves, a crosshatch denoting the normalized cumulative energy expenditure surface was generated in each of Figures 7.2 and 7.3.

The first conclusion from both Figures 7.2 and 7.3 was that each surface admitted a minimum at the optimum alignment as defined by both prosthetists. This was compatible with Hughes (1967), who concluded that minimum energy expenditure should be the criterion used to control the amputee's gait rather than trying to make the prosthesis bearing leg walk symmetrically with respect to the normal leg.

The next curves drawn were two iso-energy curves per amputee, Figures 7.4 and 7.5. These curves show that as the alignment angle tended towards optimum, each amputee could go farther on a given amount of energy. This approach was simply a way of restating the results of Figures 7.2 and 7.3.

Figure 7.6 shows, by iso-midstance curves, the energy expenditure gradient for the non-prosthetically dominated step, step 1, and for the prosthetically dominated step, step 2. These curves depicted cumulative energy expenditure at mid-stance. In this context, mid-stance was defined as the exact mid-point in time of the single stance phase. The footswitch data were used to

established these mid-points in time.

It is possible to see from Figure 7.6 that during the non-prosthetically dominated step, energy expenditure was perturbed by the alignment angle applied to the other leg. It seemed reasonable to expect some interaction, since the two legs were physically connected. However, a closer look at Figure 7.6 shows that cumulative energy expenditure even on the non-prosthetically dominated step increased with alignment angle. This gave the conclusion that the minimum energy state occurred along optimum alignment for both the prosthetic and the non-prosthetically dominated steps.

The final remarks concerned amputee versus non-amputee energy expenditure. In Figure 7.7, at optimum alignment, Subjects 2 and 3, the amputees, exerted about the same normalized cumulative energy. This was about half the normalized cumulative energy expenditure of Subject 1, the non-amputee. This graph sparked discussion and prompted a review of other means of determining gait energy expenditure. It was hoped such a review would help explain the results shown. The discussion is presented in the next section.

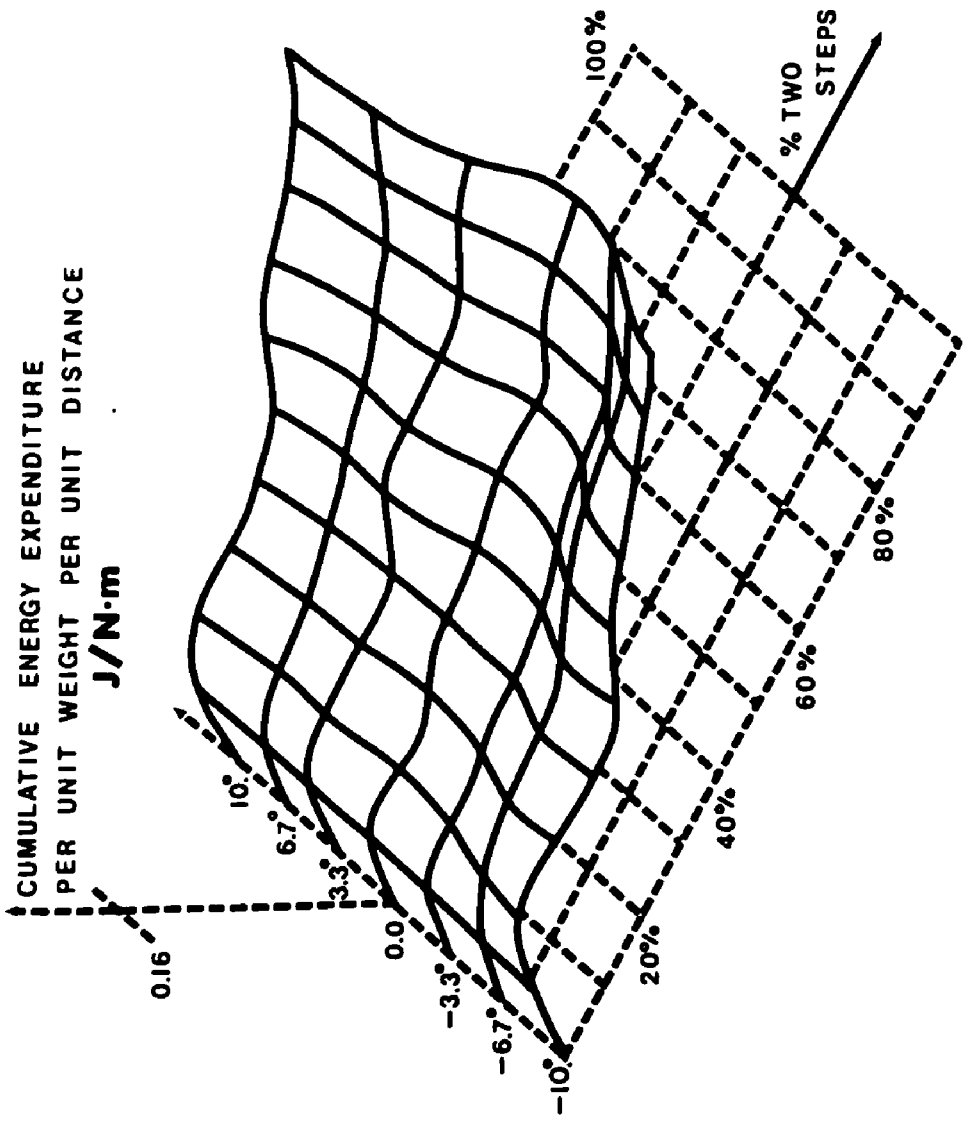


Figure 7.2.--Subject 2: Cumulative Energy Expenditure

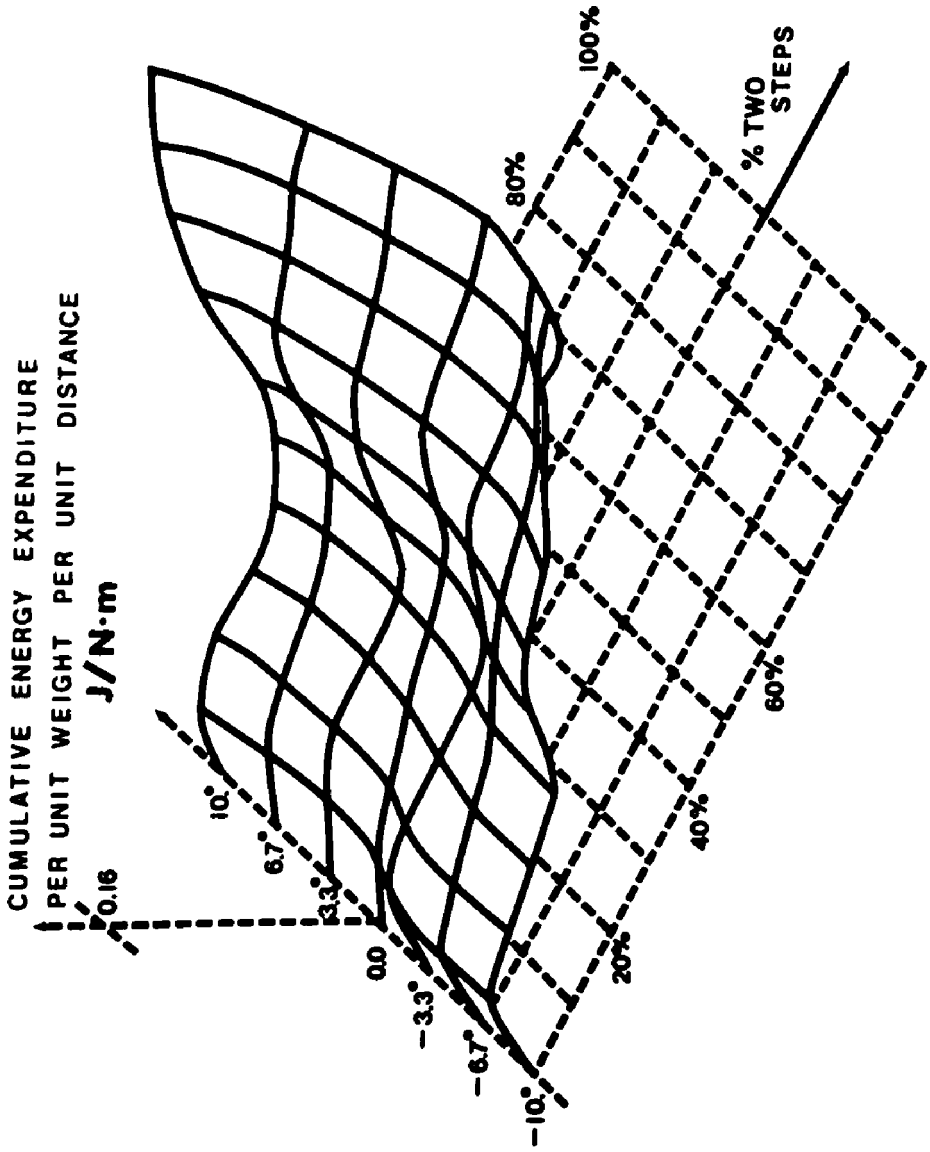


Figure 7.3.--Subject 3: Cumulative Energy Expenditure

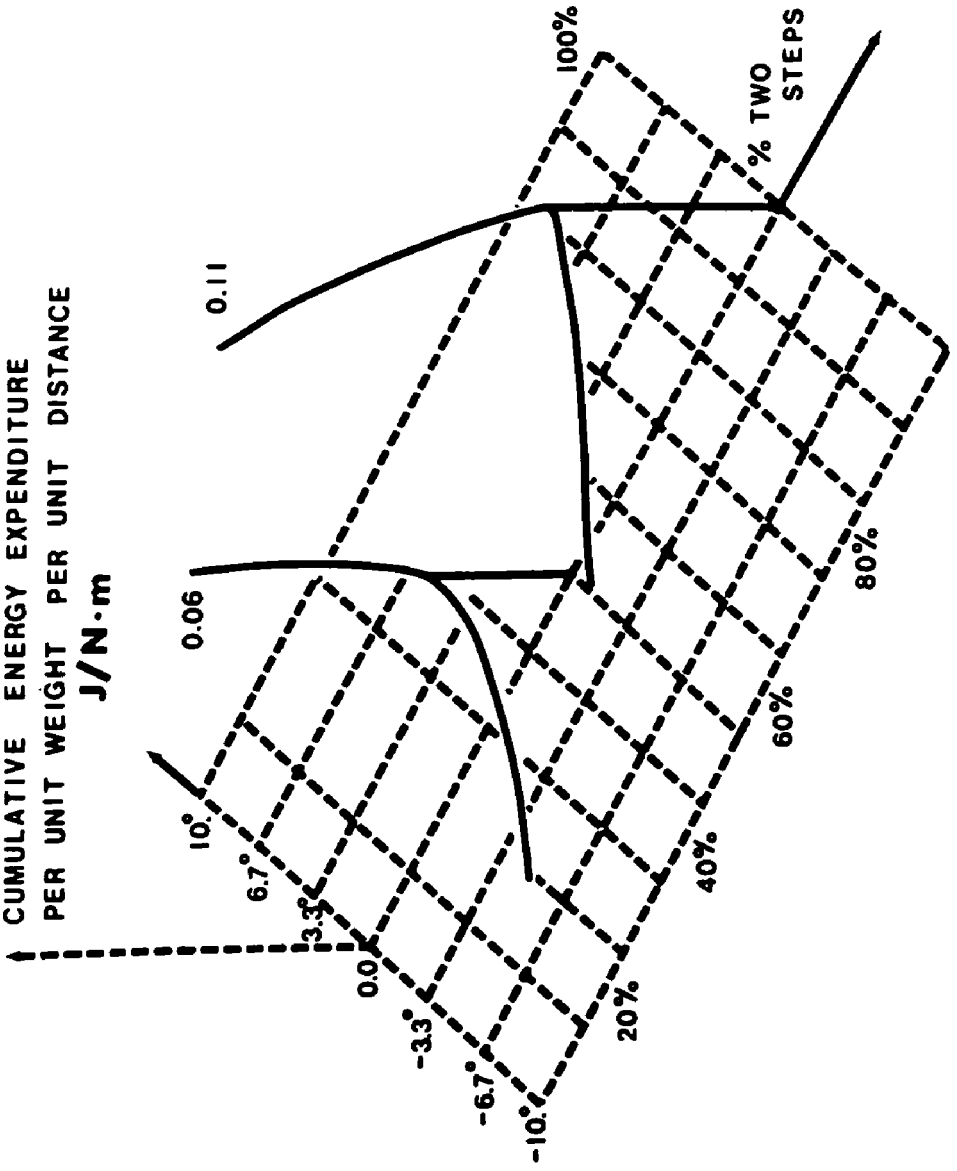


Figure 7.4.--Subject 2: Iso-Energy Lines

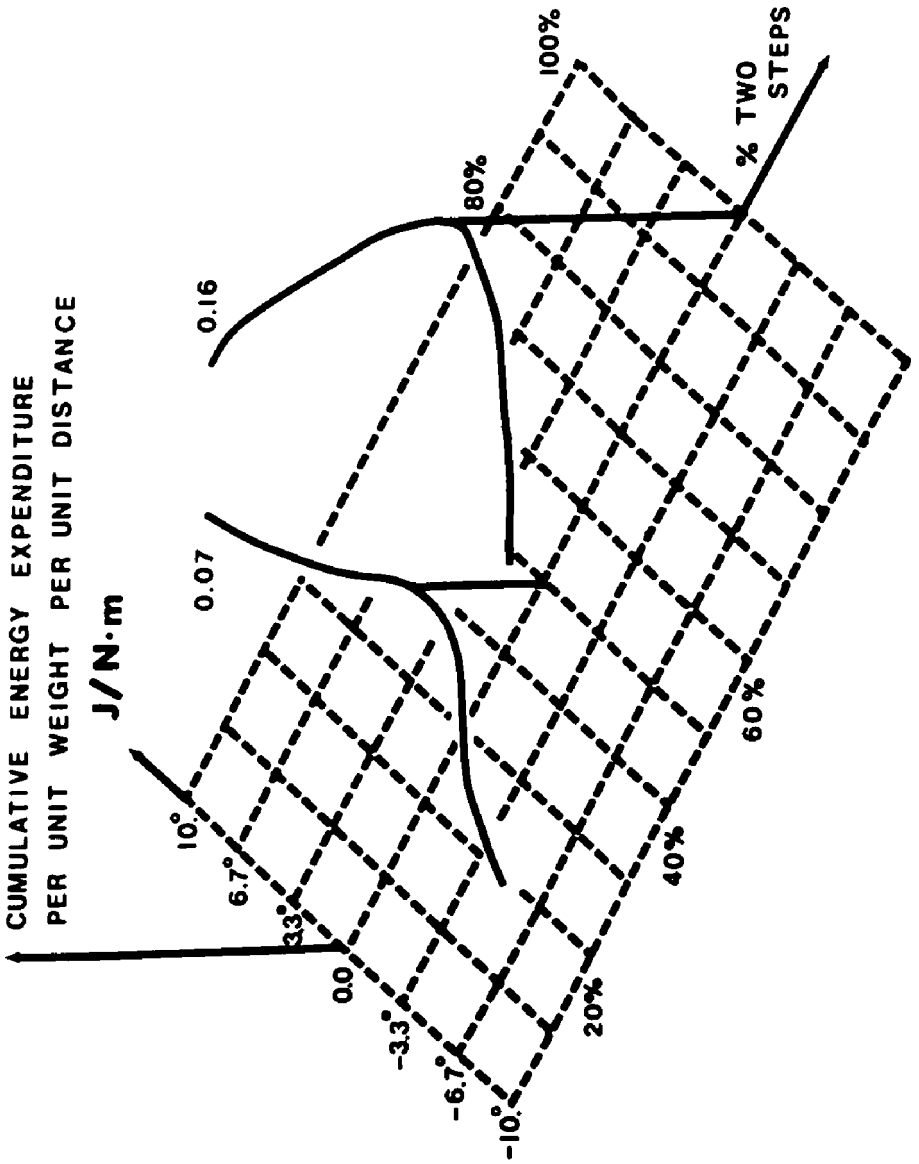


Figure 7.5.--Subject 3: Iso-Energy Lines

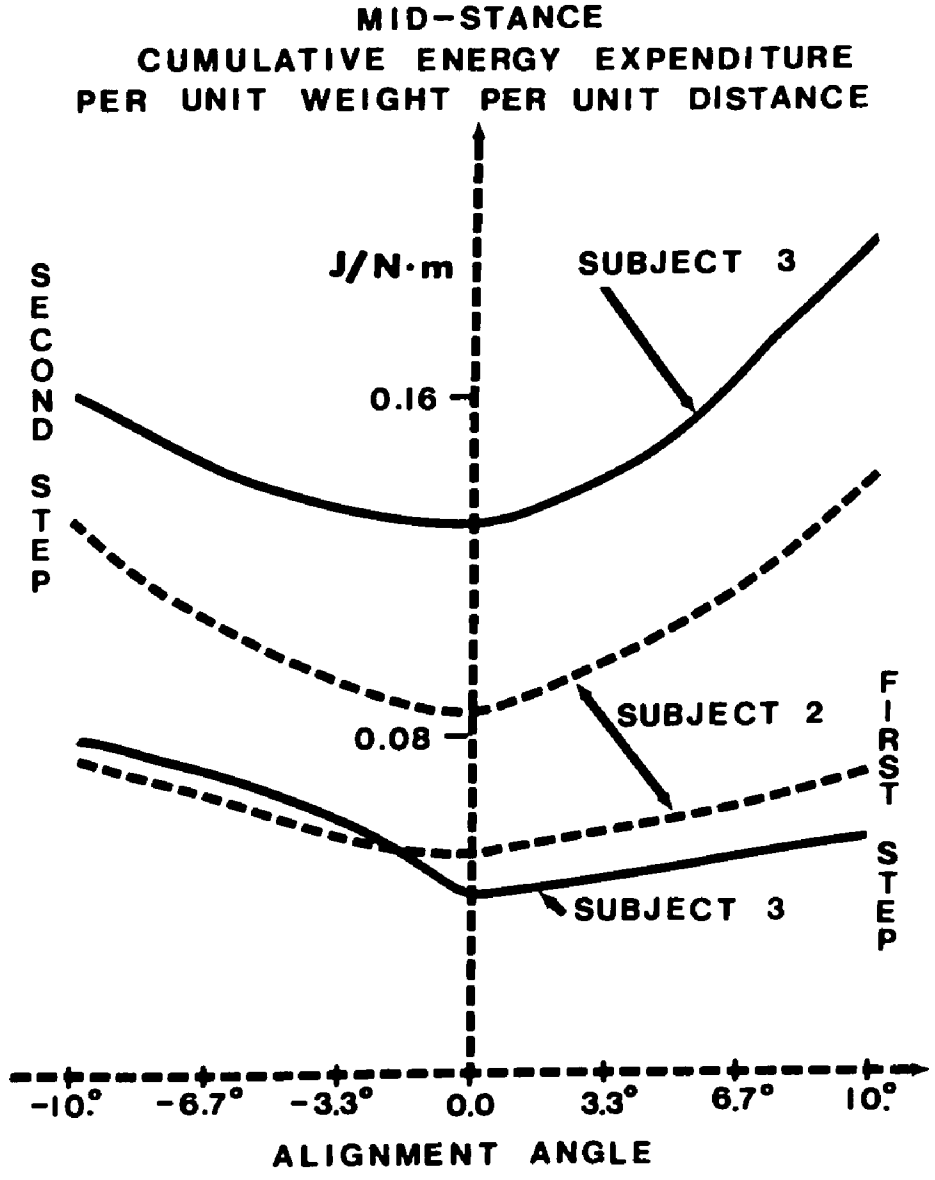


Figure 7.6.--Subjects 2 and 3: Leg Interactions

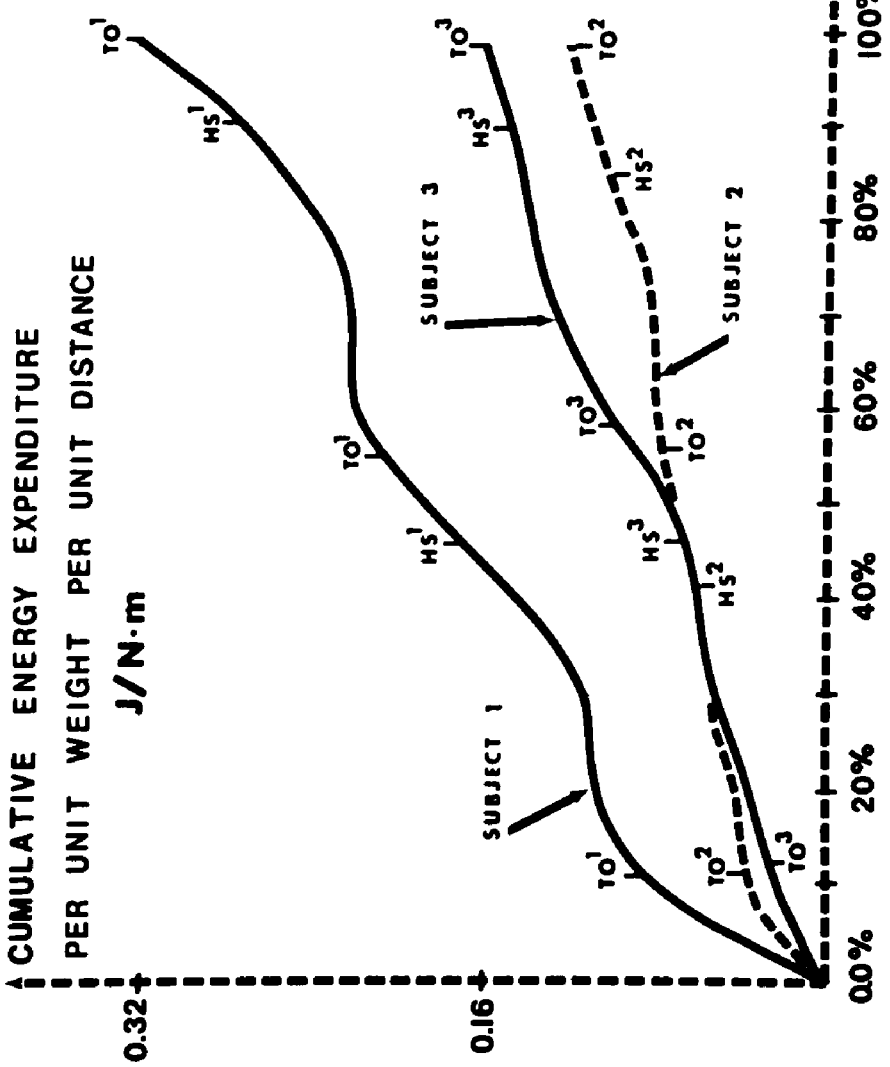


Figure 7.7.--Subject 1, 2, and 3: Comparisons

Other Research Efforts

The primary effort in determining gait energy expenditure is by means of pulmonary ventilation and oxygen consumption. Bard and Ralston (1959), Dink (1964), Ganguli, et al, (1973) and (1974), Gonzalez, Corcoran, and Rayes (1974), Lukin (1963), Margaria (1975), Malhotra (1966), Ralston (1958) and (1964), Ralston and Lukin (1969), and Traugh, Corcoran, and Reyes (1975) are some of the researchers who used this experimental method.

It should be pointed out that one of the key contributions to gait study was that there exists an optimum average velocity of forward motion in which energy expenditure is minimized. The works of Bard and Ralston (1959) will be used, since Ralston appeared to have the deepest foundation in this area.

There were two important reasons that energy expenditure as defined by the classical definition of work was applied rather than the popular metabolic approach. One reason was best expressed by the work of Margaria, et al, (1975). This article described the onset of exercise as a step-function disturbance and increase in body oxygen consumption as the response. The wave form of this response was similar to that of a first order system, exponential.

Associated with this function was a half reaction constant of 30 seconds. To have studied metabolic energy expenditure by this method would have meant requiring the

subject walk for several minutes before data were taken. This requirement applied at the extreme angular perturbations would have been extremely demanding.

Furthermore, it appeared that the maximum rate that oxygen consumption could be monitored was in 60 second intervals. This approach certainly could not give much detail or resolution in its portrayal of energy expenditure as a function of time.

For these rather sound reasons, the method of determining energy expenditure in a rather short period of time (2 complete steps) and at a frequency (32 Hz) to give a description of the instantaneous nature of energy expenditure was chosen. This was the calculation of cumulative energy expenditure based on the classical definition of work. Since this was different from the metabolic approach, it was necessary to establish a relationship with Ralston's results. This precipitated a velocity analysis.

An average velocity of forward motion was calculated for each subject over 7 different tests. Each of these 7 average velocities was defined as the mean instantaneous velocity of forward motion.

Once 7 average velocities of forward motion were derived for each subject, it was possible to calculate the mean and standard deviation of these 7 average velocities for each subject. The results are in Table 7.2 and are in metres/second.

It is interesting to note in Table 7.2 that the standard deviation, as expressed as a percentage of the mean, is quite small in all three cases. This meant that each set of velocities were compactly distributed about their respective means. This compactness and the relatively large difference in the three means meant that the three distributions were also quite distinct.

The probability of interference or overlap of these normal distributions was analyzed by studying two distributions at a time with the standardized normal variate.

The new mean was the difference between the means of the two normal distributions being studied. The new standard deviation was the square root of the sum of the squares of the two standard deviations.

This approach showed that the greatest probability of interference was between the velocity distributions of subjects 2 and 3. However, this greatest probability was only on the order of 0.02%.

TABLE 7.2

Foward Velocity Statistics

Subject	V(mean)	V(S.D.)	V(S.D.)/V(mean)
1	1.324 m/s	0.038 m/s	2.87%
2	0.914 m/s	0.048 m/s	5.25%
3	1.085 m/s	0.048 m/s	4.42%

From the data in Table 7.2, it was possible to compare the means of the the average velocities of forward motion of each of the three subjects. The results are in Table 7.3 and are expressed in non-dimensional terms. This was done by dividing by the lowest mean velocity, that of Subject 2.

It should be noted that this non-dimensional ratio is quite pronounced. Subject 1, the healthy non-amputee walked much faster than either amputee. Subject 3, the younger, stronger amputee, walked somewhat faster than the older, less able Subject 2.

The square of this velocity ratio is presented in Table 7.3 also. It was apparent that if the square of the velocity ratio were used to normalize Winarski's results in Figure 7.7, that all three curves would fall on about the same line. This would not be totally unreasonable to do, since the ratio^2 relates to a kinetic energy ratio, certainly something which was inherent in the differential equations of motion.

TABLE 7.3

Mean Velocity Ratios

Subject	Ratio	Ratio^2
1	1.45	2.10
3	1.19	1.41
2	1.00	1.00

The results of McDonald (1961) suggested that the normalization of energy with respect to weight of the subject would tend to artificially inflate values of normalized energy expenditure for light subjects. This was another workable solution for why Subject 1, the lightest of the three subjects, had a rather large normalized energy expenditure.

The final velocity analysis is presented in Tables 7.4 through 7.6. These tables represent the conclusion that the velocity distribution for each subject was compact, or tightly grouped. The significance of this is that each subject was asked to walk at his comfortable speed. No external constraints, such as a treadmill, were put in use.

This speed could be labeled his Ralston optimum. The observation is that the velocities are so closely grouped, that the minimum energy surfaces could not have been created by velocity changes alone.

If one believes the formulas given in Bard and Ralston (1959), one can calculate the first and second derivative of energy expenditure with respect to velocity.

Now, by use of a Taylor series expansion about each subject's Ralston-optimum, the change in energy as a function of velocity perturbation rather than alignment perturbation can be calculated. This is actually an easy problem. Since the series is calculated about the minimum, the first derivative vanishes. The second derivative is twice the slope of the power curve times the square root of Ralston's power curve slope divided by its intercept. As a matter of fact, this means that Ralston's second derivative is twice the reciprocal of Ralston's value of optimum velocity, 2.0/(73.8 metres/minute) or 0.615 seconds/metre, times the slope of the power curve.

This gives a value of the second derivative as
0.0001436 calories/(metre-kilogram)/(metre/minute)^2

In S.I. units this is:
0.052719 joules/(Newton-metre)/(metre/second)^2

This second derivative was determined from metabolic energy expenditure. To convert it to a classical definition of work value required establishing a relationship between the two quantities. This relationship is called efficiency. It is defined as the ratio of work output to energy input.

Ralston and Lukin (1969) established this gait efficiency at about 23%. This meant that the classical definition of work done in ambulation is about 23% of the

metabolic energy expended. Taking this factor into account gave the desired second derivative shown below.

0.01212531 joules/(Newton-metre)/(metre/second)^2

The Taylor series, in its truncated form, is written below:

$$E(V + dV) = E(V) + dV*E'(V) + dV^2*E''(V)/2.0$$

In simpler form, using the S.I. units of m/s for velocity and J/lim for normalized energy:

the change in normalized energy =

Change in Velocity^2*(0.00606266)

In Tables 7.4 through 7.6, there were no changes in velocity away from the mean (Ralston optimum) velocity sufficient to change the normalized energy expenditure enough to affect the results by more than 0.032% for either amputee. This velocity independence of the energy surfaces certainly added to establishing their credibility.

TABLE 7.4

Subject 1: Forward Velocity

Velocity	V-V(mean)	# S.D.	V/V(mean)
1.297 m/s	-0.027 m/s	0.71	0.98
1.332 m/s	+0.008 m/s	0.21	1.01
1.392 m/s	+0.068 m/s	1.79	1.05
1.353 m/s	+0.029 m/s	0.76	1.02
1.306 m/s	-0.018 m/s	0.47	0.99
1.298 m/s	-0.026 m/s	0.68	0.98
1.287 m/s	-0.037 m/s	0.97	0.97

TABLE 7.5

Subject 2: Forward Velocity

Velocity	V-V(mean)	# S.D.	V/V(mean)
0.922 m/s	+0.008 m/s	0.16	1.01
0.932 m/s	+0.018 m/s	0.38	1.02
0.921 m/s	+0.007 m/s	0.15	1.01
0.927 m/s	+0.013 m/s	0.27	1.01
0.990 m/s	+0.076 m/s	1.58	1.08
0.861 m/s	-0.053 m/s	1.10	0.94
0.844 m/s	-0.070 m/s	1.46	0.92

TABLE 7.6

Subject 3: Forward Velocity

Velocity	V-V(mean)	# S.D.	V/V(mean)
1.108 m/s	+0.023 m/s	0.48	1.02
1.036 m/s	-0.049 m/s	1.02	0.95
0.994 m/s	-0.091 m/s	1.90	0.92
1.089 m/s	+0.004 m/s	0.08	1.00
1.177 m/s	+0.092 m/s	1.92	1.08
1.091 m/s	+0.006 m/s	0.13	1.01
1.097 m/s	+0.012 m/s	0.25	1.01

Reliability

In an effort to relate each amputee's actual sample of 7 iso-alignment curves to his theoretical population of iso-alignment curves, a statistical analysis was undertaken.

First, a "failure-success" study of the concave pattern of the iso-alignment curves was done. This was accomplished by comparing each set of 7 iso-alignment curves, Figures 7.2 and 7.3, to the theoretical surface sketched before the experiment, Figure 1.1. Due to the discrete, binary, "acceptable-not acceptable" decision making criteria for each iso-alignment curve, the theory of the binomial distribution was applied.

It was noted that there were 7 curves per amputee. For each amputee, all 7 were "accepted." Based on these data, a 90% confidence level, and the F-distribution, the

upper and lower reliability limits were calculated.

The upper limit for each amputee was 100%, the lower limit, 72%. This meant that if 100 iso-alignment curves would have been calculated for each subject, there was 90% confidence that between 72 and 100 of them would have fallen into the same general concave pattern as Figure 1.1. It was felt that this was sufficient to show the existance of a test-population minimum energy expenditure for each subject.

Once the reliability of the existance of a concave surface for each subject's test-population was established, it was necessary to discuss the location of this test-population minimum for each subject. No confusion was made between the test-population for each subject and the subject-population of all B/K amputees.

The experimental effort showed each surface to possess a minimum along the 0.0 degree iso-alignment curve. It was decided to apply two appropriate energy expenditure standard deviations about this 0.0 degree line and see how many degrees of alignment adjustment were necessary to breech this volume of ambiguity. Two standard deviations were chosen because this gave a confidence level of slightly more than 95%. The work of Lukin (1963) was chosen for the source of this standard deviation. His values were expressed as a percent of the mean energy expenditure. It was decided to use the high value of standard deviation, 5%, which corresponded to intense

exercise. This figure corresponded nicely to the normalized standard deviation of the average forward velocity of the two amputees, Table 7.3.

The conclusion of this second effort showed that at 95% confidence, the test-population minimum energy expenditure alignment was determined within 1 angular perturbation (3.3 degrees) for each of the 2 amputees.

It should be noted that both the above observations did not concern the repeatibility of the experiment. Instead, the approach was to discuss the reliability of the actual experiments performed.

This entire project was done on an analytical basis, hence, statistical extrapolation of the concept of minimum energy from the sample of two amputees to the entire B/K amputee-population was never intended.

Conclusions

This was not the first time anyone has studied energy expenditure. Elftman (1939), Ganguli (1973) and (1974), Ralston (1958), and Saunders (1953) studied the manner in which energy was expended in human ambulation. Vukobratovic (1973) studied power expenditure requirements for an actual robot. Beckett and Chang (1968) studied the energy requirements of mathematically synthesized gait. In addition to these studies, Anderson and Weener (1974) studied energy expenditure, but for the purpose of man-powered flight.

However, this was the first time that energy expenditure was studied as a function of alignment adjustment and time. In more analytical terms, it was the first time that the effect of perturbations were characterized in terms of the energy demand to respond to those disturbances. It was then concluded that a state of minimum energy expenditure existed at the optimum alignment as defined by the prosthetists.

Second, it was concluded that the minimum energy surfaces were derived under reasonably constant velocity conditions. This situation occurred for both amputees without the use of a treadmill. This meant the surfaces were generated by alignment changes rather than velocity changes.

Thirdly, it was felt that this constant velocity situation had something to say about energy expenditure. In particular, the subject's forward velocity is a major determinant in his energy expenditure. An optimum velocity exists for each subject as a function of his size, mass distribution, and strength. One may perturb the system (alignment angle) and the system will respond (concave energy surface). However, the system will still minimize its energy output with respect to velocity by staying quite close to the optimum velocity.

APPENDIX A

System International Units

Name	Symbol	
Ampere	A	1 ampere = 1 coulomb/second
Coulomb	C	6.24 giga-giga electrons
Gravity	g	g = 9.806 metres/second^2
Hertz	Hz	1 Hz = 1 cycle/second
Joule	J	1 joule = 1 Newton-metre
		1 joule = 0.2390 gram calories
Kelvin	K	K = ^{0}C + 273.16^0
Kilogram	kg	1 kg = 2.204623 pounds mass
Metre	m	1 metre = 3.280840 feet
Newton	N	1 Newton = 0.2248 pounds force
Ohm	Ω	1 ohm = 1 volt/ampere
Pascal	Pa	1 Pascal = 1 Newton/metre^2
		6.89 kPa = 1 p.s.i.
Radian	rad	1 radian = 57.2957795 degrees
Volt	V	1 volt = 1 joule/coulomb
Watt	W	1 watt = 1 joule/second

APPENDIX B

Powers of Ten

Power	Prefix
+12	tera
+9	giga
+6	mega
+3	kilo
-3	milli
-6	micro
-9	nano
-12	pico
-15	femto
-18	atto
-21	flato

APPENDIX C

Anatomical Reference Points

1. Acromion: /ah-kro'-me-on/ The outward extension of the spine of the scapula, forming the point of the shoulder.

2. Condyle: /kon'-dil/ A rounded projection of a bone, usually for articulation with another bone.

3. Crest of Ilium: /il'-e-um/ The thickened and expanded upper border of the pelvis.

4. Gastrocnemius: /gas''-troc-ne'-me-us/ Muscle (1) Caput mediale: origin at the upper back part of the medial condyle and of the femur. (2) Caput laterale: origin at the lateral condyle and posterior surface of the femur. Both unite with the tendon of soleus to form the tendon of Achilles.

5. Greater trochanter: a broad flat process at the upper end of the lateral surface of the femur to which several muscles are attached.

6. Ilium: /il'-e-um/ The hauch bone, distinct in fetal life, but becoming the wide upper portion of the

innominate, or nameless, bone.

7. Lateral malleolus: /mal-e'-o-lus/ A rounded process on the lateral side of the ankle joint.

8. Medial condyle: At distal end of femur articulating with the superior surfaces of the head of the tibia.

9. Olecranon process: /o-lek'-rah-non/ A curved process of the ulna at the elbow.

10. Process: A slender projecting point.

11. Scaphoid: /skaf'-oid/ The outer bone of the first row of carpal or wrist bones.

12. Sternum manubrium: /ster-num' mah-nu'-bre-um/ Uppermost portion of the sternum or breast bone.

13. Trochanter: /tro-kan'-ter/ Either of the two great processes below the neck of the femur.

14. Ulna: /ul'-nah/ The inner and larger bone of the forearm, on the side opposite to the thumb. The other bone of the forearm is the radius.

APPENDIX D

Subjects' Reactions

These notes on Subject 2 were published with the kind permission of the author, Mr. Joseph P. Giacinto, C.P.

3.3° Extension (Pylon Flexion)

In extension. Patient feels the ball of the foot more at 3.3° extension. Socket is pushed against patient's patellar tendon and also has a double jiggle movement. Second test run: patient walked with a smoother gait. (Compensated?)

6.7° Extension (Pylon Flexion)

Patient spends more time on his heel and has a double jiggle from heel strike to mid-stance. Second run: patient has a smoother gait.

10.0° Extension (Pylon Flexion)

Patient knee is falling backward and is compensating by arching his back - and spends more time on his heel. Second run: patient compensated and had a smoother gait. He stills has a double jiggle from heel strike to mid-stance. Third run: patient gait smoother still.

ANNOTATED BIBLIOGRAPHY

Anderson, William J., and Weener, Earl F. "Human Power Production in a Caged Situation." American Institute of Aeronautics and Astronautics/ Massachusetts Institute of Technology/ Soaring Society of America 2nd International Symposium on the Technology and Science of Low Speed and Motorless Flight. Cambridge, Massachusetts, September, 1974, Paper AIAA 74-1027.

The authors calculated mechanical efficiencies for a human doing work in a standing and stooping cycle while enclosed in a cage. Similar to Beckett and Chang (1968), the classical definition of work was augmented by expressing differential displacement as the product of velocity and differential time.

One of the interesting portions of this paper was the acknowledgement that losses in the cycle were due to gravity and human inability to store energy in the unloading portion of the cycle.

Allum, J.H.J. And Young, L.R. "The Relaxed Oscillation Technique for the Determination of the Moment of Inertia of Limb Segments." Journal of Biomechanics, Volume 9, Number 1 (1976), 21-25.

Forced sinusoidal oscillations of the forearm, when the appropriate muscles were relaxed, were used for determination of its moment of inertia. The elbow rested in a cup in an effort to fix its anatomical position. The forearm was displaced sinusoidally at known frequencies and amplitudes. Recording of the forces needed to so accelerate the forearm allowed determination of the moment of inertia.

Results agreed favorably with an experimental model.

Appoldt, Frank; Bennett, Leon; and Contini, Renato. "Stump Socket Pressure in Lower Extremity Prostheses." Journal of Biomechanics, Volume 1, Number 4 (December, 1968), 247-257.

Two types of pressure transducers were used to measure normal interface pressures on two above knee amputees.

According to the authors, alignment adjustments produced small, difficult to discern pressure changes.

The authors were trying to detect these pressure changes statistically. The pressures varied from step to step, the distribution of which possessed a mean and a standard deviation. The

subject's response to an alignment change might have been to change the mean and/or the standard deviation. Statistical detection of this shift was "burdensome."

ASME Orientation and Guide for Use of Metric Units. New York: The American Society of Mechanical Engineers, 1972.

This text was the basis for all S.I. measurements listed in this dissertation. It was kindly donated by Professor Joseph E. Shigley, of The University of Michigan.

Ayoub, M.M.; Walvekar, Arun; and Petruno, Mike. "A Biomechanical Model for the Upper Extremity Using Optimization Techniques. Society of Automotive Engineers, February, 1975, Paper 740272.

Using the differential equations of motion and the constraint of minimum power expenditure, the joint torques and segment motions were determined for various tasks. Empirically based comparisons were offered.

Bard, Gregory, and Ralston, H.J. "Measurement of Energy Expenditure During Ambulation, with Special Reference to Evaluation of Assistive Devices." Archives of Physical Medicine and Rehabilitation, Volume 40, Number 10 (October, 1959), 415-420.

This particular article, similar to Ralston (1958), was a well written and highly informative documentation of metabolic energy expenditure as a function of average speed, V, of forward motion. This average speed, V, was measured in metres/minute.

The linear power equation used was:

1. $\text{calories/minute/kilogram} = 29.0 + 0.0053V^2$

Integration of this power equation with respect to time and normalizing it with respect to distance traveled gave the energy equation:

2. $\text{calories/metre/kilogram} = 29.0/V + 0.0053V$

This particular equation was the sum of a hyperbola and a straight line. It was disappointing that the authors kept referring to this sum of two distinct curves as totally hyperbolic in nature. As it was, when V tended to zero, the curve was hyperbolic; but as V tended to a large number, the curve was a straight line.

Six above knee amputees were studied. They used their usual assistive devices and then

alternate assistive devices. Energy expenditure was measured exactly in the same manner as in Ralston (1958) at various subject average speeds of forward motion. Plots of the energy expressed in calories/metre/minute versus velocity in metres/minute showed curves of the same nature as equation 2. The one difference was that these A/K curves were always higher than the predictions of equation 2. This tended to say that handicapped people had to expend more energy per unit mass and unit distance than the "normal" or non-handicapped.

However, the authors did present one example of a person handicapped by a cerebral accident, whose normalized energy versus velocity was below the values predicted by equation 2.

Finally, an A/K subject was shown to have an optimum velocity energy expenditure of 1 calorie/metre/kilogram. This corresponded well to Winarski's second Subject who had an optimum alignment energy expenditure of 1.1 calories/metre/kilogram.

Bartz, J.A., and Gianotti, C.R. "Computer Program to Generate Dimensional and Inertial Properties of the Human Body." American Society of Mechanical Engineers, November, 1973, Paper 73-WA/BIO-3.

This paper constituted a thorough study of anthropometry. A comprehensive literature search was discussed. Then values from previous experimenters' works were compared to a computer's predictions.

The previous experimenters mentioned were Dempster, Drillis and Contini, Harless, Braune and Fischer, Lepley, Palten and Theiss at Calspan, Hanavan, Kroemer, as well as Chaffin and Snyder, Conway, Sahley, Stoudt, Clauser, and Damon.

The extensive listings of experimental and analytical data were extremely informative as well as a firm guide to anyone hoping to model the human body.

Beckett, Royce, and Chang, Kurng. An Evaluation of the Kinematics of Gait by Minimum Energy." Journal of Biomechanics, Volume 1, Number 2 (July, 1968), 147-159.

A model of a human leg was derived for motion confined to the sagittal plane. This modeled leg consisted of four parts; a thigh, a shank, a foot, and a toe.

The analysis was confined to the swing phase of the gait cycle.

In addition to the derivation of the differential equations of motion, the authors

presented several model constraints. These restrictions involved the path of the hip, the path of the toe, the curve of hip moment versus time, and cadence (walking speed).

The combination of these constraints and the differential equations of motion enabled the authors to solve for joint torques and limb displacements as functions of time. Then the classical definition of work was applied to calculate energy expenditure. It was evaluated in terms of the product of moment and limb relative angular velocity integrated with respect to time. Expressing energy expenditure as a function of cadence (steps/minute) showed the existance of a minimum energy expenditure. This defined an optimum cadence quite close to the work of Ralston (1958).

Bender, M., et al. "Error Analysis of Object Space Motion Determination with Camera-Analyzer System." Proceedings of the Society of Photo-optical Instrumentation Engineers. Edited by Brown and Patrick. Volume 34 (November, 1972), 59-64.

A mathematical model of a single degree of freedom compound pendulum was derived. Displacement data from a free swinging pendulum of known mass, length, mass moment of inertia, and small amplitude time of period was used as input to the mathematical model.

The output of this analysis was a theoretical value of the acceleration of gravity. It was within 5.2% error of the actual value.

The paper also presented smoothing and numerical differentiation techniques.

Bink, B. "Additional Studies on Physical Working Capacity in Relation to Working Time and Age." Proceedings of Second International Congress on Ergonomics. Dortmund (1964), 83-86.

Although inconclusive, this study did present a graph relating the decrease of maximum oxygen intake with increase in age.

Blacharski, P.A.; Somerset, J.H.; and Murray, D.G. "A Three-Dimensional Study of the Kinematics of the Human Knee." Journal of Biomechanics. Volume 8, Number 6 (1975), 375-384.

Five cadaver knee joints were used to document a three dimensional kinematic study. Interesting graphs of the path of the instantaneous center of rotation were presented.

Bobbert, A.C. "Energy Expenditure in Level and Grade

Walking." Journal of Applied Physiology, Volume 15, Number 6 (November, 1960), 1015-1021.

This author really studied power expenditure as a function of two treadmill variables, speed and incline.

Bouisset, S., and Goubel, F. "Integrated Electromyographical Activity and Muscle Work." Journal of Applied Physiology, Volume 35, Number 5 (November, 1973), 695-702.

"The relationship between the integrated EMG of the biceps (Q) and mechanical work (W) was examined for isolated flexion movements of the elbow."

Bouisset, S., and Lestienne, F. "The Organisation of a Simple Voluntary Movement as Analyzed from its Kinematic Properties." Brain Research, Volume 71 (1974), 451-457.

"The correspondence of the movement performed with the instructions given to the subject, independently of the inertia condition obtained, suggests that the neuro-muscular system functions as an adaptive system."

Bowden, Frank P., and Tabor, David. Friction: An Introduction to Tribology. Garden City: Anchor Press, 1973.

Bowman, B.M.; Schneider, L.W.; and Foust, D.R. "Simulated Occupant Response to Side-Impact Collisions." Society of Automotive Engineers, (1975), Paper 751155.

Bressler, B., and Frankel, J.P. "The Forces and Moments in the Leg During Level Walking." Transactions of the American Society of Mechanical Engineers, 1948, Paper 48-A-62

A summary of the works of Wilhelm and Edward Weber, 1836, E.J. Marey, 1873; Otto Fischer, 1894-1904; N.A. Bernstein, 1935; and H. Elftman, from 1934 on, constituted the initial portion of this paper.

The authors then studied ankle, knee, and hip moments and forces. Force plate data was gathered simultaneously with the taking of 35mm motion pictures at 40 frames per second. Fischer's coefficients gave the weight of each segment, distance from the proximal joint to the center of mass of the segment, and radius of gyration of each segment of the subject's lower extremities.

Combination of the force plate, displacement, and anthropometric data allowed

solution for the ankle, knee, and hip moments and forces.

Cappozzo, Aurelio; Leo, Tommaso; and Pedotti, Antonio. "A Generalized Computing Method for the Analysis of Human Locomotion." Journal of Biomechanics, Volume 8, Number 5 (September, 1975), 307-320.

This article was used to describe human gait analysis via a kinematic analysis. The model used was that of a single leg confined to move in the sagittal plane. The leg itself was comprised of 3 rigid segments articulated at the hip, knee, and ankle. The five coordinates used to describe the planar motion were the two linear coordinates describing the path of the hip and the absolute angle of each of the three segments.

Displacement data was gathered photographically by superimposing images. Illumination was provided by strobe lights flashing at 20 Hz. This rather low value was chosen because the authors felt uncertainty in the data gathering process put an upper bound on the sampling rate.

Although Winarski disagreed with this very low rate, he felt its basis was excellent.

Sensitivity coefficients were derived as part of an error analysis. The range of error was portrayed graphically about the kinematically evaluated joint torques.

Cappozzo, A. Et al. "The Interplay of Muscular and External Forces in Human Ambulation." Journal of Biomechanics, Volume 9, Number 1 (1976), 35-43.

The authors applied a model confined to motion in the sagittal plane. The model itself consisted of a HAT, a pelvis, a thigh, a shank, and a foot. The motion of one leg was recorded photographically and each test subject bore four markers positioned at the hip, knee, ankle, and metatarso-phalangeal joints facing the camera. A force plate gave floor reactions and muscle activity was monitored via three EMG surface mounted electrodes.

The work done by the muscular forces on each joint was calculated by integrating joint torque as a function of limb displacement. This quantity was then erroneously compared to the body potential and kinetic energies via the equation stating that work equals the change in potential and kinetic energy.

It appeared that the authors should have applied the first law of thermodynamics to show that the cyclic change in potential and kinetic energies was negligible. In the real world, most muscular work was converted to heat. Furthermore,

all muscles expend energy. None store it. Hence, the authors' state of negative muscular work never existed.

Carnahan, Brice, and Wilkes, James O. Digital Computing Fortran IV, WATFIV, and MTS. Ann Arbor: University of Michigan, 1973.

Carnahan, Brice, and Wilkes, James O. Introduction to Algorithms and Numerical Methods. Ann Arbor: University of Michigan, 1969.

Cavagna, G.A., and Margaria, R. "Mechanics of Walking." Journal of Applied Physiology, Volume 21, Number 1 (January, 1966), 271-278.

Cavagna, G.A.; Saibene, F.P.; and Margaria, R. "External Work in Walking." Journal of Applied Physiology, Volume 18 (1963), 1-9.

The authors measured external work done on the body at the product of the mass of the body and the inner product of the center of mass acceleration and displacement vectors.

Cavanagh, Peter R., and Gregor, Robert J. "Knee Joint Torque During the Swing Phase of Normal Treadmill Walking." Journal of Biomechanics, Volume 8, Number 5 (September, 1975), 337-344.

The model used consisted of a rigid shank-foot connected to the thigh by a frictionless pin joint. Only the swing phase was investigated.

The authors photographed the swing phase of 5 subjects with a LoCam camera at 97 fps. Simulatneous E.M.G. data was gathered at 5 kHz.

The single segment model was free to roam in the sagittal plane. Three equations of motion were derived for it. The photographed motion was used as kinematic input for these equations. The authors then solved them in an algebraic manner for knee forces and flexion-extension torque.

The authors reached two conclusions. "Firstly, both the torque and E.M.G. Results show a dependence upon speed which emphasizes the importance of considering the speed of walking as a major variable in locomotion studies. Secondly, the results indicate that the variations in the magnitude of torque due to muscular activity produced at the knee can be rationalized between subjects if the relevant body segment parameter - in this case the moment of inertia of the shank and foot about the knee - is considered."

The interesting thing about this article was the use of surfaces to portray functions of the two

variables, time and speed.

Cavanagh, P.R., and Grieve, D.W. "The Release and Partitioning of Mechanical Energy during a Maximal Effort of Elbow Flexion." Proceedings of the Physiological Society, (June, 1970), 44-45.

Chace, M.A., and Angell, John C. Users Guide to Dram (Dynamic Response of Articulated Machinery). Ann Arbor: University of Michigan, 1973.
This manual was used in the graduate level course on advanced dynamics at The University of Michigan, Mechanical Engineering 540.

Chaffin, Don B. "A Computer Biomechanical Model - Development of and Use in Studying Gross Body Actions." American Society of Mechanical Engineers, June 1969, Paper 69-BHF-5.
A computerized model of the human body was developed for analysis of joint torques and forces exerted during various material handling tasks.
The model consisted of eight parts, including ground, resting in the sagittal plane. There were no differential equations of motion as inertial forces were neglected. Only the weights of the seven body segments and the forces required to quasistatically move the test materials were considered. Hence, the description of this model was static sagittal plane.
Application of this model to lifting tasks performed by thirteen volunteers gave results concerning lifting and the human spine.

Chao, Edmund Y.S. "Functional Evaluation of Total Knee Replacement Patients through Gait Analysis." American Society of Mechanical Engineers, June, 1975, Paper 75-APMB-5.
In order to study the gait of patients before and after total knee replacement, a gait laboratory was constructed. Impressive, in nature, the laboratory consisted of a specially designed walkway with overhanging moveable data tracks, a dynamic force plate mounted flush and concealed in the middle of the walkway, knee electrogoniometers, footswitches, and a knee muscle torque tester.

Chao, Edmund Y.S., et al. "The Application of 4x4 Matrix Method to the Correction of the Measurements of Hip Joint Rotations." Journal of Biomechanics, Volume 3, Number 4 (July, 1970), 459-471.
The authors' goniometer linkage and the subject's femur formed a seven link spatial mechanism. This mechanism had two ball joints, two

hinged rotational contacts, and one sliding contact.

In order to mathematically model the seven link mechanism, the two ball joints were each replaced by three hinged rotational contacts. The model was then represented by nine 4x4 matrices: eight matrices represented hinged rotational contacts; one matrix, a sliding contact.

The basis for the 4x4 matrix method was that the product of these nine matrices, in order of their position around the closed loop comprising the seven link mechanism, must be the 4x4 identity matrix.

Iterations similar to the Newton Raphson method of successive approximations were able to correct the measurements of hip joint rotations.

Chao, Edmund Yee-Su, and Rim, Kwan. "Application of Optimization Principles in Determining the Applied Moments in Human Leg Joints During Gait." Journal of Biomechanics, Volume 6, Number 5 (September, 1973), 487-510.

In order to solve kinematic problems, the authors proposed an iterative process alternating between dynamic and kinematic solution of the differential equations of motion.

The initial step was to kinematically solve the differential equations of motion based on observed displacement data and suitable numerical differentiation techniques. This kinematic analysis solved for the set of torques and forces which produced the observed displacements.

This output from the kinematic analysis became the input to a dynamic analysis, the output of which was displacement data.

Now a comparison of the displacement data used by the kinematic analysis and the displacement data predicted by the dynamic analysis was made. This comparison was based on the time integral of the sum of the squares of the error observed in each coordinate.

This measure of error and a set of adjoint equations enabled the iteration process to modify the displacement data originally observed until there existed reasonable convergence between the displacement data used as kinematic input and the displacement data obtained as dynamic output.

The model used in this integration process consisted of a planar, three degree of freedom, stick figure.

Chino, Naoichi; Pearson, J. Raymond; Cockrell, James L.; Mikishko, Harry A.; and Koepke, George H.

"Negative Pressures During Swing Phase In Below-Knee Prostheses with Rubber Sleeve Suspension." Archives of Physical Medicine and Rehabilitation, Volume 56 (January, 1975), 22-26.

Since there does not exist a state of negative pressure, an applicable title would have involved description of pressures less than ambient or that of gage pressures.

The article did go on to describe that pressures less than ambient did exist in the small space between the distal stump and the below-knee prosthesis socket during the swing phase of a series of 9 subjects.

A molded rubber sleeve was found to act as a seal. By preventing air leakage between the prosthesis and stump, pressures less than ambient were maintained during the entire swing phase. This tended to improve overall comfort and function for the amputee.

Chow, C.K., and Jacobson, D.H. "Further Studies of Human Locomotion: Postural Stability and Control." Mathematical Biosciences, Volume 15 (1972), 93-108.

In this paper, equations of motion for a three dimensional, six degree of freedom, inverted pendulum were derived. Next, the authors simplified these equations of motion and wrote them in first order, non-linear, canonical form using four state variables. Feedback stabilization of the otherwise unstable inverted pendulum was proposed. Liapunov's second method was used to show that the inverted pendulum with feedback was asymptotically stable under all initial conditions. Finally a numerical example was discussed.

Chow, C.K., and Jacobson, D.H. "Studies of Human Locomotion via Optimum Programming." Mathematical Biosciences, Volume 10 (1971), 239-306.

The study of the human gait cycle was classified into two major areas by Winarski. The first area consisted of physiological studies of the muscles and the manner and duration they acted in order to allow locomotion to occur. Winarski termed this microscopic level of analysis as kinetic.

Winarski's second area of study of the human gait cycle was of a postulated model of the human body. Under this major area fell three categories of analysis. Winarski classified each of these categories as to the manner in which the set of torques and forces and the set of coordinates describing the motion were manipulated.

One category occurred when, in the

differential equations of motion, the known quantities were the set of torques and forces and the unknown quantities were the set of coordinates describing the motion. This category of analysis, known as dynamics, was characterized by the numerical methods of matrix inversion or augmented matrices in hermite normal form in order to solve for the unknowns.

The second category contained the analysis in which the set of coordinates decribing the motion were known and the set of torques and forces were treated as unknowns. This second category was labeled kinematics.

The final category was the one chosen by the authors. The set of torques and forces and the set of coordinates describing the motion were treated as unknowns. The authors' approach was to develop enough restrictions and relationships so that the system of differential equations was solved for both the set of torques and forces and the set of coordinates describing the motion.

The model in this analysis was planar, resting in the sagittal plane.

Coblentz, A., and Ignazi, G. Anatomical and Functional Measurements, Criteria for Human Adaptation within a Driver's Control Area." Society of Automotive Engineers, February, 1975, Paper 750422.

The authors studied the extremes of the anthropometric measurement distribution. The foremost measurement considered was stature.

Contini, Renato. Body Segment Parameters (Pathological) New York, New York University, 1970.

With only the subject's weight and height, it was possible to statistically determine the length of the subject's body segments, the mass and the location of the center of mass of these segments, and the segment's mass moment of inertia at the center of mass about an axis normal to the sagittal plane.

However, in order to reduce error inherent in the application of statistics to individuals, Winarski considered it desirable to measure limb lengths and circumferences directly and to use skinfold measurements to determine the body density.

The procedure for gathering this data directly from the subject was to repeat Contini's first twenty three measurements. This information and the subject's age yielded more accurate information about the subject than a purely statistical approach. Furthermore, favorable

comparisons of empirical versus theoretical segment lengths gave credibility to the use of statistical parameters for determination of segment mass and segment center of mass.

Corcoran, P.J., and Brengelmann, G.L. "Oxygen Uptake in Normal and Handicapped Subjects, In Relation to Speed on Walking beside Velocity-Controlled Cart." Archives of Physical Medicine, Volume 51, Number 2 (February, 1970), 78-87.

Rather than using a motorized treadmill, which the authors felt would give misleading results, the SCRAM was developed. This was the Speed-Controlled Respirometer for Ambulation Measurement.

The authors compared their results for power expenditure with the results of other researchers.

Ralston (1958) $P = 6.00 + 0.367V^2$

Bobbert (1960) $P = 6.72 + 0.325V^2$

SCRAM (1970) $P = 6.15 + 0.336V^2$

Power (P) was oxygen consumption in millilitres/kg/minute; velocity (V), feet/second.

This article was a highly informative comparison of the thrusts of three independent research groups on the power requirements of gait. It also gave an energy equation by integration of the SCRAM power equation with respect to time.

$E = 0.337/V + 0.0184V$

Energy (E) was oxygen consumption measured in millilitres/kg/metre; velocity (V), feet/second.

Twice differentiation of this equation with respect to velocity showed V = 4.28 feet/second to be the optimum velocity where energy expenditure was minimized.

Corell, Edward B. Normal Pressure Distributions Applied by Total Contact Prostheses. Technical Report 9. The University of Michigan Medical School, 1969.

The author discussed the implementation of a system to monitor B/K stump normal pressures. Included in his work was the data acquisition program Sample, the precursor of Sample8 as used by Winarski.

The instrumentation used in this research included the Kulite normal pressure transducers, footswitches, hip and knee goniometers, and the Linc-8/PDP8 mini-computer.

Craig, Homer Vincent. Vector and Tensor Analysis. McGraw Hill, New York, 1943.
Starting from real numbers, this well written text covered invariance, vector algebra and calculus, integral theorems, general coordinates, introductory differential geometry, tensor and extensor analysis, Riemannian geometry, and invariance.

Cramer, H.J.; Liu, Y. King; and von Rosenberg, D.U. "A Distributed Parameter Model of the Inertially Loaded Human Spine." Journal of Biomechanics, Volume 9, Number 3 (1976), 115-130.
The authors derived a one-dimensional continuum model of the spine. It was considered a curved homogeneous beam-column and was subjected to a distributed eccentric inertial loading of the human torso. The model was then studied in terms of small strain, large deflection.

Cunningham, Don M., and Brown, G. Wayne. "Two Devices for Measuring the Forces Acting on the Human Body During Walking." Proceedings of the Society of Experimental Stress Analysis, Volume 9, Number 2 (1952), 75-90.
The purpose of this research was to instrument the pylon of an above knee amputee with strain gages and compare the force and torque results to simultaneous force plate readings. This paper became the foundation for the contribution of Mr. James Wisner to Winarski's research.

Datta, S.R.; Chatterjee, B.B.; and Roy, B.N. "The Relationship between Energy Expenditure and Pulse Rates with Body Weight and the Load Carried During Load Carrying on the Level." Ergonomics, Volume 16, Number 4 (July, 1973), 507-513.
In a study of load carrying on the head, the authors collected expired air in a Douglas bag, determined oxygen consumption from a Haldane gas apparatus, and determined energy expenditure by the method of Weir (1949).

Dean, G.A. "An Analysis of the Energy Expenditure in Level and Grade Walking." Ergonomics, Volume 8, Number 1 (January, 1975), 31-47.
A piecemeal model was thrown together to try to analyze gait energy expenditure. One of the many unpalitable assumptions made involved immobilizing the knees to make the legs rigid.

Dearing, LeRoy M. "Film Data Correlation - An Overview

Utilizing Timing Systems." Proceedings of the Society of Photo-optical Instrumentation Engineers. Edited by Brown and Patrick. Volume 34 (November, 1972), 129-132.

Time marking of motion picture film for later correlation with data gathered from other sources was considered. Instrumentation discussed included light emitting diode writeouts, binary coded decimal patterns, and time pulse generators.

The light emitting diode writeouts were complex and informative in nature, giving date-time groups in alphanumbric symbols. The binary coded decimal patterns were less informative. These printouts needed decoding to be understood. The least informative of all was the time pulse generator, which simply marked increments of time on the film.

Dempster, W.T. "Free Body Diagrams as an Approach to the Mechanics of Human Posture and Motion." Biomechanical Studies of the Musculo-skeletal System. Edited by Evans, et al. Charles. C. Thomas, Publisher, 1961, pp. 81-135.

The importance of the article rested in the discussion of each segment of the body and its relationship to the whole body center of gravity. This was especially apparent when the subject bended at the waist and the whole body center of gravity was external to the body.

Dempster, W.T. Space Requirements of the Seated Operator. Wright Air Development Center Technical Report 55-159, Wright-Patterson Air Force Base, Ohio, 1955.

Important in this rather large report are the parameters denoting segment length, mass, and center of mass for the human body.

Dempster, W.T., and Caughran, George R.L. "Properties of Body Segments Based on Size and Weight." Americal Journal of Anatomy , Volume 120 (1967), 33-54.

This article was an extremely important contribution to the study of anthropommetry, giving more statistically formulated anthropometric parameters based on cadaver studies than any of Dempster's previous articles.

Easthope, C.E. Three Dimensional Dynamics. London: Butterworth & Company Ltd., 1964.

Eberhart, Howard D., and Inman, Verne T. "An Evaluation of Experimental Procedures Used in a Fundamental Study of Human Locomotion." Annals of the New York Academy of Sciences. Volume 51, Article 7 (January,

1951), 1213-1228.

The following procedures for gathering data about the human gait cycle were discussed in detail; motion pictures using ordinary light, motion pictures using X-rays, electromyography, force plate studies, and instrumented pylon studies.

Elftman, Herbert. "A Cinematic Study of the Distribution of Pressure in the Human Foot." The Anatomical Record, Volume 59 Number 4 (1930), 481-487.

The pressure transducer used in this study was a black rubber mat studded with pyramidal projections upon a heavy glass plate. The author studied the plantar surface pressure on human feet by noticing the increase in area of each pyramid in contact with the glass. A reflecting fluid was introduced between the black pyramids to allow photographic recording of the pressure distribution at 72 frames per second.

Elftman, Herbert. "Dynamic Structure of the Human Foot." Artifical Limbs, Volume 13, Number 1 (Spring, 1969), 49-58.

This article was a very informative description of the human foot's role in the gait cycle. Of particular importance was the discussion of the traverse of the center of pressure from the heel to the great toe and isobars depicting the distribution of the plantar surface pressure.

Elftman, Herbert. "Forces and Energy Changes in the Leg During Walking." The American Journal of Physiology, Volume 125, Number 2 (February, 1939), 339-350.

This author discussed the various forces and the required muscle work as a function of the change in the sum of the potential and kinetic energies. Although the author correctly surmized that muscles must do the work to increase the body's sum of potential and kinetic energy, he erroneously felt that muscles were capable of storing energy when the body energy level was reduced.

What actually happened was muscles expended energy to reduce or dissipate the body's energy level. Muscles couldn't be considered as elastic springs, but as living entities which exert a force only with a continuous metabolism.

If one were really serious about Elftman, all we would have to do would be to fly to a Colorado mountain top, run down it, and have all of that nice potential energy stored in our muscles.

That way we would all look like Hercules until we used the stored up energy to leap tall buildings in a single bound, etc.

Elftman, Herbert. "Skeletal and Muscular Systems: Structure and Function." Medical Physics. Edited by Otto Glasser. Chicago: Yearbook Publishers, 1944, pp. 1420-1430.

This article was an informative summary of the structure physical properties, functions, and energy demands of the human musculoskeletal system.

Elftman, Herbert. "The Basic Pattern of Human Locomotion." Annals of the New York Academy of Sciences, Volume 51, Article 7 (January, 1951), 1207-1212.

This author presented a verbal description of what would be mathematically described by the set of differential equations of motion of a biped.

Elftman, Herbert. "The Function of Muscles in Locomotion." The American Journal of Physiology, Volume 125, Number 2 (February, 1939), 357-366.

Four ways muscles participated in energy exchange were by receiving energy, dissipating energy, doing work, and "tendon action." By tendon action, the author referred to the paradox of the muscle exerting a tension and thereby using energy, however, with zero displacement, no work being done.

Elftman, Herbert. "The Work Done by Muscles in Running." American Journal of Physiology, Volume 129, Number 1 (April, 1940), 672-684.

By means of photographic analysis and graphical differentiation of the displacement data, the static and inertial forces and torques exerted by a runner were determined. The runner's work rate was calculated by multiplying the instantaneous joint torques by the rate at which the relative angles between the appropriate segments were changing.

This article ended with an acknowledgement that muscles expended energy in order to exert a contractive force. This energy expenditure occurred whether the total body potential and kinetic energy level was raised or lowered. Unlike Elftman (1939), lowering of the total body energy was not stored in the muscles.

Noted in this article was that the figure of 0.4 was used to convert so-called negative work into a positive energy expenditure. Anderson (1974) used a figure of 0.33.

Erdman, W.J.; Hettinger, T.; and Saez, F. "Comparative Work Stress for Above-Knee Amputees using Artificial Legs or Crutches." American Journal of Physical Medicine, Volume 39, Number 6 (December, 1960), 225-232.

This author measured energy consumption via the Max Plank Institute gas meter. His values were in the ball park of 1 calorie/metre/kg.

Erickson, Lester, et al. "The Energy Cost of Horizontal and Grade Walking on the Motor-Driven Treadmill." American Journal of Physiology, Volume 145, Number 3 (January, 1946), 391-401.

These authors documented energy expenditure on a treadmill as a function of two variables. The first was the average speed of the treadmill; the second, its incline.

Fenn, Wallace O. "Frictional and Kinetic Factors in the Work of Sprint Running." American Journal of Physiology, Volume 92, Number 3 (April, 1930), 583-609.

This author gathered motion picture data for his analysis. Runners passing in front of a reference lattice were photographed by a hand cranked camera. Frame timing was achieved by the dropping of wooden croquet balls.

This author wisely described the reduction of body energy by its conversion to heat.

Also tabulated in this article was a useful list of conversion factors for assigning positive energy expenditure to instances of so-called negative work.

Fenn, Wallace O. "Work Against Gravity and Work Due to Velocity Changes in Running." The American Journal of Physiology, Volume 93, Number 2 (June, 1930), 433-462.

Gage, Howard. "Accelerographic Analysis of Human Gait." American Society of Mechanical Engineers, December, 1964, Paper 64-WA/HUF-8.

This author simultaneously gathered recordings of the horizontal and vertical components of acceleration of the human trunk and the angular acceleration of the shank. The data was reduced via harmonic analysis and normalized first by finding the root mean square of the sine and cosine coefficients of each harmonic.

The final normalization procedure was to set the root mean square coefficient of the second harmonic to 1.000 and recompute the remaining root mean square coefficients as a percentage of this

second harmonic.

Ganguli, S., et al. "Ergonomics Evaluation of Above Knee Amputee Prosthesis Combinations." Ergonomics, Volume 17, Number 2 (March, 1974), 199-210.

This article was quite similar to the one written by Ganguli, 1973, about below knee amputee ergonomics evaluation. As before, it was well written and informative.

Ganguli, S., et al. "Performance Evaluation of an Amputee - Prosthesis System in Below Knee Amputees." Ergonomics, Volume 16, Number 6 (1973), 797-810.

The authors measured pulmonary ventilation, oxygen consumption, energy expenditure, and peak heart rate during below knee amputees performance of four types of standard pysical activities and one standard exercise test.

It was observed that the oxygen consumption and energy expenditure were determined by collecting expired air in a Douglas bag and analyzing it in a Haldane gas apparatus.

The four types of standard physical activities were sitting upright in a chair, standing erect from the sitting position, walking on level ground, and ascending stairs. The standard exercise test was the steps test at the rate of fifteen steps per minute on a twenty centimeter high stool for a period of ten minutes.

Gavrilovic, M.; Maric, M.; and Vukobratovic, M. "An Approach to the Synthesis of Lower Extremity Control." International Symposium on External Control of Human Extremities. Belgrade, Yugoslav Committee for Electronics and Automation, 1967, pp. 206-211.

Two differential equations of motion were derived for a fixed hinge, planar, two degree of freedom, compound pendulum. A constraint to the compound pendulum's motion was that the lower part could not penetrate the straight line which constituted the floor. An analog computer simulated the problem dynamically.

Ghosh, T.K., and Boykin, W.H. "Dynamics of the Human Kip-up Maneuver." American Society of Mechanical Engineers, November, 1974, Paper 74-WA/AUT-7.

A gymnastic exercise was modeled using a three link, planar representation. Anthropometric measurements were made in accordance with Hanavan.

The subject was observed doing the kip-up maneuver from two planes, the frontal and sagittal. During this observation, film speeds were at 32 and

64 frames per second. Direct differentiation of the displacement data was limited by using Hamilton's canonical equations in the derivation of the differential equations of motion.

The purpose of this research was to compare an analytical and empirical minimum event completion time. The discussion of the automatic control theory used in the analysis was most interesting and informative, as was the rest of the article.

Gonzalez, Erwin G.; Corcoran, Paul J.; and Reyes, Rodolfo, L. "Energy Expenditure in Below-Knee Amputees: Correlation with Stump Length." Archives of Physical Medicine and Rehabilitation, Volume 55, Number 1 (March, 1974), 111-119.

Oxygen consumption during ambulation was determined for nine B/K amputees. According to the authors, stump length had no influence on the comfortable walking speed. However, they did notice that the shorter the B/K stump, the more energy spent during gait. Comparisons with previous efforts concerning bilateral B/K and unilateral A/K amputees are then offerred.

Energy expenditure was expressed as a function of velocity in the typical manner, as the sum of a straight line and a hyperbola.

$$\text{Energy} = 0.337/V + 0.0184V$$

In this expression, energy was measured in millilitres of oxygen consumption per metre traveled per kilogram of subject mass. Velocity (V) was measured in feet per second. The equation did have a minimum energy expenditure velocity, at 4.28 feet/second.

Grevsten, Sven, M.D., and Eriksson, Uno. "Stump-Socket Contact and Skeletal Displacement in a Suction Patellar-Tendon Bearing Prosthesis." The Journal of Bone and Joint Surgery, Volume 56-A, Number 8 (December, 1974), 1692-1696.

The authors' study of the suction, patellar-tendon-bearing prosthesis showed that there was better contact between the stump and the socket of the prosthesis than there was with a strap-suspended patellar-tendon bearing prosthesis.

The average difference in stump/socket local displacement between the two was 1.15 cm. This occurred primarily due to the difference in the movement of the soft tissue. Reduced movement with the suction socket tended to give a better sense of stability and a smoother gait.

The experimental method consisted of roentgenographic examination of 22 subjects in 4 discrete weight bearing positions. Values of stump linear and angular displacement were given for each of the four positions. Since the analysis was static, viscoelastic effects important during gait were ignored. Still, this paper was most illuminating, having documented stump static deformations.

Grieve, D.W. "The Assessment of Gait." Physiotherapy, Volume 55 (November, 1969), 452-460.

This author discussed plotting thigh absolute angles (with respect to the vertical) versus the thigh-shank relative angles. These plots resembled state space variable plots and appeared to be quite useful. The author used cine film to gather his required displacement data. The possible future use of a polgon, or polarized light goniometer, to gather the limb angular displacements at a frequency of fifty times a second was briefly mentioned.

Grieve, D.W., and Arnott, A.W. "The Production of Torque During Axial Rotation of the Trunk." Journal of Anatomy, Volume 107, Part 1 (July, 1970), 147-164.

Subjects exerted torques on a horizontal bar which was suspended at chest height. The bar was driven through an angular displacement of 120 degrees. The torques which the body had to exert in order to accelerate the bar through the first 60 degrees of bar movement were then calculated.

The movements of the bar and pelvis were recorded at 48 frames per second on 16mm cine film. A 25mm lens was used. The camera was 6 meters from the subject with the optical axis horizontal and perpendicular to the frontal plane.

Grieve, D.W., and Cavanagh, P.R. "The Influence of Electrode Position upon Quantitative Description of Surface Electromyograms in the Lower Limbs." Journal of Anatomy, Volume 111, Part 2 (February, 1972), 350.

This article discussed the comparison of the EMG reading from a pair of electrodes with what was really happening in the lower limb muscles. A 3x4 matrix of 12 simultaneously read bipolar EMG readings were used as the reference. Factors considered were signal gradient and underlying anatomy.

Grieve, D.W., and Gear, Ruth J. "The Relationships Between Length of Stride, Step Frequency, Time of Swing,

and Speed of Walking for Children and Adults." Ergonomics, Volume 5, Number 9 (September, 1966), 379-399.

This work was an attempt to statistically establish relationships between step frequency, stride length, time of swing, and walking speed of humans from 1 to 35 years of age.

Gubina, Ferdinand; Hemami, Hooshang; and McGee, Robert B. "On the Dynamic Stability of Biped Locomotion." I.E.E.E. Transactions of Biomedical Engineering, Volume BME-21, Number 2 (March, 1974), 102-108.

The model used consisted of a trunk possessing mass and mass moment of inertia which was supported by massless, extensible legs. This biped was confined to motion in the sagittal plane, giving it two translational and one rotational degrees of freedom. Combining a set of restrictions with the differential equations of motion allowed solution for both the set of torques and forces and the set of coordinates describing the motion.

Hanavan, E.P., Jr. "Personalized Mathematical Model of the Human Body." Journal of Spacecraft, Volume 3 (1968), 446-448.

Hatze, H. "A Method for Describing the Motion of pbiological Systems." Journal of Biomechanics, Volume 9, Number 2 (1976), 101-104.

The author extolls the virtues of a Fourier analysis on biomotion. Convergence of Fourier series was guaranteed, since biomotions always satisfy the Dirichlet Conditions. Furthermore, a least square approximation was easily achieved by simple truncation of the series.

Hill, A.V. "The Absolute Mechanical Efficiency of the Contraction of an Isolated Muslce." Journal of Physiology, Volume 46, Number 6 (August, 1913), 435-469.

This article, in Winarski's opinion, had certain hard to swallow concepts. One of them was that a contracted muscle, in tension, was "simply a new elastic body of changed elastic properties, and possessing potential energy which can be turned into work if we desire , but which otherwise will degerate in a few hundreths of a second into heat."

Still, it was rewarding to realize that early bio-experimenters considered that muscle metabolism would be converted into work or heat.

Hill, A.V. "The Energy Degraded in the Recovery Process of

Stimulated Muscles." Journal of Physiology, Volume 46, Number 1 (March, 1913), 28-80.

The major transducer in this article was the iron-constantan thermocouple . It was interesting that this was the one transducer that Winarski wanted for monitoring the B/K stump temperature rise during his experiments. Unfortunately, overburdened with monitoring many other transducers, Winarski was unable to gather this temperature rise.

The whole idea of Hill's experiment was to excite a muscle and then measure the heat output of the muscle as a function of time.

This meant Hill measured the global heat expenditure of the muscle. It was a function of the local heat production of the muscle fibers and heat transfer dynamics.

Hill, A.V. "The Maximum Work and Mechanical Efficiency of Human Muscles, and their Most Economical Speed." The Journal of Physiology, Volume 56, Numbers 1 and 2 (February, 1922), 19-41.

An interesting conclusion concerning the mechanical efficiency of human voluntary movement was discussed. It showed that there was a certain optimum speed of movement below which muscle efficiency falls slowly, and above which it falls rapidly.

Unlike Hill (1913), muscles were considered as thermo-elastic entities.

It was interesting to compare Hill's maximum muscle efficiency of 26% to Ralston and Lukin's (1969) efficiency of walking, 23%.

Hubbard, Robert P. "Anthropometric Basis of the GM ATD 502 Crash Test Dummy." Society of Automotive Engineers, February, 1975, Paper 750429.

In discussing the guidelines for selection of the specific dummy segment lengths, weights, and centers of gravity, special emphasis was placed on the human spine. Recommendations for further anthropometric reasearch were also presented.

Hughes, H. "Dynamic Gait Studies Applied to the Design of Lower Limb Prostheses." International Symposium on External Control of Human Extremities, Belgrade, Yugoslav Committee for Electronics and Automation, 1967, pp. 183-192.

In this analysis, ground reaction forces of the subject were measured with a force platform. A simulation of the subject was made using a kinematic model. Data was gathered using two cameras: one camera viewing the sagittal plane; the

other, the frontal plane. Fischer's coefficents were used to give approximate weights of limb segments in terms of the total weight of the subject. The photographic displacement data was twice differentiated to give velocity and acceleration. In addition to determining the gravitational forces, Fischer's coefficients were used to determine the inertial forces. With this information it was then possible to calculate the forces and torques exerted at the hip joint of both normal and amputee subjects.

It was concluded that minimum energy expenditure should be the criterion used to control the amputee's gait rather than trying to make the prosthesis bearing leg walk symmetrically with respect to the normal leg.

Huston, R.L., and Passerello, D.E. "On the Dynamics of a Human Body Model." Journal of Biomechanics, Volume 4, Number 5 (October, 1971), 369-378.

The model consisted of sixteen rigid parts, including ground, in Euclidean three space. The equations of motion were then derived with the aid of vector and tensor analysis. Extensive hand computation of the differential equations of motion was avoided by programming the computer to derive the differential equations as well as solve them.

Three practical applications were then discussed.

Ingels, Neil B.; Rush, Stephen K.; and Thompson, Noel P. "Analytical Stop Motion Stereo Photogrammetry." The Review of Scientific Instruments, Volume 40, Number 3 (March, 1969), 487-492.

This paper described a method of locating points in Euclidean three space by using a triangularization procedure and two cameras observing two distinct views.

Isman, R.E., and Inman, V.T. Anthropometric Studies of the Human Foot and Ankle. Biomechanics Laboratory: University of California, San Francisco and Berkeley, May, 1968, Technical Report 58.

A study of the talocrural (ankle) and talocalcaneal (subtalar) joints was made in order to acquire information of the exact locations and orientations of their axes of rotation. The study used 46 cadaver legs.

Jacobs, N.A.; Skoreckl, J.; and Charney, J. "Analysis of the Vertical Component of Force in Normal and Pathological Gait." Journal of Biomechanics, Volume 5, Number 1 (January, 1972), 11-34.

These authors focused their attention on the vertical component of force exerted in walking by normal and pathological subjects. Fourier analysis was used to reduce the data.

Further analysis required the following information for the left and the right foot: vertical force plot; period, stance phase, swing phase, both double support phases (when foot either leading or trailing), single support phase, phase of applying weight, phase of relieving weight, phase of carrying weight, body weight, weight carried by foot, maximum force, minimum force, inertial contribution, rate of applying weight, rate of relieving weight, and area of trace.

Jensen, R.H., and Davy, D.T. "An Investigation of Muscle Lines of Action about the Hip: A Centroid Line Approach vs the Straight Line Approach." Journal of Biomechanics, Volume 8, Number 2 (March, 1975), 103-110.

A comparison was presented of representing skeletal muscles as centroid lines and as straight lines. The centroid lines were determined by tracing the area midpoint of muscle cross sections from the muscle origin to insertion.

It was interesting to compare the centroid approach of this article to the straight line approach of Seireg and Arvikar (1975).

Juricic, Davor, and Vukobratovic, Miomir. "Mathematical Modeling of a Bipedal Walking System." American Society of Mechanical Engineers, November, 1972, Paper 72-WA/BHF-13.

The model described in this paper consisted of sixteen rigid parts, including ground, moving in Euclidean three space. Rather than defining a vector from an inertial reference frame to the body's torso, these authors defined this important vector's location to be from the inertial reference frame to one of the biped's feet.

Kane, T.R.; Headrick, M.R.; and Yatteau, J.D. "Experimental Investigation of an Astronaut Maneuvering Scheme." Journal of Biomechanics, Volume 5, Number 4 (July, 1972), 313-320.

This paper described the experiments used to verify and apply the theoretical predictions of Kane, 1970, concerning an astronaut maneuvering scheme.

Kane, T.R., and Scher, M.P. "Human Self-rotation by Means of Limb Movement." Journal of Biomechanics, Volume 3, Number 1 (January, 1970), 39-49.

By using models consisting of a rigid trunk and two rigid limbs free to move in the absence of external forces, the authors showed how it was possible for astronauts to orient themselves.

The differential equations of motion for both models presented were derived in Euclidean three space.

Kinzel, G.L., et al. "Measurement of the Total Motion Between Two Body Segments - II: Description of Application." Journal of Biomechanics, Volume 5, Number 3 (May, 1972), 283-293.

Using a seven bar linkage with six hinged rotational contacts, the authors were able to ascertain all six degrees of freedom of a dog's scapula-humerus joint.

Link 1 was the scapula; link 7, the humerus. The other five links were metal. Six rotary potentiometers measured the angular displacements of the six hinged, rotational contacts.

Klopsteg, P.E., and Wilson, P.D. Human Limbs and Their Substitutes. New York: Hafner Publishing Company, 1968.

Krouskop, T.A., and Newell, P.H., Jr. "A Review of Biomaterials Engineering." American Society of Mechanical Engineers, July, 1973, Paper 73-WA/BIO-36.

A well written survey article, the authors did much to explain the many facets in the selection of appropriate biomaterials. A substantial bibliography was then presented.

Lamoreux, Larry W. "Kinematic Measurements in the Study of Human Walking." Bulletin of Prosthetics Research, Volume BPR 10-15 (Spring, 1971), 3-84.

This author took his data in relative coordinates. The data gathered was averaged over forty steps.

The author did present a very useful foot switch for the subject's heel.

LaVigne, A.B. "A New Arthrograph for More Accurate Measurement of Joint Motion and Stiffness." Journal of Biomechanics, Volume 8, Number 5 (September, 1975), 345-346.

This author described his arthrograph, a mechanism for measurement of joint motion and forces required to produce that motion.

Leader, Lynn R., and Salisbury, Richard A. Subroutine and Macro Descriptions, Volume III of The Michigan

Terminal System. Ann Arbor: The University of Michigan Computing Center, 1973.

Leavitt, Lewis, A. "A Quantitative Method to Measure the Relationship Between Prosthetic Gait and the Forces Produced at the Stump Socket Interface." American Journal of Physical Medicine, Volume 49, Number 3 (1970), 192-202.

The author described the unique digital to analog voltage outputs for each of eight possible states of his binary weighted foot switches.

Three foot switches and one three bit D/A converter were used on each of the subject's feet. The number two raised to the power of the number of bits, in this case three, gave the number possible output states.

Lewin, T. "Anthropometic Studies on Swedish Industrial Workers When Standing and Sitting." Ergonomics, Volume 12, Number 6 (1969), 883-902.

The basis for this study was the adjustment of work space requirements in accordance with the individual's anthropometry. Understandably, only segment lengths were discussed.

Libertiny, George A., and Catz, Jerome. "Development of a Spindle Force-Moment Transducer," Society of Automotive Engineers, October, 1974, Paper 740938.

The authors described a recently developed load cell. It was capable of measuring the three components of the force vector and two components of a moment vector.

The load cell was in the shape of a spindle. The article described the careful strain gage mounting process, load cell calibration, and expression for output voltage as a function of input loads.

The expression for output voltage as a function of input loads was presented in matrix form. Two matrices were used: one for individual bridge sensitivities; the others, a bridge interaction correction matrix.

Lippert, Frederick G. "The Feasibility of Photogrammetry as a Clinical Research Tool." Journal of Biomechanics, Volume 6, Number 5 (September, 1973), 459-473.

This paper described photogrammetry, a technique which discerned small three dimensional movements on a moving subject. Basically, this method was one of stereo photography.

Two cameras were used to study internal tibial fracture motion during the gait cycle.

Targets were pinned directly to the tibia on each side of the active fracture.

Lipson, Charles, and Sheth, Narendra, J. Statistical Design and Analysis of Engineering Experiments. New York: McGraw Hill, Inc., 1973.

Lukin, L. "Instability of the 'Steady-State' during Exercise." International Zeitschrift fuer angewandte Physiologie einschliesslich Arbeitsphysiologie. Volume 20 (1963), 45-49.

This study showed that the steady state oxygen consumption during exercise fluctuated. Furthermore the degree of fluxuation, expressed in terms of standard deviation, was minimal at moderate work load. It was greater at either extreme.

Malhotra, M.S., et al. "Physical Work Capacity as Influenced by Age." Ergonomics. Volume 9, Number 4 (July, 1966), 305-316.

Experiments conducted with 879 healthy soldiers showed that all the physical functions tested started to show decline on those over 30 years old.

Included in the testing were speed of running, abdominal muscle strength, agility, arm and shoulder muscle strength, and capacity for short bursts of activity.

Margaria, R., et al. "The Kinetics of the Oxygen Consumption at the Onset of Muscular Exercise in Man." Ergonomics. Volume 8, Number 1 (January, 1975), 49-54.

The authors considered exercise as a step-function disturbance and body oxygen consumption as the response. This response was exponential in nature with a half reaction constant of 30 seconds.

Marks, Morton. "Gait Analysis of the Hemiplegic Patient: Force Plate and Stroboscopic Studies." Quarterly Review of Pediatrics. Volume 7, Number 1 (February, 1952), 15-24.

In his studies, Dr. M. Marks used a force plate, interrupted light camera, and an optical oscillograph.

The force plate was composed of two right triangular sections fitted so as to form a square forty inches on a side. The mounting for each plate half consisted of six supported beams, each attached to the plate top with a turnbuckle link. This arrangement allowed measurement of forces via strain gage transducers on uniaxial rather than

complex biaxial loading.

McDonald, Ian "I: Statistical Studies of Recorded Energy Expenditure of Man. II: Expenditure of Walking Related to Weight, Sex, Age, Height, Speed, and Gradient." Nutrition Abstracts and Reviews, Volume 31, Number 3 (July, 1961), 739-762.

The author related gross metabolic cost (H) in Kcal/minute, average speed (V) in metres/minute, and subject mass (M) in kilograms for men and women in level walking. He used a logarithmic form in base 10. The equation for men is listed first; women, second.

$$\text{Log}(H) = 0.0028V + 0.000020V^2 + 0.0033M + 0.072$$

$$\log(H) = 0.0053V + 0.0065M - 0.257$$

It should be noted that mass appeared explicitly in the above equations. This mass dependency and logarithmic expression was quite different from the form presented by Ralston (1958) or Bard and Ralston (1959).

The logarithmic representation of the power expenditure lent itself to numerical evaluation of energy expenditure rather than finding an expression in closed form by integration of power with respect to time.

This numerical procedure was done on the Michigan Terminal System. For a 100.4 kg subject, minimum energy expenditure was 0.706 cal/kg/m at 75 m/min. For a 79.4 kg subject, minimum energy expenditure was 0.761 cal/kg/m at 75 m/min. Finally, for a 58.3 kg subject, minimum energy expenditure was 0.883 cal/kg/m at 75 m/min.

It was interesting to notice the increase in normalized energy requirement at constant optimum velocity with decrease in mass. This increase in normalized energy requirement occurred even though the power/metre requirement decreased with the decrease in mass.

McDonald's results suggested that normalization with respect to 10 to the exponent 0.0033*Mass was more appropriate than mere division by mass (or weight) alone.

McGhee, R.B. "Some Finite State Aspects of Legged Locomotion." Mathematical Biosciences, Volume 2 (1968), 67-84.

Many definitions and theorems were presented, all based on a notion of a binary output state. One state was the swing phase; the other, the stance phase with the leg pushing backwards.

McLeod, P.C. "Measurements of Repetitive Activities of the Knee." Journal of Biomechanics, Volume 8, Number 6 (1975), 369-373.

A most interesting method of gathering knee angular displacement data was discussed. Each subject wore a electrogoniometer. However, transducer output was stored on a small, subject mounted tape recorder. This allowed data to be gathered as each subject went about his normal, daily routine outside of the laboratory.

Minns, R.J.; Bremble, G.R.; and Campbell, J. "The Geometric Properties of the Human Tibia." Journal of Biomechanics, Volume 8, Number 3/4 (July, 1975), 253-255.

Four left human tibiae were obtained from cadavers. They were cross sectioned at 7 equidistant positions along the transverse axis. The cross sectional area, the orientation of the principal axes (eigenvectors), and the values for the second moments of cross sectional area referred to those principal axes (eigenvalues) were determined.

It was interesting to observe the eigenvalues decrease as the cross section location approached the ankle.

Molen, N.H., and Boon, W. "Measurement of Momentary Velocity in a Study of Human Gait." Journal of Biomechanics, Volume 5, Number 3 (May, 1972), 273-276.

These authors compared the tachogenerator and a system using magnetic tape for use in measuring the forward velocity of a subject.

Although the authors claimed to be measuring the forward velocity of the subject's center of gravity, the authors were actually measuring the forward velocity of the point of attachment with respect to an inertial reference frame.

Morris, J.R.W. "Accelerometry - A Technique for the Measurement of Human Body Movements. Journal of Biomechanics, Volume 6, Number 6 (November, 1973), 729-736.

A study of the movement of the shank, or lower leg, using five accelerometers was reported.

An appendix showed how signals from six accelerometers could have been used to completely specify the movement of a body in space. The author omitted one accelerometer since he did not desire to measure transverse rotations of the shank.

Morrison, J.B. "Functions of the Knee Joint in Various Activities." Biomedical Engineering. Volume 4, Number 12 (December, 1969), 573-580.

The analysis was basically the same as described in Morrison, 1970.

Morrison, J.B. "The Mechanics of the Knee Joint in Relation to Normal Walking." JFournal of Biomechanics. Volume 3, Number 1 (January, 1970), 51-61.

Ground reactions forces of the subject were measured with a force platform. A simulation of the subjects was made using a kinematic model. Displacement data for this kinematic analysis was gathered by the use of a camera viewing the sagittal plane and a camera viewing the frontal plane. With this information, the author solved for the torques and forces exerted at the knee joint. Having also gathered electromyographic data on the major leg muscles simultaneously with the force plate and motion picture data, the author was able to calculate the forces exerted on particular muscles, ligaments, and articular surfaces.

In particular, the forces exerted on the articular surfaces on the knee joint were a function of three classes of forces; static, inertial, and muscular. The static and inertial forces exerted on the knee joint were calculated by the differential equations of motion. However, these equations would not take into account how the required knee torques were produced.

Including the electromyographic data revealed the various muscle tensions which must have been exerted to satisfy the torque requirements. When the compressive forces due to muscle tensions were added to the compressive forces due to static and inertial forces, the resultant compressive force's magnitude was several times the body's weight.

Murray, M. Pat.; Drought, A. Bernard; and Kory, Ross C. "Walking Patterns of Normal Men." Journal of Bone And Joint Surgery. Volume 46-A, Number 2 (March, 1964), 335-360.

A photographic method was described for gathering displacement data. Included in the discussion was a description of a pelvic and a thoracic belt. An L-shaped target was attached to each of these belts for the purpose of gathering the angular movement of these parts of the body.

Padgaonkar, A.J.; Krieger, K.W.; and, King, A.I.

"Measurement of Angular Acceleration of a Rigid Body Using Linear Accelerometers." American Society of Mechanical Engineers, November, 1975, Paper 75-APMB-3.

The authors showed that in theory a minimum of six linear accelerometers were required for a complete definition of the three dimensional kinematics of a rigid body. However, this result was contingent upon zero cross-axis sensitivity of the transducers. In reality, there was always some interaction.

The authors then showed that an alternate method based on a special nine-accelerometer configuration was able to compensate for accelerometer cross-axis sensitivity. The stability and superiority of this particular approach were shown theoretically and empirically.

Page, Edward W., and Infante, Joseph. "A New Instrumentation Technique for Gait Analysis." Proceedings of the 26th Annual Conference on Engineering in Medicine and Biology, Minneapolis, Minnesota, 1973, 126.

This paper was an abstract of the authors' previous work entitled "A New Method of Determining the Orientation of Biological Limb Segments."

Page, Edward W., and Infante, Joseph. "A New Method of Determining the Orientation of Biological Limb Segments." Proceedings of the Fifth Annual Southeastern Symposium of System Theory, North Carolina State University, March, 1973, pp. 254-259.

In order to determine the angular orientation of biological limb segments, the authors proposed the use of the magnetometer.

When in a magnetic field, such as the Earth's, the magnetometer measured its angular orientation with respect to the magnetic vector.

In spherical coordinates, two angles were needed to define the direction of a line in space. The magnetometer gave only one such angle. Hence, there was a cone of ambiguity associated with this type of instrumentation.

At the end of this paper, very interesting state variable plots were presented.

Page, R.L. "The Position and Dependence on Weight and Height of the Center of Gravity of the Young Adult Male." Ergonomics, Volume 17, Number 5 (1974), 603-612.

The objective of this study was to explore the statement that the average height of the young

adult male's center of gravity was 55% of his height.

A well written statistical analysis showed the height to be more influential than the weight in determining the position of the center of gravity and the approximate nature of the 55% figure.

Passerello, C.E., and Huston, R. L. "Human Attitude Control." Journal of Biomechanics, Volume 4, Number 2 (March, 1971), 95-102.

Using the conservation of angular momentum, equations of motion were derived for a biped consisting of ten rigid parts free to move in the absence of external forces.

The application of this model to pitch, yaw, and roll movements of astronauts was then discussed.

Paul, John P. "Comparison of EMG Signals from Leg Muscles with the Corresponding Force Actions Calculated from Walkpath Measurements." Proceedings Conference "Human Locomotor Engineering," Institution of Mechanical Engineers, University of Sussex (1972), 1-15.

The use of EMG signals to understand muscle activity was subjected to close observation in this article. It was pointed out that skin surface EMG readings might respond to more than one muscle. Futhermore, both skin surface and hypodermic needle transducers were only capable of measuring local phenomena in the muscle. Since the global muscle action was actually a function of intermitted local action of its components, these locally evaluated EMG readings could not be used directly to understand overall muscle activity.

The author proposed augmenting raw EMG data with muscle length and leg configuration data.

Paul, John P. "Design Aspects of Endo-prostheses for the Lower Limb." Perspectives in Biomedical Engineering. Edited by R.M. Kenedi. 1973, pp. 91-94.

The author discussed the force and moment loading on hip joint and knee joint prostheses. The work was based on calculations by the author and Dr. J.B. Morrison.

Paul, John P. "Forces Transmitted by Joints in the Human Body." Proceedings Conference: Institution of Mechanical Engineers, Volume 181 (1967), Paper 8.

This paper was concerned with the same subject and method of analysis as Paul, John P.

"Load Actions on the Human Femur in Walking and Some Resultant Stresses."

Paul, John P. "Load Actions on the Human Femur in Walking and Some Resultant Stresses." Experimental Mechanics, March, 1972, pp. 1-5.

By direct measurement of ground reaction, forces and limb configuration forces and moments exerted at the leg joints were calculated. From electromyographic data of leg muscle activity and the locations of these muscles, the tension in relevant muscles and ligaments was then calculated.

Pearson, J. Raymond. "Need for Research in Fundamental Biomechanical Studies." Artifical Limbs, Volume 11, Number 2 (Autumn, 1968), 24-27.

This article's description of bio-problems which needed engineering solutions helped guide Winarski's resarch.

Pearson, J. Raymond, et al. "Pressures in Critical Regions of the Below Knee Patellar Tendon Bearing Prosthesis." Bulletin of Prosthetics Research, BPR 10-19 (Spring, 1973), 52-76.

This paper summarized the study of the variation of normal interface pressures under perturbations in prosthesis alignment. The four types of perturbations were 5mm and 10mm anterior-posterior and medial-lateral linear adjustments and 5^0 and 10^0 flexion-extension and abduction-adduction angular adjustments. These adjustments were performed with the V.A.P.C. B/K adjustable shank alignment coupling.

The maximum mass of a subject was 96.9 kilograms. This gave an indication that the chosen alignment coupling would tolerate a rather heavy person.

A Kulite pressure transducer was placed in the region of the patellar tendon, distal anterior tibia, lateral tibial condyle, and medial tibial condyle. Footswitches, as well as knee and hip goniometers, were placed on the subject.

This paper underscored the experimental technique used in Winarski's later research.

Pelzer, Edward; Wright, Donald W; and Mason, Carl. "Human Locomotion." Bulletin of Prosthetics Research, (Fall, 1969), 48-105.

This article was written in a very general, non-technical, easy to read manner. Material included a description of work exerted by the body and the change in the sum of the potential and kinetic energies. A 24 gage instrumented below

knee pylon built by the Veteran's Administration Prosthetics Center was also mentioned.

Pertuzon, E., and Bouisset, S. "Instantaneous Force-Velocity Relationship in Human Muscle." Medicine and Sport: Biomechanics III. Volume 8 (1973), 230-234.

Using a goniometer to measure angular displacement and an accelerometer to measure angular acceleration, a hyperbolic relationship was established between instantaneous velocity of muscle shortening and force.

Plagenhoef, Stanley. "Computer Programs for Obtaining Kinetic Data on Human Movement." Journal of Biomechanics. Volume 1, Number 3 (August, 1968), 221-234.

Two computer programs for the purpose of kinematic analysis of the human body were presented and discussed.

The first program determined angular velocities and angular accelerations from observed angular displacement data. The output from this first program became the input for the second program.

The second program did the actual kinematic analysis; calculating the joint forces and torques, position of the body's center of gravity, and comparative contribution of each body segment to the total movement.

Radcliffe, Charles W. "Biomechanical Design of a Lower-extremity Prosthesis." American Society of Mechanical Engineers. November, 1960, Paper 60-WA-305.

After a thorough discussion of the factors involved in the human gait cycle, the design of an above knee prosthesis was proposed.

The prosthesis consisted of a four bar linkage knee which was hydraulically controlled.

Radcliffe, Charles W. "Functional Considerations in the Fitting of Above Knee Prostheses." Artificial Limbs. Volume 2, Number 1 (January, 1955), 35-60.

A comprehensive summary of the anatomy of the amputee's stump as well as the construction, fitting, and wearing of the above knee prosthesis.

Radcliffe, Charles W. "The Biomechanics of Below Knee Prostheses in Normal, Level, Bipedal Walking." Artificial Limbs. Volume 6, Number 2 (June , 1962), 16-24.

In an attempt to understand the transmission

of forces across the stump-shell interface, this author studied forces exerted on the shell of the prosthesis socket.

Rakic, M. "A Possibility of Controlling the Human Lower-Extremity Prosthesis." International Symposium on External Control of Human Extemities. Belgrade, Yugoslav Committee for Electronics and Automation., 1967, pp. 212-217.

The analysis proposed a system for controlling the human lower extremities prostheses. The principle of this system was in the use of ten servosystems which would transfer the movements of arms and convert them into the movements of the leg prostheses.

Ralston, H.J. "Effects of Immobilization of Various Body Segments on the Energy Cost of Human Locomotion." Proceedings of Second International Congress on Ergonomics." Dortmund (1964), 53-60.

The purpose of this study was to obtain the average percent increase in gait energy expenditure due to simulated fusion of specific joints.

On normal subjects, either the ankle, knee, hip, torso, or arms were immobilized. Pulmonary ventilation was measured with Max-Planck-type respirometers, and oxygen analyses performed with Beckman-Pauling analysers. Energy expenditure was calculated by method of Weir (1949). The results of this well written article were appropriately summarized in it.

Ralston, H.J. "Energy-Speed Relation and Optimum Speed during Level Walking." Internationale Zeitschrift fuer angewandte Physiologie einschliesslich Arbeitsphysiologie. Volume 17 (1958), 277-283.

Twelve male and seven female normal, untrained adults were studied. The subjects walked around a rectangular track. As usual, this author used a Max-Planck-type respirometer, a Beckman-Pauling oxygen analyzer, and the method of Weir (1949) to derive energy expenditure.

His results were summarized under three headings. First, the level walking power expenditure normalized with respect to subject mass was a linear function of the square of the average speed of forward motion. Second, if this expression were integrated with respect to time and twice differentiated with respect to velocity, a minimum energy expenditure appeared at 73.8 metres/minute (1.230 m/s). Finally, people naturally tended to walk at this optimum velocity.

It should be noted that the second

derivative mentioned above was not very large at all. This meant that the minimum was not so well defined that everybody had to walk at exactly 73.8 metres/minute. Another observation was that the linear approximation to the power function appeared to be inappropriate at high velocities. Furthermore, the slope of this line was best described as a statistical distribution rather than a simple number. Hence, 73.8 metres/minute could not be accepted as the gospel truth for every last human being.

Ralston, H.J., and Lukin, L. "Energy Levels of Human Body Segments During Level Walking." Ergonomics, Volume 12, Number 1 (1969), 39-46.

The subjects walked on a power treadmill at three speeds; 48.8, 73.2, and 97.6 meters per minute. Microswitches built into the shoes of each subject denoted heel contact. The experimental method consisted of attaching cords to the HAT, head-arms-trunk, thighs, shanks, and feet. Potentiometric transducers converted the cord motion into electric signals.

Using Dempster's data and their own measurements, the authors calculated theoretical mechanical energy expenditure versus time. Simultaneous metabolic measurements, using the method of Weir (1949), showed the average efficiency of walking to be about 23%.

This article was certainly well written and highly informative.

Reynolds, Herbert M., et al. "Mass Distribution Properties of the Male Cadaver." Society of Automotive Engineers, February, 1975, Paper 750424.

The authors conducted and reported a study of the inertia tensor properties of the human body and its segments. Six adult male cadavers were involved.

No attempt was made to reduce the data into a set of parameters for population estimates. Instead, the results were simply listed. Still, this paper was important for its addressing the problem of determining the three dimensional inertia tensor at a suitable orientation for the human body and its segments.

An important comparison between the authors' experimental method and that of previous researchers was presented. The paper's bibliography was also impressive and encompassing.

Rodenberger, C.A., "Forces Produced by Ground Contact Rotational Orthotic Devices." American Society of

Mechanical Engineers, November, 1973, Paper 73-WA/BIO-2.

Inability to rotate at the ankle could cause large, possibly painful torques to be exerted on the tibia or stump. The author suggested two devices to allow the necessary deformation, thereby reducing these "tibia" torques. One was a circular rib rotational device; the other, a parallel rib rotational device.

Rubenstein, R.L. "Development of Anthropomorphic Child Dummies." Society of Automotive Engineers, October, 1974, Paper 740937

This article described the actual construction of a pair of crash-test dummies in the 50th percentile three-year-old and six-year-old sizes. The fundamental basis for this design was a compilation of recent child anthropometry surveys.

Sard, Arthur, and Weintraub, Sol. A Book of Splines. New York: John Wiley & Sons, Inc., 1971.

Sato, Satoru. The Application of Photographic Techniques to the Kinematic Analysis of the Human Gait Cycle. Unpublished. The University of Michigan, Ann Arbor, Michigan, 1975.

This project encompassed selection and implementation of a means of recording limb movements during ambulation.

Saunders, H.B. Dec. M.; Inman, V.T.; and Eberhart, H.D. "The Major Determinants in Normal and Pathological Gait." The Journal of Bone and Joint Surgery, Volume 35-A, Number 3 (July, 1953), 543-558.

The basis of this analysis was on the pathway of a particular point of the body, the normal position center of gravity.

During the gait cycle, being a function of the distribution of the subject's mass, the actual center of gravity was constantly changing its location with respect to the subject's body. The actual center of gravity might even have been a point external to the subject's body. Hence, the actual center of gravity and the fixed location, normal position center of gravity were entirely different.

In the derivation of the differential equations of motion of a biped, a vector from an inertial reference frame must be drawn to at least one point on the biped. This meant that the normal position center of gravity, as discussed in this paper, could serve as a useful and vital point of interest.

The authors included an excellent discussion of potential energy, kinetic energy, and work transfer during the human gait cycle.

Schwartz, R. Plato, and Heath, Arthur L. "The Definition of Human Locomotion on the Basis of Measurement." The Journal of Bone and Joint Surgery, Volume 29, Number 1 (January, 1947), 203-214.

Six pressure sensitive discs were applied to the plantar surface of each subject's foot at the great toe, the first, third, and fifth metatarsal, the medial heel, and the lateral heel.

These pressure transducers consisted of carbon discs sandwiched between phosphor bronze electrodes. The output from these transducers was recorded on an optical oscillograph.

This paper concluded that normality in human locomotion could be defined by the pressure graphs plotted as function of time.

Schwartz R. Plato; Heath, Arthur L.; and Wright, John N. "Electrobasographic Method of Recording Gait." Archives of Surgery, Volume 27 , Number 5 (November, 1933), 926-934.

Gait records of normal and pathological subjects were documented by foot switch analysis.

There were three positive polarity contact discs on each foot. These discs were made of brass and were three forths of an inch in diameter.

When the subject walked on the negative polarity aluminum base plate, the duration of the time of weight bearing on the heel, the fifth metatarsal, and the great toe was determined.

Schwartz, R. Plato, And Vaeth, William. "A Method for Making Graphic Records of Normal and Pathological Gaits." The Journal of the American Medical Association, Volume 90, Number 2 (January, 1928), 86-89.

The basograph was used, a mechanism which recorded pelvic motions. The instrumentation involved consisted of a moving strip of paper and a pendulum.

Scranton, P.E., and McMaster, J.H. "Momentary Distribution of Forces Under the Foot." Journal of Biomechanics, Volume 9, Number 1 (1976), 45-48.

The difference in distribution of forces between walking and running was compared in ten normal subjects and several patients with either pes cavus, pes planus, or a tibio-fibular synostosis. The transducer used was a flexible plastic sheet enclosing cholesteric, or shear

sensitive, liquid crystals. The pressure induced color changes were documented on a Bolex camera.

The transducer was not calibrated and no attempt was made to translate color wave lengths into pressure readings. Furthermore, the authors mentioned that the device did respond to the local pressures and to the global deformation of the plastic sheet.

The information depicted photographically was still most informative.

Sears, Francis Weston. Optics. Addison-Wesley, Massachusetts, 1958.

Material included in this book covered thin lenses, thick lenses, and compound lens optics. The Lensmaker's Equation as well as the Gaussian form of the lens equation were presented.

Selreg, A., and Arvikar, R.J. "A Mathematical Model for Evaluation of Forces in Lower Extemities of the Musculo-skeletal System." Journal of Biomechanics, Volume 6, Number 3 (May, 1973) 313-326.

This paper described the development of a linear mathematical model capable of evaluating lower extremity muscle forces and joint reactions for different static postures.

The analytical results were compared to electromyographic signals for three stance positions; standing, leaning, and stooping.

Selreg, A. And Arvikar, R.J. "The Prediction of Muscular Load Sharing and Joint Forces in the Lower Extremities During Walking." Journal of Biomechanics, Volume 8, Number 2 (March, 1975), 89-102.

The human body was modeled as the HAT and seven rigid segments which comprised the lower half of the body. The body was kept in dynamic equilibrium by 62 muscle forces. The selection of which muscle acted was accomplished by a linear programming technique which utilllized the criterion of minimumization of an object function. The object function was the weighted sum of all muscle tensions plus 4 times the sum of all joint moments.

For this paper, the body was assumed to be in a state of quasistatic equilibrium during each observed point in time. In other words, all dynamic forces were ignored. 4x4 transformation matrices were used to describe the motion of the seven segments.

The solutions for muscle activity compared favorably with typical EMG patterns. This solution

also permitted the calculation of joint forces as the sum of gravitational and muscular induced forces.

The final topic was the results obtained when the object function was modified to include terms requiring the minimization of a joint reaction. Results showed that a joint reaction could not be reduced significantly without the subject altering his gait.

Selby, Samuel M., ed. Standard Mathematical Tables. Cleveland: The Chemical Rubber Company, 1972.

Shames, Irving H. Dynamics. Volume II of Engineering Mechanics. Englewood Cliffs: Prentice-Hall, Inc., 1966.

Sharkey, B.J.; McDonald, J.F.; and Corbridge, Lynn G. "Pulse Rate and Pulmonary Ventilation as Predictors of Human Energy Cost." Ergonomics, Volume 9, Number 3 (May, 1966), 223-227.

This study compared the usefulness of pulse rate and ventilation rate as indicators of oxygen consumption.

Four young men were exercised on a motor driven treadmill as six different inclines. During the last two minutes of each test, expired air was collected in a Douglas bag. A sample was collected in a Bailey bottle for duplicate analysis in a Scholander apparatus. Pulse and ventilation rates were also documented.

Comparisons of the pulse and ventilation predicted oxygen consumption rates to the actual consumption rate were made. Errors up to 39% for the pulse rate predictions and 26% for the wentilation rate predictions were apparent. This prompted the authors to state that more research in this area was required.

Shoup, T.E. "Optical Measurement of the Center of Rotation for Human Joints." Journal of Biomechanics, Volume 9, Number 4 (1976), 241-242.

By use of photo-optics, the author studied the instantaneous center of rotation of the human shoulder complex. This measurement was a manifestation of the motion of the humerus, scapula, and clavicle bones as articulated at the acromioclavicular, glenohumeral, and sternoclavicular joints.

The actual experimental technique involved double exposure images in order to obtain Moire fringes. By use of two targets, the instantaneous center of rotation of the shoulder complex was

determined.

Skorecki, J. "The Design and Construction of a New Apparatus for Measuring the Vertical Forces Exerted in Walking: A Gait Machine." Journal of Strain Analysis, Volume 1, Number 5 (1966), 429-438.

This scientist described the design and construction of an eleven foot long vertical force measuring platform. His device made use of the elastic displacement of steel diaphragms, viscous damping of air free oil, and the deflection of beams of ultra-violet light by mirrors actuated by the diaphragms.

Permanent records were obtained on photosensitive paper. The static and dynamic accuracy of this device was within 1%. Independent vertical force measurements were made for both the right and left leg. A wall separated these two transducers and protruded above floor level.

It appeared the subject might alter his gait somewhat due to the presence of the wall.

Smidt, Gary L. "Biomechanical Analysis of Knee Flexion and Extension." Journal of Biomechanics, Volume 6, Number 1 (January, 1973), 79-92.

The data was gathered in two phases; radiographs of knee flexion, in fifteen degree increments, and measurements of the torque generated by knee flexors and extensors. From this data, the path of the knee's instant center of rotation was graphed and the required tension in the flexor and extensor muscles against known external loads was determined.

Smith, Karl W.; McDermid, Charles D.; and Shideman, Frederick E. "Analysis of the Temporal Components of Motion in Human Gait." American Journal of Physical Medicine, Volume 39, Number 4 (August, 1960), 142-151.

These authors recorded only the duration of time each foot was in contact with the ground or in the swing phase.

Part of the data gathering consisted of passing 100 microamperes of current through the subject's body.

Smith , P.G., and Kane, T.R. "On the Dynamics of the Human Body in Free Fall." Journal of Applied Mechanics, Volume 35 (March, 1968), 167-168.

The coordinates used to specify the body's motion were presented in two classifications; external and internal.

External coordinates were used to describe

the location of a point on the body with respect to an inertial reference frame. Location of the parts of the body was then described using internal coordinates with respect to the point on the body described by the external coordinates.

Stidham, Phillip H., and Colortran, Berkey. "How to Layout Lighting Systems for Test Labs." Proceedings of the Society of Photo-optical Instrumentation Engineers. Edited by Brown and Patrick. Volume 34 (November, 1972), 125-128.

The authors discussed choosing film speed, ASA number, exposure time, and footcandles of illumination intensity.

Stribley, Richard Frederick. The Quantification of Balance in the Stance of Asymptomatic Normals and Pathological Patients. Ann Arbor: The University of Michigan Medical School, 1974.

System /370: Scientific Subroutine Package. New York: International Business Machines, 1970.

Thornton-Trump, A.B., and Daher, R. "The Prediction of Reaction Forces from Gait Data." Journal of Biomechanics. Volume 8, Number 3/4 (July, 1975), 173-178.

A seven segment model in three space was used to predict hip and knee moments as well as floor reactions. All seven segments were rigid and possessed mass and mass moment of inertia. Actual displacement data were not gathered. Instead, the locomotion studies of Bressler, et al. (1950) and Murray, et al. (1964) were used. Anthropometric data came from Dempster (1955). Since motion was model input, joint torques and forces were model output.

Perturbations in this constructed gait were done via phase shifts. The perturbations showed definite effects on the derived hip and knee moments and on the floor reactions. The effects these perturbations had on the knee joint were used to assist the design of an external prosthetic polycentric knee joint.

The authors felt their studies were more valuable than force plate studies, since subconsious aiming at the force plate by the subject might give misleading results.

It should be noted that the authors did state the value of treating the entire body as a system, requiring a complete set of displacement data and anthropometric measurements. This was somewhat inconsistent with their use of the HAT

representation of the upper half of the body and the use of averaged out data from previous experimenters.

Timoshenko, S., and Young, D.H. Advanced Dynamics. New York: McGraw-Hill Book Company, Inc., 1948.

Tomovic, R. "Multilevel Control of Mechanical Multivariable Systems as Applied to Prosthetics." I.E.E.E. Transactions on Automatic Control (Short Papers). Volume AC-13 (February, 1968), 72-74.

The main purpose of the author's efforts was to present the use of algorithmic methods for the optimum control of human extremities.

In state space analysis, optimal meant the selection of a definite trajectory on the basis of a performance criterion and a set of available controls. Instead of selecting a trajectory, the author proposed the selection of an algorithm out of the set of allowable algorithms on the basis of four finite joint states, the duration of these states, and a minimum energy consumption criterion.

The four finite joint states were: State (Y), consisting of inactive joint tendons; States (A, B), one sided activity only; and State (Z), locked joint due to fully active, equal and opposite joint tendon forces.

The duration of the joint states was specified by a 4xn matrix; the row index gave the finite joint state; the column index, the joint number.

The author gave an example demonstrating why it required less energy to walk with knee flexion than to walk with a stiff, fully extended knee.

Townsend, M.A. "A Practical Stability Analysis of a Class of Bipedal Locomotion Prostheses." American Society of Mechanical Engineers. November, 1975, Paper 75-WA/BIO-13.

By observing the actual operation of prosthetic walking devices, the author identified a class of motion. This class was defined by dominant motions in the parasagittal plane, negligible torso rotation and swing leg lift, and pivot turning.

His force analysis lead to a stability graph relating forward velocity, in meters per second, and turning angular velocity, in radians per second.

Townsend, M.A., and Lepofsky, R.J. "A Powered Bipedal Walking Prosthesis," American Society of Mechanical Engineers. November, 1975, Paper 75-WA/BIO-14.

The authors discussed the design and application of a walking machine. Its purposes were to allow a crippled person to walk in an upright, stable fashion for therapy or for locomotion in relatively restricted environments.

Townsend, M.A. "A Practical Stability Analysis of a Class of Bipedal Locomotion Prostheses," discussed the stability of this mechanism.

Townsend, M.A., and Sereg, A. "The Synthesis of Bipedal Locomotion." Journal of Biomechanics, Volume 5, Number 1 (January, 1972), 71-83.

The model proposed consisted of a single, rigid part possessing both mass and mass moment of inertia supported by two extendible legs.

The double stance phase was completely ignored in the analysis, as well as the vast differences between an extensible leg and a leg with knee and ankle joints.

Townsend, M.A., and Tsai, T.C. "Biomechanics and Modelling of Bipedal Climbing and Descending." Journal of Biomechanics, Volume 9, Number 4 (1976), 227-240.

By a trial and error method, the climbing and descending gaits for a sagittal plane model were synthesized. The model itself had 7 rigid segments connected by 5 planar-hinged joints. The legs and trunk were treated as possessing distributed mass according to the anthropometry of Hanavan (1968).

Townsend, P.R., et al. "The Distribution and Anisotropy of the Stiffness of Cancellous Bone in the Human Patella." Journal of Biomechanics, Volume 8, Number 6 (1975), 363-367.

The authors derived a simple sheet and strut model of cancellous bone tissue. Analytical modulus values were favorably compared to empirically derived values.

The experimental values were obtained from cubes cut from seven normal cadaver knee caps. These cubes were then place in a specially constructed compression tester.

Traugh, George H.; Corcoran, Paul J.; and Reyes, Rudolpho, L. "Energy Expenditure of Ambulation in Patients with Above-Knee Amputations." Archives of Physical Medicine, Volume 56, Number 2 (February, 1975), 67-71.

"Oxygen consumption during ambulation was measured in nine middle-aged or elderly subjects having above-knee amputations, in an effort to

determine the mode of ambulation and prosthetic prescription requiring the least energy expenditure. Studies revealed that crutch walking and prosthetic ambulation require the same energy expenditure; that there is no significant difference between the energy requirements of ambulating with knee locked or unlocked; and that 65% more energy is required at approximately one-half the normal speed of ambulation for above-knee amputees as compared to normal persons."

Normal subject power expenditure was evaluated from the formula:

$$\text{Power} = 10.03*\text{Mass} + 0.03V^2$$

The units of power were kcal/minute; mass, kg; and velocity, miles/hour.

A more useful way of writing this formula was:

$$\text{Power} = 167.1666*\text{Mass} + 2.5019*V^2$$

The units of power are now cal/second; mass, kg; and velocity, metres/second.

This formula did contrast to Ralston's (1958) which did not have an explicit mass term.

This formula did portray an optimum velocity - minimum energy point as did Ralston's. Integration with respect to time to yield energy gave an expression which was the sum of a hyperbola and a straight line. Twice differentiation of this energy curve with respect to velocity showed there was a point where the first derivative vanished and the second derivative was positive. This defined an optimum velocity where energy expenditure was minimized. Furthermore, this velocity was a function of the mass of the subject.

Unfortunately, this optimum velocity was an unusually large number and bore no physical significance.

Tse, Francis S.; Morse, Ivan E.; and Hinkle, Rolland T. Mechanical Vibrations. Boston: Allyn and Bacon, Inc., 1963.

Vukobratovic, M., et al. "Analysis of Energy Demand Distribution within Anthropomorphic Systems. Transactions of the American Society of Mechanical Engineers. December, 1973, Paper 74-AUT-D.

This paper described the development of the equations of motion for a biped consisting of nine rigid parts, including ground, free to move in Euclidean three space.

The origin of the inertial reference frame was attached to one of the biped's feet at a point called the zero moment point. It appeared the zero moment point was actually the center of pressure of the vertical force distribution exerted on the foot. These equations were then applied to the movements of a robot. Power expenditure was calculated as the integral of joint moment with respect to limb angular velocity. The conclusion of this article was in its discussion of required power expenditure as a function of various prescribed robot gaits.

Vukobratovic, M.; Frank, A.A.; and Juricic, D. "On the Stability of Biped Locomotion." I.E.E.E. Transactions on Bio-medical Engineering. Volume BME-17, Number 1 (January, 1970), 25-36.

Three categories of stability were introduced in the paper: body stability, the stability of the orientation of the body with respect to the inertial reference frame; body path stability, the stability of the body's forward velocity and gait stability, the stability of the manner in which the legs were moved.

The analysis was conducted by observing a simplified mechanical model of a biped.

Many assumptions were made, including the existance of a compensating point mass.

Vukobratovic, M., and Juricic, D. "Contribution to the Synthesis of Biped Gait." I.E.E.E. Transactions on Bio-medical Engineering. Volume BME-16, Number 1 (January, 1969), 1-6.

The gait of a biped was synthesized by the combination of a kinematic and a dynamic analysis.

The lower extremities of the biped were controlled kinematically. In order to insure the stability of the biped, a compensating point mass on an inverted, three dimensional pendulum attached to the pelvis was controlled dynamically.

The objective of this compensating mass was to introduce into the system the torques prescribed by the kinematic analysis. Stability, the sum of the kinematic moments equalling the sum of the dynamic moments, was thus achieved. The actual position of the compensating mass while it was supplying the required dynamic torques bore no anthropometric significance.

Vukobratovic, M. ; Juricic, D.; and Frank, A.A. "On the Control and Stability of One Class of Biped Locomotion Systems." Journal of Basic Engineering. Transactions of the American Society of Mechanical

Engineers, Volume 92, Series D (June, 1970), 328-332.

Two types of controls were introduced, the force controller and the position controller. Then, the concept of the algorithmic system and the condition of repeatibility was discussed.

Vukobratovic, M.; Maric, M.; and Gavrilovic, M. "On Stability of Locomotion Automata." International Symposium on External Control of Human Extremities. Belgrade, Yugoslav Committee for Electronics and Automation, 1967, pp. 218-220.

The system analyzed consisted of a planar inverted pendulum with two degrees of freedom. The coordinates used consisted of the angle of swing of the planar pendulum and the linear, horizontal translation of its pivot point. The two differential equations of motion were then simplified using small angle approximations. Proportional plus derivative control was used to stabilize the inverted pendulum.

This method of control proved satisfactory for restoring the inverted pendulum to its original vertical position despite the small angular perturbations which constituted the initial conditions of the problem.

Waters, Robert L.; Morris, James; and Perry, Jacquelin. "Translational Motion of the Head and Trunk During Normal Walking." Journal of Biomechanics, Volume 6, Number 2 (March, 1973), 167-172.

Displacement and acceleration data of the head and the T-10 and S-2 of the trunk were recorded in two separate experiments. This displacement data was gathered in the vertical, anterior-posterior, and medial-lateral directions; acceleration data, vertical and anterior-posterior directions only.

Cords affixed to the body and displacement transducers were used to measure the respective displacements; accelerometers, accelerations.

Weir, J.B. De V. "New Method for Calculating Metabolic Rate with Special Reference to Protein Metabolism." Journal of Physiology, Volume 109, Numbers 1 and 2 (August, 1949), 1-9.

Equations were derived from first principles for the calorie value of a litre of oxygen metabolizing a mixture of carbohydrate, protein, and fat.

This paper was referred to by many subsequent researchers in their study of gait energy expenditure.

Wilson, A. Bennett. "Limb Prosthetics - 1967." Artifical Limbs. Volume 2, Number 1 (Spring, 1967), 1-46.
This article was both an extremely interesting and informative description of amputations and prostheses for the upper and lower extremities. Much United States Manufacturing Company equipment was displayed and discussed.

Winarski, Daniel James. A Computer Simulation of the Human Gait Cycle for Medical Applications. Ann Arbor: The University of Michigan Medical School, 1974.

Winter, David A., et al. "Kinematics of Normal Locomotion - a Statistical Study Based on T.V. Data." Journal of Biomechanics. Volume 7, Number 6 (November, 1974), 479-486.
The television system gathered the displacement data of hemispheres targets affixed to a subject. The television record was digitized in real time via a T.V. - computer interface. The computer then calculated the subject's limbs displacement, velocity, and acceleration in absolute coordinates.

Winter, David A.; Greenlaw, Robert K.; and Hobson, Douglas A. "A Microswitch Shoe for Use in Locomotion Studies." Journal of Biomechanics. Volume 5, Number 5 (September, 1972), 553-554.
This technical note described the instrumentation of each shoe of a subject with five microswitches. One digital to analog converter was assigned to each set of five microswitches. In this manner, only one data collecting channel per shoe was used to delineate the various portions of the ground contact phases of each foot.

Winter, David A.; Sidwall, H. Grant; and Hobson, Douglas A. "Measurement and Reduction of Noise in Kinematics of Locomotion." Journal of Biomechanics. Volume 7, Number 2 (March, 1974), 157-159.
The authors proposed the electronic filtering of displacement data before differentiating it. The data was passed though a low pass filter twice, once in real time and once in reverse time.

Winter, David A.; Quanbury, Arthur O.; and Reimer, Gary D. "Analysis of Instantaneous Energy of Normal Gait." Journal of Biomechanics. Volume 9, Number 4 (1976), 253-258.
The authors studied human gait with a TV-

computer system documented by Winter (1974). From this motion analysis, it was possible to study the potential, kinetic, and the sum of the potential and kinetic energies for discrete body segments. No relationships between work and the change in the sum of the potential and kinetic energy were drawn.

This article was certainly well written and informative.

Wisner, James. Mechanical Engineering 490 Final Report. Unpublished. The University of Michigan, Ann Arbor, Michigan, 1975.

This project encompassed the design, construction, and calibration of the force and moment transducer used by Winarski.

Wyndham, C.H. "Inter- and Intra-Individual Differences in Energy Expenditure and Mechanical Efficiencies." Ergonomics, Volume 9, Number 1 (January, 1966), 17-29.

"Differences in oxygen consumption between individuals and also within individuals on 4 different tasks and at 2 rates of energy expenditure were examined.

Yamashita, T., and Katoh, R. "Moving Pattern of Point of Application of Vertical Resultant Force During Level Walking," Journal of Biomechanics, Volume 9, Number 2 (1976), 93-99.

Using a force plate sensitive only to normal loads and their point of application, the gait of four normal men was studied.