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Effects of Semi-Rigid Orthotics on  
Abduction-Adduction Moments at the Knee  
Joint During Walking

presented by

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ABSTRACT

EFFECTS OF SEMI-RIGID ORTHOTICS ON  
ABDUCTION-ADDUCTION MOMENTS AT THE KNEE DURING WALKING

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ABDUCTION-ADDUCTION MOMENTS AT THE KNEE  
JOINT DURING WALKING

By

Renee A. Peltier

THESIS

Submitted to

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and Human Performance

1990



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The results were highly reproducible among subjects.

Analysis of the results revealed no significant differences in

# ABSTRACT

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By

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**Renee A. Peltier**

study allowed for a comparative analysis of the effects of

orthotics on knee joint mechanics.

The purpose of this study was to examine the effects of semi-rigid orthotics on the abduction-adduction moment acting at the knee joint during the stance phase of walking. A three-dimensional gait analysis of five subjects walking under two conditions, shoe alone and shoe plus orthotic was performed. The orthotics used were prescribed to control excessive pronation. Kinematic data were collected using two LOCAM high speed cameras. An AMTI force platform was used to simultaneously collect the ground reaction forces and moments about the platform center. The approach taken in this study utilized the concepts of rigid body dynamics and the Newton-Euler equations of motion. The abduction-adduction moment acting at the right knee was calculated from the three-dimensional kinematic and kinetic data. The abduction-adduction moments at the knee were analyzed for general

patterns and trends.

The results were highly reproducible among subjects. Analysis of the results revealed noticeable differences in the abduction-adduction moment between the two conditions. Comparison of the magnitudes shown by the abduction-adduction moment between the two test conditions revealed that the adduction moment was decreased in the condition with the orthotic. The methods used in this study allowed an objective analysis of the effects of orthotics on knee joint mechanics.

Conclusions

To my Mother  
and to Testing

**Dedication**

**To my Mother, Corbitt and Maurel  
Ann M. Peltier**

# TABLE OF CONTENTS

## ACKNOWLEDGMENTS

LIST OF TABLES.....13

LIST OF FIGURES.....14

INTRODUCTION

I wish to express my appreciation to Dr. Dianne Ulibarri for her guidance and advice. I wish to thank Dr. Mary Verstraete for her assistance with the data post-processing procedures. Appreciation should also be extended to Brook Shoes, Inc., whom supported this study in part by a grant and to Ken Scholton and Mhairi Blacklock of Gary Nederveld & Associates, whom provided flexibility in my work schedule enabling me to complete this study.

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## TABLE OF CONTENTS

|                                                 |     |
|-------------------------------------------------|-----|
| LIST OF TABLES.....                             | 11  |
| LIST OF FIGURES.....                            | 111 |
| INTRODUCTION.....                               | 1   |
| Table I. NEED FOR THE STUDY.....                | 5   |
| II. PURPOSE OF THE STUDY.....                   | 7   |
| Table III. RESEARCH PLAN.....                   | 8   |
| Table IV. ASSUMPTIONS AND LIMITATIONS.....      | 8   |
| V. SIGNIFICANCE OF THE STUDY.....               | 9   |
| VI. DEFINITIONS.....                            | 10  |
| REVIEW OF LITERATURE.....                       | 11  |
| I. GAIT ANALYSIS.....                           | 11  |
| A. KINEMATIC STUDIES.....                       | 12  |
| B. FOOT-GROUND REACTION FORCE AND PRESSURE..... | 14  |
| C. JOINT AND MUSCLE FORCE ANALYSIS.....         | 17  |
| II. ORTHOTIC THERAPY.....                       | 18  |
| METHODS.....                                    | 24  |
| I. DATA COLLECTION PROCEDURES.....              | 25  |
| II. ANALYTICAL METHODS.....                     | 27  |
| RESULTS AND DISCUSSION.....                     | 34  |
| I. CONDITION I: WALKING WITHOUT ORTHOTIC.....   | 35  |
| II. CONDITION II: WALKING WITH ORTHOTIC.....    | 37  |
| III. DISCUSSION AND COMPARATIVE ANALYSIS.....   | 40  |
| CONCLUSIONS.....                                | 47  |
| APPENDIX.....                                   | 52  |
| LIST OF REFERENCES.....                         | 53  |



# LIST OF TABLES

|           |                                             |    |
|-----------|---------------------------------------------|----|
| Figure 1. | Force Plate Coordinate Axis and the         |    |
| Table 1.  | Mean Values of the Abduction-Adduction      | 35 |
|           | Moment Without Orthotic.....                | 35 |
| Figure 2. | Schematic Representations of the Shank..... | 36 |
| Table 2.  | Mean Values of the Abduction-Adduction      |    |
| Figure 3. | Moment With Orthotic.....                   | 40 |
|           | After Walking Without an Orthotic.....      | 40 |
| Table 3.  | Differences in Mean Peak Values of the      |    |
| Figure 4. | Abduction-Adduction Moment Across at the    |    |
|           | Conditions.....                             | 44 |
| Figure 5. | Subject 1 Abduction-Adduction Moment        |    |
|           | Across Conditions.....                      | 42 |
| Figure 6. | Subject 2 Abduction-Adduction Moment        |    |
|           | Across Conditions.....                      | 43 |
| Figure 7. | Subject 4 Abduction-Adduction Moment        |    |
|           | Across Conditions.....                      | 43 |

## LIST OF FIGURES

### INTRODUCTION

|           |                                                                                         |    |
|-----------|-----------------------------------------------------------------------------------------|----|
| Figure 1. | Force Plate Coordinate Axis and the<br>There Ground Reactions.....                      | 29 |
| Figure 2. | Schematic Representations of the Shank.....                                             | 30 |
| Figure 3. | Abduction-Adduction Moment acting at the<br>Knee Walking Without an Orthotic.....       | 36 |
| Figure 4. | Abduction-Adduction Moment acting at the<br>injuries Knee Walking With an Orthotic..... | 39 |
| Figure 5. | Subject 1 Abduction-Adduction Moment<br>Across Conditions.....                          | 42 |
| Figure 6. | Subject 2 Abduction-Adduction Moment<br>anatomical Across Conditions.....               | 43 |
| Figure 7. | Subject 4 Abduction-Adduction Moment<br>Across Conditions.....                          | 45 |

runners, it seems that many of these problems are related either directly or indirectly to foot structure and function during the support phase of the gait cycle. Osternig, Mason, & Jansen, 1998. The most common unbalanced foot structure is the flat footed foot, which results in an abnormal pattern of excessive pronation during some component of the gait cycle of gait. There are many disorders associated with excessive pronation including heel spurs, plantar fasciitis, shin splints, chondromalacia patellae, iliotibial band syndrome and low back pain (McPoil & Brackley, 1998). These disorders may be compounded with the additional stresses of sport activities.

As overuse injury itself can result as a reaction of the tissue to stress from the excessive motion. According to a survey conducted by Eggold (1981), the most common overuse injuries were:

#### INTRODUCTION

There are approximately 30 to 40 million people jogging or running in the United States today (Krakauer, 1987). Despite the great advances in the manufacture and design of running shoes, the average runner will have one or two injuries per year, due mostly to an unbalanced foot structure, overuse, or a combination of these two conditions (Subotnick 1980). Although a number of anatomical factors must be considered in the diagnosis and evaluation of lower extremity problems encountered by runners, it seems that many of these problems are related, either directly or indirectly, to foot structure and function during the support phase of the activity (Bates, Osternig, Mason, & James, 1979). The most common unbalanced foot structure is the flat (pronated) foot, which results in an abnormal amount of subtalar joint pronation during some component of the stance phase of gait. There are many disorders associated with abnormal pronation including heel spurs, planter fasciitis, shin splints, chondromalacia patellae, excessive leg fatigue and low back pain (McPoil & Brocato, 1985). All these disorders may be compounded with overuse and the additional stresses of sport activities.

An overuse injury itself can result as a reaction of the tissue to stress from the excessive motion. According to a survey conducted by Eggold (1981), the most common overuse injuries resulting from abnormal function of the foot were chondromalacia patellae, shin splints, plantar fasciitis, and Achilles tendinitis. He discussed the use of orthotics in treating the abnormal structure of the foot responsible for these symptoms. Eggold concluded that orthotic control can have significant therapeutic value in treating these conditions. The use of orthotics has become popular as a means of preventing and treating the stress related injuries found in long distance running. However, this is not the case. In fact, there are Knee pain was the major complaint among the runners studied by several investigators (Newell & Bramwell, 1984; Pretorius, Noakes, Irving, & Allerton, 1986; Bates & Osternig, 1978; Clancy, 1980; Buchbinder, Napora, & Biggs, 1979; Blake & Denton, 1985), and many of these runners were able to return to their running activities pain-free with foot orthotics. Thus, it would appear that orthotics may effectively alter the mechanics of the knee joint during gait. However, the majority of the work in the area of orthotic therapy has focused on the subtalar joint, not the knee joint. A study conducted by Burkett, Kohrt, and Buchbinder (1985), investigated the effects of orthotics on frontal plane knee kinematics during the support phase of running. Results of this study were



inconclusive.

A biomechanical or functional orthotic device is frequently prescribed by clinicians in treating lower extremity problems. Newell and Bramwell (1984) reported that in 11 months, 50 percent of their athletic patient population had knee pain complaints. Orthoses were prescribed for 82 percent of these patients. The cost of a pair of custom made orthotic appliances is between \$200 and \$350, not including the physicians' office visit fees (Krakauer, 1987). When one considers the frequency and cost of orthotics one would expect to find a large number of scientific studies supporting the effectiveness of orthotics. However, this is not the case. In fact, there are many in the medical community who remain skeptical and some who suggest that orthotics may not only be ineffective, but possibly harmful (Krakauer, 1987). There are many different foot orthotics on the market today claiming to relieve pain and enhance foot function. Unfortunately, minimal research has been conducted investigating the effectiveness of orthotics in the adult population. The effects of the orthotic device remain poorly understood.

The role of the biomechanical orthotic is to control excessive and potentially harmful subtalar and midtarsal joint movement during the stance phase of gait (D'Ambrosia & Drez, 1989). Therefore, research in the area of orthotic therapy has focused on the effects of orthotics



in controlling subtalar joint motion. To date, research performed on the effects of orthotics has been primarily kinematic investigations of the foot in which inconsistent findings have been reported.

More important is that during the stance phase of gait the lower extremity functions in a closed kinetic chain, i.e. all of the forces and moments are transmitted through a fixed end segment (the foot) cephally through the lower extremity. Consider that the lower extremity consists of three segments which are successively united by joints. Since during the stance phase of locomotion, the foot may be considered to be fixed distally, the shank and the thigh above will alter their function in response to the foot's requirements. It is logical to assume that any deviation of the foot from normal function would affect the more proximal joints. In the case of an unbalanced foot, misdirected forces and moments could cause the lower extremity to compensate in some manner. The resultant symptoms caused by compensation of the lower extremity could vary widely, but it is felt that the most common symptoms are related to the knee joint (Tiberio, 1987).

To date, the few kinetic investigations of orthotics which have been performed, have focused on center of pressure patterns. A center of pressure pattern is a graphic representation of the location and magnitude of the resultant force system between the foot and the force

plate (Soutas-Little, Beavis, Verstraete, & Markus 1987). Clearly, the position of the foot and the resulting direction of the forces and moments acting at the foot affects the gait pattern of the entire lower extremity. Center of pressure information is critical to an understanding of the mechanics of gait. Since accurate assessment of foot motion inside a shoe is difficult, the center of pressure is primarily used for studying the effects of conditions involving plantar type pain on the progression of forces along the foot during the stance phase of gait. At this present time there have been no reported investigations into the effects of orthotic devices on the forces and moments acting at the joints of the lower extremity which originate from the ground reaction force system.

In the case of the structurally unbalanced foot, the direction of the resultant forces and moments may be detrimental to proper mechanics in the joints of the lower extremity, namely the knee. Thus, an orthotic device which was designed to balance the foot would also change the direction and the effects of the resultant forces and moments. By changing the direction of the resultant forces and moments on the lower extremity, the detrimental stresses would be reduced.

#### I. NEED FOR THE STUDY

Experiments directed at the analysis of the effects of orthotics on the forces and moments acting on the knee

joint and in the surrounding connective tissue are lacking. The forces and moments acting at the knee joint which would most likely be affected by orthotics are those acting in the medial or lateral directions in the frontal plane. According to Morrison (1970), moments of abduction or adduction acting on the knee joint are resisted by the collateral ligaments. Abnormally large moments of abduction or adduction may be responsible for the clinical symptoms of knee pain which are effectively treated by the use of orthotics. It may be possible that if an individual were to continue their knee pain producing activity, over a long enough period of time without any treatment, abnormal knee joint motion might occur. Abnormal knee joint motion would be present at the end stages of pathology as a result of the cumulative trauma effects of misdirected or abnormally large abduction-adduction moments. Once there is measurable abnormal knee joint motion the damage to the restraining structures has occurred and orthotics would not be an effective treatment.

Presently, there has been limited research performed on the effects of orthotic devices on the forces and moments at the knee joint. Consider the frequency of orthotic therapy: the incidence of knee injury has been reported to range from 29 to more than 50 percent in the 30 to 40 million Americans who run (Krakauer, 1987) and nearly 80 percent of the runners who complained of knee



pain were eventually treated with orthotics (Newell & Bramwell, 1984). Furthermore, orthotic therapy is becoming more prevalent in sports other than running and even nonsport related rehabilitation programs. Considering the paucity of kinetic investigations on the effects of orthotics, it appears that orthotic devices are poorly understood. This lack of kinetic information may be a contributing factor to inappropriate and excessive prescription of orthotic devices. Consequently, investigation of the effects of orthotics on the resulting forces and moments of the lower extremity during the stance phase of gait needs to be performed. This study investigated the effects of orthotics on the abduction-adduction moment acting at the knee joint during the support phase of walking gait.

## II. PURPOSE OF THE STUDY

The interdependency of foot function and knee function have been well established (Buchbinder, Napora, & Biggs, 1979). Certain changes in foot structure, and thus its resulting function, cause changes in knee joint function. Therefore, the purpose of this study was to perform a three dimensional investigation to examine the effects of a clinically prescribed, semirigid, orthotic on the abduction-adduction moment at the knee during the stance phase of walking.



### III. RESEARCH PLAN

A three dimensional biomechanical analysis of five subjects walking under two different conditions, shoe alone and shoe plus orthotic, was performed. Experimental data were simultaneously collected using high speed cinematography and a force plate. For this study, patients who had excessive pronation and had been successfully treated by a podiatrist with a semirigid orthotic served as subjects. All subjects wore the same type of shoes (Brooks Tempo) for testing. This shoe was chosen because it incorporated a design which would not have any interactive effect on the test conditions. The methods used in this study incorporated rigid body mechanics and the Newton-Euler equations of motion. The abduction-adduction moment at the knee joint was calculated for each subject from the three dimensional kinematic and kinetic data. The abduction-adduction moment at the knee joint was given in Newton-meters and displayed graphically as a function of stance phase. The graphs were then compared for general patterns and trends in the abduction-adduction moment.

### IV. ASSUMPTIONS AND LIMITATIONS

It was assumed that during each trial the subjects exerted a similar effort and that the orthotic material was consistent for all subjects. The lower extremity was assumed to be a linkage of physically united rigid bodies transmitting forces and moments. In this manner all

physical connections of one body segment with the other allowed the equations of rigid body mechanics to be used to determine the resultant forces and moments transmitted across the joint surface which separates the two adjacent body segments.

The limitations of the study were, first, the small sample size which was obtained from a single podiatric practitioner, and second, that the abduction-adduction moment was calculated from kinematic and kinetic data, not measured directly.

#### V. SIGNIFICANCE OF THE STUDY

The findings of this study made a valuable contribution by filling a void in the present body of knowledge on orthotics. This study provided an objective analysis of the biomechanical effects that orthotic appliances have on knee joint mechanics. A better understanding of these biomechanical effects could lead to better design of orthotic appliances and make the prescribing and fitting of orthotic appliances more cost and time efficient, relying less on trial and error. Moreover, this study also provided much needed information on an objective technique in which to evaluate lower extremity dysfunction in a dynamic, closed kinetic chain condition. This study provided information which could be useful to practitioners. Specifically, the information may contribute to more consistent success in controlling abnormal foot function, the elimination of resulting knee

symptoms and the prevention of the development of abnormal knee motion.

## VI. DEFINITIONS

Abduction. Movement of a body part away from the midline of the body.

Adduction. Movement of a body part towards the midline of the body.

Moment. The measure of the tendency of an applied force or forces to produce rotation about some fixed point or axis.

Normal Walking. Normal is referred to as the nonpathological or uninjured state. Hence, normal walking is locomotion whereby one foot is always in contact with the ground in the absence of signs and symptoms of injury.

Orthotic. An orthotic is a straightening or balancing device that is custom made from a neutral cast of the foot. It can be made from semiflexible or rigid material depending on the amount of control needed. An orthotic is prescribed by a physician.

Stance phase of gait. Stance phase is that period during the gait cycle occurring between heel contact and toe off for a single limb.

technique of stereophotogrammetry into the studies of human gait (Cappozzo, 1984). In the early 1900's reproducible measurements of joint forces began using pneumatic force

## REVIEW OF THE LITERATURE

Two main areas in the literature were identified as relevant to the purpose of this study and were therefore reviewed. First, an overview of the branch of biomechanics defined as gait analysis, followed by current research on orthotic therapy, will be discussed.

### I. GAIT ANALYSIS

Gait analysis is defined as a quantitative description of human locomotion (Chao, 1985). Human locomotion is the action by which one's body is propelled through space. Locomotion is achieved by coordinated movements of the body segments employing an interplay of internal and external forces (Cappozzo, 1984). Gait analysis studies began as early as 1680 by Borelli, in the form of simple observational methods. The development of cinematography in 1887, facilitated the growth of gait studies not only by providing a systematic method which permitted detailed, objective observations of the gait phenomenon but, also by providing a means to "capture" these observations for later viewing. The quantitative aspects of gait analysis studies began in 1895, when Braune and Fisher began to represent the human body segments as rigid bodies while introducing the



technique of stereophotogrammetry into the studies of human gait (Cappozzo, 1984). In the early 1900's reproducible measurements of foot forces began using pneumatic force transducers which measured the three components of force from both the right and left feet. During the same time, the point of force application or center of pressure utilizing the centroid of vertical forces, was first determined (Cavanagh, 1982). Since then human walking has been extensively studied by many investigators (Andriacchi, Olge, & Galante, 1977; Dec, Saunder, Inman, & Eberhart, 1953; Murray, 1967; Simon, Trieshmann, Burdett, Ewald, & Sledge, 1983).

Gait analysis studies can be divided into three general areas: (A) kinematic study of segment and joint motion, (B) foot/ground reaction forces and center of pressure analysis and, (C) internal joint and muscle force prediction. This study incorporated information from each of these areas in the description of the effects of orthotics on the mechanics of the knee joint during the stance phase in walking gait. Appropriate consideration was given to each area as it pertained to this study.

#### A. KINEMATIC STUDIES

The kinematic area of gait analysis consists of many diversified studies which have attempted to quantify normal and abnormal (pathological) gait characteristics. For example, several investigators (Murray, Drought, & Ross, 1964; Murray, 1967; Murray, Kory, & Sepic, 1970;

Saunders et al., 1953) have thoroughly described the temporal and distance patterns of normal gait. Other kinematic studies have shown that many gait parameters were velocity dependent and that a change in one of the parameters, such as cadence, produced changes in the overall gait pattern (Andriacchi et al., 1977; Dillman, 1974; Smith, McDermid, & Shideman, 1960). These descriptions of gait patterns and parameters served as the fundamental base on which joint motion, ground reaction, joint, and muscle forces could be evaluated.

Description of lower extremity joint motion in the sagittal plane during gait has been studied extensively (Dec et al., 1953; Murray, 1967). However, normal walking involves a complex three dimensional motion of a multiple linkage system. The techniques which allowed for description of this three dimensional motion of the human body have been available since the end of the last century, but the clinical application of the information has been difficult. Therefore, many of the kinematic studies performed on gait have been two dimensional in nature in order to provide clinically understandable and useful information. Within the last decade, investigators (Andriacchi, Anderson, Fermier, Stern, & Galante, 1980; Andriacchi, Galante, & Fermier, 1982; Harrington, 1983; Simon et al., 1983), have been successful at providing clinically understandable three dimensional analyses of joint motion in normal and pathological states.

In order to describe three dimensional joint motion in the lower extremity during gait, localized coordinate axes must be defined on the basis of underlying skeletal structures and properly placed surface markers attached directly on the skin. Once the spatial locations of these markers and their relative relationship with the bone orientation are known, the joint motion can be calculated following the classical mechanical method (Chao, 1985). In using this method it must be assumed that joint translation is very small. Furthermore, all joint motion recorded in kinematic studies of walking or running gait should be classified as gross motion since the motion does not reflect the true articulating joint surface motion (Chao, 1985).

#### B. FOOT-GROUND REACTION FORCE AND PRESSURE

The second area in gait analysis is the study of foot-ground reaction and center of pressure patterns. Many of the studies in this area measure the relative forces between the foot and the ground utilizing a piezo electric or strain gage force plate capable of recording the vertical, antero-posterior, and medio-lateral components throughout the stance phase of gait (Chao, 1985). The resultant force point of application (center of pressure) can be determined in reference to the force plate coordinate system or to the foot-shoe itself. If the positions of the lower extremity (kinematic data) are known during the stance phase when this force information



is recorded, then the external moment due to the foot-ground reaction force vector can be assessed at a joint. Measurement of the ground reaction force using a force plate does not give pressure distribution on the foot but only location and magnitude of the resultant force of the pressure system (Soutas-Little, Ulibarri, Goodman, & Hull, 1987). In order to measure pressure distribution beneath the foot, discrete transducer elements are either mounted in a walkway or applied directly to the plantar aspect of the foot (Cavanagh & Ae, 1980). This type of device provides information concerning the area of contact and the intensity of contact pressure in localized regions of the foot.

Evaluating gait using the ground reaction forces has significant clinical value, since it is thought that many lower extremity and low back problems are associated either directly or indirectly with foot structure or foot function during the support phase of gait (James, Bates, & Osternig, 1978). Many investigators, (Cavanagh & Ae, 1980; Cavanagh & LaFortune, 1980; Grundy, Tosh, McLeish, & Smidt, 1975; Stokes, Stott, & Hetton, 1974) have attempted to describe normal foot-ground reaction forces and center of pressure patterns in walking and running. Other studies, (Beekman, Louis, Rosich, & Coppola, 1985; Jacobs, Skorecki, & Charnley, 1972; Katoh, Chao, Laughman, Schnieder, & Murray, 1983) have compared the foot-ground reaction forces and center of pressure patterns in normal



and pathological gait patterns, finding distinct patterns of similarity in normal gait and significant abnormalities in the pathological states. Bates, et al. (1983) investigated the interactive nature between subjects and different types of shoes with respect to ground reaction forces in an attempt to evaluate the effects of several types of shoes on the ground reaction force in runners. They concluded that there was no best shoe for all runners due to the unique anatomical structure and biomechanical characteristics of individuals. Therefore, the attempt to find an optimal running shoe design appear to be unimportant. More important in a running shoe design would be a shoe which contributed to the effectiveness of an orthotic device.

The evaluation of orthotics through using of center of pressure calculations is limited by the difficulty of establishing normal patterns of the center of pressure for a general population. There is an inherent variability in the center of pressure distribution and ground reaction forces for individuals which complicates the establishment of normal patterns (Cavanagh & LaFortune, 1980). Cavanagh and LaFortune (1980) have suggested that more powerful techniques for the identification of pathology would be a comparison of the results from the same individual in an injured and uninjured state. For an individual needing a prescribed orthotic, the injured state would be gait without the orthotics in the shoes and the uninjured state

would be gait with the orthotic in the same shoes. The resultant shoe-ground reaction forces may be different between these two conditions, i.e. the injured and uninjured states. Aclun and Brull (1976) reported, that the quantitative effects of orthopaedic appliances and devices may be evaluated from dynamic foot-ground forces.

### C. JOINT AND MUSCLE FORCE ANALYSIS

The final area of gait analysis to be discussed deals with the forces that caused the observed motion. This area is called kinetic analysis of gait. There are two groups of forces acting on the body: internal and external. The internal forces are those transmitted by body tissues and consist mainly of muscular forces, tension forces in ligaments, and joint contact forces. The external forces consist of those which result from the physical interaction between the environment and the body. Examples of these external forces are the gravitational and ground reaction forces. The external forces can be easily measured or reliably calculated, relative to the measurement of internal forces which often requires invasive techniques. In order to avoid invasive techniques, the internal forces must be predicted based on motion data, ground reaction force data and inertial properties. In reference to kinetic analysis, Chao (1985) concluded: "the incomplete and imprecise kinematic data of limb movement during gait and the extremely complex anatomical geometry and inertial properties of the

musculoskeletal systems make the force analysis in gait one of the most challenging tasks in biomechanics" (p. 237). Five, an orthotic device, excessive, inefficient.

The force analysis problem is often solved using many assumptions and over simplifications in modeling and may involve complicated computations. Practical, clinically useful information is difficult to obtain. However, it has been concluded by Andriacchi, et al. (1980), that useful information in terms of joint moments appear to be valuable in assessing functional effects of therapeutic and rehabilitation procedures. In a second study, Andriacchi et al. (1982), used the moments of the knee joint to evaluate total knee replacement design. The moments at the knee joint were due to the resultant ground reaction force. Therefore, slight changes in the resultant ground reaction force may be significant in the pathological knee abduction-adduction moment. Since knee pain may result from a pathological abduction-adduction moment, comparisons of the knee abduction-adduction moment in an injured and uninjured state would be of major clinical relevance.

## II. ORTHOTIC THERAPY

The intent of orthotic therapy is to support the foot in its neutral position in order to prevent excessive forces and motion while allowing normal joint function and pain free activity. The orthotic device itself is a straightening or balancing device with its main objective



to maintain the foot in its neutral position throughout the walking and running gait cycle. In accomplishing this objective, an orthotic prevents excessive, inefficient, compensatory motion and the abnormal forces associated with compensation (Eggold, 1981). The major purpose when treating patients with orthotics is to restore "normal" biomechanical function of the lower extremity, i.e. asymptomatic biomechanical function. The ability to quantify any biomechanical changes in the function of the lower extremity as a result of the use of orthotics is of major clinical importance. The ability to establish objective biomechanical information depends on utilization of efficient measurement techniques and data analysis methods. Joint motion of the lower extremity and ground reaction force parameters during the stance phase of gait are important in the biomechanical assessment of the effects of orthotics on the lower extremity during gait, since the ground reaction forces represent the total effect of functional and mechanical alterations in the lower extremity linkage (Knutzen & Bates, 1988). These parameters can be identified and analyzed through the techniques and methods of gait analyses previously discussed.

James, Bates, and Osternig (1977), suggested that many of the lower extremity problems encountered by runners were either directly or indirectly related to foot function during the support phase of gait. It could also



be argued that this concept applies to nonrunners who are experiencing chronic, nontraumatic lower extremity problems, such as chondromalacia patella. It has been reported (Eggold, 1981; James, 1975; Nevall & Bramwell, 1981) that many of these lower extremity problems could be alleviated by the use of an orthotic appliance. However, few studies have quantitatively investigated the use of orthotics. Most of the studies on the effectiveness of orthotics have either been based on results of surveys of runners who have been treated with orthotics (Eggold, 1981), or descriptive case studies (Nevall & Bramwell, 1984). D'Ambrosia and Douglas (1982), stated "orthotic treatment, in the form of foot supports, will relieve the symptoms, and combat the causes of chondromalacia patella, and at the same time allow the distance runner to get back on the road again" (p. 161). In addition, Subotnick (1975) reported that rigid orthotics were very successful in relieving symptoms of chondromalacia.

The most common abnormal foot structure is the pronated or flat foot. A variety of orthotic devices have been developed to manage the pronated foot. Clinically, orthotics are prescribed to alter foot and leg motion, through redirection of the ground reaction forces, and to aid in shock absorption (Smith, Clarke, Hamill, & Santopietro, 1986). There are three categories of orthotics: rigid, semirigid, and soft.

Rigid orthotics are made from a non-weight bearing

neutral cast of the foot or they also may be molded directly on the patient's foot. Rigid orthotics are made of rigid plastic material which reduce the shock attenuation ability and energy absorbing movements of the foot (Lockard, 1988). For this reason, rigid orthotics are often utilized to accommodate foot structural deformities and they are best suited for walking and edge control sports such as skiing. Rigid orthotics are not usually well tolerated by runners. Semirigid orthotics are made from a non-weight bearing neutral cast of the foot. The semirigid orthotics are used to provide some softness and reduce impact forces. However, semi-rigid orthotics are more commonly used to balance the malaligned foot in a neutral position to reduce abnormal foot or leg movements (Lockard, 1988). These orthotics use rubber posts at the rear-foot and fore-foot to control the motion and provide some cushioning. A semirigid orthotic is often used in sport applications where correction is required under high-impact loading; e.g. basketball, track, baseball, etc.. Soft orthotics are composed of felt and other soft material that will adapt to the shape of the foot in order to provide cushioning to improve shock absorption, to reduce plantar surface shearing or to control abnormal motion of the foot in cases of mild biomechanical imbalances (Lockard, 1988). Soft inserts do not require custom molding and are available commercially. These are often temporary devices, used to

determine whether permanent orthotics are needed (Subotnick, 1983).

The most frequent method of quantifying the effects of orthotics has been kinematic analysis of rearfoot motion. This rearfoot motion consists of eversion or inversion of the calcaneus relative to the lower leg and is thought to be an accurate predictor of how much pronation is occurring during the support phase of gait (Smith, et al. 1986). However, there exist many problems with this rearfoot method of analysis which may lead to inaccurate conclusions. Some of these problems are: the assumption that pronation can be accurately predicted by calcaneus eversion when pronation is a triplanar movement; attempting to measure motion occurring at the subtalar joint by external, surface markers; that the rear foot motion being described is actually rear shoe motion; and in two dimensional sagittal plane analysis one encounters distortion of the projected angle between the calcaneus and the lower leg (Soutas-Little, et al., 1987).

Hamill and others (1986), used an in shoe pressure sensor system to evaluate the interaction between orthotic appliances and running shoes and concluded that the addition of a semirigid orthotic appeared to control the foot at impact better in a rigid midsole shoe than in a soft midsole shoe. In a study by Burdett, Kohrt, and Buchbinder (1985), the effects of orthotics on frontal plane knee kinematics were investigated. They stated that

since past studies have indicated orthotics were beneficial in controlling excessive pronation at the subtalar joint, it may be likely that an orthotic device would also control a similar motion at the knee, i.e. excessive abduction-adduction. However, the results from their study were inconclusive.

Orthotics are frequently prescribed in the management of lower extremity problems. Considering the paucity of kinetic investigations on the effects of orthotics, it appears that the mechanics of orthotic devices are poorly understood. This lack of kinetic information may be a contributing factor to inappropriate and excessive prescription of orthotic devices.

None of the subjects had any current or previous lower extremity injuries (i.e. fractures, sprains, or surgeries). The study was approved by the Institutional Review Board and a graduate student, who was not involved in the study, informed about the study and asked the subjects to participate in the study. The subjects were targeted for filming and were not compensated for their participation (see Appendix). Anthropometric measurements were taken prior to the trials. These measurements included total body weight and height. All of the subjects were filmed during the trials. The subjects were asked to



complete two satisfactory trials under two conditions. Satisfactory trials were determined by this investigator based on whether the normal foot corrected the force plate while the subject walked. **METHODS**

The purpose of this study was to determine the effects of an orthotic appliance on the abduction-adduction moment occurring at the knee joint during the stance phase of walking gait. Five male volunteers served as subjects for this study. Four of the volunteers were obtained from a podiatrist's patient population based on the following requirements: all subjects were diagnosed as excessive foot pronators with past complaints including knee pain, the subjects were currently using prescribed, semirigid orthotics during walking or running activities, and the subjects had no history of previous, unrelated lower extremity injuries (i.e., fractures, sprains or surgeries). The fifth volunteer was an asymptomatic runner and a graduate student. All subjects were fully informed about the study and signed a written consent form to participate in the study. Each subject's right leg was targeted for filming according to the research protocol (See Appendix). Anthropometric measurements were taken prior to the trials. These measurements consisted of total body weight, and width and length of the right tibia. All of the subjects wore a similar shoe (Brooks Tempo) during the trials. Each subject was required to

complete two satisfactory trials under two conditions. Satisfactory trials were determined by this investigator based on whether the complete foot contacted the force plate while the cameras were properly timed in capturing the movement. The first condition consisted of walking without the orthotic in the shoes and the second condition consisted of walking with the orthotic in the shoes. These conditions produced the pathological and non-pathological states for each of the subjects, respectively. The order of these two conditions was consistent for all subjects. All of the subjects received similar instructions prior to the trials and as many practice trials as necessary were allowed.

#### I. DATA COLLECTION PROCEDURES

The shank and foot were considered to be a rigid body. The targeting scheme is shown in the Appendix. Four targets were placed on the surface of the shank over the following bony landmarks: [d] lateral malleoli, [e] distal tibia just proximal to the tibialis anterior tendon, [f] tibial tuberosity, and [g] lateral tibial condyle. This targeting scheme was established to facilitate recording of kinematic parameters (linear positions, velocities, and accelerations) of the shank and also to locate the center of mass of the shank segment and the center of pressure relative to the shank. The placement of the targets allowed for the solution of the inverse dynamics problem encountered in analyzing the

motions of a rigid body when the forces and moments required to create those motions are unknown. The objective of this procedure was to determine the unknown forces and moments acting at the knee joint.

Two 16 mm LOCAM high speed cine cameras were used to film the subjects as they walked across a force plate and recorded the kinematic activity of the right lower leg. The cameras were placed at angles to the plane of gait allowing both cameras lateral-anterior views of the right lower extremity. The cameras were started and stopped simultaneously. Film speed was set at 100 frames per second and the shutter angle was set at 120 degrees allowing for a  $1/300$  of a second exposure. Prior to filming the trials, a calibration structure was placed on the force plate and filmed in order to provide control points for the direct linear transformation (DLT) technique used (Walton, 1981). Synchronized timing lights accurate to  $1/1000$  of a second were placed in the field of view of both cameras so that both sets of kinematic data could be time-matched. An AMTI Force-Torque (Model OR6) dynamometer was used in conjunction with an IBM 9000 dedicated computer for simultaneous collection of three dimensional force and moment data at a frequency of 1000Hz. The kinetic data were stored on a floppy disk, then transferred for analysis to the Prime computer in the Computer Aided Design Laboratory in the College of Engineering at Michigan State University. The digitized

kinematic data were also transferred to the Prime Computer for analysis.

## II. ANALYTICAL METHODS

Data analysis consisted of the evaluation of a single right stance phase for all subjects over two trials for the two described conditions. The three dimensional kinetic and kinematic experimental data were then post-processed.

Film data from both cameras were quantified using an ALTEK Datatab rear projection system. Every other frame of data was digitized and began 10 frames prior to right heel strike and ended 10 frames after right toeoff. The order of digitization of the targets on the right lower extremity was done according to the study protocol (See Appendix). The coordinate data for the two views was time matched by linear interpolation and then three dimensional coordinates of the right lower extremity targets were computed using the DLT technique. The fixed laboratory coordinate axes were defined by the calibration structure filmed during the experimental protocol which defined the z-axis as perpendicular to the ground, positive in the superior direction. The x-axis was formed perpendicular to z and positive in the direction of motion, and the y-axis was derived perpendicular the z and x axes and was positive in the medial direction with respect to the right leg.

The force plate measured the resultant force vector



and the moment vector acting on the platform by the foot during the stance phase of gait. The direction of the forces and moments recorded by the force plate were reversed to obtain the forces and moments acting on the foot by the ground. The resultant force and moment vectors,  $\bar{R}$  and  $\bar{M}$ , were recorded with respect to the origin and axes of the force plate coordinate system, as shown in Figure 1. The moment vector was resolved through vector analysis into two vectors; one perpendicular to the resultant force vector and the other parallel to the resultant force vector. The intercept of the resultant force vector and the parallel moment vector with the surface of the force plate defined the center of pressure (Soutas-Little, et al., 1988). The center of pressure was computed with respect to the force plate coordinate system. The resultant force vector, the parallel moment vector, and the center of pressure were then transformed by a series of matrix equations to the lab coordinate system and then to the local or body coordinate system.

The abduction-adduction moment acting at the right knee was computed in terms of a local coordinate system aligned with the principal axes of the shank. The origin of this local coordinate system was chosen to be located at the segmental center of mass of the shank defined by Dempster (1955). The center of mass of the shank was located from the kinematic and anthropometric data obtained in this study. The schematic representation of the shank segment is shown in Figure 2, including the

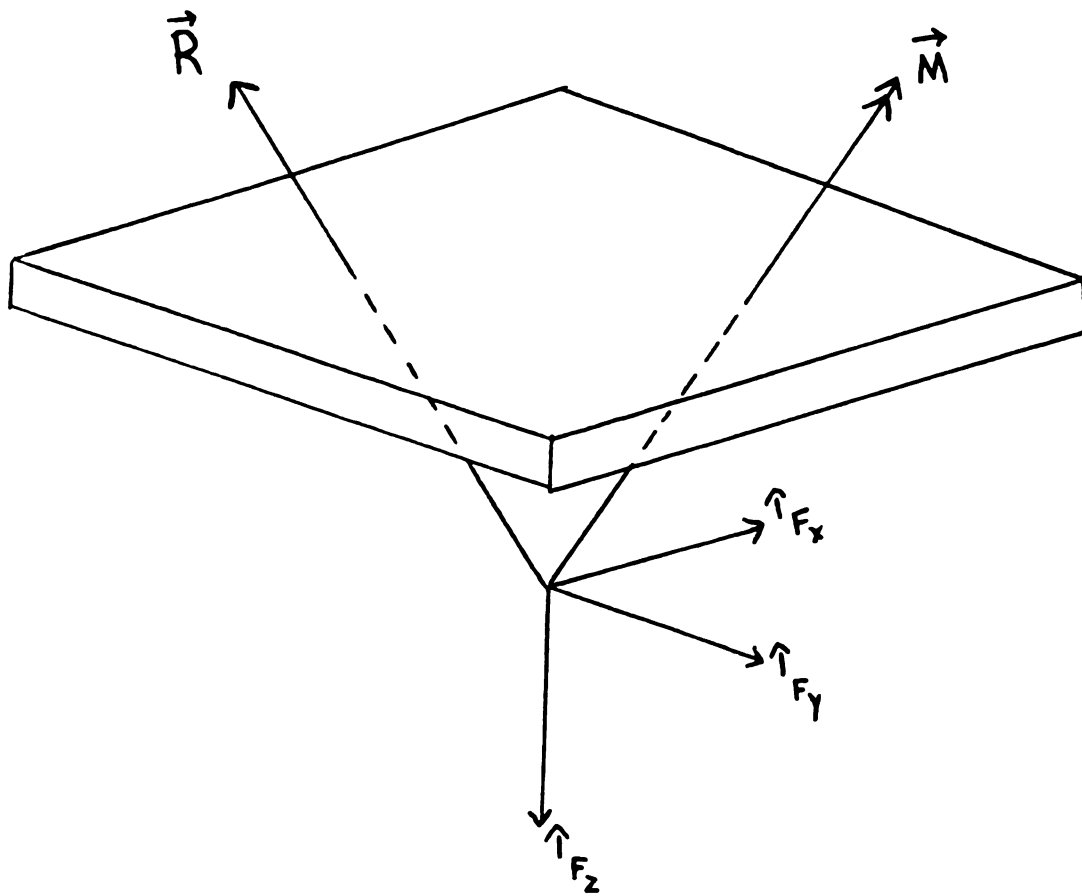


Figure 1. The Force Plate Coordinate Axis and the Ground Reactions.

**Figure 2. Schematic Representation of the Shank.**

local coordinate system located at the center of mass of the shank, and the components of the external forces and moments acting on the segment; with reference to the direction of motion, position of the cameras and the lab coordinate system. The Newton-Euler equations of motion for the shank which were used to calculate the moments and forces acting at the knee joint were as follows:

$$R_{j1} = m(a_{cm1} - g_1) - R_1 \quad [1]$$

$$R_{j2} = m(a_{cm2} - g_2) - R_2 \quad [2]$$

$$R_{j3} = m(a_{cm3} - g_3) - R_3 \quad [3]$$

$$\begin{aligned} M_{j1} = & I_1 \dot{\omega}_1 - \omega_2 \omega_3 (I_2 - I_3) - M_1 \\ & - (r_{cpcm2} R_3 - r_{cpcm3} R_2) - r_{pcm2} R_{j3} + r_{pcm3} R_{j2} \end{aligned} \quad [4]$$

$$\begin{aligned} M_{j2} = & I_2 \dot{\omega}_2 - \omega_3 \omega_1 (I_3 - I_1) - M_2 \\ & - (r_{cpcm3} R_1 - r_{cpcm1} R_3) - r_{pcm3} R_{j1} + r_{pcm1} R_{j3} \end{aligned} \quad [5]$$

$$\begin{aligned} M_{j3} = & I_3 \dot{\omega}_3 - \omega_1 \omega_2 (I_1 - I_2) - M_3 \\ & - (r_{cpcm1} R_2 - r_{cpcm2} R_1) - r_{pcm1} R_{j2} + r_{pcm2} R_{j1} \end{aligned} \quad [6]$$

where:

$a_{cm1}$  were the principle components of the acceleration of the center of mass of the shank,

$I_1$  were the principle components of the inertia tensor,

$\dot{\omega}_1$  were the principle components of the angular acceleration,



and  $v_1$  were the principle components of the angular velocity.

A numerical method, based on the method of least squares as described by Verstraete and Soutas-Little (1990), was used to determine the angular acceleration and angular velocity of the shank from the three-dimensional kinematic data. The Newton-Euler equations were expressed with respect to the principal axes and center of mass of the shank, where 1, 2, and 3 refer to the principal directions of the coordinate system. A positive force and moment convention was assumed. The vector  $\bar{r}_{cpcm}$  represented the relative position vector between the center of pressure and the center of mass of the shank. The vector,  $\bar{r}_{pcm}$ , represented the relative position between the center of mass and the knee joint center.  $M_{j1}$ , in equation (4), was the abduction-adduction moment. The inertial effects during walking of the foot on the knee abduction-adduction moment were considered insignificant (Verstraete, 1988) and were therefore excluded in this analysis. The knee joint was assumed to consist solely of the proximal portion of the tibia and the distal end of the femur. The center of the knee joint was defined to lie on the contact surface of the tibial plateau, at a point midway between the lateral and medial tibial condyles (knee width). In this analysis the knee joint center was used as a reference point of contact between the shank and thigh segments. The forces and

moments in the tibia were resolved with respect to this central linkage point defined on the proximal surface of the tibia. The computed moments acting at this linkage point, the defined center of the knee joint, were computed in Newton-meters and graphed as a function of percent of stance phase. The graphs were then compared for general patterns and trends in the abduction-adduction moment.

## RESULTS AND DISCUSSION

The results of the analysis of the three-dimensional kinematic and kinetic data consisted of the computed abduction-adduction moment acting at the defined center of the knee joint. These results were obtained from test on the right lower leg of four male subjects during the stance phase of walking gait with and without the application of an orthotic appliance. Subject 5 was not included in these results due to a technical error in data collection. Since Subject 3 was an asymptomatic runner, he was not tested walking with an orthotic and thus served as a control. The results were expressed in Newton-meters and presented graphically as a function of percent of stance phase. The action of this moment corresponded to the rotational motion of the distal shank segment relative to the thigh segment. According to the positive convention assumed in this study, a positive value for the computed moment would indicate abduction, while a negative value would indicate adduction. Therefore, an adduction moment would tend to cause a varus angulation at the knee and an abduction moment would cause a valgus angulation. The graphs were analyzed for general patterns and trends across all trials for both conditions for

each subject. The results will be presented in two parts, first for the condition without the orthotic appliance and second for the condition with the orthotic appliance. The discussion of these results follows and includes comparisons between the two conditions.

#### I. CONDITION I: WALKING WITHOUT ORTHOTIC

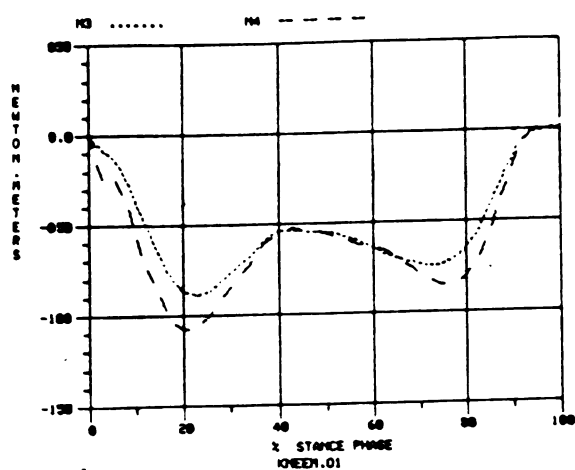
The abduction-adduction moment acting at the linkage center of the knee joint for both trials for each subject walking without an orthotic appliance in the shoe is displayed in each graph in Figure 3. In all subjects, the abduction-adduction moment displayed a predominant adduction action during the stance phase. The pattern of the abduction-adduction moment among the subjects was a consistent trend which could be depicted as a "W" shaped curve. Each curve was found to have three peaks or points of change in direction. The mean values of the peaks of the abduction-adduction moment curves for both trials for each subject are given in Table 1. For each subject,

TABLE 1. Mean Values of the Abduction-Adduction Moment Without Orthotic\*

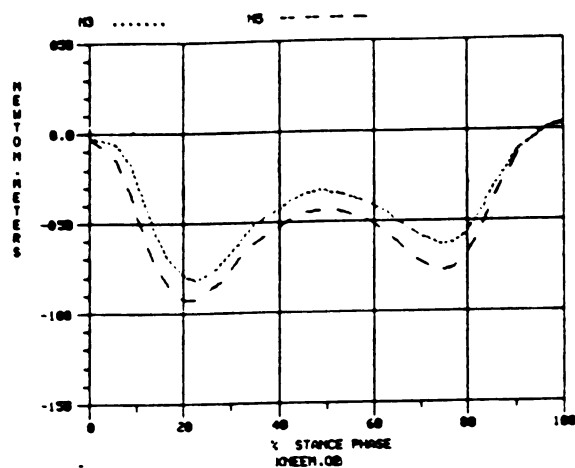
|       | <u>PEAK 1</u> | <u>PEAK 2</u> | <u>PEAK 3</u> |
|-------|---------------|---------------|---------------|
| SUB 1 | -100          | -55           | -80           |
| SUB 2 | -90           | -40           | -75           |
| SUB 3 | -65           | -40           | -55           |
| SUB 4 | -110          | -80           | -100          |

\*expressed in Newton-meters

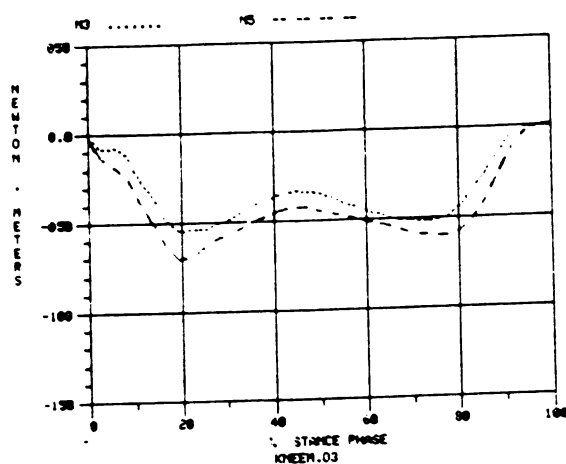




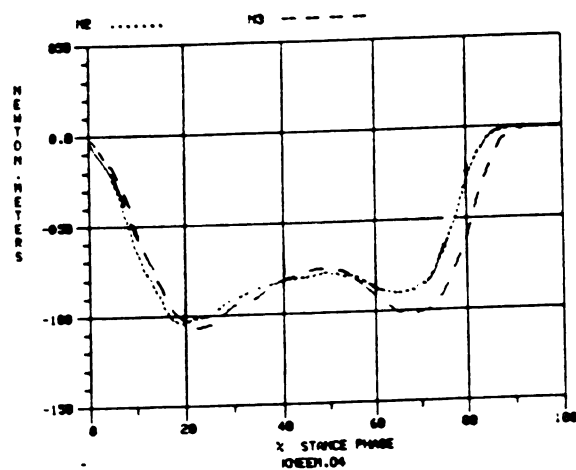
Subject 1



Subject 2



Subject 3



Subject 4

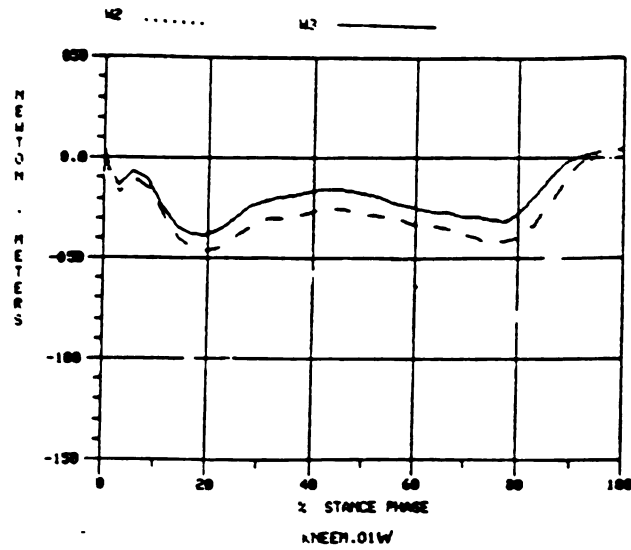
**Figure 3. Abduction-Adduction Moment acting at the Knee Walking Without an Orthotic.**

peak 1 was the largest and peak 2 was the smallest. Comparison of the mean values among subjects showed that Subject 3, the control, had overall lower values than the other, pathological, subjects. The numbers in Table 1, as well as the graphs in Figure 3, showed that the abduction-adduction moment tended to progress in a predictable manner during the stance phase. Analysis of the graphs, specifically the magnitude of the adduction moment, revealed the following pattern: there was a increase in the adduction moment during the first 30% of the stance phase, then an inconsistent decrease to 50-60% of the stance phase and then again an increase, until 90% of the stance phase. At 90% of stance phase the adduction moment showed a consistent decrease to toe off. The graphs were found to be highly reproducible within subject trials. There was slight variation in the graphs between the trials for each subject which could be attributed to a variation in walking speed between trials. The only noticeable variation in the "W" pattern was the amount of fluctuation between the peaks of the abduction-adduction moment among the subjects. The general pattern of the abduction-adduction moment displayed by these results were consistent with those described by Andriacchi and Strickland (1985) and Verstraete (1988).

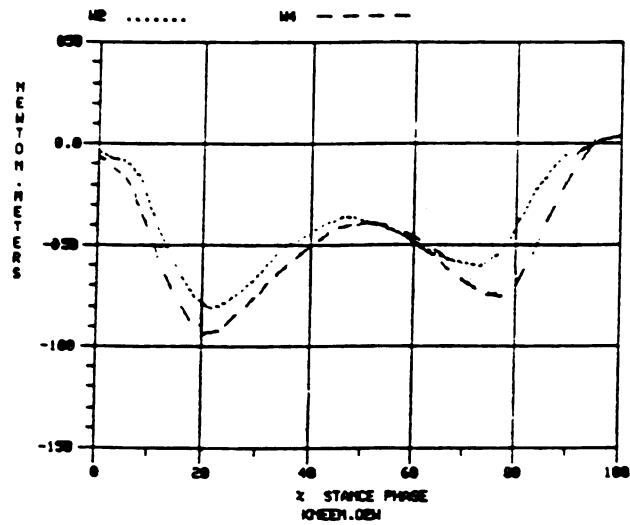
## II. CONDITION II: WALKING WITH ORTHOTIC

The resultant abduction-adduction moment computed at the knee joint for the condition of walking with an

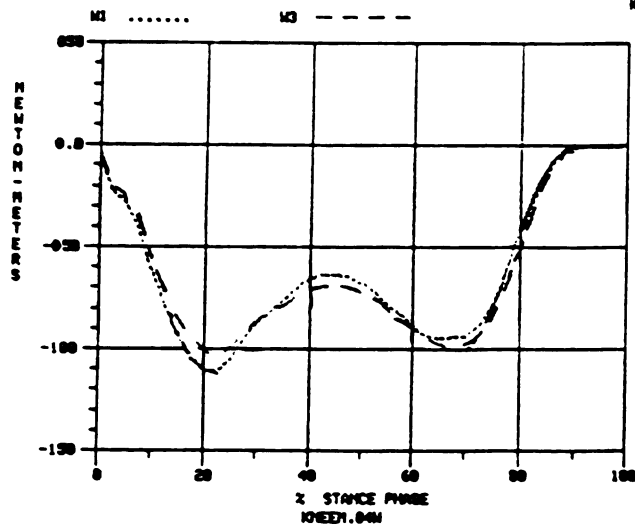
orthotic appliance for each subject over both trials are displayed in Figure 4. Subject 3, the control, was not tested walking with an orthotic. The actions of the abduction-adduction moment corresponded to the same rotational motions of the shank segment relative to the thigh segment as presented earlier in this chapter. Again, the abduction-adduction moment was analyzed for general patterns and trends in the magnitudes. Over all trials, the abduction-adduction moment displayed a predominant adduction moment. Similarly, the progression of the abduction-adduction moment computed for these trials exhibited a predictable trend. The adduction moment increased during the first 25% of stance phase, which was followed by a decrease in the adduction moment until 55% of stance phase. Next the adduction moment increased until 90% of stance phase, and then the adduction moment steadily decreased to toe-off. This general pattern was very similar to the pattern displayed for the abduction-adduction moment in the trials without the orthotic. The pattern of the abduction-adduction moment was also a "W" shaped curve. The only noticeable variation in this "W" pattern was the amount of fluctuation between the peak values of the abduction-adduction moment among the subjects, with the most noticeable differences occurring between Subject 1 and Subjects 2 and 4. The mean values of the peaks of the



Subject 1



Subject 2



Subject 4

Figure 4. Abduction-Adduction Moment acting at the Knee Walking With an Orthotic.

abduction-adduction moment curves for both trials for each subject are presented in Table 2. Here again, the mean values of peak 1 were the largest and the mean values of peak 2 were the smallest. The peak values for Subject 1 showed an overall considerable decrease when compared to the numbers in Table 1. This could reflect the effects of a properly fitted orthotics. The mean values for Subjects 2 and 4 were essentially unchanged from Table 1, This could reflect unnecessary or improperly fitted orthotics which had no effect on the abduction-adduction moment.

TABLE 2. Mean Values of Abduction-Adduction Moment  
With Orthotic\*

|       | <u>Peak 1</u> | <u>Peak 2</u> | <u>Peak 3</u> |
|-------|---------------|---------------|---------------|
| SUB 1 | -45           | -20           | -40           |
| SUB 2 | -90           | -40           | -70           |
| SUB 4 | -110          | -70           | -100          |

\*expressed in Newton-meters

### III. DISCUSSION AND COMPARATIVE ANALYSIS

It is important to realize that variations between the two conditions discussed here may be due to natural responses of the subjects, environmental effects of the testing conditions, and/or experimental errors. Furthermore, another factor limiting the accuracy of the calculation of the abduction-adduction moment occurring at the knee was the definition of the actual



knee joint center and the location of the center of mass of the shank from external markers. Nevertheless, the approach taken in this study was similar to those taken in other related studies that have been described in the literature. Moreover, since a consistent definition of the center of the knee joint and center of the mass of the shank were used, the results are comparable.

The graphs in Figure 5 show the differences in the abduction-adduction moment between the trials without the orthotic (N3 and N4) and the trials with the orthotic (W2 and W3) for Subject 1. The most obvious difference was that the adduction moment was much less in the trials with the orthotic. Also, the shape of the "W" curve is much less evident in the trials with the orthotic. There was less fluctuation in the adduction moment in trials with the orthotic.

The differences in the abduction-adduction moment between the trials without the orthotic (N3 and N5) and the trials with the orthotic (W2 and W4) for Subject 2 are shown in the graphs in Figure 6. There was much less differences between the two conditions for this subject. However, when the mean value of the moment over the two trials for each condition was compared to each other, there was a subtle, yet, consistent difference whereby the adduction moment was slightly less in the trials with the orthotic. The shape or the fluctuation of the "W" curve was not noticeably different between the two conditions

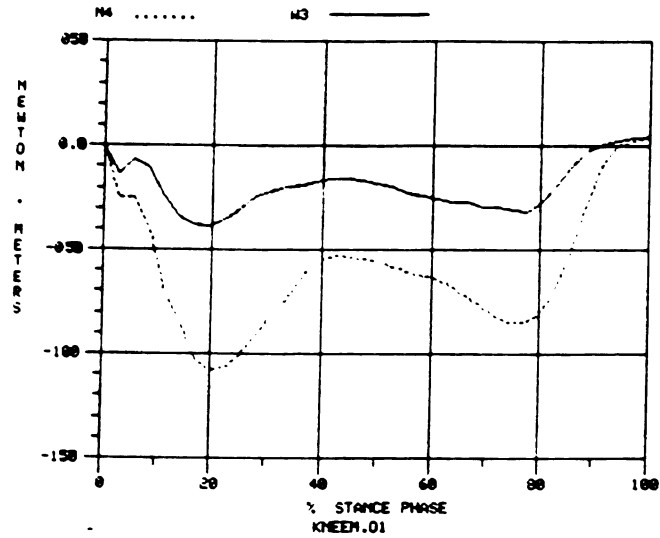
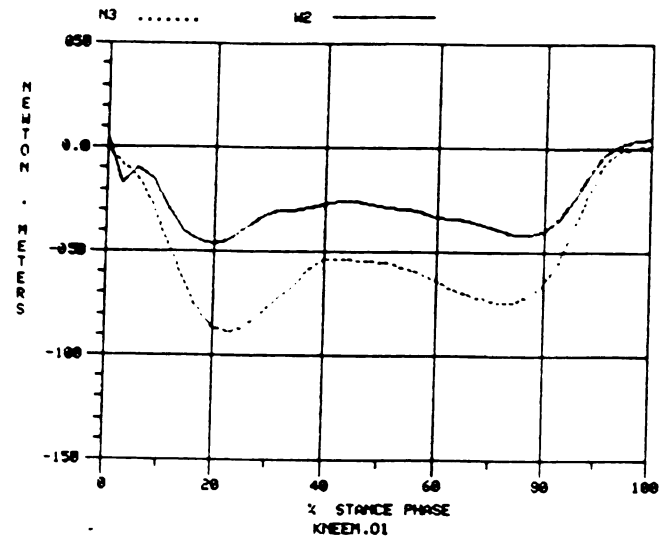


Figure 5. Subject 1 Abduction-Adduction Moment Across Conditions.

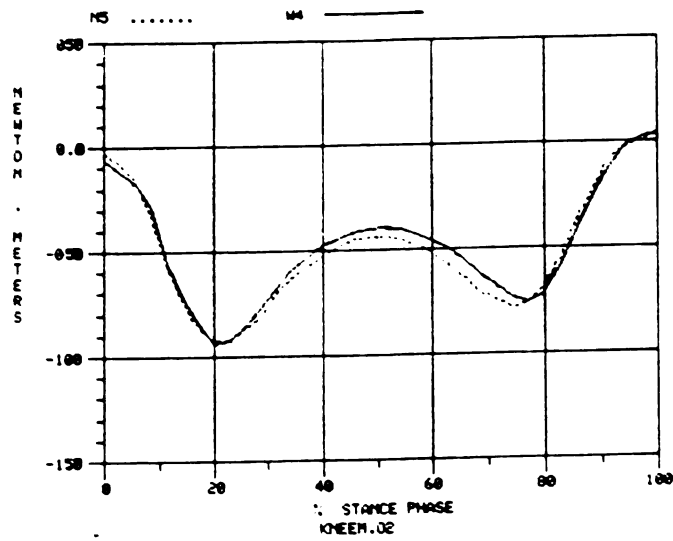
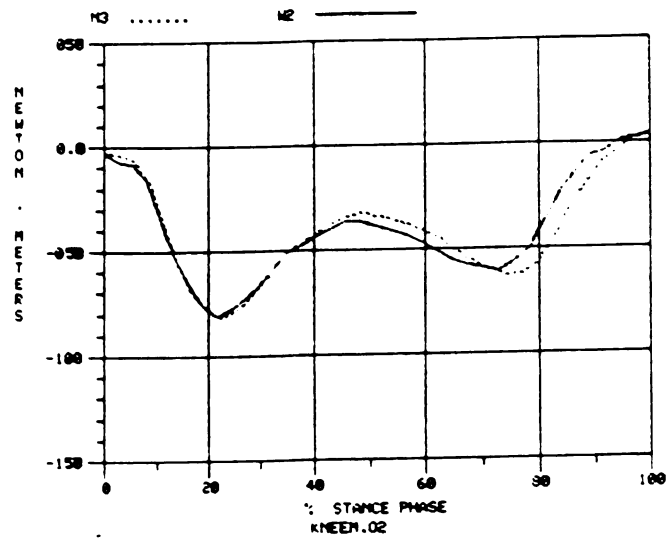


Figure 6. Subject 2 Abduction-Adduction Moment Across Conditions.

for this subject.

The graphs in Figure 7 show the differences in the abduction-adduction moment between the trials without the orthotic (N2 and N3) and the trials with the orthotic (W1 and W3) for Subject 4. The difference between the two conditions for this subject was not noticeable. The "W" pattern of the curve of the abduction-adduction moment was more evident in the trials with the orthotic. There was slightly more fluctuation in the second peak of the adduction moment in the trials with the orthotic, but this difference was minimal.

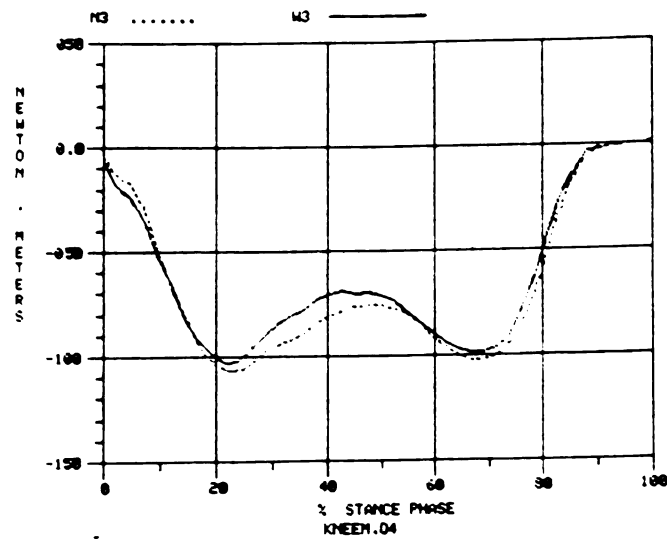
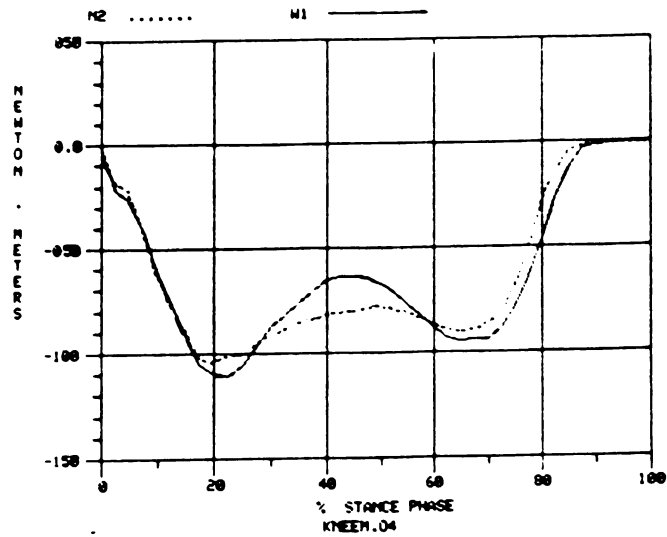
The numbers in Table 3 show the differences in the mean value of the peak magnitudes of the abduction-adduction moment across the two conditions for each subject. A positive difference corresponded to a decrease

TABLE 3. Differences in Mean Peak Values of the Abduction-Adduction Moment across Conditions

|       |     | PEAK 1  | PEAK 2 | PEAK 3 |
|-------|-----|---------|--------|--------|
| SUB 1 | +60 | +35 +45 |        |        |
| SUB 2 |     | 0       | +5     | +5     |
| SUB 4 |     | 0       | +15    | 0      |

expressed in Newton-meters

in the adduction moment magnitude. All of the measurable differences between the two conditions were positive values, indicating that the adduction moment influence was less in the trials with the orthotic.



**Figure 7. Subject 4 Abduction-Adduction Moment Across Conditions.**



An adduction moment at the knee joint would be balanced by tension in the lateral ligaments and a reactive joint contact force medially (Harrington, 1983). The clinical significance of a decrease in the adduction moment would be the decrease in the compression forces in the medial compartment of the knee joint. This could account for the decrease in knee pain which has been reported with orthotic therapy. But that remains unknown at this time. The comparison between the two conditions for Subject 1 showed a noticeable decrease in the adduction moment walking with the orthotic. This could imply a properly prescribed and fitted orthotic. Whereas, the comparisons between the two conditions for Subjects 2 and 4 showed very minimal to no noticeable differences and this could imply inappropriately prescribed or improperly fitted orthotics.

## CONCLUSIONS

The prescribing of foot orthoses has been effective in reducing the lower extremity problems, including knee pain, in recreational athletes (Donatelli, et al., 1988). The application of an orthosis has been an effective external support mechanism that altered the lower extremity function by influencing the medio-lateral function of the foot (Bates, 1979). It has been well established by previous research that the mechanical behavior and forces occurring at the foot plays an integral role in the function of the entire lower extremity during the stance phase of gait (Bates, et al., 1982; Katch, et al., 1982; McPoil & Knecht, 1985; Morris, 1976). Furthermore, the relationship between foot function and knee joint pathology has been well substantiated (Buchbinder, et al., 1979; Davies & Malone, 1980; Eggold, 1981; Lutter, 1980). Kinetic variables have been used primarily to identify gait abnormalities, evaluate shoe performance, and in describing general mechanical concepts of individual joints of the lower extremity (Bates, et al., 1979; Morrison, 1970; Simon, et al., 1983). Recently, attempts been made to examine kinetic alterations with respect to knee joint kinematic

parameters (Knutzen & Bates, 1988). Yet to date, there have been no studies which have specifically attempted to relate knee resultant forces and function with ground reaction forces. Basic mechanical principles require that the external moments acting on a joint, such as the knee joint, be counteracted, by equal and opposite internal moments. In most cases, the external moments would most likely be balanced by a combination of muscle forces, ligamentous tension and joint contact forces. However, if these forces were unable to restrain the external moments, there would then exist abnormal compensatory motions. Another mechanism by which abnormal compensatory motions could develop would be the result of prolonged, repetitive misdirected external forces acting on a joint causing weakening of the internal restraining structures. Consequently, it is plausible that determination and possibly even prediction of knee joint pathology should be evident in analysis of moments occurring at the knee joint, when computed from three-dimensional kinetic and kinematic data collected on the lower extremity during gait.

Quantitative research on the effect(s) of orthotic devices on the resultant forces in the lower extremity is severely lacking. Therefore, the mechanical effect of orthotic therapy on the lower extremity is not known. Determining the kinetic parameters which contribute to knee joint pathology is important for clinicians who are

concerned with controlling foot function to reduce knee joint injuries (Knutzen & Bates, 1988). Subtalar joint pronation is characterized by medial-lateral kinematic and kinetic parameters occurring in the frontal plane. The correlating motions to subtalar joint pronation in the knee joint are abduction-adduction. Considering that mechanically, the stance phase forces represent any functional alterations in the total lower extremity linkage, then a relationship between the kinetic parameters and the resulting function of the joints of the lower extremity would seem logical. Therefore, the effect of the resultant ground reaction force and moment that is transmitted to the knee joint is of considerable interest for clinicians who seek a more objective assessment of gait deviations and subsequent treatment interventions, including foot orthoses. If kinetic parameters can be used to predict knee joint mechanics, the clinical importance would be significant for therapists, physicians, and podiatrists who seek a more objective assessment of knee joint pathology and evaluation of orthotic therapy. Analysis of the general patterns and trends in the abduction-adduction moment acting at the knee joint during the stance phase of walking gait with and without an orthotic appliance was performed to determine if there were any noticeable differences between the two conditions.

The concepts of rigid body mechanics and the

Newton-Euler equations of motion were utilized to perform a three-dimensional gait analysis. This analysis examined the effects of foot orthoses on the computed abduction-adduction moment acting at the defined knee joint center for the right lower extremity during the stance phase of walking gait. The abduction-adduction moment acting at the knee joint was computed from kinematic and kinetic data collected on four male subjects walking under two conditions, shoe alone and shoe plus orthotic. A fifth male subject who was a asymptomatic runner served as the control and was only analyzed walking without orthotics. The abduction-adduction moment was examined for differences in general pattern and trends between two conditions. The general pattern of the abduction-adduction moment between the two test conditions among all subjects was found to be a consistent "W" shape and to be predominantly a negative value throughout the stance phase of gait. According to the convention used in the methods, a negative value corresponded to an adduction moment. There was noticeable variation of the fluctuation of the "W" shape for all subjects but, there was no trend in this variation. Comparison of abduction-adduction moment between the two test conditions revealed that the adduction moment was noticeably decreased in the condition with the orthotic for Subject 1. This would suggest a decrease in the medial joint compression forces at the knee, which may result in decrease knee pain. Subject 1

may have a properly fitted orthotic. Comparison of the abduction-adduction moment between the two test conditions for Subjects 2 and 4 showed very minimal decrease to no noticeable differences in the adduction moment. This may reflect improperly fitted orthotics which have no effect on the mechanics of the lower leg. In addition to these findings, the methods used in this study allowed an objective analysis of the forces and moments acting at the knee joint and may provide physicians, podiatrists and physical therapists with a method for objective assessment of orthotic therapy.

Further studies in this area would be recommended. Tiberio (1987) presented a theoretical model which described compensatory internal rotation of the femur as a result of excessive pronation leading to pathomechanics of the knee joint. The moment of interest in this situation would occur about the inferior-superior axis of the shank, with its potential action occurring in the transverse plane. Consequently, future studies should include the moments acting about the inferior-superior axis of the shank, as well as utilizing a larger sample size, controlling the speed of walking, and obtaining the medical history including specific information on effectiveness of each subject's orthotics.



## **APPENDIX**

## APPENDIX

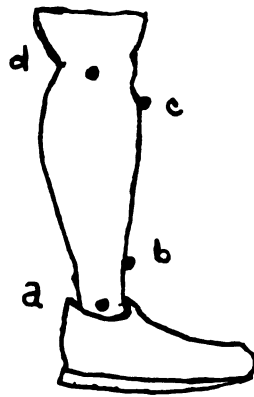
### SUBJECT PREPARATION PROCEDURES

#### CONSENT FORMS

Consent forms were obtained from each subject.

#### TARGETTING SCHEME

The right lower leg of all subjects were targetted with 4 cloth pom-pom targets as follows: [a] at lateral malleoli, [b] at distal tibia, [c] at tibia tuberosity, [d] at tibial condyle. These targets were also digitized in this order (a to d).



#### ANTHROPOMETRIC MEASUREMENTS

Each subject had the following anthropometric measurements recorded prior to completing their trials:

- body weight

- distance between targets at distal and proximal tibia
- width at malleoli and tibial condyles.

#### SUBJECT INSTRUCTIONS

Subjects received instructions to walk at a normal-comfortable pace across the walkway without looking down. The subjects completed three "good" trials without their orthotics in their shoes first, and then completed three "good" trials with their orthotics in their shoes. The shoes to be worn during the trials were Brook's Tempo running shoes.

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